TEKTRONIX

Biophysical Measurements

$10.00

Measurement Concept Series

BIOPHYSICAL MEASUREMENTS

BY
PETER STRONG

Illustrations
by
DOROTHY FREED

TEKTRONIX

MEASUREMENT CONCEPTS

CONTENTS

ACKNOWLEDGMENTS

CAUTION

Engineers and nonmedically qualified personnel should not perform biophysical measurements on human subjects unless such measurements are conducted under professional supervision. Many of the measurement techniques described in this book, while incorporating medically accepted procedures, are not entirely free from risk. These risks may be minimized by following the precautions outlined in Chapter 17, "Grounding - Safety."

One or more "current limiting adapters for protection from electric shock" as described in Chapter 29, Section 29.12, should be used between human subjects and electronic instrumentation to protect the subject from electric shock should a failure occur within this instrumentation.

SCOPE

It is the intent of this publication to familiarize engineering personnel with electronic measurements associated with the biophysical sciences. As such, a developed knowledge of electronics and electronic measuring techniques is assumed, but biological and physiological aspects are presented from first principles and simplified to a level consistent with the understanding of the basic principles involved.

This book will be found to differ somewhat from other publications on "medical electronics." Many of these publications are authored by, and directed toward, medical personnel. As such, engineering personnel may find the electronics aspects somewhat oversimplified and the physiological aspects taken for granted. The opposite approach has been taken here. The following books are suggested as general references on Biophysical Measurements. Specific references are given, where appropriate, in the text throughout this book.

Dewhurst, *Physical Instrumentation In Medicine and Biology*, London: Pergamon, 1966.

Dickinson, *Electrophysiological Technique*, London: Electronic Engineering, 1950.

Donaldson, *Electronic Apparatus For Biological Research*, London: Butterworth, 1958.

Geddes and Baker, *Principles of Applied Biomedical Instrumentation*, New York: Wiley & Sons, 1968.

Kay, *Experimental Biology*, London: Chapman and Hall, 1964.

Stacey, *Biological and Medical Electronics*, New York: McGraw-Hill, 1960.

Suckling, *Bioelectricity*, New York: McGraw-Hill, 1961.

Yanof, *Biomedical Electronics*, Philadelphia: Davis, 1965.

Engineers interfacing with the medical profession are encouraged to learn as much as possible about medical and hospital practice and in particular about the physiology of the human body. It is only by gaining such an understanding that they can communicate intelligently with members of the medical profession. One is often approached by a doctor with the request, "can you supply me with this item of equipment?" The tactful but firm reply must be, "what do you want it to do?" Very often the requirement, once understood, can be met far more effectively by some more modern development than that embodied in the original request. In this way a real service to the medical profession is provided and the engineer becomes a colleague rather than a graduate technical assistant.

Information is separated into the following categories:

SECTION I -- Physiology and generation of bioelectric potentials within man.

SECTION II -- Measurement techniques required to perform various biophysical measurement:

SECTION III -- Instrumentation required to implement the measurement techniques covered in Section II.

A reader wishing to study, for example, electrocardiographic techniques should thus first study Section I, Chapter 2, to determine the source of the bioelectric potential referred to as the electrocardiogram; he should then study Section II, Chapter 5, to determine the measurement techniques necessary in recording this electrocardiogram and, finally, he should study Section III to determine the instrumentation necessary to implement these measurement techniques.

Wherever possible, the biophysical measurements covered assume man as the subject. Although, in practice, many of these measurements are performed on laboratory animals, such as rats, cats, dogs, and monkeys, the principles involved and the results obtained relate directly to man. No division is attempted between biophysical measurements performed in a clinical environment and biophysical measurements performed in a research environment as they are, from the electronic engineering viewpoint, essentially similar. In general, only commonly accepted and widely used measurement techniques are covered in this publication as it would be impossible to document the infinite variety of unique, specialized, biophysical measurements being performed by different clinicians and researchers in the biophysical sciences. This publication should not be interpreted as a general text on medical electronics; it is, as its title implies, limited to measurement techniques.

SECTION I

PHYSIOLOGY AND GENERATION OF
BIOELECTRIC POTENTIALS WITHIN MAN

The following four chapters (1-4) cover the basic physiology of the cell, the heart, the muscular system and the brain and the generation of electrical activity within these physiological systems.

The intent of these four chapters is to familiarize engineering personnel with the physiological aspects associated with the various bioelectric generators within the body. In general, the information is presented in the simplest possible form. Thus, while satisfying the intent of this publication, the professional physiologist may find the subject material somewhat oversimplified.

Interested readers will find the following texts invaluable for further study:

Brazier, *The Electrical Activity of the Nervous System*, Baltimore: Williams and Wilkins, 1968.

Eccles, *The Neurophysiological Basis of Mind*, London: Oxford, 1953.

Eccles, *The Physiology of Nerve Cells*, New York: John Hopkins, 1957.

Hoffman and Cranefield, *Electrophysiology of the Heart*, New York: McGraw-Hill, 1960.

Katz, *Nerve, Muscle and Synapse*, New York: McGraw-Hill, 1966.

1

THE CELL AS A
BIOELECTRIC GENERATOR

The cell is the basic source of all bioelectric potentials. A bioelectric potential may be defined as the difference in potential between the inside and the outside of a cell; in other words, the difference in potential existing across the cell wall or membrane. A cell consists of an ionic conductor separated from the outside environment by a semipermeable or selectively permeable cell membrane. Many different types of cells comprise any one species of living matter. Human cells may vary from 1 micron to 100 microns in diameter, from 1 millimeter to 1 meter in length and have a typical membrane thickness of 100 Angstrom units. One micron is equal to 10^{-4} centimeters and one Angstrom unit is equal to 10^{-8} centimeters. Bioelectricity is studied both from the viewpoint of the source of electrical energy within the cell and also from the viewpoint of the laws of electrolytic current flow relative to the remote ionic fields produced by the cell. In electrophysiology we may, in some cases, penetrate a cell to investigate its internal potential. More commonly, however, we make measurements external to a group of cells while these cells are supplying electrolytic current flow.

1.1 THE SOURCE OF INTERNAL CELL POTENTIALS

Experimental investigations with microelectrodes have shown that the internal resting potential within a cell is approximately -90 millivolts with reference to the outside of the cell. This potential changes to approximately +20 millivolts for a short period during cell activity. Cell activity results from some form of stimulation as described later in this chapter. The Hodgkin-Huxley theory, initially postulated during the 1950's, is generally considered to give the best explanation as to the source of these potentials and provides equations that give an emperical mathematical fit to experimental data. This theory is briefly described as follows.

The interior of a cell primarily contains concentrations of sodium and potassium ions. These concentrations within a cell differ markedly from the concentrations of these ions in the space outside the cells (see Fig. 1-1). Elementary ionic theory states that, under suitable conditions, any uneven distribution of ionic concentration in an aqueous solution will result in a potential difference between the regions of different concentration. If, for example, solutions containing unequal concentrations of ions are separated by a membrane semipermeable to these ions, a potential will be found to exist (see Fig. 1-2).

chemical
gradient

This potential, which is referred to later as the chemical gradient, is given by the Nernst relation:

Nernst
relation

$$\text{Potential (mV)} = 61.6 \, \text{Log} \, \frac{\text{Concentration, one side of membrane}}{\text{Concentration, other side of membrane}}$$

Or, for uni-univalent ionic solutions, the Nernst relation simplifies to:

$$\text{Potential (mV)} = 61.6 \, \frac{U - V}{U + V}$$

where   U = Mobility of the negative ions (anions) through the membrane

         V = Mobility of the positive ions (cations) through the membrane

Referring to Fig. 1-2, for a 10:1 activity (concentration) ratio at 37°C, the relative mobilities of the chloride and sodium ions are 65.4 and 43.6 respectively. Applying these values to the Nernst relation gives:

$$\text{Potential (mV)} = 61.6 \, \frac{65.4 - 43.6}{65.4 + 43.6}$$

$$= 12 \, \text{mV}$$

This can be confirmed with a voltmeter as shown in Fig. 1-2. This potential will, of course, run down as diffusion proceeds, unlike that of a living cell.
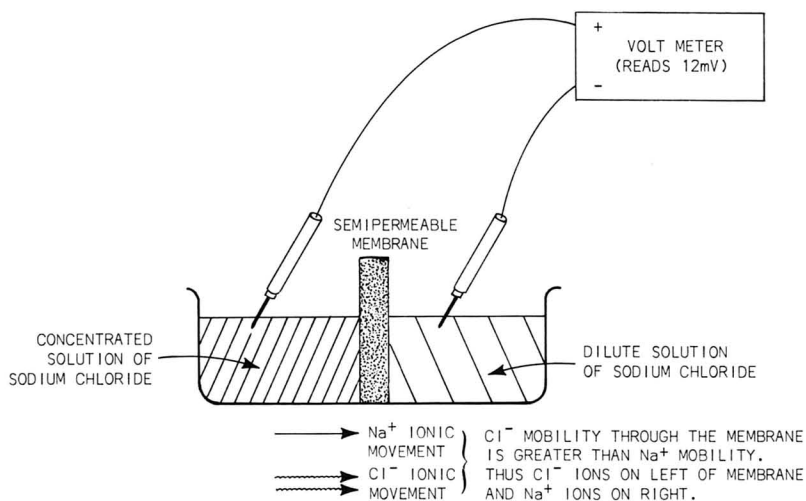
Fig. 1-1. Potential generated by ionic concentration difference between two solutions.
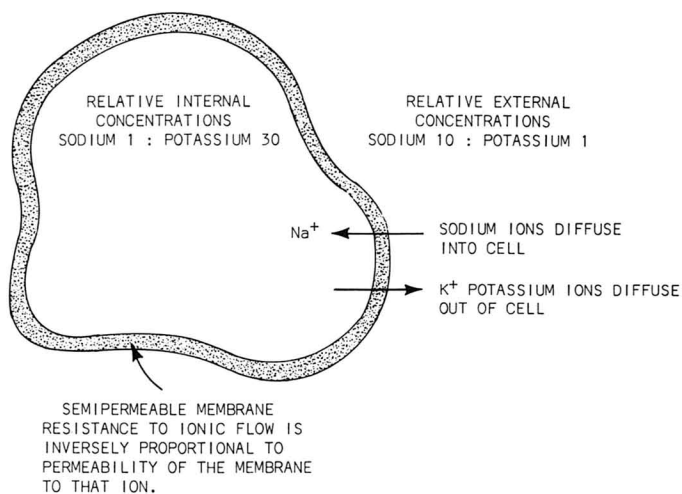
Fig. 1-2. Typical concentrations of sodium and potassium ions within a cell.

MICROELECTRODE
INTERNAL TO
CELL

CELL
MEMBRANE

VOLT METER
(READS -90mV)

INSIDE
OF CELL

ELECTRODE EXTERNAL
TO CELL

LARGE NET
SODIUM GRADIENT
ACROSS MEMBRANE

SODIUM IONIC CURRENT
LOW PERMEABILITY

SMALL NET
POTASSIUM GRADIENT
ACROSS MEMBRANE

POTASSIUM IONIC CURRENT
MEDIUM PERMEABILITY

SODIUM IONIC CURRENT PRODUCED
BY A LARGE NET Na$^+$ GRADIENT
AND LOW PERMEABILITY IS BAL-
ANCED BY A SMALLER NET K+
GRADIENT AND A HIGHER
PERMEABILITY.

(A) IONIC CURRENTS INVOLVED IN A POLARIZED CELL

MICROELECTRODE
INTERNAL TO
CELL

CELL
MEMBRANE

VOLT METER
(READS +20mV)

INSIDE
OF CELL

ELECTRODE EXTERNAL
TO CELL

LOW NET
SODIUM GRADIENT
ACROSS MEMBRANE

SODIUM IONIC CURRENT
HIGH PERMEABILITY

LARGE NET
POTASSIUM GRADIENT
ACROSS MEMBRANE

POTASSIUM IONIC CURRENT
MEDIUM PERMEABILITY

SODIUM IONIC CURRENT PRODUCED
BY A SMALL NET Na$^+$ GRADIENT
AND HIGH PERMEABILITY IS BAL-
ANCED BY THE POTASSIUM IONIC
CURRENT PRODUCED BY A LARGE
NET K+ GRADIENT AND MEDIUM
PERMEABILITY.

(B) IONIC CURRENTS INVOLVED IN A DEPOLARIZED CELL

Fig. 1-3. Cell ionic currents.

The ionic current produced by ion movement through
a semipermeable membrane depends on the permeability
of the membrane and also on the "gradient" that
net forces the ion through the membrane. This gradient,
gradient referred to as the net gradient, consists of both
a chemical gradient and an electrical gradient. A
chemical gradient is formed due to a difference in
concentration producing a potential gradient as given
by the Nernst relation. An electrical gradient is
formed as a result of a potential that may exist
across the membrane due to some other source.

Experimental investigations have shown that a marked
difference in concentrations of both sodium and
potassium ions exists across a cell membrane. In
mammalian nerve cells as shown in Fig. 1-1, the
cell concentration of potassium ions is in the vicinity
concentra- of 30 times higher inside the cell than in the
tions fluid external to the cell. On the other hand,
sodium ions are approximately 10 times more
concentrated in the fluid external to the cell than
in the fluid within the cell.

Consider a cell in its resting or polarized state
(Fig. 1-3A). In this state the membrane is
moderately permeable to potassium ions, that is,
cell in potassium ions can pass fairly readily through the
resting membrane as the membrane offers medium resistance.
state This membrane is, however, almost impermeable to
sodium ions and, thus, offers a high resistance to
the passage of these ions. A large net gradient
affects the movement of sodium ions into the cell.
This net gradient consists of a chemical gradient
produced by the 10-to-1 concentration difference
between sodium ions on each side of the membrane and
a 90 mV electrical gradient produced by the standing
net potential within the cell. The net gradient
gradient affecting the movement of potassium ions out of the
cell is considerably less than the net sodium
gradient. This gradient consists of a large
chemical gradient due to the 30-to-1 concentration
difference across the membrane; however, this
chemical gradient is opposed by the electrical
gradient produced by the 90 mV standing potential
within the cell. Thus, although the membrane is
almost impermeable to sodium ions, the net sodium
gradient is high. Conversely, although the membrane
is moderately permeable to potassium ions, the net
potassium gradient is low. The net result is that

12

-90 mV
resting
level
polarized
cell

the sodium and potassium currents are equal; the
sodium current balances the potassium current with
a resultant current of zero. Since the net current
through the membrane is zero, the cells internal
potential will not change and will remain at its
-90 mV resting level. Indeed, this -90 mV resting
level is determined by the internal cell potential
required for sodium and potassium current balance.

cell after
stimulus

When the cell receives a stimulus from an outside
source, the characteristics of the membrane at the
point of stimulation will be markedly altered and,
thus, the ionic currents will also change. After
stimulation, the membrane permeability to potassium
ions is unaltered but the permeability to sodium ions
is increased. A much lower resistance is offered to
the flow of sodium ions, thus increasing the sodium
ionic current. This increased sodium ionic current
causes more positive ions to pass into the cell than
are passing out of the cell, causing the internal
cell potential to drop from -90 mV in an attempt to
achieve sodium current and potassium current balance.
As this potential decreases, the net sodium gradient
across the membrane decreases and the net potassium

depolari-
zation

gradient across the membrane increases, causing the
currents to decrease and increase, respectively.
This process continues until current balance is
again obtained, at which time the internal cell
potential is +20 mV. The cell is then referred to
as being in a depolarized state.

repolari-
zation

By the time the cell has fully depolarized the
characteristics of the membrane have begun to revert
back to their prestimulus state. This causes the
sodium ionic current to be considerably lower than
the potassium ionic current; the internal cell
potential thus begins to go negative with the
process continuing until the -90 mV resting
potential of the cell is once again obtained.

An electrical analogy to a cell membrane is shown in Fig. 1-4. This circuit cannot strictly be referred to as an equivalent circuit as the electronic current flow in an electrical circuit and the ionic current flow through a cell membrane cannot be said to be equivalent. After assigning resistance values inversely proportional to the relative permeability of the membrane and assuming potassium and sodium concentration ratios, then the intracellular potential for both a polarized cell and a depolarized cell can be determined. The values assumed are analogous to actual values found in a cell.

cell
electrical
analogy
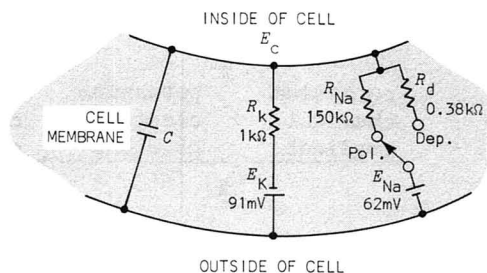
E_c = INTRACELLULAR POTENTIAL WITH RESPECT TO THE OUTSIDE OF THE CELL.

E_K = NERNST POTENTIAL DUE TO THE POTASSIUM ION CONCENTRATION DIFFERENTIAL ACROSS THE MEMBRANE

E_{Na} = NERNST POTENTIAL DUE TO THE SODIUM ION CONCENTRATION DIFFENRENTIAL ACROSS THE MEMBRANE.

R_K = RELATIVE PERMEABILITY OF THE MEMBRANE TO THE FLOW OF POTASSIUM IONS THROUGH IT.

R_{Na} = RELATIVE PERMEABILITY OF THE MEMBRANE TO FLOW OF SODIUM IONS THROUGH IT WHEN THE CELL IS POLARIZED.

R_d = RELATIVE PERMEABILITY OF THE MEMBRANE TO THE FLOW OF SODIUM IONS THROUGH IT WHEN THE CELL IS DEPOLARIZING.

C = CAPACITY OF THE CELL.

Fig. 1-4. An electrical analogy to a cell membrane.

Assume:
  Relative values for R_K, R_{Na} and R_D are 1 kΩ,
  150 kΩ and 0.35 kΩ respectively.
  Potassium ion concentration ratio of 30:1 inside
  to outside.
  Sodium ion concentration ratio of 10:1 outside to
  inside.

Then:

$$E_K = 61.6 \text{ Log} \frac{30}{1} = 91 \text{ mV} \quad -- \text{ by Nernst relation.}$$

$$E_{Na} = 61.6 \text{ Log} \frac{10}{1} = 62 \text{ mV in opposite polarity}$$

$$\text{to } E_K \quad -- \text{ Nernst.}$$

For a polarized cell:
  net potassium current + net sodium current = 0

$$\frac{\text{net potassium gradient}}{R_K} + \frac{\text{net sodium gradient}}{R_{Na}} = 0$$

$$\frac{\begin{array}{c}\text{potassium} \\ \text{chemical} \\ \text{gradient}\end{array} + \begin{array}{c}\text{potassium} \\ \text{electrical} \\ \text{gradient}\end{array}}{R_K} + \frac{\begin{array}{c}\text{sodium} \\ \text{chemical} \\ \text{gradient}\end{array} + \begin{array}{c}\text{sodium} \\ \text{electrical} \\ \text{gradient}\end{array}}{R_{Na}} = 0$$

$$\frac{E_K + E_c}{R_K} + \frac{-E_{Na} + E_c}{R_{Na}} = 0$$

$$\frac{91 \times 10^{-3} + E_c}{1 \times 10^3} + \frac{-62 \times 10^{-3} + E_c}{150 \times 10^3} = 0$$

Solving:
  $E_c = 90 \times 10^{-3} = -90$ mV (polarized)

Similarly, for a depolarized cell, R_{Na} is replaced
by R_D.

  $E_c = +20$ mV (depolarized)

If a microelectrode were inserted into the cell as
shown in Figs. 1-3A and 1-3B and a stimulus were
applied to the cell, the output of the
microelectrode would appear as shown in Fig. 1-5.
cell action   This waveform is known as the "cell action
potential   potential." It should be noted that the currents
involved in bioelectricity are unlike the currents
involved in electronics. Bioelectric currents are

due to positive and negative ion movement within a
conductive fluid. As these ions possess finite mass
and encounter resistance to movement within the
fluid their speeds are limited. The cell action
potential, thus, shows a finite "risetime" and
"falltime."

sodium-
potassium
pump

The ionic concentration gradient across the cell
membrane is maintained by virtue of metabolic
energy expended by the cell in "pumping" ions
against the ionic gradient formed by the differing
ionic concentrations between the inside and outside
of the cell. This action has been referred to as
the "sodium-potassium pump."
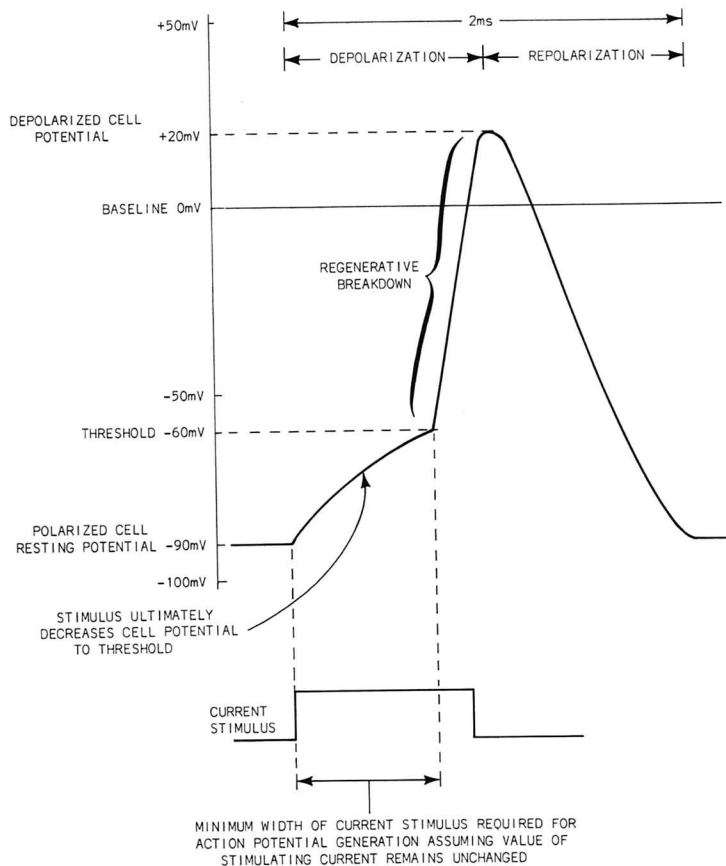
Fig. 1-5. Cell action potential (internally recorded with microelectrode).

1.2 CELL STIMULATION AND STIMULUS THRESHOLD

A cell may be stimulated, or caused to depolarize and then repolarize, by subjecting the cell membrane to an ionic current. This current may be produced by other cells, it may be produced by ionic currents existing as nerve impulses, or it may be artifically produced by some external current stimulus. A cell will be stimulated when sufficient positive ions are added to the inside of the cell to cause its resting potential to be decreased from its -90 mV level to approximately -60 mV. Once this threshold level is reached, the cell depolarizes without requiring the addition of any further positive ions to the inside of the cell

stimulus
threshold

from the stimulus source. Unless a stimulus above a certain minimum value is received, known as the stimulus threshold, the cell will not be depolarized and no action potential will be generated. The stimulus required to exceed the threshold is a function of both current and time; the threshold may be exceeded by a short, high-current pulse or by a longer, lower-current pulse (see Fig. 1-5).

refractory
period

Since the energy associated with the action potential is developed from metabolic processes within the cell itself and not from the stimulus, a finite period of time, known as the refractory period, is required for metabolic processes within the cell to return the cell to its prestimulus state. This refractory period has been observed in most cells found in the nervous system. Closer study reveals that the refractory period has two parts: The first in which no stimulus, however strong, will cause depolarization (the absolute refractory period) and the second when depolarization occurs only if the stimulus is of more than normal threshold strength (the relative refractory period).

1.3 CURRENT FROM A SINGLE CELL AND THE RESULTANT EXTERNALLY RECORDED ACTION POTENTIAL

We have previously discussed the depolarizing and repolarizing action of cells and the resulting potential existing within the cell. As stated earlier, this potential may be recorded with a microelectrode; however, in most bioelectric

external
electrodes

measurements, this potential is recorded by
electrodes external to the actual cell. These
external electrodes typically would record the net
action of many hundreds of cells, but for the time
being, we will consider only a single cell. When
recording with external electrodes, an action
potential is produced between these electrodes
during periods of current flow; that is to say, no
potential exists when cells are either in their
depolarized or repolarized state. A potential
exists only while the cell is changing from one
state to another. As the external action potential
is generated by the external current that flows
during cell activity, the shape of the action
potential is related to the variation of this current
with time.

external
action
potential

The external potential field rises to its maximum
value sometime during the regenerative breakdown
phase of the membrane. The external action
potential that is recorded from the cell is somewhat
similar to a mathematical time derivative of the
transmembrane potential. This potential is detected
with maximum amplitude when one electrode is placed
as near as possible to the active area and the other
electrode is located in a completely inactive or
remote area. It is detected with reduced amplitude
as the electrodes are placed closer to each other
so they intercept smaller elements of potential
difference.

depolari-
zation
current

Consider a single polarized cell; the inside of the
cell is negative with respect to the outside
environment which may be regarded as a reference.
As stated previously, the net ionic current flow
across the membrane is zero; thus, ionic current
flow to and from the cell is zero. Should these
conditions be altered due to the presence of a
stimulating current through the cell membrane, then
regenerative membrane breakdown will occur and the
cell will depolarize. During the depolarization
process the net current through the membrane is not
zero; there is a net positive ionic current into the
cell through the cell membrane. This current may be
detected as a potential difference between two
electrodes placed in the vicinity of the cell with
the potential difference being produced across the
finite resistance of the fluids external to the cell.

This current continues until the cell is fully depolarized, thus, the potential will appear between the two electrodes while the cell is undergoing depolarization. For the short time that the cell exists in a depolarized state, the net current through the membrane is once again zero, thus, no potential will appear between the electrodes.

Almost immediately after depolarization the cell begins to repolarize again. During repolarization, positive ionic current flows from the cell membrane, that is, in the opposite direction to ionic current flow during depolarization. During this period of current flow, a potential will be detected between the electrodes and, since this current is in the opposite direction to the depolarization current, the potential produced between the electrodes

repolari-
zation
current
during repolarization will be of the opposite polarity to the potential produced during depolarization. The area included by the depolarization and the repolarization potential waveforms is the same since the quantity of current involved in each process is the same. It should be noted that the process described above is somewhat theoretical as it describes the external action potential generated by a single cell. In the following discussion we show that single cell depolarization invariably results in depolarization of the adjacent cells and, hence, this externally recorded action potential will be the net summation of the results obtained from these cells.

1.4 EXTERNALLY RECORDED ACTION POTENTIAL FROM A GROUP OF CELLS, THE TRAVELLING WAVE OF DEPOLARIZATION

The preceding discussion must be modified to allow for many cells in close proximity to one another and to allow for the appreciable length of many of these cells. Consider a group of cells in close proximity to one another as shown in Fig. 1-6 and 1-7. Under certain conditions of stimulation these cells may all depolarize at the same time

synchronous
depolari-
zation
(synchronous depolarization); however, the repolarization process is random. Repolarization of the individual cells will occur at different times. The resultant externally recorded action potential is shown in Fig. 1-6. Once again, the area included under each wave is the same since the quantity of current involved in each process is the same.

STIMULATION OF ALL CELLS
WITHIN A GROUP OF CELLS

STIMULUS

INDIVIDUAL ACTION POTENTIALS
OF THE 6 CELLS

SUMMATION OF
INDIVIDUAL ACTION
POTENTIALS 1 THROUGH 6

AREA A = AREA B

TIME

DEPOLARIZATION   REPOLARIZATION
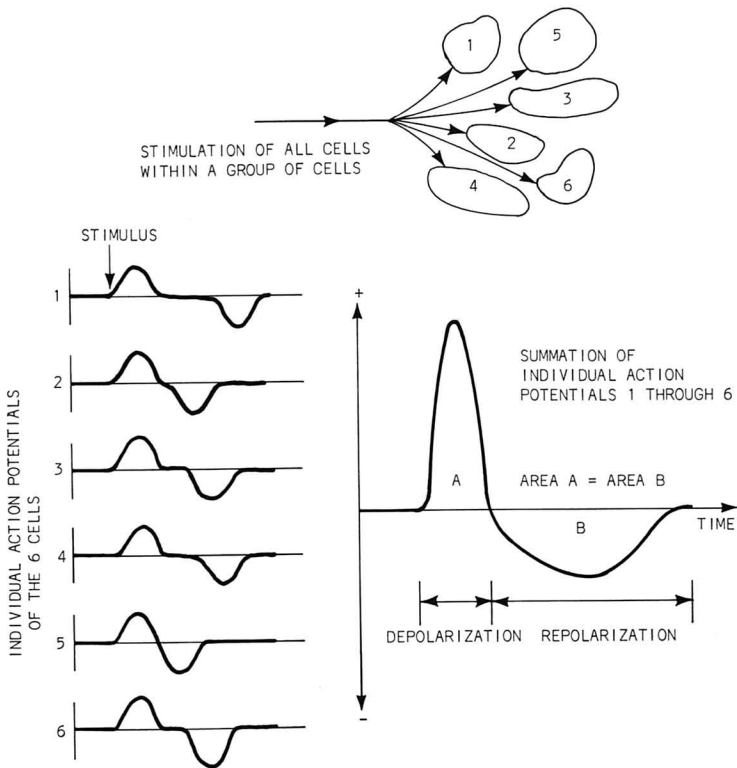
Fig. 1-6. External action potential produced by a group of cells all
depolarizing as a result of one stimulation (synchronous
depolarization).
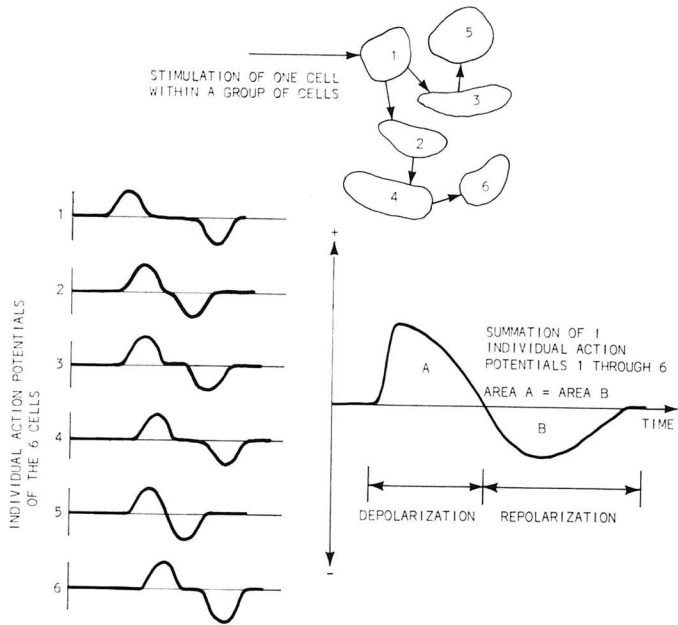
Fig. 1-7. External action potential produced by a group of cells depolarizing due to stimulation by adjacent cells (asynchronous depolarization).
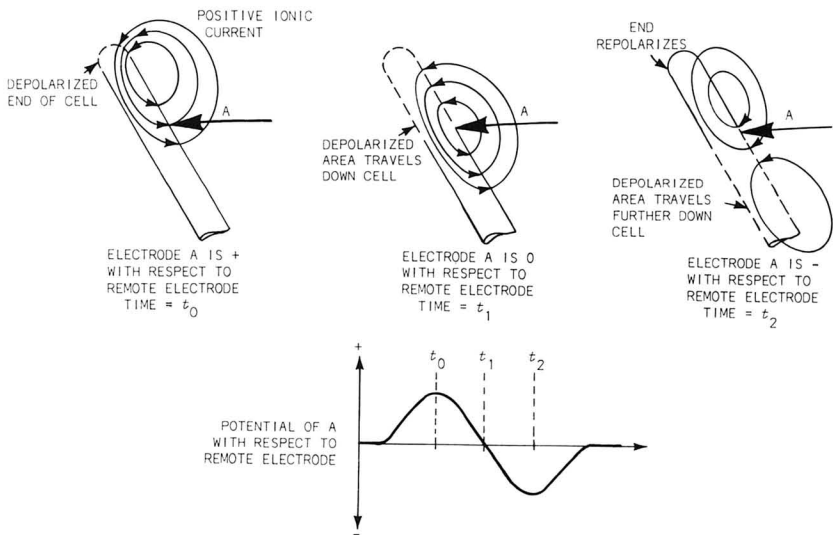
Fig. 1-8. "Traveling wave of depolarization" resulting from depolarization of successive section of a long cell.

<p>asynchronous depolari-zation</p>

Under other conditions of stimulation, the group of cells described previously will not all depolarize at the same time (asynchronous depolarization). The stimulation may result in one cell depolarizing; the action of this cell depolarizing will then act as a stimulus on its adjacent cell causing it to depolarize also. This chain reaction would proceed until all cells in a particular area have depolarized. The resultant externally recorded action potential would appear as shown in Fig. 1-7. In practice, combinations of synchronous and asynchronous depolarization occur in a group of cells.

<p>travelling wave within cells</p>

In the same way as the depolarization of one cell causes adjacent cells to depolarize, depolarization of a localized area of an individual cell will cause depolarization of other parts of the same cell. Thus, the depolarization process will appear to travel along the length of the cell causing a travelling wave of depolarization as shown in Fig. 1-8.

As most bioelectric potentials are recorded as external cell action potentials, the results obtained are a summation of the action of many cells. The action potential waveform may be modified by the number of cells, the shape of these cells, and the type of stimulation applied.

2

THE HEART AND THE
CIRCULATORY SYSTEM

myocardial
infarction

Coronary heart disease is the greatest single cause
of death in most developed countries. In the United
States alone, over a million persons die every year
from heart disease; the most frequent cause of
death is the acute myocardial infarction. Thus, a
major percentage of medical effort, both in the
research and clinical fields, is directed toward
understanding of cardiovascular system and the
subsequent prevention of heart disease.

2.1   THE CARDIOVASCULAR CIRCULATORY SYSTEM

cell
revitali-
zation

In a complex multicellular animal with a high
density of cell population and a relatively small
volume of tissue fluid bathing these cells, a
continuous and rapid revitalization process must
take place. This revitalization is the fundamental
purpose of the cardiovascular circulatory system.
The previous chapter discussed the potential
produced within a single cell. This potential
exists as a result of continuous metabolism within
the cell. This metabolism process requires
nutrients and excretes waste products; the
circulatory system provides these nutrients and
removes these waste products.

The major component of the circulatory system is the
heart which supplies the power required to circulate
blood throughout the body. The heart is two pumps
in series; the smaller, right-hand section provides
the power required to force blood through the lungs,
and the larger, more powerful left-hand section
provides the power required to force blood through
the body. A simplified block diagram is shown in
Fig. 2-1.
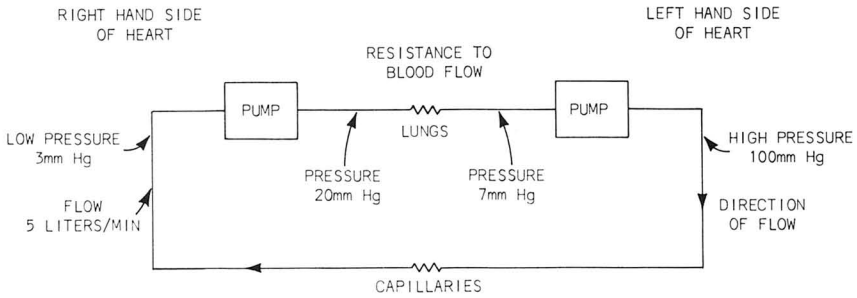
Fig. 2-1. Simplified block diagram of the circulatory system.
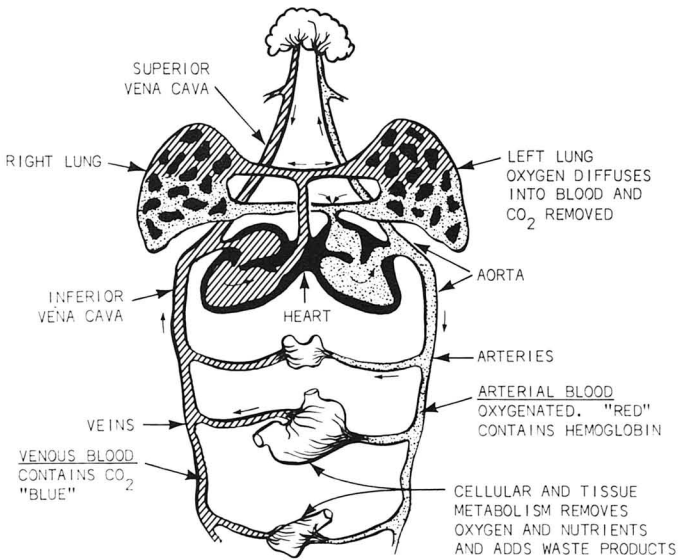
Fig. 2-2. The cardiovascular circulatory system.

blood flow

Elaborating on Fig. 2-1 and referring to Fig. 2-2, the blood flows from the heart into the aorta. The aorta curves in an arch up from the heart, down along the back bone and into the abdomen; from it other large arteries lead to the head, the digestive organs, the arms and the legs. From these arteries branch the smaller arterioles (or small arteries) and from these, branch billions of tiny capillaries. By the time the blood has reached the capillaries it is moving slowly along channels that are only about 10 microns in diameter. Here the blood discharges its load of dissolved food and oxygen to the body cells. These cells in turn deposit waste materials such as carbon dioxide into the blood stream. In yielding oxygen and taking on waste, the blood turns color from bright red to dull red or "blue." The blood now starts back to the heart passing from the capillaries into the venules. The venules converge into larger veins and then into the two largest veins just above and below the heart known as the vena cava. The blood empties into the right atrium, is pumped into the right ventricle and then moves out through the pulmonary artery to the lungs. The lungs supply the blood with fresh oxygen. The blood passes from the lungs to the left atrium, then is pumped into the left ventricle and passes out via the aorta to begin the circulation process again. This general flow throughout the body is known as the "systemic circulation;" the flow to and from the lungs is known as the "pulmonary circulation." Local circulations within the systemic system include the renal (to the kidneys), the hepatic portal (to the liver), the cerebral (to the brain) and the coronary (to the heart itself). The waste products contained in the blood are removed by the kidneys and liver (see Fig. 2-2). The average quantity of blood in man is about five liters and is completely circulated through the body in approximately one minute.

2.2 THE HEART

The heart itself weighs less than a pound, is about
six inches long at its maximum dimension and lies
pointed downward in the chest cavity to the left
of the mid-center body line. The walls of the heart
are made entirely of muscle; within these walls are
four hollow chambers, a left and a right receiving
chamber (atrium) and below them a left and a right
pumping chamber (ventricle).

The cardiac cycle is characterized by the following
mechanical events. Between beats, the heart
mechanically rests and this is known as the period

cardiac   of diastole. During diastole the heart assumes its
cycle   maximum size and fills with oxygenated blood
diastole   returning from the lungs and venous blood returning
             from the body. The heart's period of mechanical
systole   activity is known as systole. The onset of systole
is initiated by contraction of the muscles
surrounding the atria which propels additional blood
into the ventricles. The ventricles then begin to
contract, thereby causing a rise in pressure within
the ventricles. This increased pressure shuts the
two atrioventricular valves and, with further
contraction, the pressure continues to rise. Once
the pressure of the systemic and pulmonary
circulations are exceeded, a phase of ventricular
ejection is begun. The aortic valve is forced open
and blood is squeezed into the aorta and thence into
the systemic circulation. Likewise, the pulmonary
valve is forced open and blood is supplied to the
pulmonary circulation. After the ventricular
contents are partially ejected, the muscles
surrounding the ventricles relax and the
ventricular pressures fall. As soon as these
pressures fall below the pressures sustained in the
circulatory systems, the aortic and pulmonary
valves close, signalling the onset of diastole.

2.3 ELECTRICAL POTENTIALS GENERATED WITHIN THE HEART - GENERATION OF THE ELECTROCARDIOGRAM WAVEFORM (ECG)

The preceding section covers the mechanical activity
of the heart and states that mechanical activity is
initiated by contraction of the muscle surrounding
the atria. The detailed relationship between cell,
nerve and muscle producing this contraction is
covered in Chapter 3. It is sufficient at this
stage to state that muscle contraction is initiated
by stimulation.
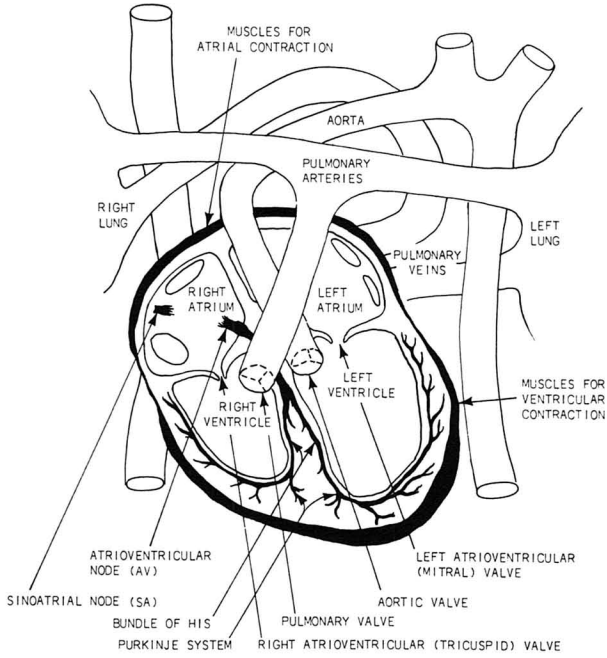
Fig. 2-3. The heart.
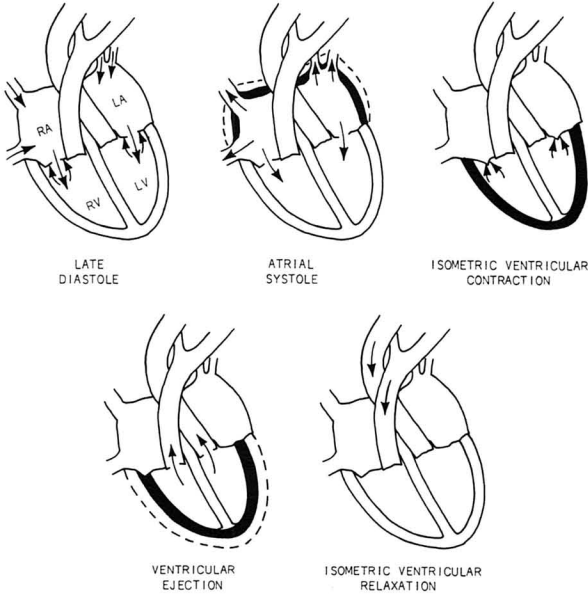
Fig. 2-4. Mechanical cycle of the heart.

MUSCLE
ACTION

DIASTOLE = "RELAXATION"
SYSTOLE = "CONTRACTION"

ATRIAL
SYSTOLE

VENTRICULAR
SYSTOLE

VENTRICULAR
DIASTOLE

120mm Hg

ARTERIAL
PRESSURE

ATRIAL
DIASTOLE

ARTERIAL
SYSTOLIC
PRESSURE

100mm Hg

ARTERIAL
DIASTOLIC
PRESSURE

80mm Hg

100 90 80   70   60   50   BEATS/min

R

ECG   0.5mV   80ms

20mm/mV
50mm/s

P   T

TA   U

Q S   U WAVE
UNKNOWN ORIGIN

P WAVE – ATRIAL DEPOLARIZATION

TA WAVE – ATRIAL REPOLARIZATION MAY OCCUR

PR INTERVAL – ATRIO-VENTRICULAR
CONDUCTION TIME 0.12-0.22s

QRS SEGMENT – VENTRICULAR DEPOLARIZATION   0.1s MAX

T WAVE – VENTRICULAR REPOLARIZATION

ST SEGMENT – ISO ELECTRIC

QT INTERVAL –

Fig. 2-5. The ECG waveform and related heart action.

sinoartrial
node

The right atrium contains a bundle of nerves known
as the sinoatrial node (abbreviated SA node). This
type of nerve cell is found nowhere else in the body.
Its function is to start the heart beat and set its
rhythm or pace. The electrical and mechanical
output from the heart is initiated by stimulation
from this node which results in contraction of the
various heart muscles. Although the basic rhythm of
the heart is self-sustaining through synchronization
from the SA node within the heart, this rhythm is
modified by certain nerve fibers external to the
heart that affect the SA node. These nerves have a
function in the normal control of the heart rate to
respond to increased or decreased demand for blood
by the body.

atrio-
ventricular
node

muscle
action
sequence

Impulses generated by the SA node stimulate
contraction of the muscles comprising the atria.
These impulses also travel along conducting fibers
in the atrium to the atrioventricular node or AV
node, stimulating depolarization of this node
(Fig. 2-3). Stimulation of the atrioventricular
node causes impulses to be sent to the myocardium or
muscle comprising the ventricules via the bundle of
His and the Purkinje conducting system resulting in
contraction of this muscle. Thus, the muscular
contractions necessary to maintain the heart's
pumping action are initiated by depolarization and
repolarization of the SA node and then
depolarization and subsequent repolarization of the
AV node.

These depolarizations and repolarizations generate
external action potentials which can be recorded at
the surface of the body as covered in detail in
Chapter 1. These external action potentials
generated from within the heart are known as the
electrocardiogram or ECG. It is common, also, to
refer to this waveform as the EKG, derived from the
German spelling of electrocardiogram. The ECG
waveform (Fig. 2-5) is shown in relationship to the
mechanical action of the heart and the resultant
arterial pressure.

Electrical activity of the heart is, as stated
earlier, initiated by depolarization of the SA node
and a resulting contraction of the muscles
surrounding the atria. The resulting external action
potential is known as the P wave. Immediately
following this depolarization, repolarization of the
atria occurs; however, for some reason, this does
not generate a pronounced action potential. This
potential is known as the TA wave and is rarely
observed in practice. Electrical activity produced
by depolarization of the SA node travels through
fibers within the atrium to the AV node. The time
taken for this electrical stimulation to travel
from the SA node to the AV node is known as the
atrioventricular conduction time and is typically
between 120 ms and 220 ms. When this stimulation
reaches the AV node, this node depolarizes and the
depolarization is conducted down through the bundle
of His to the myocardium muscle causing ventricular
depolarization. The external action potential is
referred to as the QRS complex. Immediately
following this depolarization the cells concerned
repolarize which results in ventricular
repolarization or the T wave. Many ECG waveforms
also show an additional wave occurring after the T
wave. This is designated the U wave and its origin
is unknown.

The ECG waveform is recorded at maximum potential
when one electrode is placed slightly above the
heart and to the right and the other electrode is
placed slightly below the heart and to the left;
thus, the potential output of the heart can be said
to be generated along this axis. Chapter 5
discusses variations in the ECG waveform with
different electrode positions.

3

MUSCLE ACTION —
AND THE SENSORY SYSTEM

Fundamental relationships exist between muscle
function, the sense receptors, the brain, the
nervous system and the peripheral nerves. The
discussion of muscle function that follows includes
a discussion of the physiology of the nervous system
and the peripheral nerves. The brain is, at this
stage, regarded as a processor and is considered in
more detail in Chapter 4.

3.1 THE MOTOR UNIT

motor
end-plates

The motor unit, as the name implies, is the
biological unit of muscle function. A motor unit
consists of a motor nerve arising from motoneurons
in the brain-stem or spinal cord and branching into
various motor end-plates. These motor end-plates
are each connected to an individual muscle fiber;
stimulation of these motor end-plates causes
contraction of the single muscle fiber attached to
it, as shown in Fig. 3-1. The number of motor units
varies between the different muscles of the body;
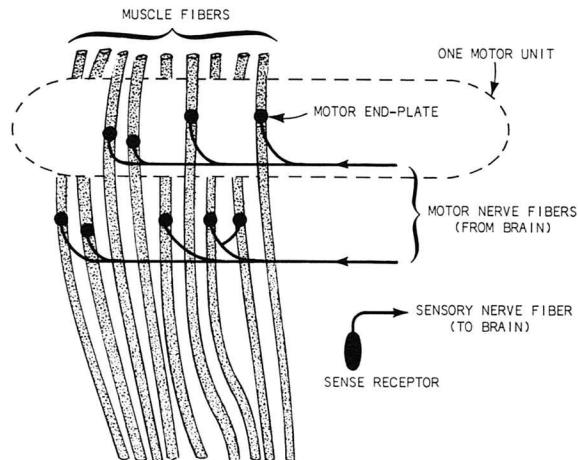
Fig. 3-1. Relationship between nerve and muscle.

generally speaking, the larger the muscle, the more
motor units will be found in that muscle. The size
of the motor unit, that is, the number of muscle
fibers activated by the same nerve fiber, can be
quite different for different muscles; in man, one
motor unit may contain from 25 to 2000 muscle fibers.
The force developed by a motor unit may range from
0.1 to 250 grams weight. The muscle fibers of a
motor unit are not clumped together in one part of
the muscle, but rather the muscle fibers of
different units are interlaced as shown in Fig. 3-1.

3.2 MUSCLE ACTION

Chapter 1 stated that a cell can only exist in its
polarized or depolarized state; that is, it is a
bistable device and intermediate potential levels
are not stable. Motor nerves are also cellular in
nature, thus, any individual motor nerve can only
exist in a polarized or depolarized state and will
transmit only two potential levels to the motor
end-plates causing a bistable "on-off" action of
the muscle fibers. Thus, the individual muscle
fibers of one motor unit can exist in only two
states, a relaxed state and a tensed state. Normal
movement   muscular activity is characterized by smoothness
smoothness   of movement, steadiness and precision. These
characteristics are due to the large number of motor
units comprising any one muscle. If a small
muscular effort is required, only one motor unit
will be called into action; as increasing muscular
effort is required, many more motor units are called
into action until the muscle is providing maximum
effort at which time all motor units connecting to
this muscle are being used. In this way, some
smoothness of movement is obtained.

Additional smoothness of movement is obtained by
modulating the number of muscle fiber contractions
per unit time. Although an individual motor unit
can result in only one level of muscular
contraction, the number of times that this
muscle   contraction occurs per unit time (the number of
modulation   depolarizations and repolarizations executed by the
motor end-plate cells) will effectively increase the
power of these muscle fibers. Thus, the smoothness
of movement of a muscle is controlled both by the
number of motor units activated and by the rate at
which these motor units are being activated.

FEEDBACK FROM SENSE RECEPTOR CONTROLS
MUSCLE AS IN SERVO SYSTEM
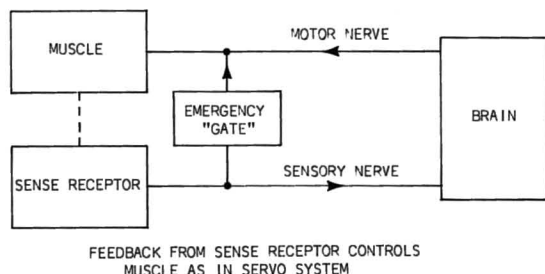
Fig. 3-2. Block diagram of the nervous system.

3.3 THE MUSCULAR SERVO-MECHANISM

brain
control

A highly simplified block diagram of the nervous
system controlling muscle action is shown in
Fig. 3-2. The system is similar to a servo-
mechanism control system: A sense receptor or
transducer produces a position or velocity signal
which is sent to the brain via the sensory nerve.
The brain in turn initiates an "error" or control
signal by comparing the measured position with the
desired one stored in the memory. This signal is
sent via the motor nerve to the muscle to control
its action.

150 ms
reflex
response

This servo-system can be demonstrated by the
following experiment. When one's finger is placed
onto a cool object, the sense receptor in the finger
senses the temperature and relays this information
to the brain. The brain interprets this signal as
coming from a cool object and, thus, does not
necessarily initiate any signal to the motor nerves.
If the finger is then placed on a warm object, the
brain will interpret the information received from
the sensory nerves as relating to a warm object
and will activate the motor nerves controlling the
muscles in the arms and the hand causing the finger
to be lifted from the warm object. There is a time
delay of several hundred milliseconds between the
time that the sense receptor feels the warm object
and the time that the finger is lifted from the
warm object. This delay is governed largely by the
degree of attention that the subject is paying to
the warm object. Now, if the finger is placed on a
hot object, a reflex response is obtained and the
finger is removed from the hot object in about
150 ms. This reflex is active at all times although
it is most marked for the hot object.

3.4 REFLEX RESPONSE

emergency
gate

An emergency gate has been shown in Fig. 3-2; this
emergency gate is not markedly involved in normal
receptor/muscle operation. When a reflex response
is called for, however, this emergency gate bypasses
the signal path to and from the brain and initiates
a reflex response. This emergency gate is usually
located within the spinal cord. Thus, a reflex
response results from a "large" signal, i.e., a
high repetition rate signal, being received from a
sense receptor. This signal bypasses the brain in
the initiation of muscle action. These reflex
responses protect the body from serious damage.

3.5 THE POTENTIAL GENERATED DURING MUSCLE ACTION

Depolarization initiated within a sense receptor
travels to the brain along the sensory nerve fiber
as a series of traveling depolarization waves. The
brain then initiates another series of traveling
waves of depolarization along the motor nerves to
cause a series of depolarizations of the motor end-
plates. Depolarization of the motor end-plates
depolarizes cells within the muscle fiber causing
contraction of these fibers. The actual internal
cell potentials involved here are the normal cell
polarized potential of -90 mV and the normal cell
depolarized potential of +20 mV.

recording
action
potential

In dealing with muscles and nerves, it is unusual to
use microelectrodes to record the action potential
within individual cells. More commonly, needle
electrodes are used to record the net result of a
number of cells such as one motor unit, or surface
electrodes are used to record the results of many
motor units. If a microelectrode were inserted into
a muscle cell to observe the depolarization and
repolarization process, the total process would
occur in less than one millisecond. If, however, a
needle electrode is placed near these muscle cells,
it will detect current flow from many fibers of the
corresponding motor unit. These fibers are being
fired at their motor end-plates at practically the
same instant by the branching nerve. The different
fibers of a motor unit do not, however, develop their
action currents simultaneously; small time
variations between the different fibers occur. These

varying delays are due to the varying lengths of the terminal branches between the motor nerves and the muscle. Thus, the excitation as it travels along is slightly ahead in some fibers compared with others. The result is that current flow in any small area from the cells comprising one motor unit lasts from 2 to 5 milliseconds, which is several times the duration of the current from any single muscle fiber. This asynchronous action helps produce smoothness of muscle action.

With anything other than microelectrodes, the basic electrical activity detected in any muscle is the single motor-unit action potential. These potentials from single motor units are best detected by concentric-needle electrodes or by insulated-needle electrodes as shown in Fig. 3-3. Single motor unit activity may sometimes be detected by placing a small electrode on the skin over the muscle. The volume of muscle influencing such an electrode is, however, considerable and the actual recording is typically the total activity from many motor units having random relationships to one another. The potential produced by muscle action, whether recorded by needle electrodes or by surface electrodes, is known as the

EMG   electromyogram or the EMG. Measurement techniques for electromyographic recordings are covered in Chapter 12.
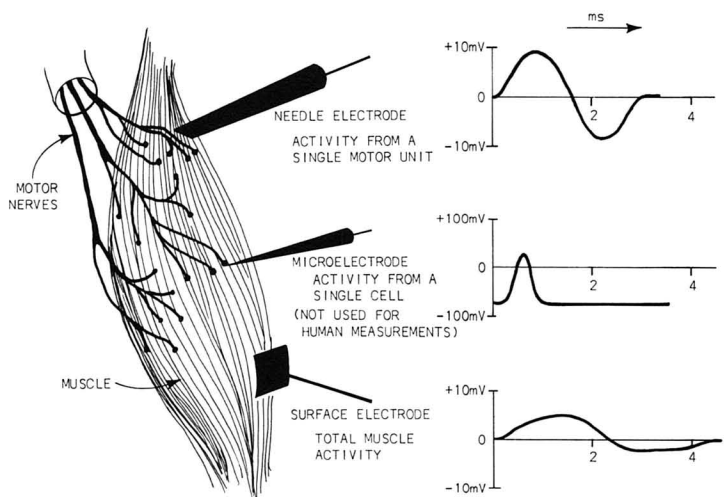
Fig. 3-3. The EMG obtained with various electrode types.

3.6 THE SENSE RECEPTORS

logarithmic
response

If a sense receptor cell or "transducer cell" is activated by a physical distortion, a temperature change, or some other effect, a series of nerve impulses are produced. The time between the individual nerve impulses become greater and greater as various factors come into play. After a while, despite continuation of the physical stimulus, no nerve impulses are generated. It is believed that the sense receptors generate discrete potential levels which are more or less logarithmically related to the stimulus strength. These potentials in turn produce a series of nerve impulses in the nerve fibers. Impulse repetition frequencies from sensory elements vary from a few impulses per second to about 1000 per second; rates above 50 per second are most unusual in human subjects. Since the sense receptors generate an approximate logarithmic response to the stimulus, they can respond over an enormous range ($1:10^7$ or more) of stimulus energy levels.

3.7 THE POTENTIAL GENERATED BY SENSE RECEPTOR STIMULATION

detecting
sense
receptor
activity

It is almost impossible to detect the electrical activity associated with most of the sense receptors on the body; they are small and are not located in clusters of sufficient size to allow detection of the electrical activity associated with a group of receptors. Most sense receptors are not electrical in nature; we measure the electrical activity associated with them only because we do not have the techniques available to us which enable us to observe underlying biochemical mechanisms. There are two groups of sensors that do produce detectable electrical activity: The middle ear and the retina. Potentials associated with the hearing mechanism and the sight mechanism can be detected in the middle ear and on the retina, respectively, because a large number of sensory cells are packed closely together and can be stimulated simultaneously. Only by such an arrangement will enough depolarization current be produced to make detection possible.

ear
response

When a sound reaches the ear, it causes vibration of a membrane within the inner ear known as the basilar membrane. Vibration of this membrane in turn stimulates a large number of sense receptors known as hair cells. If the sound reaching the ear is fairly loud, a large number of these hair cells will develop action potentials at the same time and the current produced will be strong enough to be detected in the middle ear. Electrical potentials precisely following the shape of the stimulating sound waves can be detected by an electrode in the middle ear. This part of the inner ear acts as a transducer and generates electrical activity from sound.

electro-
retinogram

If a bright light is projected onto a substantial area of the retina, many light-sensitive cells within the retina will be stimulated simultaneously and will develop a considerable synchronous response which can be detected from the outside of the retina. The electroretinogram (the external electrical response to light stimulation) may be detected by an electrode consisting of a small, flat silver plate fitted to the inner surface of a small contact lens as discussed in Chapter 11, Section 11.8.
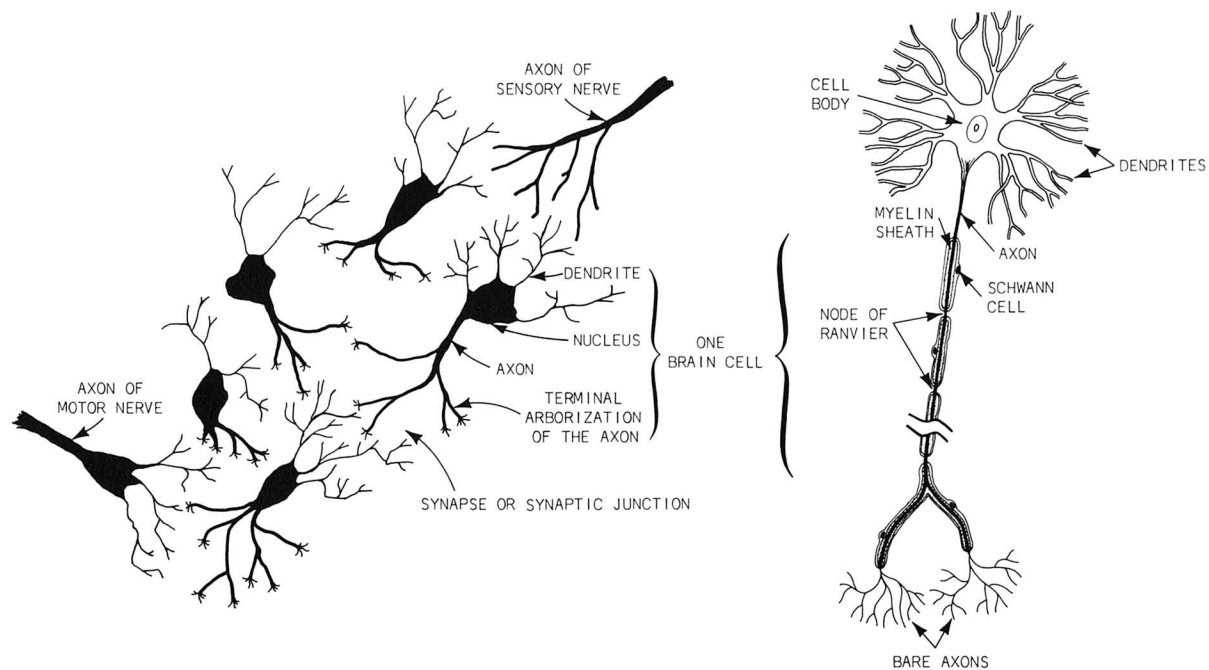
Fig. 4-1. Brain cell interconnections — highly simplified.

4

THE BRAIN AND THE
CENTRAL NERVOUS SYSTEM

Chapter 3 discussed sense receptors, muscle action
and the potentials associated with the peripheral
motor and sensory nerves. To simplify the
presentation, Chapter 3 referred to motor and
sensory nerves as connections to the brain; strictly
speaking, however, most of these nerves connect to
spinal nerves in the spinal cord and these *spinal*
nerves carry information to and from the brain.

4.1 NERVE CELLS IN THE CENTRAL NERVOUS SYSTEM AND IN THE BRAIN

cell
components

Although nerve cells are located throughout the body
and carry information between all body locations,
most of the nerve cells are located in the brain
and in the spinal cord. The cells comprising the
spinal cord and the brain are similar; the cell body
is known as the soma and surrounds a region known
as the nucleus. The cell body has an extension
known as the axon and this may comprise part of the
spinal cord and interconnect with sensory and motor
nerves or may branch into many small fibers known
as the terminal arborizations of the axon. Brain
cells consist also of projections of the cell body
known as the dendrite which sense information from
adjoining cells. The terminal arborizations
transmit information to adjoining cells either
directly in spinal cells or via the dendrite in
brain cells. These cell interconnections (synaptic
junctions or the synapse) allow electrical impulses
to flow throughout the brain and the central nervous
system; one cell acting as a trigger to influence
neighboring cells. See Fig. 4-1.

The mechanism by which information is carried is exceedingly complex. As we saw in Chapter 1, the migration of potassium and sodium ions is a primary factor. However, there are many other chemical interchanges involved which cannot be discussed in this text which is specifically concerned with electrical phenomena. It must be emphasized, however, that we measure the electrical activity only because of its *convenience*; it is *not* the fundamental form of nervous activity.

4.2 THE BRAIN

Only a very cursory treatment of the brain is given in this chapter. It is a complex structure, comprised of very large numbers of nerve cells which are interconnected among themselves and which also receive data from the various sensory organs. In a

control unit

loose sense it can be called a supervisory control unit for it can override most reflexes and direct the body to act in a coordinated manner. Until recently it was thought that most of the functions of the brain resulted from the interaction of nervous impulses at the synapses of the cells (as discussed in Section 4.3) and many analogies with digital electronic computers were formulated. It is now known that this is only part of the story. For example, the substance in which the cells are

neuroglia

embodied, the neuroglia (from a Greek word meaning glue), is known to play an important part in brain function and may be vitally concerned with memory. It is thought that oscillatory electrochemical

memory

mechanisms are responsible for short term memory but that these impulses gradually modify the chemical structures of parts of the brain so that the permanent storage of information is chemical rather than electrical.

Anatomically, the brain is divided into several parts with the general appearance shown in Fig. 4-2. When sectioned along the midline, the appearance is as in Fig. 4-3. Speaking generally we may say that the deeper structures of the brain, that is those nearest the spinal cord, are those which evolved first and are largely responsible for the more primitive and less easily controlled parts of human behavior (anger, fear, etc.). The cortex (outer structure) is the part of the brain which is highly developed in man and which has enabled him to dominate all other species.
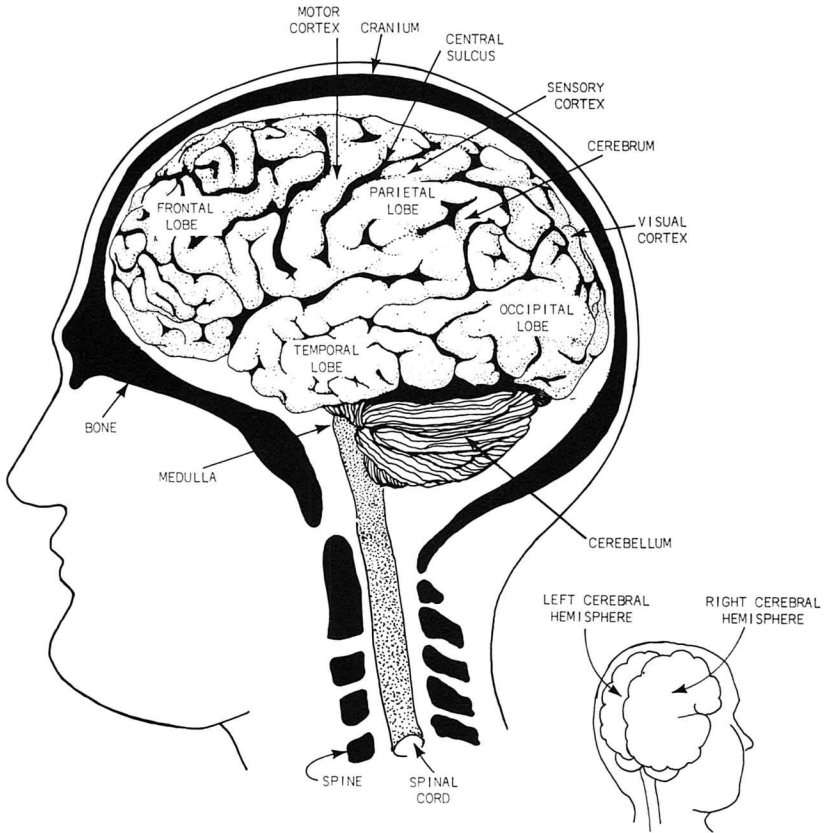
Fig. 4-2. Anatomical sections of the brain.
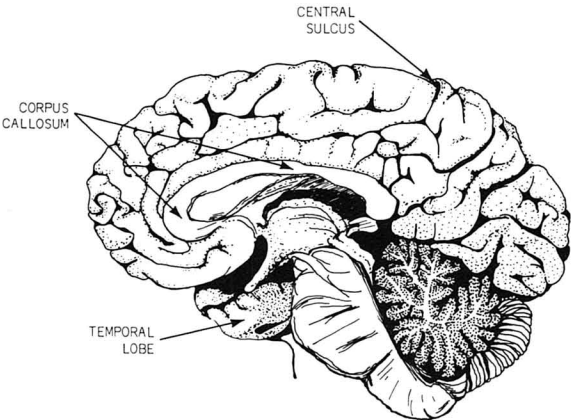
Fig. 4-3. Section of the brain.

Some areas of the cortex serve specific parts of the body. For example, in Fig. 4-2, sensory input is handled in the area marked sensory cortex while motor output proceeds from the motor cortex. Likewise, vision is handled at the rear of the brain in the visual cortex. It is less easy to define areas which serve intellectual functions although the frontal areas are at least in part responsible. There is a large bundle of fibers which interconnect the left and right hemispheres; this is known as the corpus callosum and its location can be seen in Fig. 4-3.

4.3 EXCITATION AND INHIBITION POTENTIALS

The following discussion is a simplified analysis of excitation and inhibition from one scientific school of thought. The properties of excitation and inhibition are, however, highly conjectual and, as stated previously, tell only part of the story. There is no general scientific agreement on the subject.

cell
potentials

In Chapter 1, cells were referred to as bistable devices existing in only two states; a polarized state of -90 millivolts and a depolarized state of +20 millivolts. Intermediate potentials were regarded as having little significance. Nerve cells in the central nervous system and the brain exist in a polarized state of between -70 millivolts and -110 millivolts and the exact cell potential within this range appears to be significant as it determines the cell's vulnerability to a stimulus that would result in regenerative breakdown.

Consider a group of cells producing zero activity; all cells within this group would rest at the normal polarized level of -90 millivolts. Assuming that one cell is depolarized from an external stimulus, this depolarization will affect adjoining cells via synapses, causing a change in the resting potential of these adjoining cells. The effect on these adjoining cells can be either to raise or to lower their resting potential depending on the type of fiber involved in the synaptic junction. If the resting potential is increased, breakdown of the cell membrane will not occur and the impulse can be

inhibitory effect
thought of as having had an inhibitory effect. If the impulse arriving at this synapse decreases the resting potential, the cell is then more susceptible to depolarization and the impulse can be thought of as having had an excitatory effect. With repeated

excitatory effect
excitatory stimuli at a sufficiently high rate, the membrane will eventually fully depolarize and a new impulse will be propagated. Any single cell is influenced by synapses from many other cells; some are excitatory and some inhibitory. The play of these two opposite effects on any area may or may not cause depolarization of the cell; the deciding factor is the number of impulses being received per unit time and the balance between excitation and inhibition. Although the effect at a synapse of any subthreshold impulse dies away rapidly, it does not die away immediately. This allows almost coincident impulses from different sources or a rapid rate of impulses from a single source to build

frequency is a parameter
up their effect; thus frequency becomes an important parameter. Fig. 4-4 shows the effect of various excitation and inhibition signals on any one individual cell, finally resulting in depolarization of the cell once the -70 millivolt threshold is obtained.
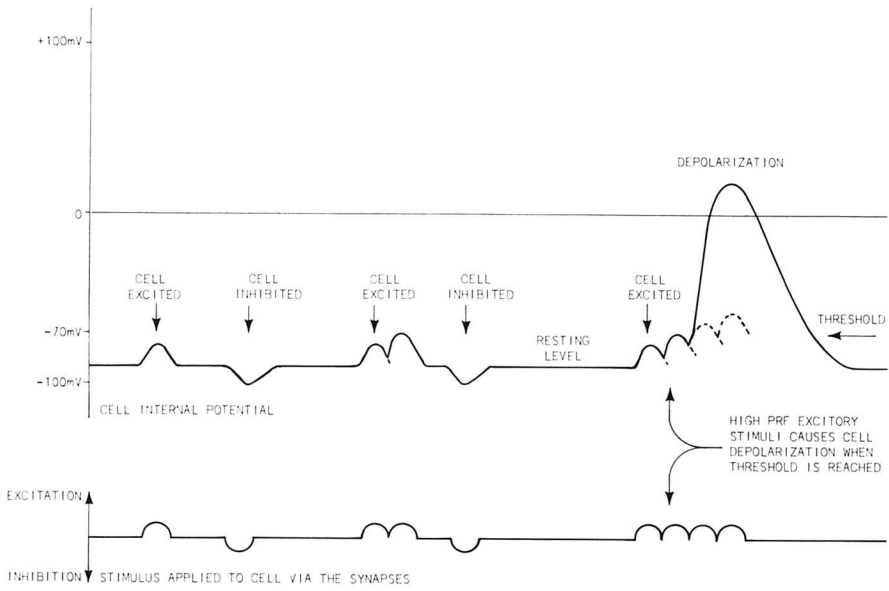
Fig. 4-4. Stimulus changing cell resting potential may result
in depolarization of the cell.
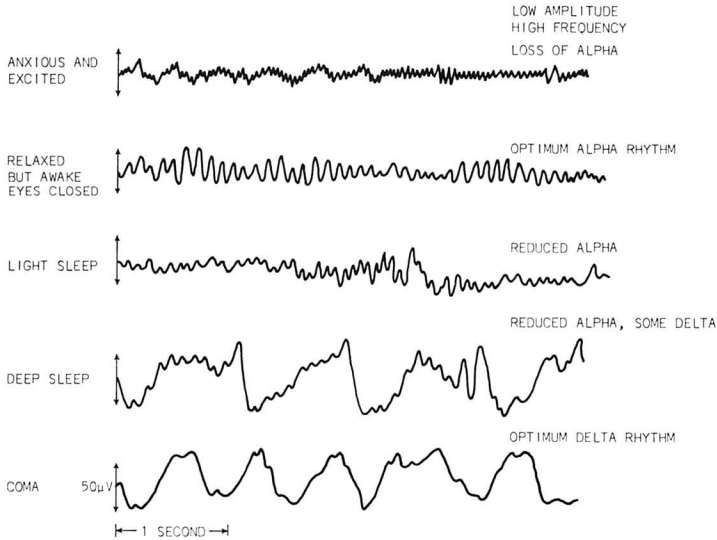
Fig. 4-5. Physiological states and the resultant EEG.

4.4 EVOKED POTENTIALS

producing
evoked
potentials
As shown in Fig. 4-2, parts of the surface of the brain have been found to be associated with various sensory systems. If an appropriate stimulus is applied to a sense receptor, the corresponding sensory area of the brain responds by producing an electric potential known as the evoked potential. The evoked potential, as it appears at the surface of the brain, is the integrative result from the action of many cells. When recorded externally on the scalp, the evoked potential is of the order of 10 μV or so. It is not yet known completely which cortical elements are responsible for these evoked potentials. Often they are camouflaged by the electroencephalogram, thus it may be necessary to remove the electroencephalogram by an averaging technique when attempting to record the evoked potential. Averaging is discussed in more detail in Chapter 11.

4.5 THE ELECTROENCEPHALOGRAM (EEG)

alpha and
delta
rhythms
As briefly discussed in Section 4.2, the brain can be regarded as a highly developed biochemical factory. The electrical activity which results from so much chemical change is known as the Electroencephalogram (EEG, see Chapter 10) and is in a sense a useful byproduct of nervous action since it allows us to make nonmutilating measurements on an organ which is singularly resentful of external interference. The electroencephalogram, as recorded from the surface of the head, consists of rhythmical, slow sinusoidal waveforms between 10 and 100 microvolts in amplitude. The electroencephalogram varies in both form, amplitude and frequency; the basic frequency of around 10 hertz is known as the alpha rhythm. Should a subject lapse into deep sleep, the alpha rhythm will also disappear and will be replaced by a slower high-amplitude signal known as the delta rhythm. The electroencephalogram produced by a subject in various conditions is shown in Fig. 4-5.

SECTION II

MEASUREMENT TECHNIQUES

The following chapters (5-14) cover some of the more
common measurement techniques used in the
biophysical sciences. It is intended that this
material be used by engineering-oriented personnel
as a basis for developing measuring techniques
that are uniquely suited to particular requirements.
As these requirements are defined by biomedically
oriented personnel who will operate the measurement
systems, cooperation between the engineer and these
personnel is necessary for refinement of the
techniques. A working knowledge of electronic
measurement techniques is assumed; this material
simply applies standard electronic measurement
techniques to physiological situations. Typical
results are shown for normal subjects and little
attempt has been made to define the limits of this
normality or to present results that may represent
abnormality.
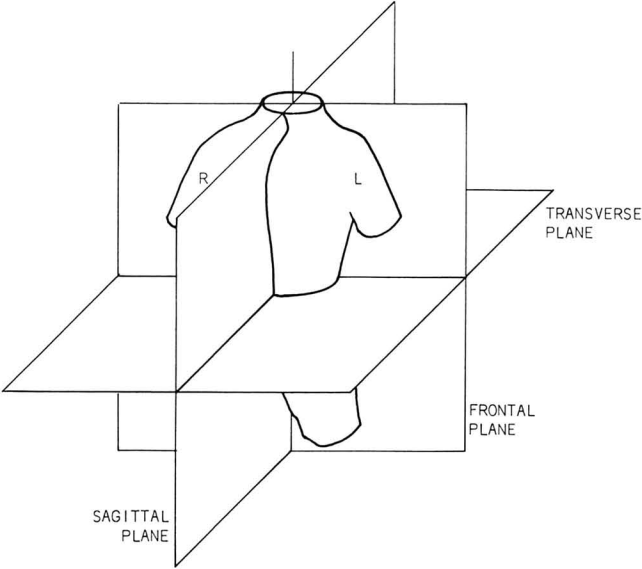
Fig. 5-1. Electrocardiographic planes.

5

ELECTROCARDIOGRAPHY (ECG)

Generation of electrical activity within the heart
is discussed in Chapter 2. Electrocardiography is
the art of analyzing this electrical activity by
measuring potentials at the surface of the body
resulting from this electrical activity within the
heart. This is achieved by applying electrodes to
certain positions on the body and recording the
potentials generated between various combinations
of these electrodes with an amplifier and CRT
display or strip chart recorder.

5.1 ELECTROCARDIOGRAPHIC PLANES

measuring
cardiac
potential

The heart can be regarded as an electrical generator
enclosed in a volume conductor, the torso. As this
electrical generator is completely enclosed by the
torso, a direct measurement of the generator output
voltage is impossible without resorting to surgery.
The electrocardiographer measures the potential
existing between various points on the surface of the
volume conductor and uses the information obtained
to determine the clinical condition of the heart.
The art of electrocardiography is simplified
somewhat by considering that at any one time the
cardiac potential is projected along axes existing
on each of three reference planes: The Frontal
plane, the Transverse plane and the Sagittal plane,
as shown in Fig. 5-1. This projection of the
cardiac potential is referred to as the
electrocardiogram (abbreviated ECG or EKG). Various
techniques are used to measure the projection of the
cardiac potential along axes existing on each of the
three planes; these techniques are considered
separately.

5.2 FRONTAL PLANE ECG MEASUREMENTS

cardiac
vectors

The electrical potential generated within the heart as projected along axes existing on the frontal plane of the body is known as the frontal-plane cardiac vector, as shown in Fig. 5-2. As with vector measurements in the physical sciences, the relative amplitude and angular position of this vector at any instant cannot be measured with a single measurement. Two separate measurements are required and the results of these two measurements are plotted on a vector diagram to determine the relative amplitude and angular position of the vector. In the physical sciences the two measurements usually determine the projection of the vector along two axes at 90° to one another. In electrocardiography the projection of this vector is recorded along two axes at 60° to one another as it allows the limbs to be used for the attachment of electrodes; the results obtained are relatively independent of where on the limbs the leads are placed.

Einthoven
triangle

cardiac
vector at
R wave peak

Although only two measurements are theoretically necessary to determine the relative amplitude and angular position of the frontal plane vector, it is common electrocardiographic practice to record at least three projections of the frontal plane vector along three axes at 60° to each other. The triangle formed by these three axes is known as the Einthoven triangle as shown in Fig. 5-2. Willem Einthoven, a Dutch physiologist, pioneered many of the electrocardiographic techniques in use today and during the first quarter of this century he developed this vector approach to electrocardiography. Einthoven's law states that the vector sum of the projections of the frontal plane cardiac vector at any instant onto the three axes of the Einthoven triangle will be zero. This is a well-known physical law; however, Einthoven realized that it did in fact also apply to cardiology. Although the above applies to the cardiac vector at *any* instant, cardiologists are usually only interested in the cardiac vector at the peak of the R wave. Thus when the term cardiac vector is used, it implies "Cardiac vector at the peak of the R wave."

Fig. 5-2. The Einthoven triangle.

Fig. 5-3. Einthoven's triangle assuming a homogeneous torso, and Burger's triangle with the effects of the lungs and spine.

More recent advances in electrocardiography have pointed out the well known, but previously ignored, fact that the human torso is neither homogeneous nor triangular and that this leads to a distortion of the electrical field. Thus, the angle of the vector deduced from Einthoven's triangle would be in error. Burger attempted to allow for the inhomogeneities of the torso by introduction of the distorted triangle shown in Fig. 5-3 which compensates for the effect of the lungs and the spine. Although Burger's triangle is perhaps a more accurate representation of the frontal-plane cardiac vector, it is not widely used and cardiologists still prefer Einthoven's approach. They realize that Einthoven's vectors are only an approximation but use them largely for convenience. The clinical interpretation of ECG's is quite empirical in practice, being done by reference to the enormous number of records which have been correlated with known cardiac disorders, usually at autopsy.

Burger's
distorted
triangle

5.3   BIPOLAR LIMB LEAD FRONTAL-PLANE ECG MEASUREMENTS

The three potential measurements commonly used to determine the frontal plane vector in conjunction with the Einthoven triangle are:

1. Potential between the right arm and left arm.

2. Potential between the right arm and left leg.

3. Potential between the left arm and left leg.

These three ECG measurements are known as the standard frontal-plane or bipolar limb lead measurements and are more commonly referred to as Lead I ECG, Lead II ECG, and Lead III ECG as shown in Fig. 5-4.
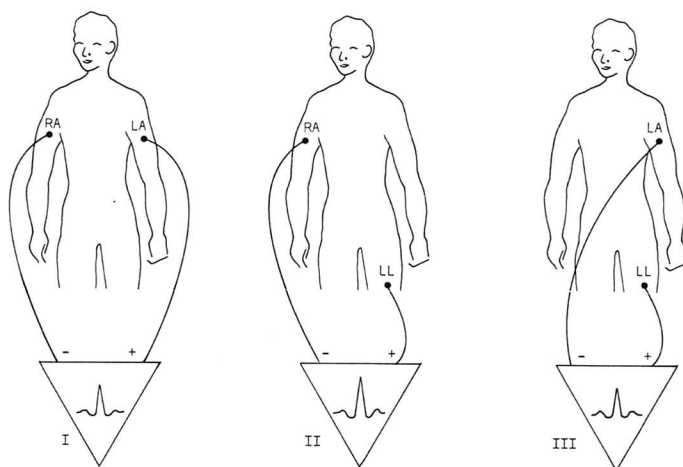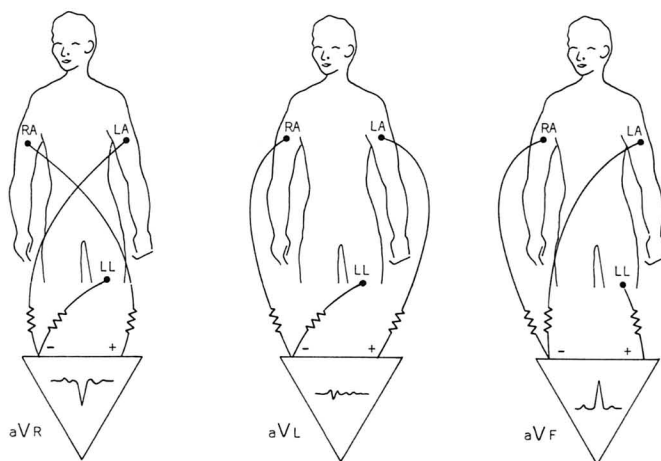
Fig. 5-4. Electrode positions — frontal plane ECG.

5.4 UNIPOLAR LIMB LEAD FRONTAL-PLANE ECG MEASUREMENTS

The unipolar limb-lead ECG measurements are a somewhat recent addition to electrocardiography and permit the unipolar registration of the electrical potential in each of three extremities (the left arm, the right arm and the left leg) by creating an "indifferent electrode" or "central terminal" by combining the signal obtained from the other two extremities. The three unipolar limb-lead

augmented
vectors

measurements are referred to as augmented vector right, augmented vector left and augmented vector foot (abbreviated aVR, aVL and aVF, respectively). Augmented measurements provide the same waveshape but 50% more potential output than the now unused, nonaugmented, unipolar limb-lead measurements, VR, VL and VF. The VL, VR and VF measurements were outdated and replaced by aVL, aVR and aVF when recording equipment sensitive to voltage changes rather than current changes became available, as shown in Fig. 5-5. The unipolar limb-lead electrode configurations are simply a projection of the same frontal-plane cardiac vector onto three different 60° axes rotated 30° from the Einthoven configuration, as shown in Fig. 5-6. The three unipolar leads bear a direct vector relationship to the three bipolar standard limb leads.

$$aVR = -\frac{I + II}{2} \qquad aVL = \frac{I - III}{2} \qquad aVF = \frac{II + III}{2}$$

aVR, aVL,
aVF

The aVR unipolar measurement refers to the potential at the right arm using the left arm and left leg to form the indifferent electrode. The aVL measurement refers to the potential at the left arm using the right arm and left leg to form the indifferent electrodes and aVF measurement refers to the potential at the left leg using both arms to form an indifferent electrode. It should be noted that in these three measurements the indifferent electrode is formed at the negative input of the amplifier in all cases. The ECG waveform will thus be positive for aVL, positive for aVF, but negative for aVR. Fig. 5-4 shows the electrode positions for recording all frontal plane ECG's.

OLDER, NONAUGMENTED UNIPOLAR CONFIGURATION:
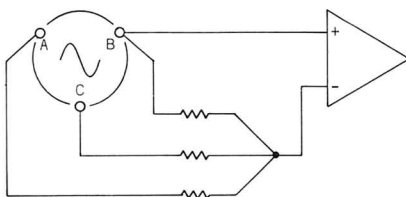
$$\begin{array}{l}\text{POTENTIAL INPUT}\\\text{TO AMPLIFIER}\end{array} = B - \left(\frac{A + B + C}{3}\right) = \frac{2B - A - C}{3}$$

AUGMENTED UNIPOLAR CONFIGURATION:
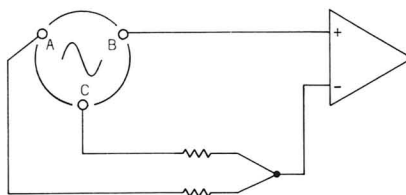
$$\begin{array}{l}\text{POTENTIAL INPUT}\\\text{TO AMPLIFIER}\end{array} = B - \left(\frac{A + C}{2}\right) = \frac{2B - A - C}{2} \left\{\begin{array}{l}\text{SAME ABC RATIO BUT}\\\text{50\% MORE POTENTIAL}\\\text{THAN NONAUGMENTED}\end{array}\right.$$

Fig. 5-5. Equivalent circuit of unipolar measurements.

THE ALGEBRAIC SUM OF THE R AND S WAVE FOR EACH OF THE SIX FRONTAL PLANE ELECTRODE CONFIGURATIONS IS TRANSPOSED TO THIS VECTOR DIAGRAM CONSISTING OF TWO 60° TRIANGLES.

ALGEBRAIC SUM IS A - B
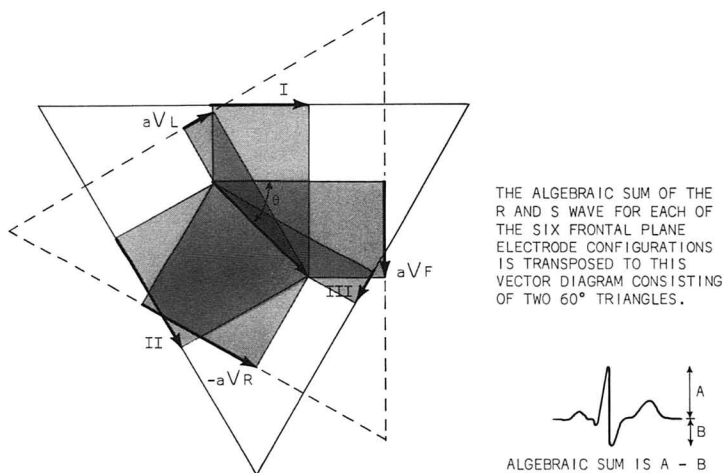
Fig. 5-6. Relationship of frontal plane ECG's to the frontal plane axes of the QRS complex.
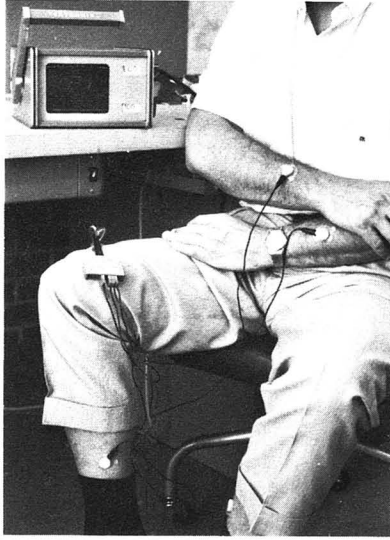
Fig. 5-7. Extremity electrodes: for fully dressed subjects,
somewhat "noisy" at times, used in surgery.
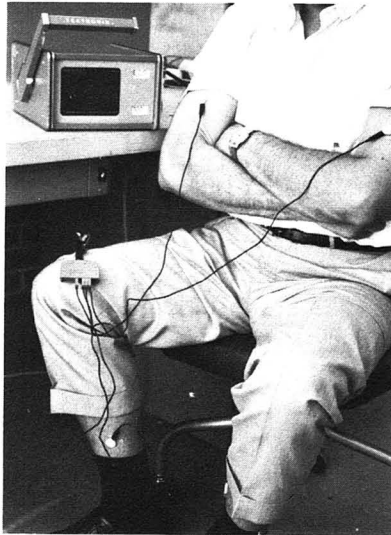
Fig. 5-8. Shoulder electrodes: preferred in most applications,
noise free, used in surgery.

5.5 FRONTAL-PLANE AXIS DERIVATION

As stated earlier, only two ECG measurements are necessary to determine the axes of the frontal-plane cardiac vector. It is, however, common to record all six frontal-plane ECG measurement configurations (I, II, III, aVR, aVL, and aVF). A trained electrocardiologist utilizes all six recordings when analyzing the ECG waveform as each ECG recording will differ in shape as well as in amplitude; certain complexes within the ECG waveform are more clearly visible in one configuration than in another.

cardiac
vector
versus
P, QRS and
T waves

To this stage we have simply referred to the electrical output of the heart when recording frontal-plane ECG's as "the frontal plane cardiac vector." This electrical activity is, however, composed of three discrete waveforms as discussed in Chapter 2. These are the P wave, the QRS complex and the T wave. The relative amplitudes of each of these waveforms can be measured separately from the six standard frontal plane ECG measurements and plotted on a vector diagram to determine the electrical axis of a particular segment of the frontal-plane cardiac vector. There is usually a difference of 30° or more between the frontal plane axis of the P wave, the frontal plane axis of the QRS complex and the frontal plane axis of the T wave (see Chapter 6, Vectorcardiography).

5.6 FRONTAL-PLANE ELECTRODE POSITIONING

electrode
placement

Hitherto, the three electrode positions used to record the frontal-plane ECG have been referred to as right arm, left arm and left leg. In actual practice the electrodes can be placed anywhere on the limb or, in fact, anywhere in the general area on the torso near the limb. When recording the ECG of a fully dressed subject, as is usually the case in a doctor's clinic, it is convenient to use the wrists and left ankle as shown in Fig. 5-7. This configuration does restrict movement. Thus, if the subject is to move his body or if the electrodes are to be left in place for several hours, it is more convenient to place the arm electrodes on the upper arms near the shoulders as shown in Fig. 5-8.

If the subject has no upper garment, they are placed
on the shoulder blade. Accurate results are also
obtained with the electrodes placed in an Einthoven
triangle configuration on the chest as shown in
Fig. 5-9. This "chest cluster" configuration is
used extensively for subject monitoring in intensive
care wards and when the subject is exercising during
recording. The right leg electrode shown in Fig. 5-7,
5-8 and 5-9 is a ground electrode as discussed in
Section 5-9.

"noise" on
waveform

In general, the closer the electrodes are to the
heart the better, as any muscle between the
electrodes produces "noise" on the ECG waveform.
Thus if possible, it is desirable to use the chest
configurations to avoid the "noise" generated by
the powerful arm muscles. A typical series of
frontal-plane ECG measurements is shown in Fig. 5-10.
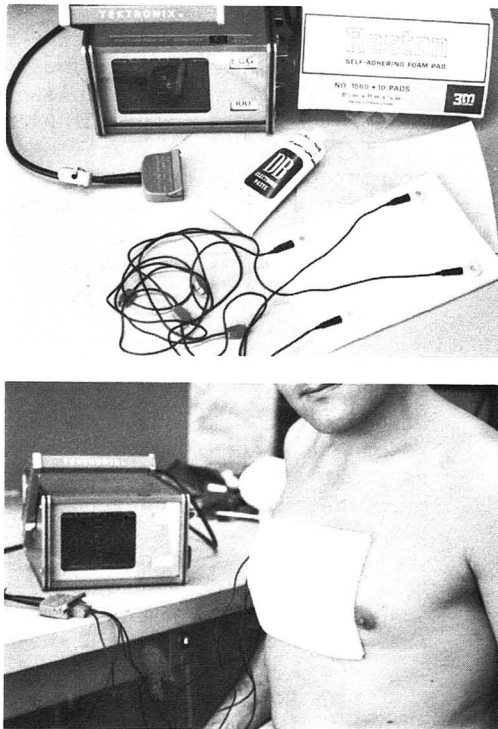
Fig. 5-9. Chest cluster electrodes: for intensive care or exercising
subject, difficult to apply in many cases.
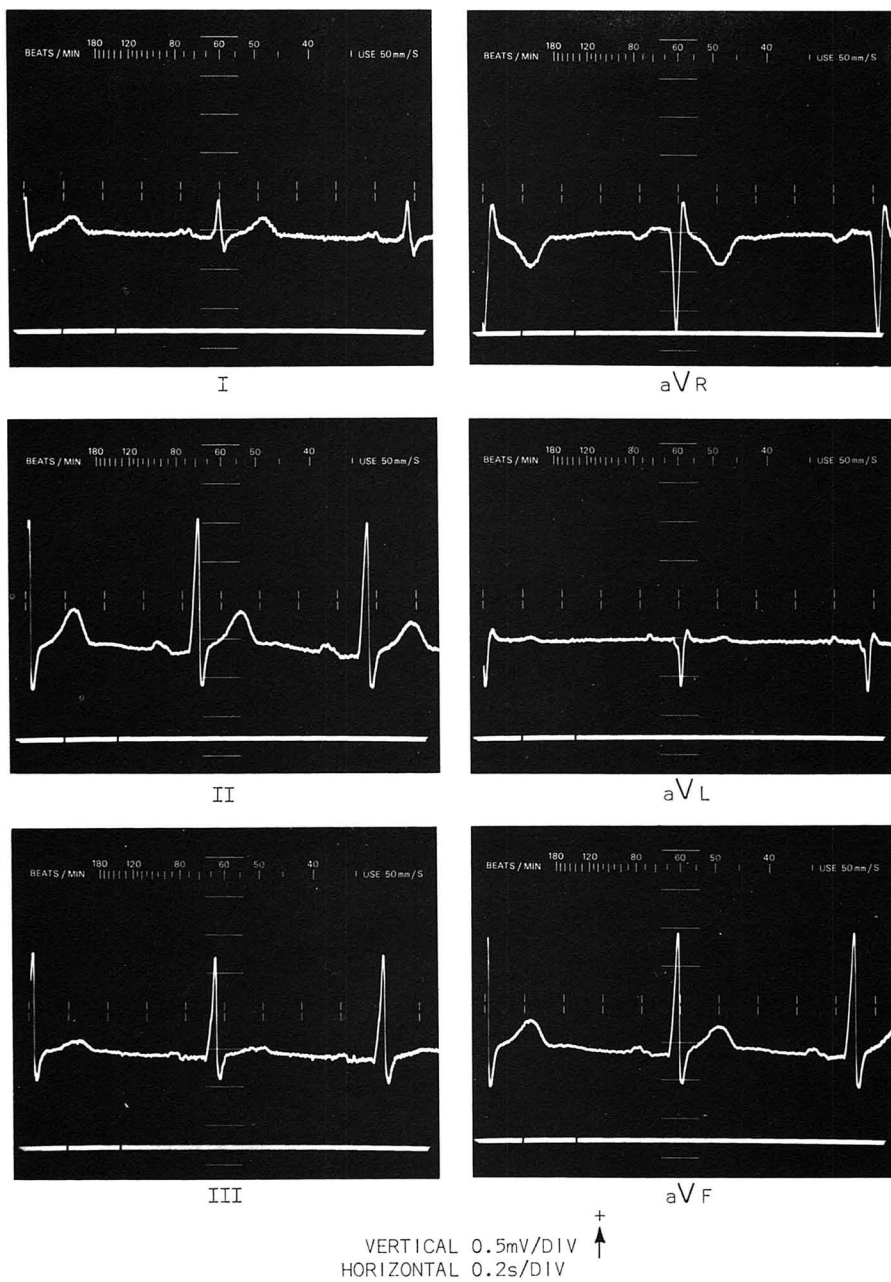
Fig. 5-10. Typical frontal plane ECG's (recorded with Type 410
Physiological Monitor).

5.7 TRANSVERSE-PLANE ECG MEASUREMENTS

We have previously discussed the measurement of the frontal-plane projection of the cardiac vector; it is common electrocardiographic technique to also record the transverse-plane projection of this cardiac vector. The transverse-plane is shown in Fig. 5-1. Many of the unipolar limb lead measurement techniques discussed earlier apply to transverse-plane electrocardiography. An indifferent electrode is formed by summing the potential at the right arm, the left arm and the left leg and the measurement consists of recording the potential between this indifferent electrode and a chest electrode placed at various positions on the torso. Transverse-plane ECG measurements are known as "the V lead measurements;" V_1, V_2, V_3, V_4, V_5 and V_6. The subscripts 1 through 6 refer to the position on the torso of the chest electrode as shown in Fig. 5-11. Although the standard transverse-plane ECG configurations are V_1 to V_6 many other transverse-plane configurations are used. The more common of these other configurations are V_7, V_8 and V_9 with the chest electrode placed on the left-hand side of the subject towards the back. If the chest electrode is placed in a position somewhat above or below a standard V position, then the configuration is subscripted H or L; thus, if the chest electrode were placed on the midline at the base of the rib cage, it would be referred to as V_{2L}. Occasionally V configurations are recorded with the electrodes on the right-hand side of the subject. Typical transverse-plane ECG's are shown in Fig. 5-12.
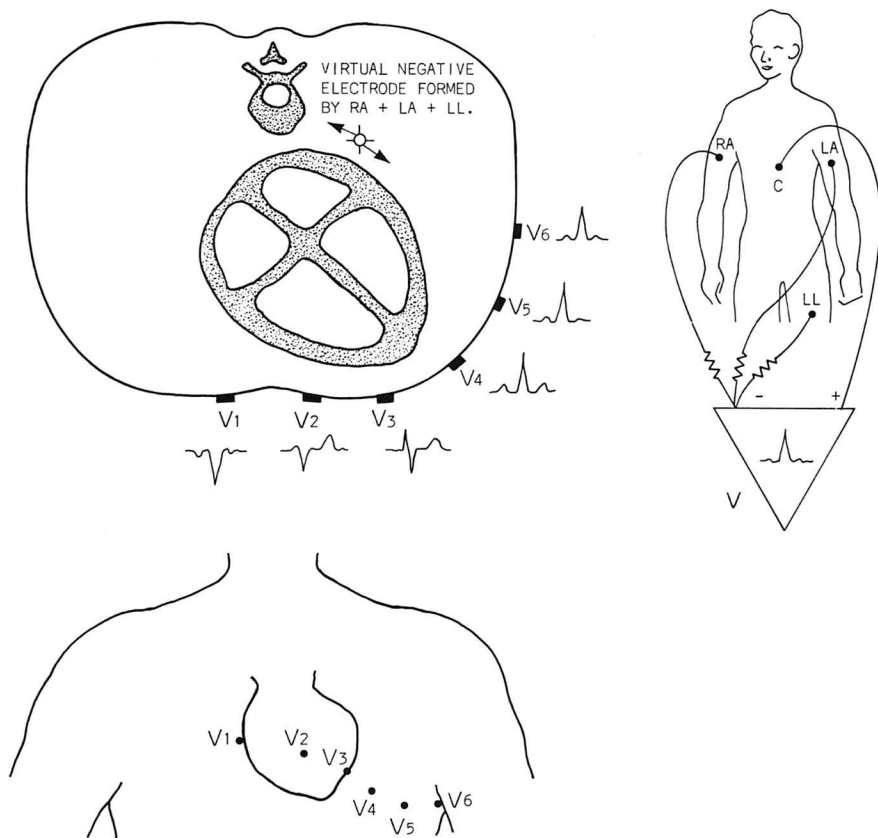
Fig. 5-11. Transverse plane ECG.

62

VERTICAL 0.5mV/DIV $\updownarrow$
HORIZONTAL 0.2s/DIV $\updownarrow$
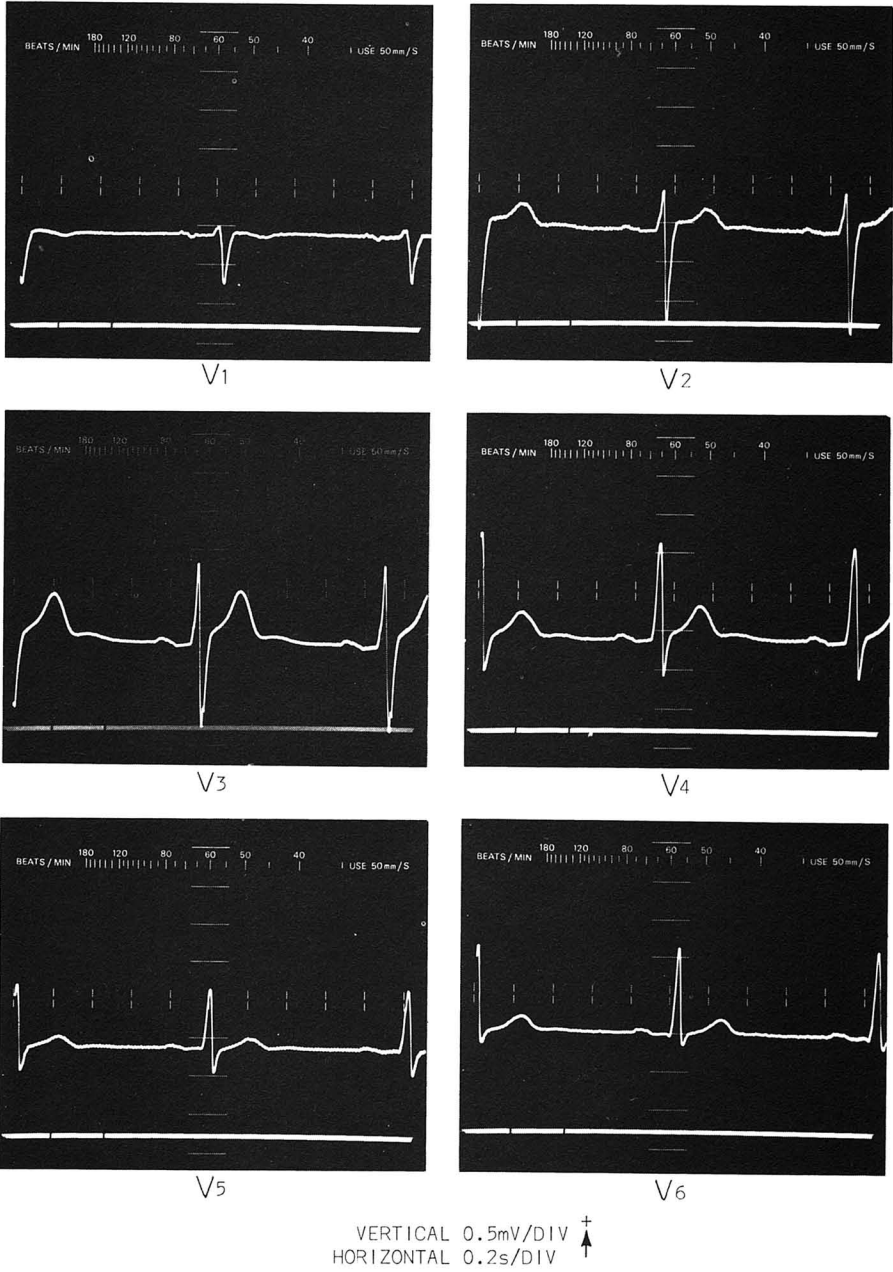
Fig. 5-12. Typical transverse plane ECG's (recorded with Type 410
Physiological Monitor).
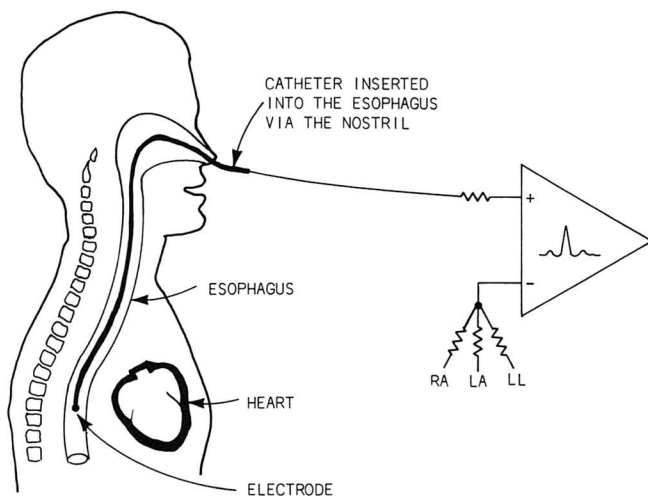
Fig. 5-13. Electrode positions —— sagittal plane ECG's.

5.8 SAGITTAL PLANE ECG MEASUREMENTS

esophageal
lead

The measurements of the cardiac vector projection in the sagittal plane are referred to as "the unipolar esophageal lead ECG measurements" or the "E lead ECG measurements." Although the unipolar esophageal lead ECG is rarely recorded these days, it can be recorded by attaching an esophageal lead to the positive input of the amplifier and forming an indifferent negative electrode using both arms and the left leg as with the transverse-plane V leads. Esophageal lead measurements are recorded from within the esophagus using a catheter, or fine rubber tube, threaded with a wire with an electrode attached to its tip, as shown in Fig. 5-13.

5.9 ECG INSTRUMENTATION REQUIREMENTS

The previous discussion has been concerned with electrode placement and electrode configurations. The potential appearing at these electrodes must be amplified by a differential amplifier and recorded on either a strip chart recorder, a cathode ray tube or, occasionally, magnetic tape.

64

display
devices

Commercial instruments usually include electrode selection, amplification and either a CRT display or a strip chart recorder. These instruments are commonly referred to as ECG monitors or physiological monitors if they utilize a CRT display and as electrocardiographs if they utilize a strip chart recorder display. The commercial electrocardiographs preceded commercial physiological monitors by 50 years or so; thus, many of the features on physiological monitors are carried over from electrocardiographs even though they may not be optimally suited for use with CRT displays.

electro-
cardiograph

Electrocardiographs almost invariably use graph paper with horizontal and vertical lines at 1 millimeter intervals and a heavier line at 5 millimeter intervals as shown in Fig. 5-14. In routine electrocardiography the recording speed is 25 millimeters per second; thus, the 1 millimeter horizontal intervals each represent 0.04 seconds and the 5 millimeter intervals represent 0.2 seconds. The sensitivity of electrocardiographs is typically 10 millimeters per millivolt.

CRT
displays

When referring to CRT displays, it is normal practice in the electronics industry to refer to millivolts/centimeter vertically and seconds/centimeter horizontally. However, due to the historical influence of electrocardiographs, the reciprocal of these dimensions is used in electrocardiography; i.e., millimeters/millivolt and millimeters/second.
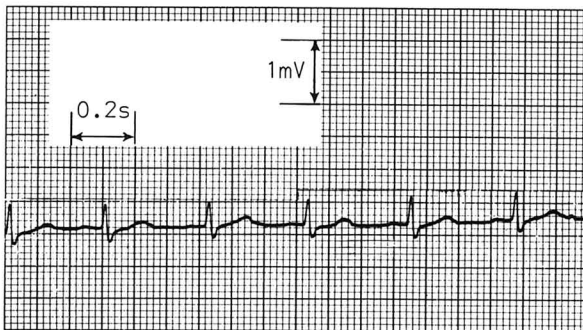
COMPUTER INSTRUMENTS CORP.   HEMPSTEAD, NEW YORK   RP-120-

Fig. 5-14. A typical electrocardiograph tracing.

Physiological monitors commonly use twice the
vertical sensitivity and twice the sweep speed of
electrocardiographs; that is, 20 millimeters/
millivolt and 50 millimeters/second.

As high input impedance differential amplifiers are
almost invariably used in modern electrocardiographs
and physiological monitors, some ground reference
must be maintained between the subject and the
amplifier. This is usually accomplished by attaching
right leg   an electrode to the subject's right leg and
ground   connecting this electrode to the ground of the
electrode   amplifier. For reasons covered in Chapter 16
(Electrodes) and Chapter 19 (Amplifiers), amplifiers
used to record the ECG should have a common mode
input impedance of at least 10 megohm per input and
should have a common mode rejection ratio of at
least 10,000:1. Frequency components within the ECG
extend from .05 Hz to approximately 80 Hz; most
commercial amplifiers will exhibit adequate high
frequency response, however many amplifiers do not
exhibit a low frequency response to .05 Hz or better.

It is highly desirable that any amplifier used to
record ECG's, or used to record other potentials
from the human body, should incorporate internal
subject   circuitry to protect the subject against electrical
protection   shock should the amplifier fail. Protection circuits
are covered in Chapter 17 (Safety).

5.10   INTERPRETATION OF THE ELECTROCARDIOGRAM

Interpreting the results obtained from the various
ECG measurements discussed previously is an art in
itself and should only be attempted by suitably
trained medical personnel. Although it is not within
the scope of this publication to cover ECG
interpretation, it is desirable that the reader have
some concept of what is considered normal and what
is considered abnormal and appreciate the meanings
of the more important terms used by cardiologists
when discussing the shape of the ECG waveform.

The "normal" ECG has come about by observing the
distribution in many thousands of healthy subjects.
The ECG's presented in Fig. 5-11 and Fig. 5-14 are
"normal" ECG's from a healthy adult male. Any

abnormalities variation from the normal rhythm of an ECG is known
as an arrhythmia. As arrhythmias may be more
evident in certain ECG measurements, a complete
set of frontal-plane and transverse-plane ECG's
is usually required for a complete analysis of the
electrocardiogram.

A fast heart rate in excess of 100 per minute is
known as tachycardia; a slow heart rate of less than
60 per minute is known as bradycardia. An
excessively large and continuous ECG with no
recognizable P, QRS or T waves is a sign of
ventricular fibrillation, a condition where the
ventricular muscle goes into local oscillation in
a "circus movement," waves of depolarization going
round and round the ventricular tissue. A random
and changing phase relationship between P and QRS
and T indicates adioventricular rhythm, in which a
complete bundle block occurs in the bundle of His,
and the auricles and ventricles are beating
independently, with the ventricles slower.
Arrhythmias also appear as extra beats, known as
ectopic beats, on an otherwise normal
electrocardiogram. A good reference for the study
of cardiac arrhythmias is *Principals of
Electrocardiography* by J. Goldman or *Coronary Care
Unit Nursing, Part I* by H. A. Braun and G. A.
Diettert.

6

VECTORCARDIOGRAPHY (VCG)

derived from ECG measurements

The electrical activity of the heart as projected along various axes on three planes of reference is referred to as the ECG and is discussed in Chapter 5. Vectorcardiography is the art of analyzing the electrical activity within the heart by obtaining ECG's along three axes at right angles to one another and displaying any two of these ECG's as a vector display on an X-Y oscilloscope. This display is referred to as a vectorcardiogram or VCG. The three axes chosen are designated X, Y, and Z as shown in Fig. 6-1. The planes bounded by any two of these axes are also shown in Fig. 6-1.
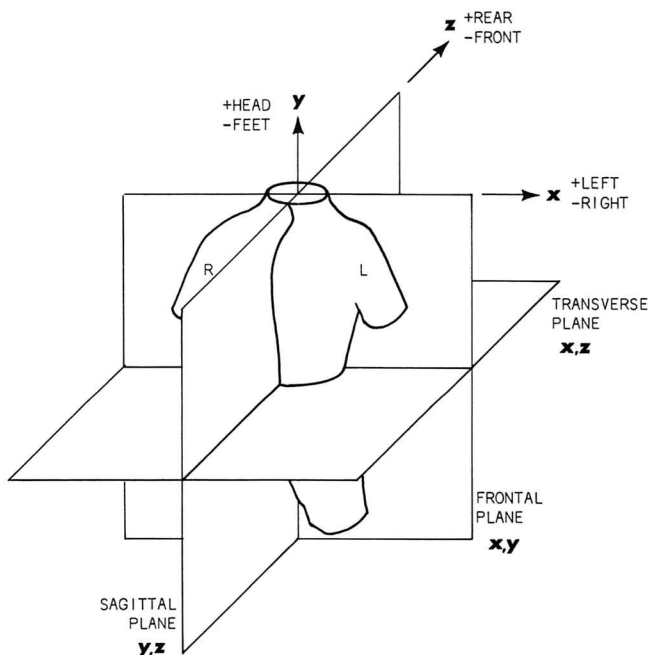
Fig. 6-1. VCG planes.

The electrocardiogram displays the electrical
potential in any one single axes, however the
vectorcardiogram displays the same electrical events
simultaneously in two perpendicular axes. At
present, vectorcardiography is used only in medical
centers and by specialty medical groups. Its value
is limited for routine clinical practice due to the
expense of the necessary equipment, the time
involved, and because no definite standard as to the
best lead system to be used or as to what is
"normal" has been established.

6.1 THE SPATIAL VECTORCARDIOGRAM

The spatial vectorcardiogram may be defined as the
record of the time variations of the instantaneous
vectors which represent the electrical activity of
the heart. Fig. 6-2 shows the spatial
vectorcardiogram and the resulting projections of
this spatial vectorcardiogram onto the frontal
plane, the transverse-plane and the sagittal plane.
The projections are referred to as the frontal VCG,
the transverse VCG and the sagittal VCG.
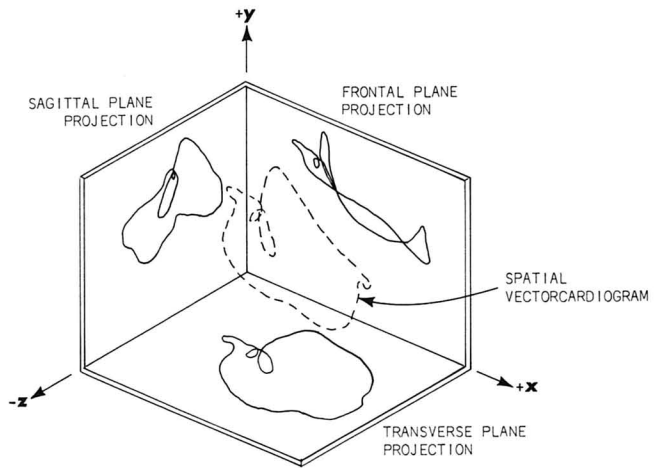
Fig. 6-2. VCG projections.

6.2 ELECTRODE PLACEMENT

The location of electrodes on the body to allow accurate recording of the ECG's along the X, Y and Z axes is extremely important. To obtain an accurate vectorcardiogram, the electrodes must be placed in positions that permit accurate recording of the three vectors. The body and the heart, however, exhibit characteristics which cannot be circumvented for practical purposes. This makes absolutely accurate recording of the vectorcardiogram almost impossible as any system of electrode placement is somewhat imperfect.

absolute
accuracy
difficult

The vectorcardiogram can be recorded by obtaining
the potential between two points located under the
arms for the X or left to right component, between
the chest and the back for the Z component and
between the head and left leg for the Y component.
Simply displaying any two of these three signals on
an X-Y oscilloscope with the same sensitivity
for both the X and the Y amplifiers would result in
considerable errors as the potential generated within
the heart is attenuated by different amounts in the
X, Y and Z directions and, as the torso is not
homogeneous, these signals are generated along axes
which are not exactly at 90° to one another.

6.3 FRANK ELECTRODE SYSTEM

Various attempts have been made to compensate for
the above attenuation and torso inhomogeneities.
The Frank system of vectorcardiography attempts to
seven   achieve isotropy as well as orthogonality by using
electrodes   seven critically placed electrodes and a resistive
                 network. The Frank electrode positions and the
compensating   Frank attenuation and compensation network is shown
network   in Fig. 6-3.
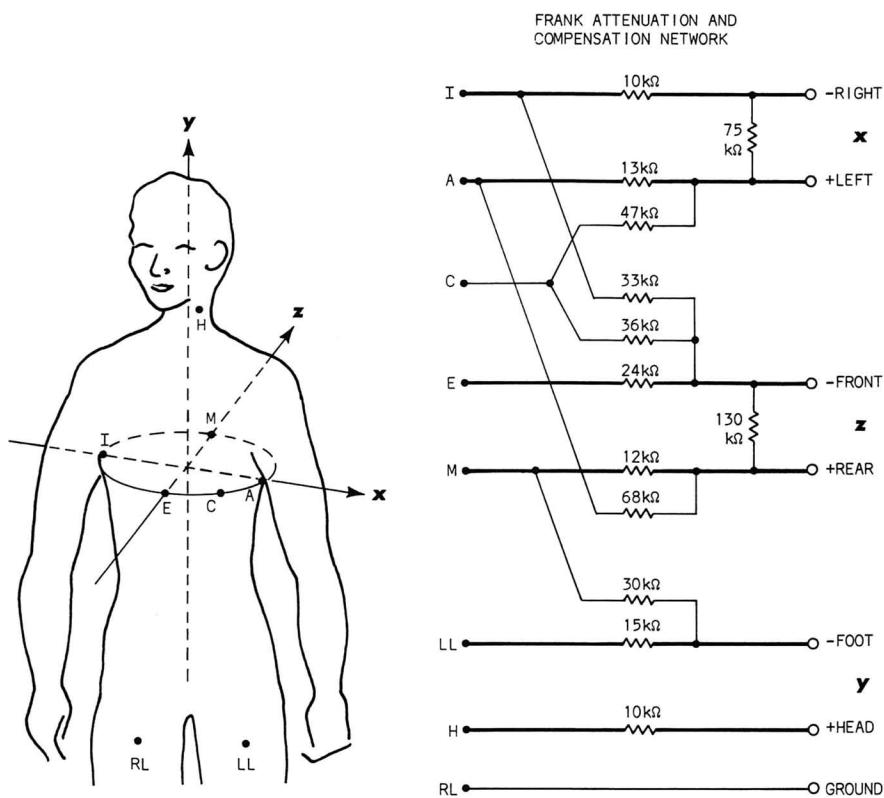
Fig. 6-3. Frank electrode positions.
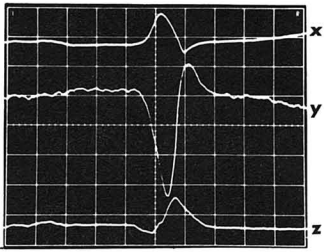
THE **x, y** AND **z** OUTPUTS OF
THE FRANK ELECTRODE NETWORKS

.04s/DIV
0.5mV/DIV

THREE TRACE DISPLAYS SHOWING THE THREE
QRS COMPONENTS OF THE VECTOR DISPLAYS
SHOWN BELOW.
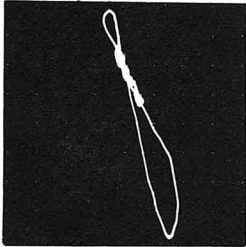
y VERTICAL
x HORIZONTAL

z VERTICAL
x HORIZONTAL
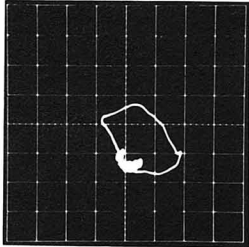
y VERTICAL
z HORIZONTAL

FRONTAL

TRANSVERSE
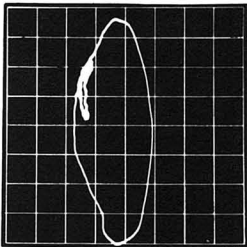
SAGITTAL
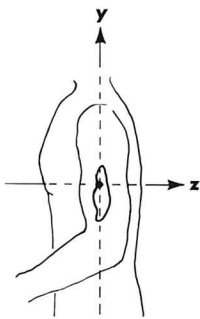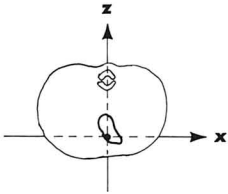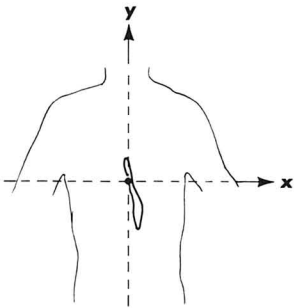
Fig. 6-4. The Frank vectorcardiogram.

6.4 POLARITY CONVENTION

+X to left
side, +Y to
head, +Z to
back

Fig. 6-3 shows a positive direction for X towards the subject's left-hand side, a positive direction for Y towards the head and a positive direction for Z towards the back. Since all modern oscilloscopes adopt the convention of a positive potential change deflecting the beam upwards and towards the right-hand side of the screen, it is necessary to adopt this polarity convention for the three axes to present the three vectorcardiograms superimposed on the human body as viewed from logical viewpoints; i.e., from the front, from the top and from the left-hand side. The polarity of the ECG waveform in the X and Z direction will thus appear "positive," however the polarity of the normal ECG waveform in the Y direction will be inverted, as shown in Fig. 6-4. Frank's original network suggested a positive Y direction towards the feet which resulted in the three separate ECG components being represented as "positive" signals. This resulted in inversion of the frontal and sagittal projections of the vectorcardiogram, making correlation to the actual position of the heart somewhat difficult. The conventions adopted in this chapter are nowadays regarded as standard.

6.5 OTHER ELECTRODE SYSTEMS

Fig. 6-5 shows three other electrode configurations in common use, the axial electrode placement, the tetrahedron electrode placement and the cube electrode placement. Although none of these three configurations are considered as accurate as the Frank system of vectorcardiography, they do not require complicated resistive networks and can utilize two standard ECG monitors and an X-Y oscilloscope to obtain satisfactory results. Indifferent electrodes are formed by summing the effects of various electrodes in the same way as for electrocardiography as covered in Chapter 5.

6.6 THE NORMAL VECTORCARDIOGRAM

Vectorcardiograms obtained from a healthy, male Caucasian subject using the Frank system are shown in Fig. 6-4. The upper CRT display shown in Fig. 6-4 represents the QRS segment of the three separate ECG's used to form the vectorcardiograms shown in the lower CRT photographs. Careful inspection will reveal that the three QRS segments are not exactly in phase; if they were, the vectorcardiogram loops resulting from the QRS segment would all be straight lines. Although each vectorcardiogram should theoretically exhibit three loops showing the vector orientation of the P wave, the QRS axis and the T wave, the loop obtained from the QRS complex predominates and an increase in horizontal and vertical deflection factors is normally necessary to adequately view the loops resulting from the P wave and T wave. It should be noted that the VCG is very dependent on the phase of the respiratory cycle in which it is recorded. The posture of the subject is also very important.

QRS loop
predominant

(A) AXIAL VECTORCARDIOGRAM

(B) TETRAHEDRON VECTORCARDIOGRAM

(C) CUBE VECTORCARDIOGRAM
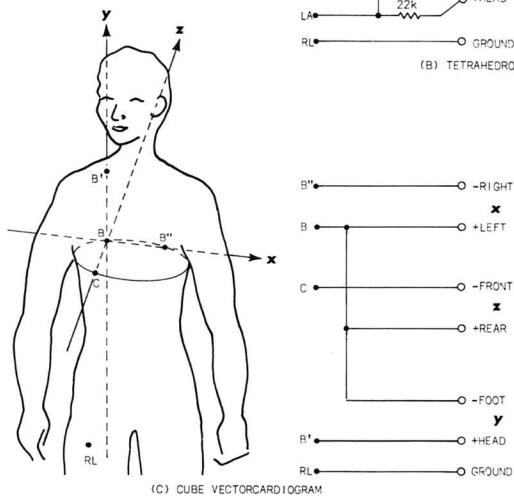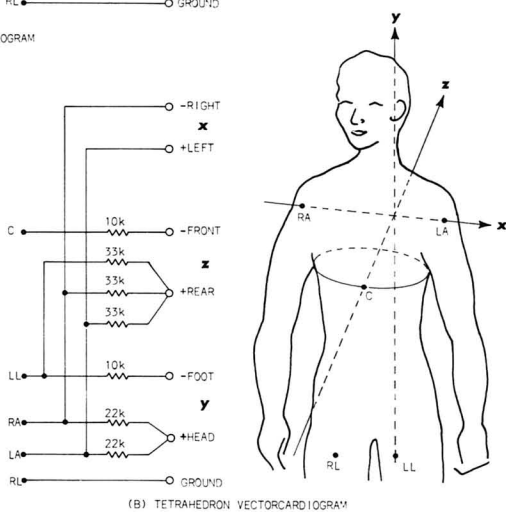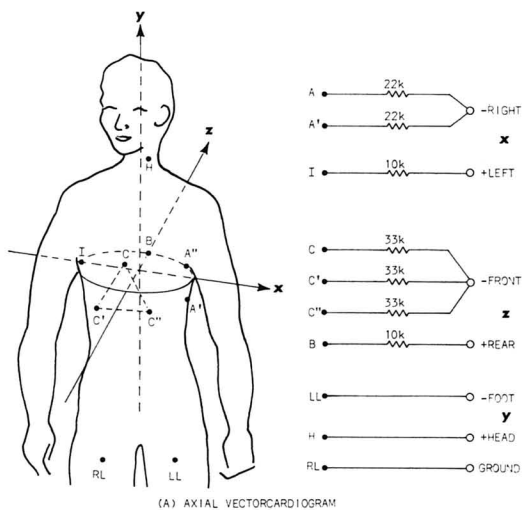
Fig. 6-5. Axial, cube and tetrahedron electrode positions.
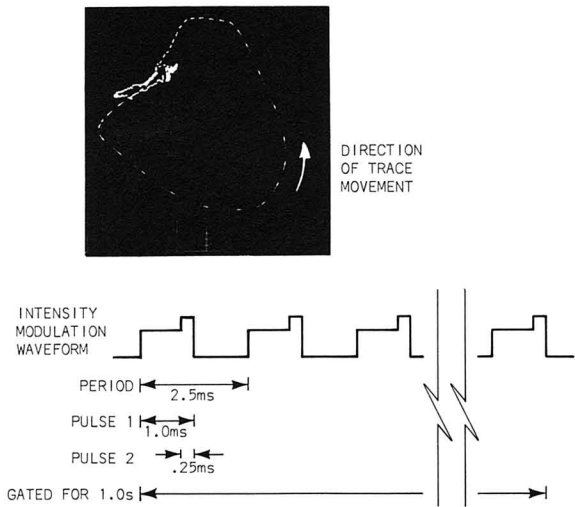
Fig. 6-6. Sagittal VCG with 2.5-ms time-duration markers.
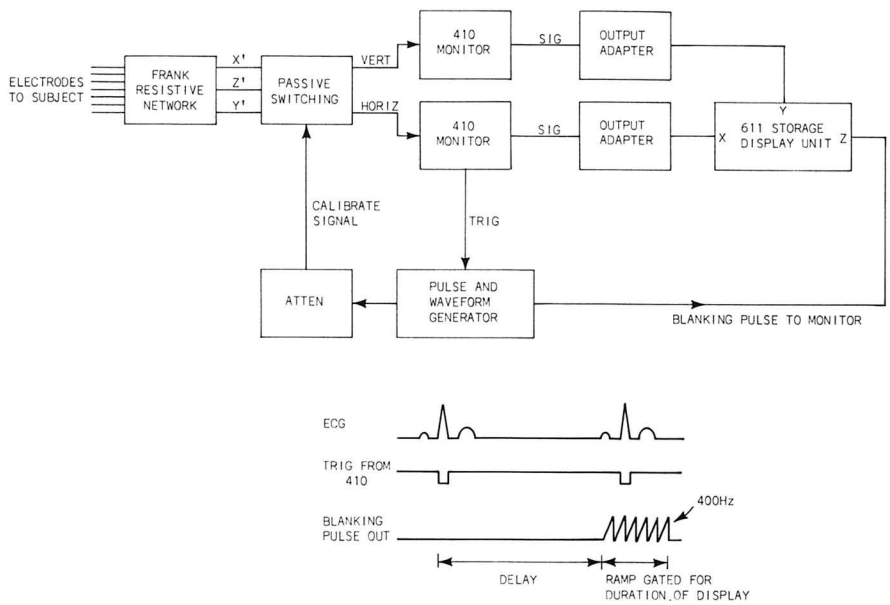
Fig. 6-7. VCG system using two monitors and incorporating intensity modulation.

6.7 TIMING AND DIRECTION REFERENCE

intensity
modulation

The vectorcardiograms shown in Fig. 6-4 give no
information as to the direction of movement of the
CRT beam when inscribing the vector loops, nor do
they provide any indication of the time taken to
inscribe the loops. As this information is often
desirable, some form of intensity modulation is
usually incorporated into VCG instrumentation
to provide a time scale and direction reference.
The single sagittal vectorcardiogram shown in
Fig. 6-6 includes intensity modulation at 400-Hz
rate to provide a timing reference of 2.5
milliseconds between each consecutive brightened
area on the display. Two discrete intensity
modulation levels are used for each intensifying
pulse providing an "arrowhead" effect to indicate
the direction of movement of the CRT beam when
inscribing the vector loops. As with the VCG's
shown previously in this chapter, only the QRS loop
is clearly displayed. Some VCG systems utilize
sawtooth intensity modulation in place of the two-
step intensifying pulse.

6.8 INSTRUMENTATION REQUIREMENTS

A complete vectorcardiography system is shown in
Fig. 6-7. Many of the ECG instrumentation
techniques discussed in Chapter 5 are used when
recording the VCG. VCG measurement requires two
separate ECG measurement systems as well as a Frank
resistive network, an X-Y storage oscilloscope
and intensity modulation circuitry.

The circuit of the Frank resistive network is shown
in Fig. 6-3. Further details of this network are
given in Chapter 29. The passive switching network
selects the correct combination of the X, Y and Z
signals from this Frank network for displaying
either the frontal, transverse or sagittal
vectorcardiogram. The two ECG's selected are
displayed on individual 410 monitors and the output
of each of these monitors is coupled via passive
output adapters to the 611 storage display unit.

78

The output adapters are necessary to decrease the signal output of the 410 to the level required by the 611 and to allow calibration of the system from an external calibration signal. For details of these adapters refer to Chapter 29. A combination of Tektronix pulse and waveform generation modules (Refer to Chapter 22) is triggered by the trigger signal available from either of the 410 monitors to provide blanking pulses to the CRT. This blanking information adds time and direction information to the vector display. These modules also provide a calibration signal which is fed to the passive switching network via an attenuator. A C-10 camera may be used to record the VCG from the 611 display unit.

410 with 611

5031

The Tektronix 5031 storage oscilloscope operated in the X-Y mode is ideally suited to vectorcardiology. The 5031 may be used as a complete vectorcardiograph system by using the cube electrode placement given in Fig. 6-5. Normally the internal time base of the 5031 is disabled when operating in the X-Y mode. However, if this disabling circuit is removed by removing pin Z from the sweep circuit board, the internal sweep circuitry can be triggered from the calibrator and used to provide a multilevel intensity modulation as shown in Fig. 6-6. If desired, a Frank resistive network as detailed in Chapter 29 could be used ahead of the 5031. Any oscilloscope camera may be used to record the results.
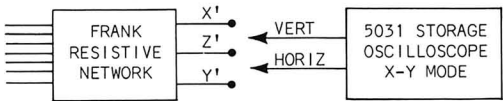
Fig. 6-8. Simplified VCG system.

Many other vectorcardiograph systems are in use.
One of the more common alternate techniques is to
561B or   use a 561B or 564B oscilloscope with 3A74 four-trace
564B with   units in both the vertical and horizontal channels.
3A74   When used in conjunction with three preamplifiers,
all three projections of the spatial vectorcardiogram
can be displayed simultaneously. The display can
then be photographed and the photograph enlarged for
convenient viewing. Additional information on the
components required for these VCG systems are given
in various chapters in Section III. The custom
instrumentation required for the above systems are
discussed in Chapter 29.

It should be appreciated that the Tektronix 410
monitor is the only Tektronix product incorporating
410 subject   subject protection features. While other Tektronix
protection   products include some protection and are inherently
highly reliable, current limiting adapters as
described in Chapter 29, Section 29.12, should be
used when using these instruments in conjunction
with human subjects. Safety is covered further in
Chapter 17.
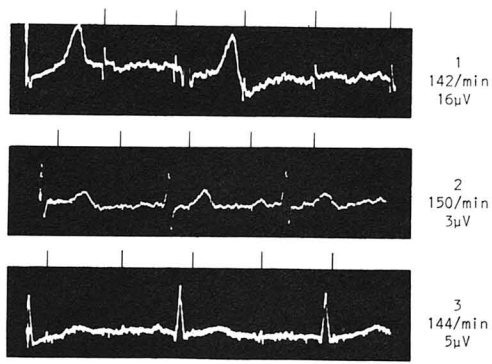
Fig. 7-1. Fetal ECG's at 16 weeks on several subjects.
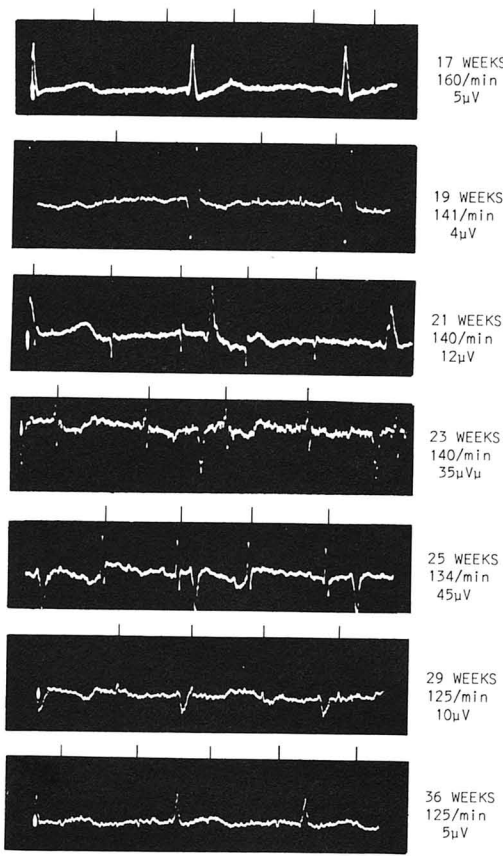
Fig. 7-2. Fetal ECG variations with gestation time.

7

FETAL
ELECTROCARDIOGRAPHY

Fetal electrocardiography is the art of recording the ECG of a fetus in utero by placing electrodes on the mother's abdomen and using recording techniques similar to those used for normal electrocardiographic recording. The procedure is complicated by the existence of the mothers's ECG and muscle action potentials and by the severe attenuation suffered by the fetal ECG when recorded at the mother's abdomen. The fetal ECG is normally abbreviated F-ECG.

7.1 THE EXISTENCE OF THE FETAL ECG

best from
16 to 24
weeks

The fetal ECG may be recorded as early as the 11th week of gestation in some subjects and can be recorded in almost all cases after 16 weeks gestation. Fig. 7-1 shows the variation in fetal ECG amplitude that may be expected at 16 weeks. By the 18th week the fetal ECG is invariably pronounced and continues to become more evident to about the 24th week. At this stage, amniotic fluids begin to build up in the mother which reduces the amplitude of the fetal ECG. This reduction slowly continues for most of the remaining weeks of pregnancy. Just prior to delivery the amniotic fluids are released and the fetal ECG can again be prominently recorded. The variation in the fetal ECG obtained from a subject at various gestation times is shown in Fig. 7-2.
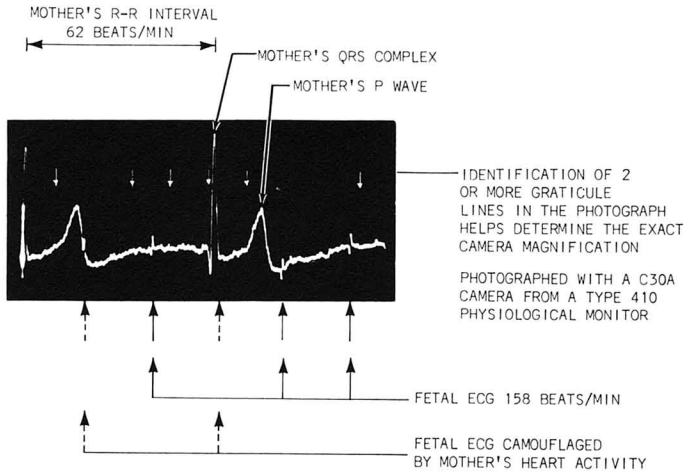
Fig. 7-3. A typical ECG display at 18 weeks.

7.2 THE NORMAL FETAL ECG

maternal ECG
versus
fetal ECG

only R wave

As previously stated, the fetal ECG amplitude varies greatly with gestation time and also varies from one subject to another. A typical fetal ECG obtained at 18 weeks is shown in Fig. 7-3. The mother's ECG is clearly evident and, being many times greater in amplitude than the fetal ECG, may camouflage the fetal ECG. The fetal ECG can clearly be observed during the isoelectric period between the mother's ECG. The techniques used to record this fetal ECG are covered late in this chapter. From this recording, it is evident that the mother's heart rate is 62 beats per minute and the fetal heart rate is 158 beats per minute. In the early stages of pregnancy the fetal heart rate is normally between 140 to 160 beats per minute. If the mother is relaxed during the recording of the fetal ECG, it is expected that her heart rate would be under 80 beats per minute. Due to the low amplitude of the fetal ECG, the fetal R wave is normally the only part of the fetal ECG complex that is evident.

7.3 SUBJECT PREPARATION

As the electrical potential generated by muscular activity within the mother's abdomen will clearly be present using abdominal electrodes when attempting

relaxed
state
imperative

to record the fetal ECG, it is imperative that the mother be situated comfortably and in a state of complete rest. A bed or couch is normally preferred to a clinical examination table. To reduce the mother's muscular activity in the abdominal area as much as possible, it is desirable that the subject does not eat for several hours prior to recording and that the subject empty her bladder shortly before recording.

7.4   ELECTRODE PLACEMENT

Many electrode positions have historically been used to record the fetal ECG. It is now generally accepted that abdominal electrodes are preferable to most other configurations. Due to the variation expected in the fetal position, many electrodes should be applied to the abdomen and the fetal ECG recorded between various combinations of these electrodes until a positive recording is obtained. The most common fetal ECG lead configuration is the

Blondheim
configuration

Blondheim configuration shown in Fig. 7-4. This configuration consists of three electrodes (A, D and F) placed at the vertices of a 60° triangle and an additional three electrodes (B, C and E) placed at the vertices of an inverted 60° triangle. Fig. 7-4 also shows two electrodes on the back of the subject (G and H). It has been found that, during the early stages of pregnancy, a recording from one of these two back electrodes to any of the abdominal electrodes produces more reliable results. It is important that care be exercised when applying the electrodes.
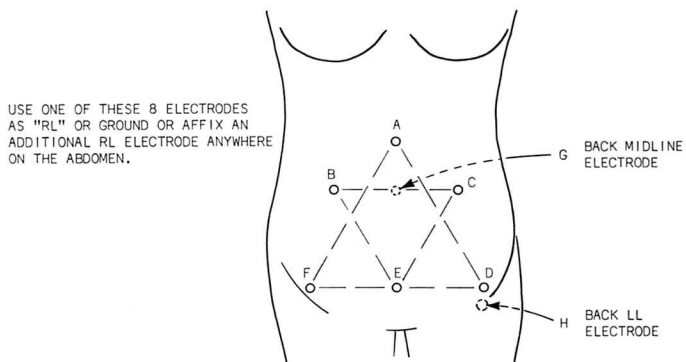
Fig. 7-4. Fetal ECG electrode positions.

ELECTRODES AFFIXED
TO ABDOMEN
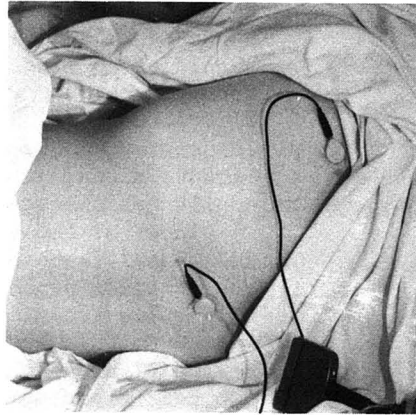
TWO BACK ELECTRODES AFFIXED AND
ADAPTER CABLES FITTED TO THEM
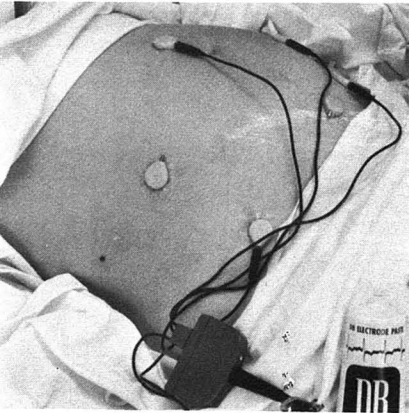
ADAPTER CABLES FITTED TO
ABDOMINAL ELECTRODES

Fig. 7-5. Fetal ECG electrode placement.

resistance
attachment
necessary

Electrode placement techniques providing low source resistance, as covered in Chapter 16, must be used. The abdominal and back electrodes affixed to a subject are shown in Fig. 7-5. It has been reported that, during early stages of pregnancy, better results may be obtained by exerting considerable pressure onto the electrodes so as to move these electrodes closer to the actual fetus.

7.5 ELECTRICAL INTERFERENCE

Particular care must be taken to eliminate line frequency interference from the fetal ECG recordings. Adequate instrumentation can eliminate any line frequency signal appearing as a common mode signal, however it is impossible to eliminate line frequency interference if it appears as a differential signal. Filters cannot be used, as the prime frequency content of the fetal ECG is between 20 Hz and 80 Hz. Line frequency interference can usually be eliminated by recording the fetal ECG with the Type 410 Physiological Monitor operated from its internal batteries and by completely disconnecting other

away from
power lines

power-line-operated devices within the near vicinity. Such devices should be completely disconnected rather than simply turned off to eliminate electrostatic interference if the switch is located in the power-line neutral lead. Interference can be eliminated in many cases by orienting the patient in a particular position within the room. If this is not successful, it is normally preferable to locate another recording site rather than to invest in costly electrostatic and electromagnetic shielding. The fetal ECG recording may still show evidence of fetal ECG activity even though line frequency interference is present. The leads connecting the patient to the monitor should be as

short leads

short as possible, preferably no longer than two feet.

7.6 RECORDING TECHNIQUES

Electrodes should be placed on the mother's abdomen
as shown in Fig. 7-4. If it is intended that the
fetal ECG from many patients be recorded, it is
desirable that some form of custom electrode selection
switching unit be built to eliminate the necessity of
having to make separate connections to the recording
equipment for each electrode configuration. A
typical set of fetal ECG's obtained by recording all
combinations of the electrodes shown in Fig. 7-4 is
shown in Fig. 7-6. Normally it is not necessary to
record all possible combinations of these electrodes
and the procedure is usually considered complete
when a fetal ECG has clearly been observed in two or
three configurations.

electrode combinations

In early stages of pregnancy the following electrode
combinations appear to give the greatest possibility
of observing the fetal ECG: C-F, B-D, A-F, A-D and,
if the G and H electrodes are used, G-F, G-D and
B-H. In later stages of pregnancy when the fetal
ECG is pronounced, it is normally only necessary
to record from either of the three sets of
electrodes forming the 60° triangles of the
Blondheim configuration, that is, electrodes A, F
and D or electrodes B, C and E. As the amplitude
of the fetal ECG is normally between 5 and 50
microvolts, and, as it is preferable to use battery-
operated equipment to record the fetal ECG for
reasons discussed in the previous section, most
commercial electrocardiographs or physiological
monitors are not compatible with fetal ECG recording.
It is common practice to use an EEG recorder or a
physiological monitor operated in the EEG position
to monitor the fetal ECG. This technique is
recommended when recording the fetal ECG with the

using the 410

Type 410 Physiological Monitor. The EEG position
on the function selector switch is selected and the
fetal ECG electrodes are connected to the EEG
position on the patient cable. A standard 410, as
well as any other commercially available EEG
recorder, may give satisfactory results, however
most of these instruments have a low frequency
response extending below .1 Hz. Muscle activity
within the mother's abdomen will probably result in
severe interference causing the trace to be off
screen most of the time. This problem can be
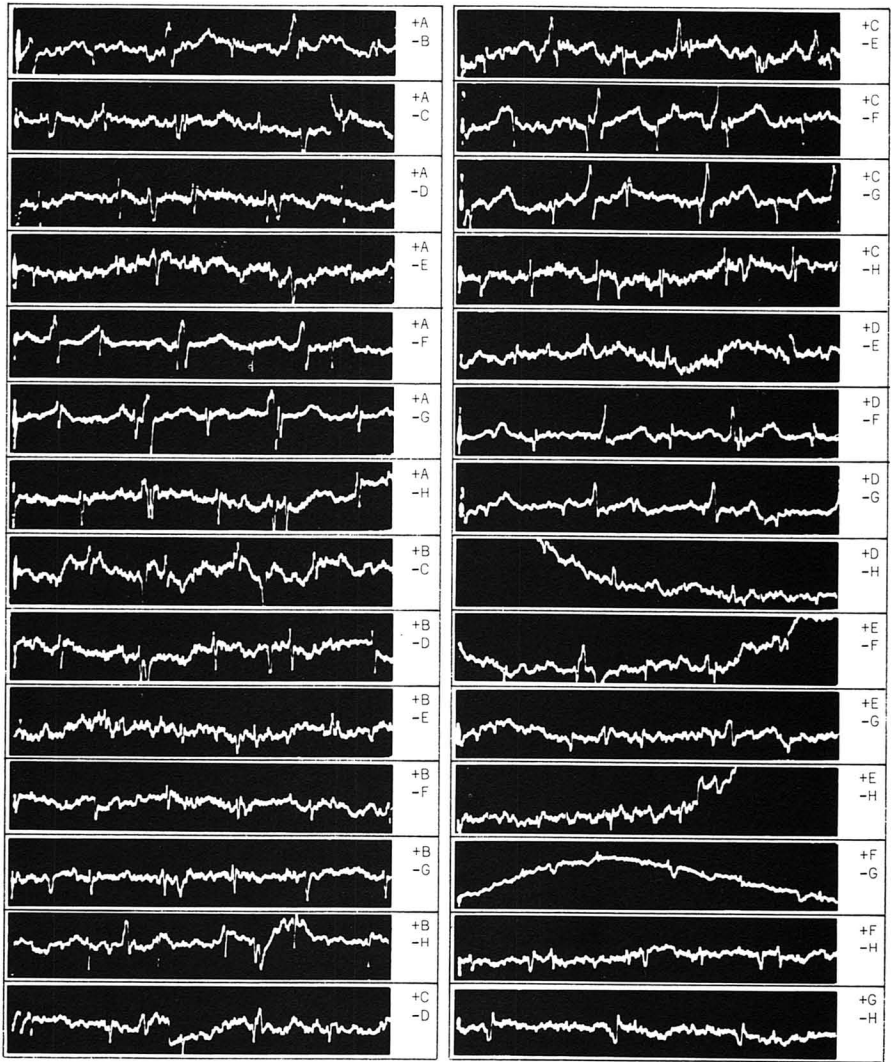
Fig. 7-6. Fetal ECG variations with various electrode positions
(23 weeks gestation).

alter low-
frequency
cutoff

overcome by altering the low frequency cutoff
characteristics of the monitor. A modification that
may be added to the 410 monitor to increase the low
frequency response in the EEG position from below
.1 Hz to approximately 1 Hz is given in Chapter 29.
This modification also reduces the gain in the EEG
position from 50 to 100 microvolts per centimeter.
The gain can, however, be increased with the
variable control by a factor of 3 to give a maximum
gain of 33 microvolts per centimeter. All fetal
ECG photographs shown in this chapter were recorded
at a sensitivity of 33 microvolts per centimeter and
at a sweep speed of 50 millimeters per second.

photography

As detailed analysis of the waveform is often
necessary to detect the fetal ECG, particularly in
the earlier stages of pregnancy, it is desirable
that a photograph be taken of the information
appearing on the screen of the 410 monitor.
Although no rigid mounting bezel is available to
adapt the 410 monitor to a Tektronix trace-recording
camera, a Tektronix C-30A camera can be held against
the faceplate of the 410 monitor and satisfactory
results can be achieved. All photographs shown in
this chapter were recorded in this manner. When
photographing from the 410 with the C-30A camera, the
shutter should be in the B position and opened
manually for approximately 2 seconds. Optimum
results are obtained using an aperture of f/16 and
a magnification of .9X. Certain nonlinearities exist
when photographing the curved faceplate CRT used in

compensating
for nonlinear
photo

the 410, however these nonlinearities can be
compensated for by examination of the trace to
identify two or more graticule lines in the
photograph. Once these graticule lines are
identified, the ratio of the spacing between these
graticule lines to the spacing between the fetal ECG
R waves gives the fetal ECG heart rate as given by
the following formulas:

-- for fetal ECG spacings measured
near the center of the screen using
a sweep speed of 50 millimeters per
second.

$$\text{Fetal heartrate} = \frac{300 \times \text{graticule mark spacing}}{\text{fetal ECG R wave spacing}} \times .9$$
Beats per min.

-- for fetal ECG spacings measured
near the edge of the screen using a
sweep speed of 50 millimeters per
second.

$$\text{Fetal heartrate} = \frac{300 \times \text{graticule mark spacing}}{\text{fetal ECG R wave spacing}}$$
Beats per min.

Due to the nonlinearities mentioned earlier, a
correction factor of .9 is necessary for information
at center screen and no correction factor is
necessary for information at the edge of the screen.
Chapter 24 discusses photography of the 410 in more
detail and gives the reasons for these correction
factors.

7.7   INTERPRETATION OF THE FETAL ECG

fetal
ECG versus
stethoscope

When attempting to record the fetal ECG, positive
results give a positive indication that the fetal
ECG does exist; however negative results are
nonconclusive and give no indication as to the
viability of the fetus. Although the fetal ECG can
be recorded as early as the 11th week and can be
recorded in almost all cases by the 16th week, it
is generally accepted that fetal heart activity
cannot be detected by a standard stethoscope earlier
than approximately the 20th week. Thus the fetal
ECG can give a positive indication of the existence
of a live fetus several weeks earlier than can be
obtained with a stethoscope.
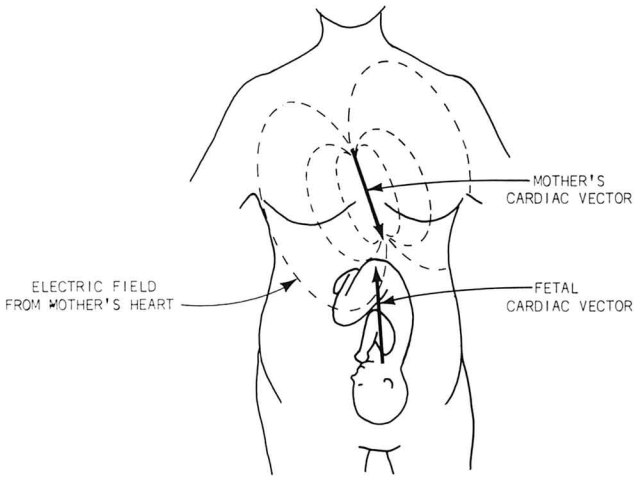
Fig. 7-7. Fetal cardiac vector vertex presentation at full term.

Fig. 7-7 shows the orientation of the electrical
axis of the fetal heart compared to the orientation
of the electrical axis of the mother's heart. As
the electric field generated by the mother's cardiac
mother's
ECG vector
vector produces "mother's ECG" activity when
recording the fetal ECG, the axes of this "mother's
ECG" activity will be tangential to this field and
not necessarily parallel to the direction of the
mother's cardiac vector. Within a few weeks after
the fetal ECG is first detected, its amplitude is
great enough for accurate measurement and the
results can be plotted on a vector diagram as shown
in Fig. 7-8 to determine the electrical axis of the
fetal heart. Fig. 7-8 shows the electrical axes
obtained from the recording shown in Fig. 7-6. As
only two fetal ECG's along two axes are necessary
fetal ECG
vector
to plot the fetal ECG vector, the other four vectors
shown on Fig. 7-8 can be used to determine the
accuracy of the measurement technique. An error
triangle is obtained by plotting the intersection
of all vectors; the head of the fetal ECG vector
must lie somewhere within this error triangle.

Fetal electrocardiography is perhaps most useful
when attempting to diagnose multiple pregnancy.
X-ray techniques are known to be harmful to the
fetus; however, fetal ECG techniques are completely
multiple
pregnancy
harmless and can detect multiple pregnancy as early
as the 16th week. Fig. 7-9 shows a fetal ECG
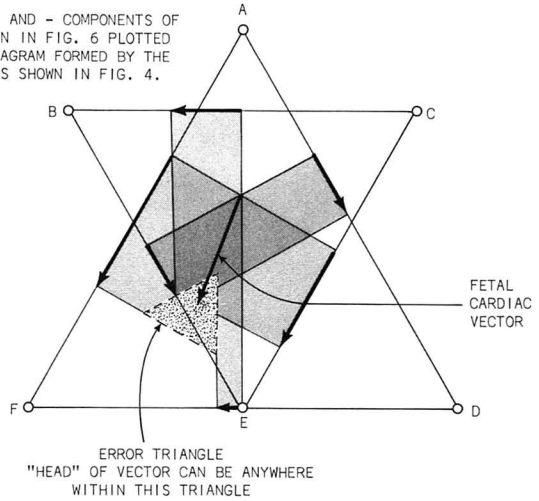
Fig. 7-8. Vector determination of fetal position.
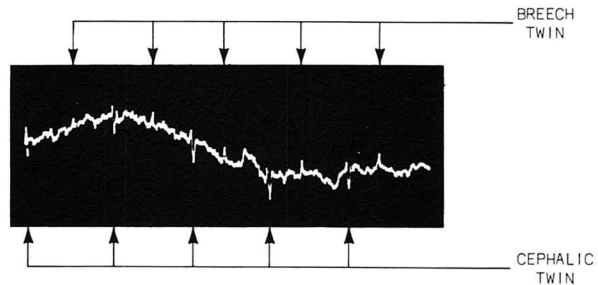
Fig. 7-9. Twin gestation —— fetal ECG at 21 weeks.

recording of twins at 21 weeks. As the fetal R waves are of opposite polarity, it is obvious that their electrical vectors are in opposite directions, i.e., one twin is breech and the other is cephalic. Fetal R waves in the same direction as the mother's QRS complex represent the breech twin and fetal R waves in a direction opposite to the direction of the mother's QRS complex represent the cephalic twin.

The fetal ECG is often recorded during fetal delivery to indicate whether the delivery will be from the vertex or the breech position and to give an indication of fetal distress during labor.

8

BLOOD PRESSURE AND FLOW

Blood pressure measurements can be classified into three discrete groups: direct, indirect and relative. Direct blood pressure measurement involves gaining access to the circulatory system and measuring the pressure in the system directly with some form of pressure transducer. Indirect blood pressure measurement involves application of pressure external to the circulatory system and observation of the effect of this external pressure on the system. Relative blood pressure measurements are uncalibrated indirect measurements usually performed with simpler, and inherently more convenient, instrumentation. Blood flow is derived from the measurement of blood velocity which necessitates gaining access to one or more of the primary arteries within the body using surgical techniques. Information on blood pressure and blood flow is often adequate for analysis of the circulatory system, however a complete hydrodynamic analysis of the circulatory system can only be accomplished by also measuring blood volume.

8.1 DIRECT BLOOD PRESSURE MEASUREMENT

Direct blood pressure is measured by inserting a pressure transducer somewhere within the circulatory system: Figs. 2-1, 2-2 and 2-3 of Chapter 2 refer to this system. Some transducers are designed to be inserted directly into the circulatory system, however it is more common to connect to the circulatory system with either a catheter or a hypodermic needle and to record the pressure with a pressure transducer attached to the catheter or needle.
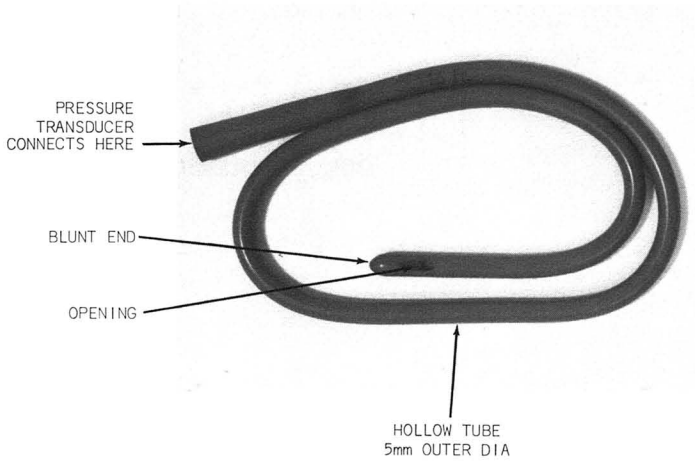
Fig. 8-1. A 5mm cardiac catheter.
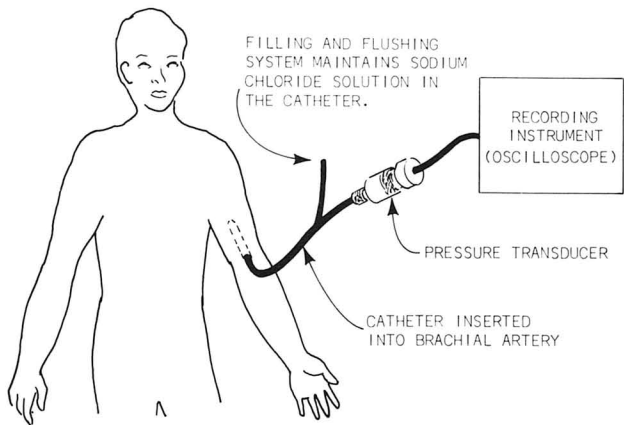
Fig. 8-2. Direct blood pressure measurement.

cardiac
catheter

A cardiac catheter consists of a rubber, teflon or polyethylene tube with one end formed to a smooth bullet-like shape to allow its easy introduction into veins and arteries. A small opening is formed about 1 centimeter from this smooth end as shown in Fig. 8-1. The catheter shown in Fig. 8-1 is approximately 40 centimeters long and 5 millimeters in diameter. Catheters are generally available from 1 to 10 millimeters in diameter and in various lengths. Cardiac catheters are also often described by their circumference.

With the catheter/pressure-transducer configuration shown in Fig. 8-2, pressures can be measured in almost any portion of the circulatory system. By gaining entry to the circulatory system at either the arms or legs a catheter may be manipulated throughout the circulatory system. It should be noted that the pressure measured will be the pressure at the blunt end of the catheter and not the pressure at the point where the catheter enters the circulatory system.

entry to circulatory system

It is possible, with careful manipulation, to actually introduce the smooth end of the catheter into the right atrium via the inferior vena cava. Cardiac defects, including holes in the heart, can then be explored. This procedure is referred to as cardiac catheterization. Investigation of the left ventricule by catheterization via the aorta, referred to as arterial catheterization, almost always causes trouble with arterial damage and leakage into the tissues.

cardiac catheterization

The most common direct pressure measurement is arterial pressure which is usually measured by introducing the catheter into the brachial artery at the elbow of either arm.

arterial pressure

A typical blood pressure transducer for use with a catheter is shown in Fig. 8-3. As little difference in pressure is encountered throughout the arterial system, it is unnecessary to introduce the catheter any great distance into this circulatory system.
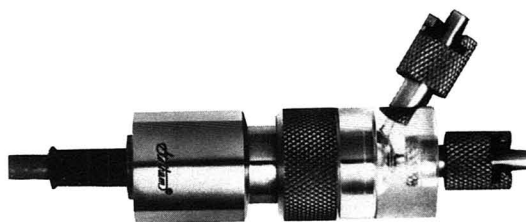
transducer

STATHAM INSTRUMENTS INC. - MODEL P23De
LENGTH - 2.5"
OUTPUT WITH 7.5VRMS EXCITATION = 75µVRMS/mmHg
FOR USE WITH CATHETERS.

Fig. 8-3. A conventional blood pressure transducer.

96

TRANSDUCER   NEEDLE

STATHAM INSTRUMENTS INC. - MODEL P37
TRANSDUCER LENGTH ≈ 1.2"
OUTPUT WITH 7.5VRMS EXCITATION = 37μVRMS/mmHg
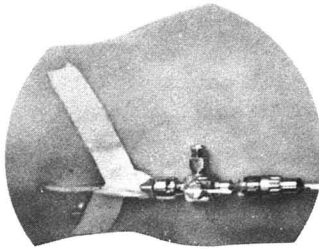
TRANSDUCER CONNECTED AT BRACHIAL
ARTERY PUNCTURE SITE.

Fig. 8-4. A needle blood pressure transducer.

STATHAM INSTRUMENTS INC. - MODEL SP25
OVERALL LENGTH ≈ 9"
OUTPUT WITH 7.5VRMS EXCITATION = 37μVRMS/mmHg

Fig. 8-5. A syringe blood pressure transducer.

The catheter/transducer combination may be replaced by a needle and transducer as shown in Fig. 8-4. The needle/transducer combination is also available in a syringe configuration as shown in Fig. 8-5. These combinations, using needles rather than flexible catheters, are obviously far easier to introduce into the subject's circulatory system and are thus preferable for direct blood pressure measurement. They are normally introduced into the femoral artery near the groin as this artery is close to the surface at this point. The syringe configuration allows administration of an arterial injection while recording arterial pressure.

needle/
transducer

For measurement of blood pressure within the heart, very small diameter (1 millimeter) catheters, known as drift catheters, are required so as not to interfere with the normal operation of the heart. These small catheters are inserted into the venous system and "drift" with the blood flow, eventually reaching the heart. They may actually be passed through the heart into the pulmonary circulatory system. The primary disadvantage of these small catheters is the damping effect they have on the pressure pulse waveform which limits the high frequency response of the system and gives erroneous readings. To avoid this problem, the catheter tip transducer shown in Fig. 8-6 incorporates a small pressure transducer built into the end of the catheter, the combination being less than 2 millimeter in diameter. This allows introduction of the actual transducer into the heart and thus avoids the damping problem inherent with small catheters.

drift
catheters

catheter tip
transducer
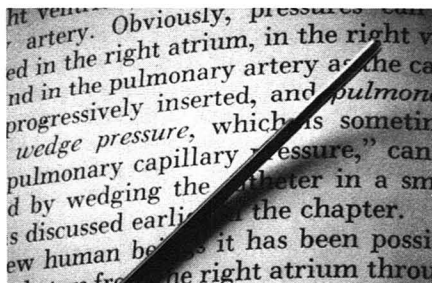
STATHAM INSTRUMENTS INC. - MODEL P866
TRANSDUCER TIP - .065" DIAMETER
OUTPUT WITH 7.5VRMS EXCITATION = 5µVRMS/mmHg

Fig. 8-6. A catheter tip pressure transducer.

bridge-type
transducers

Recording from transducers is covered in more detail in Chapter 18. The transducers shown in this chapter are all bridge-type transducers and, when excited at their maximum rated excitation voltage, they produce an output in the low millivolt or microvolt region, the smaller transducers in general producing less output. While it is possible to use DC excitation and record the output directly using a high sensitivity amplifier, superior results are obtained using a carrier amplifier system such as the Tektronix 3C66 plug-in.

micro-
electrode
pressure
transducer

Extremely small direct blood pressure "transducers" have recently been developed for research work. These transducers have an effective diameter of 10 microns. The transducer consists of a 10 micron microelectrode filled with an electrolyte having a different resistivity than that of blood or body fluids. A pressure system is connected to the microelectrode and the impedance between the center of this electrode and the body is monitored. A servo control system controls the pressure in the microelectrode driving pressure system to keep the impedance between the electrode and the body constant by exerting a pressure on the electrolyte in the microelectrode equal to the pressure external to the microelectrode tip. This pressure exerted on the electrolyte is therefore a measure of the blood pressure at the tip of the microelectrode.

measurement
terminology

Arterial pressures are normally measured as millimeters of mercury, mm/Hg, and venous pressures are usually measured as centimeters of water, cm/H_2O. These units are derived from older blood pressure measuring techniques using either mercury or water manometers. A typical arterial blood pressure waveform is shown in Fig. 8-7. The peak pressure is referred to as the systolic pressure and the minimum pressure is referred to as the diastolic pressure.

It is common terminology to refer to an arterial blood pressure of, say, 120 millimeter systolic and 80 millimeter diastolic as "120 over 80" or 120/80. As stated in Chapter 2, the pressure in the arteries is many times greater than the pressure in the veins, typical arterial pressure being 120/80 and typical venous pressure being 9/5 mm/Hg or, in the more common terminology, 12/7 cm/H_2O.

While direct blood pressure is relatively easy to measure, it does require an incision in an artery for introduction of a catheter or insertion of a hypodermic needle. Since this often results in the permanent loss of the artery used, it is regarded with some disfavor in most countries except the U.S.A. Results accurate enough for clinical use are usually obtainable by indirect methods.

Fig. 8-7. Arterial blood pressure waveform.

SPHYGMOMANOMETER

| PNEUMATIC CUFF | PRESSURE GAGE | RELEASE VALVE | HAND PUMP |

STETHOSCOPE PLACED AT THE ELBOW OVER THE BRACHIAL ARTERY
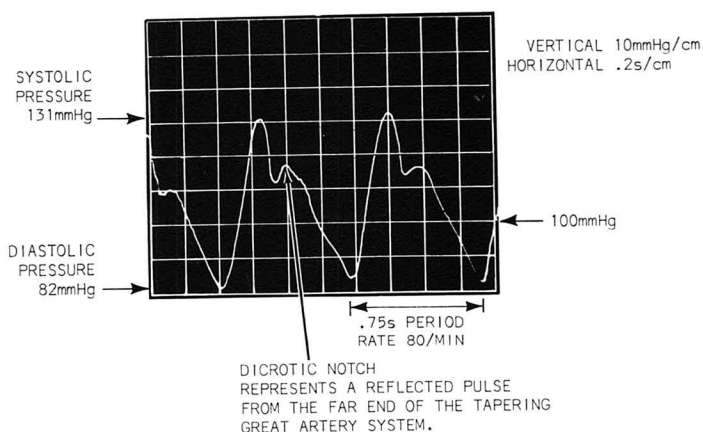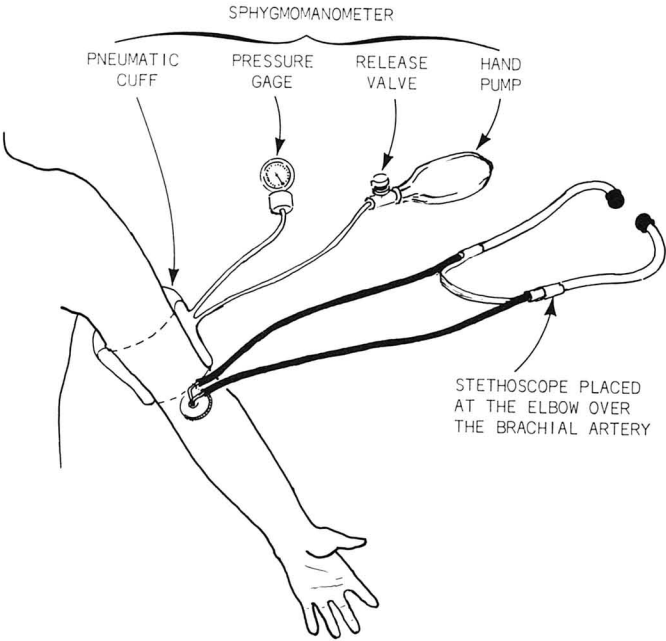
Fig. 8-8. Indirect blood pressure measurement with a sphygmomanometer.

8.2 INDIRECT BLOOD PRESSURE MEASUREMENT

sphygmoma-
nometer

By far the most common form of blood pressure measurement is the indirect measurement using the familiar pressure cuff, hand pump and pressure dial device, used by all physicians, referred to as a sphygmomanometer. The sphygmomanometer, as shown in Fig. 8-8, incorporates a pneumatic cuff encircling the upper arm. An inflatable section of this cuff is inflated by a small hand pump and the pressure in the system is indicated by a mechanical pressure gauge or, in some models, a mercury manometer. The cuff is inflated to a pressure greater than the blood pressure in the large brachial artery of the arm. This pressure thus collapses the artery and occludes (cuts off) blood flow to the arm. As the pressure in the cuff is gradually released using a release valve built into the hand pump, a point is reached where the cuff pressure and the peak or systolic arterial pressure are the same. At a pressure slightly below this level the peak arterial pressure slightly exceeds the cuff pressure and blood is able to squirt through the compressed segment of the brachial artery. This squirting blood results in turbulence within the artery creating sounds known as

Korotkoff
sounds

"Korotkoff" sounds. These sounds are usually detected with a stethoscope placed over the brachial artery. As the pressure in the cuff is further decreased, Korotkoff sounds continue until a point is reached where no further turbulence is produced as no constriction exists in the brachial artery. This point represents the diastolic blood pressure. As it is somewhat difficult to detect the pressure where the Korotkoff sounds begin and cease, this sphygmomanometer technique cannot be relied upon to produce an accuracy of much better than about 10 millimeters of mercury. While the technique is inaccurate, it is simple to perform and very little discomfort is felt by the patient. In the hands of a skilled operator highly repeatable results are obtained and, since the clinician is usually more interested in trends than exact numbers, the technique is entirely appropriate.

PUMP MAY BE OPERATED FOR A SINGLE CYCLE
OR MAY BE PROGRAMMED TO CYCLE AT PRESET
INTERVALS.

A CONTACT CLOSURE DURING INFLATION CAN
BE USED TO ERASE A STORAGE OSCILLOSCOPE.

PIEZOELECTRIC
CRYSTAL PROVIDES
HIGH OUTPUT LEVELS
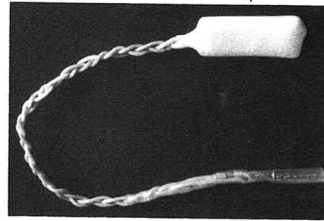
E & M INSTRUMENT CO. INC. 92-201-70

E & M INSTRUMENT CO. INC. 95-300-70
Fig. 8-9. An automatic
cuff pump.

Fig. 8-10. A Korotkoff sounds
microphone.

automatic
measuring
equipment

The sphygmomanometer technique may be automated by replacing the hand pump with an automatic cuff pump as shown in Fig. 8-9. The automatic cuff pump may be operated by pushing a panel-mounted button to produce a single cycle of inflation and deflation or it may be set for repeat cycles at various intervals for continuous monitoring of blood pressure over long periods of time. The stethoscope may be replaced by a Korotkoff-sound microphone as shown in Fig. 8-10. This microphone consists of a small piezoelectric transducer specifically designed to efficiently reproduce Korotkoff sounds. The pressure indicating dial may also be replaced with a pressure transducer similar to the transducers used for direct pressure measurement.

using an
oscilloscope

The subject's indirect blood pressure may be conveniently recorded using the system shown in Fig. 8-11. The vertical channel of an oscilloscope displays the output from a Korotkoff sound microphone while the horizontal channel displays the output from a pressure transducer. A typical display produced by this system is also shown in Fig. 8-11. The horizontal axis is calibrated in pressure; the points at which vertical information appears and then disappears are the systolic and diastolic pressures. This system is capable of somewhat greater accuracy than the conventional cuff/ stethoscope as it removes human judgment in determining the presence of Korotkoff sounds. In some cases clearer displays have been obtained by rejecting all information from the microphone below 150 Hz using a high-pass filter between the microphone and the oscilloscope vertical channel. The prime frequency content of this information appears to be between 400 Hz and 500 Hz.

KOROTKOFF
SOUNDS MICROPHONE

PNEUMATIC
CUFF

564B STORAGE SCOPE

EITHER
AUTOMOTIVE
CUFF PUMP

OR

HAND PUMP WITH
RELEASE VALVE

PRESSURE
TRANSDUCER

VERT

HORIZ

2A63
VERTICAL AMP

AC 50mV/DIV

3C66
CARRIER AMP

10mmHg/DIV

200
µSTRAIN/DIV

REMOTE ERASE FROM CONTACT
CLOSURE WITHIN THE CUFF PUMP

DIASTOLIC 85mmHg

SYSTOLIC 143mmHg

CRT
DISPLAY

50mV/DIV
AC COUPLED

VERTICAL − OUTPUT FROM
KOROTKOFF SOUNDS MICROPHONE

60   100   160

← 10mmHg/DIV →
HORIZONTAL − PRESSURE

Fig. 8-11. Automatic indirect blood pressure measurement.

8.3   INDIRECT RELATIVE BLOOD PRESSURE MEASUREMENT

In many cases, it is unnecessary to measure the
absolute value of arterial blood pressure; all that
is required is an indication of blood flow throughout
the body. If blood flow throughout the body
diminishes for some reason, the principal areas of
the body deprived will be the fingers and the toes,
thus monitoring the presence of blood flow in these
areas will insure that blood is indeed flowing
throughout the body. The simplest technique of
recording the presence of blood flow in the
peripheral arteries is to use a plethysmograph.
Plethysmography is the art of monitoring the
physical changes in size of part of the body as
modified by the flow of blood within it. Various
techniques are used to detect this change in size.
The Pulse Sensor utilizes a photoelectric technique.
Other techniques, such as impedance measuring, are
also used to indicate relative change in size.

plethysmo-
graphy

Pulse
Sensor

The Pulse Sensor uses a light source and
photodetector to record a change in opacity of the
flesh as blood is pumped through it. This Pulse
Sensor relies on light being reflected by some
reflecting medium such as bone, etc. For adequate
output potentials to be obtained, a fairly high
concentration of arteries near the surface is
required. The Pulse Sensor produced by Tektronix
for use with the Type 410 Physiological Monitor is
ideally suited for application to the finger and is
less suitable for toe, ear or nose use. If the
subject is in a drugged state, or if he is
particularly cold, then vasoconstriction will occur
which tends to limit blood flow to the limbs. Under
these conditions it may be preferable to tape the
Pulse Sensor to the forehead of the subject. It is
imperative that the Tektronix Pulse Sensor be used
with resting subjects and, as it detects changes
in physical size, it is definitely not suited to
exercising subjects. The Tektronix Pulse Sensor is
shown in Fig. 8-12, and is applied to the finger
and forehead as shown in Fig. 8-13. A typical
plethysmogram recorded with this Pulse Sensor and
the Tektronix Type 410 Physiological Monitor is
shown in Fig. 8-14.

Tektronix
Pulse Sensor

FINGER PLACED IN
CONTACT WITH
WINDOW IN SENSOR

PHOTO RESISTOR   LOW POWER LIGHT SOURCE

RESISTOR AND LIGHT SOURCE
ARE SEEN THROUGH "WINDOW"
IN THE PULSE SENSOR

LIGHT FROM THE
BULB IS REFLECTED
BY BONE TO THE
PHOTO RESISTOR

TO AMPLIFIER

TO ±6 VOLTS   VOLTAGE REGULATOR
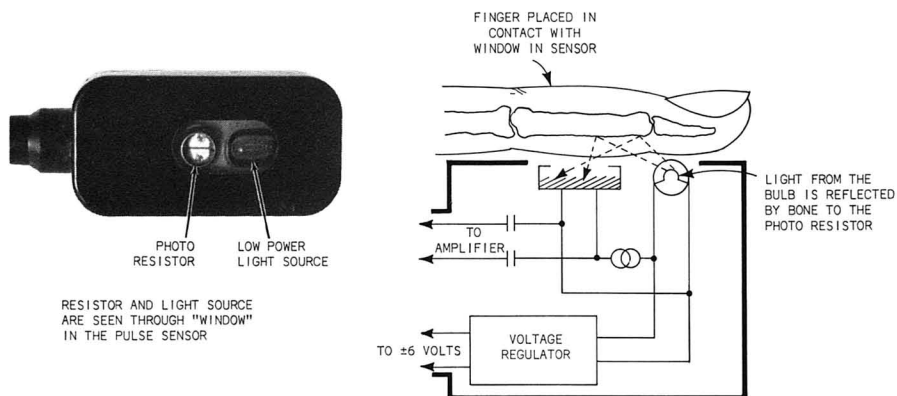
Fig. 8-12. The Tektronix plethysmograph (pulse sensor).

PLETHYSMOGRAPH HELD IN
PLACE BY TAPE OR BY AN
ELASTIC STRAP AROUND
THE HEAD.

PLETHYSMOGRAPH HELD IN PLACE
BY THE TEKTRONIX FINGER HOLDER.
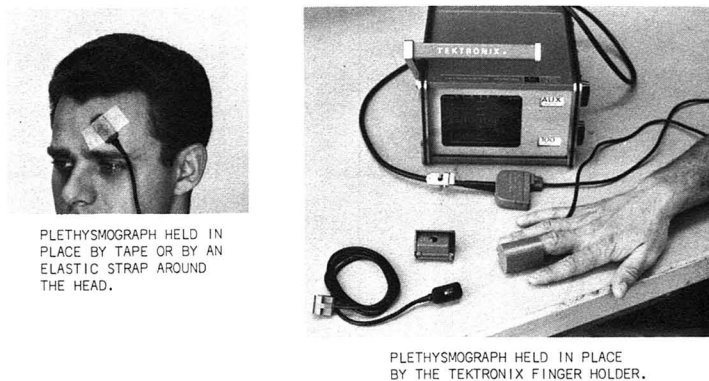
Fig. 8-13. Placement of the Tektronix plethysmograph.

RATE 78/MIN

VERTICAL 2mm/mV
HORIZONTAL 50mm/s

DICROTIC
NOTCH

410 TRIGGER CIRCUIT TRIGGERS DISPLAY.
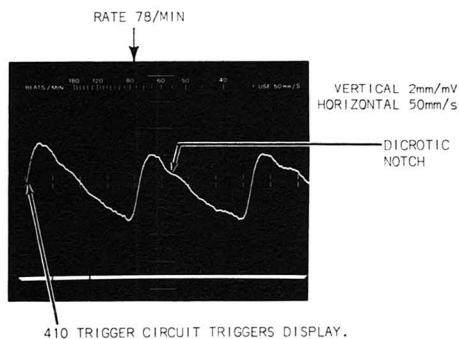
Fig. 8-14. Plethysmogram obtained with Tektronix Pulse Sensor
(attached to finger) and 410 Monitor.
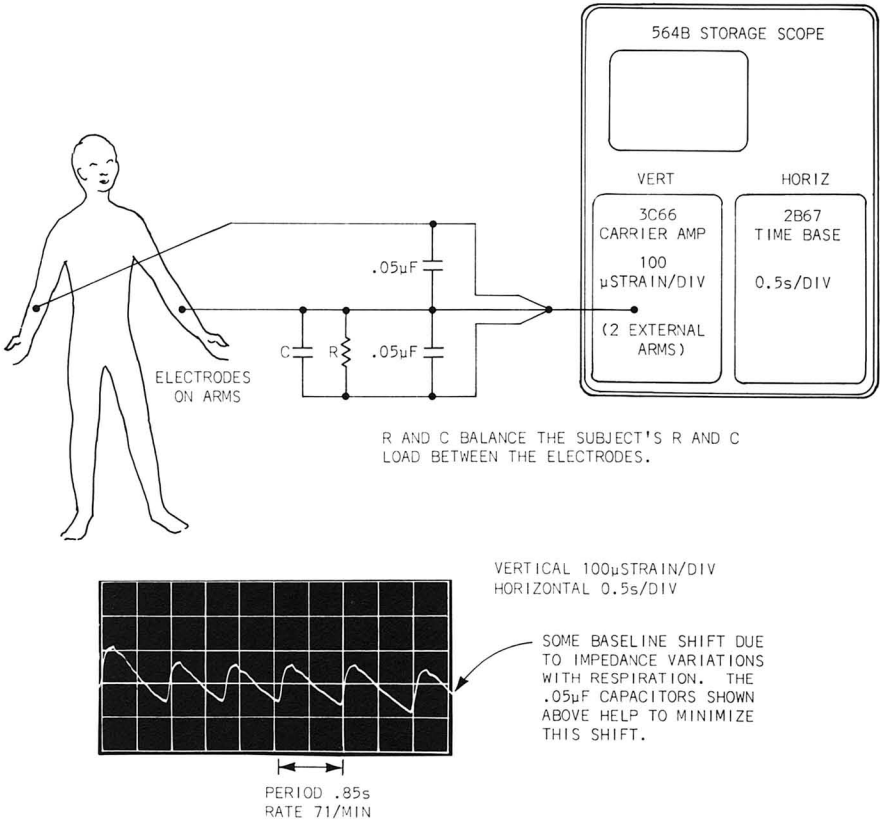
Fig. 8-15. Impedance plethysmography with Tektronix 3C66 Plug-In.

impedance
measuring

As stated above, impedance measuring techniques
may also be used to indicate relative changes in
size resulting from blood flow. If the impedance
between the arms is monitored, it will be found
that the impedance changes due to the action of
the heart and due to respiration. Extensive studies
have shown that the sensation level of arm to arm
electrical current increases with frequency: some
sensation may be felt with 1 mA of DC current
however, at about 25 kHz, ten times this current is
required to produce sensation. Thus audio frequency
currents of at least an order of magnitude less than
1 mA are used to monitor impedance. These
frequencies and low currents also avoid cell
stimulation. It has been found that when measuring
the impedance between the arms, cardiac action
primarily changes the resistive component of this
impedance and respiratory action primarily changes
the capacitive components. Thus, to detect cardiac
action, one must use an audio frequency system
sensitive only to resistive changes, to avoid
respiratory effects.

Such a system can be incorporated using the
Tektronix Type 3C66 Carrier Amplifier with a storage
oscilloscope and time base as shown in Fig. 8-15.
With this system capacitive impedance changes are
almost completely eliminated by paralleling the
impedance to be measured with a large fixed
capacitor. A similar capacitor is required to
balance the bridge, as is a balancing R and C to
simulate the R and C of the subject. The 3C66 is
operated in a Wheatstone Bridge configuration with
two of the arms of the Wheatstone Bridge external
to the instrument, the other two internal. A
typical impedance plethysmograph recorded with the
above technique is also shown in Fig. 8-15. The
slight change in base line evident in this recording
is due to the influence of respiration. Further
details on impedance measuring techniques are given
in Chapters 9, 13 and 18.

FLOWMETER

564B SCOPE

RANGE
ZERO

VERT AMP

SWEEP 2B67

SIGNAL VOLTAGE

MAGNET CURRENT

ELECTROMAGNET

MAGNETIC FIELD

SIGNAL SENSING ELECTRODES

BLOOD VESSEL

FLOW

v = MEAN FLOW VELOCITY (METERS/SECOND)

B = FLUX DENSITY OF MAGNETIC FIELD (WEBERS/SQUARE METERS)

l = SPACING BETWEEN SENSING ELECTRODES (METERS)

V_f = VOLTAGE BETWEEN ELECTRODES (VOLTS)

F = FLOW (CUBIC METERS/SECOND)

a = INTERNAL CROSS SECTIONAL AREA OF BLOOD VESSEL (SQUARE METERS)

$\frac{a}{Bl}$ IS CONSTANT FOR A SYSTEM.

$$V_f = Blv$$
$$F = va$$
$$\therefore F = \frac{a}{Bl} V_f$$
$$\therefore F \propto V_f$$

TYPICAL VALUES: B = .02
l = .01
a = .05 × 10^{-3}

V_f = 20µV per cm/s VELOCITY

V_f = 66µV per 100 cc/min FLOW

Fig. 8-16. Electromagnetic flowmeter – theory.

8.4 BLOOD FLOW MEASUREMENT

electromagnetic velocity transducer

Blood flow is measured by placing a mean velocity transducer in an artery having a known cross-sectional area. Blood flow is the product of the mean velocity and area. Many types of mean velocity-sensitive blood flowmeters have been developed; the electromagnetic blood flowmeter is now extensively used. A diagrammatic representation of an electromagnetic blood flowmeter is shown in Fig. 8-16. The theory of electromagnetic flowmeters is based on Faraday's law. When a conductive fluid, such as blood, traverses the lines of force of a magnetic field, an electromotive force is generated in the fluid which is perpendicular to both the magnetic lines of force and the direction of motion of the fluid. This electromotive force is directly proportional to the intensity of the magnetic field, the distance between the sensing electrodes and the fluid velocity.

The electromagnetic blood flow transducer consists of an electromagnet to generate a magnetic field and two electrodes to sense the flow signal. They are encapsulated in epoxy in a form to allow them to fit around the blood vessel. The lumen or inside diameter fixes the cross-sectional area of the

flow-rate measurements

vessel, changing the transducer to a flow-rate measuring instrument although basically it is a velocity transducer. Electrodes make contact with the vessel wall. The flow transducer is connected

flowmeter

to a flowmeter. This flowmeter supplies energizing current to the electromagnet, amplifies the flow signal, discriminates it from artifacts and makes it available for display on an oscilloscope.

The above theory assumes a DC magnetic field which would produce a DC flow signal. Since it would be impossible to differentiate this signal from electrode offset potentials, amplifier drift, etc.,

AC excitation

commercial blood flowmeters use AC excitation; either sinewave or squarewave. When revising the above theory for AC signals a $\frac{\delta B}{\delta t}$ component appears in the output voltage formula. Commercial systems measure the output voltage only during the period

sinewave

when this $\frac{\delta B}{\delta t}$ component is equal to zero. In sinewave excitation, the $\frac{\delta B}{\delta t}$ component is effectively zero only for a relatively short time at the peak of the excitation wave. The voltage sensing circuit must gate on during this time and correct adjustment of

squarewave

the gate is critical. In the squarewave excitation, the $\frac{\delta B}{\delta t}$ component is always zero except during short switching periods, thus larger variations in gate characteristics are permissible.

signal-to-noise ratio

Signal-to-noise ratio is proportional to the peak-to-peak value of the excitation voltage. For sinewave and squarewave excitation of the same peak-to-peak voltage, the power required for sinewave excitation

power requirements

is only 50% of the power required for squarewave excitation. The chief disadvantage of squarewave probes is, therefore, that they are larger and must operate at higher temperatures than sinewave probes to maintain an acceptable signal-to-noise ratio.

frequencies

Both systems require excitation frequencies high enough to permit effective sampling, typically from 200 Hz to 1000 Hz.

commercial flow probes	Some commercial electromagnetic blood flowmeters are shown in Fig. 8-17. The flowmeters are of the intracorporeal type which have a slot opening to allow placement around the blood vessel. These openings are usually fitted with a cover to maintain a constant blood vessel size. Extracorporeal transducers have tubular sleeve or cannula extensions and are applied by cutting the blood vessel and inserting the tubular sleeve in series with this vessel. Blood is in direct contact with the sleeve lumen in extracorporeal units.
commercial flowmeters	Blood flowmeters suitable for use with the above transducers are produced by several manufacturers. In general, these instruments provide either sinewave or squarewave excitation at between .1 and 1 amp and provide an output for use with an oscilloscope or other recording device of between .1 volt and 1 volt. This recording device should have a response from DC to 100 Hz. The instruments are calibrated to read in cubic centimeters per minute and usually cover a range from 1 to 100,000 cubic centimeters per minute. Blood velocity in the pulmonary artery may reach 100 centimeters per second. As this artery has a cross-sectional area of about 1.8 square centimeters, this velocity results in a peak flow of 100,000 cubic centimeters per minute. Blood flow systems may be calibrated by allowing blood to escape into a graduated flask for a known brief period or by using an external calibration system providing a known flow. A zero flow baseline may be obtained by occluding flow for a brief period. A typical blood flow waveform is shown in Fig. 8-18, and a complete measuring system is shown in Fig. 8-19.
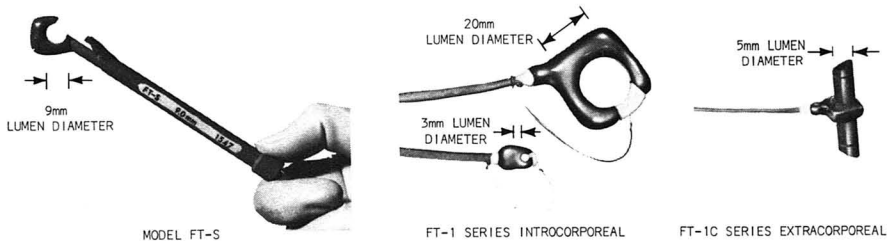
Fig. 8-17. Commercial blood flow meters. (In Vivo Metric Systems, Los Angeles, Calif.)
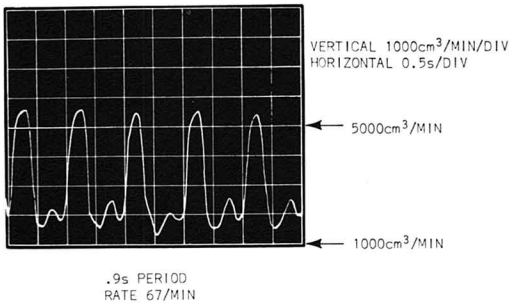
Fig. 8-18. A typical blood flow record from an electromagnetic flowmeter.
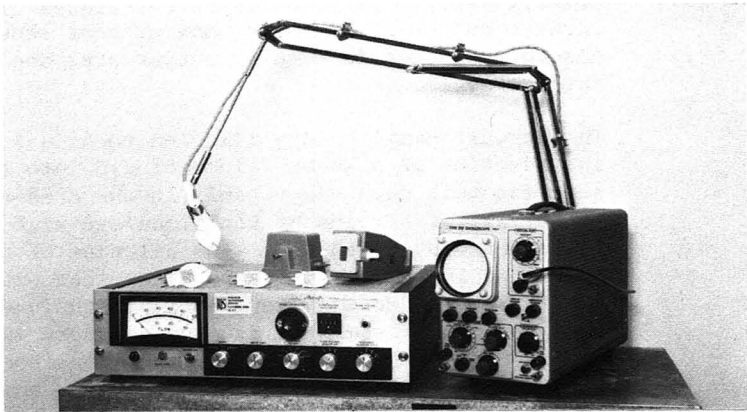
Fig. 8-19. A commercial blood flow measuring system.

| other flow-measuring devices | Other forms of blood flow-measuring devices are occasionally used. The isothermal flowmeter places a thermistor within the blood flow. The flowing blood tends to cool the thermistor and a measurement of blood flows is obtained by recording the increase in thermistor excitation required to maintain the thermistor at a constant temperature (constant resistance). The electroturbinometer inserts a minute rotational generator into the blood flow system; this generator is driven by a small propeller driven by the blood flow. Ultrasonic blood flow measuring techniques are becoming increasingly popular; these techniques involve the detecting of small phase differences between ultrasonic signals propagated in the direction of flow and in the opposite direction. |

8.5 CARDIAC OUTPUT

| direct determination | Blood flow may be measured in almost any part of the circulatory system by using the blood flow measurement techniques discussed previously. When blood flow is measured in the pulmonary artery or the aorta, this blood flow, integrated over a cardiac cycle, represents the total amount of blood flowing through the heart and is referred to as cardiac output. Thus cardiac output may be determined by integrating the results obtained from conventional blood flow measurement at the pulmonary artery or aorta. As this obviously involves critical surgical procedures, cardiac output is more commonly |
| dilution techniques | determined using dilution techniques as discussed briefly below. These dilution techniques give total cardiac output whereas flow measurement shows the discrete changes in cardiac output over one complete cardiac cycle. |

The Stewart-Hamilton dye dilution technique involves introduction of a known volume of dye into the superior vena cava via vessels in the arms and withdrawal of samples of the blood/dye mixture from the aorta via a catheter. The dilution of dye contained in these samples is determined with a densitometer. Modern dye dilution techniques use radioisotopes instead of dye and counters instead of densitometers. The output from the densitometer or counter is coupled to an analog computer which computes cardiac output directly. Cardiac output in an adult male is approximately 5000 cubic centimeters per minute.

8.6 BLOOD VOLUME

Blood volume is measured using a modified dye
dilution technique. A known amount of dye is
introduced into the system and allowed to circulate
for many cycles. After several minutes, a blood
sample is taken and the dilution of the dye in this
sample is noted. Total blood volume is then a
product of the dilution ratio and the original
quantity of dye injected. Blood volume in an adult
male is approximately 5500 cubic centimeters or one
and one-half U.S. gallons.

114
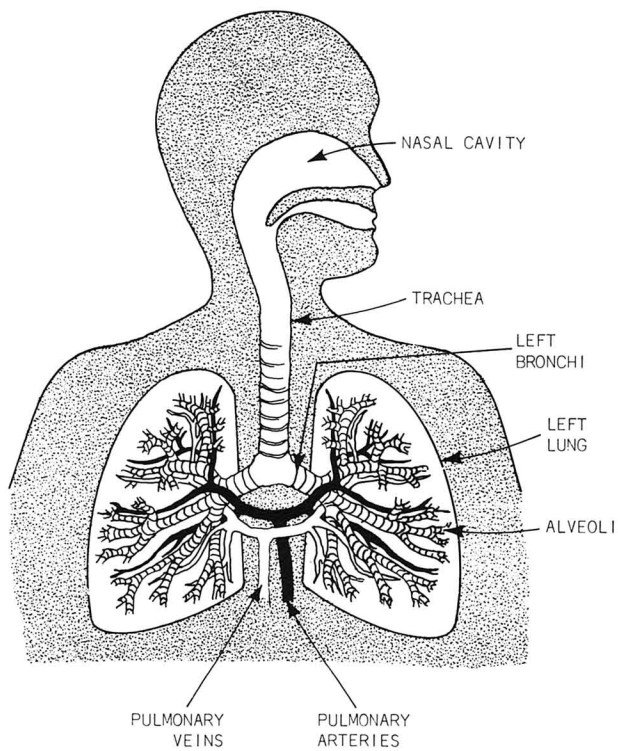
Fig. 9-1. Respiratory system.

⑨

RESPIRATION AND TEMPERATURE

9.1 PHYSIOLOGICAL CONSIDERATIONS

respiratory
functions

The physiology of the respiratory system is not covered in Section I; thus the more important physiological aspects of this system are discussed below. The primary functions of the respiratory system are to oxygenate the blood, i.e., to dissolve oxygen into the blood, and to remove carbon dioxide from the blood. If blood is not oxygenated due to failure of the circulatory system, the oxygen content of the blood will rapidly decrease and after about 60 to 90 seconds the subject will become unconscious, death occurring in 4 to 5 minutes.

lung
structure

The lungs are the major component of the respiratory system and it is in the lungs that oxygenation of the blood occurs. When the lungs are forced to expand by muscular contraction of the diaphragm and expansion of the thoracic cage by contraction of the rib muscles, air enters the lungs via the bronchi and is diverted to millions of small air sacs known as alveoli. The membrane comprising the alveoli is moist and the oxygen contained in the air is dissolved by this moisture. Interspersed with the alveoli are fine capillaries, branching from the circulatory system, through which blood is continually flowing. The oxygen dissolved in the moist surface of the alveoli diffuses into the blood stream via these capillaries. The carbon dioxide contained in the blood stream is also diffused through the alveolar membrane to be expelled with expired air. Some of the principal components of the respiratory system are shown in Fig. 9-1.
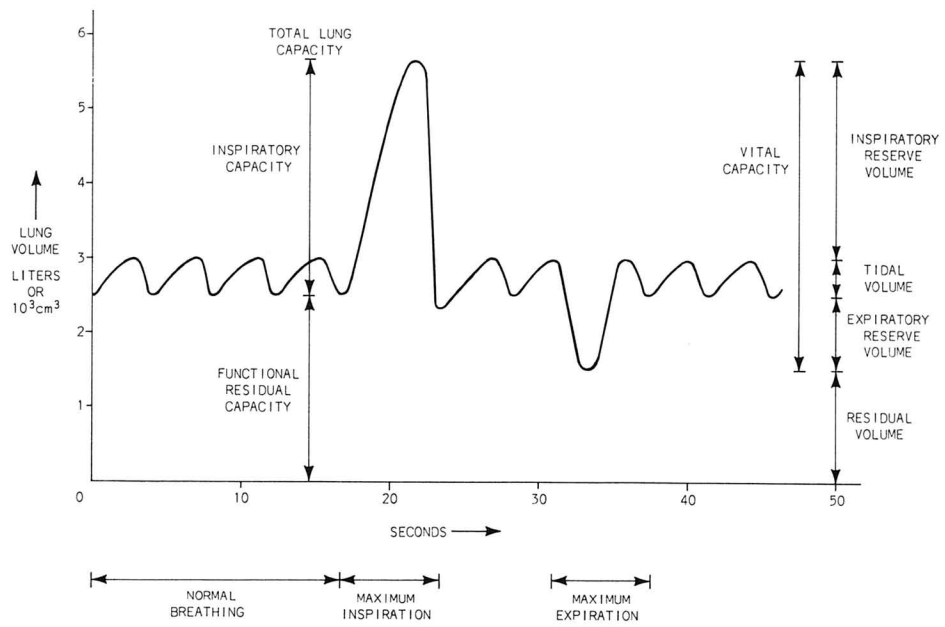
Fig. 9-2. Respiratory changes in lung volume.

	A grown man at rest inhales about 500 cubic
lung	centimeters of air with each breath and takes between
capacity	12 and 18 breaths per minute. During periods of

lung
capacity

A grown man at rest inhales about 500 cubic centimeters of air with each breath and takes between 12 and 18 breaths per minute. During periods of moderate exercise he may inhale one liter or more with each breath and take up to 25 breaths per minute. At maximum exercise the vital capacity of 4 to 5 liters is reached.

lung
volume

The action of breathing is controlled by muscular action causing the volume of the lung to increase and decrease. During normal breathing the lung does not contract to its minimum possible volume, nor does it expand to its maximum possible volume. Minimum lung volume is obtained when exhaling with extreme effort to expel the maximum possible amount of air and maximum lung volume is obtained when inhaling with maximum effort. Fig. 9-2 shows diagrammatically the changing lung volume that may be expected for a resting man and the lung volume that may be achieved during maximum inspiratory and expiratory effort.

spirometer

pneumograph

Lung volume is measured by a spirometer and the recording of lung volume changes with time is known as a spirogram. Instruments that simply detect respiratory activity are referred to as pneumographs and the resulting recording of respiratory activity changes with time is known as a pneumogram. A spirogram is normally only required when attempting to analyze the respiratory system or to detect a malfunction of this system and is rarely used for routine monitoring. A pneumogram may be used for routine monitoring and is basically used to indicate the fact that the subject is breathing. Breathing rate may be obtained from either a spirogram or a pneumogram.

9.2 RESPIRATORY ACTIVITY

Relative respiratory activity may be detected by either detecting the physical changes in the torso associated with breathing or by detecting the flow of air through the nostrils. Since no absolute measurements are required, the measurement techniques involved have been simplified to allow easy application of the devices concerned to the subject.

thermistor
pneumograph

Perhaps the simplest form of pneumograph employs a thermistor placed in the outer nasal passage to detect the temperature differential between inspired cool air and expired warm air. This technique fulfills the majority of clinical needs including those of operative and post-operative subjects. If the subject breaths through his mouth, or if he wishes to converse, the thermistor may be placed in the mouth or in such a position as to detect flow from either the nose or the mouth. The thermistor concerned should be supplied from a constant current source at a low enough current to maintain thermistor self-heating below one degree centigrade or so. Adequate sensitivity can usually be obtained with 5 milliwatts of thermistor dissipation.

temperature
consider-
ations

Excessive thermistor heating can cause subject discomfort, thus thermistor dissipation should be limited to 40 milliwatts for small bead-type thermistors. A thermistor pneumograph suitable for use with a Tektronix 410 monitor, together with a typical pneumogram obtained from this system, is shown in Fig. 9-3. Further details are given in Chapter 29.

If the temperature of the outside air is the same as the temperature of the expired air (body temperature) the above system is unsatisfactory. In this case, enough current should be passed through the thermistor concerned to raise its temperature in still air to a temperature somewhat above body temperature but below a temperature that may cause subject discomfort. This can usually be achieved with thermistor dissipations of between 5 and 25 milliwatts. The flow of both inspired and expired air over this thermistor will tend to cool it, thus producing a resistance change. Since both inspiration and expiration decreases the thermistor temperature, the resulting output will be at twice the respiratory frequency.

410 MONITOR WITH
STANDARD PATIENT CABLE

AUX

25

THERMISTOR   CONSTANT
IN HOLDER   CURRENT SOURCE

THERMISTOR PNEUMOGRAPH (FOR DETAILS SEE CHAPTER 29)

RATE $\frac{44}{2}$ = 22 BREATHS/MIN

VERTICAL AUX 2mm/mV = 5mV/DIV
HORIZONTAL 25 mm/s =.4s/DIV
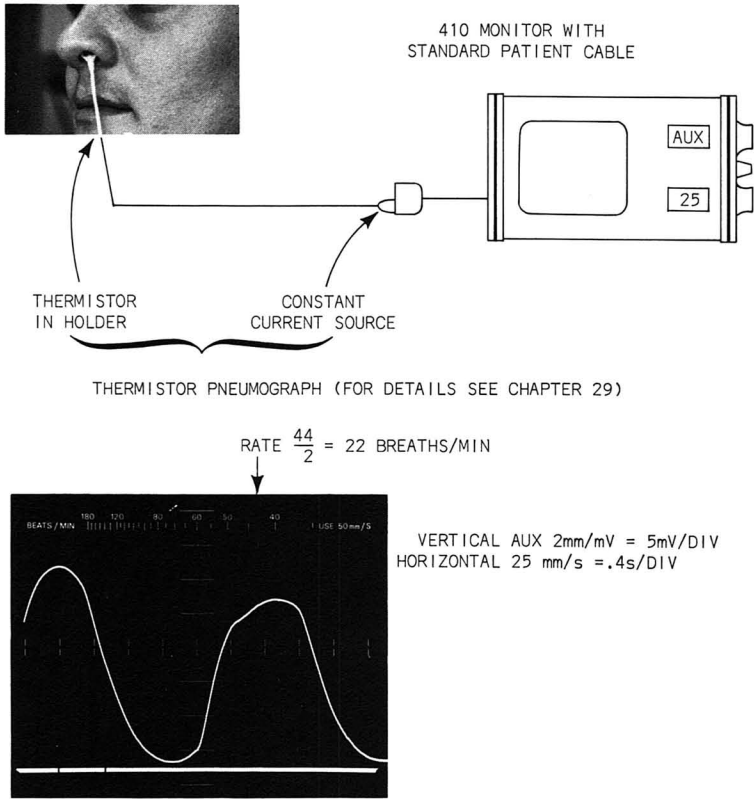
Fig. 9-3. Thermistor pneumogram with the Tektronix 410 Monitor.
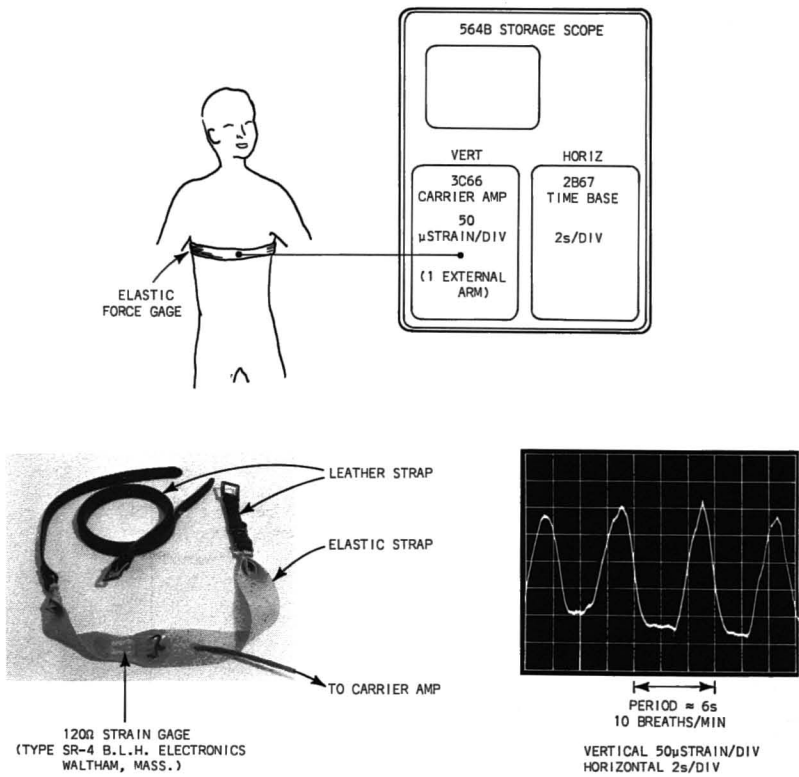
Fig. 9-4. Elastic force gage pneumogram with a Tektronix 3C66 Plug-In.

strain gauge detects chest size changes

Changes in the physical size of the torso with respiration may also be detected to indicate respiratory activity. A strain gauge attached to a piece of elastic may be used to form a band around the chest. Since chest circumference varies with the respiration, the elastic band will stretch causing a change in resistance in the strain gauge. This resistance change may be detected by a DC-excited Wheatstone Bridge or, preferably, by a carrier amplifier such as the Tektronix 3C66 system. Such a system and the resulting pneumogram obtained is shown in Fig. 9-4.

resistance changes with chest circumference

Changes in chest circumference may also be detected by a rubber tube filled with mercury, fastened firmly around the chest. As the chest expands, the rubber tube increases in length and thus the resistance of the mercury from one end of this tube to the other changes. This resistance change may be detected using a constant-current source in the same way as the thermistor resistance change was detected in Fig. 9-3. The principal disadvantage of this mercury-filled rubber tube pneumograph is its extremely low resistance, requiring large currents and sensitive detecting instruments. These disadvantages can be largely overcome by replacing the mercury with a conductive solution, such as copper sulphate solution (with copper plugs in the ends of the rubber tube) or with some of the less-viscous types of electrode paste commonly used to apply electrodes to the skin. Commercial electrode pastes vary greatly in resistivity; however, a 45-inch long 1/8 inch diameter rubber tube filled with electrode paste should have a resistance of between 1,000 and 100,000 ohms.
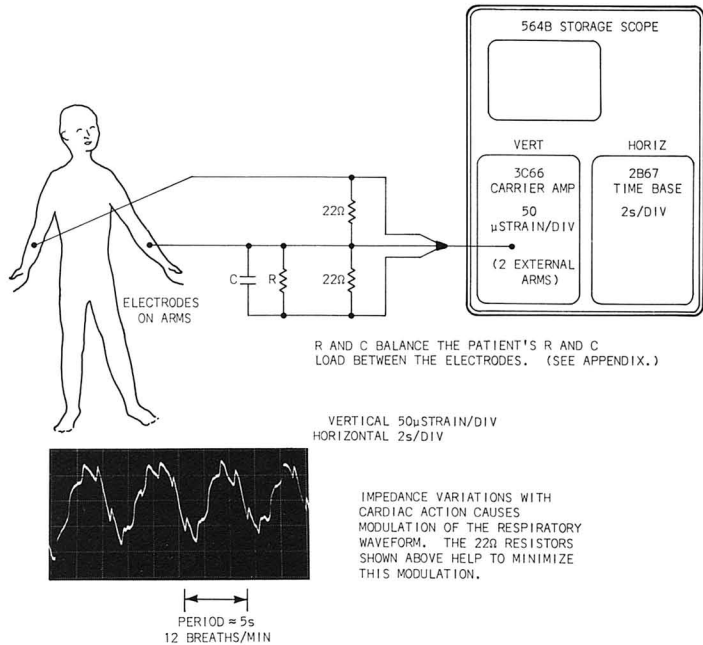
Fig. 9-5. Impedance pneumogram with a Tektronix 3C66 Plug-in.

torso
resistivity

Respiratory activity may also be detected by measuring the change in resistivity across the torso. This technique is similar to the technique used to measure cardiac activity as described in Chapter 8, Section 8.14. However, whereas the resistive components of this impedance are detected to measure cardiac activity, the capacitive component of this impedance is detected to indicate respiratory activity. A typical impedance pneumograph system together with a typical impedance pneumogram obtained from this system is shown in Fig. 9-5. There appears to be little practical use for this system as the variations produced by the heart in the record prevent a good recording from being obtained.

9.3 RESPIRATORY AIR FLOW

pneumotach

Respiratory flow is invariably measured with a pneumotachograph (commonly referred to as a pneumotach), consisting of a hydraulic resistance head, and a differential pressure transducer. The pneumotach consists of a one-inch diameter tube containing a fine mesh screen, as shown in Fig. 9-6.

This mesh screen offers slight resistance to air flow; this resistance to flow produces a pressure differential across the mesh screen which is proportional to the mean flow velocity. This pressure differential is detected by a differential pressure transducer. The sensitivity of the device can be varied by varying the size of the mesh screen or by using several mesh screens, however total airway resistance offered to a subject should never exceed about 1 cmH$_2$0 if normal respiration is not to be affected by the measuring device. A typical pressure differential across the screen would be 0.09 cmH$_2$0 per 10 liters per minute flow using 2 inch diameter, 400 per inch, stainless steel gauze in the head.

measurement
system
calibration

The differential pressure transducer can be used in conjunction with a Tektronix Type 3C66 Carrier Amplifier in a Tektronix Type 564B Storage Oscilloscope. The complete respiratory flow measuring system, i.e., pneumotach, differential pressure transducer, 3C66 carrier amplifier and oscilloscope, may be calibrated with a known flow.
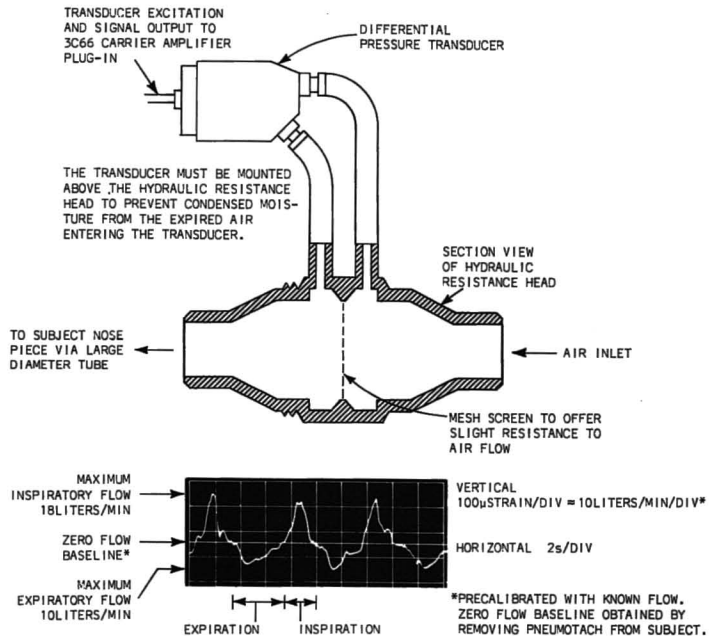
Fig. 9-6. Pneumotach air flow transducer.

FROM
564B CALIBRATOR

≈13mA

1MΩ

10kΩ

10kΩ

① FLOW

PRESSURE
TRANSDUCER

1μF

1MΩ

② RESPIRATORY
VOLUME

PNEUMOTACH

DIFFERENTIAL
PRESSURE
TRANSDUCER

SIGNAL

EXCITATION

564B STORAGE SCOPE

VERT   HORIZ

3A8   2B67
OPERATIONAL   TIME BASE
AMP

CAL OUT
40V DC

1MΩ

FROM
TRANSDUCER

1kΩ

1kΩ

① FLOW
(INVERTED)

1μF

10MΩ

FROM 564B
CALIBRATOR

22MΩ

② DC LEVEL
ONE MINUTE AFTER
SWITCH IS OPENED
GIVES VOLUME

BOTH DIODES
GE-SE416
TEK P/N 152-0324-00

CONFIGURATION FOR MINUTE VOLUME MEASUREMENT.

STORAGE DISPLAY SHOWING
BOTH FLOW AND LUNG VOLUME
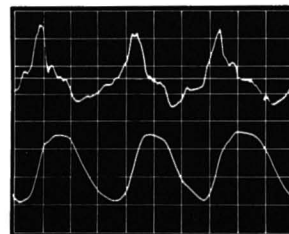
VERTICAL ① FLOW (REFER TO FIG.6)
10LITERS/MIN/DIV

HORIZONTAL 2s/DIV

VERTICAL ② LUNG TIDAL VOLUME
WAVEFORM 1 INTEGRATED
0.2LITERS/DIV
(PRECALIBRATED WITH
KNOWN FLOW FOR 5s.)

Fig. 9-7. A spirogram obtained with an integrating pneumotach.

Commercial pneumotachs usually provide a calibration
of the differential pressure produced per 10 liters
per minute flow. This information, when related to
the sensitivity of the pressure transducer, can be
used to provide a calibration factor, as indicated in
Chapter 18. A typical flow pneumogram is shown in
Fig. 9-6. A zero base line was added to this display
to provide zero flow reference by simply removing
the subject from the pneumotach system and adding an
additional sweep to the previously stored pneumogram.

Respiratory air flow measurement is frequently used
to estimate a subject's respiratory function. Flow
measurement also allows respiratory volume to be
easily obtained.

9.4 RESPIRATORY VOLUME

Since air flow is simply a measurement of volume per
unit time, respiratory air flow information may be
integrated to provide respiratory volume. Such a
system is shown in Fig. 9-7. A pneumotach and
differential pressure transducer produces an output
proportional to respiratory flow as discussed in the
previous section. To simplify the instrumentation
requirements, the pressure transducer is operated
from a ≈13 milliamperes DC source (ocilloscope
Calibrator set to 40 V DC) and the resulting output
amplified by one of the operational amplifiers in
the Tektronix Type 3A8 Operational Amplifier. The
amplified flow signal is then integrated using the
second operational amplifier in the 3A8 unit. The
output from this operational amplifier is thus an
indication of respiratory volume. With the above
system, the resistors and capacitors associated with
the operational amplifiers can be selected from the
front panel of the 3A8, with the exception of one of
the 10,000 ohm resistors in the amplifier which must
be added between the appropriate terminals on the
front of the 3A8.

transducer
signal
amplified
then
integrated

Minute respiratory volume may be measured by
modifying the above procedure. Minute respiratory
volume is the amount of air that a subject inhales
in a one minute period. It may be measured by
integrating *inspiratory flow* only, over a one-minute
period. Referring to Fig. 9-7, the output of the
second operational amplifier in the 3A8 unit will
register minute volume if the flow signal is coupled
to the integrating circuit via a diode and the
integrator is gated ON for a one-minute period.

minute
respiratory
volume
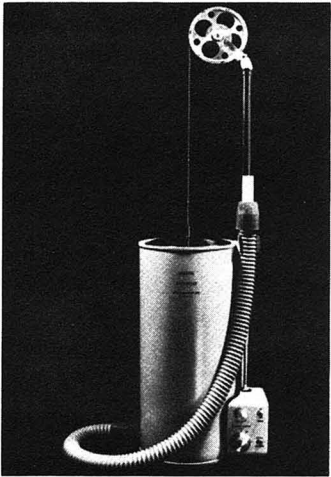
A COMMERCIAL SPIROMETER

PULLEY — CALIBRATION

OUTPUT CABLE

HIGH LINEARITY
POTENTIOMETER
COUPLED TO PULLEY
PROVIDES RESISTANCE
CHANGE PROPORTIONAL
TO BELL MOVEMENT.

BELL

TO SUBJECT'S
MOUTH OR
NOSEPIECE

COUNTERWEIGHT

FLEXIBLE
HOSE

WATER

CONTAINER
(6 LITER CAPACITY)
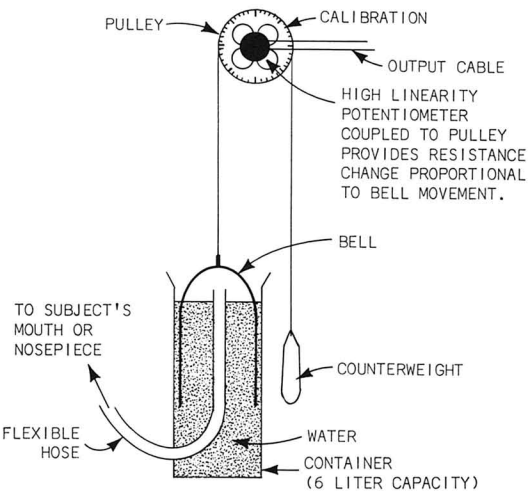
SPIROMETER — FUNCTIONAL DETAILS

INSPIRATION RESERVE VOLUME
2.5 LITERS

TIDAL VOLUME
0.6 LITERS

EXPIRATION RESERVE VOLUME
0.8 LITERS

VERTICAL LUNG VOLUME
0.5 LITERS/DIV
HORIZONTAL 2s/DIV
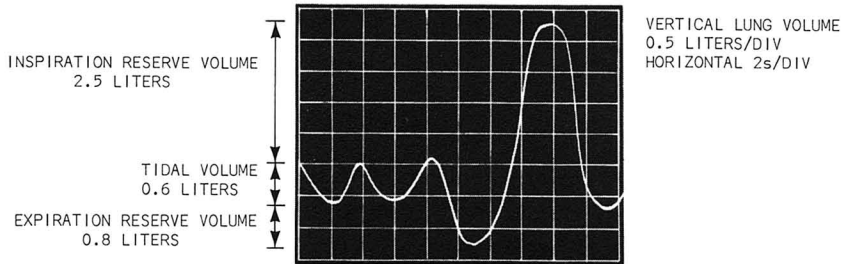
Fig. 9-8. A spirogram obtained with a spirometer.

128

YELLOW SPRINGS INSTRUMENT CO. TELE-THERMOMETER
WITH AN OUTPUT FOR A RECORDER.
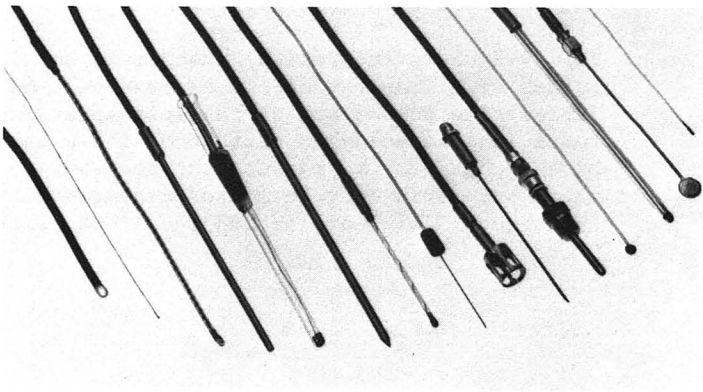
VARIOUS THERMISTOR PROBES AVAILABLE
FROM YELLOW SPRINGS INSTRUMENT CO.

Fig. 9-9. A commercial thermometer for use with thermistors.

9.5 TEMPERATURE

thermometer
using a
thermistor

In most cases, temperature does not vary at any appreciable rate and, thus, may be displayed using a moving coil meter. Such a thermometer, using thermistors in conjunction with a moving coil meter, is shown in Fig. 9-9. For some applications, however, the meter display is inadequate and thus an oscilloscope or chart recorder is required. The meter shown in Fig. 9-9 provides an output for use with an oscilloscope. Perhaps the main application where temperature does vary rapidly is in the recording of respiration, as covered earlier in this chapter.

While other temperature sensing devices, such as thermocouples, are feasible they are rarely used and small thermistors are used almost exclusively for temperature detection. These thermistors are available in a wide variety of sizes and mounting styles as shown in Fig. 9-9.

10

ELECTROENCEPHALOGRAPHY

The following is necessarily a short and therefore
somewhat incomplete survey of the EEG and EEG
measuring techniques. Interested readers will find
W. Grey Walter's text, *The Living Brain*, 1953,
Duckworth and Co., invaluable for further study.

Electroencephalography, conveniently abbreviated
EEGy, is the study of the electrical activity of the
brain. Usually this activity is recorded from
electrodes placed on the scalp, although some
relatively rare diagnostic procedures require
electrodes on or beneath the cerebral cortex. The
EEG has been known for some 40 years and has made
many contributions to man's knowledge of brain
function. For reasons which are examined later, it
has been of greater help to neurology (the study of
brain function) than to psychiatry (the study of
mental processes).

10.1 THE CHARACTERISTICS OF THE NORMAL EEG

The EEG of a normal adult human, normal being used
in the everyday sense of the word, is relatively
easily described. When the subject is relaxed, but
not drowsy, a relatively smooth oscillation, whose
frequency is seldom less than 8 Hz or more than 13 Hz,
can be recorded from the area of scalp immediately
over the occipital lobes. (Refer to Chapter 4,
α rhythm   Fig. 4-2.) Typically this oscillation, the α rhythm,
has an amplitude of 50 μV peak-to-peak, although
in rare subjects it may be twice this amplitude and
in about 10% of the population it is absent or very
small. This rhythm is responsive to mental activity;
in most subjects attempting a task such as mental
arithmetic will attenuate or abolish it.
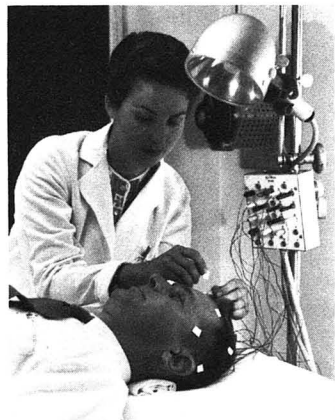
(A) ELECTRODES ARE APPLIED TO THE SCALP AND PLUGGED INTO THE JUNCTION BOX.

(B) A SWITCH SELECTOR ALLOWS THE DESIRED ELECTRODE CONFIGURATION TO BE CHOSEN.
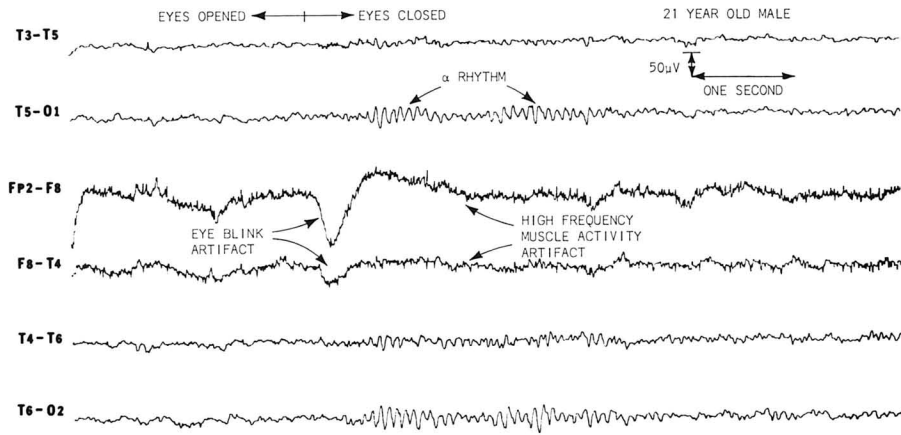
(C) A SEGMENT OF THE RECORD OBTAINED SHOWING SIX OF THE SIXTEEN CHANNELS RECORDED.

Fig. 10-1. A typical adult EEG from a normal subject.

recorder

Most EEGs are recorded using multi-channel ink writing oscillographs, as shown in Fig. 10-1, historically because they were widely available to physiologists. If some more sophisticated method of display is used it is found that more than one generator is involved, that there are generally several different frequencies, that there are differences in responsiveness between the cerebral hemispheres and that often the frequency of the signal measured in a transverse plane is consistently different from that observed with laterally placed electrodes.

frequency

Frequency information is particularly significant since the basic frequency of the EEG varies greatly with different behavioral states. To assist in the EEG analysis, the normal frequency range of the EEG (.5 Hz to 30 Hz) has been subdivided into five bands:

delta	(δ).	0.5 Hz	-	4 Hz
theta	(θ).	4 Hz	-	8 Hz
alpha	(α).	8 Hz	-	13 Hz
beta	(β).	13 Hz	-	22 Hz
gamma	(γ).	22 Hz	-	30 Hz

Various techniques for signal display are discussed later in this chapter. Although the α rhythm is the most prominent activity in the EEG of healthy adults, it is not seen in very young children and its absence does not indicate a lack of mental health or any deficiency in intelligence.

A segment of an EEG record from a normal adult male is shown in Fig. 10-1. Six of the 16 channels commonly recorded by an EEG instrument are shown. The tracing is read from left to right. Initially, the subject's eyes were open but after about 2.5 seconds he was asked to close them. The large downward deflection in leads FP2 - F8 and the

artifacts

smaller one in F8 - T4 are the "eye blink artifact". The α rhythm can be seen in the occipital channels T5 - 01 and T6 - 02 after the eyes were closed. Although the subject was completely normal, the α rhythm is somewhat smaller and less persistent than usual. The high frequency component in the two middle tracings is an artifact due to muscle activity and is not from the brain. The EEG shown in Fig. 10-1 represents only about eight seconds of recording, however, in practice, a recording may be maintained for an hour or more, producing a vast quantity of information for analysis.

10.2 EEG ELECTRODE CONSIDERATIONS

From an engineering standpoint the design of an EEG instrument and its accessories (electrodes etc.) is nowadays a routine matter requiring little more than ordinary care and attention to detail. As is so often the case in electronic design, the overall system limitations are almost all in the input devices, (the electrodes) which interface the equipment to the subject, and in the methods of storing the output data.

input electrodes

The input electrodes are the most critical components of the recording chain. To be of use for routine EEG recording they must be small, be easily affixed to the scalp with minimal disturbance of coiffure, cause no discomfort and remain in place for extended periods of time. They must also have some fairly rigid electrical specifications if the signals are to be recorded with acceptably low levels of distortion.

We have noted that many EEG signals are of microvolt levels and it should be remembered that the signal is arising not at the scalp but in the cerebral cortex which is separated from the scalp by the cerebral spinal fluid (in which the brain is suspended) and by the skull. Parenthetically, we should note that engineers often suppose the skull to be an insulator because they usually see it dried and mounted. The living brain, however, is encased in living bone which is well permeated with conducting fluid. The amplifying system thus sees

signal sources

signals which arise in generators which have large, complex and variable source impedances. There may be large electrode offset potentials of the order of many millivolts developed between the electrode and the scalp unless a suitable electrode material is used. The high common-mode rejection ratio of the modern EEG amplifier will cancel the common-mode part of this signal but in practice small movements of the subject's head can cause substantial variations in the standing potential and if these are different in each lead they will of course appear as differential signals.

line
interference

A further cause of problems in EEG recording is the presence in the modern clinic of many pieces of line operated equipment so that there are substantial magnetic and electric fields at the line frequency. The CMMR of the amplifier can, in principle, reduce these signals to insignificance but only if the entire system, including the electrode impedances, is balanced with respect to the common (ground) point on the amplifier. Thus electrode resistance must be reduced as far as possible; with good technique interelectrode resistances of $1 - 2$ kΩ can be obtained. The alternative technique of reducing the line interference by the use of a shielded cage is not generally satisfactory since the degree of physical isolation it entails can be an emotionally upsetting experience, especially for a child. A relaxed subject is a necessity if good recordings are to be obtained.

electrode
construction
and
connection

The most widely used electrodes are small silver pads electrolytically coated with silver chloride and attached to the scalp with a quick drying adhesive, usually collodion. A harness of rubber straps is also often used to hold the electrodes in place. Before the electrodes are applied, the scalp area is degreased and cleaned with alcohol and the surface resistance reduced by the use of a conducting paste. These electrodes are satisfactory for most recordings in the range $1 - 60$ Hz. If, however, the low frequency limit is to be extended, which is the case in some research applications of the EEG, then electrodes which more closely approximate truly nonpolarizable electrodes, such as Tektronix Ag/AgCl electrodes must be used. Electrodes are generally placed at standard locations on the scalp to facilitate communication between electroencephalographers. These positions, with their usual designations, are shown in Fig. 10-2. The usual abbreviations are: F = frontal, T = temporial, C = central, P = parietal, O = occipital.
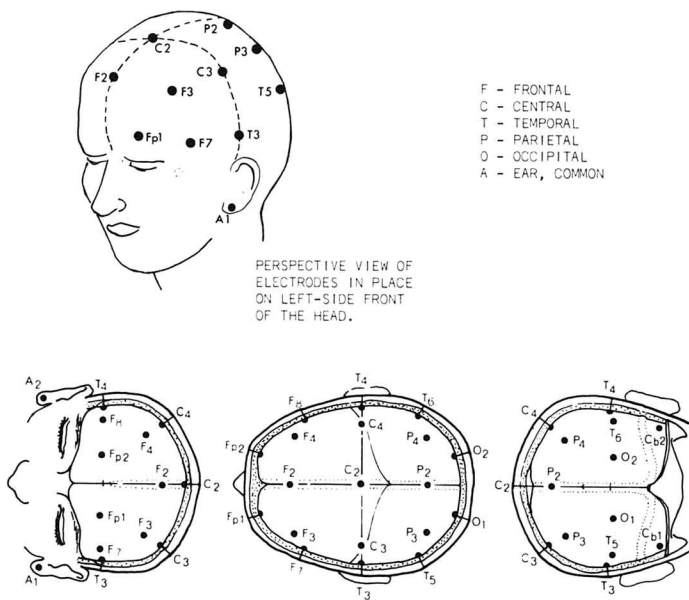
Fig. 10-2. EEG electrode positions.

10.3 EEG RECORDING INSTRUMENTS

oscillographs

Turning now to the output end of an EEG recording system, multi-channel ink-writing oscillographs are used for many reasons. The recording material is relatively cheap, the record is available for inspection as it is being written, the electroencephalographer can quickly flip through a long recording and obtain an "eyeball" impression of its contents and he may study interesting or complex parts of the record for as long as is necessary. Other media such as magnetic tape do not possess these properties and have not become popular in the routine clinical laboratory. If visual analysis is to be supplanted by computer or other automated data-processing techniques, then the written record must be supplemented by a magnetic recording or a curve reader (e.g., the multichannel high-speed curve reader described by Barlow in 1968) must be used. The frequency response of most EEG systems is limited by the characteristics of the recorder to something of the order of 60 Hz but this is adequate for most clinical purposes.

system
frequency
response

EEG recording systems are usually self-contained units consisting of electrode switching networks, high-gain differential amplifiers and graphic recorders.

multichannel   Multichannel recording is almost invariably used, the number of channels ranging between 6 and 32 with 8 or 16 channels being the numbers preferred for routine work.

A multiplicity of electrodes is affixed to the scalp as shown in Fig. 10-2 and the recording channels are connected to them via a switching network. The amplifiers are invariably designed to accept differential inputs and their design is usually optimized for low noise and good common-mode rejection. The low-frequency response usually extends to about 0.1 Hz, however the high-frequency response need not be in excess of about 100 Hz due to the limited high-frequency response of the graphic recorder following the amplifier. Since most EEG activity occurs below 50 Hz, a notch filter tuned to

line   line frequency is often included in EEG instruments
frequency   to minimize line frequency interference, but its use
filter   is strongly discouraged except as a last resort.

Although the gain of most modern EEG instruments is stable to within a few percent, a 50 μV squarewave calibrator may be included in the instrument. Although the sensitivity of the amplifier may be adjusted to suit particular subjects, the electroencephalographer rarely changes the

sensitivity   sensitivity during the recording of an EEG. He usually selects a gain that makes the initial record "look right" and uses the same gain throughout all phases of recording the EEG. Since the electroencephalographer is concerned with relative amplitudes between the channels, it is necessary that each channel has the same sensitivity. It is desirable to standardize on both sensitivity and paper speed to achieve aspect ratio consistency, which allows comparison with other EEG's recorded from other subjects. No firm standard exists here, however many workers prefer a sensitivity of about 7 mm per 50 μV for adult subjects, a somewhat lower sensitivity for children and a somewhat higher sensitivity for the aged. The range of sensitivities used is usually within the range 4 mm per 50 μV to 15 mm per 50 μV.
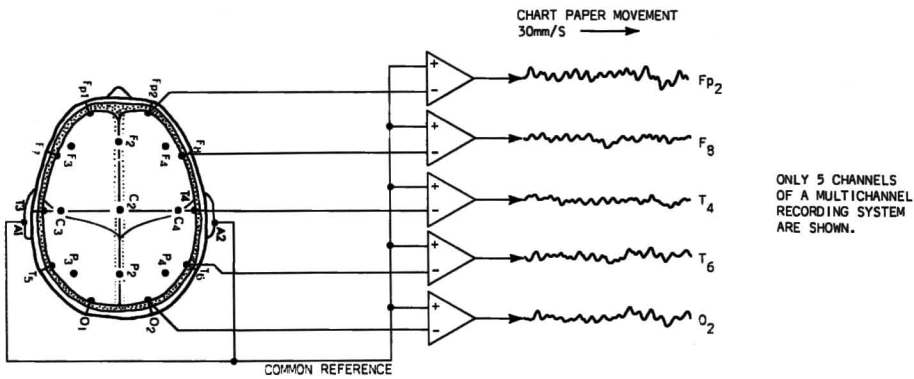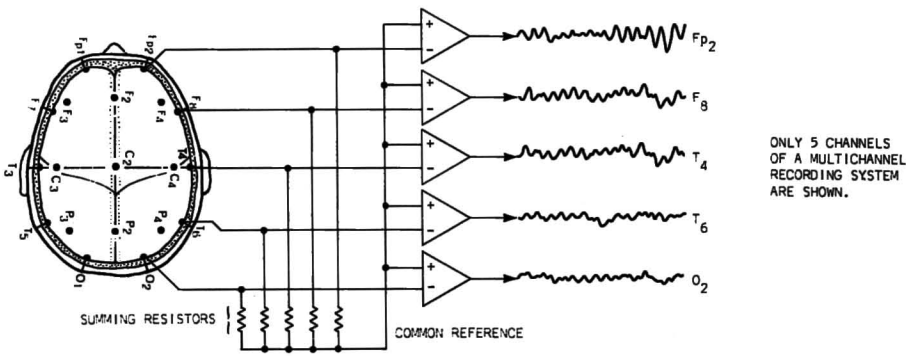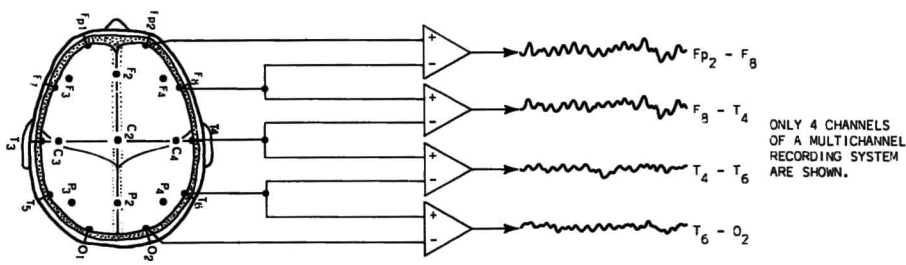
Fig. 10-3. EEG recording modes.

Most EEG instruments provide auxiliary outputs from all amplifiers to be used with other equipment, such as oscilloscopes, tape recorders or display devices.
paper speed A paper speed of 30 mm per second (25 mm per second in some instances) is often used for electroencephalographic recording. Many EEG instruments may also add time markers to the EEG recording on a separate channel and may also incorporate electrode contact resistance measurement facilities.

10.4 EEG RECORDING MODES

Three modes of recording are used in the routine EEG laboratory as shown in Fig. 10-3. They are known as unipolar (often improperly called monopolar), averaging reference and bipolar recordings.

unipolar mode In the unipolar mode one electrode is common to all channels (Fig. 10-3A). Ideally, this common electrode is regarded as electrically inactive, however in practice, electrical activity near this electrode will appear in all channels and invariably there are problems in selecting the site for this common electrode. The ear, or both ears connected together, are sometimes used as being generally close to regions of the brain with little on-going electrical activity. If a subject has a localized discharge, for simplicity we will assume a spike discharge, then successful localization of the spike will be dependent on its amplitude in the various channels. With some loss of scientific vigor we may say that the amplitude will be greatest in the channel with its active electrode nearest the source of the spike. If the common electrode is near the spike focus, localization is either not possible or very ambiguous. Although one electrode is common to all channels, to reduce interference and artifacts it is desirable to not ground this common electrode and a separate ground electrode is often connected between the subject and the instrumentation ground.

average electrode mode	In the average electrode system one input lead of all amplifiers is taken to the common point of a summing network in which equal (high) resistors are taken to each electrode (Fig. 10-3B). The recording will now indicate deviations from the mean instantaneous potential of the electrode system and thus an isolated feature (e.g., the spike) will, if it is sharply localized, stand out in one, or at worst, a small number of channels. This recording mode can be loosely compared with the recording configuration used for unipolar ECG's as described in Chapter 5, Sections 5.4 and 5.7. A resistive summing network is used to create a common point in each case.
bipolar mode	In the bipolar mode, the channels are connected in series between electrode pairs (Fig. 10-3C). By noting the change in the recorded EEG between these electrode pairs, very sharp localization of discharges is possible. The electrode immediately over the spike generator will cause a positive deflection in one recording channel and a negative deflection in the adjacent recording channel so that the electroencephalographer will see an apparent 180° phase difference between them. This "phase reversal focus" is accepted as the most reliable means of localization of discrete phenomena.

10.5 UNUSUAL EEG DISPLAY MODES

The voltage/time graph was used originally to display the EEG because oscillographs were available to the electrocardiographer. There is no certainty that these are the optimum ordinates to use in studying the EEG and a number of other techniques have been proposed. Some of these give a form of map-like presentation and allow the potential gradient over the head to be studied either in a "snapshot" fashion as used by Remond or as a synoptic display over a short time interval (Shipton). Such devices,

generally speaking, are good indicators of change in
the EEG activity but poor for quantification,
especially in terms of amplitude. It can be
plausibly argued that amplitude is not the parameter
of greatest interest and that what we are really
concerned with is the way that the brain handles
sensory data in terms of their temporal or spatial
succession. The use of more sophisticated methods
of display is increasing because the availability
of on-line, real-time digital computers which have
permitted data transformation in a number of
interesting ways.

spectrum
analysis

A spectrum analysis system has been occasionally
used in research applications to present the EEG
using amplitude/frequency coordinates. Most
conventional sampling-type instrumentation spectrum
analyzers are unsuited for direct EEG analysis as
their low-frequency performance characteristics are
inadequate and they do not lend themselves to the
analysis of continuously changing data such as an
EEG. The most desirable form of spectrum analyzers
is a "real-time spectrum analyzer," however, such
instruments are inherently rather expensive.
Frequency analysis is rarely used as a clinical
procedure; it masks much useful information which a
human operator, using our superb pattern recognition
abilities, can see at a glance. Mathematically, the
difficulty of spectrum analysis of an EEG is that
it is not time-invariant for a period long compared
with the lowest frequencies present (refer e.g.
Broadman & Tukey *Measurement of Power Spectra*).
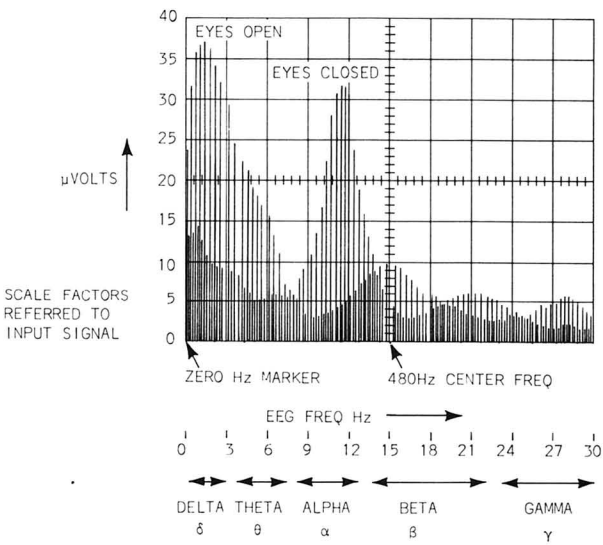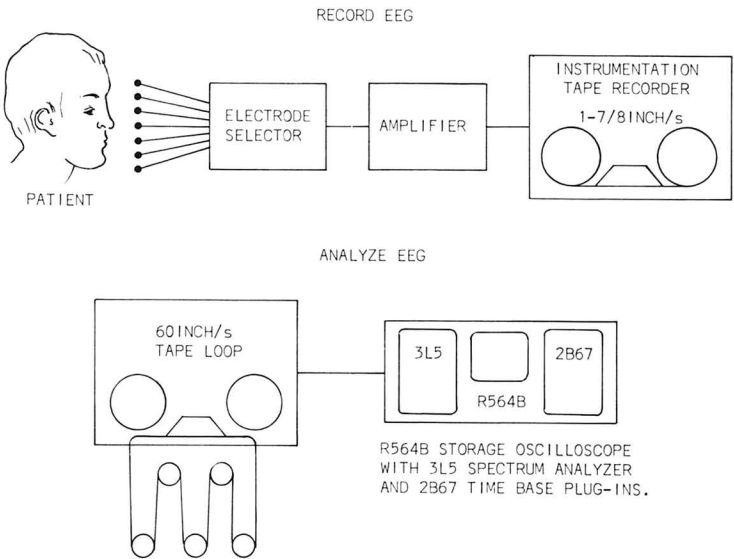
Fig. 10-4. EEG spectrum analysis.

tape
recorder

Spectrum analysis of the EEG has been performed with conventional instrumentation and sampling-type spectrum analyzers by utilizing an instrumentation tape recorder as indicated in Fig. 10-4. With this system a standard instrumentation tape recorder operating at 1 7/8" per second records the EEG and a small section of this recorded EEG is formed into a tape loop. This tape loop is then played back at 60" per second to effectively increase the frequency of the recorded information and to allow the use of a sampling-type spectrum analyzer, such as the Tektronix Type 3L5 Spectrum Analyzer in conjunction with a Tektronix Type 564B Storage Oscilloscope. The simulated CRT display shown in Fig. 10-5 represents two separate spectrum analyses of two separate tape loops, one recorded with a subject's eyes open and one recorded with a subject's eyes closed but with the subject awake. This display clearly shows the predominance of alpha activity with the eyes closed and the shift from alpha predominance with the eyes open.

10.6   INTRA-CRANIAL ELECTRODE PLACEMENT

In some diagnostic procedures the EEG recording electrodes are placed directly on the exposed surface of the brain. Under these conditions the output voltage will be considerably greater than the voltage obtained with normal EEG electrode placement, thus the gain of the recording instrument must be correspondingly reduced. Standard EEG electrodes cannot be used under these conditions as they are nonsterile and physically unsuited, thus special electrodes are used.

insulated
needle
electrodes

During neurosurgery insulated needle electrodes are often used to place an electrode deep within the subject's brain. These "deep electrodes" may consist of a needle insulated over its entire length with the exception of a small area at the tip or they may consist of concentric needles of varying length to effectively provide many electrodes at regular intervals along the length of the needle.

10.7 APPLICATIONS OF THE EEG

The EEG is primarily used in clinical neurology for partially assessing a subject's neurological state. Such applications are covered in more detail in the following section when discussing the abnormal EEG.

anesthetic
level

As well as its utility in the neurological clinic, there are other uses for the human EEG. Brain cells are, for example, affected by anesthetic agents and the EEG is a sensitive indicator of the depth of anesthesia. Some workers indeed have used the EEG signal in a closed-loop controller to keep a constant anesthetic level.

monitoring
during
surgery

In many surgical procedures involving the heart, the ECG waveform cannot be monitored, thus the EEG signal is used as an indication of subject well being. With these procedures the verification of death can no longer be related to the activity of the cardiac system, thus the presence of EEG activity is, in part, a useful indicator. EEG monitoring during surgery does not require the use of multichannel EEG instrumentation. The EEG signal can be monitored using a single channel recorder or, more commonly, by using an oscilloscope. Many surgical monitors, including the Tektronix Type 410 Physiological Monitor, designed primarily for cardiac monitoring, include EEG monitoring facilities. The 410 monitor simply requires two electrodes placed on the head over the occipital regions for EEG recording and one ground electrode placed anywhere on the subject.

alertness
monitor

The EEG is also a very subtle estimator of the differences between sleep and wakefulness. Much of our present knowledge of sleep phenomena - and sleep is much more complex than it seems at first sight - we owe to the EEG observation of sleeping subjects. A number of states can be distinguished. For those who must be alert at specified times during a long task, piloting a spacecraft is a case in point, the EEG can be made the basis of a reliable "state of alertness" monitor.

stimuli
responses

Recently there has been great interest in the slow (circa 0.1 Hz) potential changes in the EEG. Modern amplifier techniques and nonpolarizable surface electrodes have established that these shifts in the DC potential are associated with voluntary responses to stimuli. It is very probable that such studies will extend the use of EEG techniques into the realm of psychophysiology; already a number of tentative relationships between mental state and variation of the slow "expectancy waves" have been established.

10.8 THE CHARACTERISTICS OF THE ABNORMAL EEG

epilepsies

In describing the EEG as it is seen in various disease states, it is important to remember that very few single clinical tests are by themselves sufficient to make a diagnosis. Thus, the EEG is only one of many procedures used by the clinician in assessing the neurological state of a subject. The most common condition in which the EEG is valuable is epilepsy. Strictly speaking, we should refer to "the epilepsies" since many varieties are found. The EEG is of great help in forecasting the outcome of an epileptic illness and is valuable in establishing the optimum course of treatment.

brain
injury

For example, injury to a specific region of the brain can leave a permanent scar on the cerebral cortex. Such scar tissue is electrically inert but has an irritative effect on nearby healthy cortex. The EEG will often show a localized spike discharge and will suggest, among other possibilities, surgical removal of the damaged tissue.

inborn
epilepsy

When epilepsy is inborn, and some forms of this disease are hereditary, the abnormal electrical activity generally contains signals at many frequencies in the range 1 - 50 Hz. There is no consistent phase relationship between the various components so that the EEG presents, to the eye, a "noisy" appearance. The signals are generally a good deal larger than the α rhythm and usually cannot be localized to any specific region of the brain.

"petit mal"   In the so-called "petit mal" epilepsy, in which the
               manifestation of the illness is often a transient
               loss of consciousness or some automatic motor
               behavior, the signals are wideband but have
               remarkably consistent phase relationship between each
               component. The signal is thus seen as a regular
               pattern in which a sharp spike appears superimposed
               on a smooth low frequency (1 - 3 Hz) wave. Although
               many hypotheses have been advanced to account for the
               remarkable phase consistency seen in this "spike and
               wave" phenomenon, none are entirely convincing and
               all are outside the scope of this chapter.

tumors   When the brain is invaded by some forms of tumor, a
               considerable portion of the active nervous tissue
               may be displaced by the electrically inert new growth.
               If this is very large, its presence can be inferred
               from the absence of organized electrical activity
               from the region of the tumor. Usually, however, a
               tumor large enough to be associated with a
               detectable area of "electrical silence" is large
               enough to manifest itself in other ways so that this
               technique of tumor detection is of limited practical
               value. There are, however, other ways in which the
               EEG can help in the diagnosis of brain tumors at a
               much earlier state of their development. The
               expanding new growth can interfere with the blood
               supply to neighboring areas and the consequent
               malfunctioning of nerve cells around it manifests
               by a large, slow discharge - the δ rhythm. There
               may also be significant differences in the electrical
               activity in the affected hemisphere, perhaps as a
               result of interference with the internal
               communicating pathways within the brain. The extent
               to which the EEG is of value in tumor localization
               depends on many factors. Some of the more important
               are the rate at which the tumor is growing, its
               special relationship to the recording electrodes and
               the skill of the electroencephalographer.

               The conditions noted above are those in which the
               EEG is most frequently used. In other circumstances
               such as certain toxic conditions and some
               psychological states, the EEG can add to the overall
               amount of clinical information and be of significant
               benefit to the subject and physician.

10.9 INTENTIONAL MODIFICATION OF A SUBJECT'S EEG

Up to this point we have assumed that the
electroencephalographer plays an entirely passive
role and is content to study the brain in its normal
physiologic milleau. In practice a number of
techniques are used to increase the yield of
meaningful information; some of these apply external
stimuli to the subject and note their effect on the
EEG. Because the excitability of various parts of
the nervous system is critically dependent on the
acid-base balance (and thus on the oxygenation of the
blood) it is general practice to modify this balance
by asking the subject to hyperventilate, that is to
say breathe rapidly and deeply while at rest. In
the normal subject this procedure produces a nominal
slowing of the α rhythm and some small increase in
the overall signal level. If, however, the subject
is an epileptic, the record is dramatically changed
to the extent that a seizure may be provoked. Some
epileptics are markedly affected by lowering their
blood sugar and for this reason many clinical records
are obtained from fasting subjects.

Another important means of modifying the EEG is the
use of rhythmic sensory stimulation. One or more of
the senses is stimulated by brief repetitive stimuli;
light flashes are the most commonly used partly
because they are easy to generate and partly because
the visual cortex is large and the source of the
α rhythm. Sensory stimulation of this kind can
emphasize latent abnormalities in the resting EEG
and help in the interpretation of the tracing.

external
stimuli

148
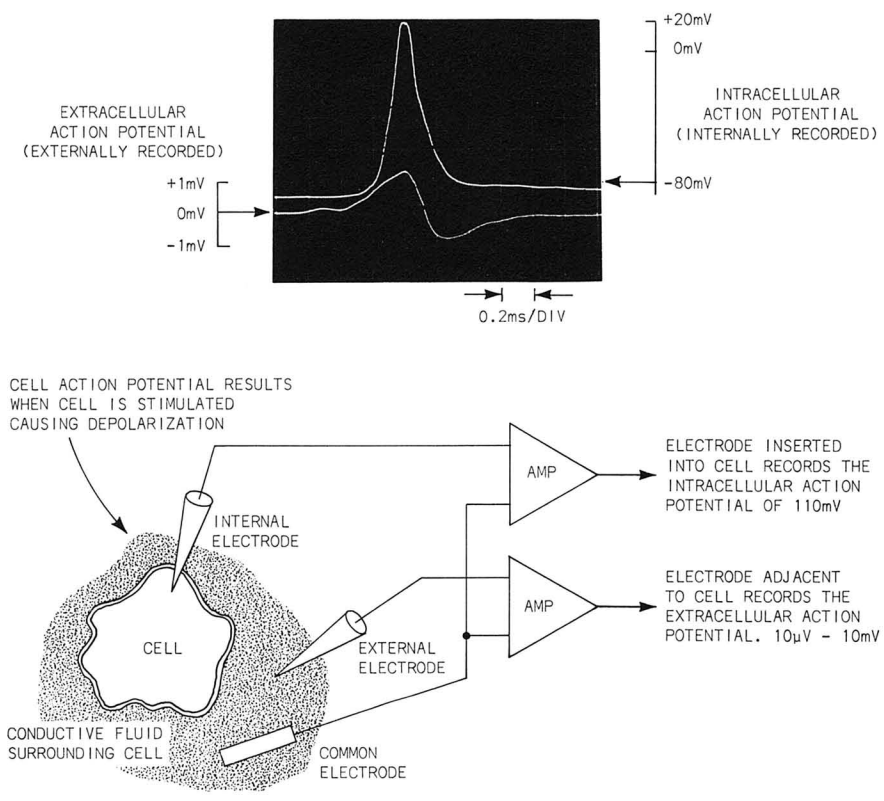
Fig. 11-1. Cell action potential — internally and externally recorded.

11

EVOKED CORTICAL RESPONSES

The electroencephalogram referred to in Chapter 10
is a measure of the over-all electrical activity
of the brain with the subject essentially at rest.
The electroencephalogram is probably associated with
the computation process continuously active within
the brain. Evoked potentials are also potentials
generated within the brain, however these potentials
result from a stimulus being applied to the body's
sensory system and are localized to a particular
area of the brain. These potentials are said to be
"evoked" by the stimulus.

11.1 EVOKED ACTION POTENTIAL

As stated above, stimulation of a subject's sensory
system produces electrical activity in a localized
area of the brain. When attempting to analyze the
effect of the stimulus, it is necessary to record
the electrical potentials generated within *individual*
intracellular brain cells by using intracellular recording
recording techniques. These techniques are only used in
research applications on nonhuman subjects. External
electrodes will record the electrical activity of a
cell, however, as many adjacent cells may also be
producing electrical activity, the results obtained
could not be attributed to any *particular* cell.
Under certain conditions, particularly when recording
from the spinal cord rather than from the brain, it
may be possible to isolate individual cell activity
extracellular using external electrodes to record the extracellular
recording action potential. Fig. 11-1 shows the action
potential generated by a *single* cell when recorded
with both internal and external electrodes and shows
the time and voltage relationship between these two
recording techniques.

150
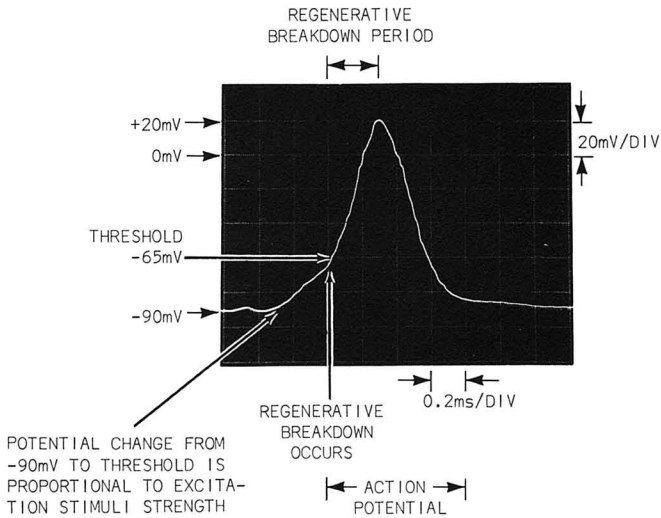
Fig. 11-2. Action potential showing threshold.

intracellular
action
potential

When recording the intracellular potential with an internal electrode, about 110 millivolts of signal is generated during a cell depolarization/ repolarization process and this signal is known as the intracellular action potential. It is, however, possible to record lower amplitude signals generated within the cell as a result of excitation and inhibition stimuli as discussed in Chapter 4. The intracellular action potential shown in Fig. 11-2 is clearly preceded by a period where the cell is receiving excitatory stimuli which decreases the cell's resting potential at a linear rate until the cell threshold is reached, at which time the rate of change of potential increases, indicating regenerative depolarization. The cell subsequently repolarizes.

11.2 MICROELECTRODE TECHNIQUE

The preceding discussion covers intracellular recording and shows typical intracellular action potentials. In practice, intracellular recording requires highly specialized measurement techniques, and results similar to the action potentials shown in Fig. 11-1 and 11-2 are difficult to achieve.

When recording the intracellular action potential it is, of course, necessary to insert an electrode into the cell concerned. If the results obtained are to serve any practical purpose, it is also necessary that this electrode have a negligible effect on the characteristics of the cell concerned. It is thus desirable to use an electrode with dimensions much smaller than the dimensions of the cell concerned; requiring electrodes with tip diameters well under one micron (10^{-4} centimeters). These small electrodes are known as microelectrodes. Both glass microelectrodes, often referred to as micropipettes, and metal microelectrodes are available.

micro-
electrodes

The basic difference between these is that in the metal microelectrode the metal is in direct contact with the biological tissue whereas in the glass microelectrode an electrolyte is interspersed between the tissue and the metal electrode. Thus, metal microelectrodes have a lower resistance, but they polarize with smaller amplifier input currents, their resistance may increase, and they may develop unstable electrode offset potentials. Unless extreme precautions are taken, they are, therefore, unreliable for steady-state potential measurements.

metal

The glass microelectrode interposes an electrolyte between the tissue and the metal electrode which results in improved stability as the metal and the electrolyte can be chosen so that small, steady currents can pass through their junction without modifying the electrical properties of the electrode. The surface contact area between the electrolyte and the metal is large so that the current-carrying capacity of the electrode is substantial. The glass microelectrode is, therefore, usually preferred.

glass

Fig. 11-3 shows typical tip dimensions of a glass microelectrode, having an over-all tip diameter of about 0.6 microns with an internal diameter of 0.2 microns. The glass microelectrode is formed by heating special glass tubing and drawing it out over several stages of reduction. Although the tubing is reduced to less than a micron in diameter, it still remains hollow. Potassium chloride solution is introduced into the microelectrode as the electrolyte.

As it is not possible to fill the microelectrode by pressure (surface tension) or by capillary action, boiling with or without reduced pressure is often used. Ideally, one would like to use an electrolyte within the microelectrode having the same concentration as the potassium chloride within a typical cell (0.1 Normal). It is, however, impossible to use an electrolyte of this concentration

as the resistance of the microelectrode would be well over 1,000 megohms. It is, thus, common practice to use 2 or 3 Normal potassium chloride in the microelectrode. This electrolyte has a resistivity of 3.3 ohm centimeter which gives the microelectrode similar to that shown in Fig. 11-3 a typical resistance of 10 megohms. Special purpose microelectrodes having smaller tip diameters and/or using a less concentrated electrolyte may have a typical resistance of 100 megohms. As will be shown later in this chapter, special recording techniques are necessary to accommodate this extremely high electrode series resistance.

Fig. 11-4 shows a typical microelectrode inserted into a cell and the equivalent electrical circuit formed between this microelectrode and a common electrode elsewhere on the subject. This equivalent circuit can be further simplified to the microelectrode resistance with distributed capacity, and an RC load on the microelectrode formed by interconnection capacity, amplifier input capacity and amplifier input resistance. Since the dimensions of the microelectrode tip are extremely small, and as most of the microelectrode resistance is located within one millimeter of the microelectrode tip, this distributed capacities associated with the electrode resistance are less than one picofarad.

MICROELECTRODE OVERALL DIMENSIONS:

6 5 4 3 2 1 0 cm

SILVER WIRE

GLASS TUBING

KCl ELECTROLYTE - 3N CONCENTRATION
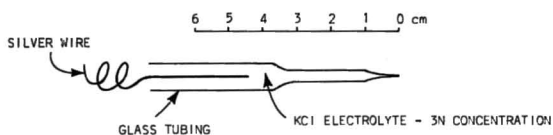
MICROELECTRODE TIP DIMENSIONS:

ONE MICRON = 10^{-4} cm

MICROELECTRODE FILLED WITH
3N POTASSIUM CHLORIDE (KCl)
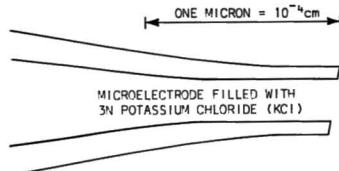
Fig. 11-3. Glass microelectrode geometry.

MICROELECTRODE

AMP

CELL

COMMON
ELECTRODE

APPROX.
EQUIVALENT TO

TO AMP

TIP OF
MICROELECTRIC
INSIDE CELL

MICROELECTRODE
ELECTROLYTE

C_{mf}

C_{mc}

R_m

R_c

FLUID INSIDE
CELL

E

C_m

R_m

CELL MEMBRANE

R_f

CONDUCTIVE FLUID
SURROUNDING CELL

COMMON ELECTRODE

APPROX.
EQUIVALENT TO

R_c, R_m, R_f AND
C_m ARE NEGLIGIBLE

R_m

C_{mc}

C_{mf}

R C

E

AMPLIFIER WITH R AND C INPUT LOAD.
(C INCLUDES INTERCONNECTION CAPACITY)

R_m = TIP RESISTANCE OF MICROELECTRODE

C_{mf} = CAPACITANCE - MICROELECTRODE TIP TO FLUID
SURROUNDING CELL

C_{mc} = CAPACITANCE - MICROELECTRODE TIP TO FLUID
INSIDE CELL

R_c = "RESISTIVITY" OF FLUID INSIDE CELL

E = BIOELECTRIC ACTION POTENTIAL GENERATOR

R_m = MEMBRANE RESISTANCE

C_m = MEMBRANE CAPACITANCE

R_f = "RESISTIVITY" OF FLUID SURROUNDING CELL

Fig. 11-4. Cell-microelectrode equivalent circuit.

154

EQUIVALENT CIRCUIT FROM FIG. 11-4.

TYPICAL VALUES:

C_{mc} = 0.1pF

C_{mf} = 0.2pF

R = 1000MΩ

C = 1pF

R_m = 10MΩ

E = −90mV TO +20mV

t_r = 0.1ms t_f = 0.2ms

DURATION − 0.5ms

INPUT PULSE E

OUTPUT PULSE FOR VARIOUS VALUES OF R_m

10MΩ

30MΩ

100MΩ 300MΩ

0.2ms/DIV

R_m COMPRISES THREE RESISTORS − EACH 1/3R_m

PULSE GENERATOR

0.1pF

0.2pF   1000MΩ   1pF

MICROELECTRODE EQUIVALENT CIRCUIT MODELED WITH IMPEDANCES SCALED BY 10^3.
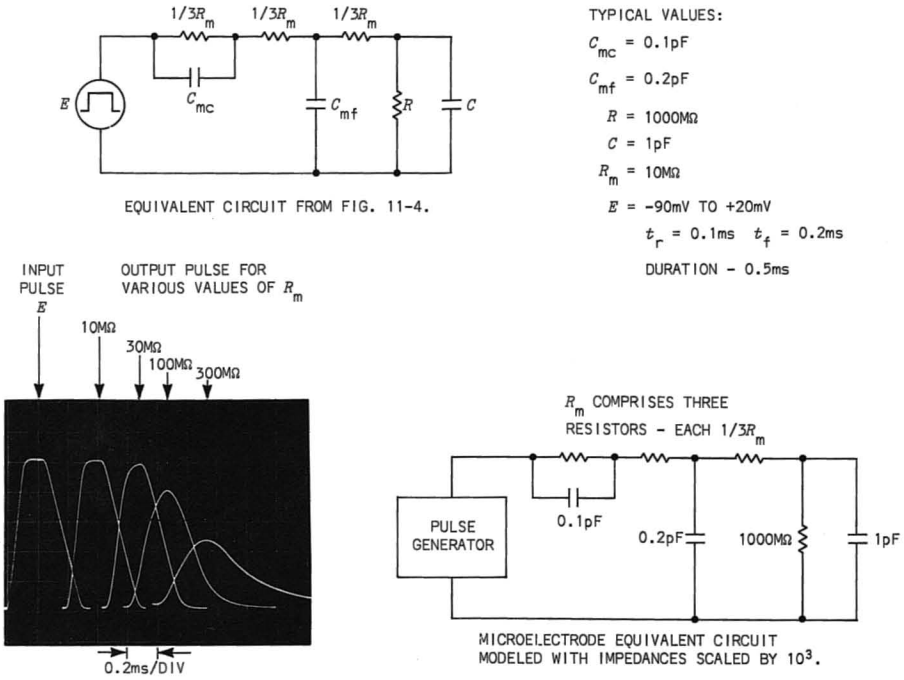
Fig. 11-5. Microelectrode model and resulting pulse response.

Referring to Fig. 11-5, a typical microelectrode equivalent circuit is shown together with typical values for the impedances concerned. This equivalent circuit represents a low-pass filter and attenuator and it is necessary to know the characteristics of this circuit to determine the fidelity expected from the system when recording the action potential. This equivalent circuit can be modeled using a pulse generator in place of the cell and using discrete impedances in place of the microelectrode resistance and capacity. Fig. 11-5 shows the degradation that can be expected from an input "action potential" having a risetime of 0.1 millisecond and a falltime of 0.2 millisecond for various values of electrode internal resistance when using an amplifier having an equivalent input capacity of one picofarad and an equivalent input resistance of 1,000 megohms.

circuit
determines
fidelity

As can be seen from Fig. 11-5, a 100 megohm microelectrode used in the above system reduces the amplitude of the action potential by 20 percent and degrades its risetime and falltime. It is thus desirable to either reduce the resistance of the microelectrode to 30 megohms or perhaps even 10 megohms to increase fidelity or, alternatively, to increase the input impedance of the amplifier and interconnection network. Since decreasing the microelectrode resistance usually means increasing the size of the microelectrode, the latter alternative is preferable.

156

564B STORAGE SCOPE

CAL

VERT
3A8
OPERATIONAL
AMPLIFIER

HORIZ
2B67
TIME BASE

LO PASS
FILTER

INPUT
NEUTRALIZED
ADAPTER

PULSE
SHAPING
CIRCUIT

MICRO-
ELECTRODE

COMMON
ELECTRODE

WITH CALIBRATOR "OFF", COMMON
ELECTRODE IS GROUND.

WITH CALIBRATOR AT "40V", PULSE
SHAPING CIRCUIT ADDS A 50mV PULSE
IN SERIES WITH THE MICROELECTRODE.

R_m

R_o
18Ω

C1
<1pF

R1
>10⁹Ω

0.2ms/DIV

50mV
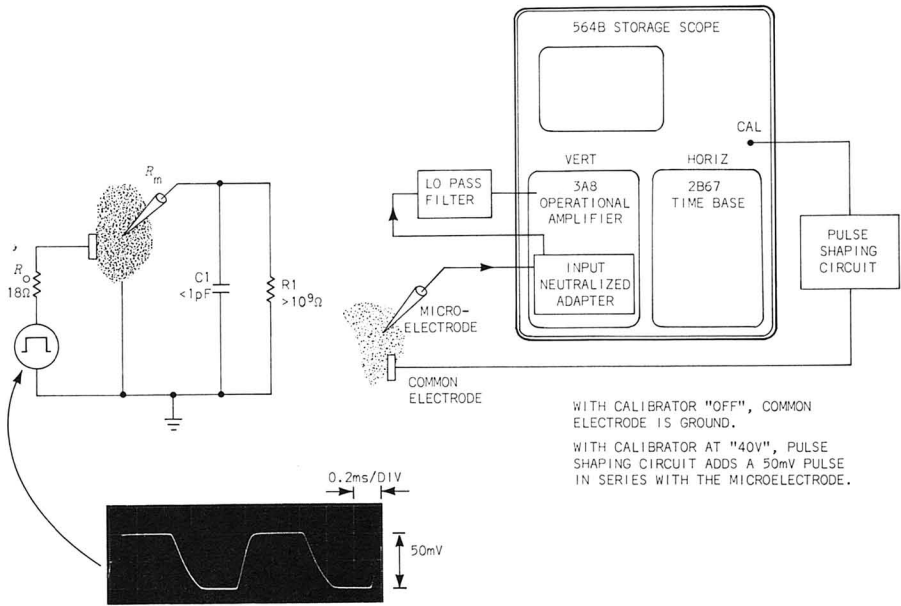
OUTPUT WAVEFORM FROM PULSE SHAPING
CIRCUIT USED FOR SYSTEM CALIBRATION.

PERFORMANCE

WITH 10MΩ
MICROELECTRODE

WITH 100MΩ
MICROELECTRODE

CALIBRATOR "ON", MICROELECTRODE
REPLACED WITH WIRE (0Ω)

MICROELECTRODE IN PLACE AND
NEUTRALIZATION CONTROLS ADJUSTED
FOR OPTIMUM RESPONSE

AS ABOVE WITH BANDWIDTH LIMITING
ADJUSTED FOR "LO" NOISE WITH
MINIMUM DISTORTION

EFFECT OF AMPLIFIER ADJUSTMENTS

WITH 100MΩ
MICROELECTRODE

INPUT RESISTANCE TOO LOW
APPROX. 30MΩ

INPUT CAPACITY TOO HIGH
APPROX. 5pF

INPUT CAPACITY OVER-NEUTRA-
LIZED, PRODUCING RINGING
AND EVENTUALLY OSCILLATION

CORRECT ADJUSTMENT

FOR CIRCUIT DETAILS
REFER TO CHAPTER 29

Fig. 11-6. An input neutralized amplifier and scope system.

11.3 INPUT NEUTRALIZED AMPLIFICATION

feedback
increases
input Z

The impedance load on the microelectrode can be increased by using active feedback amplification to provide input impedance neutralization. Careful adjustment of such amplifiers can achieve an input resistance in excess of 10,000 megohms with equivalent input capacities of the order of 0.1 picofarad. Fig. 11-6 shows a typical microelectrode system utilizing a Tektronix Type 3A8 Operational Amplifier plug-in unit operating in an "input neutralized" mode. This requires the use of an input neutralizing adapter in conjunction with the 3A8. Details of this adapter are given in Chapter 29.

system
calibration

This microelectrode recording system also incorporates a series calibration waveform. By switching the 564B calibrator from OFF to 40 V, a 50 millivolt, 1000 Hz, signal is added in series with the microelectrode. This signal has a controlled risetime of 0.1 ms and a controlled falltime of 0.2 millisecond and is used in conjunction with the two input neutralization controls (-R and -C) to check the amplitude and time calibration of the system. When the calibrator is returned to its OFF position, this signal source is essentially replaced by a short circuit. This calibration waveform and the response of the system for various settings of the input neutralization controls are shown in Fig. 11-6. In practice, the input neutralization should be adjusted to the point at which overshoot and ringing are reduced to less than 10 percent. Further reduction of the overshoot and ringing will degrade the risetime of the system.

amplifier
input
current

Although the dynamic input resistance of a neutralized amplifier may be above 10^{10} ohms, the system may have excessive amplifier input current and will thus not operate. Since the dynamic input resistance of any device is determined by the change in input current for an incremental change in input voltage; $R = \frac{\Delta V}{\Delta I}$; and as amplifier input current does not change with an incremental change in input voltage, this amplifier input current is not reflected by the input resistance specification of the amplifier and must be specified separately. Any neutralized amplifier should have an amplifier input current below 10^{-12} amperes.

The system shown in Fig. 11-6 has an amplifier input current of approximately 10^{-13} amperes. These extremely small currents are necessary when using the amplifier for intracellular recording as the current will cause potassium ions to migrate from the microelectrode into the cell. This upsets the intracellular concentration, which may change the cell resting potential. Small changes in resting potential due to extremely small currents cannot be avoided, and may not be significant. However, it is obviously desirable to maintain the current below the level required to cause cell depolarization.

tangential
noise

The noise in any neutralized amplifier is proportional to the degree of neutralization required; a tangential noise of several millivolts is not uncommon. "Tangential" noise is discussed in Chapter 19, Section 19.7. The system shown in Fig. 11-6, when correctly neutralized, produces a tangential noise of 1 millivolt when used with an electrode having a resistance of 10 megohms and 2.5 millivolts when used with an electrode having a 100 megohms resistance. This noise is insignificant when recording intracellular potentials as these potentials are in the order of 100 millivolts. For extracellular measurements however, the extracellular action potential may be as low as 10 microvolts and the amplifier noise will completely swamp any extracellular signals. This situation also arises when using intracellular recording at high sensitivity in an attempt to observe excitation and inhibition potentials. The signal-to-noise ratio

lowpass
filter

can be substantially improved by a lowpass filter following the neutralized amplifier. Such a filter is included in the system shown in Fig. 11-6 and further details of this filter are given in Chapter 29. Even using a filter for extracellular measurements, the noise level obtained may still completely swamp the signal and, at best, will be extremely objectionable. Thus, particularly when working at sensitivities below 100 microvolts per division, some more effective noise reduction technique is almost always necessary.

11.4 NOISE REDUCTION - AVERAGING

signal
averaging

A single response evoked by one stimulus is usually too small to be seen above the on-going EEG activity and the noise inherent in the system. However, if we assume that the response always follows the stimulus after a fixed delay, "signal averaging" computers can improve the signal-to-noise ratio. Many such signal averaging computers have been marketed in the past few years and, while they differ in detail, they all use the same basic principle.

Signal averaging extracts a wanted signal from a background of unwanted noise. It can only be used effectively if the desired signal, with its accompanying noise, can be generated a number of times, either periodically or aperiodically. In addition, a trigger pulse is required that has a fixed time relationship to the desired signal. The stimulus, or trigger pulse, initiates a scanning device which samples the signal at fixed intervals. These time-sequential samples are stored in discrete storage channels or memory locations, each channel collecting data over a small time segment. When the stimulus is repeated, the responses are added to the values stored at each location. During readout, each of the channels displays the sum of all the previous samples fed into it. If the background activity is random with respect to the stimulus, the time-locked response will add linearly with the number of samples (n) while the background will add only as the square root of n. If the stimulus is repeated 64 times (a fairly common number), then the response to noise ratio will have been improved by a factor of $\frac{n}{\sqrt{n}} = \sqrt{n} = 8$. Filtering to the maximum permissible extent should always precede the averaging device.

160
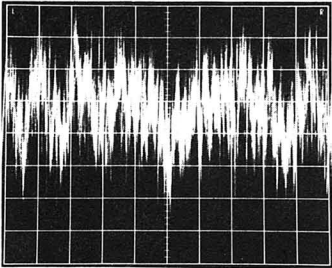
RAW SIGNAL OUTPUT FROM
NEUTRALIZED AMPLIFIER
VERTICAL 100µV/DIV (REFERRED TO INPUT)
HORIZONTAL 0.2ms/DIV

SIGNAL AVERAGER AND READOUT OSCILLOSCOPE

CONVENTIONAL SIGNAL PROCESSING

SIGNAL AVERAGING

NUMBER OF
SWEEPS
OF AVERAGER

AMPLIFICATION WITH BANDWIDTH
LIMIT (10kHz) AND GAIN REDUCTION
(200µV/DIV REFERRED TO INPUT)

1 SWEEP
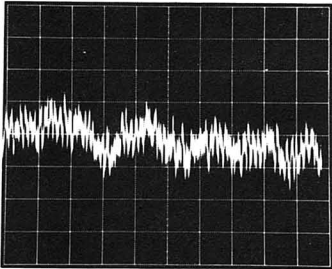
4 SWEEPS

64 SWEEPS

ORIGINAL SIGNAL WITHOUT NOISE.
IDEAL SIGNAL PROCESSING SHOULD
PRODUCE THIS WAVEFORM.
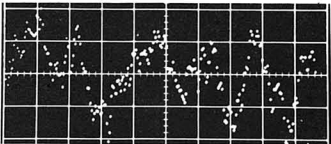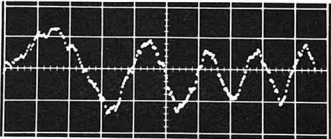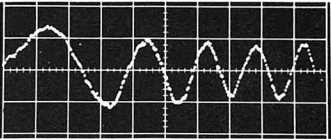
256 SWEEPS

100µV/DIV   0.2ms/DIV

100µV/DIV   0.2ms/DIV

Fig. 11-7. Signal to noise ratio improvement with averaging.

signal-to-
noise ratio
improvement

Typical signal-to-noise ratio improvement via
averaging is shown in Fig. 11-7. The noise has a
peak value about equal to that of the signal,
equivalent to an RMS S/N ratio of about 4. Averaging
is an ideal technique under these conditions. For
greater noise levels the signal content is reduced,
because the peak noise plus signal must still not
overdrive the averager; difficulty then arises
because the signal component alone is not being
digitized to a sufficient number of bits to give
a usable output.

triggering
on signal

In many circumstances, averaging can be carried out
in the absence of the trigger pulse if it is
possible to use part of the response itself as the
stable point. This is an aspect of averaging which
is currently of great interest to physiologists.

averager
character-
istics

The resolution of the signal averager is limited by
the number of addressable channels; commercial units
may contain from 50 to 1,000 channels. The "speed"
of the system is limited by the length of time
required to store information into any one channel.
Commercial averagers are able to sweep through all
channels in less than 5 milliseconds, providing an
effective sweep speed of 0.5 milliseconds per
centimeter for the readout oscilloscope. Since the
sweep speed is limited by the individual channel
access time, the use of fewer channels can increase
the effective sweep speed at the expense of
resolution. The channel access time for most modern
averagers is about 10 microseconds; however, some
earlier averagers designed specifically for
biological use have an access time in excess of 100
microseconds.

Most signal averagers contain the necessary circuits
for averaging but require a separate X-Y
oscilloscope to display the output information.
Normally, calibration facilities are incorporated
into averagers to allow the monitor to be
calibrated for both voltage and time reference.
This calibration is particularly important as the
"level" of each channel in the averager varies with
the number of sweeps used and all absolute amplitude
information is, therefore, meaningless unless a
calibration system is provided. Some time averagers
are now available in which the *average* rather than
the *sum* of the sweeps is displayed.

The signal averager relies on a trigger pulse being available having a known time relationship to the desired signal. When recording evoked potentials, it will be recalled that these potentials are generated in response to a stimulus being applied to the patient, this stimulus can be used as the trigger for the averager.

using stimulus as trigger

The averaging technique described above can introduce considerable error if time inconsistencies (jitter) exist between the trigger pulse and the evoked response. The response shown in Fig. 11-8 was produced after averaging over 256 sweeps, each sweep being initiated by light flashes at an approximate 2 second rate; thus 512 seconds or about 8.5 minutes are required to produce the response. During this long time interval, both the shape of the evoked response and the timing relationship between this evoked response and the light stimulus may vary and the simple averager will produce a response of a longer duration and shorter amplitude than will be the case if these time inconsistencies did not exist.

time jitter

More sophisticated signal-averaging techniques, using digital computers, although inherently more expensive than the techniques described above, are somewhat more versatile as they also incorporate facilities to analyze statistically the results obtained. The Digital Equipment Corporation Lab 8 signal averager is typical of a computer-type signal averager.

digital computer averagers

Many other computational aids are in use to improve the quality of signal recovery. For example, the presence or absence of a rhythmic component in a signal can be determined by cross-correlating the signal from one amplifier channel with that in another. This technique is a sensitive indicator that a signal is present but does not give information about its amplitude nor necessarily about its phase or time relationship.

correlating two inputs to detect signal

Although the signal averager is primarily intended to detect a wanted signal in the presence of unwanted noise, it can also be used to detect a wanted signal in the presence of other unwanted signals. When recording the EEG, for example, it is possible to detect small changes in the EEG when flashing a light at the subject.

averaging evoked responses from unwanted signals

Using conventional amplification techniques, it is possible to detect the response of the brain to this light flash as this response is camouflaged by EEG activity produced within other parts of the brain. The signal averager can be used to detect this response by regarding the EEG as an unwanted signal which bears no time relationship to the stimulus. Fig. 11-8 shows a typical EEG recorded with a light being flashed at the subject at approximately two-second intervals. The EEG shown appears normal and it is almost impossible to detect the effect of the light flash. The same EEG signal, when analyzed by an averaging computer triggered from the light flash, clearly shows the electrical response of the brain to the light flash as seen in Fig. 11-8.

EEG RECORDED WITH SCALP ELECTRODES DURING REPEATED LIGHT FLASHES OF 1ms DURATION AT APPROX. TWO SECOND INTERVALS.

RESPONSES TO THE LIGHT FLASHES ARE BURIED BY THE EEG AND BY NOISE

$50\mu V/DIV$

$0.5s/DIV$

LIGHT FLASH   LIGHT FLASH   LIGHT FLASH
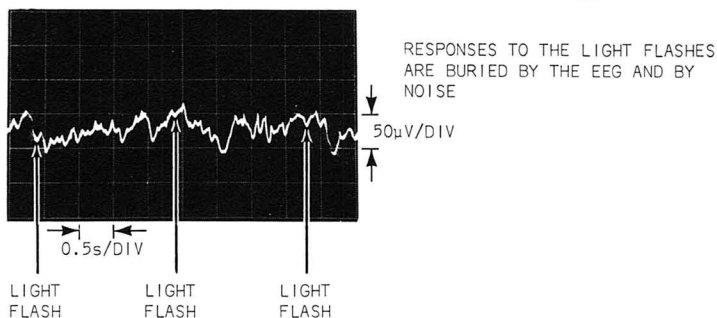
THE ABOVE WAVEFORM AFTER ANALYSIS BY A SIGNAL AVERAGING INSTRUMENT TRIGGERED FROM THE LIGHT FLASH.

25 MICROVOLT RESPONSE CAN CLEARLY BE DETECTED AFTER AVERAGING WITH 256 SWEEPS

$5\mu V/DIV$

$0.1s/DIV$
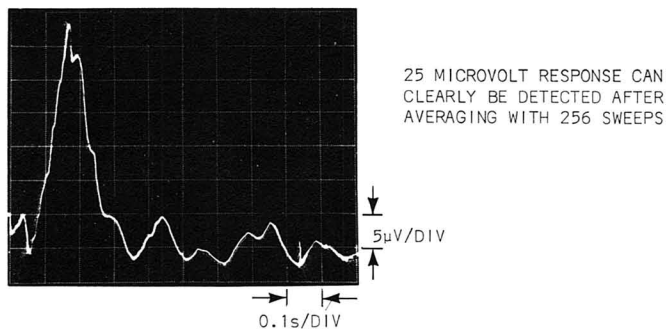
Fig. 11-8. Use of averaging to detect a response in the EEG.
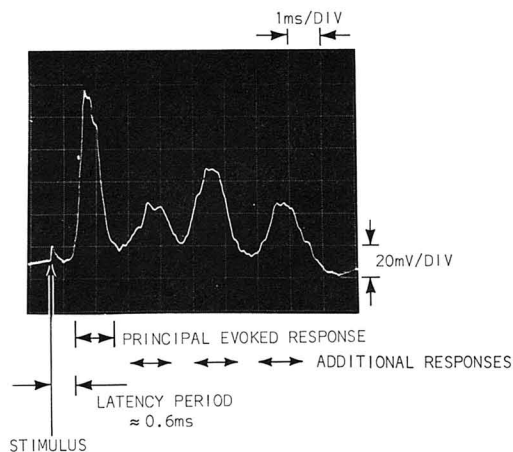
Fig. 11-9. Evoked response showing latency and additional responses.

11.5 TYPICAL EVOKED RESPONSES

stimulus/
response
delay

A typical evoked response obtained with a
microelectrode on a laboratory animal is shown in
Fig. 11-9. This response waveform is preceded by a
small amount of stimulus artifact. Although it is
desirable to reduce the level of stimulus artifact
appearing in recordings, it is often undesirable to
completely suppress it as it does provide some time
relationship in the recorded evoked potential. The
evoked potential begins sometime after the stimulus
is received; this delay is known as latency and is
approximately 0.6 milliseconds in the waveform shown
in Fig. 11-9.

Typical evoked potentials obtained with an
extracellular microelectrode with stimulation of the
visual cortex of a laboratory animal are shown in
Fig. 11-10. For the reasons covered above, stimulus
artifact has deliberately been added to these
recordings to provide a time reference. Fig. 11-10
shows the evoked response obtained for a single
light flash and for multiple light flashes separated
by various periods from one another. The 50
millisecond pair of flashes is seen by the subject
as one, whereas two flashes are perceived for the
150 and 250 millisecond intervals. Unlike the
evoked response shown in Fig. 11-9, these responses
were obtained with extracellular electrodes and show
the effect of many cells within the visual cortex;
thus the "time scale" of the response is greatly
increased as many cells are involved.

RESPONSE TO SINGLE AND DOUBLE FLASHES OF LIGHT. EACH
FLASH IS 10μs DURATION. (ARROWS INDICATE LIGHT FLASH.)
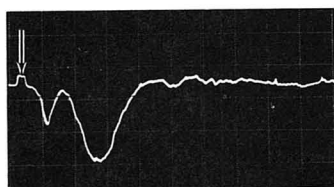
SINGLE FLASH

SINGLE RESPONSE OBTAINED
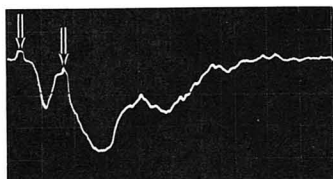
DOUBLE FLASH
50ms APART

SINGLE RESPONSE OBTAINED
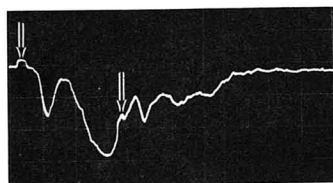
DOUBLE FLASH
150ms APART

DOUBLE RESPONSE OBTAINED
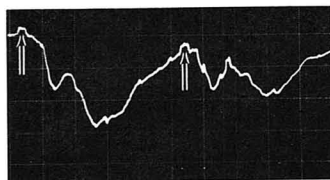
DOUBLE FLASH
250ms APART

DOUBLE RESPONSE OBTAINED

50ms/DIV   100μV/DIV

Fig. 11-10. Evoked potentials from the visual cortex.

11.6 STIMULATION

This chapter has previously discussed potentials produced as a result of stimulation, however the techniques used to provide this stimulation have not been covered. Stimulation is covered in detail in Chapter 12 when discussing the Electromyogram and Nerve Conduction. The principles applied to stimulating the peripheral nerves and muscles also apply to stimulating the cortex. It should be noted, however, that cortex stimulation may be carried out with the brain exposed and with stimulation electrodes spaced by only a few millimeters so as to provide extremely localized stimulation. Under these conditions, a greater percentage of the stimulation current passes through the cells of interest and, thus, the overall stimulus current required to produce a response is far lower than is the case when stimulating through skin and bone with electrodes placed many centimeters apart. Generally, when stimulating with the cranium opened, stimulators capable of outputs up to ten volts are adequate. Stimulus electrodes are normally needles or concentric needles -- the concentric needle provides stimulation between two "electrodes" less than one millimeter apart and also provides some degree of shielding to minimize the level of stimulus artifact appearing in the recorded response. Since electrical stimulation may create stimulus artifact potentials which camouflage the response, and since it appears to be desirable to stimulate via a sensory modality rather than to use direct stimulation, light stimulation or other sensory receptor stimulation is often used.

11.7 STEREOTAXIC INSTRUMENTS

The instrumentation system shown in Fig. 11-6 simply depicts a microelectrode inserted into a specimen. In practice, when recording evoked potentials, the skull is held firmly in a jig and the microelectrode is positioned with a micromanipulator or stereotaxic instrument, usually with the aid of a microscope or X-ray techniques, as shown in Fig. 11-11. Although stereotaxic instruments are available commercially, many are constructed from precision microscope stages.
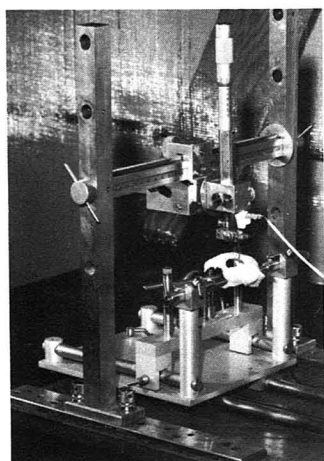
STEREOTAXIC INSTRUMENT

RIGID SUPPORT FOR ANIMAL'S SKULL

A TYPICAL STEREOTAXIC
INSTRUMENT AND SKULL
SUPPORT
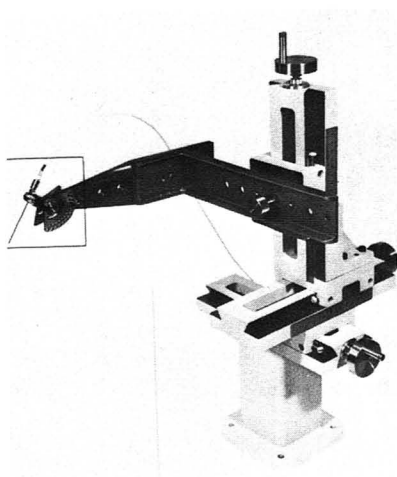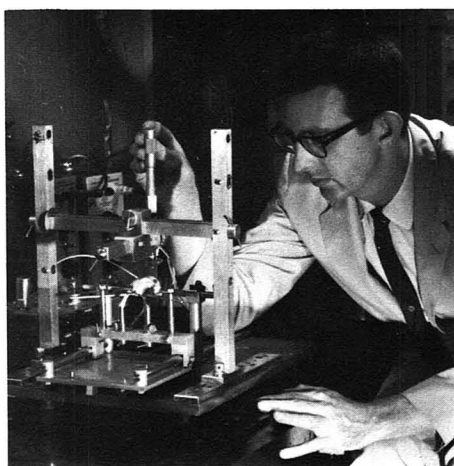
THE FRY* PROBE AND ELECTRODE
POSITIONING UNIT

VERNIER SCALES ON THE
X,Y AND Z AXES PERMITS
POSITIONING TO 0.05mm

*INTERSCIENCE RESEARCH INST.
CHAMPAIGN, ILLNOIS

Fig. 11-11. Sterotaxic instruments.

CONTACT LENS WITH
ELECTRODE EMBEDDED
BELOW VISION AREA

CHAMBER OF
CONTACT LENS
FILLED WITH
SALINE

Ag/AgCl ELECTRODE

AMPLIFIER AND OSCILLOSCOPE

COUPLING CAPACITORS
ELIMINATE DC CURRENT

COMMON ELECTRODE ON
SUBJECT'S FOREHEAD
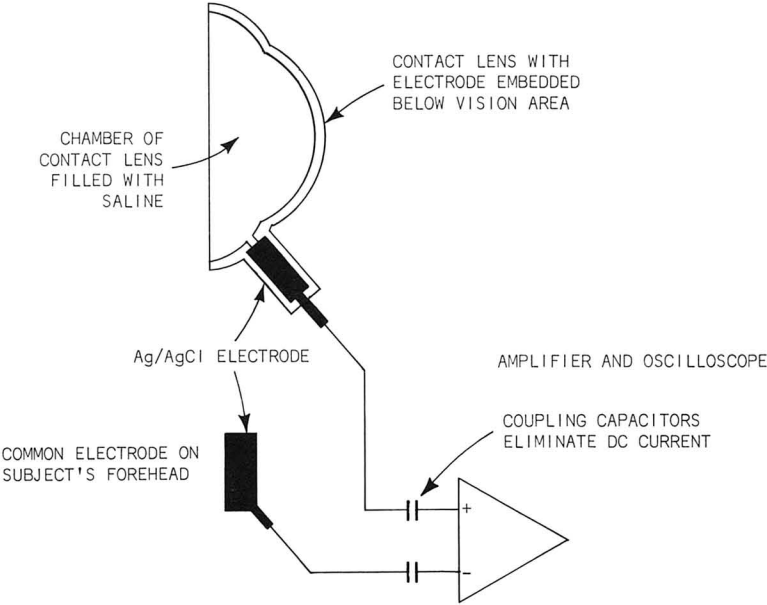
STROBE UNIT

Fig. 11-12. Contact lens ERG electrode.

11.8 THE ELECTRORETINOGRAM -- ERG

As stated in Chapter 3, it is almost impossible to detect the electrical activity produced by most of the sensory receptors on the body as they are too small and are dispersed. The notable exceptions to this are the potentials resulting from stimulation of the middle ear by sound waves and of the retina by light. If a bright light is projected into the eye, the retina will be stimulated and will generate action potentials known as the electroretinogram, or ERG, which can be detected by an electrode placed on the outside of the retina.

electrodes

stimulus

monitor

caution

The human electroretinogram is recorded with the aid of a silver/silver-chloride electrode imbedded in a contact lens, and a "common" silver/silver-chloride electrode placed on the subject's forehead, as shown in Fig. 11-12. Application of the contact lens electrode is painless and it is thus suited to clinical use. The subject is placed in front of a light source which may simply be an incandescent light coupled to a photographic shutter or, more commonly, a stroboscopic light as shown in Fig. 11-12. The General Radio Company Type 1504 P4 Stroboscope is ideal for this application. Stimuli are presented in the form of flashes of light and the responses, which are obtained by means of the contact lens electrode, are displayed on a medical monitor, such as the Tektronix Type 410 Physiological Monitor. The 410 is operated in the EEG mode with the "EEG electrodes" formed by the contact lens and common electrode. Careful technique is required in affixing ERG electrodes and recording the ERG. It is of paramount importance that the input current of the amplifier used to record the ERG be extremely low as series damage can occur to the eye if DC current is allowed to flow through an ERG electrode. Since stimulus artifact is not involved, AC-coupled amplifiers can be used and a capacitor may be used in series with the ERG electrode to eliminate DC currents.

A typical ERG is shown in Fig. 11-13 together with a diagram showing the general characteristics of the ERG. Variations in the characteristics of the stimulating light, as well as variations in the level of the state of light adaptation of the retina, will affect the response characteristics. The electroretinogram shown in Fig. 11-13 was recorded after the subject had become adapted to room lighting. The general characteristics of the electroretinogram consist of an initial A wave, followed by a positive B wave, a more slowly developing positive C wave (which is dependent upon the duration of the stimulation) and a D wave or off-effect.

components
of ERG
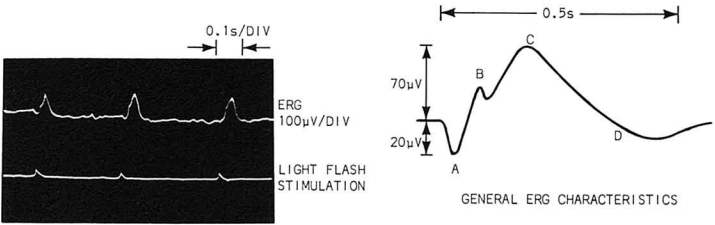
Fig. 11-13. A typical ERG.

12

STIMULATION – ELECTROMYOGRAPHY
– NERVE CONDUCTION

12.1 STIMULATION

Chapter 1 covers the bioelectric potentials generated within individual cells when these cells are stimulated. Stimulation refers to an external force being applied to the cell which results in the cell depolarizing and then repolarizing. Cells in the eye are sensitive to light stimulation, cells in the ear are sensitive to sound stimulation and nerve cells are sensitive to electro-chemical stimulation. All cells are sensitive to one degree or another to artificial electrical stimulation: The passage of electric current through the cell will cause it to depolarize.

artificial
stimulation

When analyzing biomedical phenomena it is often desirable to artificially stimulate a group of cells. Such artificial stimulation is achieved by using a pulse generator to pass current though the cells concerned for a brief period of time. If, for example, one wishes to cause the ulnar nerve in the arm to propagate a depolarization pulse, this nerve can be stimulated by passing current through the arm with the aid of two surface electrodes placed so that part of the current flow between them passes through the ulnar nerve.

In general, it is impossible to localize stimulating current to a particular cell or small group of cells. Current is passed through the bulk tissue surrounding these cells and a finite, *but unpredictable*, portion of this current will pass through the cells. Since it is *impossible* to *localize* current flow to a particular cell, it is impossible to know how much current will flow through it.
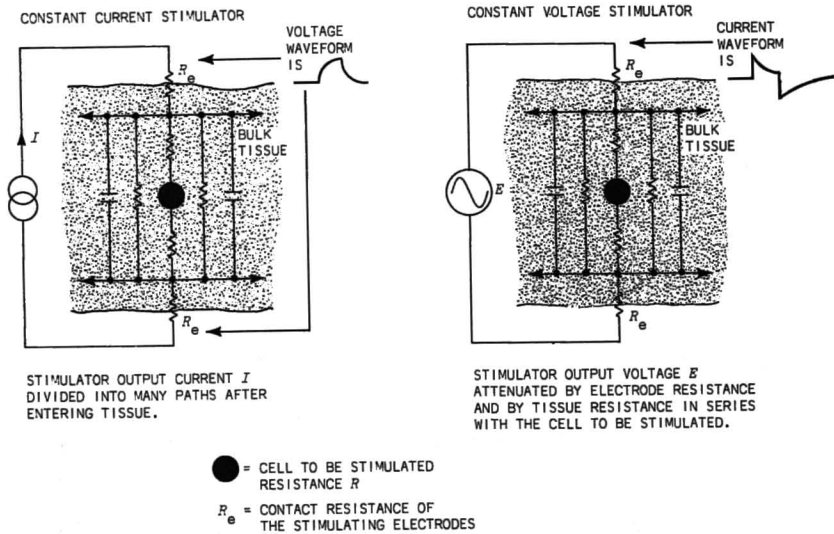
Fig. 12-1. Constant current and constant voltage stimulation.
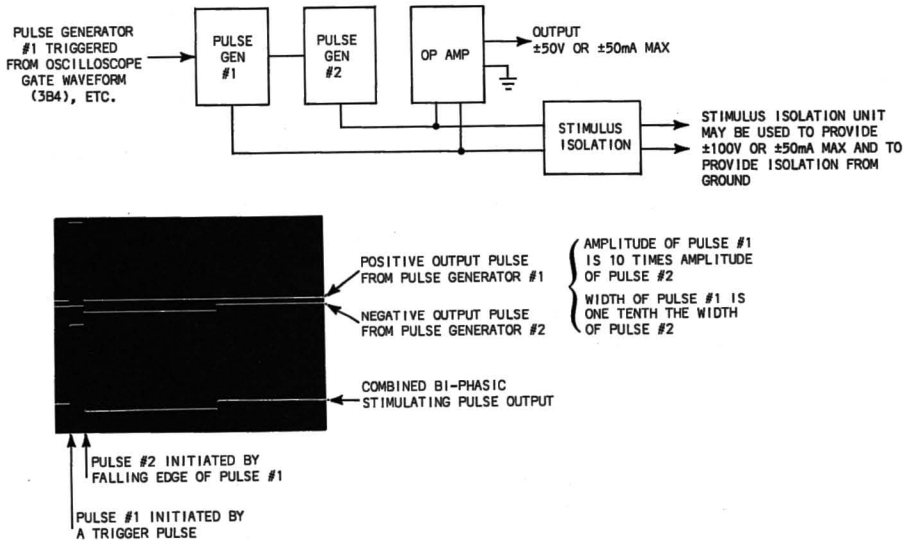
Fig. 12-2. Biphasic stimulation pulse generator.

stimulator
types

Stimulators provide an output from either a constant current source ($Z_0 > 10,000$ ohms) or from a constant voltage source ($Z_0 < 50$ ohms). Both constant current stimulators and constant voltage stimulators are extensively used and arguments for and against constant current or constant voltage stimulation have existed since the very early days of physiology. Although this controversy as to the best type of stimulator for routine use still exists, neither type allows an accurate prediction of the amount of current passing through the cells concerned. Satisfactory results can be obtained with either type. Many modern pulse generators intended for tissue stimulation provide both constant current and constant voltage output characteristics. A pair of stimulating electrodes immersed in tissue fluid typically has an impedance of about 500 ohms at the frequencies involved in the stimulus pulse.

depolari-
zation

biphasic
stimulation

If a stimulus current of insufficient intensity to cause cell depolarization is applied to a cell, the cell membrane resting potential will be reduced from its normal -90 millivolts. If an additional stimulus pulse of the same intensity is then applied to the cell within the next few milliseconds or so, the cell may then depolarize, as the cell membrane potential has not yet returned to its normal resting potential, as discussed in Chapter 1. The cell would thus appear to be more sensitive to stimulation from the second pulse than from the first. It has been suggested that cell recovery from the effect of a stimulus current can be hastened by the passage of an opposing, lower intensity, current for an appreciably longer period so that the net quantity of electricity is zero. This feature is incorporated in the biphasic stimulator shown in Fig. 12-2. Biphasic stimulation also neutralizes recording electrode polarization where silver/silver chloride recording electrodes are not used and helps to maintain a fixed baseline on DC-coupled recording systems. In the waveform shown in Fig. 12-2, the stimulating pulse is followed by a pulse of opposite polarity, of one-tenth the amplitude and ten times the width.

RECORDING
ELECTRODES

AMPLIFIER

STIMULATOR

STIMULATING
ELECTRODES

EQUIVALENT TO

Z_s = STIMULATOR IMPEDANCE TO GROUND
(ZERO FOR GROUNDED STIMULATOR)

Z = IMPEDANCE LEAD TO STIMULATOR

Z_1 = TISSUE IMPEDANCE BETWEEN STIMULATOR
AND AMPLIFIER + INPUT

Z_2 = TISSUE IMPEDANCE BETWEEN STIMULATOR
AND AMPLIFIER - INPUT

Z_i = INPUT IMPEDANCE OF AMPLIFIER

STIMULATOR
OUTPUT VOLTS

Z_1

Z_2

Z

Z_s

e

DIFFERENTIAL
AMPLIFIER
INPUT IMPEDANCE Z_i

E = OUTPUT VOLTAGE OF STIMULATOR

e = DIFFERENTIAL SIGNAL BETWEEN
AMPLIFIER + AND - INPUTS

$e \propto \dfrac{(Z_2 - Z_1)}{Z_s + Z_i}$ SINCE $Z_2 - Z_1$ WILL NEVER BE ZERO

e CAN BE MINIMIZED BY MAKING Z_s AND Z_i LARGE:

TYPICAL Z_s = 20pF, $10^9 \Omega$ AND Z_i = 30pF, $10^7 \Omega$

EQUIVALENT TO

E

Z

Z_1

Z_2

Z_s

Z_i

Z_i

e PRODUCED WITH 100V STIMULATION AT 1mV/DIV RECORDED 5cm FROM STIMULATOR

e STIMULUS ARTIFACT WITH GROUNDED STIMULATOR
AND 100MΩ, 2pF AMPLIFIER

e STIMULUS ARTIFACT WITH ISOLATED STIMULATOR
AND 1MΩ, 47pF AMPLIFIER

e STIMULUS ARTIFACT WITH GROUNDED STIMULATOR
AND 1MΩ, 47pF AMPLIFIER

1mV/DIV

1ms/DIV

Fig. 12-3. Equivalent circuit of stimulator, subject and amplifier.

12.2 STIMULUS ISOLATION

The previous section referred to stimulation.
Stimulation may be produced directly from a pulse
generator or it may be produced via a stimulus
isolation unit. The results of stimulation may be
detected with an oscilloscope and it is important
that the pulse produced by the stimulator does not
interfere with the response obtained on the
oscilloscope. It is thus necessary to insure that
the stimulating current is limited to the area
between the two stimulating electrodes and that
little or no stimulating current appears in the
region of the recording electrodes.

reducing
stimulus
artifact

Any stimulating current flowing near the recording
electrodes will cause a potential difference between
these electrodes which will appear as an out-of-phase
signal and thus not be rejected by the differential
amplifier. The equivalent circuit involved in
stimulating and recording is shown in Fig. 12-3. In
this circuit, most of the stimulating current flows
through the impedance formed by the tissue between
the stimulating electrodes. It is apparent, however,
that two alternative current paths exist; via the
tissue impedances Z_1 and Z_2 and the amplifier input
impedances to ground. If it were possible to make
Z_1 equal to Z_2, then no in-phase signal would be
produced at the recording electrodes. These
impedances are not controllable however, and it is
impossible in practice to balance them by careful
placement of the recording electrodes.

isolated
stimulator

high input
Z amplifier

Since it is not possible to balance the tissue
impedances, then one must reduce the stimulating
current passing through these impedances to as low
a value as possible in an effort to reduce the
differential signal appearing at the recording
electrodes. To reduce these currents, it is necessary
to either increase the input impedance of the
differential amplifier or to increase the impedance
between stimulator and ground. In practice one
attempts to make both these impedances as large as
possible in an effort to achieve the maximum
reduction in the level of stimulus signal appearing
at the recording electrodes. With a nonisolated
stimulator, the impedance between stimulator and
ground is zero, however with an isolated stimulator
this impedance can be made very high, typically 20
picofarads and 10^9 ohms. A typical differential
amplifier input impedance would be 30 picofarads and
10^7 ohms.

If a grounded stimulator is used on tissue, the ground electrode should *always* be placed between the "hot" stimulating electrode and the recording electrodes, to minimize stimulus artifacts. The upper trace shown on the waveform in Fig. 12-3 shows an acceptable level of stimulator pulse appearing at the recording electrodes when the recording electrodes are approximately 5 centimeters from the stimulating electrodes. This trace shows a 100 volt stimulator pulse reduced to less than 0.5 millivolts, a reduction of 200,000:1, which is entirely acceptable. The second trace shown on the waveform in Fig. 12-3 is also acceptable as, although the level of stimulator pulse appearing at the recording electrodes is substantially greater than shown on the upper trace, the amplifier quickly returns to zero after the stimulating pulse returns to zero. The lower trace shown on the waveform in Fig. 12-3 is unacceptable. In this instance, the stimulating pulse has severely overloaded the differential amplifier and the amplifier takes longer than 10 milliseconds to return to zero. Any response occurring in this 10 millisecond period would be either camouflaged by, or distorted by, this amplifier overload characteristic.

acceptable artifacts

amplifier overload

In practice, complete reduction of the stimulator pulse is rarely achieved and a severely attenuated stimulator pulse will appear on the oscilloscope trace. This attenuated stimulator pulse is referred to as stimulus artifact. Stimulus artifact is not altogether undesirable if it is not of excessive amplitude as it does provide a time reference on the CRT trace.

AC-coupled amplifiers

The above discussion assumes the use of DC-coupled differential amplifiers. If AC-coupled differential amplifiers are used, the time constant of the input coupling capacitors and the input resistance will determine the amplifiers recovery characteristics after overload.

12.3 STRENGTH/DURATION CURVES

Strength/duration curves show the excitability characteristics of muscle or nerve fibers. Stimulation of a cell is achieved by the passage of a certain *quantity* of electricity through the cell,

thus stimulation is dependent on charge rather than on current. Stimulation can be achieved by passing a large current for a short period or with a lesser current for a longer period. Since diffusion within the cell tends to oppose the stimulating current, a lower limit of current is reached below which stimulation will not occur, no matter how long the current is maintained.

For any given muscle or nerve fiber a strength/ duration curve may be drawn, as shown in Fig. 12-4, which represents the minimum stimulus or threshold required to stimulate the muscle or nerve fiber. Referring to the strength/duration curve shown in Fig. 12-4 for a normal muscle, it can be seen that a 0.01 millisecond stimulus pulse of 3.5 amplitude units applied in the region of the nerve associated with the muscle will cause the nerve to depolarize, causing muscular action. Alternatively, muscle action can be produced with a 0.03 millisecond wide stimulus pulse of 1.7 amplitude units, a 0.1 millisecond wide stimulus pulse of 1.1 amplitude units or a 0.3 millisecond wide stimulus pulse of 1.0 amplitude units. It is also apparent that, for pulse widths greater than 0.3 millisecond, no further reduction in stimulus amplitude will be effective. This amplitude, below which stimulation will not occur, no matter what the stimulus pulse width is, nerve tissue   is referred to as the nerve tissue threshold or threshold   *rheobase* for the muscle. The term is now rarely used.
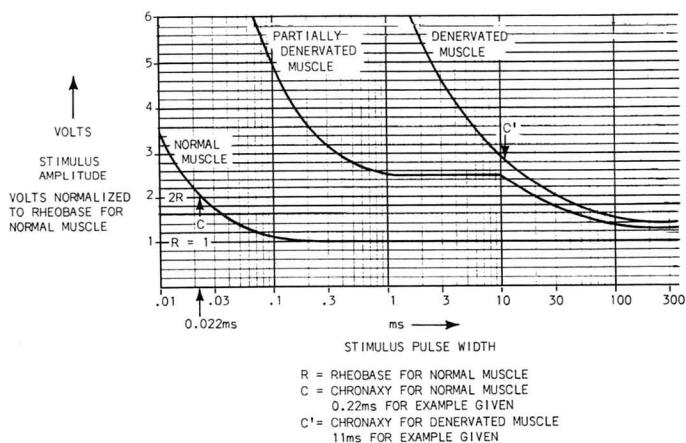
Fig. 12-4. The strength/duration or S-D curve.

Pulse widths of from 0.03 millisecond to 100 millisecond in eight steps have been proposed as an international standard for strength/duration curve determination.

In the particular example shown in Fig. 12-4 this rheobase occurred when using a constant voltage stimulator set at 30 V. Since, as discussed earlier, this absolute voltage level has little meaning,

pulse
amplitude
normalized
to rheobase

strength/duration curves are normally shown with a stimulus amplitude normalized to the rheobase, that is 30 volts in this instance is equivalent to 1.0 unit of stimulus amplitude, 2.0 units of stimulus amplitude being equivalent to 60 volts, etc.

Once a strength/duration curve has been determined, it is desirable to have a technique whereby curves for various muscles can be compared. Strength/duration curves may be compared by comparing the

chronaxy

chronaxy obtained from the curve. The chronaxy is the minimum pulse width required to excite the tissue for a stimulus of twice the amplitude of the rheobase. The chronaxy, in the example shown on Fig. 12-4, is 0.022 millisecond for normal muscle.

denervated
muscle
action

The strength/duration curve for a denervated muscle is also shown in Fig. 12-4. A denervated muscle exists when the nerve connection to the muscle has effectively been interrupted. This may be achieved by impairing the motor end plate action with drugs such as curare. It can be seen from Fig. 12-4 that a denervated muscle is less sensitive to stimulus than a normal muscle. In a normal muscle the stimulating current excites the more sensitive nerve fibers which in turn excites the muscle whereas in a denervated muscle the stimulating current must stimulate the muscle directly. Thus the strength/duration curve for a normal muscle shows the excitability characteristics for a single motor neuron, however, the strength/duration curve for a denervated muscle shows the excitability characteristics of a single muscle fiber.

Fig. 12-4 also shows a strength/duration curve for a partially denervated muscle. It can be seen, for pulse widths up to 10 milliseconds, muscle action is due to stimulation of the nerve whereas for pulse widths greater than 10 milliseconds muscle action is due to stimulation of the muscle directly. The strength/duration curve is, thus, displaying the excitability characteristics of the component which has the lower threshold for a specific pulse width.

muscle
reaction to
stimuli

When preparing strength/duration curves, muscle action is determined by viewing the muscle concerned in a well-lit environment. When stimulating with a 100 millisecond pulse, the muscular contraction produced is brisk and pronounced when the contraction is due to excitation of the nerve fibers and is sluggish and wormlike when the contraction is due to direct excitation of the muscle fibers. When stimulating with narrower pulses, muscle action is characterized by a twitching of the muscle concerned. Muscle action can also be determined by using some form of force-gage attached to the muscle concerned to detect muscular movement or, more commonly, by recording the electrical activity produced within the muscle when the muscle is stimulated by the nerve.

abbreviated
strength/
duration
curves

In some instances, the plotting of the strength/duration curve is somewhat tedious and abbreviated strength/duration curves are obtained by determining the stimulus amplitude required for a 100 millisecond pulse and a 1 millisecond pulse and expressing these as a ratio. In a normal muscle this ratio should be approximately unity, in a partly denervated muscle it will be between 1.5 and 4 and in a fully denervated muscle it will be between 4 and infinity. Abbreviated strength/duration curves may also be obtained by determining the rheobase and then determining the chronaxy directly by doubling the stimulus amplitude and reducing the pulse width to a point where the tissue can no longer be stimulated. This point will represent the chronaxy.
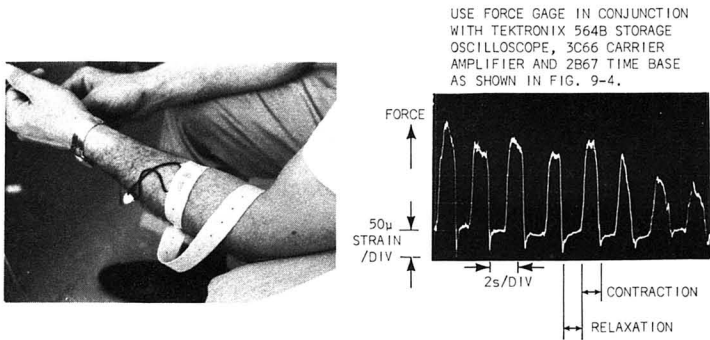
Fig. 12-5. A myrograph using an elastic force gage.

12.4 MYOGRAPHY

Myography is the study of muscular contractions and
a myograph is an apparatus for recording the
mechanical effects of a muscular contraction. A
myograph may simply consist of a displacement
transducer or a force transducer mechanically coupled
to the muscle under investigation. As shown in
Fig. 12-5, an elastic strip is placed around the
strain   muscle concerned and a strain gage is bonded to this
gage   elastic strip. Muscular contraction causes a tension
increase in the elastic strip which results in a
resistance change in the strain gage. The muscular
contraction may be initiated voluntarily or produced
by electrical stimulation. A strain gage myograph
may be used with a Tektronix Type 564B Storage
Oscilloscope and a Tektronix Type 3C66 Carrier
Amplifier as shown in Chapter 9, Fig. 9-4. The
output from such a recording system, a series of
muscular contractions over a 20-second period, is
also shown in Fig. 12-5. Force myographs are
particularly suited to exercising subjects or for
the study of muscular fatigue over prolonged periods.

GROUND
ELECTRODE

POWER
AMP

SPEAKER

DUAL TRACE
STORAGE OSCILLOSCOPE

564B

DIFF
AMP
3A9

3A72
CH1
AMP
CH2

3B4
TIME
BASE

PICKUP
ELECTRODES
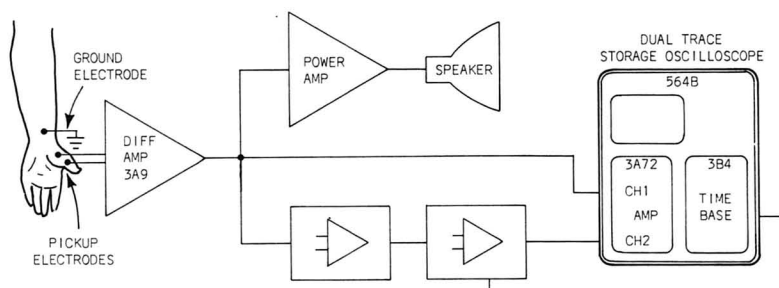
3A9 AND 3A8 USED IN ANY
TEKTRONIX 560 SERIES
OSCILLOSCOPE (e.g., 561B).

ABSOLUTE VALUE
CIRCUIT - "FULL
WAVE RECTIFIER"

INTEGRATOR GATED
WITH TIME BASE

3A8 OPERATIONAL AMPLIFIER
WITH ABSOLUTE VALUE ADAPTER
AND GATING ADAPTER.
(SEE CHAPTER 29 FOR DETAILS)

3A9   DIFFERENTIAL AMPLIFIER SET AT 1mV/DIV. PROVIDES A GAIN
OF 1000X TO THE 3A8 OPERATIONAL AMPLIFIER AND 3A72 CHANNEL 1.

3A72   CHANNEL 1 SET AT 1V/DIV PROVIDING A SENSITIVITY
REFERRED TO THE ELECTRODES OF 1mV/DIV.

3A8   OPERATIONAL AMPLIFIER #1 USED IN CONJUNCTION WITH AN
ABSOLUTE VALUE CIRCUIT AS SHOWN IN CHAPTER 29.

3A8   OPERATIONAL AMPLIFIER #2 USED IN CONJUNCTION WITH A LOW
SPEED GATING ADAPTER AS SHOWN IN CHAPTER 29.
Z_i SET AT 1mΩ, Z_f AT 0.1μF FOR INTEGRATION.

3A72   CHANNEL 2 SET AT 0.1, 0.2 OR 0.5V/cm PROVIDING VARYING
INTEGRATION SENSITIVITIES.

INTEGRATOR CALIBRATION - OSCILLOSCOPE AMPLIFIER #2 AT 0.2V/DIV

0.8mV

2.2 DIV

20ms/DIV   TOTAL INTEGRAL
PERIOD IS 0.2s
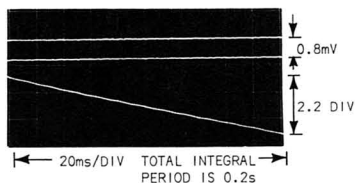
4mV CALIBRATOR SIGNAL FROM OSCILLOSCOPE CALIBRATOR
CONNECTED TO DIFFERENTIAL AMPLIFIER.

DIFFERENTIAL AMPLIFIER GAIN REDUCED BY 5X (FROM 1mV/DIV
TO 5mV/DIV) PROVIDING A CALIBRATION VOLTAGE RELATIVE TO
THE EMG SYSTEM OF 0.8mV.

CALIBRATOR DUTY FACTOR IS 50%.

CALIBRATOR INPUT SHOWN = $0.8 \times 10^{-3} \times \frac{50}{100} \times 0.2$ VOLT SECONDS

= 80×10^{-6} VOLT SECONDS.

2.2 DIV OF INTEGRATOR OUTPUT = 80×10^{-6} VOLT SECONDS
∴ INTEGRATOR CALIBRATION IS 36×10^{-6} VOLT SECONDS/DIV
WITH OSCILLOSCOPE AMPLIFIER #2 AT 0.2V/DIV.

INTEGRATOR OUTPUT =
180×10^{-6} VOLT SECONDS INPUT/VOLTS OUTPUT

Fig. 12-6. An electromyograph system.

12.5   ELECTROMYOGRAPHY -- EMG

muscle
fiber
depolari-
zation

Whereas the myograph records the mechanical effects of a muscular contraction, the electromyograph records the electrical effects of such a contraction. Muscular contraction is caused by depolarization of the muscle fibers. This depolarization produces an action potential as covered in Chapters 1 and 3. This muscular action potential is known as the electromyogram, or EMG. An electromyogram will be produced in a muscle when the muscle contraction is caused either by voluntary muscle action or by electrical stimulation of the muscle.

12.6   ELECTROMYOGRAPHY WITH VOLUNTARY MUSCULAR ACTION

instrumen-
tation

A typical system for recording the electromyograph produced by voluntary muscle action is shown in Fig. 12-6. Further details on this system are given in Chapter 29. The muscle action potential is picked up by needle electrodes inserted into the muscle or by surface electrodes placed over the muscle concerned and then amplified by a suitable differential amplifier. The EMG can then be detected audibly by using a speaker in conjunction with an audio amplifier. The EMG may also be displayed directly on an oscilloscope or may be converted to an absolute integral and then displayed on an oscilloscope. Both direct displays and integrated displays are shown in Fig. 12-7.

absolute
integrals

Referring to Fig. 12-7, the upper trace on the top photograph shows an EMG produced by a mild voluntary contraction. The action potential produced by a single motor unit can clearly be differentiated from other action potentials. The absolute integral of this activity is displayed on the lower trace in the same photograph. The integral displays the quantity of electricity involved in the muscular contraction. The quantity of electricity associated with the single motor unit action potential can also be determined.

With a more forceful voluntary contraction, as shown in the middle photograph, many motor units are involved and the EMG obtained is the result of the action potential produced by all these motor units; the resulting integral being greater than the integral obtained for a single motor unit.

By using a slower sweep speed in conjunction with the system shown in Fig. 12-6, the EMG and absolute integral for a series of contractions may be displayed as shown in the bottom photograph.
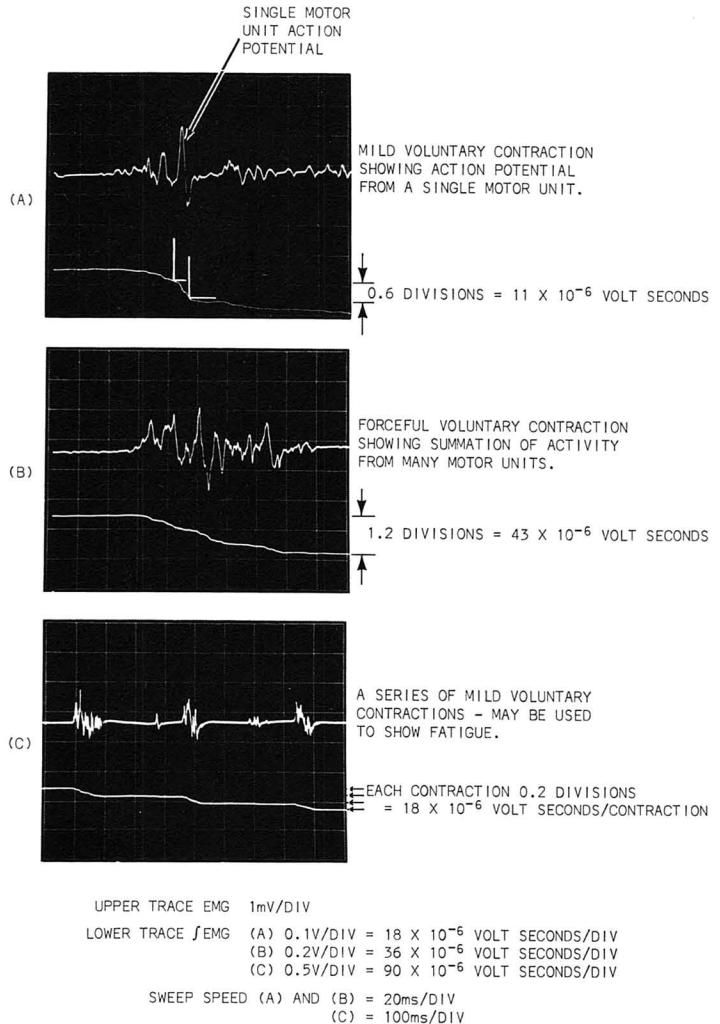
Fig. 12-7. Results obtained with the system shown in Fig. 12-6.

The quantity of electrical activity produced by a muscular contraction of a montomic function of the strength of the contraction. Since it is difficult to estimate this quantity from an observation of the EMG waveform, the absolute integral of the EMG is used as a measure of this quantity. The integrator output is calibrated in units of electrical quantity, that is, volt seconds. Since one volt second is a substantial quantity of electricity, it is preferable to refer to EMG's in smaller units, one "unit" being equal to 10^{-6} volt seconds. It can be seen from Fig. 12-7 that for a particular subject and a particular muscle 11 units are produced by a single motor unit, 18 units by a mild muscular contraction and 43 units by a forceful contraction.

electrical
quantity

volt
seconds

Referring to Fig. 12-6, the absolute integral is obtained by full-wave rectification of the electromyograph and then integrating this rectified signal with an integrator gated "on" for the duration of the oscilloscope sweep. The whole system consisting of the differential amplifier, the full-wave rectifier, the integrator and the display oscilloscope can be calibrated using an oscilloscope calibrator waveform as shown in Fig. 12-6. The system shown in Fig. 12-6 may be duplicated by using one Type 561B Oscilloscope in conjunction with a Type 3A9 Differential Amplifier and a 3A8 Operational Amplifier and one Type 564B Storage Oscilloscope. in conjunction with a Type 3A72 Dual Trace Amplifier and a 3B4 Time Base unit. Details of the absolute value adapter and gating adapter for the 3A8 are given in Chapter 29.

calibration

audio
output

The EMG is often presented audibly in clinical applications and the trained listener can judge the condition of the muscle by the volume and characteristic tones produced by the audio system during a muscular contraction.

12.7 ELECTROMYOGRAPHY DURING ELECTRICAL STIMULATION

voluntary
versus
stimulated
EMG

The EMG produced during a voluntary twitch is spread out over a period of 100 milliseconds or more as the nerve impulses to various motor units are not time coincident as the propagation delay from the spinal cord to the muscle concerned is different for all nerve fibers. Also, since the contraction is voluntary, any one motor unit may produce several action potentials, the frequency of discharge being determined within the spinal cord. Such is not the case when recording EMG produced by electrical stimulation. All neurons with thresholds above the stimulating intensity are simultaneously stimulated by the electrical impulse, thus all muscle fibers discharge simultaneously, producing substantial activity for a brief period of time, typically less than 10 milliseconds. Although the response obtained when stimulating is referred to as an EMG, it is an unnatural occurrence and should perhaps be more correctly referred to as a "myographic response" or a "muscle action potential." The stimulus pulse used to initiate this response usually has an amplitude of >100 volts and is either 0.1 millisecond, 0.3 millisecond, or, occasionally, 0.5 millisecond wide.
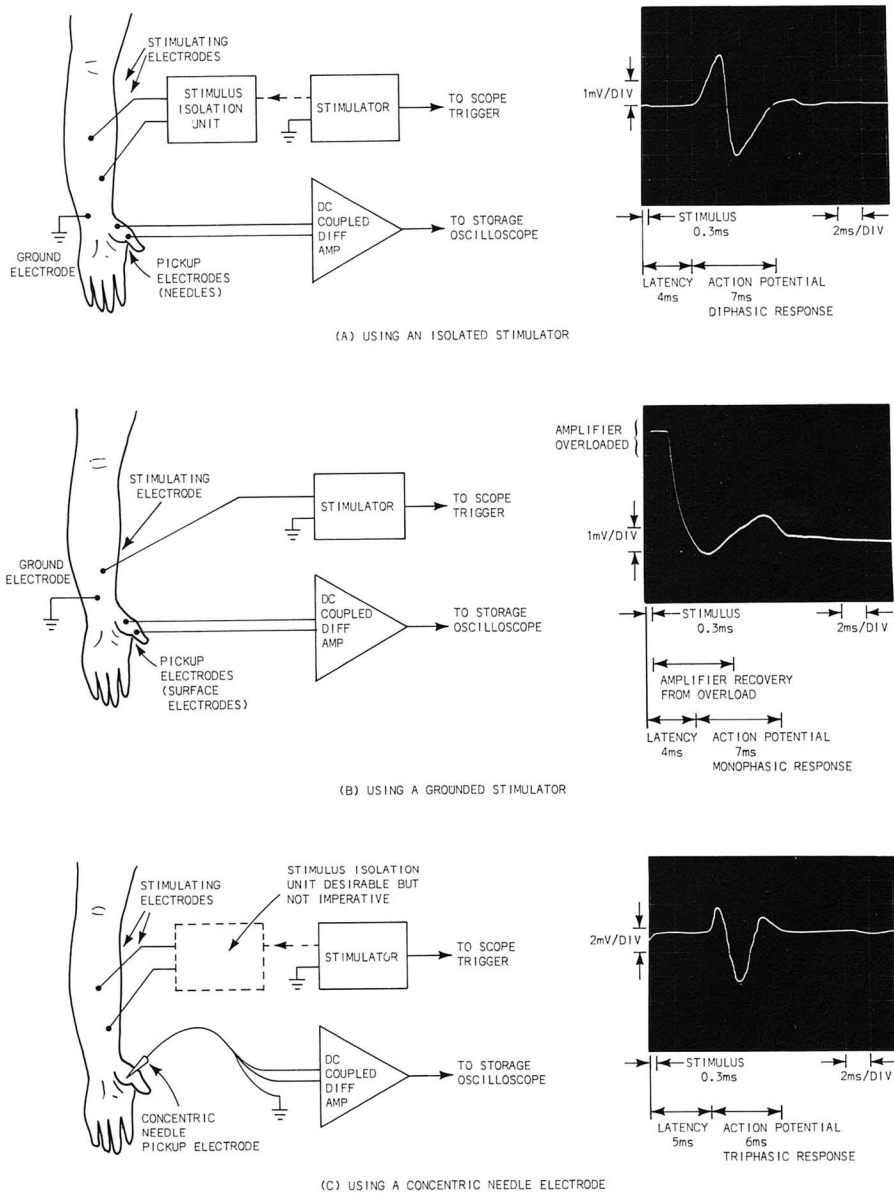
Fig. 12-8. EMG produced by electrical stimulation.

latency

A typical EMG produced by electrical stimulation is shown in Fig. 12-8A. A 0.3 millisecond stimulus pulse was used to initiate the response. A delay occurred between the stimulating pulse and the response; this delay is referred to as latency. The action potential shown in Fig. 12-8A has a latency of 4 milliseconds and produces an action potential covering the following 7 milliseconds. This action potential is referred to as diphasic as it shows a single positive deflection followed by a single negative deflection.

diphasic response

monophasic response

The action potential shown in Fig. 12-8B is referred to as monophasic as the action potential appears to be comprised of only one positive deflection. This action potential is partially camouflaged by the amplifier recovery characteristics due to the use of a nonisolated stimulator. Such amplifier characteristics were covered earlier in this chapter. If a grounded stimulator is used on tissue, the ground electrode should *always* be placed between the "hot" stimulating electrode and the recording electrodes, to minimize stimulus artifacts.

triphasic response

The EMG shown in Fig. 12-8C is referred to as triphasic as two positive deflections and one negative deflection are exhibited. This EMG was recorded using a concentric needle pickup electrode which effectively locates two electrodes less than 1 millimeter apart in the muscle in an attempt to record the action potential produced by a single motor unit rather than by the complete muscle. All three EMG's shown in Fig. 12-8 are considered acceptable and basically serve to show that the nerve and muscle relationship is functioning correctly as the latency is not excessive and an action potential is, in fact, generated.

12.8 THE H REFLEX

The muscular reflex response generated within the spinal cord was covered in Chapter 3. When recording EMG's produced by electrical stimulation, the stimulating current excites the motor nerve which in turn initiates a response in the muscle concerned.

This stimulating current also excites the sensory nerve. It is possible to decrease the stimulus level to a point where the stimulus intensity is insufficient to excite the motor nerve but is sufficient to excite the more sensitive sensory nerve. The depolarization pulse propagated in the sensory nerve as a result of this stimulation travels to the spinal cord where a reflex response is in turn propagated in the motor nerve. This reflex response propagates along the motor nerve to the muscle concerned, initiating a muscular response.

muscle
response
via
reflex
stimulation

MUSCLE ACTION PRODUCED BY STIMULUS CURRENT DEPOLARIZING MOTOR NERVE (DIRECT) OR BY STIMULUS CURRENT DEPOLARIZING SENSORY NERVE WHICH IN TURN DEPOLARIZES MOTOR NERVE DUE TO REFLEX RESPONSE GENERATED IN SPINAL CORD (INDIRECT).
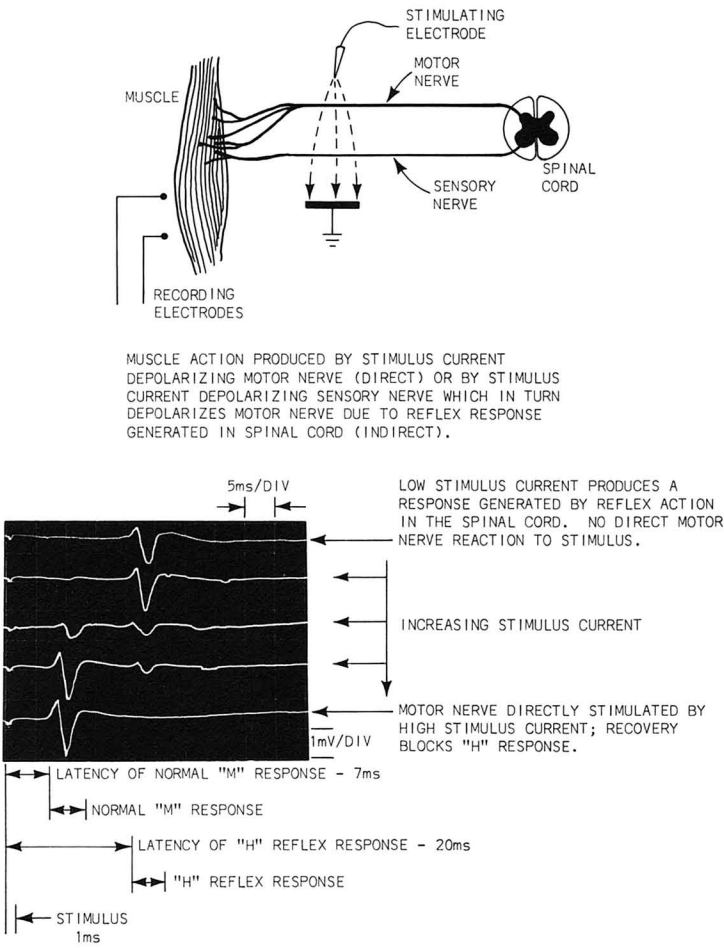
Fig. 12-9. The H-Reflex response.

The physical relationship between the muscle, the nerve and the spinal cord is shown in Fig. 12-9, together with the results obtained when recording the EMG with progressively increasing stimulus intensity. At low stimulus intensity a response having a latency of approximately 20 milliseconds is detected, known as the "H reflex" response. This latency is due to the conduction time from the stimulating point along the sensory nerve to the spinal cord and thence from the spinal cord along the motor nerve to the muscle concerned. As the stimulating current is progressively increased this H reflex response decreases and a normal or "M reflex" response appears with a normal latency of 7 milliseconds, representing the conduction time from the stimulus site, via the motor nerve, to the muscle concerned. The "H reflex" response can be used to determine the condition of the reflex system.

H reflex

M reflex

12.9 NERVE CONDUCTION

motor nerve propagation velocity

The propagation velocity of the nerve impulse along the motor nerve from the stimulus site to the muscle can be determined as shown in Fig. 12-10. In the example shown, the peroneal nerve of the left leg is stimulated behind the knee and a muscular response is detected in the foot, using either surface electrodes or needle electrodes. The response shown has a latency of 11.5 milliseconds. The stimulus electrodes are then moved to a point behind the ankle and a response obtained in the foot having a latency of 4 milliseconds. The difference between these two latencies is attributed to the conduction time required for the nerve impulse to propagate along the motor nerve from the knee to the ankle.

PERONEAL NERVE OF THE LEFT LEG
MOTOR NERVE STIMULATED AND ACTION
POTENTIAL DETECTED IN APPROPRIATE MUSCLE.

2mV/DIV

STIMULUS SITE #1
KNEE
RESULTING ACTION POTENTIAL

KNEE TO ANKLE 38cm

RESULTING ACTION POTENTIAL

STIMULUS SITE #2
ANKLE

TO OSCILLOSCOPE

2ms/DIV
TIME DIFFERENCE BETWEEN ACTION POTENTIAL - 7.5ms

CONDUCTION VELOCITY - 38cm IN 7.5ms
- 51m/s
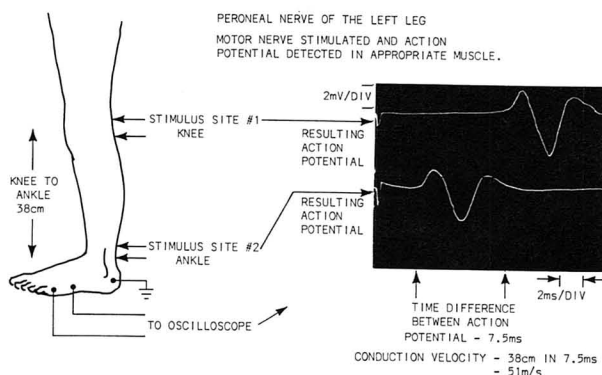
Fig. 12-10. Motor nerve conduction velocity determination.

The propagation velocity can be determined by
measuring the distance from the knee stimulation
point to the ankle stimulation point and dividing
it by the difference in latencies. A 100 volt
stimulus pulse, 0.3 millisecond wide, is usually used.
Occasionally, pulse widths of either 0.1 millisecond
or 0.5 millisecond are used. The stimulus should be
repeated several times to ensure that the response
obtained is consistent.

sensory
nerve
propagation
velocity

A similar technique can be used to measure sensory
nerve conduction velocity as shown in Fig. 12-11.
In Fig. 12-10 we moved the stimulus site with
respect to a fixed recording site, however, when
attempting to record sensory nerve conduction
velocity, it is necessary to stimulate at a fixed
sense receptor site and to record the propagation of
this stimulus pulse along the sensory nerve by
detecting the nerve impulse or "traveling wave of
depolarization" at various sites along the nerve.
Fig. 12-11 shows the results obtained when
stimulating the hand and recording the propagation
of this pulse along the ulnar nerve at four points
along the length of the nerve using a four-channel
oscilloscope. If the vertical position of the four
channels on this oscilloscope are adjusted to
represent distance in centimeters from the stimulus
site, the various latencies obtained should be
directly proportional to this distance, resulting in
the straight line shown in Fig. 12-11. Any
deviation in straightness in this line would
represent a change in conduction velocity. Injury
to the nerve will normally result in decreased
conduction velocity in the injured part of the nerve.

ULNAR NERVE OF THE RIGHT ARM

SENSORY NERVE STIMULATED AND "TRAVELING WAVE OF DEPOLARIZATION AND REPOLARIZATION" DETECTED WITH SURFACE ELECTRODES

74

16cm

58

17cm

41

28cm

13

13cm

0

RECORDING SITE #1
NECK

RECORDING SITE #2
ARMPIT

RECORDING SITE #3
ELBOW

RECORDING SITE #4
WRIST

STIMULUS SITE
AT LITTLE FINGER

DISTANCE (IN cm) ALONG
ULNAR NERVE FROM
STIMULUS SITE

cm

80
70
60
50
40
30
20
10
0

#1
#2
#3
#4

12.5ms

5ms/DIV

NEAR STRAIGHT LINE SHOWS CONSTANT
CONDUCTION VELOCITY ALONG NERVE

THE FOUR RECORDINGS ARE STACKED
SO THAT THEIR DISTANCES FROM THE
BASELINE REPRESENT THE DISTANCE
FROM STIMULUS SITE TO RECORDING SITE

CONDUCTION VELOCITY - #1 TO #4 - (16 + 17 + 28)cm IN 12.5ms
= 49m/s

Fig. 12-11. Sensory nerve conduction velocity determination.

limb nerves
investigated

When measuring conduction velocity four nerves are principally investigated: the ulnar and median nerves of the arm and the peroneal and tibial nerves of the legs. The ulnar, peroneal and tibial nerves perform both general sensory and motor functions, however the median nerve primarily performs a sensory function. Responses from the ulnar nerve are normally detected on the back of the hand or on the fingers, the median nerve on the thumb or on the thick fleshy part of the hand near the thumb, the peroneal nerve on the four smaller toes or the top of the foot and the tibial nerve at the side of the foot near the larger toe or on the larger toe.

40 to 60
m/s normal

electro-
diagnosis

Normally, conduction velocity is measured, or EMG's are recorded, in an attempt to diagnose an abnormality in the subject and, since this abnormality usually manifests itself in only one side of the subject at a time, the limb on the other side of the subject may be regarded as normal and conduction velocities or EMG's compared between the two limbs. Although conduction velocity depends on the nerve under investigation, conduction velocities in most healthy nerves fall in the range from 40 to 60 meters per second; injury or abnormality being indicated by lower conduction velocities extending to below 10 meters per second. The term electrodiagnosis is applied to the general study of normal motor unit behavior. Complete electrodiagnosis should include most of the techniques covered in this chapter.

12.10 REPETITIVE STIMULATION

recovery
character-
istics

critical
frequency

So far this chapter has dealt with the effects of a single stimulating pulse on muscle and nerve fiber. In attempting to determine the recovery characteristics of a motor unit it is necessary to stimulate with a double pulse and to determine the delay required between the two pulses for the stimulus pulse to be "seen" by the muscle as two separate stimuli rather than as only one stimulus. This minimum pulse separation, when translated to frequency, is known as the critical frequency. The critical frequency varies for different muscles in the same subject and for the same muscle at a different temperature and in a different state of fatigue. The critical frequency for most of the major muscles in the human body lies between 5 and 15 Hz corresponding to minimum pulse separations of 200 milliseconds and 66 milliseconds.

Multiple pulse stimulation, rather than single pulse or double pulse stimulation, is used to determine the fatigue characteristics of nerve and muscle fiber. Under normal conditions, the response obtained during prolonged stimulation should show little change from the response obtained from a single stimulus as long as a brief relaxation period is allowed between each pulse.

12.11 SMOOTH MUSCLE POTENTIALS

Previously in this chapter we have dealt with the electrical activity produced in muscles known as skeletal muscle. Skeletal muscle produces contractions in a series of muscle fibers, the combined effect producing continuous motion. Other muscles, particularly the muscles surrounding the major organs in the torso, produce overall continuous contraction and are known as "smooth" muscles. "Smooth" refers to the microscopic appearance of the muscle, in contrast to "striated" skeletal muscle, with its characteristic cross stripes. Electrodes placed on, in, or over smooth muscle can be employed to detect contractions in these muscles. Electrodes appropriately inserted into the stomach, bladder, etc. or placed on the surface of the body over these organs can serve to monitor the slowly varying potentials generated by the muscles in these organs. These potentials are known as smooth muscle potentials. Although these potentials are rarely monitored, the recording of the electrical activity produced by the stomach has been termed the

electro-
gastrogram

electrogastrogram. The electrogastrogram is characterized by a slowly changing "DC" potential (below 1 Hz) which would normally be recorded on a DC-coupled instrument at a sensitivity of 10 millivolts per division and at a sweep speed of perhaps 1 second per division.

AREA OF SUBJECT WITH
A HIGH CONCENTRATION
OF SWEAT GLANDS
(PALMS AND SOLES)

TO 1. IMPEDANCE MEASURING CIRCUIT
TO DETECT δR

OR

2. DC AMPLIFIER TO DETECT E

δR

E

GSR DESIGNATED BY
EITHER CHANGE IN RESISTANCE (δR)
OR GENERATION OF POTENTIAL (E)
BETWEEN ELECTRODES

AREA OF SUBJECT DEVOID
OF SWEAT GLANDS
(TRUNK, EARS, AND LIMBS)
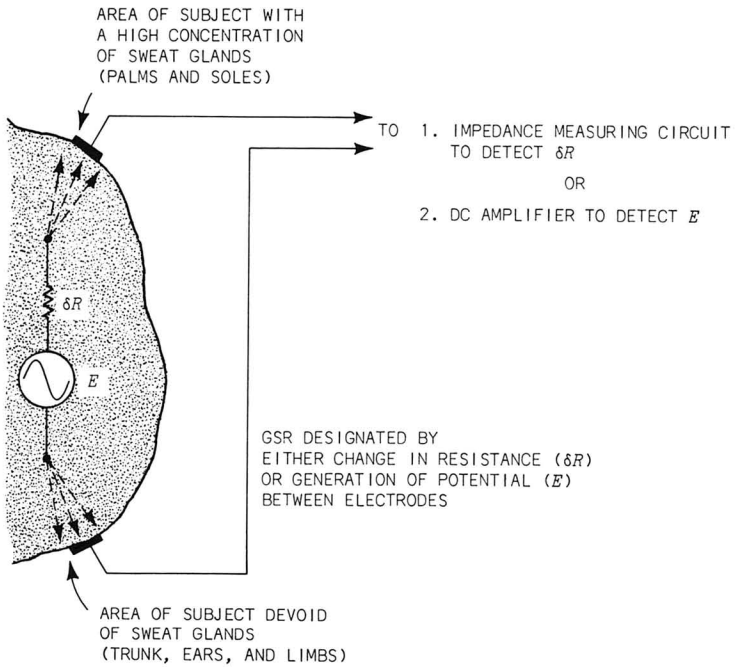
Fig. 13-1. GSR measurement.

13

GALVANIC SKIN REFLEX – GSR

13.1 THE AUTONOMIC NERVOUS SYSTEM

The human autonomic nervous system is the system within the body that regulates body functions such as temperature, respiration and glandular activity. When a subject is psychologically excited or is in some other elevated state of psychological activity, the subject perspires or "sweats." This is due to an emotional stimulus initiating a response in the autonomic nervous system which in turn produces a response in the subject's sweat glands. Detection of this sweat gland activity is, thus, an indication of the subject's psychological state or state of arousal.

sweat
gland
activity

Activity of the sweat glands is referred to by one or more of the following terms: electrical skin resistance (ESR), galvanic skin reflex (GSR), electrodermal response (EDR) and psychogalvanic reflex (PGR). Occasionally, the term GSR is referred to as the galvanic skin reaction or galvanic skin resistance. These terms all relate to one or both of the following physiological changes associated with sweat gland activity: a change in resistance and the generation of a potential between areas containing many sweat glands and areas almost devoid of them. The change in resistance is referred to

Fere
effect

as the Fere effect on the exosomatic response of the GSR. A decrease in the subject's resistance indicates arousal. Relaxation is indicated by an increase in resistance. The generation of a potential

Tarchanoff
effect

difference is referred to as the Tarchanoff effect or the endosomatic response of the GSR. This resistance change and potential generation is represented by δR and E in Fig. 13-1.

AC COUPLED
0.1Hz – 3dB LOWER FREQUENCY

3A9 AMP

CONSTANT CURRENT PULSE GEN

LEAD ELECTRODES PLACED ON HAND

10µA

0

4ms

400ms

REFER TO CHAPTER 29 FOR DETAILS OF THE CONSTANT CURRENT PULSE GENERATOR

EMOTIONAL STIMULUS

50kΩ

25kΩ

5s/DIV

AMPLIFIER GAIN AND POSITION ADJUSTED WITH 25kΩ AND 50kΩ CALIBRATION RESISTORS FOR SCALE CALIBRATION. ALLOWS FOR EFFECT OF AC COUPLING AND 3A9'S 1MΩ INPUT RESISTANCE.
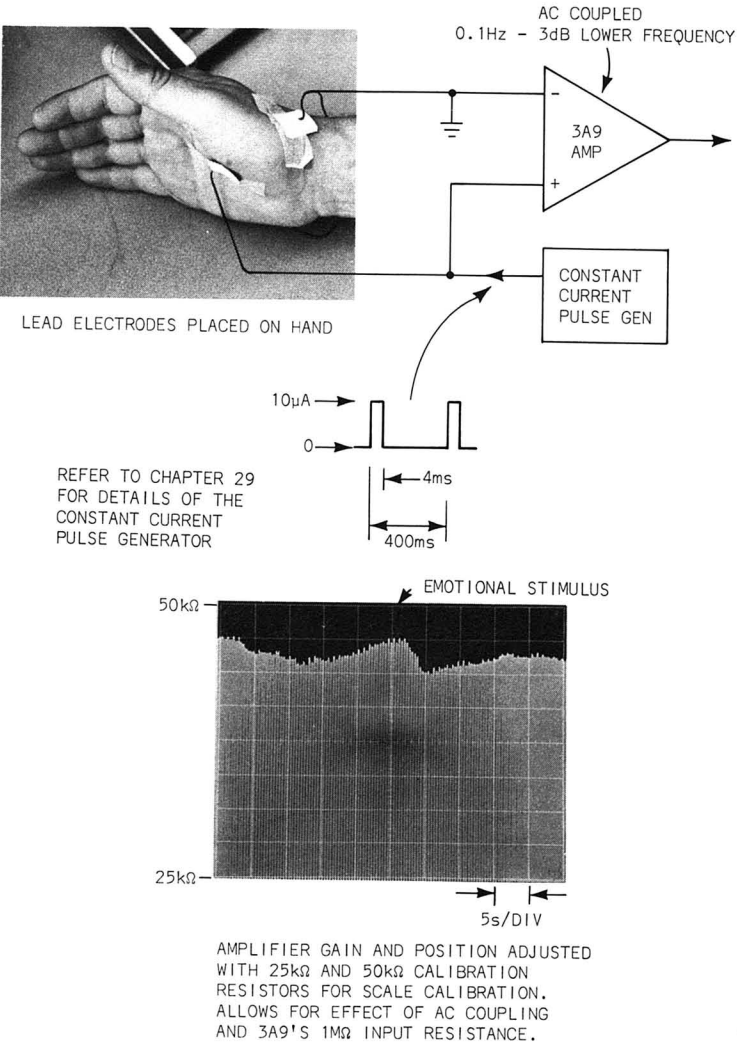
Fig. 13-2. GSR measurement.

13.2   GSR MEASUREMENT BY RESISTANCE CHANGE

GSR measurement by this technique involves the detection of an impedance change between two electrodes on the subject. The GSR primarily changes the resistive component of this impedance, it is thus important that the measurement technique used be insensitive to reactive component changes. The simplest technique would appear to be the passage of a DC current via the electrodes and the detection of the voltage drop produced between the electrodes due to this current flow. Since, however, the GSR is usually recorded for prolonged periods, the electrodes used cannot be relied upon as these electrodes will undoubtedly produce a DC offset potential of several hundred millivolts due to the passage of current for a prolonged period. This offset potential could not be differentiated from a potential produced by a change in the subject's resistance. For this reason, DC techniques have been found to be unsatisfactory.

DC techniques

Since it is desirable to employ a measurement technique sensitive primarily to resistance change, and since it is undesirable to use DC measurement techniques, very low frequency AC techniques are invariably used. These measurement techniques involve the passage of an AC current of perhaps 10 microamperes peak at a frequency of 2 or 3 Hz. The resulting voltage drop between the electrodes can then be detected as an AC signal which will be independent of any DC offset potentials generated at the electrodes; such a technique is shown in Fig. 13-2. Referring to Fig. 13-2, a constant-current pulse generator adapter, used in conjunction with a Type 561B or 564B oscilloscope's calibrator, provides pulses of 10 microamperes, with a duration of 40 milliseconds. Further details on this adapter are given in Chapter 29. Since this pulse waveform has only a 10% duty cycle, a change in subject resistance, when using an AC-coupled amplifier, will primarily alter the displayed pulse amplitude and will cause almost no shift in the oscilloscope zero level or base line.

AC techniques

electrodes

Since, when recording the GSR, we are recording resistance changes due to action of the sweat glands at the surface of the skin, electrodes should be used that make direct contact with the skin as the use of any conductive electrode paste would interfere with the action of the sweat glands. The electrodes used should also have no chemical effect on the action of the sweat glands. Thin lead plates are preferred for GSR electrodes as they meet the above requirements and they are also malleable and can be molded to suit the subject's contour. Lead plate electrodes are used in the photograph shown in Fig. 13-2; one electrode is placed on the palm of the hand in a region of high sweat gland concentration and the other electrode is placed on the back of the hand in a region almost completely devoid of sweat glands.

typical
resistances

The range of resistances encountered in normal subjects when recording between two electrodes on the hand is from 20,000 ohms to perhaps 0.2 megohms. In some instances, particularly if the autonomic nervous system is malfunctioning and the sweat glands are effectively denervated, the resistance between the electrodes will exceed 1 megohm.

13.3 GSR MEASUREMENT BY POTENTIAL DETECTION

offset
potential

GSR measurement by this technique involves the detection of a DC potential between two electrodes on the subject. This DC potential will normally be less than one millivolt. An electrode offset potential in excess of one millivolt may be produced at the electrode/subject interface and any unbalance in this offset potential between the two electrodes cannot be differentiated from the GSR potential. Although solid silver electrodes, or perhaps silver plate electrodes, are used, this offset potential unbalance is difficult to control and will invariably contribute a considerable DC potential to the GSR potential. For this reason, GSR measurement by this technique is rarely attempted.

13.4 ELECTRICAL SKIN RESISTANCE

sweat
gland
distribution

dermometer

The electrical skin resistance (ESR) is basically
the same as the GSR. The term ESR is, however,
usually reserved for measurement of the distribution
of sweat glands on the human body rather than the
actual change in the activity of these sweat glands.
The ESR is measured in the same way as the GSR,
however the ground electrode is applied by means
of a silver clip attached to the ear and the active
or exploring electrode consists of a noncorroding
silver or lead disc or a small roller wheel. Sweat
gland distribution is detected by moving the position
of this exploring electrode. The change in
resistance between a sweating and a nonsweating area
is distinct and a small movement of the exploring
electrode can result in a resistance change in
excess of 100 percent. Instruments specifically
designed for measurement of the electrical skin
resistance are referred to as dermometers.

14

ULTRASONOGRAPHY

acoustical
reflections

Ultrasonography is a technique by which ultrasonic energy is used to detect internal body organs. Bursts of ultrasonic energy are transmitted from a transducer through the skin and into the internal anatomy. When this energy strikes an interface between two tissues of different acoustical impedance, reflections are returned to the transducer. The transducer converts these reflections to an electrical signal. This electrical signal is amplified and displayed on an oscilloscope, each tissue interface appearing as a vertical deflection along the baseline of the oscilloscope at a distance proportional to the depth of the interface. This ultrasonic technique is similar to the time-domain reflectometry technique used to measure electrical cable length and the sonar technique used to detect objects under water.

advantages

While the use of pulse-echo ultrasonic energy is somewhat similar to the use of X-ray, the results obtained differ from an X-ray picture, being a cross-sectional projection or simply a linear projection rather than a profile of the area examined. Also, in contrast to X-ray, ultrasonography uses mechanical energy at a level which is not harmful to human tissue, thus, it may be used with safety on pregnant subjects and for frequent examination. Ultrasonography can detect materials that are not radiopaque, thus angiographic dyes are unnecessary. As commercial ultrasonic diagnostic instruments are easy to operate, ultrasonography is rapidly becoming a valuable diagnostic technique.
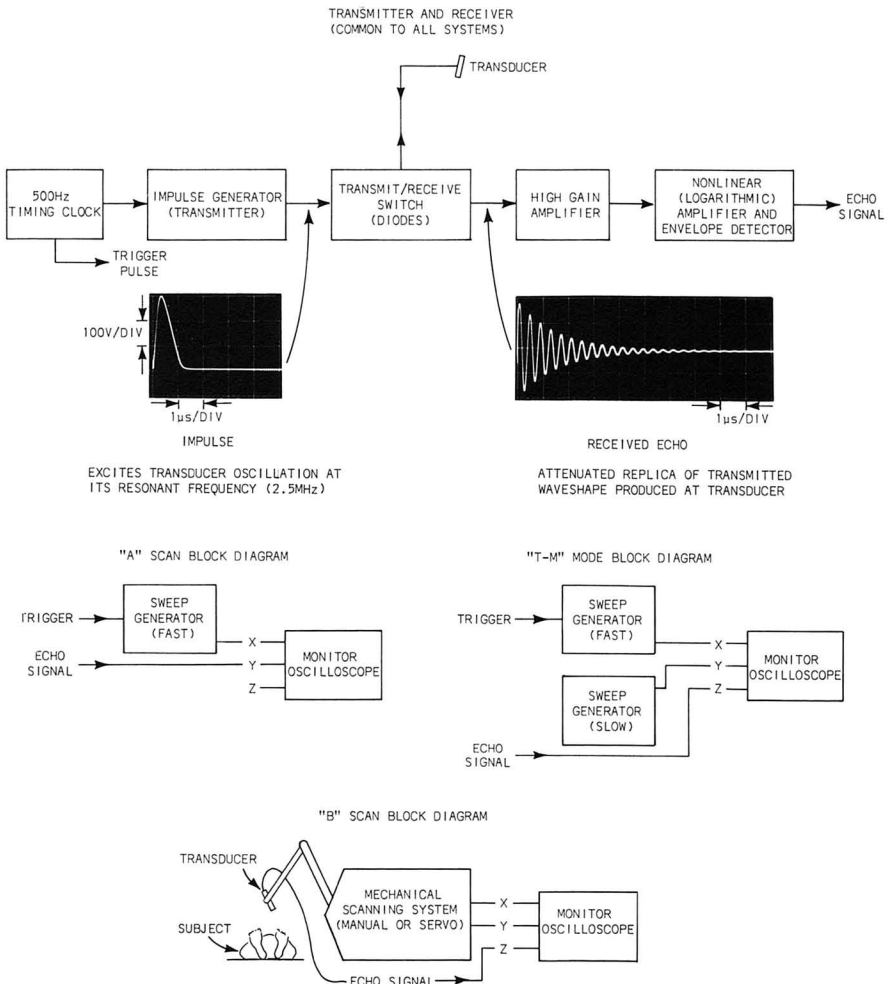
Fig. 14-1. Ultrasonic diagnostic systems - block diagrams.

While the basic function of ultrasonic diagnostic equipment is to measure distances between interfaces that separate body structures by timing the echos produced by these interfaces, this timed-echo information may be processed in various ways to produce different forms of display. The three commonly used display modes are the "A" scan mode, the "T-M" mode and the "B" mode; further details on these three operating modes are given later in this chapter.

display modes

14.1 ULTRASONIC SYSTEMS

Ultrasonic energy for use in ultrasonography is produced by exciting a piezoelectric crystal referred to as the ultrasonic transducer. These piezoelectric crystals normally have a self-resonant frequency of between 1 MHz and 10 MHz; the most common type used for ultrasonography having a self-resonant frequency of 2.5 MHz. When this crystal transducer is excited by an electrical impulse, it will ring at its self-resonant frequency and produce a train of damped oscillations. Fig. 14-1 shows the input pulse used to excite a 2.5 MHz transducer and a reflected pulse detected by the same transducer. The input pulse has an amplitude of almost 300 volts and a duration of 1 microsecond. The transducer oscillates at 2.5 MHz, the damping factor being dependent on the transducer design and on the type of tissue in the transducer's path. As this transmitted damped oscillation reaches an interface between materials having different acoustical impedances, a reflection or echo is produced. By the time this echo has returned to the transducer, the transducer is passive and is then again excited by vibrations produced by this returning echo. This echo signal is then amplified and processed in a logarithmic amplifier and an envelope detector. This process is repeated at an approximate 500 Hz rate. The time between the generation of the transducer exciting pulse and the detection of the received echo represents the time taken for the ultrasonic energy to travel from the transducer, to the interface, and thence from the interface back to the transducer. The velocity of sound waves, and thus of ultrasonic energy, in body tissue is about 0.125 centimeters per microsecond.

transducer frequencies

operational theory

sound velocity

It is rarely necessary to know this velocity accurately as relative interface distances are more important than actual interface distances. If these actual distances are required, the system can be calibrated against a known tissue/interface distance.

depth
versus
resolution

Depth and resolution of the system depends on the crystal's resonant frequency, a 1 MHz crystal provides low resolution, however, reflections can be detected for tissue 50 centimeters from the transducer. The commonly used 2.5 MHz crystal can be used to at least 20 centimeters; whereas a 10 MHz crystal provides excellent resolution, however it cannot be used above about 5 centimeters. Relating distance to velocity, a tissue interface 5 centimeters from the transducer will produce a reflection 80 microseconds after the impulse.

transducer/
tissue
coupling

It is important that the transducer be firmly held against the tissue (skin) as any air in the transducer's path will severely attenuate the ultrasonic energy. A satisfactory transducer/tissue coupling can be assured by using a liquid coupler such as water. In certain ultrasonic scanning systems it is impossible to locate the transducer against the tissue and a substantial depth of water is used as a coupling medium.

14.2 "A" SCAN ULTRASONOGRAPHY

Y-T mode

"A" scan ultrasonography displays the amplified echo signal on the vertical channel of an oscilloscope with the horizontal channel being deflected by a conventional sweep generator. This sweep generator is triggered from the impulse signal and the time delay between the beginning of the sweep and the echo appearing on the CRT screen is proportional to tissue depth. The sweep generator may be a conventional sweep generator or may be specifically calibrated in tissue depth (centimeters)

calibrated
sweep
generator

rather than in time per division. A sweep speed of 100 microseconds per division would correspond to approximately 6.25 centimeters of tissue per division. Faster sweep speeds would give correspondingly greater resolution. The fastest usable sweep speed would probably be 2 microseconds per division which would correspond to 1.25 millimeters of tissue per division. This faster sweep speed would only be usable with 10 MHz transducers.

echoencepha-
lography

The most common usage of "A" scan ultrasonography is in echoencephalography. Echoencephalography detects brain midline position and possible displacement of this midline due to an abnormal space occupying mass within one side of the skull, such as a tumor. When performing an "A" scan with the transducer held against the side of the subject's head, echos are received from the two halves of the brain as shown in Fig. 14-2. The echo produced by the midline is termed the "M" echo and should be symmetrically placed between the echos received from each side of the subject's skull.

A TYPICAL INSTRUMENT CONFIGURATION
FOR "A" SCAN ULTRASONIC DIAGNOSIS
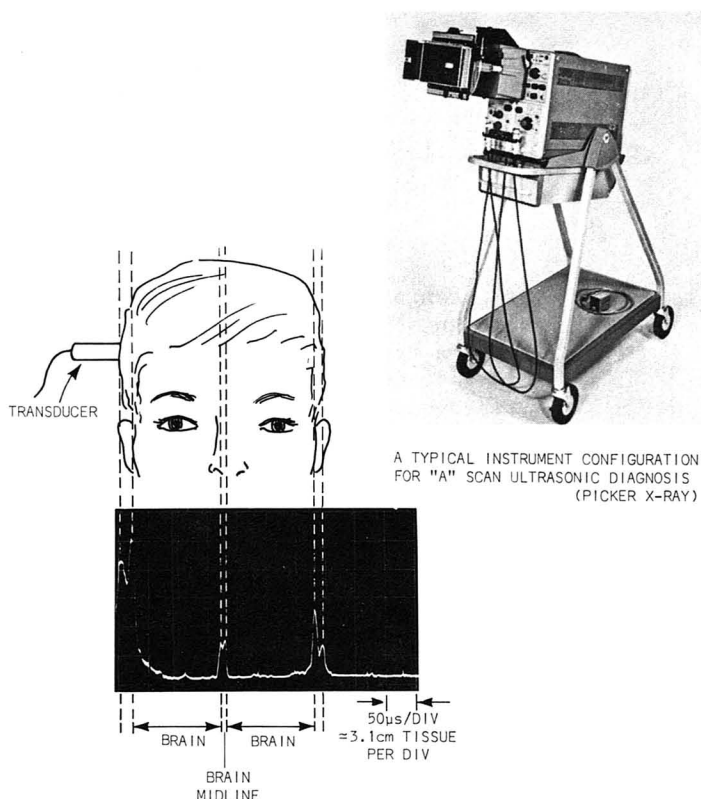(PICKER X-RAY)

Fig. 14-2. "A" scan echoencephalography.
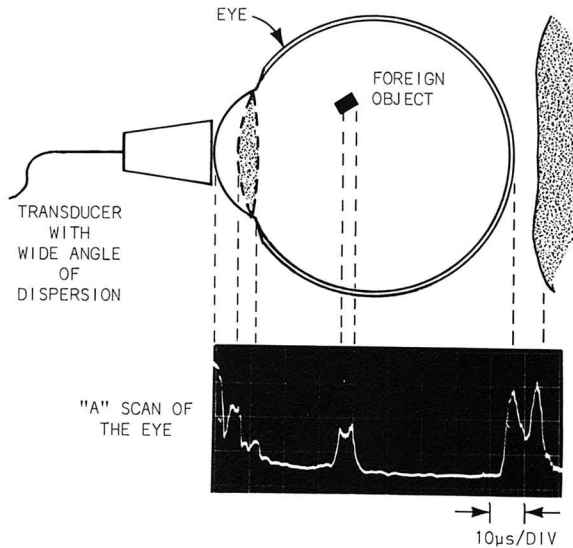
Fig. 14-3. "A" scan detection of a foreign object.

foreign
object
detection

"A" scan ultrasonography may also be used to detect foreign objects imbedded in a subject. Fig. 14-3 shows a typical "A" scan obtained when scanning an eye containing a foreign object with a 10 MHz transducer. The response attributed to the foreign object can be verified by scanning the other eye and noting the lack of any response at this time position.

Many manufacturers produce equipment suitable for "A" scan ultrasonography. A typical instrument configuration is shown in Fig. 14-2; a Tektronix Type 561B or 564B Oscilloscope is used with a Tektronix Type 2B67 time-base unit and a special vertical plug-in unit, designed for ultrasonography, produced by the Picker X-Ray Corporation. Suitable probes are also produced by the Picker X-Ray Corporation. The complete instrument may be used in conjunction with a Tektronix Scope-Mobile® Cart and Tektronix Trace-Recording Camera.

14.3 "TIME-MOTION" MODE ULTRASONOGRAPHY

Considering "A" scan ultrasonography as discussed previously; if an interface producing an echo was moving in relation to the transducer, the response obtained on the CRT would also move horizontally relative to the beginning of the sweep. If this "A" scan system was to be modified slightly by removing the echo signal from the vertical channel (Y) and connecting it to the intensifying channel (Z), an echo response would then appear as an intensified spot on the screen. Any "motion" in this echo response would appear as a horizontal motion of this spot between successive sweeps of the horizontal sweep generator. No vertical signal would be involved.

Z-T mode

In time-motion or "T-M" mode ultrasonography, the above hypothetically modified "A" scan system is used with a slow sweep generator moving the display vertically. Thus, any motion in the echo response is displayed in real time by the slow vertical sweep. This system is shown diagrammatically in Fig. 14-1, and a typical "T-M" mode ultrasonogram from motion of the mitral valve of the heart is shown in Fig. 14-4.
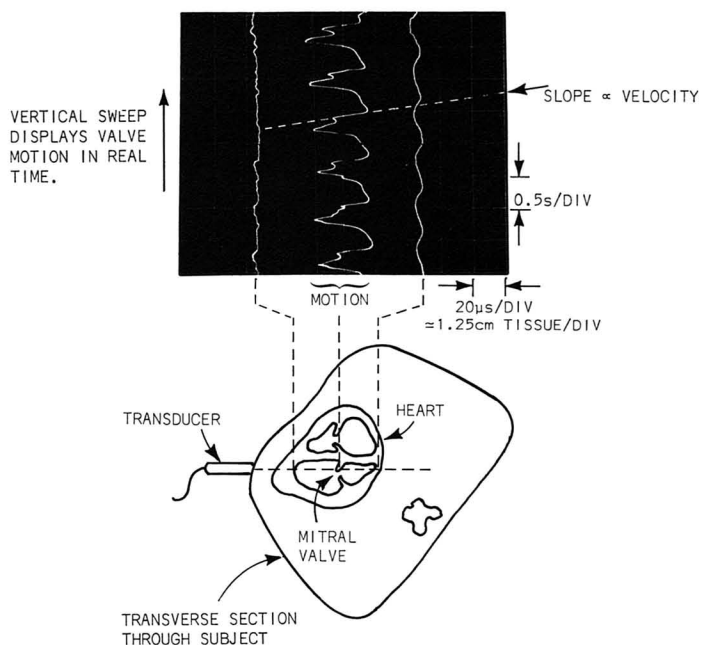
Fig. 14-4. "T-M" mode cardiography.

208

TRANSDUCER SCANNED ACROSS
SUBJECT. 6 INSTANTANEOUS
POSITIONS SHOWN.

SECTION
THROUGH SUBJECT

INTERNAL
ORGAN

X-Y SCAN INTENSIFIED BY TRANSDUCER
OUTPUT. EACH OF THE 6 INSTANTANEOUS
TRANSDUCER POSITIONS PRODUCES 3
INTENSIFIED AREAS. THESE INTENSIFIED
AREAS ARE RECORDED ON A STORAGE SCOPE
AND LINKAGE OF THE "DOTS" PRODUCES
AN OUTLINE OF THE INTERNAL ORGAN.
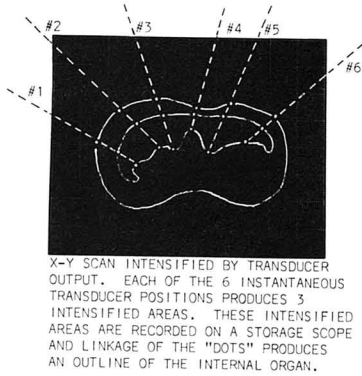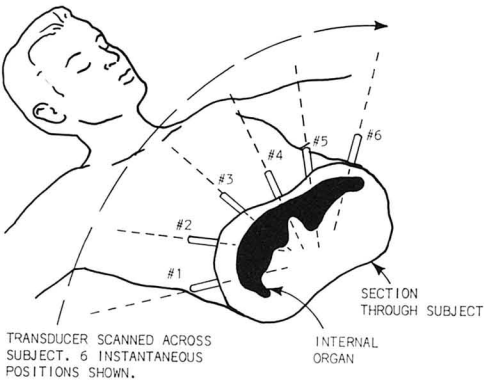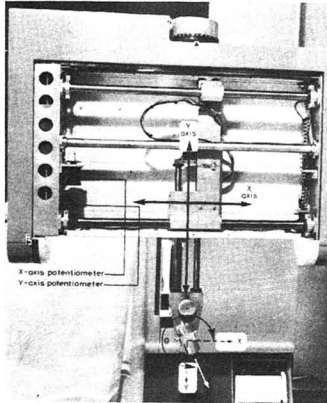
A TYPICAL MANUAL SCANNING SYSTEM PRODUCED
BY THE PICKER X-PAY CORP., LONGMONT, COLO.

TYPICAL XYθ AUTOMATIC SCANNING SYSTEM.
THE POSITION OF THE TRANSDUCER IS PRE-
SENTED AS FOUR VOLTAGES, REPRESENTING
THE DISTANCES ALONG THE X AND Y AXES
AND THE SINE AND COSINE OF THE ANGLE
BETWEEN THE LINE OF THE ULTRASONIC
BEAM AND THE X-AXIS.

Fig. 14-5. Ultrasonic scanning–"B" mode.

Referring to Fig. 14-4, the motion of the mitral valve is clearly shown on the vertically-swept trace on the face of the CRT. Fig. 14-4 also shows the motion of the walls of the heart, however echoes from other interfaces in the transducers path have been gated out in this instance. This particular

cardiography   application, known as "T-M" mode cardiography is particularly valuable in the detection of mitral stenosis. Since the actual movement of the mitral valve is shown in centimeters in real time (vertical sweep), the velocity of movement of this valve can be determined from the slope of the trace obtained. In the example shown in Fig. 14-4, the maximum velocity of the mitral valve is 16 centimeters per second, the overall movement of the valve is 2.2 centimeters and the heart rate is 86 beats per minute.

14.4 ULTRASONIC SCANNING

Ultrasonography using the "B" mode technique is referred to as ultrasonic scanning. In this mode, the echo signal is connected to the Z axis of a

mechanical   monitor to provide intensification as in the "T-M"
X-Y scan   mode, however the X and Y monitor input signals are derived from a mechanical scanning system and provide signals proportional to the position and direction of the probe to form part of a physical picture of the organs being examined. The result obtained is a two-dimensional, cross-sectional presentation of part of the subject.

In its simplest form the "B" mode ultrasonic scanner
manual   is manually scanned by an operator across the part
scanning   of the subject of interest. Referring to Fig. 14-5, with the transducer placed in each of the six positions shown, a series of intensified areas appears on the CRT face. Their position bearing an exact physical relationship to the positions of tissue interface within the subject. With only six transducer positions as shown in Fig. 14-5, only six rows of intensified dots would appear on the CRT, however, in practice, the transducer is slowly scanned over the subject and is producing echo patterns at a 500 Hz rate. The display thus appears continuous rather than composed of several dots. The manual scanning system shown in Fig. 14-5, while not offering the sophistication of an automatic scanning system, is economical and may also be used for both "A" mode scanning and "T-M" mode scanning rather than being restricted to "B" mode use.

automatic
X-Y-θ
scanning

The mechanism associated with an automatic X-Y-θ compound scanning system is also shown in Fig. 14-5. Automatic compound scanning systems can be divided into two distinct classes, in one class the scanning movements of the probe are predetermined and related to the geometry of the machine itself, in the other the scanning pattern is dictated by the contours of the particular part of the individual patient being examined. The first type is referred to as an immersion scanner since a liquid-filled space is required between the probe and the subject and the second type is referred to as a dry scanner since the transducer is in direct contact with the subject. The mechanism associated with immersion scanners is somewhat less complicated than the mechanism required for direct contact dry scanners and the requirement of having a liquid filled space is satisfied by having a plastic membrane separating the subject from the liquid in which he or she is immersed. Immersion scanners are particularly suited to areas such as the breasts and the eyes where direct contact scanning is impractical.
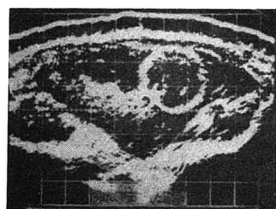
POSITION OF FETUS *IN UTERO*
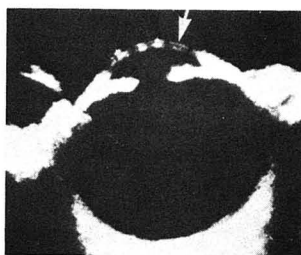
CORNEAL THICKNESS DETERMINATION
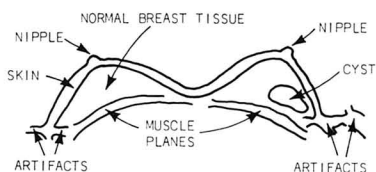
DETECTION OF CYST IN BREAST TISSUE

Fig. 14-6. Typical "B" mode displays from a compound scanning system.

ultra-
sonographic
applications

Fig. 14-6 shows typical "B" mode displays obtained
with a compound scanning system. These displays are
self-explanatory and indicate some of the potential
advantages of ultrasonography. Neither of the three
displays shown in Fig. 14-6 could have been obtained
by conventional X-ray techniques as; for the fetus
in utero, fetal development is impaired by X-ray
energy; for the breast cyst, such cyst tissue is
undetectable by conventional X-ray techniques; and,
for the scan of the eye, resolution with X-ray
techniques cannot be obtained if the X-ray plate
cannot be located directly behind the region
concerned.

14.5 DOPPLER ULTRASOUND

movement
causes
frequency
shift

Continuous ultrasonic energy, rather than bursts of
ultrasonic energy as used in ultrasonography, is
used to detect motion within a subject by the Doppler
principle. When ultrasonic energy is reflected from
a moving object it is shifted slightly in frequency,
the frequency shift being proportional to the speed
of the object. In the living body there are
numerous movements which reflect ultrasonic energy:
blood flowing through arteries, the action of the
heart, intestinal movements and passage of urine and
gastric juices. The most common application of
Doppler ultrasonics is in obstetrics to detect
movement of the fetal heart and fetal blood flow;
such fetal activity can be detected as early as the
tenth week of gestation. Another major application
for Doppler ultrasonics is in the detection of blood
flow in the peripheral circulation of the body.

SECTION III

INSTRUMENTATION

The following chapters (]5-]8) describe the
characteristics of the instrumentation required to
implement the measurement techniques covered in
Section II as well as the characteristics of
instrumentation in current use for biophysical
measurements. A working knowledge of electronic
instrumentation principles is assumed; this material
primarily discusses the instrumentation
characteristics that are unique to the biophysical
sciences.

Since the following material is generally limited to
a discussion of those instrumentation characteristics
that are unique to biophysical measurement, it should
not be regarded as a general reference source on any
one type of instrumentation. An attempt is made in
each of the following chapters to direct a reader
who is somewhat unfamiliar with general electronic
instrumentation to a general reference source. This
general reference source should first be studied by
such readers before preceding with a study of the
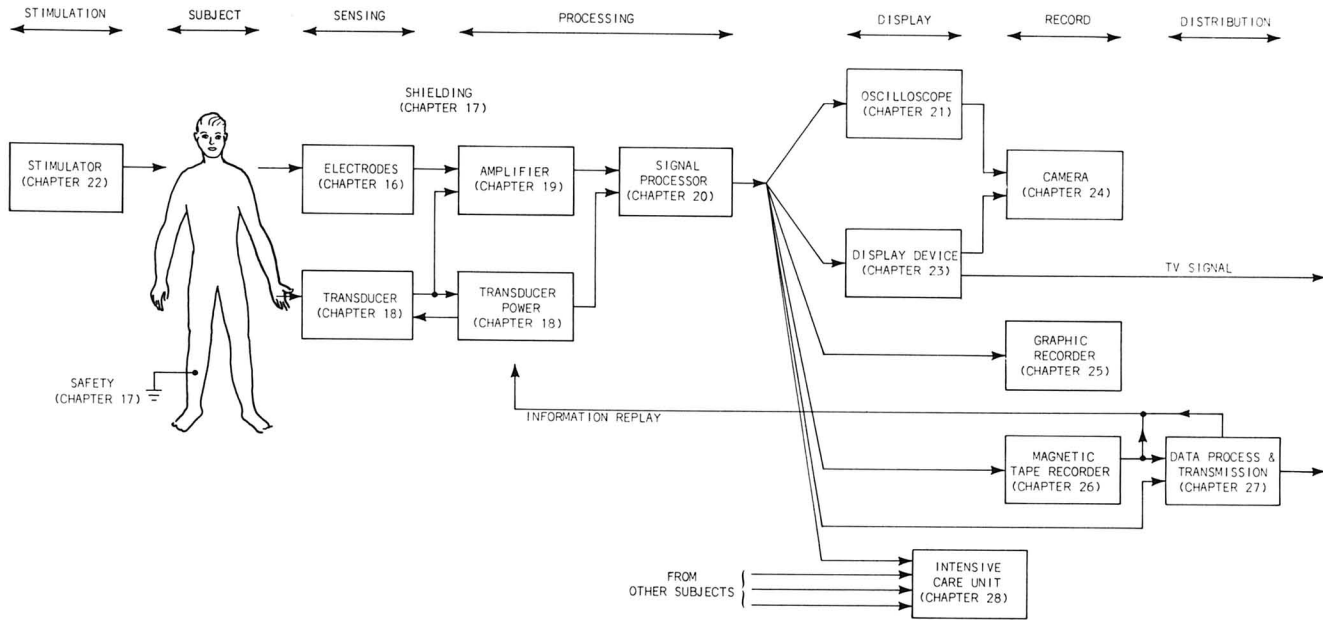material presented in the following chapters.

Fig. 15-1. Biomedical instrumentation configuration
(Showing references to appropriate chapters in this book).

15

INSTRUMENTATION SUMMARY

A block diagram of the instrumentation used for biomedical measurements is shown in Fig. 15-1. While the majority of measurement systems do not utilize all of the components shown in this diagram, the ultimate measurement system may contain all components. The following 13 chapters (Chapters 16-28) describe each of these instrumentation groups in more detail.

biomedical signal source

The biomedical signal to be measured is derived from the subject. This signal may be continuously produced by the subject (such as the ECG), it may be voluntarily initiated by the subject (such as a muscular contraction) or it may be produced by artificial stimulation of the subject (such as an evoked response). This artificial stimulation is normally produced by an electric current; however, other forms of stimulators, such as mechanical or optical stimulators, are occasionally used. Stimulators are covered in more detail in Chapter 22.

electrodes and transducers

The biomedical signal is either directly available from the subject as an electrical potential detected by using electrodes or it is available in some other form, in which case an electrical potential corresponding to the biomedical signal is produced with the aid of a transducer. Biomedical measurements such as body impedance are regarded as transducer-type measurements (although electrodes are used to perform this measurement) as the measurement involves the detection of an impedance rather than the detection of a bioelectric potential. Electrode systems are covered in more detail in Chapter 16 and transducer systems are covered in more detail in Chapter 18. Electrode shielding and subject safety is covered in Chapter 17.

amplifying
and
processing

Whether the biomedical signal is obtained directly via electrodes or produced via transducers, it invariably requires amplification and perhaps signal processing. Amplifiers are discussed in Chapter 19 and signal processes are discussed in Chapter 20. Many transducer systems incorporate the necessary amplification into the system as discussed in Chapter 18.

displaying

Once the biomedical signal has been amplified and processed, then biomedical measurement techniques can be forgotten as electronic measurement and display techniques are now employed. The amplified and processed signal may be displayed on a conventional oscilloscope (Chapter 21), on some other form of display device such as a television receiver via a scan converter (Chapter 23), on a paper chart recorder (Chapter 25) or it may be recorded on magnetic tape (Chapter 26). The signal may be directly routed to further data processing and data transmission equipment (Chapter 27), or it may be combined with signals from other patients in an intensive care unit (Chapter 28).

permanent
record

If the signal is recorded on magnetic tape or on a chart recorder, then a permanent record of the signal is obtained. When monitoring the signal on a conventional oscilloscope or a display device, a permanent record must be obtained by photographic methods as covered in Chapter 24. Use of a magnetic tape recorder or data processing and transmission equipment allows information replay through the instrumentation system for a detailed analysis of biomedical signals and, in many cases, frees the operator to concentrate on recording the biomedical information.

15.1 BIOMEDICAL SIGNALS

Fig. 15-2 lists the biomedical measurements covered in Section II of the book. This summary is necessarily very brief and references to the appropriate sections in Section II are given to direct the reader to more detailed information. Fig. 15-3, produced by courtesy of Beckman Instruments, Inc., is a more complete summary of biophysical measurements and gives the ranges and characteristics of the various biomedical signals. (Note: Not all of the biomedical measurements covered in Fig. 15-3 are included in this book.)

PHYSIOLOGICAL PARAMETER	REFERENCE (THIS BOOK)	MEASUREMENT REQUIRED	SENSING DEVICES USED	PARAMETER TYPICAL P-P VALUE	PARAMETER FREQUENCY CHARACTERISTICS	
					FUNDAMENTAL Hz	SPECTRUM F(ω) Hz
CIRCULATORY SYSTEM						
HEART POTENTIALS	CHAPTERS 2 AND 5	ELECTROCARDIOGRAM	SURFACE ELECTRODES	2mV	1.3	.05 – 80
	CHAPTER 17		HEART ELECTRODES	50mV	1.3	.05 – 80
	CHAPTER 6	VECTORCARDIOGRAM	SURFACE ELECTRODES	2mV	1.3	.05 – 80
	CHAPTER 7	FETAL ELECTROCARDIOGRAM	SURFACE ELECTRODES (MOTHER)	10μV	2.5	2 – 100
BLOOD PRESSURE	SECT 8.1	DIRECT ARTERIAL PRESSURE AT BRACHIAL ARTERY OR FEMORAL ARTERY	PRESSURE TRANSDUCER	120mmHg	1.3	DC – 20
			MERCURY MANOMETER	120mmHg	"	"
	SECT 8.1	DIRECT VENOUS PRESSURE	PRESSURE TRANSDUCER	9mmHg	1.3	DC – 20
			WATER MANOMETER	12cmH₂O	1.3	–
	SECT 8.2	INDIRECT ARTERIAL PRESSURE	SPHYGMOMANOMETER WITH KOROTKOFF MICROPHONE	120/80mmHg 150mV	– 1.3	– 30 – 500
	SECT 8.3	RELATIVE ARTERIAL PRESSURE	PLETHYSMOGRAPH	RELATIVE	1.3	.05 – 10
			IMPEDANCE PLETHYSMOGRAPH	0.1% CHANGE	1.3	.05 – 10
BLOOD FLOW (AND BLOOD VELOCITY)	SECT 8.4	PERIPHERAL FLOW	ELECTROMAGNETIC FLOWMETER	1000cc/min	1.3	DC – 50
			ISOTHERMAL FLOWMETER	"	"	"
			ELECTROTURBINOMETER	"	"	"
			ULTRASONIC FLOWMETER	"	"	"
	SECT 8.5	CARDIAC OUTPUT	FLOWMETER AT AORTA	5000cc/min	1.3	DC – 50
			DYE DILUTION	5000cc/min	–	–
BLOOD VOLUME	SECT 8.6	BLOOD VOLUME	MODIFIED DYE DILUTION	5500cc	–	–
MITRAL VALVE FUNCTION	SECT 14.3	ULTRASONIC ECHO	T–M MODE ULTRASONICS	16cm/s	1.3	DC – 50
RESPIRATORY SYSTEM	CHAPTER 9					
BREATHING	SECT 9.2	PNEUMOGRAM	THERMISTOR PNEUMOGRAPH	500cc/BREATH	0.25	.05 – 2
			IMPEDANCE PNEUMOGRAPH	"	"	DC – 2
			ELASTIC FORCE GAGE	"	"	"
RESPIRATORY FLOW	SECT 9.3	PNEUMOTACHOGRAM	PNEUMOTACHOGRAPH WITH PRESSURE TRANSDUCER	20,000cc/min	"	"
RESPIRATORY VOLUME	SECT 9.4	SPIROGRAM	SPIROMETER	4000cc	"	DC – .5

BRAIN FUNCTIONS						
ELECTRICAL ACTIVITY	CHAPTER 10	ELECTROENCEPHALOGRAM	SCALP ELECTRODES	50μV	10	.5 – 100
			INTRACRANIAL ELECTRODES	500μV	"	"
EVOKED RESPONSES	CHAPTER 11	INTRACELLULAR POTENTIALS	MICROELECTRODES	100mV		1 – 10,000
		EXTRACELLULAR POTENTIALS	NEEDLE ELECTRODES	50μV		1 – 1000
EYE RESPONSES	SECT 11.8	ELECTRORETINOGRAM	CONTACT LENS ELECTRODE	100μV		.05 – 20
BRAIN MIDLINE POSITION	SECT 14.2	ULTRASONIC ECHO	"A" SCAN ULTRASONICS			
MUSCULAR FUNCTIONS	CHAPTER 12					
MUSCLE EXCITABILITY	SECT 12.3	S–D CURVE	STIMULATE WITH SURFACE ELECTRODES			
MUSCLE STRENGTH	SECT 12.4	MYOGRAM	NEEDLE OR SURFACE ELECTRODES	300μSTRAIN	.5	DC – 50
MUSCLE POTENTIALS	SECT 12.6	ELECTROMYOGRAM	NEEDLE OR SURFACE ELECTRODES	1mV		10 – 5000
	SECT 12.7	ELECTROMYOGRAM WITH STIMULATION	NEEDLE OR SURFACE ELECTRODES STIMULATE WITH SURFACE ELECTRODES	"	"	"
NERVE CONDUCTION	SECT 12.8	H REFLEX RESPONSE	AS ELECTROMYOGRAPH WITH REDUCED STIMULATION			
	SECT 12.9	CONDUCTION VELOCITY	AS ELECTROMYOGRAPH			
SMOOTH MUSCLE ACTIVITY	SECT 12.11	ELECTROGASTROGRAM	SURFACE ELECTRODES	20mV	.25	.05 – 2
AUTONOMIC NERVOUS SYSTEM						
SWEAT GLAND ACTIVITY	CHAPTER 13	GALVANIC SKIN REFLEX	LEAD SURFACE ELECTRODES	50kΩ		
		ELECTRICAL SKIN RESISTANCE	" " "	"		
BODY TEMPERATURE	SECT 9.5	TEMPERATURE	THERMISTOR PROBE THERMOMETER	98°F		
ANATOMY						
INTERNAL ORGAN POSITION	CHAPTER 14	ULTRASONIC ECHO	ULTRASONIC SCANNING SYSTEM			

Fig. 15-2. Biophysical measurements covered in this book.

	PRIMARY SIGNAL RANGES AND CHARACTERISTICS
CARDIOVASCULAR SYSTEM	
BLOOD PRESSURE, DIRECT METHOD	FREQUENCY RANGE: DC TO 200Hz; DC TO 60Hz USUALLY ADEQUATE. PRESSURE RANGE, ARTERIAL: 40 TO 300mmHg; VENOUS 0 TO 15mmHg.
BLOOD PRESSURE, INDIRECT METHOD, INTERMITTENT SYSTOLIC AND DIASTOLIC	AUSCULTATORY CRITERION (KOROTKOFF SOUNDS): 30 TO 150Hz USUALLY ADEQUATE. PALPATORY CRITERION: 0.1 TO 60Hz. BOTH REQUIRE ADDITIONAL SIGNAL SHOWING OCCLUDING PRESSURE.
PULSE WAVES, DIRECT METHOD, ARTERIAL	(SEE BLOOD PRESSURE, DIRECT)
PULSE WAVES, INDIRECT METHOD, PERIPHERAL ARTERY	FREQUENCY RANGE: 0.1 TO 60Hz USUALLY ADEQUATE. PULSE TRACE SIMILAR TO BLOOD PRESSURE, DIRECT, BUT WITHOUT BASELINE ZERO.
PHONOCARDIOGRAM	FREQUENCY RANGE: 5 TO 2000Hz; MAJOR DIAGNOSTIC COMPONENTS LIE IN 20 TO 200Hz RANGE.
PLETHYSMOGRAM (VOLUME MEASUREMENTS)	FREQUENCY RANGE: DC TO 30Hz.
BALLISTOCARDIOGRAM	FREQUENCY RANGE: DC TO 40Hz.
HEART RATE	AVERAGE RATE, HUMAN: 45 TO 200 BEATS/min; LAB ANIMAL; 50 TO 600 BEATS/min.
OXIMETRY	FREQUENCY RANGE, 0 TO 60Hz; 0 TO 5Hz USUALLY ADEQUATE.
CARDIAC OUTPUT	FREQUENCY RANGE, 0 TO 60Hz; 0 TO 5Hz USUALLY ADEQUATE.
BLOOD FLOW	FLOW RANGE, HUMAN: 5cc/min TO 10 LITERS/min. FREQUENCY RANGE: 0 TO 60Hz; 0 TO 20 USUALLY ADEQUATE.
HEART SHUNT DETECTION	ELECTRODE SIGNAL RANGE, Po_2: 0 TO 160mmHg. HYDROGEN AND SODIUM ASCORBATE: QUALITATIVE.
ELECTROCARDIOGRAM	FREQUENCY RANGE: 0.05 TO 80Hz. SIGNAL RANGE: 10μV TO 5mV INCLUDES FETAL RANGE.
RESPIRATORY SYSTEM	
FLOW RATE (PNEUMOTACHOGRAM)	FREQUENCY COMPONENTS TO 40Hz. NORMAL FLOW RANGE: 250 TO 500ml/s; MAXIMUM 8 LITERS/s.
BREATHING RATE CALCULATED FROM RECORD (WITH APPROXIMATE RELATIVE RESPIRATORY VOLUME)	AVERAGE RATE: HUMAN, 12 TO 20 BREATHS/min; LAB ANIMAL, 8 TO 60 BREATHS/min.
TIDAL VOLUME (MEASURED PER BREATH OR INTEGRATED TO PROVIDE VOLUME/min)	TYPICAL VOLUME, ADULT HUMAN: 600ml/BREATH; 6 TO 8 LITERS/min.
CO_2, N_2O, OR HALOTHANE CONCENTRATION IN RESPIRED AIR	NORMAL RANGE, CO_2: 0 TO 10%; END-TIDAL CO_2, HUMAN: 4 TO 6%. N_2O: 0 TO 100%. HALOTHANE: 0 TO 3%.
DIFFUSION OF INSPIRED GAS (USING NITROGEN)	NORMAL RANGE OF NITROGEN CONCENTRATION DIFFERENTIAL: 0 TO 10%.
PULMONARY DIFFUSING CAPACITY (USING CARBON MONOXIDE)	NORMAL RANGE, HUMAN: 16 TO 35mlCO/mmHg/min.

DISSOLVED GASES AND pH	
PARTIAL PRESSURE OF DISSOLVED O_2, IN VIVO OR IN VITRO	FREQUENCY RANGE: DC TO 1Hz USUALLY ADEQUATE. NORMAL MEASUREMENT RANGE: 0 TO 800mmHg Po_2. HYPERBARIC Po_2 RANGE: 800 TO 3000.
pH, IN VITRO	SIGNAL RANGE: 0 TO ±700mV COVERS pH RANGE.
PARTIAL PRESSURE OF DISSOLVED CO_2, IN VITRO	NORMAL SIGNAL RANGE: 0 TO ±150mV COVERS RANGE FROM 1 TO 1000mmHg Pco_2.
BIOELECTRIC POTENTIALS	
ELECTROENCEPHALOGRAM	FREQUENCY RANGE: DC TO 100Hz; MAJOR DIAGNOSTIC COMPONENTS LIE IN 0.5 TO 60Hz RANGE. NORMAL SIGNAL RANGE: 15 TO 100μV
CEREBRAL POTENTIALS, INTRACRANIALLY RECORDED	NORMAL SIGNAL RANGE: 10μV TO 100mV. PULSE DURATION: 0.6ms TO 20ms.
ELECTROMYOGRAM (PRIMARY SIGNAL)	FREQUENCY RANGE: 10 TO 2000Hz. PULSE DURATION: 0.6ms TO 20ms.
ELECTROMYOGRAM (AVERAGED)	AN AVERAGE OF THE PRIMARY SIGNAL, AFTER FULL WAVE RECTIFICATION.
SMOOTH MUSCLE POTENTIAL (e.g., ELECTROGASTROGRAM)	FREQUENCY RANGE: 0 TO 0.6Hz. NORMAL SIGNAL RANGE: 0.5 TO 80mV.
ELECTRORETINOGRAM	FREQUENCY RANGE: DC TO 20Hz ADEQUATE. NORMAL SIGNAL STRENGTH: ½μV TO 1mV.
ELECTROCARDIOGRAM	(SEE LISTING UNDER CARDIOVASCULAR SYSTEM)
ELECTRONYSTAGMOGRAM	DIRECT: FREQUENCY RANGE, 0 TO 20Hz. TYPICAL SIGNAL STRENGTH, 100μV/10° EYE MOVEMENT. DERIVATIVE OR VELOCITY: FREQUENCY RANGE, 0 TO 20Hz. SIGNAL DERIVED FROM DIRECT READING.
PHYSICAL QUANTITIES	
TEMPERATURE	FULL RANGE OF SIGNALS.
VOICE	FREQUENCY RANGE: 20 TO 20,000Hz.
SKIN RESISTANCE (GSR)	RESISTANCE RANGE: 1k TO 500k.
ISOMETRIC FORCE, DIMENSIONAL CHANGE, BODY FLUID AND BODY CAVITY PRESSURES	FULL RANGE OF SIGNALS.

Fig. 15-3. Biophysical signal ranges

16

ELECTRODES

tissue-
electrode
interface

ionic-
electronic
currents

The electronics engineer, accustomed to measuring potentials between various points on an electronic circuit, may tend to regard two physiological electrodes simply as two probes applied to a subject in order to measure a potential difference. Such an oversimplification cannot be made as the conduction of current in tissue, as in any other liquid system, is ionic; that is to say, by the migration of positive and negative ions from point to point. To measure electrical effects in tissue it is necessary to make a transfer from this ionic conduction to the electronic conduction which occurs in the measuring circuit. This transfer is accomplished at the tissue-electrode interface.

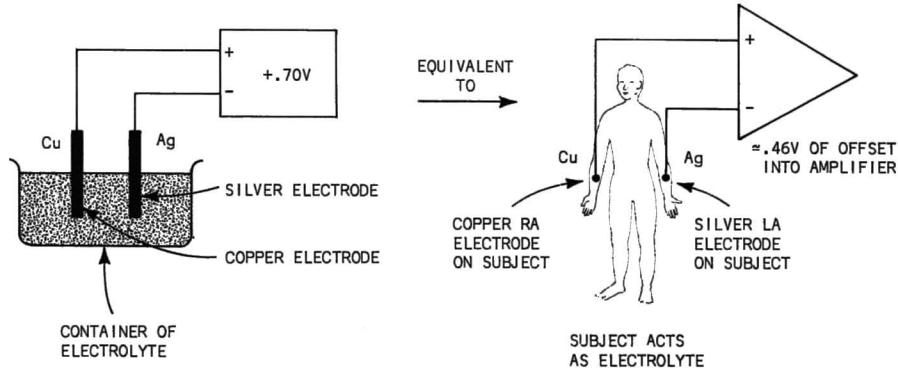## 16.1 ELECTRODE OFFSET POTENTIAL

If two dissimilar metals are inserted into a container of electrolyte, a potential difference can be measured between these metals. If one of these metals were silver and the other copper, this potential would be referred to as the silver/copper cell potential and would be in the order of 0.4 volts. This potential is comprised of two separate components added algebraically together: a "half cell" potential due to the silver electrode and a "half cell" potential due to the copper electrode. These half cell potentials, although difficult to measure, are the potentials produced across the metal-electrolyte interfaces. To a first order approximation, neglecting many chemical factors that would be important if the electrodes were used in a chemical measuring application rather than a biophysical measuring application, the half cell potential is approximately equal to the electrode potential of the metal concerned. Thus, the

METAL	IONIC SYMBOL	ELECTRODE POTENTIAL (EP)*
ALUMINIUM	Al^{+++}	-1.66VOLTS
ZINC	Zn^{++}	- .76
IRON	Fe^{++}	- .44
LEAD	Pb^{++}	- .12
HYDROGEN	H^{+}	0
COPPER	Cu^{++}	+ .34
SILVER	Ag^{+}	+ .80
PLATINUM	Pt^{+}	+ .86
GOLD	Au^{+}	+1.50

CORRODED END (ANODIC OR LEAST NOBLE)

PROTECTED END (CATHODIC OR MOST NOBLE)

*APPROXIMATELY EQUAL TO HALF-CELL POTENTIAL IN PHYSIOLOGICAL ENVIRONMENTS.

Fig. 16-1. Electrode potentials of metals, the electrochemical series.

+.70V

EQUIVALENT TO

Cu   Ag

SILVER ELECTRODE

COPPER ELECTRODE

CONTAINER OF ELECTROLYTE

Cu   Ag

≈.46V OF OFFSET INTO AMPLIFIER

COPPER RA ELECTRODE ON SUBJECT

SILVER LA ELECTRODE ON SUBJECT

SUBJECT ACTS AS ELECTROLYTE

OFFSET POTENTIAL PRODUCED BY DISSIMILAR METALS APPROXIMATELY EQUALS DIFFERENCE IN ELECTRODE POTENTIALS.

OFFSET ≈ EP Ag - EP Cu
≈ 0.80V - 0.34V
≈ 0.46V

Fig. 16-2. Electrode offset potential.

potential produced between the silver electrode and the copper electrode will be approximately equal to the difference in the electrode potentials of silver and copper, that is, approximately equal to 0.80 minus 0.34 or 0.46 volts. The exact value of this potential will depend on many chemical factors, the more important of which are the electrolyte used and the concentration of this electrolyte; however the value given is probably within about 100 millivolts of the actual value that may be measured in a physiological environment. A list of the electrode potentials for the more common metals is shown in Fig. 16-1. This listing is referred to as the electrochemical series. The example given of a silver and copper electrode is illustrated in Fig. 16-2, which shows the similarity between electrodes in a container of electrolyte and electrodes on a subject. In electrophysiology the difference in half cell potentials that can be detected between two electrodes is referred to as "offset potential."

electrode offset potential

It is important not to confuse the electrode offset potential produced between two electrodes connected to a subject and the DC-differential offset potential available in many differential amplifiers. The DC-differential-offset-potential capability in differential amplifiers is intended to be used to cancel the electrode offset potential produced between electrodes.

differential amp offset potential

In the biomedical situation, the electrode offset potential produced between electrodes may be unstable and unpredictable. Thus, it is desirable that this potential be as low as possible. In order to reduce electrode offset potential it is first necessary to understand its origin. At any electrode/electrolyte interface there is a tendency for the electrode to discharge ions into solution and for ions in the electrolyte to combine with the electrode. These chemical reactions may be represented as follows:
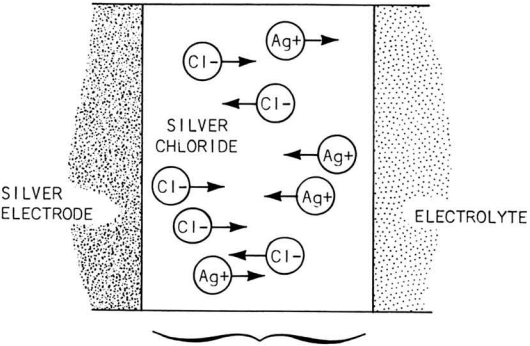
$$\text{Metal} \longrightarrow \text{electrons} + \text{metal ions}$$
(Oxidization reaction)

$$\text{Electrons} + \text{metal ions} \longrightarrow \text{metal}$$
(Reduction reaction)

222

ELECTRODE DOUBLE
LAYER FORMS
(COMPLEX)
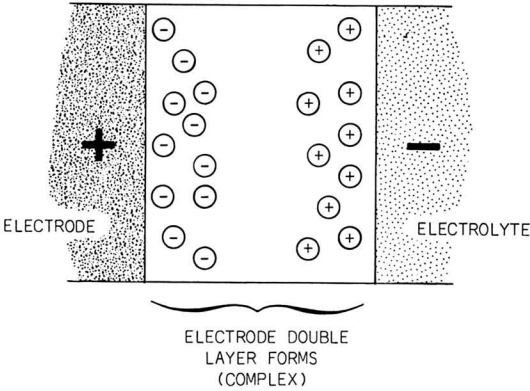
POTENTIAL EXISTS BETWEEN THE ELECTRODE AND
ELECTROLYTE DUE TO THE FORMATION OF THE
ELECTRODE DOUBLE LAYER.

(A) METALLIC ELECTRODE

SILVER CHLORIDE FORMS FREE SILVER IONS (Ag+)
AND CHLORIDE IONS (Cl-) WHICH PREVENT THE
FORMATION OF THE ELECTRODE DOUBLE LAYER.

(B) SILVER/SILVER CHLORIDE ELECTRODE

Fig. 16-3. Electrode/electrolyte interface.

The net result of these reactions is the creation of a charge gradient, the spatial arrangement of which is called the electrode double layer. This double layer may be represented, as shown in Fig. 16-3A, in its simplest form as two parallel areas of charge of opposite signs. Electrodes in which no net transfer of charge occurs across the metal-electrolyte interface are referred to as perfectly polarized or perfectly nonreversible electrodes, that is, only one of the two chemical reactions shown above can occur. Electrodes in which unhindered transfer of charge is possible are referred to as perfectly nonpolarizable or perfectly reversible electrodes, that is, both of the equations referred to above occur with equal ease. Practical electrode systems have properties that lie between these ideal limits. The electrode double layer existing between a practical electrode and electrolyte may be regarded as a battery in parallel with a reasonably high impedance as shown in Fig. 16-4. This impedance may typically be 10,000 ohms and the battery may be equivalent to a 10 μF capacitor. The potential of the battery, or the charge on the capacitor, will be the half cell potential of the electrode.
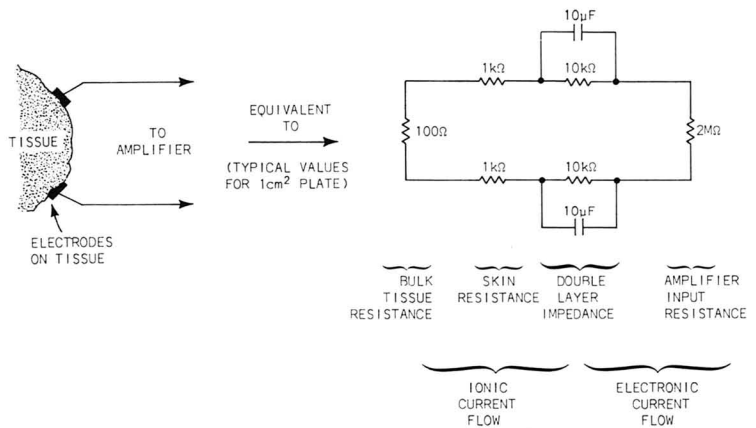
nonpolar-
izable
versus
polarizable

reversible
versus non-
reversible

partially
reversible
electrodes

Fig. 16-4. Electrode equivalent circuit for partially reversible electrodes.

The silver/silver-chloride electrode shown in
Fig. 16-3B consists of either a solid silver surface
coated with a thin layer of solid silver chloride
or, as is the case with Tektronix silver/silver-
chloride electrodes, consist of silver powder and
silver chloride powder compressed into a solid
pellet. The presence of the silver chloride allows
the electrode to behave as a near perfect
nonpolarizable or reversible electrode as it
prohibits the formation of the electrode double
layer, the silver chloride dissociating to silver
ions and chloride ions which are free to migrate
between the electrode and the electrolyte, thus
opposing the formation of the double layer. The net
result is a low impedance, low offset potential,
interface between the silver and the electrolyte.
Both silver/silver-chloride and zinc/zinc-sulfate
electrodes exhibited this characteristic.

Most electrodes form soluble metallic salts and thus
are highly toxic and can be used only as surface
electrodes on intact skin. Silver electrodes are,
however, nontoxic, as silver chloride is almost
insoluble in a chloride-containing solution, thus
very few free silver ions exist and tissue damage
from them is negligible. Thus, silver or silver/
silver-chloride electrodes are definitely preferred
for use on exposed tissue. Although, as stated
previously, zinc/zinc-sulfate electrodes produce low
offset potential characteristics similar to silver/
silver-chloride electrodes, they are highly toxic to
exposed tissue due to the passage into the tissue of
free zinc and/or sulfide ions.

A layer of silver chloride may be deposited on
silver electrodes, converting them to silver/silver-
chloride electrodes, by a process known as
chloriding. This is achieved by making the silver
electrodes positive to a solution containing sodium
chloride or saline (0.9% concentration) and passing
a current through the electrode at the rate of
1 mA/cm^2 of surface for several minutes. The silver
electrode should be cleaned to remove surface
contaminents before chloriding.

Mechanical disturbance of the electrode double layer causes electrode noise because the double layer acts as a region of charge gradient, disturbance of which gives rise to a change in capacitance and thus a change in voltage. The electrical stability of an electrode is considerably enhanced by mechanical stabilization of the electrode-electrolyte interface. This is achieved by the use of indirect-contact jelly floating electrodes which interposes an electrolyte (paste) "jelly" or "paste" between the electrode and the contact tissue.

The preceding discussion of electrode offset potential has been highly simplified, as in electrophysiology it is unnecessary to know the exact value of the electrode offset potential produced and it is therefore unnecessary for users of electrodes to understand all of the chemistry involved. Interested readers wishing to more fully understand the chemical process involved should refer to Geddes and Baker, *Principles of Applied Biomedical Instrumentation*, Wiley, 1968, Sections 9.2, 9.3, 11.1, 11.2 and 11.3 and also Dewhurst, *Physical Instrumentation in Medicine and Biology*, Pergamon, 1966, Section 21.2 (specifically Chapters 21, 22, 26 and 28).

16.2 ELECTRODE OFFSET POTENTIAL CHARACTERISTICS

The previous section discusses the electrode offset potential produced at an electrode-electrolyte interface and states that this potential is unstable and unpredictable. When electrodes are connected to a subject in order to record a bioelectric event on a DC-coupled oscilloscope, long-term changes in drift and electrode offset potential appear as baseline drift noise and short-term changes appear as noise on the trace.

If a pair of new electrodes are applied to a subject, an electrode offset potential will exist as the electrodes can essentially be regarded as two batteries. When these electrodes are then connected to a DC-coupled oscilloscope amplifier, the input impedance of this amplifier will act as a load on the batteries and tend to discharge them. Since these electrodes form a very poor battery, a typical Tektronix oscilloscope differential-input impedance of two megohms will discharge the batteries over a period of several hours. It is recommended that new electrode pairs be shorted together in 0.9% saline for a few hours before use in an effort to accelerate **discharging** this process. As these batteries are discharged **offset** their effective potential is reduced, this decreasing potential appearing as electrode offset potential drift as shown in Fig. 16-5. Although the source is represented as a battery in preceding discussions, it may equally well be represented as a capacitor with the amplifier differential-input impedance tending to discharge the capacitor as depicted in the equivalent circuit shown in Fig. 16-4. Either analogy is only a first order of approximation to the true situation.
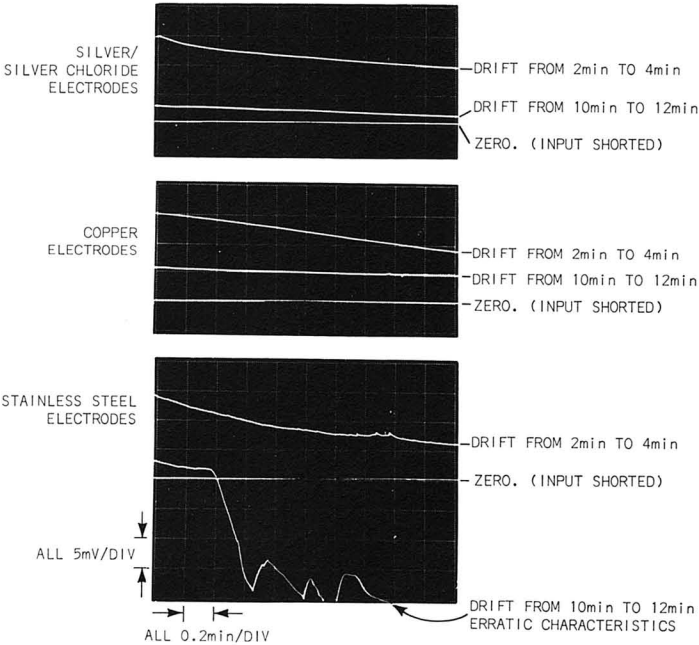

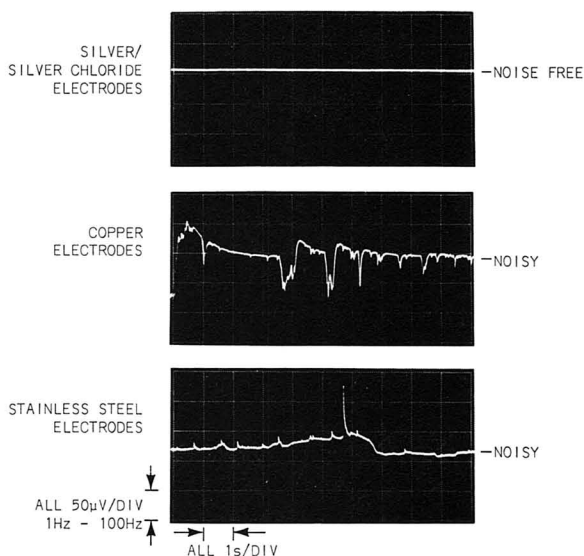Fig. 16-5. Electrode offset potential long term stability–drift.

Fig. 16-6. Electrode offset potential short term stability – noise.

drift
comparison

Referring to Fig. 16-5, the electrode offset
potential drift for silver/silver-chloride electrodes,
copper electrodes and stainless steel hypodermic
needle electrodes are shown. In each case the
electrode offset potential drift is shown over a
two-minute segment after the amplifier load had been
applied to the electrodes for two minutes and for
ten minutes. A "zero" trace is also included on each
photograph. The electrode offset potential produced
by the silver/silver-chloride electrodes and the
copper electrodes appears to decrease exponentially;
however, the stainless steel electrodes are clearly
erratic and, for the particular pair of electrodes
used for this investigation, show an abrupt change
in potential after about ten minutes. The electrode
offset potential drift for the silver/silver-chloride
electrodes and the copper electrodes would be
considered acceptable, however the erratic
performance of the stainless steel electrodes may
make them undesirable for physiological recording.

noise
comparison

Short term changes in electrode offset potential are
referred to as electrode noise. If electrodes are
used in conjunction with an AC-coupled high-
sensitivity amplifier as shown in Fig. 16-6, the
long-term drift characteristics will be rejected by
the low frequency characteristic of the amplifier.

However, short-term instabilities are clearly
displayed. From Fig. 16-6 it is clear that the
silver/silver-chloride electrodes are essentially
noise free, whereas copper electrodes and stainless
steel electrodes exhibit noisy characteristics that
may be intolerable for high-sensitivity physiological
recording.

If current is passed through electrodes, this current
will tend to charge the "battery" or "capacitor"
formed at the electrode-electrolyte junction. If
electrodes are used for tissue stimulation, this
current flow is deliberate. However, any electrical
activity associated with a subject, such as
stimulation or defibrillation, will cause some degree
of current flow in the recording electrodes. The
stored
charge
ability of the electrodes to be charged like a
battery or a capacitor by current flow is termed the
electrodes' "offset-potential stored-charge
characteristic." It is desirable that physiological
electrodes store as low a charge as possible, thereby
allowing an amplifier's differential-input impedance
to discharge this energy in as short a time as
possible. Referring to Fig. 16-7, silver/silver-
chloride electrodes, copper electrodes and stainless
steel electrodes were applied to a subject and used
in conjunction with an amplifier having a 2-megohm
differential-input impedance. A charge of 0.003
coulombs was then applied between these electrodes
by discharging a capacitor through them. It can be
seen from Fig. 16-7 that the silver/silver-chloride
electrodes recovered to within 50 mV in 0.2 seconds
and had almost completely recovered within 5 seconds.
stored-
charge
comparison
The copper electrodes exhibited inferior stored
charge characteristics, taking 3.5 seconds to recover
to 50 mV and several minutes to completely discharge.
The stored-charge characteristics of these copper
electrodes may be acceptable if no deliberate current
were applied through the subject.

The stainless steel electrodes exhibit particularly
poor stored-charge characteristics, taking 120
seconds to recover to within 50 mV and in excess of
six minutes to be sufficiently stable for ECG use.
The stainless steel electrodes may cause erratic
recording even if no current is deliberately passed
through the subject since electrostatically and
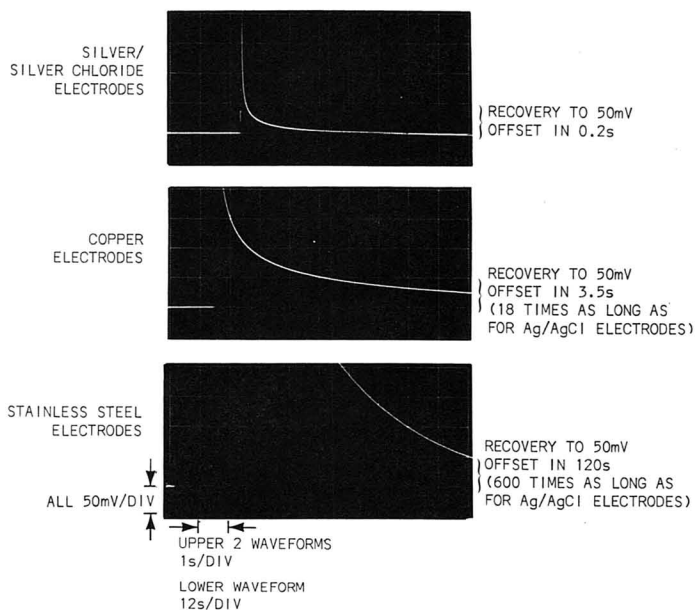electromagnetically induced currents are always
present.

SILVER/
SILVER CHLORIDE
ELECTRODES

) RECOVERY TO 50mV
) OFFSET IN 0.2s

COPPER
ELECTRODES

RECOVERY TO 50mV
) OFFSET IN 3.5s
) (18 TIMES AS LONG AS
FOR Ag/AgCl ELECTRODES)

STAINLESS STEEL
ELECTRODES

RECOVERY TO 50mV
) OFFSET IN 120s
) (600 TIMES AS LONG AS
FOR Ag/AgCl ELECTRODES)

ALL 50mV/DIV

UPPER 2 WAVEFORMS
1s/DIV

LOWER WAVEFORM
12s/DIV

Fig. 16-7. Electrode offset potential stored charge after .003 coulomb
surge – recovery.

EFFECT ON MEDIUM SENSITIVITY MEASURE-
MENT AT VARIOUS LOW FREQUENCY CUTOFF
FREQUENCIES.

— 1Hz
— 0.1Hz

— 0.01Hz
— DC

0.5mV/DIV

0.2s/DIV
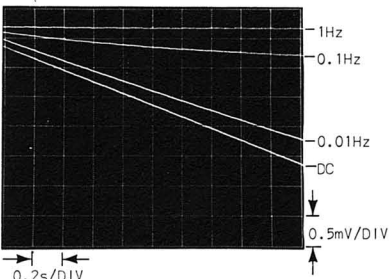
ELECTRODE DC OFFSET
DRIFTING AT 1mV/s.

EFFECT ON HIGH SENSITIVITY MEASURE-
MENT AT VARIOUS LOW FREQUENCY CUTOFF
FREQUENCIES.

10Hz —
1Hz —
0.1Hz —

50μV/DIV

0.2s/DIV
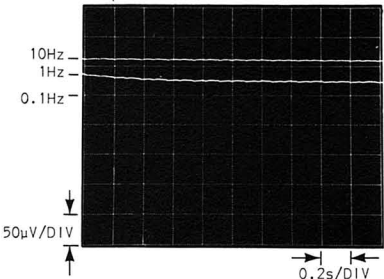
ELECTRODE DC OFFSET
DRIFTING AT 1mV/s.

Fig. 16-8. Electrode offset drift effect on AC coupled amplifier.

drift and
AC-coupling

Long term electrode offset potential changes (drift) cannot simply be ignored when using AC-coupled instrumentation as the rate of change of this potential must be related to the AC-coupling time constant in the recording amplifier. Fig. 16-8 shows the baseline changes exhibited when using electrodes whose offset potential is drifting at a rate of 1 mV/s. At a sensitivity of 0.5 mV/div, as would be used to record the ECG, a 0.01-Hz low-frequency response offers little improvement over a DC-coupled instrument. When recording at high sensitivities, say 50 µV/div as is the case with the EEG, this 1-mV/s electrode offset potential drift, used in conjunction with an amplifier having a 0.1-Hz low-frequency response, produces intolerable deflection and a 1-Hz low-frequency response must be used for satisfactory results. The drift rate of 1 mV/s chosen is typical of some of the poorer quality electrodes; however, silver/silver-chloride electrodes and many commercial electrodes can be expected to exhibit drift rates of between 10 µV/s and 100 µV/s after having been applied for more than two minutes.

electrode
offset
versus
amplifier
maximum
input speci-
fications

The absolute value of the electrode offset potential is rarely significant unless this absolute value exceeds the maximum differential-DC-input voltage characteristics of the amplifier concerned. When using DC-coupled instrumentation this electrode offset potential may exceed the maximum DC-differential offset available within the instrumentation and it would therefore be impossible to bring the trace onto the CRT screen.

check
'unknown'
electrodes

If the characteristics of an electrode are unknown, it is recommended that tests be performed on the electrode before the electrode is used for physiological recording. Electrodes may be tested *in situ* or, alternatively, the subject may be replaced by a 0.9% concentration saline solution as were most of the electrode characteristics presented in this chapter.

16.3 OTHER ELECTRODE CHARACTERISTICS

inertness

The most important electrode characteristics are the characteristics associated with electrode offset potential, that is, drift, noise and recovery. Other factors must also be considered when selecting electrodes. The electrode must, of course, be a good conductor and must be chemically inert. Inertness, or the relative inactivity of the material in chemical reaction, is approximately proportional to the metal's electrode potential as shown in Fig. 16-1. Metals with a high negative electrode potential such as aluminum tend to be more chemically active than metals with a high positive electrode potential such as silver. If an electrode is not relatively inert, the formation of ions by the electrode material and the reactions between these ions and the acidity of the subject's skin produce poisons which can irritate the subject.

mechanical character- istics

Mechanical characteristics of electrodes are important in many applications. Although it is of paramount importance that the electrode be satisfactory electrically, it must also be mechanically rugged, easy to clean and/or sterilize and easy to apply to the subject.

photosensi- tivity

Silver/silver-chloride electrodes exhibit photosensitive characteristics. It is not fully understood whether the effect of light on the silver/silver-chloride electrode actually results in a potential being generated, or whether it results simply in a change in the electrode's offset potential. Nevertheless, in some instances, it may be desirable to shield the electrode from light interference to insure satisfactory operation. So far we have been unable to detect any photosensitive characteristics in the Tektronix silver/silver-chloride electrodes.

silver purity

It should be noted that if silver is used for electrodes, either in its raw state or as a silver/silver-chloride electrode, it should be of spectroscopic grade (99.999% pure) to ensure noise-free operation. "Fine" jeweller's silver is nowhere near pure enough.

16.4 REUSABLE SURFACE ELECTRODES

reusable
versus
disposable

There are two principal categories of electrodes in use -- reusable electrodes and disposable electrodes. Reusable electrodes are intended to be used on many subjects after having been cleaned and perhaps sterilized after each use. Disposable electrodes are intended to be discarded after use on a single subject. Price is rarely a paramount factor in selecting a reusable electrode type; however, disposable electrodes must be inexpensive to be practical thereby limiting their performance. Reusable electrode types usually offer improved performance over disposable electrode types and may be expected to cost between one dollar and ten dollars for each electrode.

indirect
contact

Each of the two principal electrode types (reusable and disposable electrodes) can be further categorized into indirect contact (floating) and direct contact types. Indirect contact electrodes are built in such a way as to be spaced some distance from the skin and rely on an electrolytic bridge between the electrode and the skin. This electrolytic bridge is formed by conductive electrode paste or "jelly," such as the electrode paste manufactured by Day-Baldwin Inc. that is currently supplied as a standard accessory with the Tektronix Type 410 Monitor. Indirect contact electrodes typically produce less motion artifact than the direct contact types and their performance is somewhat more predictable as the electrode is always used in conjunction with only one type of electrolyte -- the electrode paste.

direct
contact

indirect-
contact
best
performance

Direct contact electrodes, as the terminology implies, are designed to make direct contact with the subject and they may, if necessary, be applied without electrode paste (dry). Since most direct contact electrodes are used with a small amount of electrode paste to improve electrode-skin contact impedance, they should probably more correctly be categorized between the indirect contact and the direct contact types. The electrodes supplied by Tektronix for use with the Type 410 Physiological Monitor and also the Beckman electrodes shown in Fig. 16-9 are indirect-contact silver/silver-chloride electrodes. It is generally agreed that this type electrode provides the best available electrode performance for surface electrode applications.
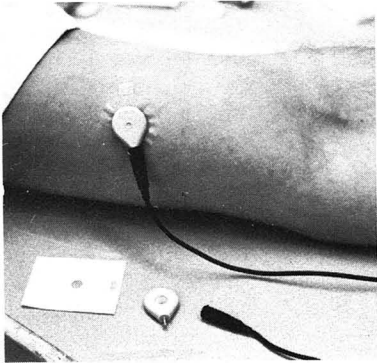
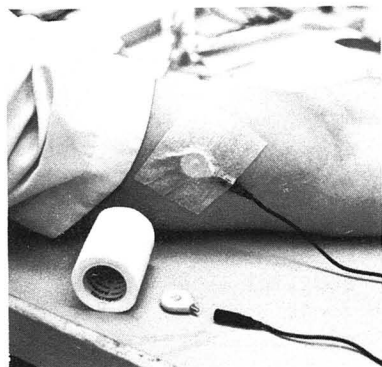ELECTRODES SUPPLIED BY TEKTRONIX FOR USE WITH THE TYPE 410 MONITOR
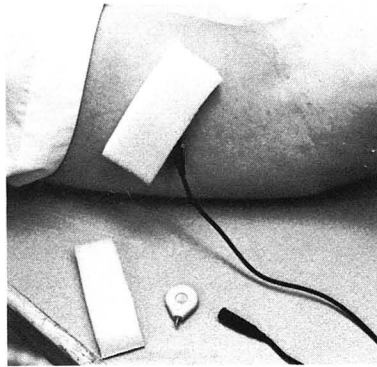
Type currently supplied
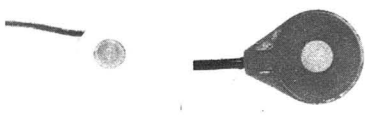
Earlier types supplied
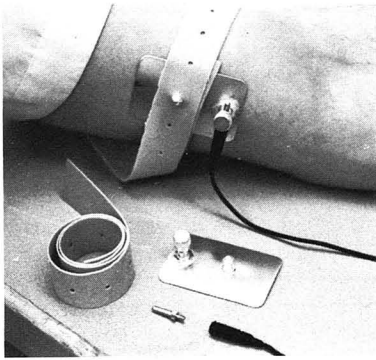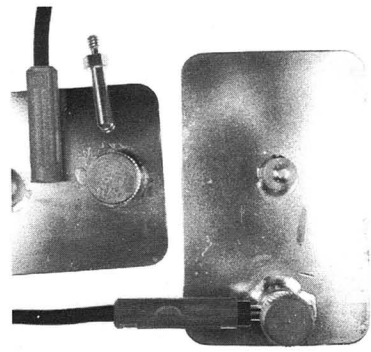
Recommended application configuration

Alternate application configuration
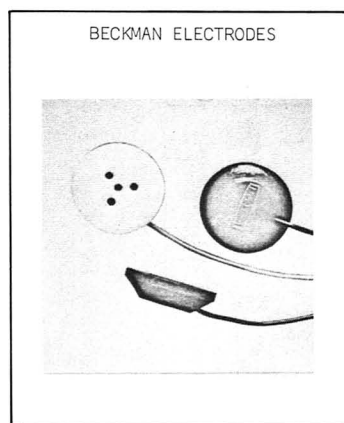
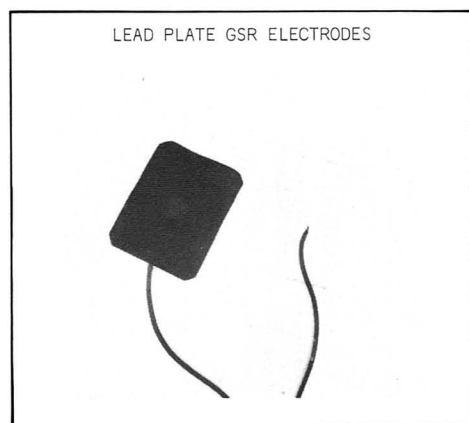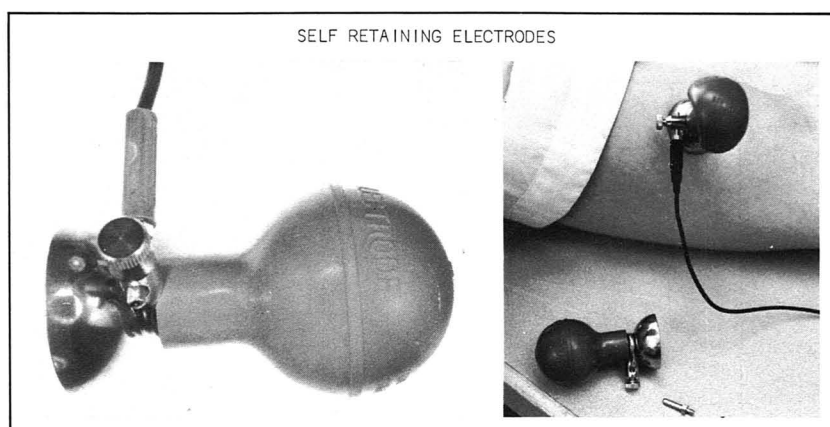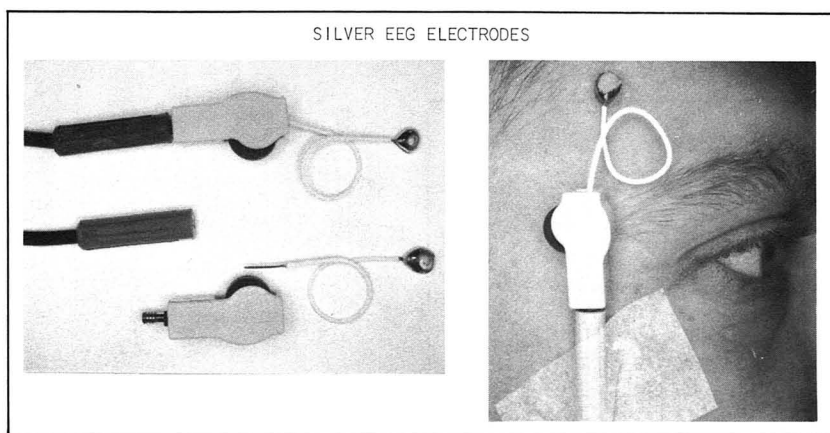Alternate application configuration

SILVER PLATE ELECTRODES

Fig. 16-9. Reusable electrode types.

THE 2 PIN PLUG USED ON THE ADAPTER
CABLES MAY BE REMOVED AND REPLACED
WITH A BNC CONNECTOR, TEKTRONIX
PART NUMBER 131-0297-00, TO ALLOW
USE WITH INSTRUMENTATION HAVING
BNC INPUT SOCKETS.

4-40 U.N.F.
THREAD USED

ELECTRODE
P/N 119-0197-00

BARE WIRE ADAPTER
P/N 103-0080-00

NEEDLE ADAPTER
P/N 103-0108-00

PLATE ADAPTER
P/N 103-0079-00

CUSTOM ADAPTERS FOR USE
WITH OTHER ELECTRODE TYPES
MAY BE FABRICATED USING
THE PLATE ADAPTER

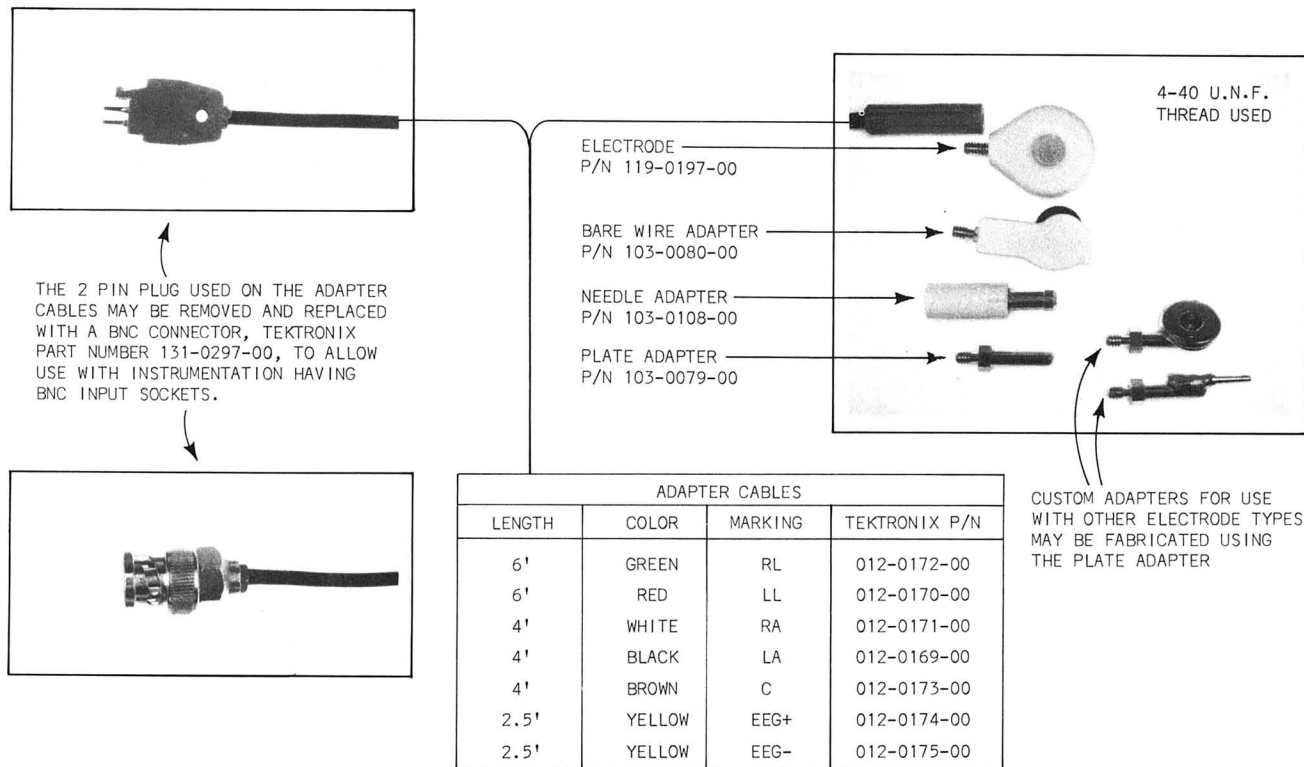| ADAPTER CABLES			
LENGTH	COLOR	MARKING	TEKTRONIX P/N
6'	GREEN	RL	012-0172-00
6'	RED	LL	012-0170-00
4'	WHITE	RA	012-0171-00
4'	BLACK	LA	012-0169-00
4'	BROWN	C	012-0173-00
2.5'	YELLOW	EEG+	012-0174-00
2.5'	YELLOW	EEG-	012-0175-00

Fig. 16-10. Tektronix electrode cables and adapters.

reusable
electrode
types

Fig. 16-9 shows various types of reusable surface electrodes. Earlier electrode types supplied by Tektronix for use with the Type 410 Physiological Monitor were permanently affixed to an electrode cable; however, the types currently supplied are detachable from the electrode cable. Fig. 16-9 shows the current Tektronix silver/silver-chloride electrode applied to a subject using the adhesive rings and electrode paste supplied by Tektronix with the Type 410 Monitor. Alternative application configurations, using surgical tape over the electrode or using an adhesive-backed foam pad over the electrode, are also shown. The large, silver-plated, copper electrodes historically used to record the ECG, the self-retaining electrodes currently preferred as a chest electrode for ECG recording, silver EEG electrodes, lead plate electrodes intended for GSR recording, and silver/silver-chloride electrodes produced by Beckman are shown. All of the electrodes in Fig. 16-9 are photographed to the same scale. These electrodes can be adapted for use with the Tektronix Type 410 Physiological Monitor using either a plate adapter, a needle adapter or a bare-wire adapter. These adapters are shown in use in Fig. 16-9 and further details on adapter cables and adapters available from Tektronix are shown in the next figure.

adapter
cables

Referring to Fig. 16-10, Tektronix supplies $2\frac{1}{2}$-, 4- and 6- foot adapter cables to allow signals to be connected from electrodes to the Type 410 Monitor. These adapter cables can be directly connected to the Tektronix silver/silver-chloride electrodes or can be connected to other electrode types via either plate adapters, needle adapters or bare-wire adapters. These are the only adapter types available from Tektronix; however, enterprising users have successfully adapted other electrode types for use with the Tektronix Type 410 Monitor. The electrode and adapter system shown in Fig. 16-10 is primarily intended for use with the Tektronix Type 410 monitor; however, if the two-pin plug supplied with these cables is replaced by a BNC connector, these cables and adapters can be used with other instrumentation having BNC input sockets.
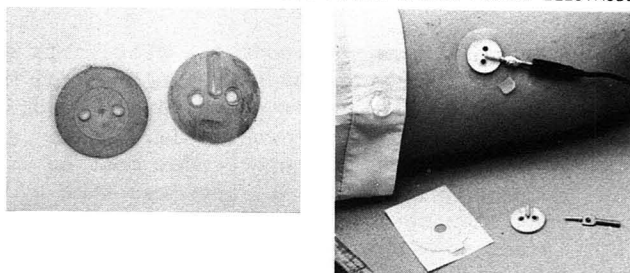
UNITED AIRCRAFT CORPORATION; HAMILTON STANDARD
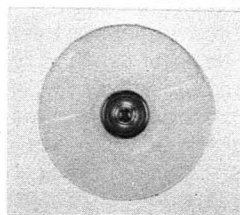DIVISION; ADHESIVE "TELECTRODE" DISPOSABLE ELECTRODE
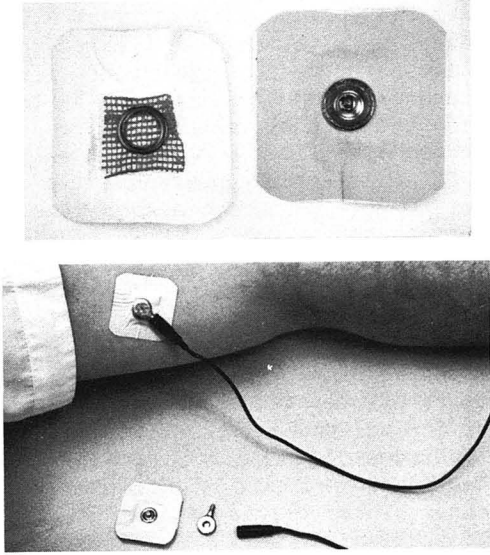
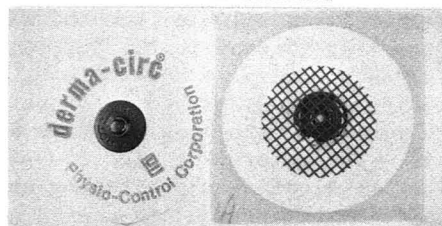ELECTRONICS FOR MEDICINE INC. DISPOSABLE PURE SILVER ELECTRODES
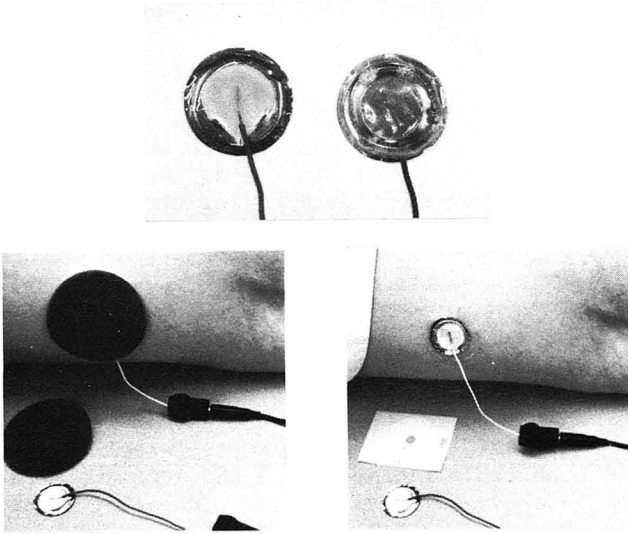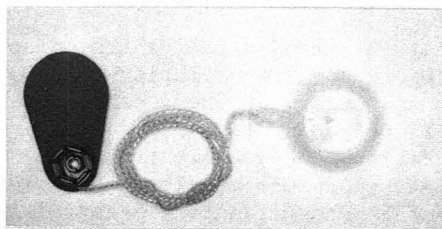
Fig. 16-11. Disposable electrode types.

16.5 DISPOSABLE SURFACE ELECTRODES

disposable
electrode
types

Tektronix does not produce a disposable electrode
for use with the Type 410 Physiological Monitor.
Disposable electrodes produced by six different
manufacturers are shown in Fig. 16-11. As stated
earlier, disposable electrodes must be inexpensive
and one would expect to pay less than a dollar for
each electrode. While disposable electrodes are
indeed convenient and offer adequate performance for
many applications, they appear to be unsuited to
applications where the subject may receive
stimulation and/or defibrillation as their electrode
offset potential stored-charge characteristics appear
to be substantially worse than for silver/silver-
chloride electrodes. While Tektronix would not wish
to recommend any particular brand of disposable
electrode, all are usable with the Type 410 Monitor
for ECG recording if the patient is not expected to
require defibrillation, stimulation or cauterization.

spray-on
electrode

A relatively new and unique disposable electrode is
the spray-on electrode developed by NASA for ECG
recording in space flight. This electrode system is
commercially available from the Hauser Research and
Engineering Company in Boulder, Colorado. These
electrodes basically consist of a film of conductive
adhesive that is sprayed onto the skin with a contact
wire imbedded into this adhesive during its drying
cycle. These spray-on electrodes are particularly
suited for long-term monitoring on subjects who are
physically active.

16.6 NEEDLE ELECTRODES

hypodermic
electrodes

Needle electrodes fall in the disposable electrode
category and are primarily used for ECG monitoring
during surgery or where extremely fast electrode
application is desirable, as in an emergency
situation. Needle electrodes may simply be stainless-
steel hypodermic needles. Stainless steel is a
somewhat unsatisfactory material for electrodes
(covered elsewhere in this chapter) as its

characteristics are inferior to most other electrode materials. Stainless steel must however, be used as it is one of the few materials that can offer sufficient mechanical strength. Typical long-needle electrodes are shown in Fig. 16-12. These needles are regular hypodermic needles but *short* hypodermic needles are preferable as it is only necessary to insert the needle just under the subject's skin.
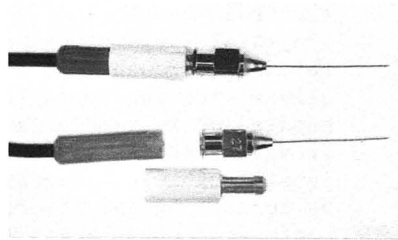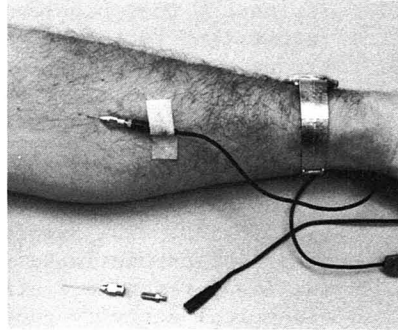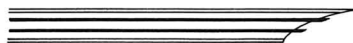

(A) HYPODERMIC NEEDLE ELECTRODES

INSULATED HYPODERMIC NEEDLE (OR SURGICAL NEEDLE)

CONCENTRIC NEEDLE ELECTRODE (HYPODERMIC NEEDLE WITH ONE CONDUCTOR THREADED THROUGH IT)

DIFFERENTIAL CONCENTRIC NEEDLE ELECTRODE (HYPODERMIC NEEDLE WITH TWO CONDUCTORS)

(B) CONCENTRIC NEEDLE ELECTRODES

Fig. 16-12. Needle electrodes (disposable).

The hypodermic needle electrode referred to previously is intended to perform similar functions to those of the surface electrodes. Special types of needle electrodes, such as concentric needle electrodes or insulated needle electrodes, may be used in applications where it is necessary to insert the needle to a bioelectric site *within* the body rather than to record the electrical activity produced near the surface of the body. Concentric needle electrodes may be two or three inches long for use in electromyography of the skeletal muscles and may be up to six inches long for certain applications in neurosurgery. Various arrangements of concentric needle electrodes are also shown in Fig. 16-12B. The simplest form of "concentric" needle electrode is an *insulated* hypodermic needle. The insulated needle effectively only records electrical activity generated in the region of the tip of the needle. A simple concentric needle electrode consists of a hypodermic needle with a length of 36-gauge enamel-covered silver wire threaded through it. The enamel covering on the silver wire insulates it from the stainless steel needle and is removed at the electrode tip area to provide a needle electrode similar to a miniature coaxial cable. A bioelectric signal may be recorded by connecting the recording amplifier between the center conductor and the outer concentric needle or, preferably, two concentric needle electrodes may be used with a recording amplifier connected between the center conductor of each and the outer needle of each acting as a grounded shield. Two pieces of enamel-covered platinum wire may be used in a hypodermic needle to produce a differential concentric needle electrode. This form of electrode is primarily used with the outer needle grounded and with the recording equipment connected between the two inner conductors. This effectively records the electrical activity produced in the region of electrodes spaced less than 1 millimeter apart. This type of differential concentric needle electrode may be used for recording the EMG and evoked responses.
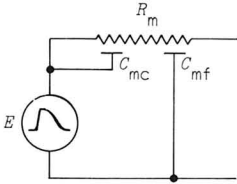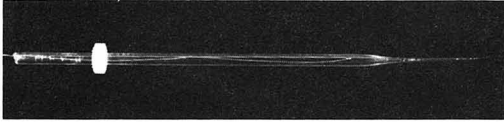
concentric electrode

differential concentric electrodes

16.7 MICROELECTRODES

Microelectrodes are used to record the electrical activity within individual cells, thus, the active electrode tips of these microelectrodes must be small enough so as not to interfere with the cell's normal function. Microelectrodes may be expected to have tip diameters within the range from 0.1 micron to 10 microns (one micron = 10^{-6} meters).

glass
micro-
electrodes

Microelectrodes can be divided into two broad classifications -- glass microelectrodes and metal microelectrodes. Glass microelectrodes consist of an electrolyte-filled glass tube that has been drawn to a capillary with a silver wire making connection to the electrolyte within the tube. The transfer from ionic conduction to electronic conduction is made at the relatively large surface area interface between the platinum wire and the electrolyte. Metal

metal
micro-
electrodes

microelectrodes consist of a finely tapered metallic rod covered with a layer of insulation. With metal microelectrodes, the transfer from ionic conduction to electronic conduction occurs at the electrode tip, that is, within the cell itself.
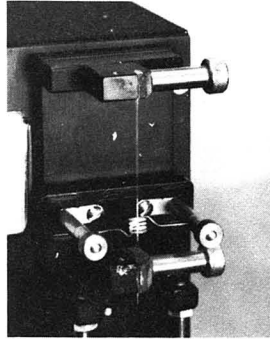
The characteristics of glass microelectrodes are covered in considerable detail in Chapter 11. The equivalent electrical circuit of the glass microelectrode is shown in Chapter 11 and is also shown in Fig. 16-13 together with the equivalent circuit for the metal microelectrode. Referring to

micro-
electrode
character-
istics

the equivalent circuit for the metal microelectrode, most of the microelectrode resistance (R_m) is located within the first few microns from the electrode tip. With the metal microelectrode inserted into the cell, this resistive area will be located within the cell. The capacity between the electrolyte within the cell and the metal microelectrode (with the microelectrode insulating material acting as the dielectric) is represented by C_{mc}.

GLASS MICROELECTRODE WITH SILVER WIRE

A GLASS "MICROELECTRODE PULLER"

TYPICAL:

C_{mc} = 0.1pF

C_{mf} = 0.2pF

R_m = $10^8 \Omega$

RESIN INSULATED METAL MICROELECTRODE

RESIN INSULATION

GOLD PLATED CONNECTOR

TUNGSTEN CARBIDE POLISHED ROD

METALLIC THIN FILM

TYPICAL:

C_{mc} = 15pF

R_m = $10^6 \Omega$

METAL MICROELECTRODE

CELL

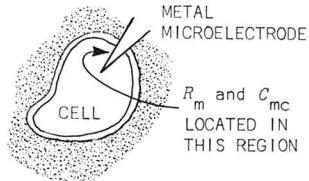R_m and C_{mc} LOCATED IN THIS REGION
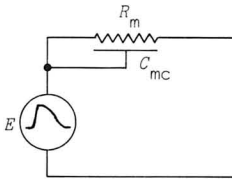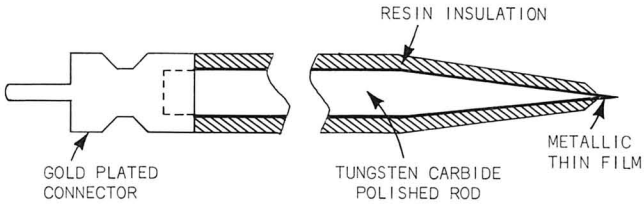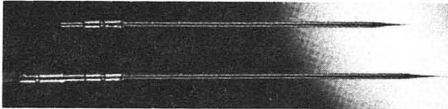
Fig. 16-13. Microelectrodes.

micro-
electrode
construction

Metal microelectrodes are formed by etching a metal
rod while the metal rod is being slowly withdrawn
from the etching solution, thus forming a tapered tip
on the rod. The etched metal rod is then dipped into
an insulating solution bath to form a layer of
insulation over the rod. This insulating layer is
prevented from covering the tip by the surface
tension action. Glass microelectrodes are formed
by heating a special quality glass tubing and then
drawing the soft glass tubing apart with controlled
tension to form a capillary. This is usually
achieved with a mechanical device known as a
"microelectrode puller" consisting of a heating
element and tensioning system. A microelectrode
puller is shown in Fig. 16-13.

16.8 ELECTRODE APPLICATION

Electrodes may be affixed to a subject in a variety
of ways and application recommendations are usually
included with electrodes at the time of purchase.
Fig. 16-9 shows various techniques for applying
many electrode types and, in the case of the
Tektronix electrodes, shows application by either
an adhesive ring, adhesive tape or an adhesive foam
pad. The silver EEG electrode shown in Fig. 16-9 is
glued to the subject's forehead with a commercial
adhesive known as collodion. All electrode types
offer better performance if applied after cleaning
and abrading the skin with a small amount of slightly
abrasive electrode paste. This tends to remove dead
skin cells from the surface and thus decreases the
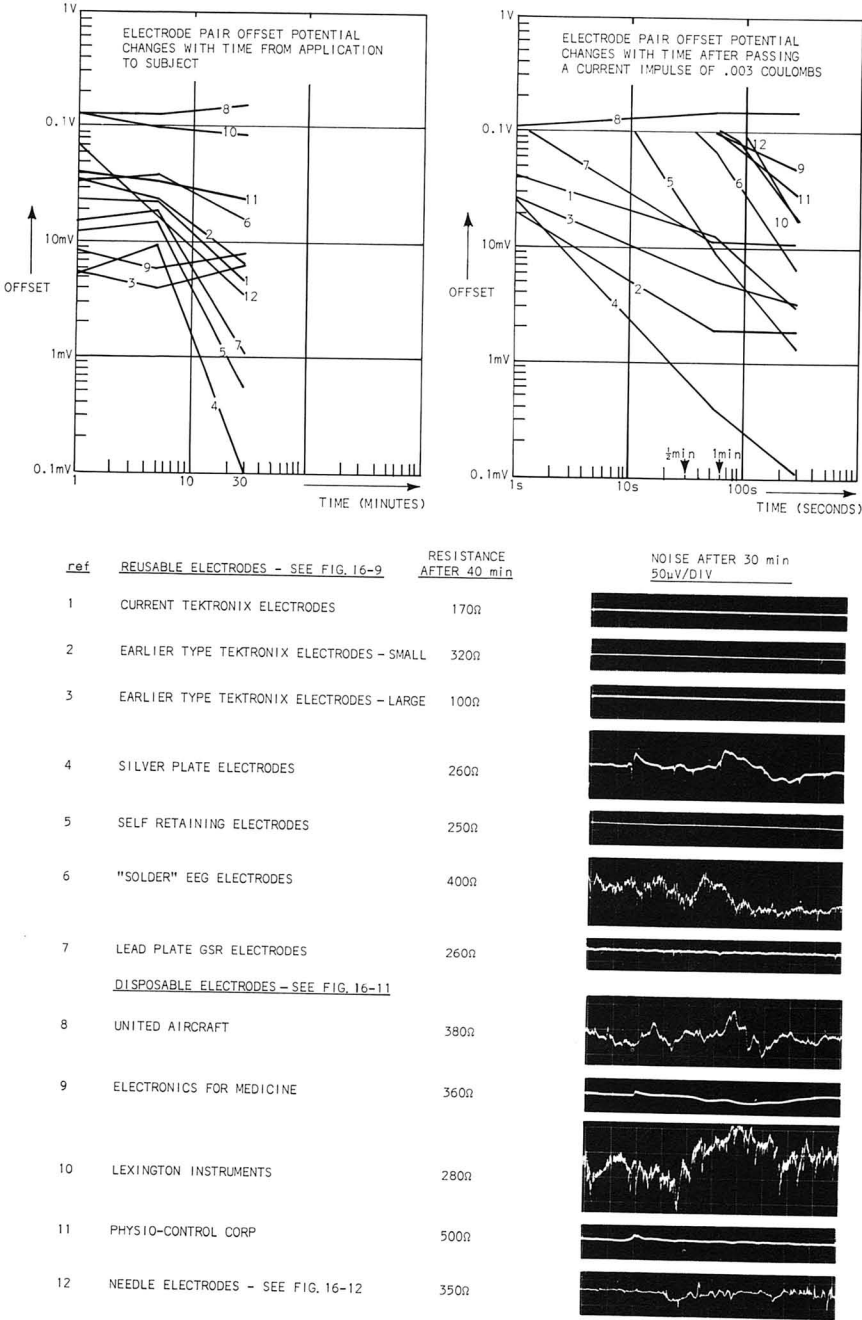electrode contact impedance.

Fig. 16-14. Characteristics of electrode types shown in Figs. 16-9, 16-11 and 16-12.

16.9   STIMULATING ELECTRODES

stimulating versus recording electrodes

Almost all of the electrode types discussed previously may be used as stimulating electrodes. When using electrodes for stimulation, current is deliberately passed between the electrodes. This current, unless it is truly biphasic, will increase the electrode offset potential, thus making the electrodes increasingly unsuitable for future use as recording electrodes, even if silver/silver-chloride electrodes are used as stimulating electrodes. It is advisable that, once any electrode has been used as a stimulating electrode, it be identified in some manner and not used thereafter as a recording electrode. Silver should be used for stimulating electrodes as it is nontoxic and does not tend to produce iontophoresis due to the passage of electric current. Ionotophoresis is the driving of external ions in through the skin which can produce considerable irritation if unsuitable electrodes are used.

16.10   COMPARISON OF ELECTRODE TYPES

30-minute drift comparison

There is undisputed evidence to suggest that silver/silver-chloride electrodes are far better than other types for recording physiological signals; however, they are not well suited to some applications and may offer unnecessary performance in other applications. Many non-silver/silver-chloride electrodes are used for physiological monitoring and are discussed in previous sections in this chapter. A comparison between 12 of the electrode types discussed in this chapter is shown in Fig. 16-14. Several pairs of each of the 12 electrode types were tested and, in each case, the pair exhibiting the poorest performance characteristics is shown in this characteristic summary. The first graph respresents the electrode offset potential drift of each of these 12 electrode types over a 30-minute period after the electrodes were applied to a subject and coupled to an amplifier having a 2-megohm differential-input impedance. In all cases, the reusable electrodes had been used several times previously.

After observing the drift characteristics of these
electrodes over a 30-minute period, a current impulse
of .003 coulombs was applied through the electrodes
by discharging a capacitor across them. The second
graph shows the change in electrode offset potential
for each of these 12 electrode types during the first
five minutes after application of the current impulse.
In all cases the electrodes were tested in an

electrolyte consisting of 0.9 percent sodium
chloride. The insertion of the electrodes into the
sodium chloride bath was adjusted to match the
electrode surface area in the bath to the electrode
surface area normally in contact with a subject.
The separation of the electrodes within the bath was
maintained at approximately three centimeters.
Fig. 16-14 shows the resistance of the electrode pair
to a 10 mA DC current after having been in the
electrolyte and connected to the amplifier for a
period of 40 minutes. Fig. 16-14 also shows the

noise produced by these electrodes over a 10-second
period after having been in the electrolyte bath and
connected to the amplifier for a 30-minute period.

While we would not wish to make a detailed analysis
of the results shown in Fig. 16-14, it is apparent
that reusable electrode types generally exhibit lower
electrode offset potential and lower noise
characteristics than the disposable electrode types.
It should also be noted that, particularly for
disposable electrodes, manufacturing techniques are
constantly changing and the performance typical of
an electrode type may change as the manufacturing
process is changed.

17

GROUNDING – SAFETY

artifact

The term artifact refers to the presence of an unwanted signal such as power line frequency interference or noise. When recording, for example, the EEG, the presence of ECG or EMG information on the recording also constitutes artifact. By far the most common source of artifact is power line interference; however, power line leakage considerations are of paramount importance from a safety viewpoint and it is not uncommon for safety considerations to seemingly contradict artifact-elimination considerations.

17.1 GROUNDING

If a subject were deliberately connected across a one-volt AC source, an AC current of perhaps 100 µA would flow through the subject due to the finite impedance ($\approx$10,000 ohms) of the subject and skin connections. Although such a connection should never be deliberate, many instrumentation systems inadvertently produce a similar situation in the form of a ground loop.

Fig. 17-1. Induced ground current from a "ground loop".

If a 10-foot length of wire is connected between a
patient's left arm and his right leg, a 60-Hz AC
current will be induced through the patient as this
wire acts as a transformer secondary, with current
magnetically induced from adjacent power wiring and
transformers. Although the magnetic coupling
material involved is air and the amount of coupling
therefore is extremely small, it is *finite* and an
EMF will be induced into the loop which will in turn
cause current to flow through the subject. This
10-foot length of wire may, in practice, be two
separate grounding wires attached to a subject
(Fig. 17-1).

induced
60-Hz

Referring to Fig. 17-1, if either the ECG monitor
or the "other monitoring equipment" is connected to
the subject, a ground loop is not produced. However,
if both instruments are simultaneously connected to
the subject, the ground connections for each
instrument form a ground loop. The ground loop shown
produces a potential between the subject's right leg
and left arm causing a current to flow through the
patient's bulk body resistance. This current flow
produces a voltage drop across different parts of
the body which would appear as a 60-Hz artifact on
the ECG monitor.

ground
loop

The ground loop produced by the configuration shown
in Fig. 17-1 may simply be eliminated by removing
one of the ground connections to the subject
whenever both items of monitoring equipment are used
simultaneously. If this is particularly
inconvenient, the ground loop may be eliminated by
adding a high impedance in series with one of these
ground leads to reduce current flow to a negligible
value or the subject may be grounded at a single
reference point so any induced current flow does not
flow through the subject's bulk body impedance. Both
of these ground loop changes are shown in Fig. 17-1
and both changes effectively eliminate the current
flow through the subject, thus eliminating 60-Hz
potential differences between various points on the
subject.

eliminating
ground
loop

17.2   INDUCED GROUND CURRENTS
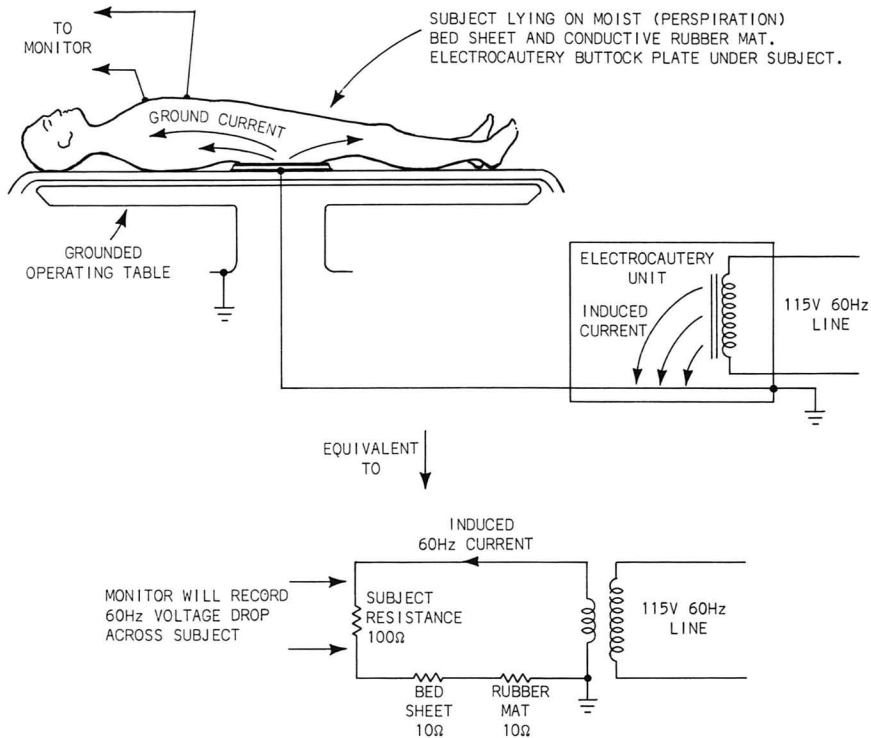
A ground loop is an example of an induced ground current, the value of which can be reduced or eliminated by modifying the grounding configuration. Another form of induced ground current is illustrated in Fig. 17-2 where a current is induced into a ground wire via an instrument's power transformer. In this specific case, an induced 60-Hz ground current is induced into the grounding circuit of an electrocautery unit. This ground current flows via the electrocautery ground plate, through the subject and to ground via a perspiration-dampened bedsheet, a conductive rubber mat and the grounded operating table.

The above type of induced ground current is particularly difficult to eliminate as safety considerations demand that the subject be grounded and that the electrocautery-unit buttock plate be well grounded to the electrocautery unit and to the subject to eliminate burns at electrode sites during electrocautery. This particular problem is often encountered when trying to monitor a subject's ECG in an operating room using an older style electrocautery unit. When these older units were designed it was not expected that the ECG may be monitored while the electrocautery unit was connected to the line ready for use. Modern electrocautery units have more carefully designed internal-grounding configurations and incorporate shielding to prevent any magnetically induced or capacitively induced current from entering the subject via the buttock plate. Perhaps the only solution that can be offered for an electrocautery unit of the older design is to discard the unit. Although this "solution" may seem harsh, the laws of physics offer no alternative and, more importantly as will be seen later in this chapter, the induced current produced by the electrocautery unit may be sufficient to cause death if grounded catheters are used on the subject.

TO MONITOR

SUBJECT LYING ON MOIST (PERSPIRATION) BED SHEET AND CONDUCTIVE RUBBER MAT. ELECTROCAUTERY BUTTOCK PLATE UNDER SUBJECT.

GROUND CURRENT

GROUNDED OPERATING TABLE

ELECTROCAUTERY UNIT

INDUCED CURRENT

115V 60Hz LINE

EQUIVALENT TO

INDUCED 60Hz CURRENT

MONITOR WILL RECORD 60Hz VOLTAGE DROP ACROSS SUBJECT

SUBJECT RESISTANCE 100Ω

115V 60Hz LINE

BED SHEET 10Ω

RUBBER MAT 10Ω

AN INDUCED 60Hz CURRENT OF 350nA RMS WILL PRODUCE 2mm OF 60Hz NOISE ON A STANDARD ECG MONITOR

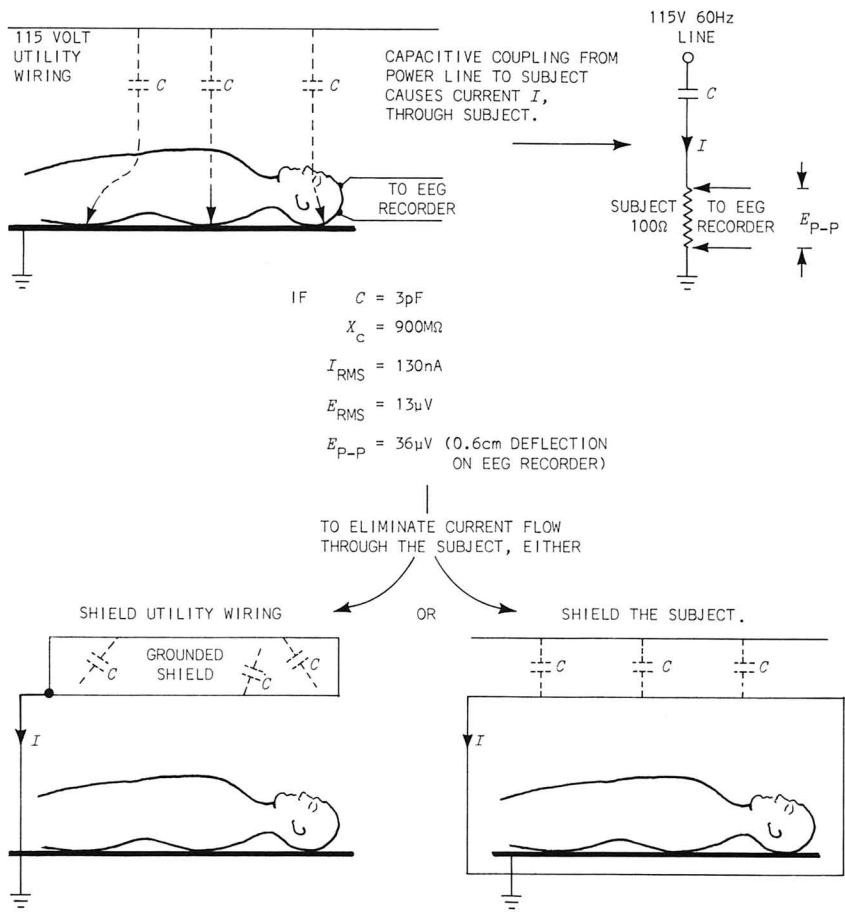Fig. 17-2. Induced ground current from electrocautery unit.

254

Fig. 17-3. Electrostatic shielding.

17.3 ELECTROSTATICALLY INDUCED CURRENTS

utility
wiring
capacitance

Capacitive coupling between the 115-volt wiring and the subject constitutes an impedance which will allow current to flow through the subject. The situation is illustrated in Fig. 17-3 which shows that a stray capacitance of 3 pF between the utility wiring and the subject can produce 6 centimeters of 60-Hz artifact on an EEG recording.

utility
wiring
shielding

To eliminate electrostatically induced currents, it is necessary to provide an alternative path for this current to flow. This may be achieved by either shielding the 115-volt utility wiring, in which case the current flows from this wiring to the shield, or by placing the subject inside an electrostatic shield (Faraday Cage), in which case the current flows to ground via the shield rather than via the subject. In either case, shielding is expensive and it may be preferable to locate a recording site in an alternative part of the hospital or laboratory where electrostatically induced current is less bothersome. In the research environment, where high sensitivity amplifiers are invariably used, the subject and parts of the recording equipment should be enclosed in a screened room to effectively isolate the subject from capacitively induced currents from any source.

recording
lead
shielding

Currents may be capacitively coupled to the leads connecting the subject to the recording equipment, thus these leads must be screened to effectively shield them. Care should be taken to ensure that these shields are, however, only grounded at one point otherwise, in attempting to eliminate interference by electrostatic shielding, one may inadvertently create 60-Hz interference by introducing a ground loop. This interference may be produced by 60-Hz power wiring or it may be high-frequency interference produced by radio and/or television transmitters.

17.4 ELECTRIC SHOCK CURRENT THRESHOLDS

When electric current is passed through the body, part of this current will pass through the heart and interfere with the normal function of the heart, perhaps causing death. In cases of death by electrocution, almost all subjects are killed by the passage of electric current through the heart rather than by some related occurrence such as burns or

muscular paralysis of the muscles controlling breathing action. If a person receives a 115-V shock, it is not the voltage of this shock that is important but rather the *current* through the body. Thus, if a person receives a 115-V "shock" standing in a dry environment insulated by normal clothing, the "shock" may hardly be felt as the current may be well below 1 mA. If, however, a person receives a 115-V shock while standing barefooted on moist ground, the person would receive a severe shock and probably be killed as the current may exceed 100 mA.

Referring to Fig. 17-4; experimental investigation on numerous subjects (both male and female) has indicated that up to 300 μA applied to the surface of the body, such as from one arm to the other, may be suggested as being reasonably safe for most subjects. It can be seen that 99.5 percent of the population require 400 μA or more of 60-Hz current to perceive the current and that the threshold of perception or sensation increases as frequency increases above 100 Hz and below 10 Hz. At 10,000 Hz and at DC, the threshold of sensation is approximately five times greater than at 60 Hz. While current above the threshold of sensation may not be detrimental to a healthy person, it may cause complications in a hospitalized patient and will certainly create anxiety which would be detrimental to the patient's general well being. From Fig. 17-4 it can also be seen that "cannot-let-go" current thresholds are an order of magnitude above sensation current thresholds and that they also increase for frequencies below 10 Hz and above 100 Hz.

Fig. 17-5 shows the physiological effects of 60-Hz arm-to-arm current and shows a threshold of sensation at 300 μA, a threshold of pain at 1 mA, a "cannot-let-go" threshold at 10 mA and a ventricular fibrillation-induction threshold at 100 mA. 99.5% of the population will have thresholds above these values. Ventricular fibrillation refers to malfunctioning of the ventricular musculature which will interfere with the normal blood-pumping action of the heart and eventually cause death. Ventricular fibrillation is produced by current directly through the heart during a specific portion of the cardiac cycle known as the "vulnerable period." The vulnerable period for ventricular muscle occurs during the upstroke of the T-wave and a single shock impulse lasting for less than 0.1 second could cause ventricular fibrillation if received during this vulnerable period.
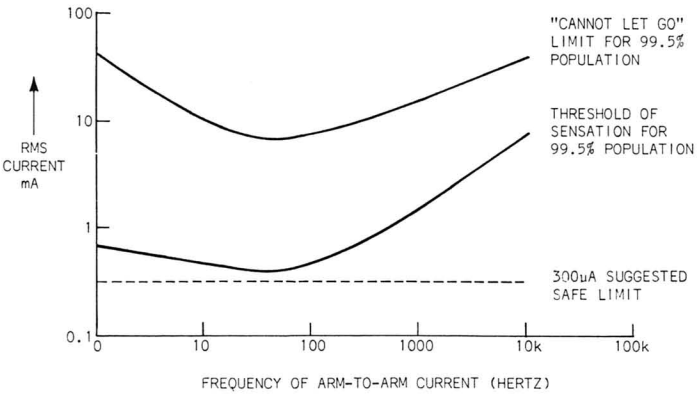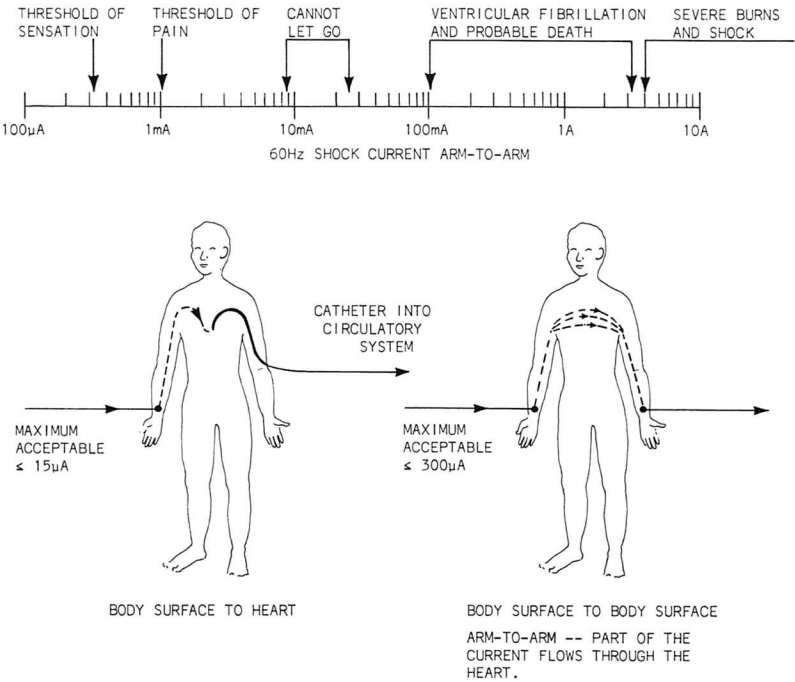
Fig. 17-4. Physiological effects versus current frequency.

Fig. 17-5. Physiological effects of 60Hz electric current.

The 300 µA threshold refers to 60-Hz current applied from one arm to the other arm and is intended as a guide to acceptable leakage current levels. The acceptable level for any one subject may be somewhat above this value and will also depend on the size of the current contact. A very small point contact can undoubtedly be felt at 0.3 mA, but a current in excess of perhaps 1 mA may not produce sensation if the contacts are somewhat larger. Depending on the size of contact, the threshold of pain may also be considerably above 1 mA, probably 10 mA if the contacts are large enough. It is thus evident that current density is also important in determining the physiological effects of electric current.

current density versus thresholds

The threshold of sensation of electric currents differs greatly between currents applied arm-to-arm and currents applied internally to the body. If a current is applied internally, a far greater percentage of the current may flow via the arterial system directly through the heart, thus less over-all current is required to produce ventricular fibrillation. Experiments with dogs have indicated that in some animals ventricular fibrillation can be produced with currents as low as 17 µA applied directly to the dog's heart. In general, as the threshold of sensation in man may not be much higher than for the dog, 30 µA of 60-Hz current through the heart may produce ventricular fibrillation and 15 µA could perhaps be considered as a safe upper limit. 5 µA has been postulated as a safe upper limit under certain adverse conditions.

internal current thresholds

A probe or electrode within the body such as a cardiac catheter or pacemaker electrode will provide a direct electrical pathway to the heart and thus the 15 µA current limit proposed in the preceding paragraph should be considered as an upper limit for current flow through these conductors. The impedance from these conductors to the subject's skin surface may typically be 1,000 ohms, thus a potential of 30 mV between these conductors and a point on the surface of the subject's body is sufficient to cause electric shock, ventricular fibrillation and possible death! It has been proposed that the introduction of the *internal electrode* has been the leading factor in the present high incidence of accidents in hospital patient care areas and operating rooms.

internal probes and electrodes

high accident rate

Cardiac catheterization for diagnostic or therapeutic purposes has become common in recent years. Sensing catheters may have electrodes or transducers at their tips for ECG recording, blood pressure measurements and other diagnostic procedures. Fluid-filled catheters are also in common use and, although they are not normally considered to be a probe or electrode within the body, the fluid in these catheters may be conductive, thus they also may provide an electrical pathway to the heart.

17.5 INSTRUMENTATION SAFETY CONSIDERATIONS

Twenty years ago the electrocautery unit was the only item of electronic equipment routinely used in the operating room. Any electrical shock hazard associated with this instrument was far outshadowed by the risk of explosion from flammable anesthetic gases. Nowadays the operating room is packed with electronic equipment, all of which is necessary for the fulfillment of today's modern surgical procedures, and the risk of explosion from flammable

unprecedented shock hazard demands system approach

anesthetic agents is considered of secondary importance. All of the instruments used in the operating room may be used alone with relative impunity however, in combination, they produce an unprecedented electrical shock hazard of giant proportions. The interaction of various units connected to the subject demand a *system* approach which may be foreign to the training of medical staff and may also not be fully understood by the manufacturers of the various items of instrumentation. Some medical instrument manufacturers tend to be oblivious to this problem and tend to regard their product as an entity within itself rather than as a part of an over-all system.

power transformer leakage currents

The major source of potentially lethal currents in any instrument is leakage current from the 115-V (or 230-V) power-transformer primary. This leakage current is largely due to capacitive coupling from the power transformer primary to other parts of the transformer and/or other parts of the instrument. Instruments are usually designed so this leakage current flows to the instrument case and then to ground via the three-wire power cord provided with the instrument. Most modern general-purpose instrumentation may be expected to produce up to 500 µA of leakage current, however instrumentation

designed for use in the medical environment should preferably have a leakage below 100 μA. By special design techniques it is possible to decrease this leakage to well below 10 μA. Although this may not always be practical, as long as the instrument is adequately grounded, the leakage current will not flow through the subject. A problem arises when instrumentation is used in an ungrounded configuration, as would be the case if the three-pin power plug provided with the instrument was replaced by a two-pin plug to allow the instrument to be used with the older style two-pin power receptacles. Under these conditions the ground leakage current may flow through the subject as shown in Fig. 17-6 and if one of the conductors is an internal electrode, such as a catheter, death will almost certainly result.

It is apparent that if the subject were "floating" or isolated, then a path to ground would not be provided for the leakage current mentioned in the previous paragraph and no current would flow through the subject. In practice, however, this does not reduce the risk as fault currents could still flow between various items of equipment without involving "ground" in the current circuit; also, if a subject inadvertently touched a grounded item, such as a bed frame or water pipe, then the isolation would be nullified and electrocution may result. It is, therefore, apparent that protection systems must be organized on a grounded system concept.
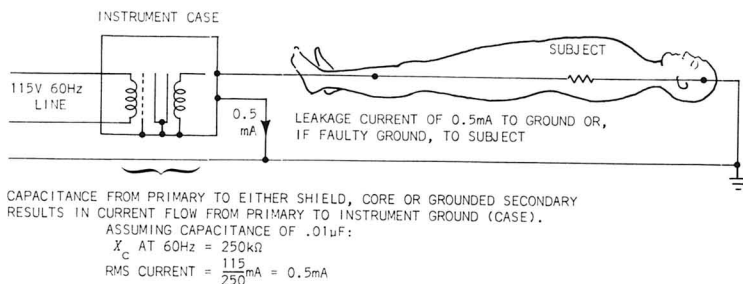
(margin note:) shock currents from ungrounded instruments

(margin note:) shock currents to isolated subjects

CAPACITANCE FROM PRIMARY TO EITHER SHIELD, CORE OR GROUNDED SECONDARY RESULTS IN CURRENT FLOW FROM PRIMARY TO INSTRUMENT GROUND (CASE).
ASSUMING CAPACITANCE OF .01μF:
X_c AT 60Hz = 250kΩ
RMS CURRENT = $\frac{115}{250}$mA = 0.5mA

Fig. 17-6. Transformer capacitive leakage currents.

The various hazards mentioned so far in this chapter primarily refer to current flowing through the subject if a fault occurs or if something abnormal happens. As in any situation, faults do occur and abnormalities do happen, thus it is desirable to offer some second order protection to the subject.

second order shock protection

Under normal operation of, for example, an ECG unit, only a small signal current should flow in any of the active electrodes or the right-leg ground electrode. Under abnormal or fault conditions, however, current may flow through any one of these paths and second order protection can be offered by incorporating some form of current limiting. The simplest form of current limiting, and a form adequate for use with the right-leg ground electrode, is to use a large value resistor in series with the electrode. Series resistance current limiting is not, however, always possible. In any amplification equipment, such as an ECG monitor, series resistance will degrade the common-mode rejection-ratio specification of the instrument to a point where 60-Hz interference may be intolerable. It is sometimes necessary to use a more sophisticated form of current limiting.

current limiting

The Tektronix Type 410 Physiological Monitor is specifically designed for operating room use and incorporates series-resistance current limiting in the right-leg ground electrode and more sophisticated current limiting in series with the active electrodes. This current limiting utilizes field effect diodes, the impedance of which is approximately 1,000 ohms when the current is less than 1 μA. However, the diodes limit fault currents to a maximum of 300 μA as their impedance increases as the current attempts to exceed the 300 μA level. It is thus apparent that current limiting within the Tektronix Type 410 Physiological Monitor provides adequate second order subject protection for "arm-to-arm" fault currents delivered by surface electrodes. It does *not* provide adequate protection to allow the Type 410 to be used with intracardiac-catheter electrodes. It should be noted that the instruction manual provided with the Tektronix Type 410 Physiological Monitor specifically states that the unit is *not intended* for use with internal electrodes of any type.

current limiting with field effect diodes

Many oscilloscopes produced by Tektronix and electronic instrumentation produced by other manufacturers are used in operating rooms and in intensive-care rooms on live human subjects. It should be noted that this type of instrumentation was designed for use in the electronics laboratory and does *not* incorporate any secondary subject protection. If a Tektronix product, other than the Type 410 Physiological Monitor, is to be used in conjunction with a human subject, we suggest that some form of current limiting be included in series with any electrode in contact with the subject. Spacelabs, Inc., Chatsworth, California, market an electrical shock protector which can be used in series with electrodes if the nominal line voltage is below 120-V AC or, alternatively, a current limiting adapter is described in Chapter 29 which is suited for use in areas with either 120-V or 240-V power. It provides secondary protection by limiting current to 300 µA. Secondary protection does not guarantee safety, it simply reduces the probability of an accident. Each protection feature added to instrumentation should reduce this probability.

current
limiting
adapters

17.6 ELECTRICAL SERVICE GROUNDING

Section 17.4 pointed out that a potential of only 30 mV between two supposedly grounded conductors may allow enough current to flow through a subject to cause death by electrocution. If multiple ground points are available in an operating or intensive-care room there is no guarantee that all of these are at the same potential and a voltage may exist between, for example, a grounded wall outlet and a nearby water pipe. These subtle ground potential differences are difficult to detect, but they may be *fatal* if left undetected. It is not uncommon to find two electrical power services available in older hospitals, the second service having been provided to handle increased power demand since the hospital was constructed. This second service may come from a different part of the hospital and the potential at its ground conductor may differ from the potential of the ground conductor of the older service by a

potentials
between
different
grounding
systems

volt or more. If subtle differences in ground
potential are detected, they can usually be almost
completely eliminated by interconnecting each ground
point with a heavy-gauge copper grounding bus. As
currents of several amperes may flow in this
interconnecting grounding bus, the copper wire used
should be at least a 1/4 inch in diameter. Such
subtle ground differences can be detected with a
differential AC voltmeter or a differential
oscilloscope.

remote
fault
currents

Referring to Fig. 17-7, it is apparent that two
ground points from the same power service may show
a potential gradient due to a fault current in
equipment connected elsewhere in the power service.
Although the fault current of 1 A shown in Fig. 17-7
is insufficient to trip a circuit breaker in the
active line of the power service, it is sufficient
to produce a voltage drop of 50 mV across 20 feet
of standard service ground wire. Once again, this
50 mV may be enough to cause a fatality. To avoid
this difficulty the service outlets should be grouped
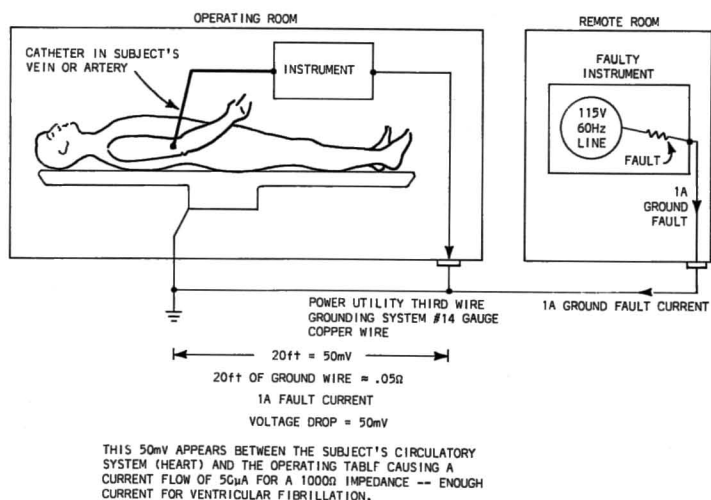together to provide a single point ground.

Fig. 17-7. Subject injury caused by a "remote" electrical fault.

response
time of
protection
devices

A catastrophic remote electrical fault may produce a fault current of perhaps 50 amperes which will cause the primary circuit breaker to trip. This circuit breaker may, however, take several hundred milliseconds to disconnect the service after the fault has occurred, thus the fault current may flow for more than 100 milliseconds. As discussed in Section 17.4, if this fault current occurred during the "vulnerable period" of a subject's cardiac cycle, any potential produced between different ground points attached to the subject may be sufficient to cause death.

17.7   OPERATING ROOM ISOLATION

static
charge
prevention

In an effort to reduce the risk of flammable anesthetic ignition in an operating theater, conductive and grounded flooring is used, the subject is connected to a grounded operating table by a conductive rubber mat and moist bedsheet and the operating personnel wear conductive clothing. These precautions prevent the build-up of static electricity and the subsequent spark caused by the discharge of this static electricity.

fault
current
monitors

Due to the presence of this extensive grounding in the operating room, the electrical service to operating rooms normally has both conductors isolated from ground so if a fault develops between one conductor and ground, no fault current can flow. The integrity of this isolation is monitored continuously by ground fault detectors installed between either side of the service and ground. These detectors normally signal any ground fault causing a current to flow in excess of about 1 mA. Unfortunately, there is no more maligned item of electrical equipment in the operating room than the ground fault monitor; if the monitor alarm sounds, the operating room personnel often as not suspect the monitor rather than heed its warning. It should also be noted that the 1 mA level at which the ground-fault monitor alarm sounds may be low enough to protect the subject against currents flowing from surface contacts to his body but is not low enough to protect the subject from shock from internal electrodes or catheters.

17.8 ELECTROCAUTERY AND DEFIBRILLATION

electro-
cautery

The electrocautery unit is extensively used in operating rooms and is used to provide a source of high-frequency RF current for either cutting tissue or welding tissue together. The electrocautery probe provides a high frequency source of 2 MHz and up to 15 kV with respect to ground. When this probe is applied to a subject during surgery, current flows between this probe and ground. The electrocautery unit is always used in conjunction with a large buttock plate underneath the subject in such a way as to provide a large surface area, good electrical contact, ground to the subject. An electrocautery current of several amperes will thus flow between the electrocautery probe and the buttock plate, however the energy dissipated will cause appreciable heating and burning only in the region of the probe as the current density flowing in the region of the large surface area buttock plate will be quite low.
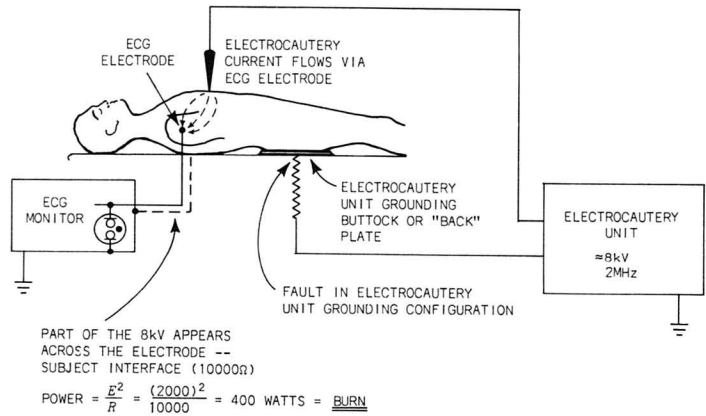

Fig. 17-8. Electrocautery burns at ECG electrode sites.

If the electrocautery unit does not provide an excellent ground to the subject, either due to poor application or due to a fault within the buttock-plate grounding circuit, then the electrocautery current will be diverted to other grounded points on the subject. It is not uncommon for ECG electrodes to provide this alternative grounding point as all ECG monitoring equipment essentially provides a low impedance path to this high-voltage, high-frequency current. Unfortunately, however, the ECG electrode does not behave in the same manner as the buttock plate as the contact area involved is smaller by a factor of at least 100, perhaps 1,000. The current density is thus increased by a corresponding factor and will almost certainly cause heating and burns at ECG electrode sites. The high-frequency current used in electrocautery is at too high a frequency to cause ventricular fibrillation. In practice, if a surgeon suspects that his electrocautery unit "is not cutting too well," he should be suspicious of the electrocautery unit ground circuitry and should have the electrocautery unit adequately checked before continuing to use it.

electrode burns during electro- cautery

The defibrillator is another high voltage device commonly used in patient-care areas. The defibrillator provides a single short-duration pulse of up to 10 kV between two large electrodes. These electrodes are normally applied to a subject's chest in cases where the heart is in ventricular fibrillation or has stopped completely. The high energy defibrillation pulse is transmitted between the electrodes and a great portion of this energy is dissipated in the heart. Defibrillation hopefully either re-starts the heart or reverts it from ventricular fibrillation to a normal beating action. Defibrillators should have both electrodes isolated from ground when the high voltage pulse is applied and can normally be expected to provide safe operation if other personnel are not in contact with the subject during the defibrillation process.

defibril- lation

Although a defibrillator's output is isolated, this output will be unbalanced capacitively to ground. Thus, unless monitoring devices, such as an ECG instrument, incorporate input circuit protection they may be damaged by the defibrillator pulse.

cardio-
version

If a subject's heart has not stopped beating altogether but is beating erratically, a defibrillator may be operated in a synchronized mode to revert the heart to a normal rhythm. This is referred to as cardioversion. The defibrillator synchronizing pulse is obtained from the subject's ECG R-wave, the defibrillation pulse occurring sometime after the ECG R-wave but avoiding the vulnerable period during the upswing of the ECG T-wave.

18

TRANSDUCERS – TRANSDUCER SYSTEMS

defined

A transducer may be generally defined as a device which converts one form of energy into another. As far as this text is concerned, a transducer is defined as a device which converts any form of energy other than electrical energy into electrical energy. Transducers are used to measure physical quantities and phenomena by producing an electrical output proportional to this quantity or phenomena. Some transducers require electrical excitation, others produce an electrical output without the aid of excitation. A complete transducer system consists of the transducer as well as the necessary excitation voltage source if excitation is required. Transducer systems may also incorporate some output signal amplification.

biophysical measurements only

This chapter should not be regarded as a general reference on transducers, it specifically discusses transducers from a biophysical measurement standpoint and makes no attempt to review some of the more sophisticated transducer systems used for high-accuracy physical measurements. A companion volume entitled *Transducer Measurement Concepts* provides a more general purpose reference source for transducers and transducer measurements.

18.1 RESISTIVE TRANSDUCER CONCEPTS

A resistive transducer is a device whose resistance
changes in proportion to some physical quantity.
This resistance change is then converted to an
electrical output signal by utilizing additional
circuitry. Various types of resistive transducers
are discussed later in this chapter. For the purpose
of this text, resistive transducers may be divided

two   into two broad classifications: Those where the
classes   change in resistance expected with a changing
physical quantity will be greater than five percent
of the transducer's nominal resistance and those
where the change in resistance will be less than
five percent. This ratio of change in resistance to
nominal resistance is normally shown as $\frac{\Delta R}{R}$. This
classification is necessary as the circuits used in
conjunction with each of these transducer types are
quite different.

Any one particular transducer may encompass both of
the above broad classifications. For example, a
thermistor used to measure body temperature must be
particularly sensitive and is only expected to be
used over a narrow temperature range. This
temperature range would not normally change the

type and   thermistor's resistance by more than five percent.
application   A thermistor used, however, to record outside air
temperature would be expected to operate over a broad
temperature range and the thermistor's resistance may
be expected to change by 50 percent or more. Thus,
the transducer type as well as the transducer's
application determines the classification.

Resistive transducers are normally used in one of
the three circuit configurations shown in Fig. 18-1.
The circuit shown in Fig. 18-1A essentially powers
the resistive transducer from a constant voltage
source and monitors the current through the
transducer. This circuit is generally the easiest

voltage   to implement as the excitation source may simply be
excitation   a low-voltage power supply or battery; however, the
circuit is nonlinear unless the value of the current
sensing resistor R2 is negligible in proportion to
the value of the transducer's resistance $R1$. The

output voltage e_o is thus very small with respect to the excitation voltage E. Since this circuit is somewhat insensitive it can normally only be used in applications where the change in resistance expected with a change in physical quantity will be greater than five percent of the transducer's nominal resistance, i.e. $\frac{\Delta R}{R} > 0.05$.
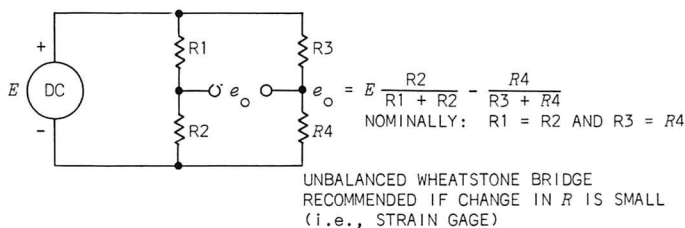
$$e_o = E\left(\frac{R2}{R1 + R2}\right)$$
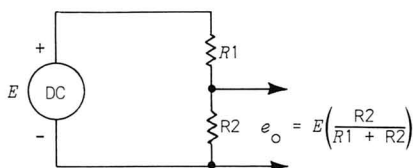
e_o IS PROPORTIONAL TO $R1$ if R2 << $R1$
THUS e_o << E.   NOT RECOMMENDED

(A) CONSTANT VOLTAGE

$$e_o = I(R)$$
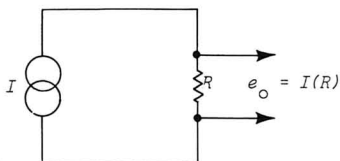
e_o IS PROPORTIONAL TO R
RECOMMENDED IF CHANGE IN R IS GREAT
(i.e., THERMISTOR)

(B) CONSTANT CURRENT

$$e_o = E\frac{R2}{R1 + R2} - \frac{R4}{R3 + R4}$$

NOMINALLY: R1 = R2 AND R3 = $R4$

UNBALANCED WHEATSTONE BRIDGE
RECOMMENDED IF CHANGE IN R IS SMALL
(i.e., STRAIN GAGE)

(C) CONSTANT VOLTAGE BRIDGE

NOTE: R = TRANSDUCER
R = FIXED RESISTOR

Fig. 18-1.   Resistive transducer circuits.

The constant current excitation circuit shown in Fig. 18-1B should also be used in applications where the change in resistance expected with a changing physical quantity is greater than five percent of the transducer's nominal resistance. It does not, however, suffer the nonlinearity due to the presence of a current sampling resistor as does the previous circuit. This circuit requires a high-input impedance-recording circuit but this is not normally an important factor with modern electronics. It derives its excitation from a constant current source which may not be as readily available as a constant voltage source.

current excitation

The Wheatstone Bridge powered from a constant voltage source shown in Fig. 18-1C is recommended in applications where the change in resistance expected with a changing physical quantity will be less than five percent of the transducer's nominal resistance, i.e. $\frac{\Delta R}{R} < 0.05$. This circuit has the advantage over the previous two circuits discussed in that the output voltage is zero when the transducer's resistance is at its nominal value. The bridge may be, however, somewhat nonlinear as discussed in the following section, thus it should be limited to applications where ΔR is small compared to R.

voltage excited Wheatstone Bridge

The feature of providing zero output when the transducer's resistance is at its nominal value is perhaps the Wheatstone Bridge's most desirable characteristic. "Standing" output in the circuits shown in Fig. 18-1A and 18-1B must often be "bucked out" by an offset potential within the amplifier or recording device for the circuits to be useful.

The gage factor of a resistive transducer is the ratio of the relative change in resistance of the transducer to the relative change in the physical quantity being measured. In the particular case of a resistive displacement transducer which, in its simplest form, may consist of a length of resistance wire, displacement can be coupled to the wire to change its length and the gage factor G is computed from the following formula:

gage factor

$$G = \frac{\frac{\Delta R}{R}}{\frac{\Delta l}{l}} = \frac{\Delta R}{R}\frac{l}{\Delta l}$$

Δl and l refers to the change in length of the resistance wire and the original length of this wire respectively. The term $\frac{\Delta l}{l}$ is referred to as strain, symbol ε.

As an example of the above, consider a 1 meter length of resistance wire having a resistance of 100 ohms. If a displacement is mechanically coupled to this wire, the length of the wire may be stretched by one cm to 101 cm. The strain would then be equal to $\frac{1}{100}$ or 0.01. In physical measurement, this strain would be considered large and would be referred to as 10,000 μstrain or 10,000 με. If in stretching the wire, its resistance changes from 100 ohms to, say 102 ohms, then the ratio $\frac{\Delta R}{R}$ is equal to 0.02 and the gage factor G is equal to 2, being the ratio of the resistance change, 0.02, to the strain, 0.01.

determining strain and gage factor

18.2 THE UNBALANCED WHEATSTONE BRIDGE

The Wheatstone Bridge circuit was discussed briefly in the previous section, however further discussion of the characteristics of this circuit is desirable due to its extensive use in transducer measurement systems. The Wheatstone Bridge can be used in either of two distinct modes: as a *balanced* bridge where an unknown resistance is measured by adjusting the value of one of the other resistors in the bridge for zero output, or as an *unbalanced* bridge where an unknown resistance is measured by measuring the output voltage produced by bridge imbalance. The balanced bridge is only used in static measurement situations, however the unbalanced bridge is used in both static and dynamic measurement situations. Most transducer applications use the Wheatstone Bridge in its unbalanced condition.

balanced -- unbalanced bridge

The following paragraph discusses the output voltage obtained from an unbalanced Wheatstone Bridge when one of the resistors comprising the bridge changes its value by ΔR. If only one resistor changes in value, the bridge is said to contain only one active element. This change in value would normally be caused by a change in a physical quantity acting on a transducer.
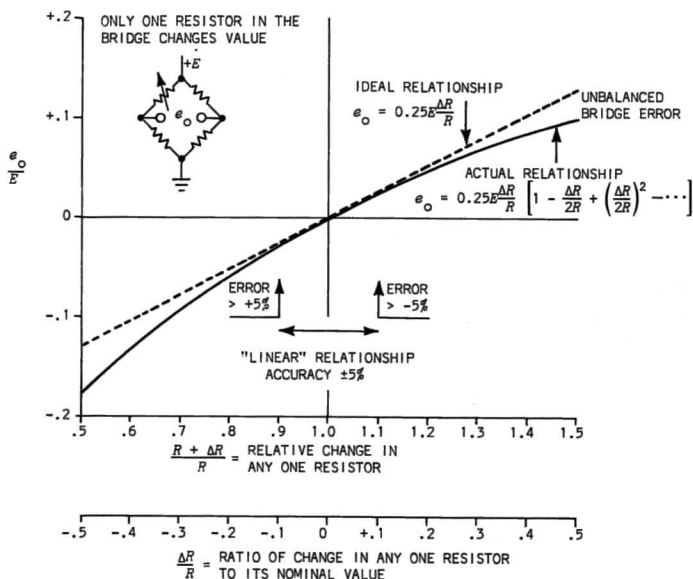
Fig. 18-2. Output voltage from unbalanced Wheatstone Bridge.

The Wheatstone Bridge circuit is shown in Fig. 18-1 together with a formula for determining the output voltage from the circuit. If R1 and R2 are equal, and R3 and $R4$ are equal, e_O will be zero. If the resistance of one of the resistors, say $R4$, is changed in value by ΔR, the output voltage may be computed from the equation given in Fig. 18-1. The algebraic solution of this equation reverts to a complex binomial function; however in most instances, second-order terms can be neglected and the equation for the output voltage can be simplified to:

calculating e_O

$e_O = 0.25\ E\ \dfrac{\Delta R}{R}$. The percentage error is approximately

±5% for ±10% changes in R, i.e., the unbalanced Wheatstone Bridge with one active element is accurate to ±5% for values of $\dfrac{\Delta R}{R}$ from −0.1 to +0.1. Fig. 18-2 shows both the actual output voltage obtained from an unbalanced Wheatstone Bridge with one active element and the "ideal" output voltage computed from the above equation neglecting second-order binomial terms. The graph presented in Fig. 18-2 confirms the accuracy statement above and also shows that for large ratios of $\dfrac{\Delta R}{R}$ the unbalanced bridge with one

active element is substantially in error if the second-order binomial terms are neglected. It should be noted that, if the changing resistive element is a thermistor, the curvature of its characteristic largely corrects the nonlinearity of the bridge over the temperature range of clinical interest.

Many transducer systems are arranged so that a changing physical quantity causes a resistance change in two of the resistors in the unbalanced Wheatstone Bridge circuit or possibly all four of the resistors in the unbalanced bridge circuit. These bridges are then referred to as having two active elements and four active elements, respectively. Re-analysis of the unbalanced Wheatstone Bridge equations will show that, if the physical quantity to be measured has an opposite effect on both active elements in a bridge containing two active elements, then the output is twice the output of a bridge containing only one active element. Similarly, the output for a bridge containing four active elements is four times the output of a bridge containing only one active element. Re-analysis for two or four active elements will also show that the solution of the equation given in Fig. 18-1 is simple and no second order binomial terms are derived; thus no errors are produced by neglecting such terms. The output voltage formulas for the unbalanced Wheatstone Bridge circuit are summarized below:

more than one active element

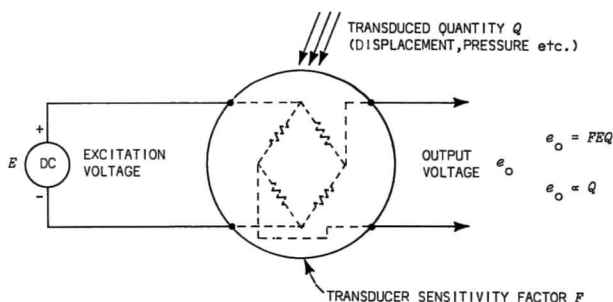If the bridge contains one active element:

$$e_o = 0.25 \ E \ \frac{\Delta R}{R} \text{ accurate to } \pm 5\% \text{ for}$$

$$-0.1 < \frac{\Delta R}{R} < 0.1$$

If the bridge contains two equally active elements:

$$e_o = 0.5 \ E \ \frac{\Delta R}{R} \text{ exactly.}$$

If the bridge contains four equally active elements:

$$e_o = E \ \frac{\Delta R}{R} \text{ exactly.}$$

EXAMPLE: PRESSURE TRANSDUCER WITH SENSITIVITY FACTOR
SUPPLIED WITH TRANSDUCER OF 118μV/V/cmHg PRESSURE.

SAY, CONVENIENT EXCITATION VOLTAGE OF 6V DC

e_o FOR ONE cmHg PRESSURE = FEQ = 118 x 10^{-6} x 6 x 1 = .708mV

THUS, WITH 6.0V OF EXCITATION, SYSTEM GIVES

.708mV OF e_o/cmHg PRESSURE.

ALTERNATIVELY: ADJUST EXCITATION VOLTAGE FOR A MORE CONVENIENT
CALIBRATION FACTOR.

SAY WE WISH 1mV OF e_o/cmHg PRESSURE;

THEN, e_o = FEQ

1×10^{-3} = 118 x 10^{-6} x E x 1

E = 8.49V

THUS, WITH 8.49V OF EXCITATION, THE SYSTEM GIVES

1.0mV OF e_o/cmHg PRESSURE.

THE EXCITATION VOLTAGE MUST BE WITHIN THE TRANSDUCER'S RATING.
FOR MOST TRANSDUCERS THE EXCITATION RANGE IS 5V TO 8V.

Fig. 18-3. Transducer system calibration using sensitivity factor F
supplied with transducer.

18.3 PRACTICAL TRANSDUCER SYSTEMS USING THE UNBALANCED WHEATSTONE BRIDGE

sensitivity
factor F

Practical transducers are self-contained units that
produce an electrical output proportional to the
excitation voltage and the physical quantity to be
measured. Although the transducer may internally
consist of a Wheatstone Bridge having one or more
active elements, it is unnecessary to know the
characteristics of the elements comprising this
bridge if a calibration factor or transducer
sensitivity factor F is supplied with the transducer
at the time of purchase. The transducer sensitivity
factor F is normally given as volts output per volt
of excitation per unit of physical quantity to be
measured. The actual output voltage is determined
as shown in Fig. 18-3.

calculating
F

If the transducer sensitivity factor for a particular
transducer is unknown, it may be determined by
subjecting the transducer to a known amount of the
physical quantity that the transducer is designed to
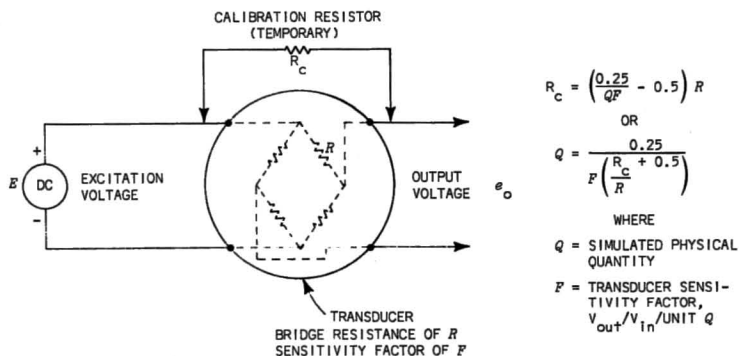measure. If the transducer is then excited from a

known source, and the output voltage measured accurately with a calibrated oscilloscope or voltmeter, the sensitivity factor may be calculated by rearranging the equation shown in Fig. 18-3.

With the above system, an accurate measurement of the physical quantity under investigation depends on the accuracy of the transducer sensitivity factor, the accuracy of the excitation voltage, and the accuracy of the output voltage indication device. Many transducer systems incorporate a calibration resistor as shown in Fig. 18-4. This calibration resistor is chosen to have the same effect on the transducer as would a certain amount of the physical quantity to be measured. It is thus only necessary to connect this calibration resistor as shown in Fig. 18-4 and to adjust the output voltage indication device for a known deflection. This resistor fully "calibrates" the transducer system and it is unnecessary to know the exact value of the excitation voltage or the exact sensitivity of the output voltage indication device. Many transducers are supplied by manufacturers at the time of purchase complete with a calibration resistor. If the transducer sensitivity factor F is known, an appropriate calibration resistor value can be selected as shown in Fig. 18-4.

calibration resistor

$$R_c = \left(\frac{0.25}{QF} - 0.5\right) R$$

OR

$$Q = \frac{0.25}{F\left(\frac{R_c}{R} + 0.5\right)}$$

WHERE

Q = SIMULATED PHYSICAL QUANTITY

F = TRANSDUCER SENSITIVITY FACTOR, V_{out}/V_{in}/UNIT Q

CALIBRATION RESISTOR (TEMPORARY)
R_c

EXCITATION VOLTAGE

OUTPUT VOLTAGE e_o

E DC

R

TRANSDUCER
BRIDGE RESISTANCE OF R
SENSITIVITY FACTOR OF F

BRIDGE RESISTANCE R IS THE NOMINAL RESISTANCE OF EACH OF THE FOUR ARMS OF THE BRIDGE. IT MAY BE OBTAINED FROM THE MANUFACTURER'S DATA SHEET FOR THE TRANSDUCER OR MEASURED WITH AN ACCURATE RESISTANCE BRIDGE BY SIMPLY MEASURING THE RESISTANCE BETWEEN THE EXCITATION TERMINALS WITH THE OUTPUT TERMINALS OPEN CIRCUITED OR VICE VERSA.

EXAMPLE: PRESSURE TRANSDUCER SENSITIVITY FACTOR SUPPLIED WITH TRANSDUCER OF 118µV/V/cmHg PRESSURE -- BRIDGE RESISTANCE 350Ω

WE WISH TO SIMULATE A PRESSURE OF 10cmHg:

$$R_c = \left(\frac{0.25}{QF} - 0.5\right) R = \left(\frac{0.25}{10 \times 118 \times 10^{-6}} - 0.5\right)350 = 74.2k\Omega$$

Fig. 18-4. Simulation of transduced quantity with a calibration resistor.
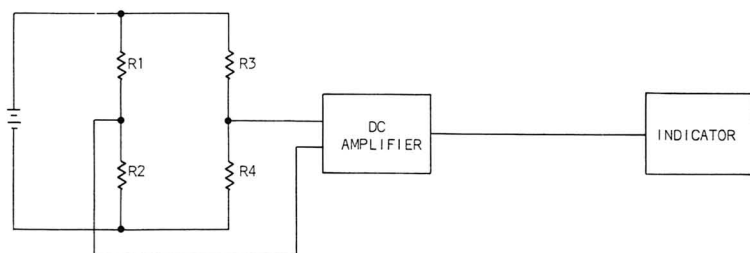
18.4 AC- AND DC-BRIDGE SYSTEMS
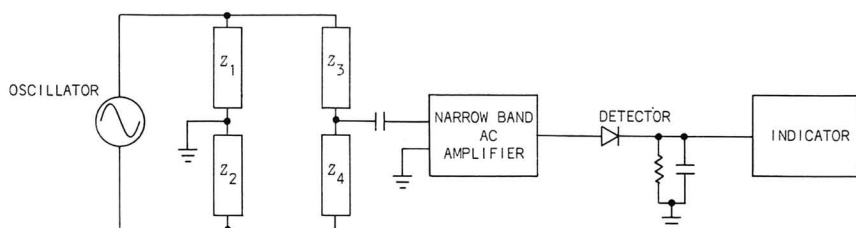
DC-bridge
systems

The previous discussion applies to transducer
systems having either DC or AC excitation sources.
In practice, both DC-bridge and AC-bridge systems
are used. The DC-bridge systems shown in Fig. 18-5
simply consist of a battery or DC power supply as
the excitation source and incorporate a DC-coupled
amplifier in the output voltage indication system.
Since the excitation is DC, this system cannot be
used with reactive transducers; that is, transducers
where the active elements consist of inductors or
capacitors rather than resistors. Until recently
this DC-bridge system was not preferred for high-
accuracy or high-sensitivity measurements due to the
relative instability of DC-coupled amplifiers.
Stable DC-coupled amplifiers, such as the Tektronix
Type 3A9 Differential Amplifier, are now available
and thus DC-excited bridge systems are regaining
usage.

simple AC-
bridge
systems

Perhaps the most economical bridge system is the
simple AC-bridge system shown in Fig. 18-5. With
this system, the bridge is excited by a sinewave
oscillator, normally within the audio-frequency range,
and the output is amplified via either a wide-band
AC-coupled amplifier or, preferably, a narrow-band
AC-coupled amplifier. Since the amplifier is AC
coupled, DC stability is unimportant. The amplified
bridge output signal is detected with a simple
detector, however, as this detector is not phase
sensitive, this simple AC-bridge system will provide
essentially "positive" output for both "positive"
and "negative" changes away from the balance
condition in the physical quantity to be measured.
Often, however, the nature of this physical quantity
overcomes this apparent ambiguity.

(A) DC SYSTEM

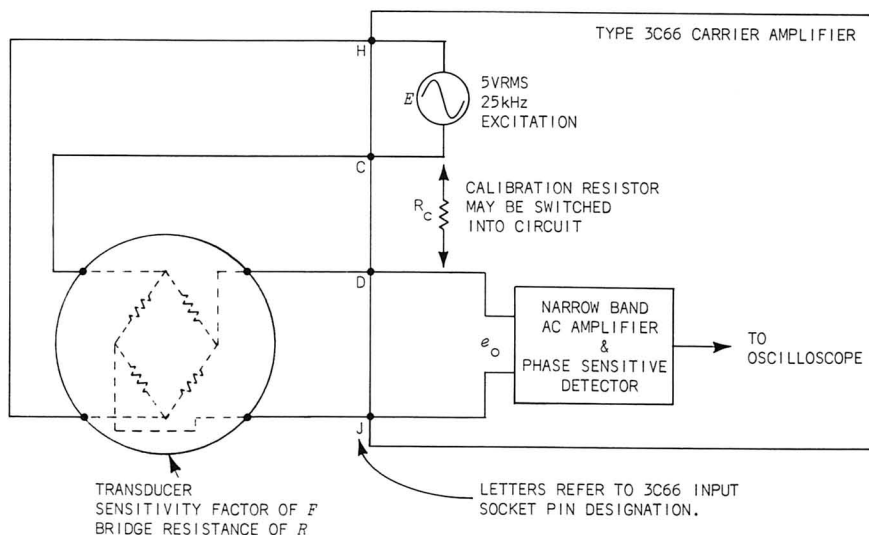(B) SIMPLE AC SYSTEM
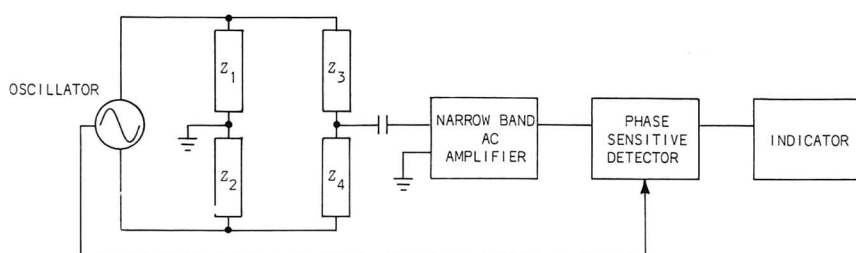
(C) PHASE SENSITIVE AC SYSTEM

Fig. 18-5. DC and AC bridge systems.

The detector used in the simple AC system discussed in the previous paragraph may be replaced with a phase-sensitive detector that solves the apparent ambiguity by producing both positive and negative outputs. Such a system is referred to as a phase-sensitive AC system or a carrier amplifier system as shown in Fig. 18-5. The Tektronix Type 3C66 plug-in unit for the Type 561B Oscilloscope or 564B Storage Oscilloscope employs this phase-sensitive AC-bridge system.

The Tektronix Type 3C66 Carrier Amplifier was primarily intended for use with strain gages and is thus calibrated in microstrain or $\mu\varepsilon$. This instrument does, however, incorporate switching circuitry to allow a built-in 150-kΩ shunt calibration resistor to be used. Thus the μ STRAIN/DIV switch may be regarded as a variable gain control on the transducer's output voltage detection circuitry and a calibration resistor may be used to calibrate the system as discussed in Section 18.3. The Tektronix Type 3C66 Carrier Amplifier system is shown in Fig. 18-6. The built-in 150-kΩ shunt calibration resistor in the Type 3C66 may be replaced with any desired value of calibration impedance. The calibration impedance must be of the same form as the impedance that is changing in the transducer, thus resistive transducers require calibration resistors, inductive transducers require calibration inductors and capacitive transducers require calibration capacitors.

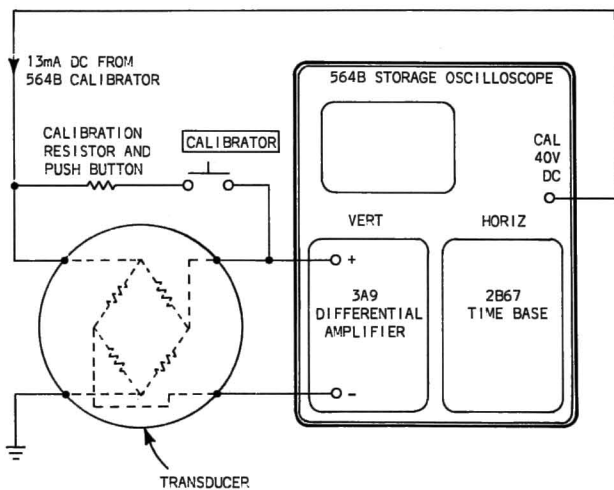Fig. 18-6. Tektronix Type 3C66 Carrier Amplifier.

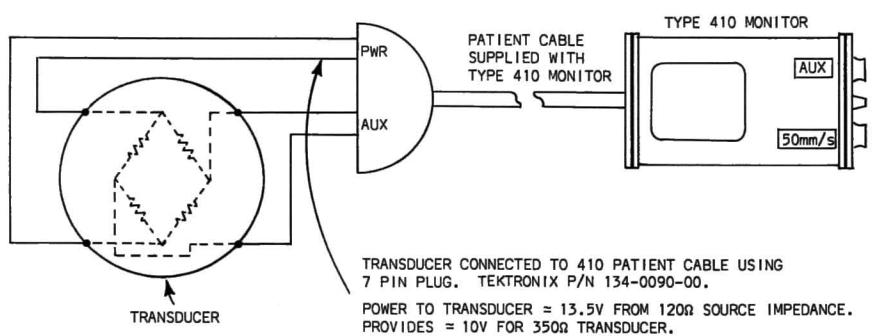**Fig. 18-7. Transducer system using conventional oscilloscope and plug-in units.**

Fig. 18-8. AC coupled transducer system with Type 410 Monitor.

564B
system

A DC-transducer system may be fabricated using a
Tektronix Type 564B Storage Oscilloscope with a
3A9 Vertical Amplifier and 2B67 Time Base unit as
shown in Fig. 18-7. This system excites the
transducer from an approximate 13 mA current source,
however since the transducer's impedance is
essentially constant, the voltage between the
transducer's excitation terminals will be constant
and the system will operate as if it were excited
from a constant voltage source. This system is
suitable for use with transducers having a nominal
bridge resistance of between 100 ohms and 2,000 ohms.

410
system

The transducer system shown in Fig. 18-8 may be used
in some limited applications where the physical
quantity to be measured is essentially an AC signal.
This system is thus suitable for recording the
difference between diastolic and systolic pressures
using a pressure transducer, however since the
Type 410 is insensitive to DC signals, it will not
show the absolute values of these pressures.

18.5 DISPLACEMENT TRANSDUCERS

A displacement transducer converts linear or
rotational displacement into an electrical output.
Various forms of resistive displacement transducers
are shown in Fig. 18-9. Each of these transducers
changes resistance due to displacement; this
resistance change is then converted to an electrical
output by using any one of the circuits shown in
Fig. 18-1.

potenti-
ometer

Perhaps the simplest resistive displacement
transducer is the common potentiometer. The
resistance of the potentiometer changes with
rotational displacement which may be obtained from a
linear displacement via either a lever system or a
pulley. This type of resistive displacement
transducer is commonly used in spirometers as
discussed in Section 9.4.

fluid-filled   The fluid-filled rubber tube shown in Fig. 18-9
rubber tube   changes its cross-sectional area and length due to
                a linear displacement or force acting on the rubber
                tube. This changing cross-sectional area and length
                causes a change in the resistance of a conductive
                fluid within the tube as measured between each end
                of the tube. Mercury has historically been used as
                the conductive fluid in this type of transducer,
                however the resistance of mercury is extremely low
                and special techniques are required to detect any
                change in this resistance. Fluids with much higher
                resistivities are preferred and are indeed used in
                many commercial fluid-filled rubber-tube transducers.

                The three strain-gage resistive-displacement
                transducers shown in Fig. 18-9 represent the basis
bonded   for most modern transducers. The bonded strain
strain   gage consists of a length of fine resistance wire
gage   bonded to a backing of paper or epoxy. As linear
                displacement is coupled to this backing material, it
                stretches and the resistance of the wire bonded to
unbonded   this material therefore increases. The unbonded
strain   strain gage is somewhat similar to the bonded strain
gage   gage with the exception that the resistance wire is
                self supporting as shown in Fig. 18-9. The bonded
                semiconductor strain gage is also somewhat similar
                to the bonded strain gage with the exception that
                the resistance wire element is replaced with a
                filament of semiconductor material, usually silicon.
semiconductor   The principal advantage of the bonded semiconductor
strain   strain gage is its high gage factor, in the order of
gage   120 compared to 2 or perhaps 3 for the bonded or
                unbonded resistance-wire strain gages.

displacement   With any of the displacement transducers shown in
inertia   Fig. 18-9, the mechanical source providing the
                displacement must also provide some force to overcome
                inertia in the transducer. The fluid-filled tube may
                require considerable force in conjunction with the
                linear displacement to be able to stretch the rubber
                tube. The unbonded strain gage requires some force
                to stretch the resistance wires comprising the gage.
                Such devices should more correctly be classed as
                force transducers that may be used for displacement
                measurements if the force driving the system is
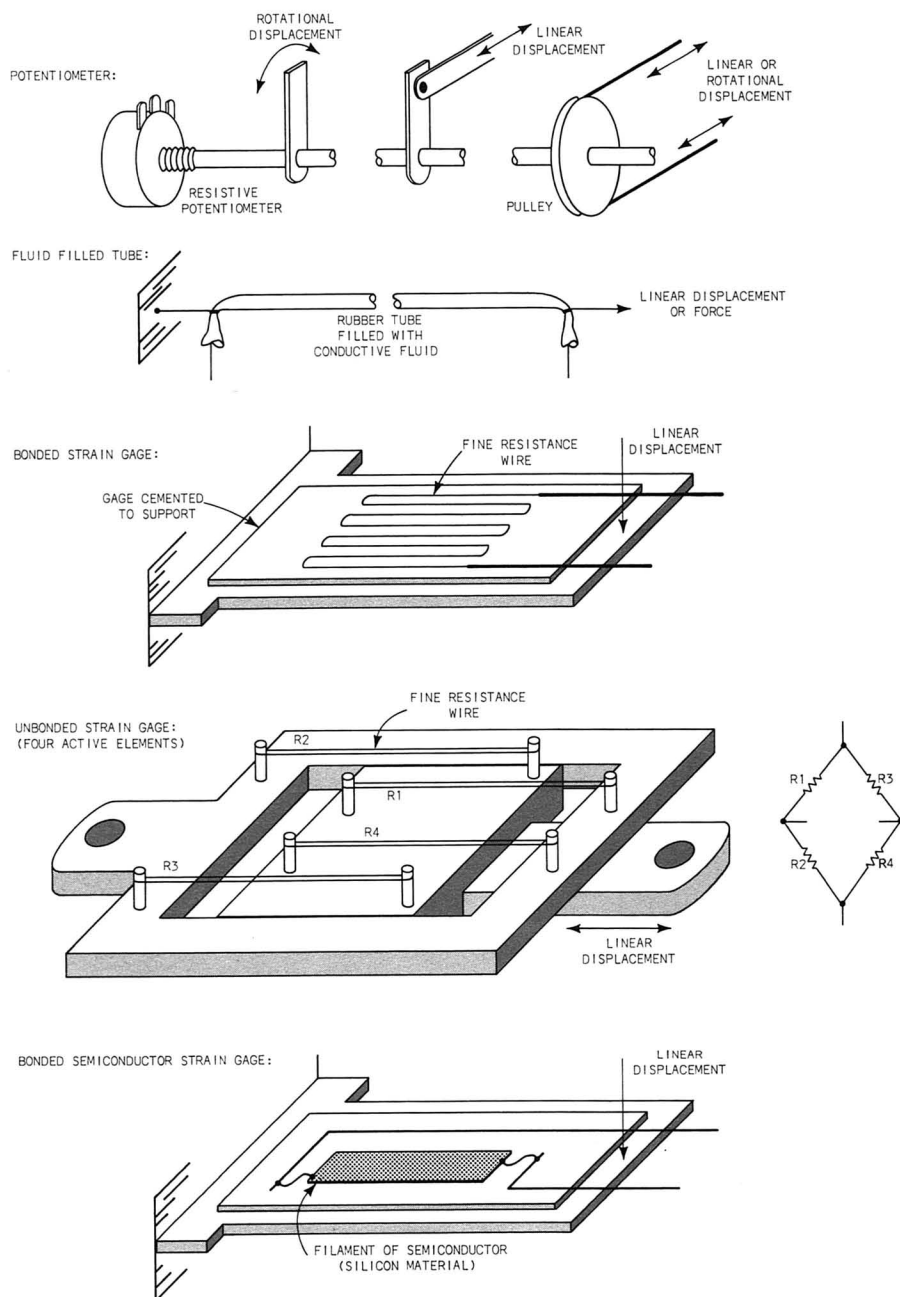   large compared with that required to drive the
                transducer.

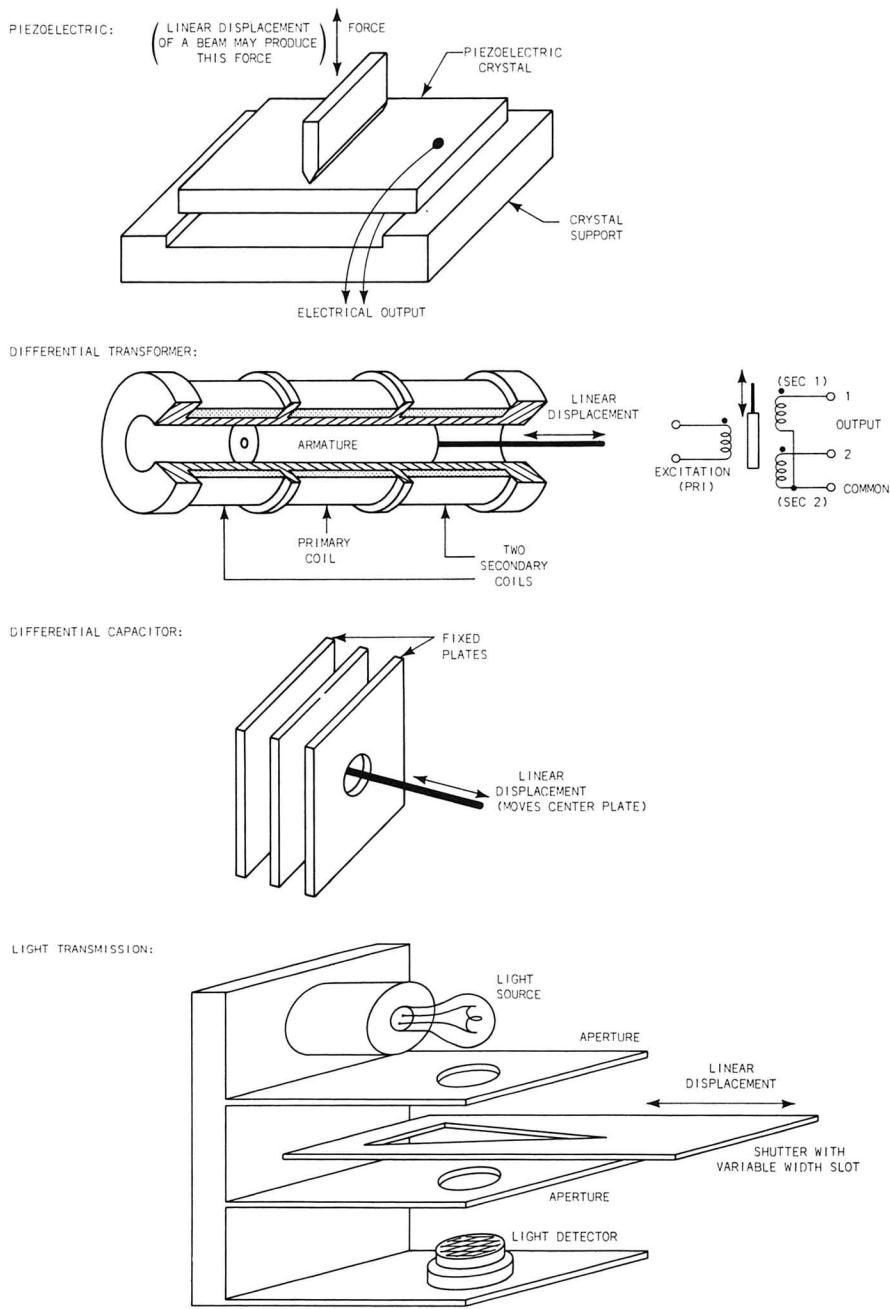Fig. 18-9. Resistive displacement transducers.

Fig. l8-l0. Nonresistive displacement transducers.

Various forms of nonresistive displacement
transducers are shown in Fig. 18-10. The
piezoelectric force transducer produces an electric
charge between the faces of a solid crystal when this
crystal is subjected to bending which may be caused
by displacement of a beam, etc. A principal
advantage of the piezoelectric crystal transducer is
that no excitation source is required, however
displacement does produce a charge rather than a
voltage and this charge is normally quickly
dissipated by the input impedance of the amplifier
used in conjunction with this transducer. It is thus
primarily suited to the measurement of AC quantities,
however the transducer can be used in conjunction
with a charge amplifier to provide good low-frequency
response characteristics. A piezoelectric
displacement transducer is used in the Korotkoff
sound microphone shown in Chapter 8, Fig. 8-10.

*piezo-
electric*

A differential-transformer displacement transducer
is shown in Fig. 18-10. The differential transformer
consists of three inductors coupled magnetically by
a common armature. One of these three inductors is
excited from an AC source (primary winding) and
this excited winding induces voltages into the other
two windings (secondary windings), which are
connected in opposition. The induced voltage is
dependent on the amount of coupling and thus on the
relative position of the armature. With the armature
coupling energy equally from the primary winding to
each of the secondary windings, voltages of equal
amplitude appear across each winding. When the core
is moved off center, however, each secondary is not
linked identically. One secondary produces a larger
voltage than the other and thus the potentials
produced in the secondaries do not cancel and an
output voltage is obtained. The two secondary
windings of a differential transformer may be
considered as two elements of an unbalanced
Wheatstone Bridge with the other two elements
consisting of fixed resistors. The differential
transformer is particularly suited for use with
either of the AC-bridge systems shown in Fig. 18-5.
They are also available in DC-to-DC versions with
all the necessary electronics built into the actual
transducer.

*differential
transformer*

288

differential   The differential-capacitor displacement transducer
capacitor   shown in Fig. 18-10 essentially consists of two
capacitors arranged so that a common plate for both
of these capacitors is coupled to a linear
displacement. Any movement of this common plate
increases the capacitance in one capacitor while
simultaneously decreasing the capacitance in the
other. These two capacitors may be used to form
two elements in an AC Wheatstone Bridge system with
the other two elements consisting of fixed resistors.

The light-sensitive displacement transducer shown in
Fig. 18-10 employs a variable-width slot coupled to
the displacement. Light is passed through this slot,
the width of the slot determining the amount of
light that can pass. The varying light intensity is
normally detected with a light-sensitive transducer
as discussed in Section 18.7.

18.6   FORCE, PRESSURE AND ACCELERATION TRANSDUCERS

force   Force, pressure and acceleration transducers are
related to displacement transducers as shown in
Fig. 18-11. A force transducer converts force to
linear displacement using a spring or some other
resilient material. Force on the transducer produces
stress in the sensing element. This stress causes
strain and the strain is then detected with a strain
gage displacement transducer.

pressure   The pressure transducer shown in Fig. 18-11 is
basically a force transducer with fluid pressure
being converted to force via a diaphragm. Since
the force produced is proportional to the fluid
pressure, the transducer's resistance change is
proportional to the fluid pressure.

acceleration The acceleration transducer shown in Fig. 18-11
simply consists of a force transducer coupled to a
known mass. This mass converts an acceleration to
a force. Although the acceleration transducer shown
in Fig. 18-11 is sensitive to acceleration in one
direction only, modifications to this principal
allow acceleration transducers sensitive to
acceleration in any direction to be constructed.
The weight transducer shown in Fig. 18-11 is
essentially a force transducer with the force being
proportional to the weight of a body applied to the
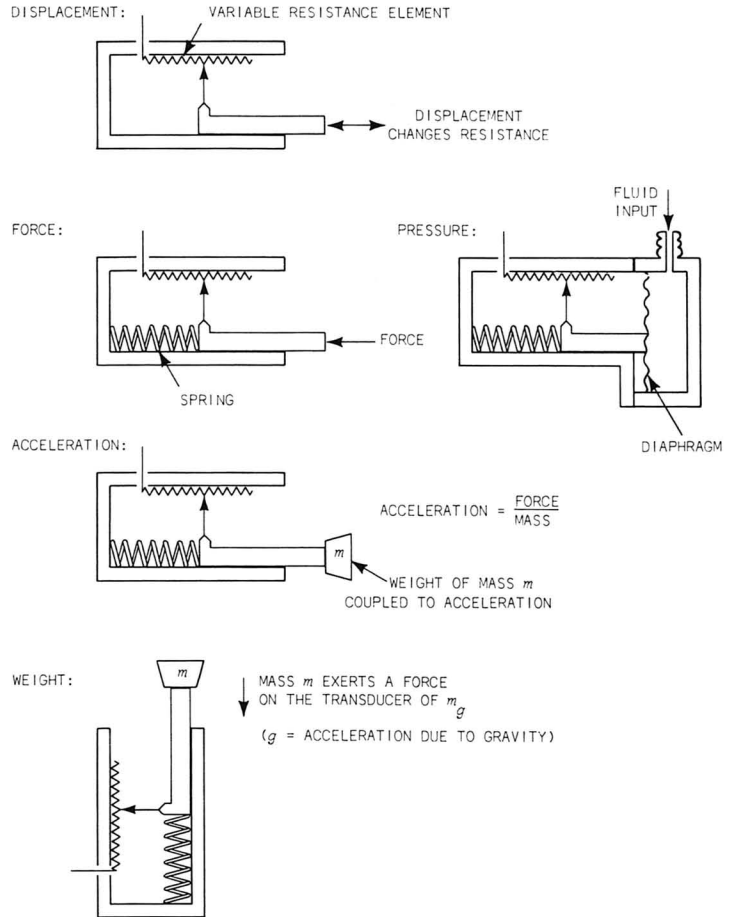transducer and to the gravity constant g.

Fig. 18-11. Transducer evolution.

velocity
pickups

Velocity pickups, consisting simply of a coil located in a magnetic field, are commonly used for vibration pickups and in dynamic microphones. Displacement and acceleration may be obtained from the output of a velocity transducer by integrating or differentiating this output voltage. Conversely, displacement can be measured with an acceleration transducer by double integration of the output voltage and acceleration can be measured with a displacement transducer by double differentiation.

The mathematical relations involved are shown here.

Differentiation processing:

$$v = \frac{dx}{dt}$$

$$a = \frac{dv}{dt} = \frac{d^2x}{dt^2}$$

x = Displacement
v = Velocity
a = Acceleration
t = Time

Integration processing:

C = Constant

$$v = C1 \int a\,dt$$

$$x = C2 \int v\,dt = C1C2 \int a\,dt$$

For sinusoidal motion, the above may be simplified to:

$$v_{peak} = 2\pi f x_{peak}$$

$$a_{peak} = (2\pi f)^2 x_{peak}$$

f = Frequency of sinusoidal motion

Calibration of systems using this indirect method can be difficult, thus these methods are primarily used when uncalibrated or relative measurements are satisfactory. Double differentiation can often result in excessive noise and thus should be avoided.

Although force, pressure, acceleration and weight transducers are all used in biophysical measurements, the pressure transducer is of particular interest due to its extensive use for monitoring the activity of the circulatory system and respiratory system as discussed in Chapters 8 and 9. An important characteristic of a pressure transducer used for biophysical measurements is its volume displacement, that is, the amount of fluid that must flow into the pressure transducer to produce movement of the diaphragm within the transducer. Volume displacement

in pressure transducers is normally measured in cubic millimeters per 100 mm Hg pressure change. Volume displacements range from below .01 cubic mm per 100 mm Hg pressure to above 20 cubic mm/100 mm Hg pressure. Transducers with high volume displacements normally have poor high-frequency response characteristics due to the finite time required to move the large volume of fluid involved. Transducers for blood pressure measurement should have a volume displacement of less than .04 cubic mm/100 mm Hg pressure.

18.7 TRANSDUCERS FOR NONMECHANICAL QUANTITIES

The range of physical quantities that can be measured with the aid of a transducer is almost limitless, as is the range of transducers that may be useful in the biophysical measurement sciences. Transducers are used extensively in the biophysical sciences to measure the following nonmechanical quantities: temperature, light, radiation, liquid chemical composition and gaseous chemical composition. These transducer types are discussed briefly below, however more detailed information is available from the principal transducer manufacturers; perhaps some of the more prominent are Statham Instruments, Inc., Oxnard, California; Beckman Instruments, Inc., Palo Alto, California; E & M Instrument Company, Inc., Houston, Texas and many others. An excellent reference for the application of transducers to biomedical instrumentation is Chapters 1 through 9 of *Principles of Applied Biomedical Instrumentation* by L. A. Geddes and L. E. Baker, published by John Wiley and Sons, Inc., New York, N.Y.

thermistor

Temperature transducers can be divided into two categories; temperature dependent resistors and thermoelectric generators. The thermistor is a temperature dependent resistor whose resistance decreases as temperature is increased. A typical thermistor may be expected to produce a one percent change in resistance for a temperature change of one or two degrees within normal environmental temperature ranges. For the resistance change to be detected, some form of resistance detection circuitry is necessary. It is important that the power level involved in this circuitry does not cause self-heating within the thermistor.

thermocouple

A thermocouple consisting of a junction between two dissimilar metals is a thermoelectric generator, however, this form of temperature transducer is rarely used for biophysical measurements due to its relative insensitivity compared to thermistors.

Optical transducers sensitive to light are used in the mechanical displacement transducer shown in Fig. 18-10 and in the plethysmographs shown in Chapter 8. Although many different forms of optical transducers are available, the most common type used in biophysical measurements is the light-sensitive resistor. Light-sensitive resistors, or photo resistors, consist of a thin film of resistive material deposited on the inside of an evacuated glass or plastic chamber. The most commonly used material is cadmium sulphide, however, cadmium selenide and semiconductor materials are used in certain applications. The response of these photo resistors is reasonably linear; the resistance decreases proportionally with an increase in the level of illumination. A typical "dark" resistance would be above 10 megohms, this resistance would decrease to a few kilohms with normal room illumination of about 100 foot-candles.

light sensitive resistor

Radiation transducers are used in biophysical measurements to check for the possibility of a radiation hazard. Radiation transducers may detect one or more of the three principal forms of radiation: alpha particles, beta particles and gamma rays. Gamma radiation is usually considered the most dangerous because of its higher energy content. Radiation transducers either detect radiation directly or detect light emitted by certain crystals when exposed to radiation. Perhaps the most common form of radiation detector is the Geiger-Muller counter which, when subject to radiation produces ionization in a sealed gas-filled tube which allows current to flow between electrodes placed in the tube.

radiation detection

Liquid chemical transducers are used in biophysical measurements to provide a continuous chemical analysis of body matter, particularly blood. Liquid chemical transducers are used to measure the pH or relative acidity of body fluids and are also used for routine blood analysis to determine the amount of dissolved oxygen or carbon dioxide in the blood. In addition, liquid chemical transducers are available to determine the relative amounts of potassium and sodium ions in body fluid. Gaseous chemical transducers are used to analyze the gaseous content of expired air, particularly the amounts of oxygen and carbon dioxide, and also are used as oxygen detectors in atmospheric warning systems.

chemical detection

19

AMPLIFIERS

Physiological signals acquired by either electrodes
or transducers are typically below 10 mV in amplitude
and must therefore be amplified to be compatible
with display devices and recorders. This chapter
discusses the characteristics of amplification
systems as seen by electrodes. The discussion will
be limited to the process of amplification, additional
signal processing is discussed in Chapter 20. In
practice, most physiological amplifiers include both
amplification and signal processing within the one
package. For the purpose of this discussion,
therefore, an amplifier is defined as a device having
a high input impedance and a low output impedance
and providing either a fixed or perhaps a variable
voltage gain. The amplifier may or may not contain
DC-offset capabilities to offset the DC level of the
output with respect to the input DC level.
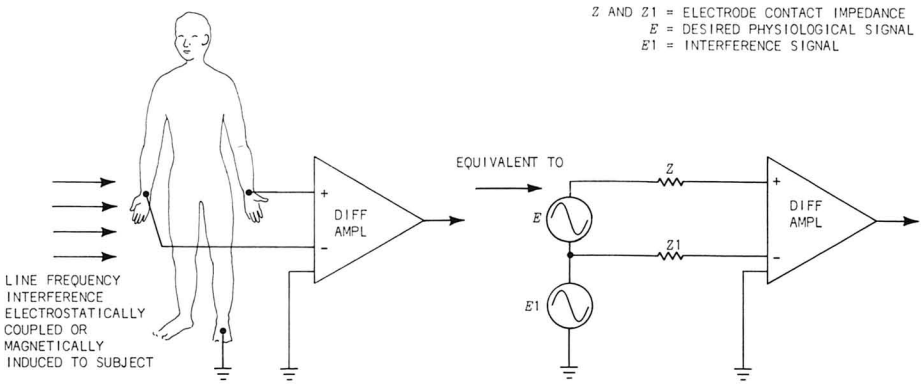
Fig. 19-1. Typical physiological measurement configuration.

19.1 THE DIFFERENTIAL AMPLIFIER USED WITH ELECTRODES

desired
and
interference
signal

electrode
contact
impedances

differential
and common
mode signal

A typical physiological measurement configuration is shown in Fig. 19-1. In almost all physiological measurement situations, two signals are produced by the subject: a desired physiological signal such as the ECG or EEG and an interference signal, typically 60 Hz, due to electrostatically coupled or magnetically induced interference. The purpose of the differential amplifier is to reject this interference signal and to amplify the desired signal. Two impedances, Z and Z1, representing electrode contact impedance are shown in Fig. 19-1. The source of these impedances is discussed in Chapter 16 when referring to the characteristics of various electrode systems. Z and Z1 will normally lie in the range from 1,000 ohms to 50,000 ohms and it is possible for the difference between Z and Z1 in any one system to be as great as 10,000 ohms. Although all modern differential amplifiers have input impedances above 1 megohm, these finite impedances Z and Z1 cannot be ignored as will be seen later in this chapter. The desired physiological signal shown in Fig. 19-1 is connected so that it appears between the two input terminals of the differential amplifier and is referred to as the differential signal. The interference signal shown appears between both inputs of the differential amplifier and ground and is referred to as the common mode signal.

19.2 COMMON MODE REJECTION

CMRR

As stated in the previous paragraph, the purpose of a differential amplifier is to reject common mode signals. The "common mode rejection ratio" or CMRR of an amplifier is its relative ability to reject common mode signals. The common mode signal is defined as the signal that is applied to both inputs in the same phase, that is:

$$\text{common mode signal} = \frac{\left(\begin{array}{c}\text{signal to}\\ \text{+ input}\end{array}\right) + \left(\begin{array}{c}\text{signal to}\\ \text{- input}\end{array}\right)}{2}$$

Common mode rejection ratio is defined as the ratio between the amplitude of the common mode signal to the amplitude of an equivalent differential signal that would produce the same output from the amplifier. As an example, consider a differential amplifier with a single 1-V source connected from ground to both inputs of the amplifier, that is, in common mode. If the output signal is equivalent to that produced by a 10 μV differential signal, the common mode rejection ratio would then be the ratio of this apparent differential signal to the true common mode signal; that is, 1V:10μV or 100,000:1. Common mode rejection is often specified in decibels; a CMRR of 100,000:1 being referred to as a common mode rejection of -100 dB. The relationship is --

Common Mode Rejection in dB = 20 $\log_{10}$ CMRR

The common mode rejection ratio calculated previously was obtained by applying the same one volt common mode signal to both inputs of the differential amplifier. Let us now consider the case where an impedance of 150 ohms is inserted between the one volt source and the positive input of the differential amplifier with the negative input of the differential amplifier connected directly to the source as previously. Considering the positive input circuit and assuming an input impedance to ground from this positive input of 10 megohms, the 150-ohm resistor in conjunction with the 10-megohm input resistance of the differential amplifier will form a small but nevertheless highly significant voltage divider. Now, instead of the full one volt of common mode signal reaching the differential amplifier, 15 μV appear across the 150-ohm resistor and only one volt minus 15 μV or .999985 volts reach the positive input of the amplifier. Since no resistor is inserted in series with the negative input, the full one volt common mode signal reaches this negative input. A common mode signal is therefore $\frac{1 + .999985}{2}$ = .9999925 volts. The signal at the +input is 7.5 μV less than this and the signal at the -input is 7.5 μV greater than this. The additional 15 μV on the negative input with respect to the positive input is a differential signal and will be amplified as a differential signal. Assuming a gain of 20,000, this 15 μV will appear as 300 mV at the output.

CMRR versus source impedance

<div style="float:left;">source
impedance
reduces
CMRR</div>

It is thus apparent that the amplifier described in the preceding paragraph with a CMRR of 100,000:1 appears to produce an additional output due to the 150-ohm "source impedance unbalance." If 200 mV of common mode signal appear at the output due to the CMRR of the amplifier and 300 mV of common mode signal appear at the output due to source-impedance unbalance, a maximum of 500 mV of common mode signal may appear at the output. These two signals would normally not add directly, but being AC signals, they must be added algebraically and it is conceivable that they may in fact tend to cancel each other. An equivalent differential signal that would produce this same output would be $\frac{500 \text{ mV}}{20,000}$ or 25 µV. The apparent CMRR is then the ratio of this equivalent differential signal to the true common mode signal; that is .9999925V:25µV or $\simeq$ 40,000:1. It is thus apparent that the 150-ohm source-impedance unbalance has reduced the CMRR of the amplifier from 100,000:1 to an apparent CMRR of 40,000:1. As these signals add algebraically and may tend to cancel each other, 40,000:1 will be a worst-case CMRR. In practice the apparent CMRR may be increased by the source impedance unbalance.

<div style="float:left;">source
impedance
unbalance</div>

Extension of the previous discussion will show that it is not the absolute value of the source impedance that determines this apparent reduction in CMRR but rather the difference between the source impedances (source impedance unbalance) between the positive and negative inputs of the differential amplifier. As stated in the preceding section, it is possible for a source-impedance unbalance of up to 10,000 ohms to occur when recording physiological signals. It is thus desirable either to attempt to minimize this unbalance, or to increase the input impedance of the differential amplifier to the common mode signal in an effort to increase the ratio between the source impedance unbalance and the common mode input impedance of the differential amplifier.

<div style="float:left;">output has
three
major
components</div>

From the preceding discussion it is apparent that the output of any differential amplifier will consist of three components --

1. A desired output component due to amplification of a differential signal by the differential amplifier,

2. An undesired component due to incomplete rejection of common mode interference signals by the differential amplifier and,

3. An undesired component due to source impedance unbalance allowing a small proportion of a common mode signal to appear as a differential signal to the amplifier.

These three output components are shown in Fig. 19-2 together with the equations used for determining the values of each of these three output-signal components.

fourth
component

When a differential amplifier is used on a subject there will also be a fourth output; an undesired output component due to amplification of an *undesired* differential input from the subject, due to the subject's finite resistance between the input terminals and the presence of AC interferring currents through this resistance. This fourth output limits the useful rejection ratio of an ECG amplifier to about 100,000:1.

calculation
of output
components

Referring back to the amplification system discussed earlier in this section, this system is illustrated in Fig. 19-3 with the one volt common mode signal consisting of a one volt P-P, 60-Hz sinewave and a desired differential signal consisting of a 100 μV P-P, 2-Hz sinewave. Applying the formula shown in Fig. 19-2 to this amplification system gives a desired output of 2 volts of 2-Hz sinewave and an undesired output of 0.5 volts of 60-Hz sinewave. This undesired output consists of 0.2 volts due to the inherent CMRR of the amplifier and 0.3 volts due to the 150-ohm source-impedance unbalance of the system.

When measuring the CMRR of modern amplifiers, it is necessary to use about one volt of common mode signal so that the output due to this signal can be discerned from noise. (1 V common mode with 100,000:1 CMRR is equivalent to a 10 μV differential input signal.) Typical common mode signals in biophysical measurements are from 1 mV to 100 mV. The CMRR to these signals will be slightly greater than the CMRR to a one volt signal due to small nonlinearities in the amplifier's input devices.

DIFFERENTIAL
AMPLIFIER

GAIN - A
INPUT IMPEDANCE - Z_{in}
CMRR - X:1

OUTPUT

THREE COMPONENTS

DESIRED COMPONENT
DUE TO PHYSIOLOGICAL
SIGNAL E

UNDESIRED COMPONENT
DUE TO INTERFERENCE
SIGNAL $E1$ AND CMRR
OF AMPLIFIER

UNDESIRED COMPONENT DUE TO
INTERFERENCE SIGNAL $E1$ AND
ELECTRODE CONTACT RESISTANCES
Z, $Z1$

IF $Z_{in} \gg Z$ AND $Z1$ AND $E1 \gg E$

$$\text{OUTPUT} = AE + \frac{AE1}{X} + AE1\left(1 - \frac{Z_{in}}{Z_{in} + Z - Z1}\right)$$

$$\text{OUTPUT} = AE + AE1\left(\frac{1}{X} + 1 - \frac{Z_{in}}{Z_{in} + Z - Z1}\right)$$

THE COMPONENT OF THE OUTPUT
IS THE UNWANTED OR COMMON MODE
SIGNAL REFERRED TO THE INPUT
OF THE DIFFERENTIAL AMPLIFIER

Fig. 19-2. Common mode rejection.

DIFFERENTIAL INPUT
100µV OF 2Hz SINEWAVE

OUTPUT

1V/DIV

150Ω

DIFF
AMPL

COMMON MODE INPUT
1V OF 60Hz SINEWAVE

E

$E1$

$A = 20,000$
$CMRR = 100,000:1$
$Z_{in} = 10M\Omega$

A - IDEAL
B - ACTUAL WITH NO SOURCE Z UNBALANCE
C - ACTUAL WITH 150Ω SOURCE Z
UNBALANCE

$$\text{OUTPUT} = AE + AE1\left(\frac{1}{X} + 1 - \frac{Z_{in}}{Z_{in} + Z - Z1}\right)$$

$$= 20,000 \times 100\mu V + 20,000 \times 1V\left(\frac{1}{100,000} + 1 - \frac{10 \times 10^6}{10 \times 10^6 + 150 - 0}\right)$$

$$= \text{2V OF 2Hz} + \text{0.2V OF 60Hz} + \text{0.3 OF 60Hz SINEWAVE}$$

| = 2V OF 2Hz SINEWAVE | + 0.2V OF 60Hz SINEWAVE DUE TO CMR OF AMPLIFIER | + 0.3 OF 60Hz SINEWAVE DUE TO 150Ω INPUT Z UNBALANCE |

Fig. 19-3. Example of common mode rejection calculation.

The principal common mode signal encountered in biophysical recording is line frequency interference at either 50 Hz or 60 Hz. Let us now examine the validity of inserting coupling capacitors in series with the differential amplifier inputs to provide an AC-coupled amplifier. In physiological situations these capacitors may be intended to block the DC-offset potentials encountered in the electrode system. Assuming a low frequency response to 0.1 Hz with 10-megohm input resistors, then 0.16 µF coupling capacitors are required. The capacitive reactance of 0.16 µF at 60 Hz is 16,700 ohms. When considering this impedance in the light of the earlier CMRR calculations, it becomes obvious that ordinary 10% tolerance capacitors will have a substantial impedance unbalance to this 60 Hz and thus will prevent a high common mode rejection ratio from being obtained. Even if the coupling capacitors are matched to ±1%, the reduction in CMRR due to their imbalance will be significant. It is thus desirable not to use coupling capacitors at the inputs of a differential amplifier but to use them in later amplification stages after common mode signals have been rejected.

coupling
capacitor
unbalance

19.3   INPUT RESISTANCE

input
resistance
to common
mode and
differential
signals

For any differential amplifier there are two significant input resistance values: the input resistance of the amplifier to a differential signal and the input resistance to a common mode signal. From the preceding discussion it is clear that the input resistance to common mode signals (common mode input resistance) should be as high as possible, definitely above 10 megohms and preferably above 100 megohms. Relating back to the discussion on electrodes in Chapter 16, electrode offset potentials are stabilized over a period of time due to the loading effect of the amplifier; that is, the input resistance of the amplifier to differential signals will tend to discharge the "batteries" formed at the electrode-tissue interfaces. It is thus desirable that a differential amplifier's input resistance to differential signals (differential input resistance) be of a value that will allow electrode stabilization within two or three minutes after application. It is undesirable however, to use too low a differential input resistance as too low resistance produces a distorted pulse response to a physiological signal as discussed in Section 16.1.

In practice, a differential input resistance of
2 megohms has proved satisfactory and could perhaps
be regarded as an optimum value for most physiological
recording or from surface electrodes.

Referring to Fig. 19-4, the input resistance
configuration of differential amplifiers determines
the ratio between the differential input resistance
and the common mode input resistance. The input
resistance configuration shown in Fig. 19-4A uses
separate input resistors to ground for each input.

DIFFERENTIAL AMPLIFIER WITH SEPARATE
INPUT RESISTANCE ON EACH INPUT

TO GATE OF
INPUT DEVICE (FET)

DIFFERENTIAL INPUT RESISTANCE = $2R$
COMMON MODE INPUT RESISTANCE = $0.5R$

(A)

DIFFERENTIAL AMPLIFIER WITH SEPARATE
INPUT RESISTANCE ON EACH INPUT AND AN
ADDITIONAL RESISTANCE TO REDUCE THE
DIFFERENTIAL INPUT RESISTANCE

DIFFERENTIAL INPUT RESISTANCE = $\frac{\cdot \times R1}{2R + R1}$
COMMON MODE INPUT RESISTANCE = $0.5R$

(B)

DIFFERENTIAL AMPLIFIER WITH AN INPUT
RESISTANCE ON EACH INPUT SHARING A
COMMON RESISTANCE TO GROUND

DIFFERENTIAL INPUT RESISTANCE = $2R$
COMMON MODE INPUT RESISTANCE = $0.5R + R2$

(C)

Fig. 19-4. Common mode and differential input resistance.

This configuration gives a high differential input
resistance with a lower common mode input resistance,
a somewhat undesirable situation. The input
resistance configuration shown in Fig. 19-4B is
similar to that shown in Fig. 19-4A with the addition
of resistance R1 to reduce the differential input
resistance to an acceptable value. This configuration
is satisfactory for use in most physiological
situations and can be achieved from the circuit shown
in Fig. 19-4A with the simple addition of one
resistor. The input resistance configuration shown
in Fig. 19-4C provides acceptable values for both
the differential input resistance and the common mode
input resistance and is particularly recommended as
it has the added advantage of having only one
physical resistor forming most of the common mode
input resistance. Typical values for this circuit
may be 1 megohm for R and 100 megohms for $R2$. This
circuit also lends itself to input guarding as
discussed in the following section.

19.4 INPUT GUARDING

The common mode input impedance of an amplifier
can be greatly increased by "guarding" or
"bootstrapping" the input circuit of the amplifier.
Input guarding can best be understood by viewing the
amplifier from the input, by considering DC conditions
and then expanding the analysis to include practical
AC conditions. A conventional single-ended input
circuit is shown in Fig. 19-5A. The input impedance
of this circuit consists of the input resistance R
in parallel with the reactance of the capacitance C
of the shielded cable used for interconnection between
the input voltage and the amplifier. Assume an input
voltage of one volt and an input resistance R of one
megohm. With the input voltage connected to the
amplifier, a current of one μA will flow. As far as
this input voltage is concerned, it is one volt
seeing a load that requires one μA of current; the
load thus appears as one megohm to the input voltage.

AMPLIFIER
INPUT CKT
GAIN A

SHIELDED CABLE
CAPACITY C

$\longrightarrow$ OUTPUT

R

INPUT IMPEDANCE
$Z_{in} = R \| X_C$

(A) CONVENTIONAL INPUT CIRCUIT

AMPLIFIER
INPUT CKT
GAIN A

SHIELDED CABLE
CAPACITY C

G
D
S

R

$\longrightarrow$ OUTPUT

INPUT IMPEDANCE
$Z_{in} = \dfrac{(R \| X_C)}{1 - A}$

TYPICAL VALUE FOR
AMPLIFIER STAGE GAIN
$A = 0.99$

(B) GUARDED INPUT CIRCUIT

AMPLIFIER
INPUT CKT
COMMON MODE
GAIN A

SHIELDED CABLE
CAPACITY C

R

R

COMMON MODE
SIGNAL OUTPUT

COMMON
MODE

COMMON MODE
INPUT IMPEDANCE
$Z_{in(cm)} = 0.5 \dfrac{(R \| X_C)}{1 - A}$

DIFFERENTIAL
INPUT IMPEDANCE
$Z_{in(dif)} = 2R$

(C) GUARDED DIFFERENTIAL INPUT CIRCUIT

Q109A

5.6µF

2.2
MΩ

10
MΩ

0.03µF

33
MΩ

10
MΩ

EMITTER
FOLLOWER
Q110
Q111

Q109B

AT DC –
COMMON MODE INPUT IMPEDANCE 38MΩ
DIFFERENTIAL INPUT IMPEDANCE 2MΩ

AT 60Hz –
COMMON MODE INPUT IMPEDANCE ≈ 750MΩ
DIFFERENTIAL INPUT IMPEDANCE 2MΩ

CONSTANT
CURRENT
SOURCE

(D) GUARDED DIFFERENTIAL INPUT CIRCUIT
AS USED IN THE TEKTRONIX TYPE 410
PHYSIOLOGICAL MONITOR

Fig. 19-5. Input guarding or "bootstrapping".

Now consider the single-ended guarded input circuit shown in Fig. 19-5B. Assume the same conditions as before with the input FET having a gain of 0.90 between gate and source. With the amplifier input resistance connected to the FET source rather than to ground, a one volt input will produce a one volt potential at the FET gate and therefore a 0.90 volt potential at the FET source. The potential therefore appearing across the input resistance R is the difference between the potentials, that is, 0.1 volt. The current through the one megohm input resistor is now 0.1 μA. Now, as far as the input voltage is concerned, it is one volt seeing a load that requires 0.1 μA of current; the load thus appears as 10 megohm to the input voltage even though the actual value of the input resistance is only one megohm. The value of the input resistance is thus amplified by a factor of 10 by connecting the input resistance in a guarded configuration rather than by simply connecting it to ground.

In actual practice, a more typical value for the FET gain between the gate and the source would be 0.990, resulting in an input resistance "amplification" of 100X. Thus, a physical 1-megohm resistor in a guarded circuit would appear as a 100-megohm resistor to the input voltage. The closer the amplifier bootstrap loop gain is to unity, the higher the input resistance as seen from the input. The above discussion applies equally well to AC signals, however, both the input resistance R and the input capacitance C formed by the shielded cable connecting the input voltage to the amplifier must be considered as the input components of the circuit. As with the DC conditions discussed previously, the input reactance of the circuit can also be "amplified" by connecting the shield of the input cable to the FET source rather than simply to ground. A formula for the input impedance of this circuit is given in Fig. 19-5B.

The above discussion can be further extended to a
differential amplifier input circuit as shown in
Fig. 19-5C. With a differential amplifier the
common mode input impedance is increased by guarding.
As discussed in the previous section, it is desirable
to have as high a common mode input impedance as
possible to minimize the effect of source impedance
unbalance.

common
mode
guarded
input

410 guarded
input

Tektronix Type 410 Physiological Monitor uses a
guarded differential-input circuit as shown in
Fig. 19-5D. Since high common mode input impedance
is primarily required to reject 60-Hz signals, the
guarding circuit can be AC coupled to the input
resistors and cable shields as it is unnecessary to
provide high common mode impedance to DC signals.
The 410 Physiological Monitor has a 38-megohm common
mode input resistance to DC signals and a common
mode input impedance to AC signals of considerably
above 750 megohms.

guarding
used within
amplifier

Guarding can also be used within an amplifier to
provide constant operating conditions for the input
FET's. In the 410 Physiological Monitor, as the gates
of both input FET's go positive, the output of the
input-guarding emitter follower, Q110 and Q111, also
goes positive. This emitter-follower output is used
to guard the various components of the input
impedance and is also used to determine the drain
potentials on each of the input FET's. The circuit
is so arranged that a constant voltage appears
between the source and drain no matter what the
common-mode input-signal level.
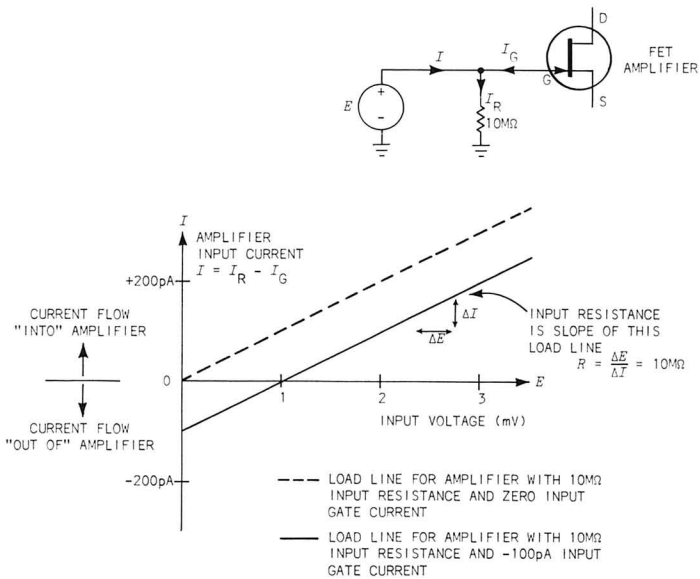
Fig. 19-6. Amplifier input current.

19.5 INPUT CURRENT

Consider the simplified amplifier configuration
shown in Fig. 19-6. The input impedance of such an
amplifier is defined as the ratio of an incremental
change in the input voltage to the resulting
incremental change in the input current. Thus, if
the input voltage shown in Fig. 19-6 changed by 1 mV
and the resulting input-current change was 100 pA,
the input resistance would be $\frac{1 \text{ mV}}{100 \text{ pA}}$ = 10 megohm.

input gate
current
    If the only component of input current was due to
the input resistance, then Ohm's law dictates that
the input current would be zero when the input voltage
is zero. In all physiological amplifiers an
additional input current, due to the amplification
device, is always present. In the case of a modern
FET amplifier, this input current would normally
be below 1 nA and would possess a negative sign
indicating that the FET is acting as a "source" of
current rather than a "sink" of current. This current
is referred to as "input gate current."

In older style amplifiers using vacuum tubes or nuvistors for the input element, the input current may be several nanoamps and may be of either polarity, indicating that the input device is either a "sink" or "source" of current. Input current in these amplifiers is referred to as "input grid current."

input grid current

The two load lines shown on Fig. 19-6 defining the relationship between input voltage and input current both represent an input resistance of 10 megohm (the slope of the line), the upper line representing a *theoretically ideal* amplifier having zero input gate current and the lower line representing a *practical* FET amplifier having an input gate current of -100 pA. Since the component of the input current associated with the input amplification device is constant and does not vary with the input voltage, it must be shown separately when specifying an amplifier and cannot in any way be included with the amplifier's input resistance specifications.

input current versus input resistance

Input gate current can be a source of undesirable offset potential if an amplifier is used in conjunction with a high impedance source. Consider the amplifier shown in Fig. 19-6 used in conjunction with a source having a 1-megohm output impedance. Assume also that this source is zero volts. The -100 pA input grid current will flow through the output impedance of the source creating a potential drop of 100 μV. The output of the amplifier would then indicate a 100-μV input voltage even though the input voltage is in fact zero. This situation is only troublesome when making DC measurements as the gate current is a DC current.

input current produces offset potential

Many amplifiers provide some degree of input gate-current compensation by adding a constant-current drain between the gates of the field effect transistors and ground. Thus, an amplifier such as the Tektronix Type 3A9 Differential Amplifier, has an input gate-current specification of ±20 pA at 25°C even though the actual internal FET-gate current may be in excess of -100 pA.

input current compensation

19.6 DYNAMIC RANGE, DC OFFSET AND RECOVERY

overloading
amplifiers

As either the common mode or differential input voltage to a differential amplifier is increased, a point is reached when the amplifier will overload and the output voltage will not be representative of the input voltage. This overload may occur within the amplifier input stages or elsewhere within the amplifier. If, however, the amplifier's dynamic range specifications are not exceeded, overload will not occur. Dynamic range characteristics for a differential amplifier are specified as follows:

1. Input differential dynamic range -- The maximum peak differential voltage that can be applied between the differential-amplifier input terminals,

2. Input common mode dynamic range -- The maximum peak common mode voltage that can be applied between the inputs of the differential amplifier and ground,

3. Output dynamic range -- The maximum peak output voltage that can be expected from the amplifier.

The amplifier is referred to as "overloaded" if any one of these three limits is exceeded.

transient
overloading
versus
recovery

In physiological applications it is not uncommon for transients to be produced that are several orders of magnitude greater in potential than the physiological signal under observation. An important characteristic of a physiological amplifier is the time taken for it to recover from such overloads. Amplifier recovery characteristics can be defined by the time taken for the amplifier to recover from a transient within the dynamic range specifications of the amplifier and by the time taken for the amplifier to recover from transients far greater than the dynamic range characteristics. DC-coupled amplifiers normally recover almost instantaneously from transients within the amplifier's dynamic range specifications as these transients are within the common mode rejection capabilities of the amplifier. Recovery time for transients that exceed the amplifier's dynamic range specifications are difficult to specify as

this recovery time depends on the precise characteristics of the overload transient. With such transients, thermal effects and the inherent unbalance produced by such thermal effects must be considered. Typical amplifiers may take several seconds to recover from severe overload.

AC coupling and transients versus recovery

AC-coupled differential amplifiers may have particularly bad recovery characteristics due to the inherent time constant involved in the AC-coupling network. Recovery time may be ten times as long as this time constant. Although AC coupling at the amplifier input is not recommended, AC coupling at a later stage in the amplifier is desirable as it eliminates electrode offset potentials, etc. Amplifiers that are AC coupled after several stages of DC-coupled amplification are referred to as "AC-stabilized" amplifiers. Many physiological amplifiers, such as the amplifier incorporated into the Tektronix Type 410 Physiological Monitor, incorporate AC coupling after several stages of DC amplification and internal limiting to improve the amplifier's recovery characteristics. The amplifier in the Type 410 Physiological Monitor has a DC-coupled input stage and is AC coupled after this input stage at a time

410 recovery

constant of two seconds. Its recovery time is less than four seconds for any overload signal. This characteristic is particularly important in an amplifier intended for ECG monitoring in intensive care and surgery as the amplifier will certainly be overloaded if the subject is defibrillated and it is highly important that the medical personnel be able to use the amplifier immediately after the defibrillation process is complete.

DC differential offset versus electrode offset

Physiological signals obtained from electrodes include a DC component due to the electrode-offset potential. A differential amplifier capable of recording these physiological signals in a DC-coupled mode must therefore, include the capability to reject this electrode-offset potential. This is achieved by incorporating DC-differential offset within the differential amplifier which effectively produces a DC-differential voltage at the differential amplifier input which cancels the electrode-offset potential.

electrode
offset
exceeding
dynamic
range

If DC-differential offset is not available within an amplifier, AC coupling must be used to reject the electrode-offset potential. As it is undesirable to incorporate AC coupling at the amplifier's input the amplifier must be AC coupled after one or more amplification stages and care must be taken to ensure that the electrode-offset potential is not exceeding the dynamic-range characteristics of the amplifier. Often the user may not be aware that the amplifier's dynamic range is being exceeded and erroneous results are obtained.

dynamic
window

The term "dynamic window" refers to the use of a differential amplifier with DC-differential offset. Dynamic window can best be understood by considering the following example. Consider a Tektronix Type 3A9 Differential Amplifier operating at its maximum gain of 100,000 with no DC-differential offset. As the output dynamic range for this amplifier is ±5 volts, the maximum differential input voltage that will not exceed this output dynamic range is $\frac{\pm 5 \text{ volts}}{100,000}$ or ±50 μV. The amplifier is able to "see" a dynamic window of ±50 μV. Now assume that the Type 3A9 is used in the DC OFFSET position. The DC OFFSET range available is the same as the differential dynamic range, that is ±1 volt. Assume that the course and fine DC OFFSET controls are set to provide +0.3 volts of DC-differential offset. Now the maximum differential input voltage that can be applied to the amplifier is ±50 μV +0.3 volts. The amplifier is now able to "see" a dynamic window of ±50 μV from 0.3 volts, that is from 0.299950 volts to 0.300050 volts.

19.7 NOISE AND DRIFT

noise
versus
drift

stability

hum

Noise and drift are both unwanted signals that occur within an amplifier. The term "noise" normally refers to unwanted signals generated at a frequency above 0.1 Hz and DC drift normally refers to slow changes in the output level at frequencies below 0.1 Hz. The term "stability" is also used to indicate relative DC-drift characteristics. The component of noise occurring at the power-line frequency and harmonics of the power-line frequency is often referred to as "hum."

The noise generated within an amplifier is often
specified in microvolts and refers to this noise as
if it were a differential input voltage. The noise
produced within an amplifier is either specified as
microvolts P-P, microvolts RMS, microvolts
tangential or by specifying an equivalent noise
qualifying   resistance. The equivalent noise resistance
noise   considers all the noise in an amplifier to be
produced by an equivalent resistance producing
thermal noise. The RMS value of this thermal noise
is related to the resistance value by the following
expression:

$$E_{RMS} = 7.4 \times 10^{-12} \times \sqrt{R \ bw \ (t + 273)}$$

or

equivalent
related to $$R = \frac{(E_{RMS})^2 \times 10^{24}}{55 \ bw \ (t + 273)}$$ R = Equivalent Noise
RMS   Resistance -- ohms

bw = Bandwidth -- Hz

t = Temperature -- °C

Thus, an amplifier specified as having an equivalent
noise resistance of 100,000 ohms would produce 7 μV
RMS of noise at 30 kHz bandwidth at 20°C. Equivalent
noise resistance relates, therefore, to RMS noise.

The RMS noise, and thus the equivalent noise
resistance, of an amplifier is measured by shorting
the inputs to ground and measuring the output voltage
at maximum sensitivity with an RMS-indicating
RMS noise   voltmeter. The RMS noise generated within the
amplifier will then be this RMS output voltage
divided by the gain of the amplifier. For a
Tektronix Type 3A9 Differential Amplifier operating
at a bandwidth of 30 kHz and at its maximum gain of
100,000, the RMS noise output was 0.32 volts. Thus,
the RMS noise produced within this amplifier is
3.2 microvolts RMS and the equivalent noise
resistance for the amplifier is 20,000 ohms.

Amplifier noise is normally specified in microvolts
RMS if the amplifier is to form part of an
instrumentation system. This noise specification
gives no information as to the peak component of
noise or to the relative appearance of a CRT display
incorporating this noise, thus either P-P noise or
tangential-noise measurements may be desirable in
some instances.
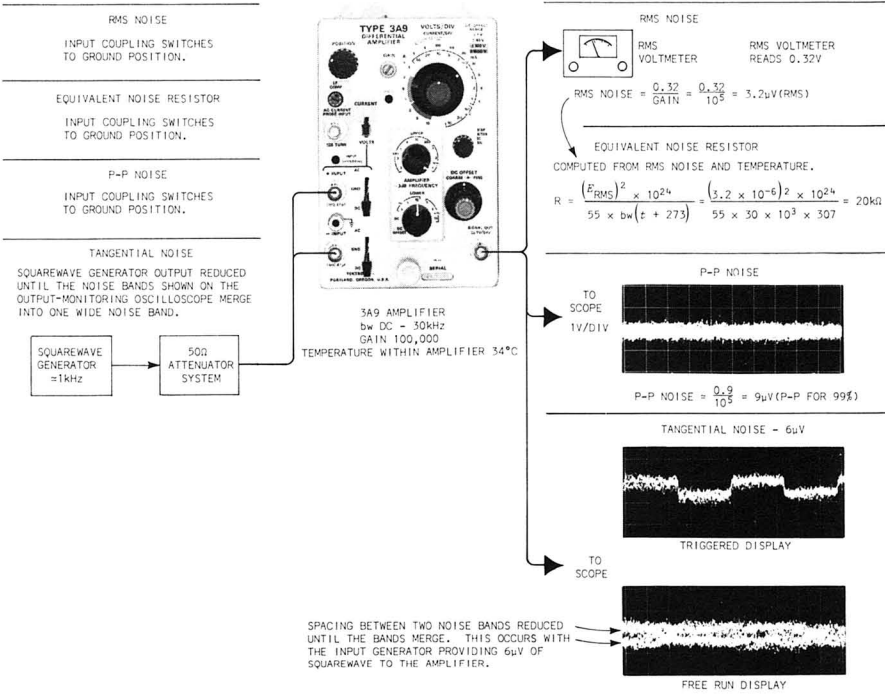
Fig. 19-7. Amplifier noise measurement.

peak-to-peak noise

If an amplifier is to be used in conjuction with peak detector or trigger circuitry, it is important to know the peak value of noise produced by an amplifier rather than simply the RMS value of this noise. Peak-to-peak noise for an amplifier can be measured in the same way as RMS noise by using an oscilloscope to display the noise output. The displayed noise for the amplifier system discussed previously is shown in Fig. 19-7. From observation of the oscilloscope display at a sensitivity of 1 volt per division, the P-P deflection shown appears to be approximately 0.9 divisions referring to an output of 0.9 volts. The P-P noise generated within the amplifier is thus 9 μV P-P referred to the input. Closer examination of the noise shows that, while 99 percent of the information appears to be within 0.9 divisions, random spikes are produced over an apparent overall range of 1.2 divisions. (Theoretically, random spikes can extend out any distance.) The P-P noise generated by this amplifier could, therefore, be more correctly stated as 12 μV subjective P-P with 99 percent of the noise below 9 μV P-P.

tangential noise

In the majority of physiological measurements an oscilloscope is used to display the output from an amplifier. It is thus desirable to know the amplifier's noise characteristics with respect to this oscilloscope display. A "tangential" measurement of displayed noise effectively measures the trace width produced by noise and therefore gives an idea of the resolution obtainable from the oscilloscope screen. The following discussion on tangential noise measurement technique will further clarify this resolution concept. Consider a low-amplitude squarewave being applied to the input of the differential amplifier as shown in Fig. 19-7. As it is necessary to know the amplitude of this squarewave, it is desirable to obtain a relatively high-amplitude squarewave for ease of measurement and then to accurately attenuate this via a 50-ohm attenuation system. Although the output from this 50-ohm attenuation system is single ended, that is, only one output source is produced with respect to ground, this output should be coupled differentially into an amplifier by connecting the positive input of the amplifier to the output source and the negative input of the amplifier to the reference ground for this output source. For the configuration

shown in Fig. 19-7, the oscilloscope will clearly show the squarewave together with the noise generated from within the differential amplifier; this display is shown in the upper photograph of the tangential noise section of Fig. 19-7. If the oscilloscope's sweep generator is then operated in a free-running mode, the squarewave information will be lost and the CRT display will appear as two bands of noise. As the output from the squarewave generator is increased in amplitude, these bands will tend to separate; as the output is decreased in amplitude, these bands will tend to merge into one. The input squarewave level at which these two bands just merge into a single band is referred to as the tangential-noise specification for the amplifier. In the particular configuration shown in Fig. 19-7, the tangential noise specification for the amplifier is 6 μV and a free-running display of this 6-μV squarewave together with inherent noise from within the amplifier is shown in the bottom photograph in Fig. 19-7.

tangential
× 2 equals
RMS

If an amplifier's frequency response is perfectly Gaussian, then a mathematical analysis of RMS-noise measuring techniques and tangential-noise measuring techniques will show tangential noise to be twice RMS noise. Since most physiological amplifiers are essentially Gaussian, RMS noise can be simply determined once tangential noise has been measured. This indirect method of determining RMS noise sidesteps the need for a true RMS-reading voltmeter.

P-P
derivation
from RMS

If the distribution of noise within an amplifier is Gaussian, and the noise distribution in most wideband amplifiers is approximately Gaussian, then the noise distribution will follow a "normal distribution" curve. From the curve we can deduce that:

RMS noise × 2 = P-P noise for 68% of the noise

RMS noise × 4 = P-P noise for 96% of the noise

RMS noise × 5 = P-P noise for 99% of the noise

RMS noise × 6 = P-P noise for 99.9% of the noise

RMS noise × 8 = P-P noise for 99.99% of the noise

Examination of the equation relating RMS noise to equivalent noise resistance indicates that RMS noise is proportional to the square root of the amplifier's bandwidth. This noise component is known as "thermal noise" or "Johnson noise." The above proportionality is essentially true for wideband amplifiers (100 kHz and above) but, however, does not hold true for narrowband amplifiers or when wideband amplifiers are subject to bandwidth limiting. Under these conditions an additional low-frequency noise component generated within the input field-effect transistors becomes an appreciable proportion of the amplifier's total noise and cannot be neglected. This extra noise component is referred to as "excess noise" (also occasionally referred to as "$\frac{1}{f}$ noise" or "flicker noise") and is discussed in more detail in Chapter 20 when discussing bandwidth-limiting considerations.

excess
noise

As previously stated, drift can be regarded as very low-frequency noise generated within an amplifier. Drift specifications, as with noise specifications, are in microvolts and refer to the drift as if it were a differential input voltage. Because of its inherent low frequency nature, drift is normally specified in P-P units. Drift specifications for most amplifiers are self-explanatory, however, a full evaluation of drift should include both long-term drift and short-term drift measurements. Short-term drift measurements are measured in microvolts per minute and indicate to the user the drift in microvolts expected over a one-minute period after the amplifier has had sufficient time to warm up. Long-term drift specifications are in microvolts per hour and indicate the total drift that may be expected over a one-hour period. Long-term drift specifications will always exceed short-term drift specifications.

short- and
long-term
drift

Both long-term drift and short-term drift assume a constant operating temperature within the differential amplifier and an additional specification must be generated to determine the characteristics of this drift with a change in temperature. Temperature drift is specified in microvolts per degree Celsius and indicates the drift that may be expected from the amplifier for a 1°C change in ambient temperature.

temperature
drift

PRINCIPAL CHARACTERISTICS ONLY SHOWN WITH THE 3A9 USED AS AN INPUT AMPLIFIER TO PROVIDE GAIN
FROM A DIFFERENTIAL INPUT TO THE SIGNAL OUTPUT CONNECTOR AT A BANDWIDTH OF FROM DC TO 30KHz.
BOTH WIDER AND NARROWER BANDWIDTHS ARE AVAILABLE AS DISCUSSED IN CHAPTER 20 - SIGNAL PROCESSORS.
GAIN SETTINGS BELOW x100 ARE NOT CONSIDERED.

CHARACTERISTICS	PERFORMANCE	* MODIFIED PERFORMANCE
GAIN - CALIBRATED	x100 TO x100,000; 10 STEPS IN A 1-2-5 SEQUENCE	SAME AS STANDARD PERFORMANCE
- UNCALIBRATED	VARIABLE CONTROL PROVIDES CONTINUOUSLY VARIABLE GAIN BETWEEN STEPS	"
CMRR	100,000:1	"
DYNAMIC RANGE	± 1 VOLT DIFFERENTIAL ± 10 VOLTS COMMON MODE	" "
RECOVERY	TO WITHIN 0.5% IN LESS THAN 10µs IF OVERLOAD VOLTAGE WITHIN DYNAMIC RANGES SHOWN ABOVE	"
INPUT IMPEDANCE	2MΩ 24pF DIFFERENTIAL 0.5MΩ 94pF COMMON MODE	" 5.0MΩ 94pF COMMON MODE
INPUT CURRENT	MAXIMUM ±20pA AT EACH INPUT AT 25°C	TYPICALLY < ±100pA AT EACH INPUT AT 25°C
NOISE	6µV MEASURED TANGENTIALLY (TYPICAL)	SAME AS STANDARD PERFORMANCE
DC DRIFT AFTER 1 HR WARMUP	5µV/MIN (PEAK TO PEAK) 10µV/HOUR (PEAK TO PEAK) 50µV/°C	"
OUTPUT DYNAMIC RANGE	±5 VOLTS FROM 100Ω SOURCE	"
DC OFFSET	±1 VOLT	"

*3A9 MODIFIED TO INCREASE INPUT IMPEDANCE
  - REMOVE LINKS ON 3A9 CIRCUIT BOARD AS
    SHOWN ON PHOTOGRAPH.
  - ADD 2.2MΩ RESISTOR BETWEEN + INPUT
    AND - INPUT IF DESIRED TO MAINTAIN
    2MΩ DIFFERENTIAL INPUT RESISTANCE.

REMOVE
LINKS

ADD 2.2MΩ
BETWEEN
THESE
POINTS
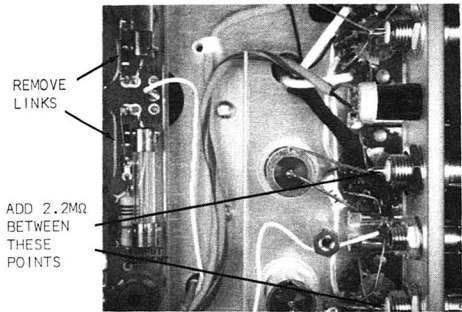
Fig. 19-8. Characteristics of the Tektronix Type 3A9 Differential Amplifier
when used as a physiological differential amplifier.

power
supply
drift

If high-stability power supplies are not used within an amplifier, drift may be caused by variations in supply voltage. This would be specified as microvolts drift for a supply voltage change of ±10% from nominal. Drift with line-voltage change is negligible in modern FET amplifiers.

19.8 SPECIALIZED AMPLIFIERS

Only the characteristics of general-purpose physiological amplifiers have been discussed in this chapter. A typical general-purpose physiological amplifier is the Tektronix Type 3A9 Differential Amplifier, abbreviated specifications for which are shown in Fig. 19-8. Many physiological-monitoring situations are particularly dependent on one or more characteristics of an amplifier and it is often necessary to use amplifiers with characteristics

optimum
character-
istics

optimized for a particular measurement requirement. An ECG amplifier intended for use during defibrillation must possess particularly good recovery characteristics. An EEG amplifier must be particularly free of noise and an amplifier intended for use with microelectrodes for recording evoked responses must have an extremely high input impedance and low input current. Amplifiers specifically intended for use with microelectrodes are discussed in more detail in Chapter 11. A relatively new concept is the use of plug-in probes that allow a physiological amplifier to be optimized for a particular measurement requirement. Thus, the one amplifier may be used for general purpose recording, for high sensitivity recording, or for use in conjunction with microelectrodes simply by changing probes.

20

SIGNAL PROCESSORS – OPERATIONAL AMPLIFIERS

The signal available as an output from a physiological amplifier may not be in a form compatible with other instrumentation. Signal processing is often required to either achieve compatibility or to improve the presentation of a physiological signal. Signal processing is also used when more than one category of information is to be obtained from the same physiological signal.

20.1 OPTIMUM BANDWIDTH FOR PHYSIOLOGICAL SIGNALS

DC-10,000 Hz
bandwidth

Spectrum analysis of the action potential generated when a single cell depolarizes indicates that essentially all of the signal is produced within the frequency domain from DC to 10,000 Hz. Since all physiological signals are the sum of one or more action potentials, the frequency domain of these physiological signals will be less than or equal to the frequency domain of the single-cell action potential. Thus, the optimum bandwidth for a physiological amplifier capable of monitoring any physiological signal is from DC to 10,000 Hz. This bandwidth is adequate for use in almost all physiological monitoring situations; however, in the research environment, some workers postulate the existence of higher frequency components and a bandwidth from DC to 30,000 Hz should be adequate to allow these postulated components to be observed.

INPUT SIGNAL

PRF = 100Hz
t_r AND t_f = 0.1ms
WIDTH = 0.6ms

FREQUENCY SPECTRUM
OF INPUT SIGNAL

1kHz   6kHz   11kHz

HIGH FREQUENCY RESPONSE LIMITING

-3dB

100kHz
30kHz
10kHz

3kHz
1kHz
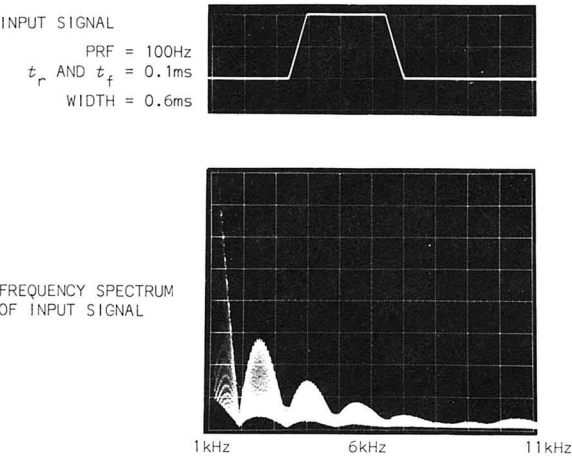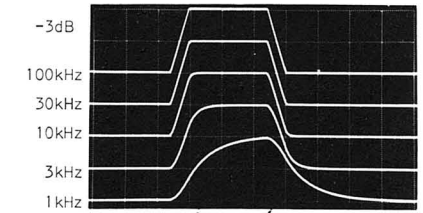
DISPLAYED RISE-
TIME OR FALLTIME

$$= \sqrt{\left(\begin{array}{c}\text{MAXIMUM}\\ \text{RISE- OR FALLTIME}\\ \text{COMPONENT IN THE}\\ \text{INPUT SIGNAL}\end{array}\right)^2 + \left(\dfrac{0.35}{\begin{array}{c}\text{HIGH FREQUENCY}\\ \text{-3dB POINT}\end{array}}\right)^2}$$

HIGH FREQUENCY -3dB CUTOFF REQUIRED
FOR NEGLIGIBLE DISTORTION, THAT IS,
NO VISUAL WAVEFORM CHANGE

$$= \dfrac{1.5}{\begin{array}{c}\text{MAXIMUM RISE- OR}\\ \text{FALLTIME COMPONENT}\\ \text{IN THE INPUT SIGNAL}\end{array}}$$

HIGH FREQUENCY -3dB CUTOFF REQUIRED FOR
ACCEPTABLE DISTORTION, THAT IS, NO VISUAL
RISETIME DEGRADATION BUT SOME ROUNDING
OF ABRUPT CHANGES

$$= \dfrac{1.0}{\begin{array}{c}\text{MAXIMUM RISE- OR}\\ \text{FALLTIME COMPONENT}\\ \text{IN THE INPUT SIGNAL}\end{array}}$$

LOW FREQUENCY RESPONSE LIMITING

-3dB

DC
10Hz
100Hz

y x

w
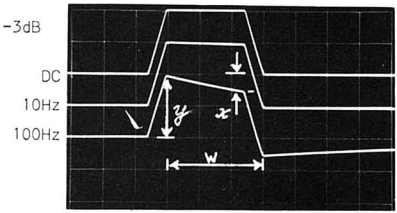
DISPLAYED "TILT" OR "SAG" IN $\% = \dfrac{x}{y} \times 100\%$

DISPLAYED TILT

$$= \begin{array}{c}\% \text{ TILT ON THE}\\ \text{INPUT SIGNAL}\end{array} + \left(630 \times w \times \begin{array}{c}\text{LOW FREQUENCY}\\ \text{-3dB POINT}\end{array}\right)$$

LOW FREQUENCY -3dB CUTOFF REQUIRED
FOR NEGLIGIBLE DISTORTION, THAT IS,
< 1% TILT ADDED

$$= \dfrac{.0016}{\begin{array}{c}\text{MAXIMUM w IN THE}\\ \text{INPUT SIGNAL}\end{array}}$$

LOW FREQUENCY -3dB CUTOFF REQUIRED
FOR ACCEPTABLE DISTORTION, THAT IS,
< 5% TILT ADDED

$$= \dfrac{.008}{\begin{array}{c}\text{MAXIMUM w IN THE}\\ \text{INPUT SIGNAL}\end{array}}$$

Fig. 20-1. Input signal degradation due to frequency response limiting.

Fig. 20-2. Amplifier noise/bandwidth relationship
(Tektronix Type 3A9 Amplifier).

Johnson noise versus bandwidth

Examination of the equation relating the Johnson noise produced within an amplifier to the amplifier's equivalent noise resistance indicates that Johnson noise is proportional to the square root of the amplifier's bandwidth. It is thus apparent that if the amplifier's bandwidth is limited, the RMS noise produced within the amplifier will also be decreased. This Johnson noise–bandwidth relationship is shown in Fig. 20-2.

excess noise versus bandwidth

The excess noise component of the total RMS noise produced within an amplifier is not mathematically related to the amplifier's bandwidth. There is little change in this excess noise component when the amplifier's bandwidth is reduced from 300,000 Hz to 3,000 Hz, thus the frequency content of this noise for modern FET amplifiers is predominantly below 3,000 Hz. The relationship between the excess noise produced within a Tektronix Type 3A9 Amplifier and the amplifier's bandwidth is also shown in Fig. 20-2. Excess noise has historically been referred to as $\frac{1}{f}$ noise.

total noise
versus
bandwidth

Since both Johnson noise and excess noise are essentially random, the total of these two noise components will be the square root of the sum of the squares of each component. The total RMS noise produced within a Type 3A9 Amplifier is shown in Fig. 20-2. It is apparent from this figure that an amplifier bandwidth reduction from 300,000 Hz to 10,000 Hz results in a 280% reduction in the noise produced within the amplifier. It is also apparent that a further reduction in the amplifier's bandwidth from 10,000 Hz to 1,000 Hz is of less value in reducing noise as the resulting decrease in the amplifier's noise is only about 33%. Many physiological signals can be recorded with bandwidths substantially less than 1,000 Hz and in such cases, limiting the amplifier's frequency response can also substantially reduce the noise generated within the amplifier.

3A9 signal
processing
features

An amplifier such as the Tektronix Type 3A9 Amplifier incorporates bandwidth limiting as shown on Fig. 20-3. The high frequency response of the amplifier can be adjusted in nine steps between 1 MHz and 100 Hz and the low frequency response of the amplifier can be adjusted in seven steps between DC and 10,000 Hz. DC differential-offset capabilities are also provided to offset the effect of electrode-tissue interface potentials as discussed in Chapters 16 and 19.
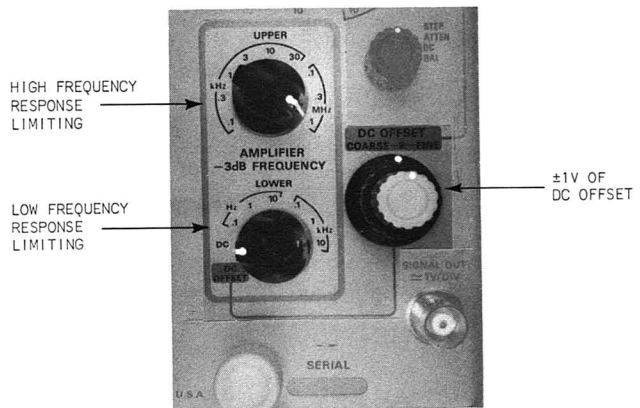
Fig. 20-3. Signal processing controls on the Type 3A9 Amplifier.

INPUT SIGNAL BEFORE ATTENUATION

5μV SIGNAL AFTER ATTENUATION

AMPLIFIER
UPPER −3dB
1MHz

1ms/DIV
ALL WAVEFORMS

0.1MHz

10kHz

ALL 10μV/DIV
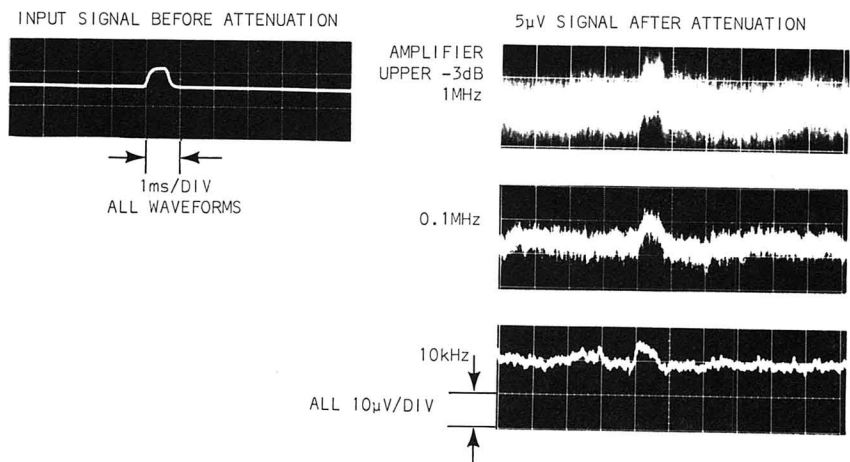
Fig. 20-4. Noise reduction by high frequency response limiting.

INPUT SIGNAL AND
BASELINE DRIFT DUE
TO ELECTRODE OFFSET
POTENTIAL DRIFT.
AMPLIFIER DC COUPLED

AMPLIFIER LOWER −3dB
CUTOFF AT 10Hz
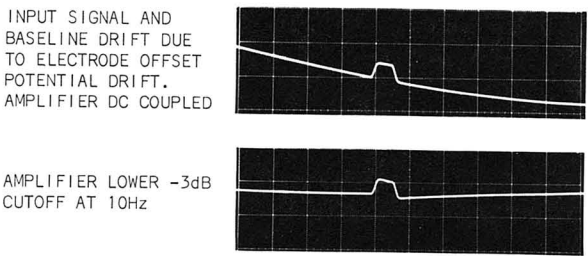
Fig. 20-5. Drift reduction by low frequency response limiting.

limiting
upper 3-dB
point

The effect of bandwidth limiting within an oscilloscope vertical amplifier on the oscilloscope's display is shown in Fig. 20-4. With the oscilloscope displaying a 5-μV pulse with an amplifier bandwidth of 1 MHz, the pulse is completely swamped by the amplifier's internally generated noise. If the amplifier's upper 3-dB cutoff is limited to 10,000 Hz the noise is substantially reduced and the input signal predominates. A further reduction in the amplifier's bandwidth has little effect on the displayed noise; however, it does produce undesirable distortion.

20.3 AMPLIFIER LOW FREQUENCY RESPONSE LIMITING

limiting
lower 3-dB
point

While limiting an amplifier's low frequency response has a negligible effect on the noise produced within the amplifier, it can be useful in eliminating unwanted low frequency signals such as drift within the amplifier or drift in the electrode-offset potential produced between two physiological electrodes. Fig. 20-5 depicts a typical physiological response obtained with a DC-coupled amplifier and one obtained using low frequency bandwidth limiting.

20.4 LINE FREQUENCY REJECTION

Some years ago line-frequency rejection filters were incorporated into physiological monitoring systems to eliminate 50- or 60-Hz line frequency interference from a physiological signal. Of course, if the physiological signal also contained frequency components at 50 or 60 Hz, these components were also eliminated and the physiological signal was distorted.

high CMRR
minimizes
interference

With the advent of the modern high CMRR differential amplifier it was found that the increased common mode rejection eliminated most line-frequency interference, indicating that this interference was from a common mode source rather than from a differential source. When using an amplifier with high common mode rejection ratio, it is rarely necessary to include any form of line frequency filter and, as such filters may cause signal distortion, line-frequency rejection filters should be considered only as a last resort.

20.5 NOISE REDUCTION BY SIGNAL AVERAGING

signal
averaging
minimizes
noise

Signal averaging is a signal processing technique
for extracting a wanted signal from a background of
unwanted noise. Signal averaging can only be used
if the desired signal can be generated a number of
times, either periodically or aperiodically, and can
be used to effectively reduce the noise accompanying
the signal by several orders of magnitude. The
subject of signal averaging is adequately covered in
Chapter 11, Section 11.4, when discussing the
techniques used to record evoked cortical responses.
Because of its inherent expense, signal averaging
is normally used only when absolutely necessary and
is primarily used for recording cortical responses
and in research applications. It is, however, an
important physiological signal processing technique
and should not be overlooked. This subject is
discussed elsewhere in this book.

20.6 OPERATIONAL AMPLIFIERS

character-
istics

Electronically, an operational amplifier is simply
a high gain inverting amplifier designed to remain
stable with large amounts of negative feedback from
output to input. An operational amplifier, by means
of this negative feedback, is capable of processing
a signal in many different ways, dependent only on
the passive components selected as the feedback
network and possibly as a series input network. The
ideal operational amplifier has infinite gain,
infinite input impedance and zero output impedance.

reference
sources

This chapter should not be regarded as a general
reference on operational amplifiers, it specifically
discusses operational amplifiers from a biophysical
measurement standpoint and makes no attempt to review
some of the more sophisticated operational amplifier
systems used within the general electronics industry.
A companion volume entitled *Operational Amplifier
Circuits* published by Tektronix, Inc. provides a more
general reference source on the use of operational
amplifiers. The instruction manual provided by
Tektronix for use with the Type 3A8 Operational
Amplifier plug-in also provides an extensive
reference on operational-amplifier signal-processing
circuits. The major manufacturers of modular or
IC-packaged operational amplifiers also produce

application handbooks containing many circuits that would be useful for biophysical measurements. In particular, we would recommend *Applications Manual for Operational Amplifiers* published by Philbrick/ Nixus Research, Dedman, Massachusetts.

negative
feedback to
current
null at
input

Referring to Fig. 20-6, an operational amplifier with negative feedback operates in a self-balancing configuration providing, through the feedback element, whatever current is necessary to hold the input at ground potential. The output signal is a function of this current and of the impedance of the feedback element. If current is applied to the operational amplifier input, it would tend to develop a voltage across the feedback impedance and thus move the operational amplifier input away from ground potential. The output, however, changes in the opposite direction, providing current to balance the input current and thus maintains the operational amplifier input at ground potential. If the feedback impedance is high, the output voltage must become quite high to provide enough current to balance even a small input current.

FEEDBACK
CURRENT
I_f

Z_f FEEDBACK
IMPEDANCE

INVERTING
HIGH GAIN AMPLIFIER
(ASSUME INFINITE
GAIN)

INPUT
CURRENT
I_{in}

OUTPUT
e_o

CURRENT NULL POINT
AT GROUND POTENTIAL
(ASSUME ZERO CURRENT
TO AMPLIFIER)

VOLTAGE OUTPUT
FROM A LOW SOURCE
IMPEDANCE
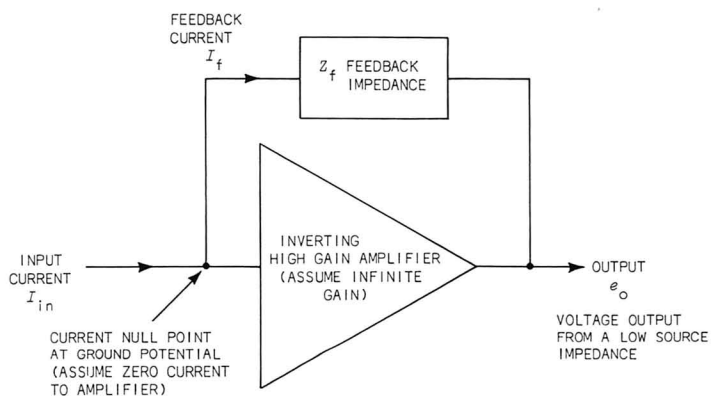
$$I_{in} + I_f = 0$$
$$\text{AND } I_f = \frac{e_o}{Z_f}$$

$$\therefore I_{in} + \frac{e_o}{Z_f} = 0$$

$$\text{OR } e_o = \frac{-I_{in}}{Z_f}$$

Fig. 20-6. Operational amplifier principle.

(A) MODULAR OPERATIONAL AMPLIFIERS USING DISCRETE COMPONENTS OR IC'S
GENERALLY LOW POWER OUTPUTS
<100mW
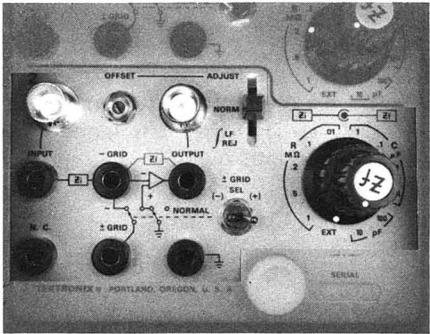
(B) TEKTRONIX TYPE 3A8 OPERATIONAL AMPLIFIER PLUG-IN UNIT

MEDIUM POWER OUTPUT
±25V   ±7.5mA   190mW

(C) KEPCO POWER OPERATIONAL AMPLIFIER

HIGH POWER OUTPUT
±36V   ±5A   180W

Fig. 20-7. Typical operational amplifiers.
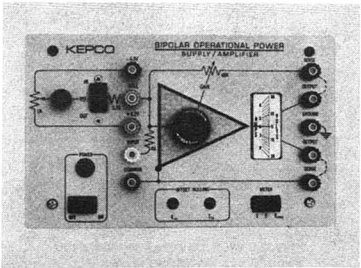
voltage
input

In physiological measurements we are more often required to deal with an input *voltage* rather than an input *current*. We thus use an additional element, referred to as an input impedance (Z_i), which converts an input voltage signal to an input current signal.

noninverting
input

The input to an inverting high-gain amplifier is commonly referred to as the inverting input or "− input." Many operational amplifiers also provide access to a noninverting or "+ input." If this noninverting input is used in an operational amplifier circuit, negative feedback must still be maintained between the output and inverting input and the operating principles remain as discussed previously with the exception that the current null point will be maintained at the potential of the noninverting input rather than at ground potential.

classifi-
cations

Many types of operational amplifiers are commercially available, some of which are shown in Fig. 20-7. Operational amplifiers can generally be classified as either general purpose or specialized, this discussion will refer only to general purpose operational amplifiers. Operational amplifiers can be further classified as to their output voltage and current capabilities.

low power

Low power operational amplifiers are available in modular or integrated circuit packages as shown in Fig. 20-7A and generally provide an output power of less than 100 milliwatts. A typical IC-packaged low-power operational amplifier may provide a maximum output of ±10 volts at ±5 milliamperes. Modular operational amplifiers are particularly suited for use in custom signal processing equipment as they need only be supplied with the necessary DC power, feedback impedance and input impedance to perform a signal processing function. The Fairchild μA741 operational amplifier is particularly suitable.

The operational amplifier shown in Fig. 20-7B is one of the two operational amplifiers incorporated into the Tektronix Type 3A8 Operational Amplifier plug-in unit. The Type 3A8 can be used with Tektronix 560-Series Oscilloscopes or with the Type 129 Plug-in Unit Power Supply. It contains two separate but identical operational amplifiers plus a display amplifier. The display amplifier monitors the output of either operational amplifier or can be used as an independent oscilloscope amplifier. The input impedance and feedback impedance for each operational amplifier can be internally selected from a range of resistors and capacitors, thus eliminating the need for external components if the operational amplifier is to be used for simple signal processing applications. If more sophisticated applications are required the necessary input impedance and feedback impedance networks can be permanently constructed on an adapter assembly (Tektronix PN 013-0048-01) which can be plugged into the operational amplifier when desired. This technique is used extensively in the various operational amplifier circuits used throughout this book and discussed in greater detail in Chapter 29. The operational amplifiers incorporated into the Type 3A8 are classified as medium-power operational amplifiers providing an output power below 1 watt. The Type 3A8 can provide a maxium output of ±25 V at ±7.5 mA.

medium power

The high-power operational amplifier shown in Fig. 20-7C has an output power capability of 180 watts. Such high power capability operational amplifiers are rarely used in biophysical measurements; however, high-power operational amplifiers having output powers of several watts may be required to drive chart recorders and other low impedance devices.

high power

20.7 OPERATIONAL AMPLIFIER APPLICATIONS

The references mentioned at the beginning of the previous section (20.6) should be consulted when considering the use of an operational amplifier for a particular signal processing application. Operational amplifier applications can, however, be classified into several broad groups as shown in Figs. 20-8 and 20-9. The following will discuss

these configurations as relating to one of the operational amplifiers in the Tektronix Type 3A8 Operational Amplifier plug-in unit. Some practical operational amplifier circuits using the Type 3A8 are discussed in Chapter 29.

noninverting amplifier

The noninverting amplifier shown in Fig. 20-8 can be used to provide gain in any physiological measuring system. It also provides a high input impedance and a low output impedance and thus can provide current gain as well as voltage gain in a system. If the Z_i and Z_f components of this noninverting amplifier are infinity and zero respectively, then the configuration becomes a unit voltage gain amplifier providing only current gain between the input and output.

oscillator

If positive feedback is added to an operational amplifier, an oscillator is formed as shown in Fig. 20-8. One or more frequency dependent elements in the positive-feedback circuit determines the frequency of oscillation, however, some negative feedback is also required to stabilize the oscillation amplitude. This negative feedback may simply consist of resistive feedback elements or often nonlinear devices are used to provide some degree of output voltage regulation.

NONINVERTING AMPLIFIER

$$e_o = E_i \frac{Z_f + Z_i}{Z_i}$$

UNITY GAIN AMPLIFIER

$$e_o = E_i$$

OSCILLATOR

NEGATIVE FEEDBACK STABILIZES OSCILLATION AMPLITUDE

POSITIVE FEEDBACK PRODUCES OSCILLATION

FREQUENCY DEPENDENT ELEMENTS
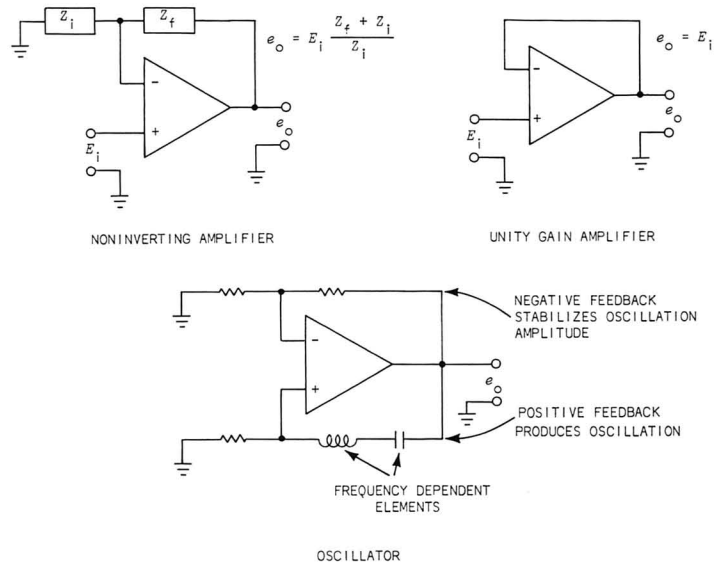
Fig. 20-8. Operational amplifier applications using the noninverting (+) input to the amplifier.

CURRENT TO VOLTAGE CONVERTER

$e_o = -I_i Z_f$
$E_i = 0$

INVERTING AMPLIFIER

$e_o = -E_i \dfrac{Z_f}{Z_i}$
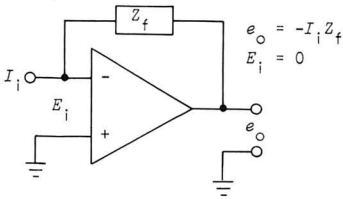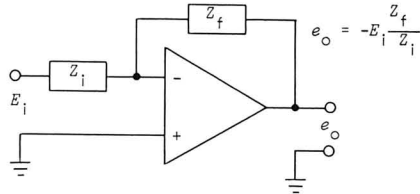
IF $Z_f Z_i$ ARE RESISTIVE, THE AMPLIFIER IS WIDEBAND WITHIN THE LIMITATION OF THE OPERATIONAL AMPLIFIERS GAIN BANDWIDTH PRODUCT.

IF $Z_f Z_i$ ARE RC COMBINATIONS THE AMPLIFIER FREQUENCY RESPONSE WILL DEPEND ON THE RC RATIO.

IF $Z_f Z_i$ ARE NONLINEAR WITH VOLTAGE OR CURRENT (DIODE) THEN THE AMPLIFIER IS NONLINEAR, EITHER COMPRESSION OR EXPANSION.

SWITCH OPENED DURING INTERVAL $t_1 - t_2$

$e_o = \dfrac{-1}{RC} \displaystyle\int_{t_1}^{t_2} E_i \, \Delta t$

VOLTS/s
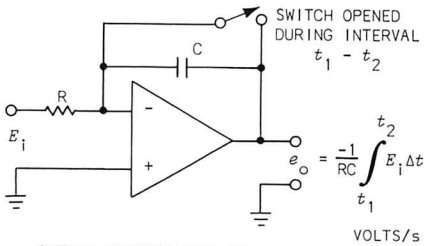
OUTPUT PROPORTIONAL TO INPUT VOLTAGE AND TIME

INTEGRATOR

$e_o = \dfrac{-\Delta E_i}{\Delta t} RC$

VOLT/s
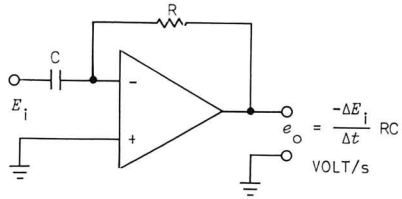
OUTPUT PROPORTIONAL TO RATE OF CHANGE OF INPUT VOLTAGE

DIFFERENTIATOR

Fig. 20-9 Operational amplifier applications using the inverting (–) input to the amplifier.

The previous circuits use the noninverting + input to an operational amplifier as well as the inverting − input. The following circuits use only the − input to the operational amplifier with the + input grounded. The basic operational amplifier and feedback impedance discussed in Section 20.6 is

current to
voltage
converter

essentially a current to voltage converter as shown in Fig. 20-9 which provides an output voltage proportional to the input current and feedback impedance.

The input current for a current to voltage converter circuit may be obtained from an input voltage by the

inverting
amplifier

use of an input impedance resulting in the inverting amplifier shown in Fig. 20-9. The output voltage is then proportional to the input voltage and to the ratio between the feedback impedance and input impedance. By selecting appropriate impedance values this inverting amplifier may also be used as an inverting attenuator.

In most applications the input impedance and feedback impedance used for an inverting amplifier are resistors and the amplifier is essentially wideband. For large $\dfrac{Z_f}{Z_i}$ ratios the gain of the circuit will be high and the use of resistors for Z_i and Z_f may produce some high frequency losses due to stray capacitance. These losses can be minimized by the

compensating
adapter

use of the Tektronix Compensating Adapter (Part Number 013-0081-00). This adapter compensates for the stray capacitance associated with the internal Z_i and Z_f resistors.

If the Z_i and Z_f components of an inverting amplifier are combinations of resistors and capacitors, then these impedances will have nonlinear frequency response characteristics and the amplifier may therefore be tailored to suit particular frequency response requirements. Various values of resistors and capacitors for the input impedance and feedback impedances can be chosen to provide high-pass

bandwidth
modification

filtering, low-pass filtering or other bandwidth modification in a biophysical measurement system.

If the input impedance and feedback impedance elements in an inverting amplifier are nonlinear with respect to voltage or current then the amplifier's gain will be nonlinear with respect to voltage or current. If the Type 3A8 is used with a Tektronix Log Adapter (PN 013-0067-00) the relationship between the input voltage and the output voltage of the operational amplifier will be

logarithmic display

logarithmic and thus allows the measurement of high amplitude signals mixed with low amplitude signals. Thus, biophysical signals spanning three voltage decades can be displayed logarithmically on an oscilloscope.

20.8 INTEGRATION AND DIFFERENTIATION WITH OPERATIONAL AMPLIFIERS

Operational amplifier circuits are particularly suited to integration and differentiation applications, that is, where the output voltage is proportional to either the integral or the differential of the input voltage.

An operational amplifier used as an integrator is shown in Fig. 20-9. The output voltage is proportional to the integral of the input voltage in volt-seconds during the time interval t_1-t_2. In

integration

integration it is necessary to specify this time interval, for the output voltage will continue to increase as long as a positive input voltage is present and, when the input voltage is removed, the output voltage will remain at a fixed level. In order to return this integrator output to zero, allowing the integrator to once more perform useful integration, the feedback capacitor of the integrator must be discharged either by a manual switch, a reed switch controlled by other circuitry or by the use of the Tektronix Gating Adapter (PN 013-0068-00). This gating adapter is designed for use with one of the operational amplifiers in the Type 3A8 and allows repetitive signals with a net integral other than zero to be integrated and displayed. The use of an integrator in biophysical measurements is further discussed in Chapter 12, Section 12.6, when discussing the electromyogram produced by a voluntary muscular action. Details of a simple gating adapter for use with an integrator are given in Chapter 29.

differenti-
ation
An operational amplifier may be used for differentiation as shown in Fig. 20-8B. The output voltage is then proportional to the differential of the input voltage, i.e., it is proportional to the rate of change of the input voltage in volts per second. Differentiation is rarely used directly in biophysical measurement systems due to the excessive noise it produces, however, indirectly it may be used to intensity compensate an oscilloscope display as discussed in Chapter 21.

The values of R and C used in integration and differentiation circuits depend upon the amplitude and frequency of the input signal and are usually selected experimentally to produce an adequate output voltage without attempting to exceed the operational amplifier's ±25-V output capability.

21

OSCILLOSCOPES

A physiological signal, after having been amplified
and processed, can be displayed and/or recorded on
many different devices as shown in Chapter 15,
Fig. 15-1. The oscilloscope is the most popular of
these devices and is incorporated within almost
every biophysical instrumentation system. In many
cases, the oscilloscope is the final component in
an instrumentation system. However, as an
oscilloscope does not provide a permanent record,
some form of recorder, such as a magnetic tape
recorder, a graphic recorder or an oscilloscope
camera, may also be necessary.

This chapter should not be regarded as a general
reference on oscilloscopes. It specifically
discusses the unique characteristics of oscilloscopes
as they relate to biophysical measurements. Many
companion volumes relating to either oscilloscope
measurement concepts or oscilloscope circuit concepts
are available from Tektronix; a complete list of
reference these volumes is given at the end of this book.
material We would also recommend *Typical Oscilloscope
Circuitry*, published by Tektronix, as a good single
reference on oscilloscope principles. Also, the
Tektronix catalog of oscilloscopes and associated
instruments provides a useful reference while listing
the characteristics of all of Tektronix' products.
This catalog includes a list of US field engineering
offices and international field offices and
distributors where additional information on
oscilloscopes and their applications are available.
Tektronix also produces an applications oriented
periodical entitled *TEKSCOPE*. For regular receipt
of this periodical, contact the before-mentioned
field office or distributor in your area.

The three principal components of a simple oscilloscope are the vertical amplifier, the horizontal amplifier/sweep generator and the cathode-ray tube. These components are discussed in detail in the before-mentioned references and the characteristics of amplifiers suited for use with physiological signals are further discussed in Chapter 19. An oscilloscope may contain an amplifier suitable for displaying a physiological signal directly from electrodes or transducers (such as the Type 5031) or it may incorporate a simpler amplifier which can be used in conjunction with a preamplifier for physiological applications.

It is important to differentiate between three essentially similar devices that present information on a cathode-ray tube: the conventional oscilloscope, the slave oscilloscope and the display device. A conventional oscilloscope is a versatile instrument incorporating a vertical amplifier and a horizontal amplifier/sweep generator in conjunction with a CRT display. Both the vertical amplifier and the sweep generator's characteristics can be varied over wide ranges by controls on the front of the oscilloscope. A slave oscilloscope's function is to provide a conventional oscilloscope with an additional CRT display. The slave oscilloscope is coupled to a conventional oscilloscope in such a way as to present the same display on its CRT as presented on the CRT of the conventional oscilloscope. The coupling between a slave oscilloscope and "master" conventional oscilloscope is usually achieved with high impedance probes limiting the physical distance between two oscilloscopes to 12 feet. A display device is somewhat similar to a slave oscilloscope; however, it is intended to be used in a remote location for the display of any signals on x-y-z coordinates. These signals may be derived via low impedance outputs from an oscilloscope or they may be derived from other instrumentation systems.

21.1 OSCILLOSCOPE VERTICAL AMPLIFIERS

To preserve linearity and focus on a CRT display, it is necessary to drive both the CRT vertical deflection plates and horizontal deflection plates with differential signals. The oscilloscope amplifier in its simplest form may simply amplify an input signal and present this amplified signal to the CRT deflection plates as a differential signal.

To obtain various deflection sensitivities the gain of this amplifier must be variable both in discrete steps and over a continuous range. A simple oscilloscope vertical amplifier, such as the Tektronix Type 3A75 Amplifier, accepts an input signal into a 1-MΩ impedance and attenuates this signal by discrete steps to 50 mV/cm of deflection on the CRT display. The 3A75 then amplifies the signal and provides a differential output of +10 V/cm and -10 V/cm to the CRT deflection plates. The CRT thus receives a total deflection voltage of 20 V/cm.

multitrace scopes

Multitrace amplifiers allow the presentation of several channels of information by switching a CRT display between several amplifiers in such a manner that the display appears to a viewer as being produced by several separate CRT deflection systems within the one CRT. The Tektronix Type 3A3 Dual Trace Differential Amplifier switches or "chops" between two amplifiers at a 0.2 MHz rate. The CRT display thus consists of a sequence of 2.5 μs segments from each channel in turn. At the maximum sweep speed commonly used for physiological signals of 0.1 ms/cm, the two traces on the CRT appear as continuous lines since 20 chopped segments occur during each centimeter of sweep and therefore the separate segments cannot be discerned. Four-trace amplifiers such as the Type 3A74 incorporate switching circuitry to switch between four separate amplifiers, presenting a four-channel display on a CRT.

21.2 OSCILLOSCOPE HORIZONTAL AMPLIFIERS/SWEEP GENERATORS

The oscilloscope horizontal system, as with the vertical system, requires a differential voltage to deflect the CRT beam and is similar to the vertical amplifier with the exception that the variable and step gain controls are often omitted (a constant level of sweep signal is usually used in conjunction with the horizontal amplifier). A sweep generator provides a ramp to the horizontal amplifier, thus causing a linear horizontal sweep across the CRT face. For physiological applications the sweep generator ramp duration should be variable from 50 seconds to less than 1 ms, providing sweep speeds from 5 s/div to 0.1 ms/div. Some applications may require faster sweep speeds; most Tektronix sweep generators are capable of sweeping to at least 1 μs/div.

A SIMULATED ECG DISPLAYED AT 0.5s/DIV
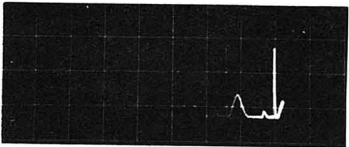
USABLE PERSISTENCE FOR 1.5s MAXIMUM

(A) OSCILLOSCOPE WITH P31 PHOSPHOR
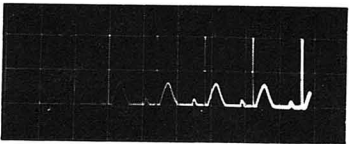
USABLE PERSISTENCE FOR 4s MAXIMUM

(B) OSCILLOSCOPE WITH P7 PHOSPHOR
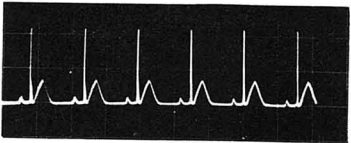
EFFECTIVELY HAS INFINITE PERSISTENCE

(C) STORAGE OSCILLOSCOPE SUCH AS
    TYPE 564B

NONSTORED
TRIGGERED DISPLAY
SHOWING NOISE →
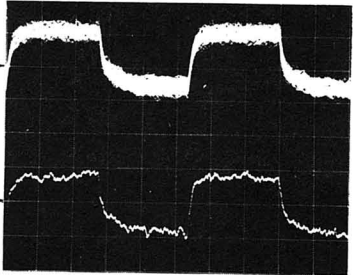
SAME SIGNAL →
AS ABOVE
STORED FROM
A SINGLE SWEEP

(D) STORAGE DISPLAYS MAY HELP TO REDUCE APPARENT
    NOISE ON A TRACE AS SHOWN BY THE NONSTORED
    AND STORED DISPLAYS PRESENTED ABOVE

Fig. 21-1. Persistence and storage.

21.3 OSCILLOSCOPE CRT DISPLAYS

Most conventional oscilloscope CRT displays are intended to be viewed at a distance from 2 feet to 6 feet and are presented in a format approximately 10 centimeters wide and 8 centimeters high. Special large screen CRT displays are available for viewing at greater distances or for viewing by a number of observers.

persistence

The persistence of a CRT display refers to the length of time that the CRT phosphor continues to emit light after the CRT beam has excited the phosphor. Referring to Fig. 21-1A, the standard phosphor supplied with many Tektronix oscilloscopes is a P31 phosphor which provides usable persistence for up to 1.5 seconds; however, for most practical purposes its persistence is considered to be under 0.5 seconds. Most Tektronix oscilloscopes are available on special order with a Type P7 phosphor (see Fig. 21-1B) which provides considerably greater persistence than the

storage

P31 phosphor. A storage oscilloscope such as the Tektronix Type 564B or Type 5031 effectively provides infinite persistence (see Fig. 21-1C) and should be considered for all oscilloscope applications requiring sweep speeds below 10 ms/div or where the information displayed is transient in nature. The storage oscilloscope stores a trace on the CRT phosphor until such time as an erase control is manually activated or activated by a remote electrical signal. As shown in Fig. 21-1D, the storage oscilloscope may also be useful in place of a conventional oscilloscope in reducing the apparent noise on a CRT display.

21.4 TYPICAL OSCILLOSCOPES

Type 5031

Oscilloscopes representative of Tektronix products
suited to biophysical measurements are shown in
Fig. 21-2, 21-3 and 21-4. The Tektronix Type 5031
Dual Beam Storage Oscilloscope shown in Fig. 21-2 is
a non-plug-in storage oscilloscope providing a
general purpose instrument particularly suited to
biophysical measurements. This oscilloscope is
referred to as a dual-beam oscilloscope, having two
separate vertical deflection systems and two separate
pairs of vertical deflection plates within the CRT.
This contrasts to the dual-trace amplifier discussed
earlier that presents two channels of information
via a beam switching circuit. Each vertical amplifier
has a maximum sensitivity of 10 μV/div with a
differential input impedance of 2 MΩ and a common
mode input impedance of 0.5 MΩ (modifiable to 5 MΩ).
No electrodes or other input coupling items are
provided as standard accessories with this product.

560 series

The Tektronix 560 Oscilloscope series shown in
Fig. 21-3 is representative of the plug-in
oscilloscope concept. Both the vertical amplifier
and the horizontal amplifier/sweep generator
characteristics in the 564B Storage Oscilloscope
and 561B conventional Oscilloscope can be varied
over broad ranges by the use of various vertical
and time-base plug-in units. Fig. 21-3 shows only
those plug-in units that are suited to biophysical
measurements. The 560 series of oscilloscopes
includes many more vertical and time-base plug-in
units for specialized, non-physiological applications.

The Type 565 Oscilloscope shown in Fig. 21-3 does
not utilize a time-base plug-in unit; however, it
is a dual-beam oscilloscope (having two separate
and independent deflection systems) utilizing two
vertical plug-in units. A Type 565, when used in
conjunction with, for example, two Type 3A74 four-
trace plug-in units, can present a total of eight
traces on its CRT.
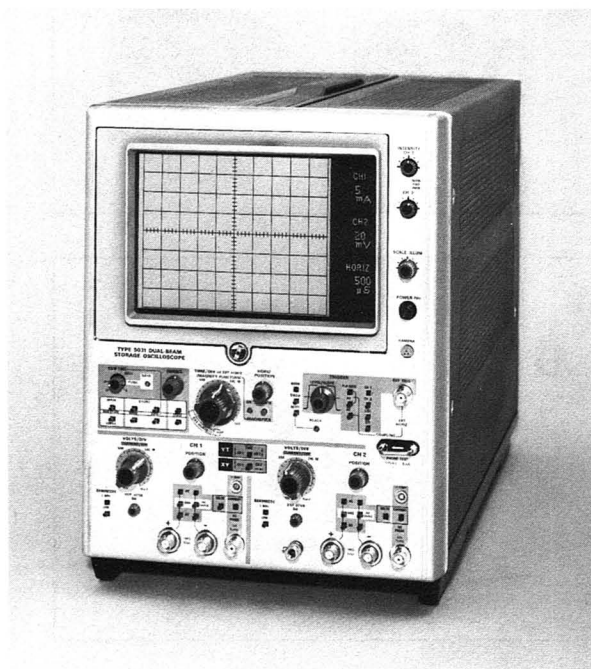
ALSO AVAILABLE IN A RACK
MOUNTING OR LOW PROFILE
CABINET CONFIGURATION.

TYPE R5031

ALSO AVAILABLE AS A
NONSTORAGE OSCILLOSCOPE
IN A RACK MOUNTING OR
LOW PROFILE CABINET
CONFIGURATION

TYPE R5030

BISTABLE SPLIT-SCREEN STORAGE AND CONVENTIONAL DISPLAYS
WITH VARIABLE VIEWING TIME AND AUTOMATIC ERASE

SCALE FACTOR READOUT ADJACENT TO THE CRT

DUAL BEAM DIFFERENTIAL. EACH BEAM TO 10µV/DIV AT 1MHz
OR 5kHz WITH 100,000:1 CMRR

SELECTABLE X-Y MODE AT 10µV/DIV

Fig. 21-2. The Tektronix Type 5031 Dual Beam Storage Oscilloscope.

344
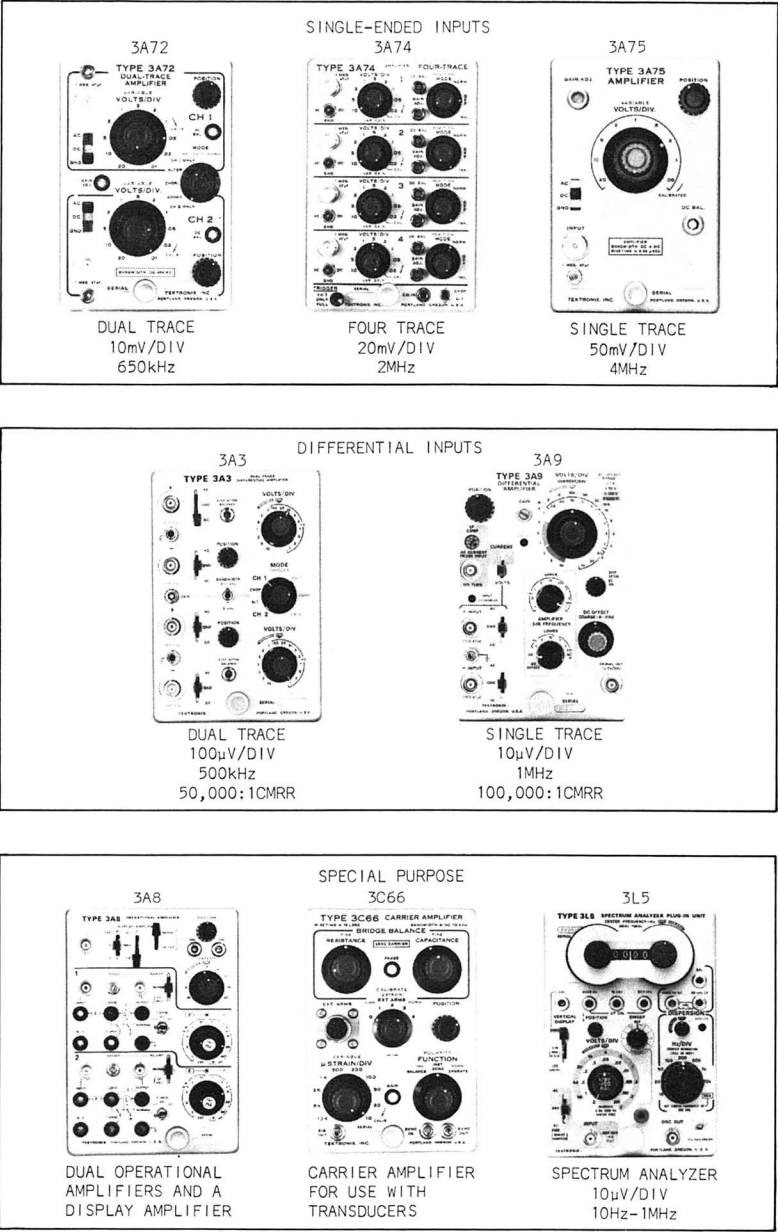
Fig. 21-3. Part of the Tektronix 560 Series of oscilloscopes.

PLUG-IN OSCILLOSCOPES

TIME-BASE PLUG-IN UNITS

564B
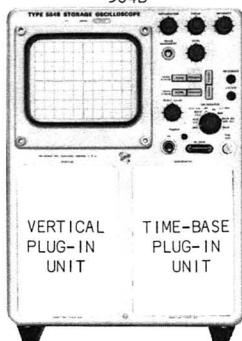
BISTABLE SPLIT-
SCREEN STORAGE
AND CONVENTIONAL
DISPLAYS

AVAILABLE WITH
VARIABLE VIEW-
ING TIME AND
AUTO ERASE

USES INTERCHANGE-
ABLE VERTICAL AND
TIME-BASE PLUG-IN
UNITS

561B
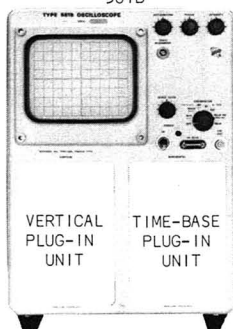
CONVENTIONAL
NONSTORAGE
DISPLAYS ONLY

USES INTERCHANGE-
ABLE VERTICAL AND
TIME-BASE PLUG-IN
UNITS

2B67
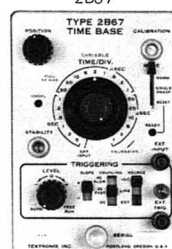
1μs/DIV-5s/DIV
5X MAGNIFIER
SINGLE SWEEP
LIMITED USE AS
AN AMPLIFIER

3B4
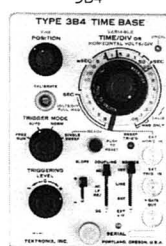
0.2μs/DIV-5s/DIV
50X MAGNIFIER
SINGLE SWEEP
AS AN AMPLIFIER
0.2V/DIV-5V/DIV

565
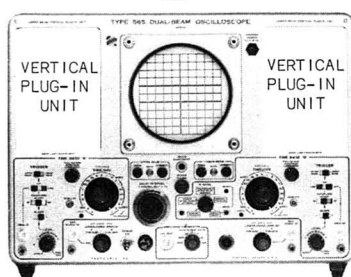
DUAL-BEAM OSCILLOSCOPE HAVING TWO
VERTICAL AND TWO HORIZONTAL SYSTEMS.

CONVENTIONAL NONSTORAGE DISPLAY ONLY

USES INTERCHANGEABLE VERTICAL PLUG-IN
UNITS

HORIZONTAL SYSTEMS - 1μs/DIV-5s/DIV
10X MAGNIFIER
SWEEP DELAY
LIMITED USE AS
AN AMPLIFIER

MONITORS ECG, EEG OR RELATIVE BLOOD PRESSURE
ALARM ON LOSS OF ECG OR PRESSURE SIGNAL
BATTERY OPERATED PORTABLE - 12.5 POUNDS
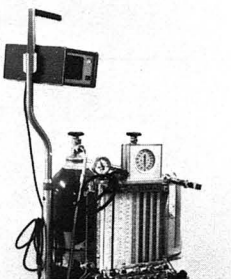
THE TYPE 410 MONITOR POSITIONED AT THE 5ft
LEVEL ON AN ANETHESIOLOGIST'S GAS
MACHINE DEMONSTRATING USE OF THE MOUNTING
STAND SHOWN AS AN OPTIONAL ACCESSORY
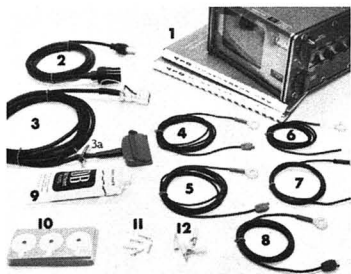
THE TYPE 410 WITH STANDARD ACCESSORIES

REF	QTY	DESCRIPTION	PART NUMBER
1	2	INSTRUCTION MANUALS	070-0658-00
2	1	POWER CABLE ASSEMBLY - 7ft	161-0058-00
3	1	PATIENT CABLE ASSEMBLY - 10ft	012-0120-00
3A		INCLUDES SHEET CLAMP	344-0146-00
4	1	ELECTRODE RL GREEN - 6ft	012-0121-25
5	1	" LL RED - 6ft	012-0121-22
6	1	" RA WHITE - 4ft	012-0121-29
7	1	" LA BLACK - 4ft	012-0121-20
8	1	" C BROWN - 4ft	012-0121-21
9	1	TUBE ELECTRODE PASTE	006-1098-00
10	1	PACKAGE 100 ADHESIVE RINGS	006-1099-00
11	5	PLATE ADAPTER	103-0079-00
12	5	NEEDLE ADAPTERS	103-0108-00
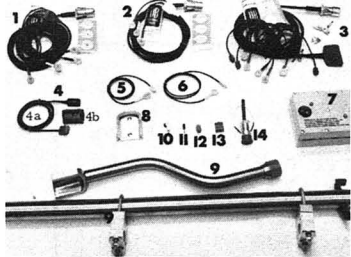
OPTIONAL ACCESSORIES FOR USE WITH THE
TYPE 410 MONITOR

REF	DESCRIPTION	PART NUMBER
1	LIMB LEAD CABLE PACKAGE LIMB LEAD CABLE WITH FOUR ELECTRODES INSTALLED, ELECTRODE PASTE AND ADHESIVE RINGS (FOR ECG MONITORING IN SURGERY)	012-0184-00
2	CHEST LEAD CABLE PACKAGE CHEST LEAD CABLE WITH THREE ELECTRODES INSTALLED, ELECTRODE PASTE AND ADHESIVE RINGS (FOR ECG MONITORING IN INTENSIVE CARE)	012-0183-00
3	PATIENT CABLE PACKAGE SAME AS STANDARD ACCESSORY PACKAGE (ABOVE) LESS POWER CORD AND MANUALS.	012-0185-00
4	PULSE SENSOR ASSEMBLY	015-0104-00
4A	CONSISTING OF PULSE SENSOR	015-0102-00
4B	AND FINGER HOLDER	015-0103-00
5	ELECTRODE +EEG YELLOW -2½ft	012-0121-24
6	" -EEG " 2½ft	012-0121-23
7	BATTERY PACK	016-0107-02
8	MOUNTING CUP	407-0393-01
9	MOUNTING STAND	016-0110-00
10	BARE WIRE ADAPTER	103-0080-00
11	PLUG - MINI-PHONE	134-0079-00
12	PLUG 2 PIN ELECTRODE	134-0089-00
13	PLUG 7 PIN AUXILIARY	134-0090-00
14	PLUG OUTPUT	131-0551-01

Fig. 21-4. The Tektronix Type 410 Physiological Monitor.

Type 410   The Tektronix Type 410 Physiological Monitor shown
            in Fig. 21-4 is a specialized oscilloscope intended
            for monitoring during surgery or in intensive care.
            The Type 410 provides a minimum of operator controls
            to allow its use by nontechnical personnel within
            the hospital environment. A full complement of
            electrodes and a patient cable are provided as
            standard accessories with many optional accessories
            available as shown in Fig. 21-4.

            Full specifications on the oscilloscopes discussed
            here are given in the current Tektronix catalog of
            oscilloscopes and associated instruments.
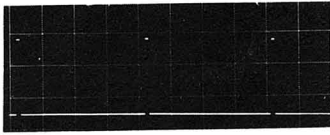
A STANDARD OSCILLOSCOPE DISPLAY OF
A 0.5ms WIDE PULSE AT A SWEEP SPEED
OF 5ms/cm (NO RATE INTENSIFICATION)
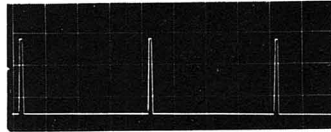
OSCILLOSCOPE DISPLAY OF THE SAME
WAVEFORM AS SHOWN AT LEFT WITH RATE
INTENSIFICATION

IF NEGATIVE INTENSITY MODULATING
SIGNAL IS REQUIRED, REVERSE DIODES
IN ABSOLUTE-VALUE AMPLIFIER
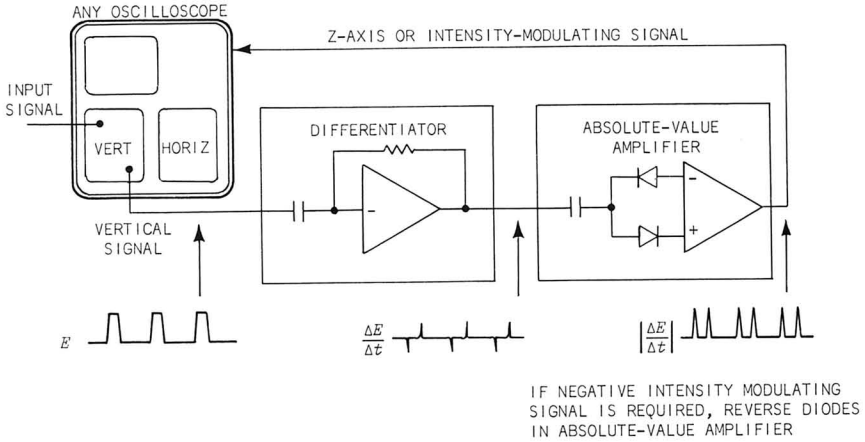
(A) PRINCIPLE
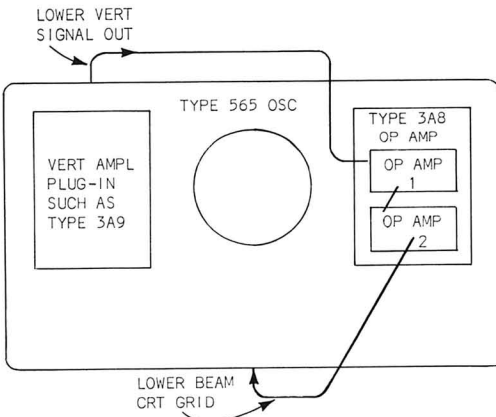
OPERATIONAL AMPLIFIER 1 OPERATES
AS DIFFERENTIATOR WITH Z_i = .001μF
AND Z_f = 200kΩ

OPERATIONAL AMPLIFIER 2 OPERATES
AS AN ABSOLUTE-VALUE AMPLIFIER
USING THE ABSOLUTE-VALUE ADAPTER
DESCRIBED IN CHAPTER 29

(B) A TYPICAL RATE INTENSIFIED SYSTEM

Fig. 21-5. Rate intensification.

21.5 RATE INTENSIFICATION

The left-hand oscilloscope waveform shown in Fig. 21-5 is typical of a pulse display on an oscilloscope. Only the baseline and top of the pulse are displayed on the CRT because during the transitions, the CRT beam is moving 100 times faster than during the baseline or pulse top periods. This difference in rate results in a difference in intensity on the CRT display.

differenti-
ation and
absolute
value
amplification
In order for the baseline, the pulse top and the transitions to be all presented on the CRT at the same intensity, the intensity of the CRT display must be made proportional to the rate of change of the vertical information. Fig. 21-5A shows the principle of rate intensification. A differentiator provides a signal proportional to the rate of change of the vertical signal. The output on the differentiator is, however, direction sensitive and must be "full-wave rectified" in an absolute-value amplifier to produce an output proportional to the absolute rate of change of the vertical signal. This output is then used in conjunction with the z-axis input of an oscilloscope to intensity modulate the CRT beam.

practical
system
The above principle can be applied in practice as shown in Fig. 21-5B. The CRT photographs shown in Fig. 21-5 were taken with this system. The differentiator used consists of one of the operational amplifiers in a Type 3A8 Operational Amplifier plug-in unit operating as a differentiator. The differentiator's input impedance and feedback impedance are selected by the controls on the front of the 3A8 at .001 μF and 0.2 MΩ, respectively. The output from this operational amplifier is then fed to the second operational amplifier on the Type 3A8 which is used as an absolute-value amplifier in conjunction with the absolute-value adapter described in Chapter 29. The output from this absolute-value amplifier is then fed into the lower-beam CRT grid to modulate the lower beam intensity. Rate intensification is particularly useful when attempting to photograph a CRT display or when many channels of information are presented on a single CRT display.
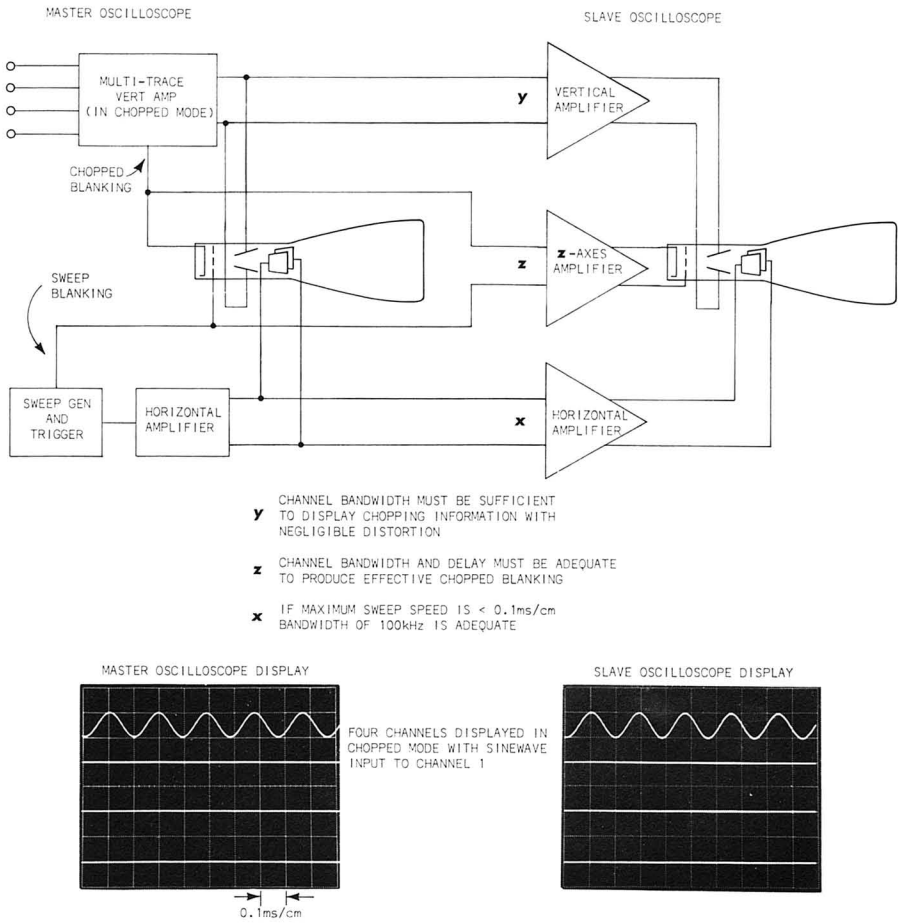
Fig. 21-6. Oscilloscope slaving considerations.

21.6 SLAVE OSCILLOSCOPES

additional
CRT display

The purpose of the slave oscilloscope, as mentioned previously, is to provide a conventional oscilloscope with an additional CRT display. A slave oscilloscope should, thus, contain only those controls necessary to alter the characteristics of the slave oscilloscope display, that is, intensity, focus and perhaps astigmatism and scale illumination. Although a slave oscilloscope will contain a vertical amplifier, a horizontal amplifier and a z-axis amplifier, it is unnecessary to vary the characteristics of these amplifiers after the slave oscilloscope has been coupled to the master oscilloscope; thus, these amplifiers need no external controls. A block diagram of a slave oscilloscope system is shown in Fig. 21-6 and the ideal relationship between a master oscilloscope display and the slave oscilloscope display is also shown in Fig. 21-6. Any change in the master oscilloscope display should be perfectly reproduced by the slave oscilloscope display.

nondistorting
coupling

high cost

When considering slave oscilloscopes, two basic rules should generally be adhered to. First, the required signal from the master oscilloscope should be coupled to the slave oscilloscope via high-impedance, low-capacity probes so as not to interfere with the characteristics of the master oscilloscope. Tektronix X10 or X100 probes are recommended as, even with 12-foot lengths, they provide a load on the master oscilloscope of 10 MΩ and less than 13 pF. Invariably, the master oscilloscope is not designed to allow low-impedance probes or direct interconnection and the use of either will degrade the performance of the master oscilloscope. The second basic rule of slave oscilloscopes relates to economy. In general, the cost of a master oscilloscope/slave oscilloscope system will be equal to or will exceed the cost of two separate oscilloscopes. Thus, the criteria for using a slave oscilloscope system should not be based on economy but rather on the desirability of having one set of operator controls controlling two CRT displays.

The slave oscilloscope block diagram shown in
Fig. 21-6 shows both the vertical (y) and horizontal
(x) signals coupled differentially from the master
oscilloscope deflection plates to the slave
oscilloscope deflection amplifiers. It is desirable
to use differential inputs to eliminate unwanted
signals that may appear on the master oscilloscope
amplifier outputs but may not appear on the master
oscilloscope display due to the inherent differential
rejection capability of the deflection plates. If
a multitrace vertical amplifier is used in the
master oscilloscope vertical, the slave oscilloscope
vertical amplifier must have sufficient bandpass to
allow the chopped vertical signal to be coupled into
the slave oscilloscope without risetime reduction.
For most Tektronix multitrace vertical amplifiers,
this necessitates a slave oscilloscope vertical
amplifier bandpass considerably in excess of 1 MHz.
The bandwidth of the slave oscilloscope horizontal
amplifier is noncritical in biophysical measurements
applications and 0.1 MHz is usually sufficient.

A master oscilloscope display uses intensity
modulation via the CRT grid and/or cathode to turn
the master oscilloscope CRT beam off during sweep
retrace and during the interval that the multitrace
vertical amplifier is switching from one channel to
another. These signals are referred to as sweep
blanking and chopped blanking, respectively. The
z-axis amplifier must be capable of accepting both
of these blanking signals to allow the slave
oscilloscope CRT beam to be turned off in
synchronism with the master oscilloscope CRT beam.
The chopped blanking signal consists of a narrow
pulse occurring during the chopping interval; this
narrow pulse has a risetime considerably less than
1 μs, thus, a z-axis amplifier bandwidth of several
megahertz is necessary to allow faithful chopped

blanking of the slave oscilloscope display. If a
single trace vertical amplifier is used, then no
chopped blanking signal is involved and only the
sweep blanking need be coupled to the z-axis
amplifier. Under these conditions the bandwidth of
the z-axis amplifier would be noncritical for
biophysical measurement applications and, as with
the horizontal amplifier, 0.1 MHz should be
sufficient.

Various compromises may be made to the basic slave
oscilloscope block diagram shown in Fig. 21-6;
however, it is important to fully understand any
performance degradation associated with these
compromises. All Tektronix oscilloscopes utilize
regulated power supplies and high quality vertical
and horizontal amplifiers. Thus, it is not
absolutely necessary to couple the vertical and
horizontal signals from the master oscilloscope to
differential the slave oscilloscope in a differential mode.
versus Negligible distortion will be observed on the slave
single-probe oscilloscope display if the signals are coupled
coupling between master and slave with single probes. It
should be noted, however, that the output from the
master oscilloscope vertical and horizontal
amplifiers will contain a standing DC level and it
is necessary for the slave vertical and horizontal
amplifiers to be able to cancel out this standing
DC level if only single-probe coupling is used.
Most Tektronix vertical and horizontal amplifiers
suitable for use with slave oscilloscopes incorporate
sufficient range in the DC BALANCE control and
POSITION control to cancel out this DC voltage. It
is, however, necessary to fully consider this and
to ensure that sufficient position range is
available in the slave oscilloscope to center the
display on the slave CRT.

354

"FUZZY" TRACES DUE
TO LACK OF CHOPPED BLANKING
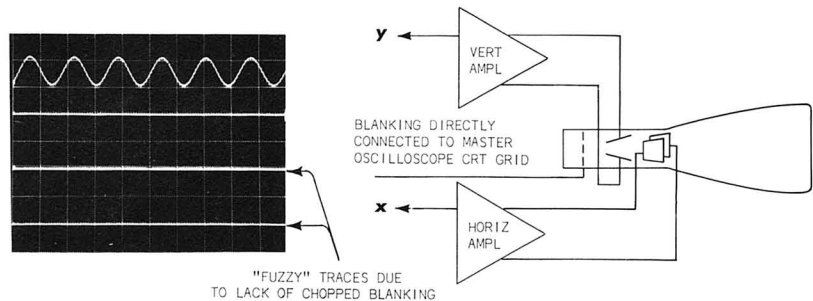
(A) DIRECT SWEEP-BLANKING INTERCONNECTION. SINGLE
PROBES USED TO PICK OFF **x** AND **y** INFORMATION.

INFORMATION DISPLAYED
DURING SWEEP RETRACE DUE
TO LACK OF SWEEP BLANKING

"FUZZY" TRACES DUE
TO LACK OF CHOPPED BLANKING
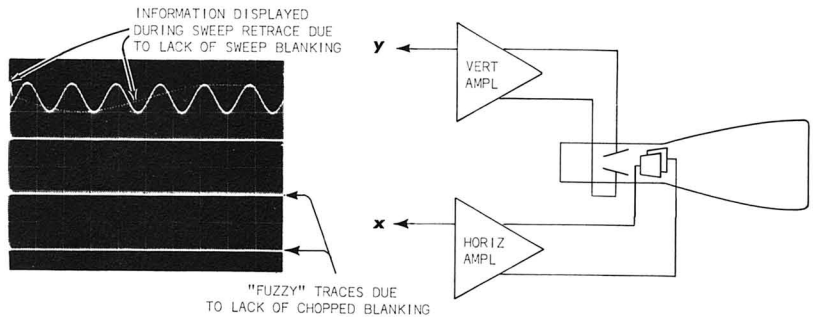
(B) NO **z** AXIS INFORMATION. SINGLE PROBES USED TO
PICK OFF **x** AND **y** INFORMATION
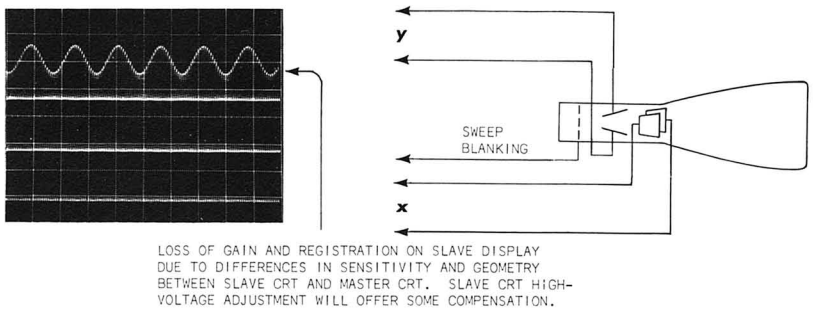
LOSS OF GAIN AND REGISTRATION ON SLAVE DISPLAY
DUE TO DIFFERENCES IN SENSITIVITY AND GEOMETRY
BETWEEN SLAVE CRT AND MASTER CRT. SLAVE CRT HIGH-
VOLTAGE ADJUSTMENT WILL OFFER SOME COMPENSATION.

MASTER OSCILLOSCOPE DISPLAY IS ALSO DEGRADED
DUE TO CAPACITIVE LOADING REDUCING THE VERTICAL
AMPLIFIER'S BANDWIDTH

(C) DIRECT INTERCONNECTION OF **x, y** AND SWEEP BLANKING BETWEEN
MASTER AND SLAVE - 12ft LONG UNSHIELDED WIRES.

Fig. 21-7. Oscilloscope slaving compromises.

Many slave oscilloscopes are used with multitrace vertical amplifiers in the master oscilloscope but with no chopped blanking signal coupled between the master oscilloscope and the slave oscilloscope. No chopped blanking is, therefore, presented to the slave CRT resulting in a fuzziness on the traces as shown in Fig. 21-7A and B. Due to the characteristics of the film used to record the waveform shown in Fig. 21-7A and B, this fuzziness does not appear to be excessive; however, a viewer would find the lack of chopped blanking quite annoying when viewing the slave oscilloscope display.

fuzziness

If a slave oscilloscope system does not incorporate sweep blanking, as shown in Fig. 21-7B, sweep retrace can clearly be observed on the slave oscilloscope CRT and under certain input signal conditions can be misleading as well as annoying.

no sweep
blanking

Many attempts have been made to create a slave oscilloscope by simply paralleling an additional CRT with the master oscilloscope CRT; the results, as shown in Fig. 21-7C, are far from ideal. This system also degrades the performance of the master oscilloscope as well as presenting a poor display on the slave oscilloscope and is, therefore, not generally recommended. Also, as the additional CRT will not have exactly the same characteristics as the master oscilloscope CRT, the slave oscilloscope display will never be a direct replica of the master oscilloscope display.

paralleling

slave
systems

Two practical slave oscilloscope systems are shown
in Figs. 21-8 and 21-9. The single-channel slave
oscilloscope system shown in Fig. 21-8 uses either
a Type 601 Storage Display Unit or a Type 602
(conventional) Display Unit as a slave oscilloscope.
The master oscilloscope consists of either a Type
561B (Conventional) Oscilloscope or a Type 564B
Storage Oscilloscope with a 2B67 Time Base plug-in
unit and any single-channel vertical amplifier
plug-in unit. The x and y bandwidth of the slave
oscilloscope is 1 MHz, sufficient for a single-
channel operation. The y input to the slave
oscilloscope is derived via a single probe from the
output of the vertical amplifier. The position
control within the slave oscilloscope is sufficient
to cancel out the standing DC level at the output
of the vertical amplifier. The x input to the slave
oscilloscope is derived in the same manner as for the
y input. As the system is only intended for single-
channel operation, no chopped blanking is required
and the z input to the slave oscilloscope must simply
provide sweep retrace blanking. The z input of the
slave oscilloscope is derived via a probe from the
blanking pulse output of the time base plug-in unit.
In order to provide sufficient gain range in the 601
or 602 display unit, R25 and R75 in these units must
be shorted out with a wire link.

probe
coupling

The Type 601 and 602 Storage Display Units have
100,000-ohm input impedance amplifiers. Tektronix
P6006 (X10) and P6007 (X100) probes are intended for
use with amplifiers having a 1-megohm imput
impedance. These probes can, however, be
successfully used with the 601 or 602 display unit;
the X10 probe actually providing an attenuation of
X100 and the X100 probe actually providing an
attenuation of X200. It should be noted that the
compensation on all probes used in any slave
oscilloscope system must be correctly adjusted.
Correct probe compensation adjustment for probes
used with the z-axis amplifier may require an
additional oscilloscope to view the z-axis waveform
during adjustment.

MASTER OSCILLOSCOPE

SLAVE OSCILLOSCOPE

564B STORAGE OSCILLOSCOPE

3A9
DIFFERENTIAL
AMPLIFIER

2B67
TIME
BASE

y

TYPES 601
OR 602
DISPLAY UNITS

13   *z*

21   21

x

R25 AND R75 IN THE 601 OR 602 DISPLAY
UNIT DEFLECTION AMPLIFIERS TO BE
REPLACED BY A SHORT CIRCUIT.

PIN NUMBER ON PLUG-IN
INTERCONNECTION SOCKET
IN 564B OSCILLOSCOPE

21
TEKTRONIX X100 PROBE
12ft LONG - TYPE P6007
TEKTRONIX PN 010-0154-00

TEKTRONIX X10 PROBE
12ft LONG - TYPE P6006
TEKTRONIX PN 010-0148-00

564B MASTER OSCILLOSCOPE DISPLAY

TYPE 602 SLAVE OSCILLOSCOPE DISPLAY

Fig. 21-8. A single channel slave oscilloscope system.
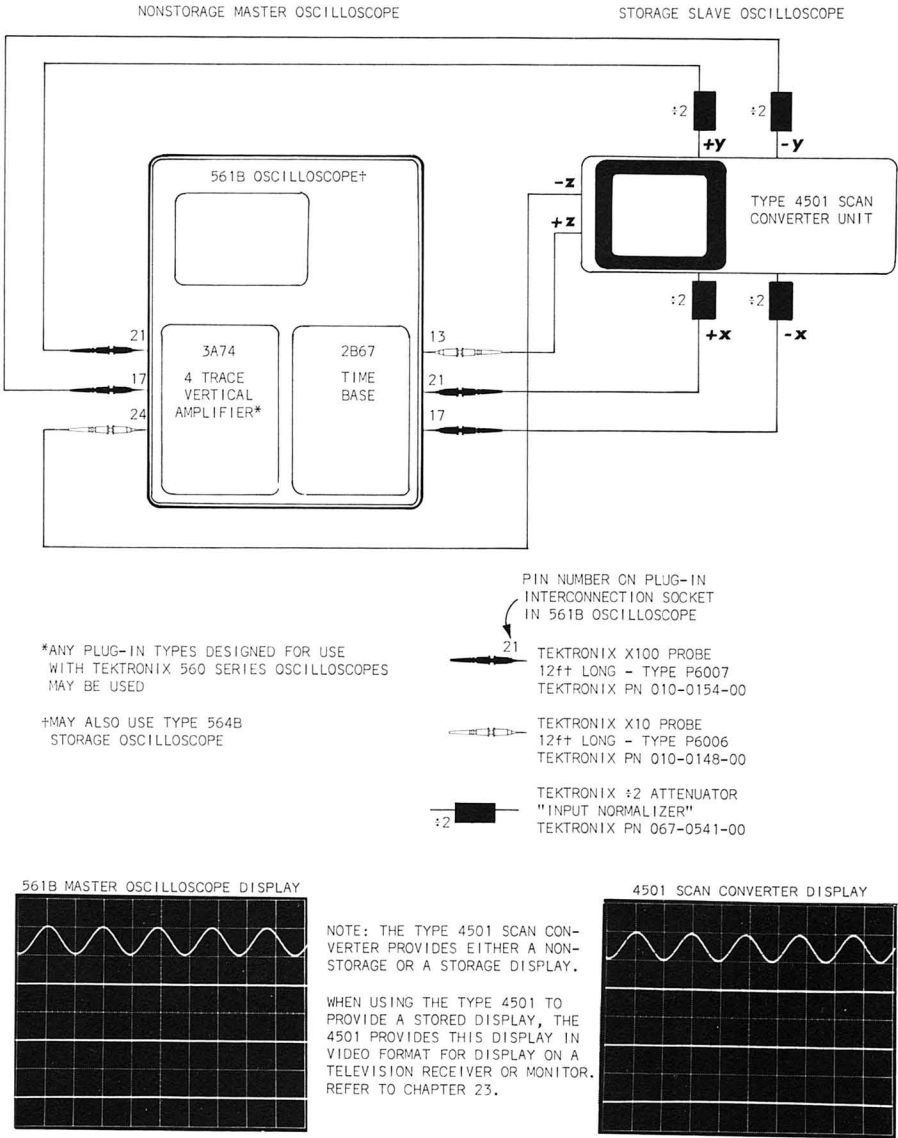
Fig. 21-9. A multichannel slave oscilloscope system.

A multichannel slave oscilloscope system is shown
in Fig. 21-9. The Type 4501 Scan Converter Unit is
basically a five-inch storage display unit having
high input impedance (1 MΩ, 47 pF) differential input
into the x, y, and z axes. The bandwidth of the x
and y amplifiers is 10 MHz and the bandwidth of the
z-axis amplifier is 5 MHz. The x, y, and z amplifier
characteristics of the Type 4501 make it particularly
suited as a slave oscilloscope as these amplifier
characteristics exceed all of the desirable criteria
discussed earlier in this section. This system shown
in Fig. 21-9 is self-explanatory; either X10 or X100
probes are used to couple the necessary signals from
the master oscilloscope to the slave oscilloscope
and, where necessary, Tektronix "input normalizers"
are used to provide an additional 2X attenuation.
The x and y signals are coupled differentially between
the master oscilloscope and the slave oscilloscope
and the differential z-axis amplifier is actually
utilized as a summing amplifier to add together the
master oscilloscope chopped blanking waveform and
sweep retrace blanking waveform.

The Type 4501 Scan Converter Unit, as well as being
basically a five-inch storage display unit, can
internally scan information stored on the storage
CRT and present this information on bright, large
screen, TV monitors or receivers. The Type 4501 Scan
Converter Unit is discussed in greater detail in
Chapter 23.
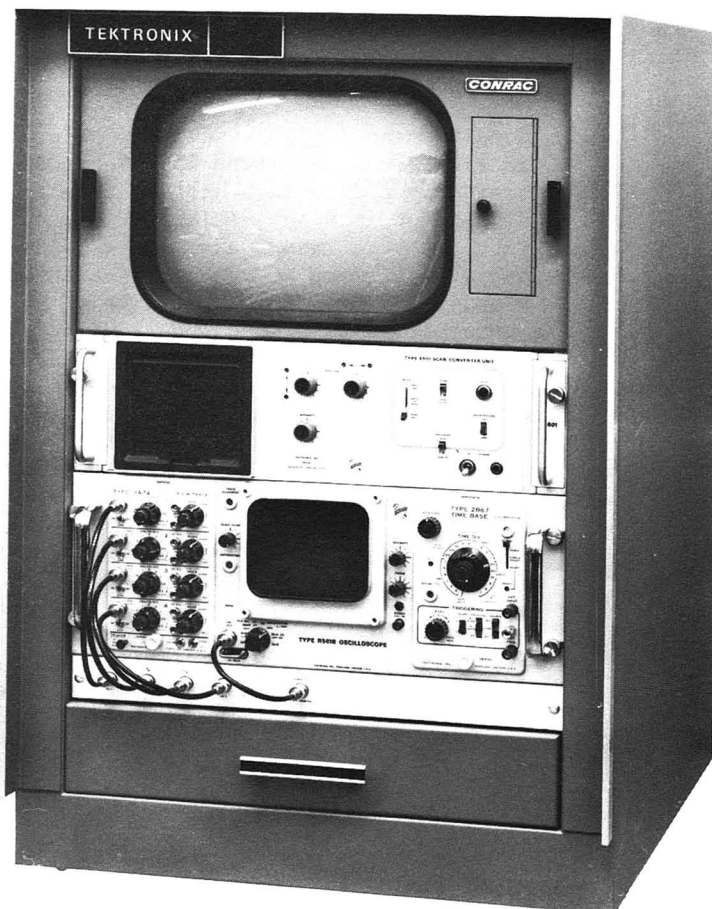
Fig. 21-10. A complete four-channel slave oscilloscope.

22

PULSE GENERATORS AND STIMULATORS

stimulation   As covered in various chapters in Section II of
this book, the passage of electric current through
various types of cells may cause these cells to
depolarize and to generate an action potential. This
passage of electric current is normally achieved
biologically by a nerve impulse propagated from within
the central nervous system; however, it may be
artifically produced by a stimulator. The stimulator
may be a self-contained instrument or it may be
controlled by a separate pulse generator. For the
purpose of this discussion of stimulus systems the
pulse generator portion will be defined as that part
that generates the pulse waveshape required for
stimulation and the stimulator will be defined as
that part that amplifies the signal to a level
sufficient for artificial biological stimulation.
Typically, a pulse generator may generate a 10 volt
pulse; this pulse is then amplified within the
stimulator to 100 volts or more for stimulation.
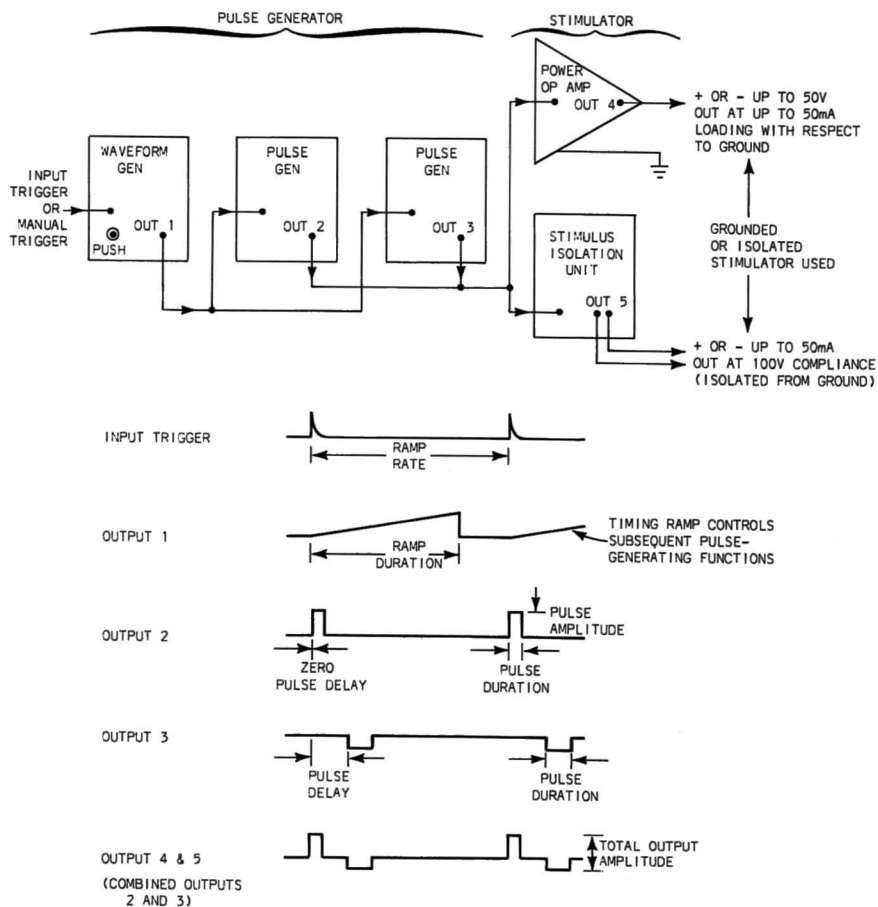
Fig. 22-1. A typical physiological stimulus system.

22.1 STIMULUS SYSTEMS

typical
system

A typical physiological stimulus system is shown in
Fig. 22-1. The pulse generator section of this
stimulus system uses a ramp generator to control two
pulse generators. These pulse generators are
triggered at discrete ramp levels, thus providing
delay between the initiation of the ramp and the
initiation of the pulse. With the system shown in
Fig. 22-1, the ramp can either be generated
repetitively, generated via an external trigger
source or can be initiated manually from a push-
button on the waveform generator. The output pulses
from the pulse generators are combined to produce 2
positive pulses or a positive/negative pulse pair.

A stimulus system suited to most biophysical
measurement requirements may have the following
characteristics:

PULSE GENERATION

-- Double pulse capability with each pulse
   independently controlled.

-- Pulse widths from .01 ms to 300 ms.

-- Pulse rise- and falltimes <1 μs.

-- Pulse repetition rate, controlled by ramp
   rate, from 1,000 Hz to 0.1 Hz with single
   pulse capability.

-- Delay between pulses in a double-pulse
   format from ≈0 to 300 ms.

-- Output pulses from constant current sources
   to allow mixing. Current amplitude and
   voltage compliance must be compatible with
   power operational amplifiers and stimulus
   isolation units. Usually 10 mA with
   compliance to 10 volts is adequate.

STIMULATION VIA POWER OPERATIONAL AMPLIFIER

-- Input configuration to allow summation of
   either voltage source or current source
   outputs.

-- Slew rate of >50 V/μs with feedback elements
   adjusted for maximum output from a 1 volt
   or 10 mA input pulse.

-- Up to ±50 V/±50 mA capability as either a voltage source or as a current source.

STIMULATION VIA STIMULUS ISOLATION UNIT

-- Tristable output capability for stimulation with pulses having positive and negative components.

-- Isolation between output and source of <5 pF, $>10^{10}\Omega$ and between output and ground of <30 pF, $>10^{10}\Omega$.

-- Up to ±100 V/±50 mA capability as either a voltage source or as a current source.

brain
stimulation

Open cortical stimulation (stimulation of the exposed brain) requires considerably less stimulus energy than peripheral stimulation (stimulation of the arms and legs). Cortical stimulation requires from 0.1 to 10 V/0.1-10 mA while peripheral stimulation requires from 10 to 200 V/1-50 mA. Cortical stimulus energy should be maintained below 1 watt peak and 0.3 μcoulombs to avoid thermal and/or electrolytic injury to brain cells. Peripheral stimulation energy should be maintained below pain thresholds whenever possible.

22.2 TEKTRONIX 160 SERIES PULSE GENERATORS

Type 162
operation

The Tektronix Type 162 Waveform Generator shown in Fig. 22-2 may be operated in one of three separate modes as shown in Fig. 22-3. When operating in the recurrent mode, the ramp will immediately reset itself to zero after completing a ramp cycle and begin a second ramp. This process will continue indefinitely. When the Type 162 is operated in the gated mode, the ramp generator will free-run for the duration of the input gate pulse, a ramp beginning when the input gate pulse goes positive and the last ramp being completed after the gate pulse again returns to zero. With the Type 162 operating in the triggered mode, the ramp generator may be triggered from external positive trigger pulses or by internal trigger pulses generated by manually depressing the trigger push-button on the front of the generator. One ramp is generated for each input trigger, thus, if only one input trigger is received, or if the front panel manual trigger push-button is depressed only once, a single ramp will be generated.

| GENERATES A TIM-ING RAMP FROM 100μs TO 10s DURATION | GENERATES + OR – OUTPUT PULSE OF UP TO 50V WITH RISE- AND FALL-TIMES ≈ 1μs. DURATION VARIABLE 10μs TO 0.1s | GENERATES + OR – OUTPUT PULSE OF UP TO 25V WITH RISE- AND FALL-TIMES ≈ 0.2μs. DURATION VARIABLE 1μs TO 10ms | DISPLAY UNIT Y = 0.05-50V/DIV X = 11-15V/DIV Z = 50V | POWER SUPPLY FOR 160 SERIES GENERATORS |
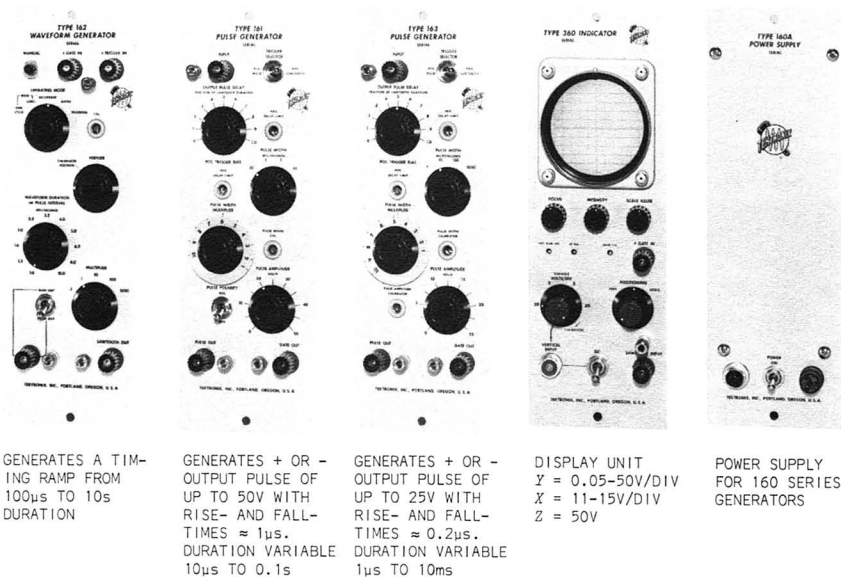
Fig. 22-2. Tektronix "160" Series Pulse Generator system.

(A) RAMP GENERATOR FREE RUNS

(B) RAMP GENERATOR GATED

(C) RAMP GENERATOR TRIGGERED
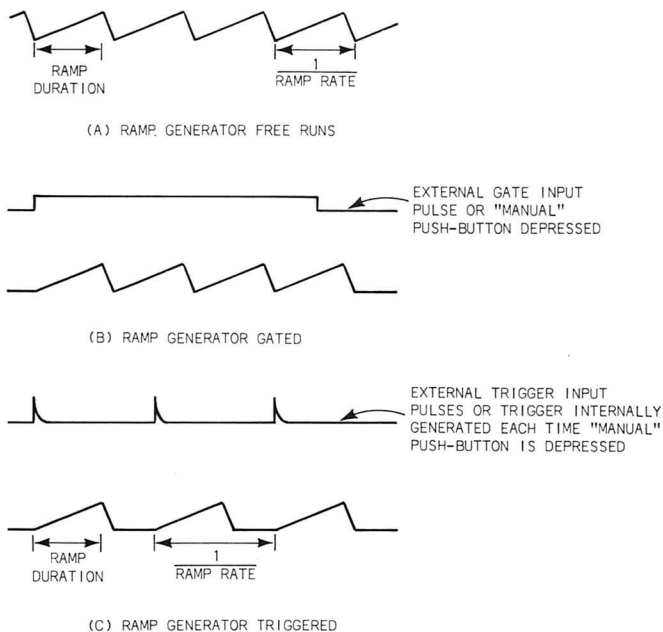
Fig. 22-3. Timing ramp control.
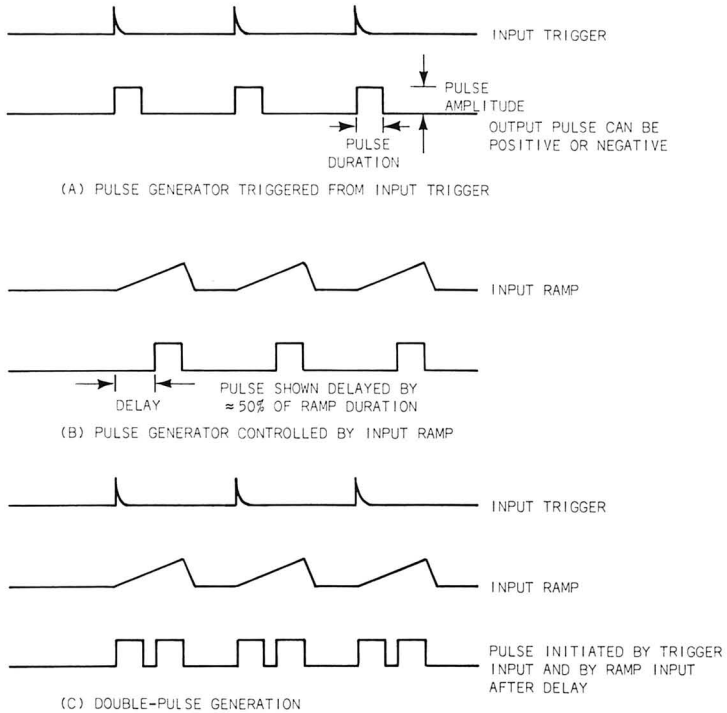
Fig. 22-4. Output pulse control.

Type 161 operation

The Type 161 Pulse Generator is shown in Fig. 22-2 and its various operating modes are shown in Fig. 22-4. The pulse generator may be triggered from an external positive trigger source, in which case the pulse output will begin when the trigger pulse goes positive. The pulse generation cycle may be initiated by an input ramp. Level-sensitive trigger circuitry within the pulse generator triggers at a precise level on this ramp and thereby provides a delay corresponding to the time taken for the ramp to reach the preset triggering level. The pulse generator may be triggered by both an input trigger and an input ramp, thus providing double pulse generation.

Types 163,
360 and
160A

The Type 163 Pulse Generator shown in Fig. 22-2 is similar to the Type 161 but provides a lower amplitude output pulse with faster rise- and falltimes. The Type 360 Indicator is basically a display device; its primary purpose being to display outputs from the 160 Series to simplify output pulse waveform adjustment. The Type 160A Power Supply provides regulated power to other 160 Series modules.

22.3 HIGH E OR I OUTPUT VIA A POWER OPERATIONAL AMPLIFIER

The output level from most pulse generators is insufficient for tissue stimulation, therefore, some form of pulse amplification is required between the pulse generator and the stimulating electrodes. A power operational amplifier, as discussed in Chapter 20, is ideally suited to this application.

constant
current
and
constant
voltage

Chapter 12, Section 12.1, discusses the characteristics of stimulators required for EMG use and points out that two different forms of stimulators are currently in use -- constant voltage stimulators and constant current stimulators. Although tissue can be stimulated with either a constant voltage source or a constant current source, it is difficult to correlate results obtained when using these different stimulating sources. Thus, while both systems are entirely adequate, the clinical electromyographer or physiologist will normally have a preference for either constant voltage or constant current stimulation, this preference probably being influenced by his earlier medical training.

INPUT PULSES FROM VOLTAGE
SOURCES

IF INPUT PULSES ARE FROM CURRENT
SOURCES, OMIT R1, R2 AND R3 (EQUAL ZERO)

$$e_o = R_f \left(I_1 + I_2 + I_3 \right)$$

$$e_o = - R_f \left(\frac{E_1}{R1} + \frac{E_2}{R2} + \frac{E_3}{R3} \right)$$

R_f CONTROLS TOTAL OUTPUT VOLTAGE
AND R1, R2 AND R3 CONTROL PROPORTION
OF EACH INPUT SIGNAL IN THE OUTPUT

(A) VOLTAGE SOURCE OUTPUT

INPUT PULSES FROM VOLTAGE
SOURCES

IF INPUT PULSES ARE FROM CURRENT
SOURCES, GROUND E_2 INPUT TERMINAL,
SUM ALL CURRENTS INTO E_1 INPUT TERMINAL
AND ADD R_c FROM E_1 INPUT TO GROUND
VALUE OF $R_c = R_s$

$$I_o = - \left(\frac{R_f}{R1} \right) \left(I_1 + I_2 + \cdots \right)$$

$$R = \frac{R_f R2}{R1} - R_s$$

$$I_o = \frac{\left(E_2 - E_1 \right)}{R_s} \frac{R_f}{R1}$$

(B) CURRENT SOURCE OUTPUT

Fig. 22-5. Pulse combination and amplification with a power operational amplifier.
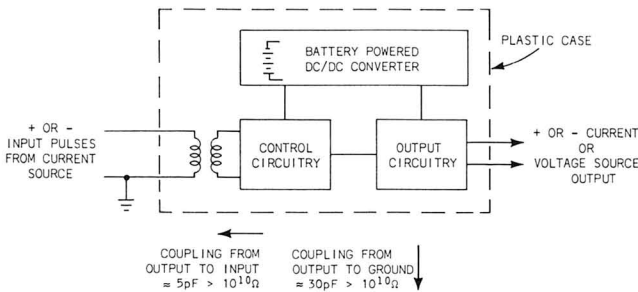
Fig 22-6. Equivalent circuit of a stimulus isolation unit.

constant
voltage

A power operational amplifier used in a voltage
source configuration for constant voltage
stimulation is shown in Fig. 22-5A. The input to
the operational amplifier can be derived from either
voltage sources or current sources and the output
from the operational amplifier exhibits essentially
constant voltage characteristics with an output
impedance of considerably less than 10 ohms in most
power operational amplifiers. Power operational
amplifiers suitable for tissue stimulation should
have an output capability of at least 50 V at 50 mA.
It should be noted that the operational amplifier
configuration shown in Fig. 22-5A inverts the output
signal with respect to the input; in most instances,
reversal of the stimulating electrodes effectively
inverts the stimulating signal back to its original
polarity.

constant
current

A power operational amplifier used in a current
source configuration for constant current
stimulation is shown in Fig. 22-5B. Since an
operational amplifier inherently provides a voltage
source output, a current sensing resistor (R_s) is
used to detect the output current and to modify
feedback around the operational amplifier to keep
this current constant. Inputs from either voltage
sources or current sources may be used. The output
exhibiting constant current characteristics should
have an output impedance typically in excess of
0.1 MΩ. An operational amplifier suitable for
constant current stimulation should have an output
capability of up to 50 mA with a voltage compliance
to at least 50 V.

22.4 HIGH E OR I OUTPUT VIA A STIMULUS ISOLATION UNIT

character-
istics

A stimulus isolation unit provides amplification of
an input pulse or pulses and isolates the output
pulse from both ground and the input pulse source.
The reasons for output pulse isolation are covered
in detail in Chapter 12, Section 12.2. An equivalent
circuit for a typical stimulus isolation unit is
shown in Fig. 22-6. Since stimulus isolation may be
used in applications requiring high amplitude
stimulation into a relatively high impedance (such
as in stimulation of the extremities using electrodes
on the surface of the skin), the output from a
stimulus isolation unit should have a greater voltage
capability with a somewhat lower current capability
than the output from a power operational amplifier.

A typical stimulus isolation unit operating in a constant voltage stimulation mode may produce a voltage output of up to 100 V with a current capability of 50 mA. Similarly, when operating in a constant current stimulation mode, it may provide a current output of up to 50 mA with compliance up to 100 V.

bistable
and
tristable

Stimulus isolation units are typically bistable or tristable devices. A bistable stimulus isolation unit offers either positive or negative outputs, having only two stable states -- output on and output off. A tristable stimulus isolation unit offers simultaneous positive and negative outputs, having three stable states -- output on positive, output off and output on negative. With a tristable stimulus isolation unit, the current or voltage provided by the "output-on-positive" pulse can be controlled independently from the current or voltage provided by the "output-on-negative" pulse.

22.5 CARDIAC PACEMAKERS

As discussed in Chapter 2 when dealing with the heart and the circulatory system, the steady rhythm of the heart is maintained by a biological "pacemaker" within the sinoatrial node of the heart. Failure of this pacemaker will cause the pumping action of the heart to be interfered with, causing a seizure and possible death.

Emergency resuscitation from a seizure can be accomplished by external electrical stimulation or

heart
stimulation

cardiac massage of the heart by trained medical
personnel. Since in many subjects seizures are
unpredictable, continuous stimulation is required
to ensure continuous reliable operation of the heart.
Long term stimulation from an external electrical
source requires a considerable level of stimulating
current to ensure that a sufficient level of current
passes directly through the heart, thus causing
considerable pain to the subject. This pain may be
avoided by using external electrical stimulation via
internal electrodes placed directly on the heart;
however, tissue irritation and rejection present a
problem that is difficult to overcome.

The modern cardiac pacemaker concept avoids tissue
irritation and rejection by surgically implanting
a pacemaker within the subject's abdomen and
connecting it to the heart via internal electrodes.
Modern internal pacemakers use mercury batteries
which can operate the pacemaker continuously for a
year or more. Since the pacemaker is placed in the
subject's abdomen rather than in the heart, a
relatively unsophisticated surgical procedure is
necessary on a routine basis to replace the pacemaker
with one having a fresh battery. The problem of

battery
life

determining just when an internal pacemaker's battery
is nearly discharged, and thus, when the pacemaker
must be replaced, has not yet been completely solved;
however, many researchers have proposed various
techniques for determining pacemaker end-of-life by
analyzing the pacemaker output pulse as recorded on
the surface of the body. At this stage there does
not appear to be sufficient evidence in favor of any
one method for it to be universally accepted.

The internal cardiac pacemaker consists of a transistorized blocking oscillator producing pulses of approximately 10 V in amplitude, a few milliseconds in width and at an approximate 60-per-minute rate. The equivalent circuit for a pacemaker and subject, together with a photograph of a commercial pacemaker, is shown in Fig. 22-7.

PACEMAKER   SUBJECT

$\approx 1.5 k\Omega$ $\approx 10\mu F$

$\approx 10\mu F$

E $\approx 10V$ $\approx 700\Omega$

EQUIVALENT
CIRCUIT

WIRES CONNECT PACEMAKER
OUTPUT TO ELECTRODES
ATTACHED TO THE HEART

ELECTRODE WIRES CAN BE
DISCONNECTED FROM PACEMAKER
TO ALLOW PACEMAKER TO BE
REPLACED BY SIMPLE SURGERY

PACEMAKER SURGICALLY
IMPLANTED IN SUBJECT'S
ABDOMINAL CAVITY

TYPICAL ECG FROM A SUBJECT WITH
AN IMPLANTED CARDIAC PACEMAKER

MYOCARDIAL
ELECTRODE
WIRES

SELF-LOCKING
SEALED
CONNECTORS

PACEMAKER
6.4 x 7.2 x 2.0c
WEIGHT = 180g

(WITH INTERNAL
NONREPLACEABLE
BATTERIES)

MYOCARDIAL
ELECTRODES

A COMMERCIAL PACEMAKER PRODUCED BY
ELECTRODYNE COMPANY INC., MASS., USA

Fig. 22-7. The implantable cardiac pacemaker.

LARGE PADDLE ELECTRODES ALLOW HIGH CURRENTS TO BE PASSED THROUGH SUBJECT WITH LOW CURRENT DENSITY AT THE ELECTRODE SITES. ALSO PROVIDES LOW ELECTRODE-SUBJECT RESISTANCE.
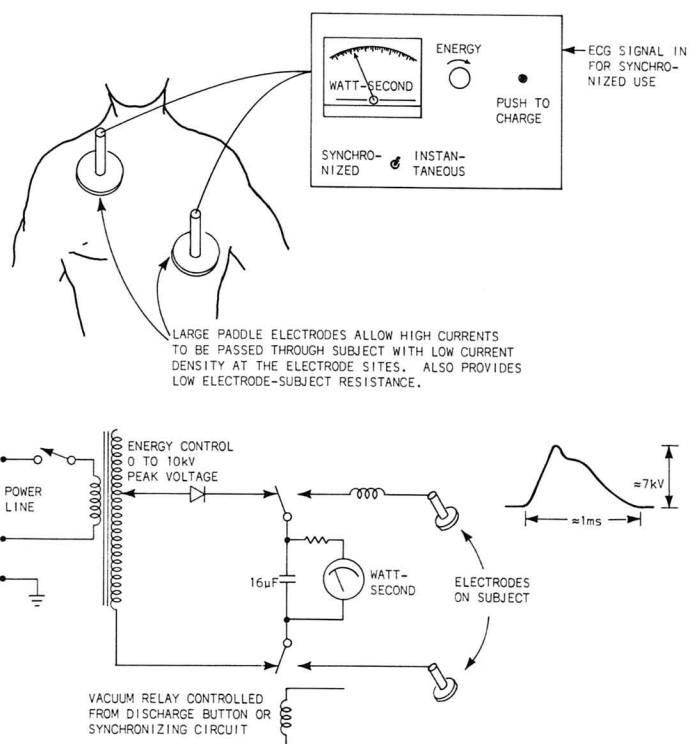
Fig. 22-8. The DC defibrillator.

22.6 CARDIAC DEFIBRILLATORS

ventricular
fibrillation

Resuscitation from either a heart seizure or from ventricular fibrillation can be accomplished by external electrical stimulation. Ventricular fibrillation is produced within the human heart due to a variety of causes, including accidental electrocution. When the heart is in ventricular fibrillation, individual portions of the ventricular muscle contract independently instead of synchronously and effective output of blood ceases. External stimulation can be achieved with a cardiac defibrillator which essentially consists of a capacitor charged to several thousand volts which is then discharged through the subject via large-surface-area "paddle" electrodes as shown in

watt-second
shocks

Fig. 22-8. The energy produced within a cardiac defibrillator is normally measured in watt-seconds

or joules and is equal to $0.5\,CE^2$, (C is the value of the storage capacitor and E is the voltage level to which it is charged). Defibrillation shocks from 200 to 400 watt-seconds are normally required to achieve satisfactory cardiac defibrillation. Not all of this energy will be available for dissipation at the subject as the efficiency of discharge will be considerably less than 100%; typically 20% to 70%.

The cardiac defibrillator may be operated in one of two modes: instantaneous or synchronized. When operated in the instantaneous mode, the energy from the charged capacitor is discharged through the subject when the discharge button (usually located on one of the hand-held paddle electrodes) is depressed. It has been found that cardiac defibrillation is most efficient if the defibrillating pulse occurs during the falling part of the ECG R-wave and that defibrillation can be detrimental to some subjects if it occurs during the T-wave.

instantaneous mode

With a cardiac defibrillator operating in the synchronized mode, the discharge pulse is not immediately applied to the subject after the discharge button has been depressed but is delayed to occur during the falling part of the following R-wave. Use of a defibrillator in the synchronized mode necessitates an ECG signal being applied to the defibrillator for synchronizing purposes. Thus, a defibrillator can only be used in the synchronized mode if the ECG signal generated by the subject is of sufficient quality to allow the synchronizing circuit within the defibrillator to detect the R-wave. Many cardiac disorders produce abnormal ECG signals having detectable R-waves. These disorders may often be remedied by using a defibrillator in the synchronized mode, the defibrillator pulse *forcing* the heart to revert to a normal operating rhythm. This process is known as cardioversion.

synchronized mode

cardio-version

23

DISPLAY DEVICES AND INDICATORS

Both display devices and indicators are used to
visually display the output from a biophysical
instrumentation system. The display device utilizes
a cathode-ray tube as the display medium, the
indicator uses some other visual device such as a
panel meter or a numerical readout device such as
Nixie tubes. The following material discusses
display devices from a biophysical measurements
viewpoint only; therefore, we would recommend a
companion volume entitled *Information Display
Concepts* published by Tektronix, Inc. as a *general
reference* source for display devices.

display
devices
versus
oscilloscopes

A display device may appear to be simply an
oscilloscope, and indeed, an oscilloscope is often
used as a display device. Display devices, however,
generally differ from oscilloscopes as the CRT
display within a display device is optimized for
display characteristics (resolution, contrast,
brightness and screen size) whereas, in contrast, the
CRT display within an oscilloscope is optimized for
measurement capability (vertical amplifier bandwidth
and writing rate). A display device is intended to
be used within a specific instrumentation system and
thus has no external controls to change the
characteristics of the display. An oscilloscope may,
however, contain numerous external controls to change
the characteristics of the display to suit the
desired measurement requirement.

5in CRT UNITS   8cm × 10cm

TYPE 601 STORAGE DISPLAY UNIT
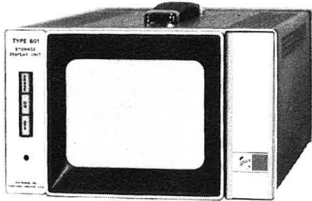
x AND y – 1V FULL–SCREEN SENSITIVITY
        – 100kHz BANDWIDTH
        – 100kΩ 50pF INPUT RC
z–AXIS – BISTABLE: ON > 1V OFF < 0.5V
        – > 100kHz EFFECTIVE BANDWIDTH
        – 100kΩ 50pF INPUT RC
CRT   – BISTABLE STORAGE
        – 12.5 STORED LINE–PAIRS/cm
         RESOLUTION
        – 5cm/ms WRITING SPEED

TYPE 602 DISPLAY UNIT
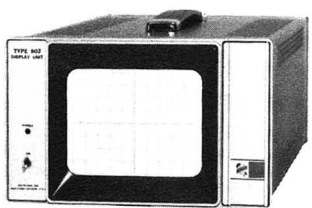
x AND y – 1V FULL–SCREEN SENSITIVITY
        – 1MHz BANDWIDTH
        – 100kΩ 30pF INPUT RC
z–AXIS – LINEAR: ON 1V RANGING TO OFF 0V
        – 1MHz BANDWIDTH
        – 100kΩ 70pF INPUT RC
CRT   – NONSTORAGE. P31 OR P7 PHOSPHOR
        – < .014 INCH TRACE WIDTH

11in CRT UNIT   16.2 × 21cm

TYPE 611 STORAGE DISPLAY UNIT

x AND y – 1V FULL–SCREEN SENSITIVITY
        – 100kΩ 60pF INPUT RC
z–AXIS – BISTABLE: ON > 1V. OFF < 0.5V
        – > 100kHz EFFECTIVE BANDWIDTH
        – 100kΩ 50pF INPUT RC
CRT   – BISTABLE STORAGE
        – 19 STORED LINE–PAIRS/cm
         RESOLUTION
        – 25cm/ms WRITING SPEED

TYPE 611 MOD 162C – HORIZONTAL FORMAT

(A)

SCAN CONVERTER - 5in CRT TO 14in OR 17in TV MONITOR

TYPE 4501 SCAN CONVERTER UNIT

TYPE 4501 WITH 17in CONRAC MONITOR

x AND y - 1V FULL-SCREEN
　　　　　　SENSITIVITY
　　　- 10MHz BANDWIDTH
　　　- 1MΩ 47pF INPUT RC
z-AXIS - LINEAR OR LIMITING:
　　　　　　ON 1V.　OFF 0V
　　　- 5MHz BANDWIDTH
　　　- 1MΩ 47pF INPUT RC
CRT　- BISTABLE STORAGE
　　　- 12.5 STORED LINE-
　　　　　PAIRS/cm RESOLUTION
　　　- 5cm/ms WRITING
　　　　　SPEED
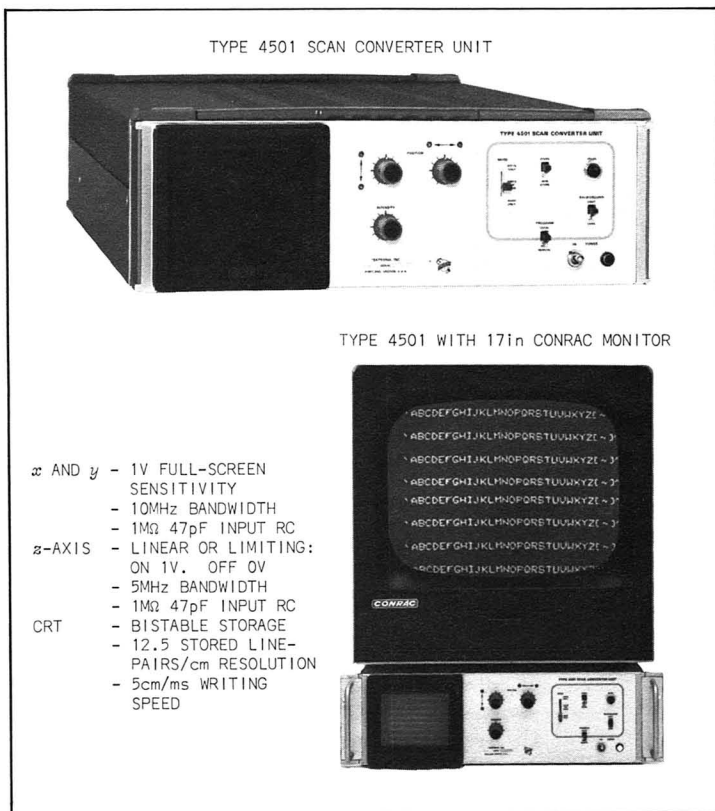
(B)

Fig. 23-1. Tektronix display units.

23.1 TEKTRONIX DISPLAY UNITS

display
unit
character-
istics

Tektronix currently produces four display units --
the Type 602 Display Unit, the Types 601 and 611
Storage Display Units and the Type 4501 Scan
Converter Unit. These four products are shown in
Fig. 23-1 together with a brief summary of their
principle characteristics. All Tektronix display
units require one volt of x and y input for full
screen deflection horizontally and vertically and
require one volt of z-axis input to turn the CRT
beam on and off. As these x, y and z sensitivities
are only adjustable over a relatively narrow range,

it is necessary to externally amplify or attenuate signals to a level consistent with the sensitivity of the display unit before coupling such signals to the display unit. Tektronix display units may be used to provide an additional CRT display for a conventional oscilloscope, that is, in a slave oscilloscope configuration as discussed in Chapter 21.

Types 601 and 602

The Type 602 Display Unit utilizes a nonstorage, high resolution, CRT providing a display area of 8 cm vertically by 10 cm horizontally. The Type 601 Storage Display Unit has similar characteristics to Type 602 Display Unit and incorporates a storage CRT to allow displayed information to be stored as discussed in Chapter 21. Erasure of stored information can either be accomplished manually with a push-button on the front of the Type 601 or may be accomplished by an electrical signal.

Type 611

The Type 611 Storage Display Unit provides four times the display area of the Type 601 by utilizing an 11-inch storage CRT. The display area of the Type 611 is 16.2 cm horizontally by 21 cm vertically; the instrument is also available in a horizontal format providing 16.2 cm vertically and 21 cm horizontally. A stored display on the Type 611 may be either erased by a push-button or remotely by an electrical signal.

Type 4501   The Type 4501 Scan Converter Unit (see Fig. 23-1B) is a unique instrument providing a large screen CRT display via a television system. The Type 4501 Scan Converter may basically be considered as a 5-inch storage display unit (similar to the Type 601) coupled with a television system to display information stored on the scan converter's CRT on a large screen television monitor or a commercial television receiver. The bright displays achieved via scan conversion are ideal for individual or group viewing under high ambient light conditions.

Readout from the storage CRT is accomplished by scanning the CRT storage target with a TV raster. All necessary scanning, sync, and video circuits are contained in the scan converter, providing a composite TV signal of the information stored on the CRT.

The TV signal is available as either EIA 525-line, 60-field format or in CCIR 625-line, 50-field format at 1 volt P-P. This signal can be used in conjunction with a studio-quality TV monitor for large screen display. Modulated RF at 55.25 MHz through 67.25 MHz is also produced to allow the use of commercial TV receivers for a display.

380

STORED DISPLAY
ON SCAN CONVERTER CRT
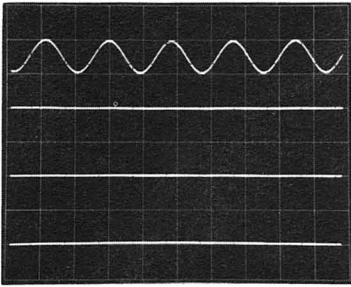
DISPLAY MAY BE REPRODUCED TO
ANY SIZE VIA TV MONITOR
PROJECTION SYSTEM OR TV
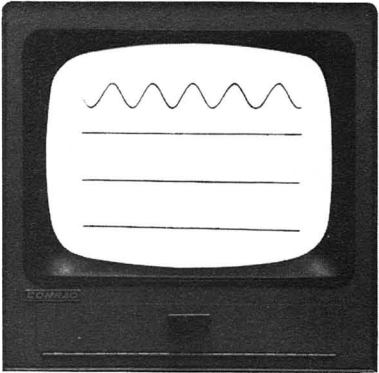
MULTI-CHANNEL OSCILLOSCOPE DISPLAYS
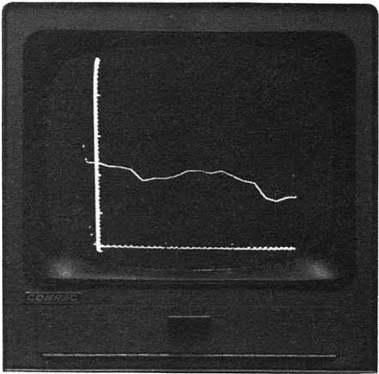
COMPUTER GENERATED DISPLAYS
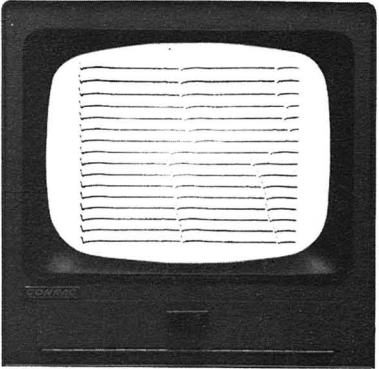
SUBJECT ECG OVER ONE MINUTE
(REFER TO FIG. 23-4)

Fig. 23-2. Type 4501 Scan Converter capabilities.

variety of
information

As shown in Fig. 23-2, the Type 4501 Scan Converter
or any other display unit can be used to present
many different types of information. The Type 4501
Scan Converter's application as a slave oscilloscope
is discussed in Chapter 21. A typical four-channel
slave oscilloscope display on a television monitor
is shown in Fig. 23-2 together with a computer
generated display of a subject's variation in
temperature over a 36-hour period and a stepped-
trace display of a subject's ECG measured over a one-
minute period using the system discussed later in
this chapter.

THE RESOLUTION OF AN OSCILLOSCOPE DISPLAY IS MEASURED IN
LINE-PAIRS AS SHOWN

OSCILLOSCOPE DISPLAY   TELEVISION

1 LINE-PAIR
(1 LINE + 1 SPACE)

2 SCAN LINES
NEEDED TO REPRODUCE
1 LINE AND 1 SPACE
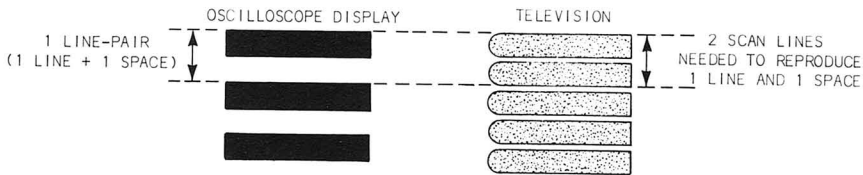
SCAN ON A TYPE 602 DISPLAY UNIT CRT SHOWING LINE-PAIR
RESOLUTION AND EFFECT OF CLOSER SCANNING

25-LINE/cm SCAN   36-LINE/cm SCAN   50-LINE/cm SCAN

RESOLUTION IS 25
LINE-PAIRS/cm AS
LINE AND SPACE ARE
THE SAME WIDTH

SCAN AT TWICE LINE-
PAIR RESOLUTION RATE
JUST MERGES INTO NON-
RESOLVABLE SCAN
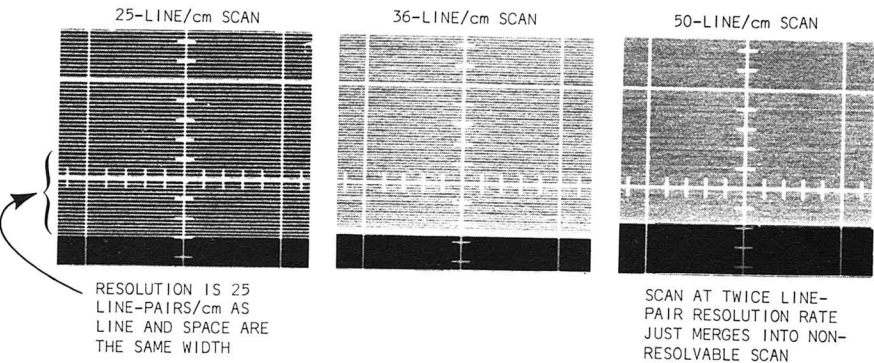
TOP LEFT-HAND QUADRANT OF HIGH RESOLUTION DISPLAYS ON
TEKTRONIX DISPLAY UNITS SHOWING RESOLUTION UNIFORMITY

TYPE 601 STORAGE   TYPE 602   TYPE 611 STORAGE
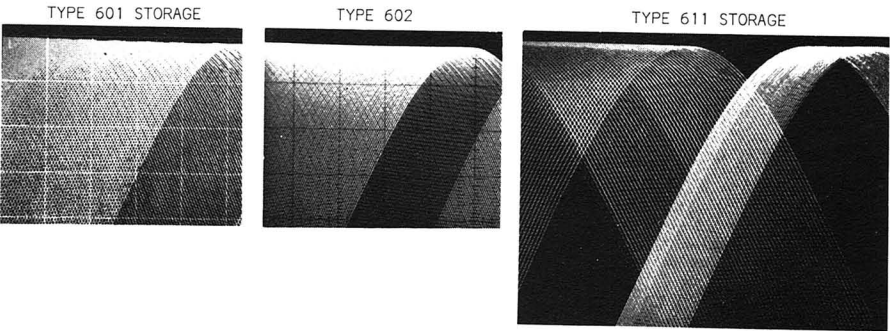
Fig. 23-3. Resolution of Tektronix display units.

23.2 RESOLUTION

line-pairs
versus
scan lines

The principal difference between the CRT display on an oscilloscope and a display unit is the increased brightness and resolution obtainable with a display unit as shown in Fig. 23-3. The resolution of a display unit or oscilloscope CRT display is specified in line-pairs, one line-pair being equal to one bright written line and one space of equal width. This specification should not be confused with scanning resolution specified in *lines* as associated with television systems; in such systems two scan lines are required to produce one line and one space.

determining
resolution

A scan of 25 lines/cm on a Type 602 Display Unit is shown in Fig. 23-3. From an observation of this display it appears that *the width of the dark space between these lines is equal to the width of the lines*, thus it can be said that this display has a resolution of 25 line-pairs/cm. If the scanning density is now increased to 36 lines/cm as shown on Fig. 23-3, the individual scanning lines can still be clearly discerned; however, it is apparent that the width of the scanning lines is somewhat greater than the width of the dark space between the scanning lines. At a scanning rate of 50 lines/cm the dark space between the scanning lines begins to disappear and the individual scanning lines begin to merge into a nonresolvable scan. The resolution of this particular CRT could be quoted as 25 line-pairs/cm or, when dealing with a television type display, as having a resolution capability of 50 lines/cm.

spot size

The spot size of the before referenced display is calculated to be .008 inches as the display is capable of 25 line-pairs/cm, that is, 50 spots/cm. It should be noted that the spot size specified for the Tektronix Type 602 Display Unit is .014 inches, almost twice the spot size calculated above. Spot size will vary from one CRT to another; it will also vary slightly over the total display area of the CRT and varies considerably with the intensity of the display. Although the typical spot size for a Type 602 Display Unit may be .008 inches, the specification of .014 inches represents the maximum spot size that would be encountered at any point on any Type 602 CRT when displayed at a relatively high intensity level.

384
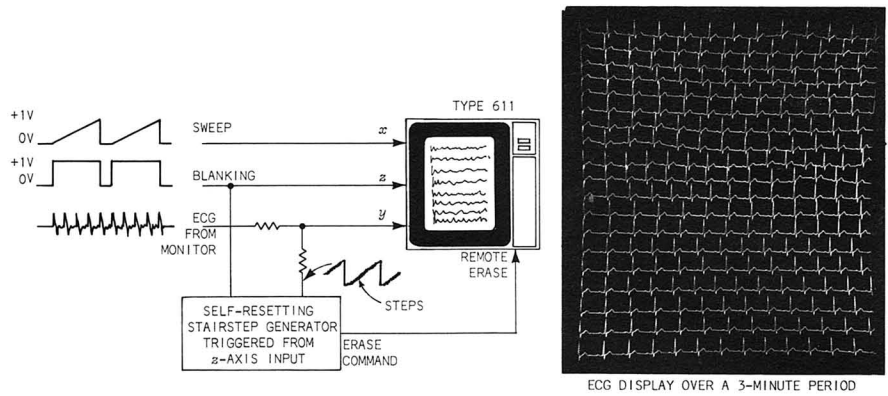
(A) USING THE TYPE 611 IN A SYSTEM

ECG DISPLAY OVER A 3-MINUTE PERIOD
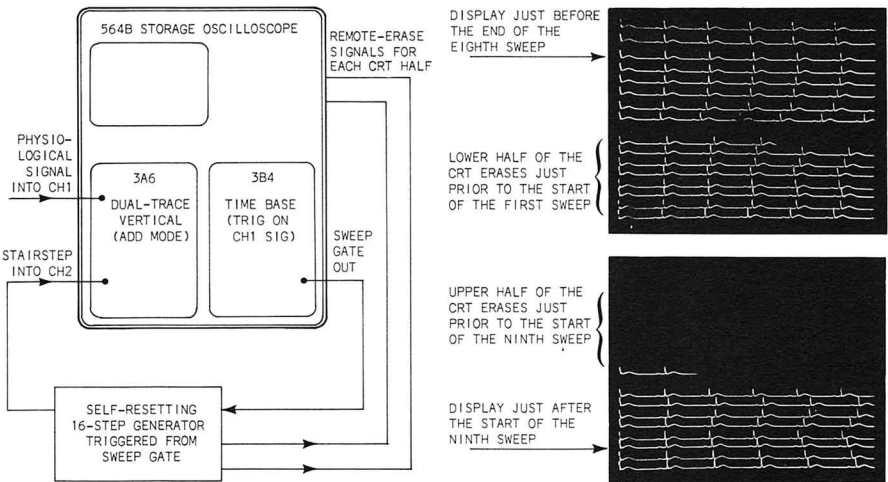
REFER TO CHAPTER 27 FOR DETAILS
OF THE STEP GENERATORS

(B) SELF-CONTAINED SYSTEM USING A TYPE 564B SPLIT-SCREEN STORAGE OSCILLOSCOPE

Fig. 23-4. Expanded x-axis storage-CRT displays.

The uniformity of resolution between the center of
a CRT display and the corners of the display is
particularly good with Tektronix display units as
shown in the lower series of photographs in Fig. 23-3.
An observation of these photographs will show that
little loss of resolution can be detected when
comparing the resolution at the center of the CRT
display to the resolution at the top left-hand edge
of the CRT display.

23.3 LARGE SCREEN EXPANDED-SWEEP DISPLAYS

The normal 5-inch CRT of most oscilloscopes limits
the amount of information that may be presented on
the CRT without confusion. The large screen
capabilities of the Type 611 Storage Display Unit
and the Type 4501 Scan Converter Unit allow a
greater quantity of information to be displayed.
This greater quantity of information may be derived
from many different channels of information or may be
derived from a single channel of information over a
prolonged period of time using vertical sweep stepping
as shown in Fig. 23-4.

sweep
stepping

The displays shown in Fig. 23-4 require the use of
a step generator capable of being updated with each
successive sweep. The step generator may also
contain an adjustment to allow the number of sweeps
to be varied to suit the display requirements.
Details of two self-resetting stair-step generators
are given in Chapter 29.

expanded
x axis

The expanded x-axis display shown in Fig. 23-4A
uses the Type 611 to store a large volume of analog
information on its 11-inch CRT. Although the display
shown represents a subject's ECG history over a three
minute period, the resolution of the Type 611 is such
as to allow similar displays to be presented over far
greater periods by using slower sweep speeds, a lower
level of ECG signal and less spacing between sweeps.

signal
density

An erase command from the step generator will erase
the CRT when the step generator resets to zero to
once again begin its stepping cycle or, alternatively,
the information may be held on the CRT for a
prolonged period or until the step generator is
manually reset to erase the information and begin
the cycle once again. While this system has the
advantage of a large screen display, all of the
information stored on the screen must be erased

before beginning a new series of sweeps, thus, the amount of "history" presented on the CRT varies from zero just after erasure to three minutes, as with the display shown in Fig. 23-4A when the step generator has almost completed its cycle and the screen is full of information. This disadvantage is overcome in the system shown in Fig. 23-4B.

The expanded x-axis storage display shown in Fig. 23-4B uses a Type 564B Storage Oscilloscope with a Type 3A6 vertical plug-in unit and a Type 3B4 horizontal plug-in unit in conjunction with a self-resetting 16-step generator. Although this system only uses a 5-inch CRT display, it has the advantage of being a self-contained system and, as described below, uses the split-screen storage feature of the Type 564B to alternately erase segments of the CRT.

The expanded x-axis display system shown in Fig. 23-4B provides 16 sweeps, the first sweep beginning at the bottom left-hand corner of the CRT and the last sweep finishing at the top right-hand corner of the CRT. Just prior to the completion of the last sweep the CRT will present the previous 16 sweeps of stored information. At the end of the last sweep the step generator will reset to zero

programmed
erase

and the lower half of the CRT will be erased, thus leaving only the previous 8 sweeps presented on the upper half of the CRT. The system will then commence to cycle through its first 8 sweeps, storing them on the lower half of the CRT. At the end of the eighth sweep the step generator will cause the upper half of the CRT to be erased, leaving only the information generated by the previous 8 sweeps on the lower half of the CRT. The step generator will then commence to cycle through the remaining 8 sweeps, storing information on the upper half of the CRT. The cycle is then repeated continuously. This system has the advantage that at least 8 sweeps are always presented on the storage CRT at any one time, thus the amount of "history" presented at any one time varies from 8 sweeps to 16 sweeps rather than, as with a system using a nonsplit-screen storage oscilloscope, from zero to 16 sweeps.

23.4 INDICATORS

moving-
coil meter

As stated previously, indicators may consist of a moving-coil panel meter or some form of digital display device. The moving-coil panel meter, while being considerably less expensive than a digital indicating device, is somewhat difficult to read and thus should only be used for the indication of noncritical parameters or where digital display devices would be uneconomical. Moving-coil meters may be directly coupled to other components in a biophysical monitoring system or may be coupled to these components via amplification circuits.

digital
meters and
indicators

Digital meters and indicators accept voltage levels from other components in a biophysical monitoring system and convert them to a format suitable for display in numeric form. They offer the advantage of showing the measured quantity in a form that can be read at a glance or at a considerable distance. The digital indicator is also free of any reading error that may occur when reading a moving-coil pointer against a fixed scale. Digital display indicators are generally custom built for specific systems. Electronics for Medicine, New York, produces a digital display unit for use in an operating room providing 6 channels of numeric display to indicate the subject's temperature, pulse rate, venous pressure, arterial diastolic pressure, arterial systolic pressure and mean arterial pressure. A numeric indicator device may consist of a CRT display unit and a character generator for characters on the CRT. The character generator may be a separate instrument or may utilize a digital computer to generate characters with the aid of digital to analog converters.

Digital displays are not the whole answer to data presentation. Particularly if the user is harassed or tired, they are easily misinterpreted. Further, they give no indication of rate-of-change of the variable shown, and the human brain seems to be able to accept rate-of-change information separately from absolute levels. It is far easier for the human brain to accept a pattern than a digital value.

The best way of looking at the matter is to ask how
many bits of information, or information levels, are
actually required. Is it just "Too much" or "Too
little," or is it "Much too much," "Too much,"
"O.K.," "Too little" or "Much too little?" A level
display which may include absolute values for
recording or closer examination if time permits is
far more valuable than just numbers.

24

OSCILLOSCOPE CAMERAS

Biophysical information displayed on an oscilloscope
or display unit must be photographically recorded
if a permanent record of this information is to be
maintained. While a storage oscilloscope does
maintain a record for an hour or more, it too must
be photographed if a more permanent record is
required.

Information displayed on a CRT may be photographed
using either conventional photographic techniques or
moving-film photographic techniques. Conventional
photography of information displayed on a CRT
involves photographing the CRT display with an
oscilloscope camera. While this technique is
entirely satisfactory for most applications, it does
not provide a continuous record of data versus time,
but only a record of data over a discrete period of
time (normally the duration of one CRT horizontal
sweep). Continuous data may be recorded by
photographing a CRT display with a movie camera;
however, the movie film can only be displayed by
projection and does not provide a convenient record
that can be studied with ease. A modification of
movie photography, known as continuous-motion
photography, provides a permanent record of data
versus time on photographic film by eliminating the
horizontal sweep from an oscilloscope and by
providing the horizontal "sweep" by moving the film
past the face of the CRT as discussed in Section 24.3.

continuous-
motion
photography
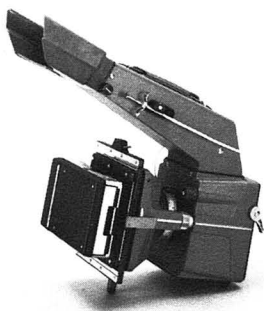
TYPE C-12 CAMERA

PROJECTED GRATICULE ACCESSORY
FOR TYPE C-12 CAMERA
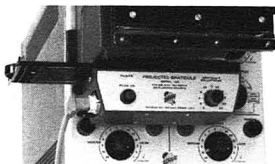
PROJECTED GRATICULE ALLOWS SPECIAL
GRATICULES OR HAND WRITTEN DATA TO
BE SUPERIMPOSED ONTO THE FILM TO-
GETHER WITH DATA FROM THE CRT.

BEAM-SPLITTING MIRROR MINIMIZES
VIEWING PARALLAX WITH CRTS HAVING
EXTERNAL GRATICULES. ACCEPTS A WIDE
RANGE OF LENSES AND FILM BACKS.

TYPE C-27 CAMERA
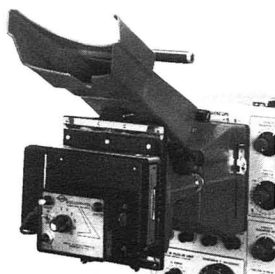
TYPE C-50 CAMERA

DIRECT VIEWING WITHOUT THE USE OF
A MIRROR, THUS TRANSMITTING MAXI-
MUM LIGHT TO THE FILM. ACCEPTS A
WIDE RANGE OF LENSES AND FILM BACKS.
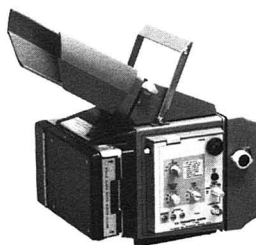
F1.9 LENS 1:0.7 MAGNIFICATION
EASE OF OPERATION AFFORDED BY BUILT
IN TRACE-BRIGHTNESS PHOTOMETER,
RANGE-FINDER FOCUSING AND ACCURATE
EXPOSURE CONTROL.

(A) FOR SIMULTANEOUS PHOTOGRAPHY AND VIEWING - FOR USE WITH MOST
TEKTRONIX OSCILLOSCOPES WITH 5in CRTS

TYPE C-30A CAMERA
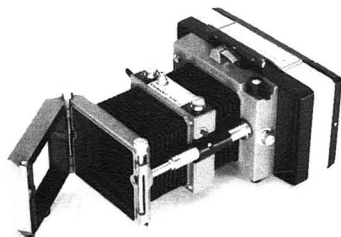
TYPE C-10 CAMERA

FOR USE WITH TEKTRONIX OSCILLOSCOPES
HAVING 3, 4, OR 5in CRTS. F1.9 LENS
MAGNIFICATION VARIABLE IN INDEXED STEPS
FROM 1:0.7 TO 1:1.5. ACCEPTS A WIDE
RANGE OF OPTIONAL FILM BACKS.
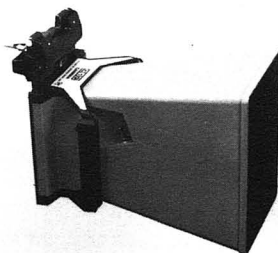
FOR USE WITH TEKTRONIX OSCILLOSCOPES
OR MONITORS HAVING 11in CRTS
F8 LENS - FIXED FOCUS - 1:0.5 MAG-
NIFICATION

(B) FOR PHOTOGRAPHY OR VIEWING WITH CAMERA REMOVED OR "SWUNG AWAY"

Fig. 24-1. Tektronix oscilloscope cameras suited to biophysical measurements.

24.1 CONVENTIONAL CRT PHOTOGRAPHY

Conventional oscilloscope photography involves taking a still photograph of the CRT. Specialized cameras are, however, required to eliminate extraneous light from the photographic record and to allow the CRT display to be photographed without distortion by a lens with a relatively short focal length. Since it is usually undesirable to wait for a film to be processed before being able to analyze data recorded on it, CRT cameras normally utilize Polaroid* film, providing a permanent record within 10 seconds. Polaroid film is produced by the Polaroid Corporation, Massachusetts. It is often desirable for an oscilloscope camera to incorporate a viewing tunnel or some other form of viewing mechanism to allow the CRT display to be viewed during photography.

Type C-12 with beam-splitting mirror

The Tektronix Type C-12 Camera shown in Fig. 24-1 uses a beam-splitting mirror between the lens and the CRT. This mirror permits part of the light from the CRT to pass through the mirror to the lens and part of the light from the CRT to be reflected by the mirror into the viewing tunnel. This permits viewing from an effective viewing position directly in front of the CRT, thus minimizing parallax error between an external graticule and the information displayed on the CRT. The beam-splitting mirror within the Type C-12 Camera also allows light projected from beneath the camera to be reflected into the lens and to pass through the mirror into the viewing system. A projected graticule accessory for the Type C-12 Camera provides a light source beneath the camera and allows various special graticules and masks to be inserted between this

projected graticule

light source and the camera. The projected graticule accessory eliminates any parallax from an external graticule CRT by providing a supplementary graticule in the same plane as the CRT phosphor. Many specialized graticule formats are available from Tektronix for use with the projected graticule; most of these special graticules include a clear area to allow the user to write information on the graticule which will then appear on the final photographic record.

---

*Trademark Polaroid Corporation

392

Type C-27   Although the Tektronix Type C-27 Camera (Fig. 24-1)
is similar to the Type C-12 Camera, it incorporates
direct angular viewing of the CRT rather than a
beam-splitting mirror. This permits the maximum
transmission of light from the CRT to the lens.
As it does not include a beam-splitting mirror, it
does not allow the use of the projected graticule
accessory. The Type C-27 Camera shown includes a
speed computer and an electric shutter; either
manual shutters or electric shutters are available
on many Tektronix oscilloscope cameras.

Type C-50   The Type C-50 Camera (Fig. 24-1) is particularly
easy to operate due to the inclusion of a built-in
trace brightness photometer, a range finder focusing
mechanism and an accurate exposure control system.
The Type C-50 provides angular viewing of the CRT
and does not include a mirror between the CRT and
the lens.

Type C-30A   The Type C-30A Camera (Fig. 24-1) does not include
provision for simultaneous photography and viewing;
it must be either removed from the oscilloscope for
viewing or it may be swung away to one side via its
hinged mounting bezel. The Type C-30A Camera is
particularly versatile, being suited for use with
most Tektronix oscilloscopes and including bellows
between the lens, CRT plane and film plane to allow
variable magnification from 1:0.7 to 1:1.5.

Type C-10   The Type C-10 Camera (Fig. 24-1) is intended for use
with Tektronix oscilloscopes or monitors having
11-inch CRT's, such as the Type 611 Display Unit or
the Type T4002 Graphic Computer Terminal.

film backs   All of the Tektronix oscilloscope cameras shown in
Fig. 24-1, with the exception of the Type C-10, are
shown with film backs for use with Polaroid Type 107
Pack Film, providing "instant" photographic records
on 3 1/4-inch X 4 1/4-inch Polaroid film. All of
these cameras can be used with Graflok* film holders
and accessories to allow the camera to be used with
cut-film holders, film-pack adapters, roll-film (120)
holders, etc. Graflok backs and accessories are
available from local camera shops. The Type C-10
Trace-Recording Camera incorporates a Graflok back
to allow its use with Polaroid Type 57 4 by 5-inch
cut film.

*Trademark Graflex, Inc.

PLAN VIEW OF CURVED FACEPLATE CRT AS USED
IN THE TYPE 410 PHYSIOLOGICAL MONITOR

MARKERS ON CRT PHOSPHOR LINE
UP WITH GRATICULE LINES WHEN
VIEWED FROM INFINITE DISTANCE

LINES ON
GRATICULE IN FRONT
OF CRT HAVE 1cm SPACING

MARKERS ON CRT
PHOSPHOR DO NOT LINE
UP WITH GRATICULE LINES
WHEN VIEWED FROM LENS

OSCILLOSCOPE CAMERA
LENS PLANE

OSCILLOSCOPE CAMERA
FILM PLANE

.82cm
.94cm

CENTER SCREEN
CAMERA MAGNIFICATION
IS 1:0.82

RIGHT EDGE OF SCREEN
CAMERA MAGNIFICATION
IS 1:0.94

$$\text{NONLINEARITY OF MARKER SPACING DUE TO CURVED CRT FACEPLATE} = \frac{.94 - .82}{.82} \approx 15\%$$

Fig. 24-2. Photography of a curved faceplate CRT.

24.2 PHOTOGRAPHY OF CURVED FACEPLATE CRT'S

All Tektronix oscilloscope cameras are intended for
use with CRT's having flat faceplates, the lenses
are thus compensated for good linearity when used
with flat-faceplate CRT's. Some CRT's, such as the
CRT incorporated within the Tektronix Type 410
curved Physiological Monitor, have curved faceplates. A
faceplates curved-faceplate CRT such as the Type 410 CRT will
produce an approximate 15% nonlinearity in the
photographic record when photographed with a
conventional oscilloscope camera as shown in
Fig. 24-2.
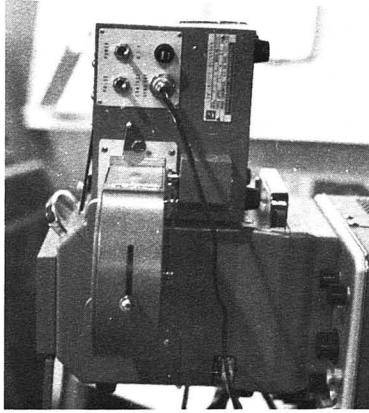
THE NIHON KOHDEN CAMERA MOUNTS DIRECTLY ON MOST TEKTRONIX
OSCILLOSCOPES, THUS ELIMINATING THE NEED FOR CUSTOM ADAPTERS

OPENS FOR SIMULT-
TANEOUS PHOTOGRAPHY
AND CRT VIEWING BY
OPERATOR
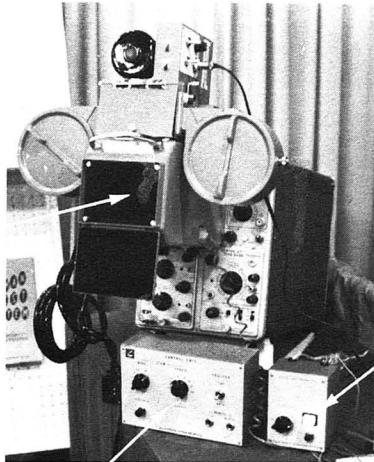
ILLUMINATES A
DATA CARD FOR
RECORDING PERM-
ANENT DATA ON
THE FILM

CONTROL UNIT SELECTS CONTINUOUS MOTION
OR SINGLE FRAME OPERATION AND FILM
SPEED FROM 100cm/s TO .05mm/s

Fig. 24-3. The Lehigh Valley Electronics, Inc., Nihon Kohden PC–2A
continuous-motion camera.

396

HORIZONTAL MOUNTING IN FRONT
OF CONVENTIONAL OSCILLOSCOPE

RACK ASSEMBLY WITH CAMERA CUSTOM MOUNTED

CUSTOM
BELLOWS

CRT

OSCILLOSCOPE CRT MAY BE
VIEWED NORMALLY BY EN-
LARGING THE OPENING IN
THE C-12 CAMERA FRAME
AND COVERING IT WITH AN
AMBER FILTER

BEAM-SPLIT-
TING MIRROR
(INCLUDED WITH
PN 122-0635-00)

PART OF TYPE C-12 CAMERA
(PN 122-0635-00) REMOVE
VIEWING TUNNEL AND REAR
FRAME MOUNTING STUDS.
LENS MOUNTING HOLE MUST
BE BLOCKED IF USED UNDER
HIGH AMBIENT ILLUMINATION.

TEKTRONIX OSCILLOSCOPE/CAMERA ADAPTER
PN 016-0226-01 FOR MOST 5-INCH
ROUND CRT'S
PN 016-0217-01 FOR MOST 5-INCH
RECTANGULAR CRT'S
REFER TO CURRENT TEKTRONIX
CATALOG FOR PN VERIFICATION

Fig. 24-4. The Grass Instrument Company, Kymograph continuous-
motion camera.

Kymograph camera

The Kymograph* Continuous Motion Camera shown in Fig. 24-4 is manufactured by the Grass Instrument Company in the United States. Film transport speeds from 1000 millimeters per second to 0.25 millimeters per second are provided. The Grass Instrument Company does not produce adapters to allow this continuous motion camera to be easily adaptable to most commercial oscilloscopes, thus some form of custom mounting, either in a horizontal or vertical format, must be developed as shown in Fig. 24-4. This custom mounting may be achieved using part of the Tektronix Type C-12 Camera to provide a beam-splitting mirror to allow the CRT to be viewed during continuous motion photography. An amber filter must be incorporated into the viewing path to block out all but amber extraneous light from entering the camera lens via the viewing path. Most recording films are blue sensitive, thus the amber light will not expose the film. If, however, panchromatic film is used, then the amber extraneous light can be rejected with a blue filter in front of the camera lens. The Nihon Kohden camera incorporates a beam-splitting mirror and filters for simultaneous viewing and photography. Both the Nihon Kohden and Grass continuous motion cameras can also be used for conventional, single-picture oscilloscope photography with the oscilloscope time base operating.

If either camera (incorporating a 110 volt motor) is to be used on a higher voltage supply by using a step-down transformer, the transformer should be of adequate rating (500 volt-amp or more) to deliver the *starting surge* current of the motor.

---

*Trademark Grass Instrument Company

25

GRAPHIC RECORDERS

A continuous visual record of physiological data versus time may either be obtained with a graphic recorder or with an oscilloscope and a continuous motion camera as covered in the previous chapter. Continuous motion CRT photography, while offering almost unlimited bandwidth and extremely fast recording rates, does not provide an immediate record of the recorded data. A graphic recorder, while inherently possessing limited high-frequency response characteristics, does provide an immediate record of the recorded data for observation. Graphic recorders are often referred to as oscillographs, oscillographic recorders, strip chart recorders or chart recorders. Graphic recorders utilize some form of stylus to traverse a strip of chart paper while the chart paper is in motion, leaving a visible tracing which is a record of the time variations of the input voltage or current.

basic
components

Most graphic recorders consist of the following five basic components:

    -- An electromechanical device to convert an electrical input signal to mechanical movement,

    -- A stylus arm to transmit the mechanical movement from the electromechanical device to the stylus,

    -- A stylus to leave a written record on chart paper as the stylus moves across the chart paper,

    -- A chart paper assembly consisting of a chart paper supply roll, a chart paper writing table and a chart paper takeup roll and

    -- A paper drive mechanism to move the chart paper across the writing table from the supply roll to the takeup roll at a constant speed.
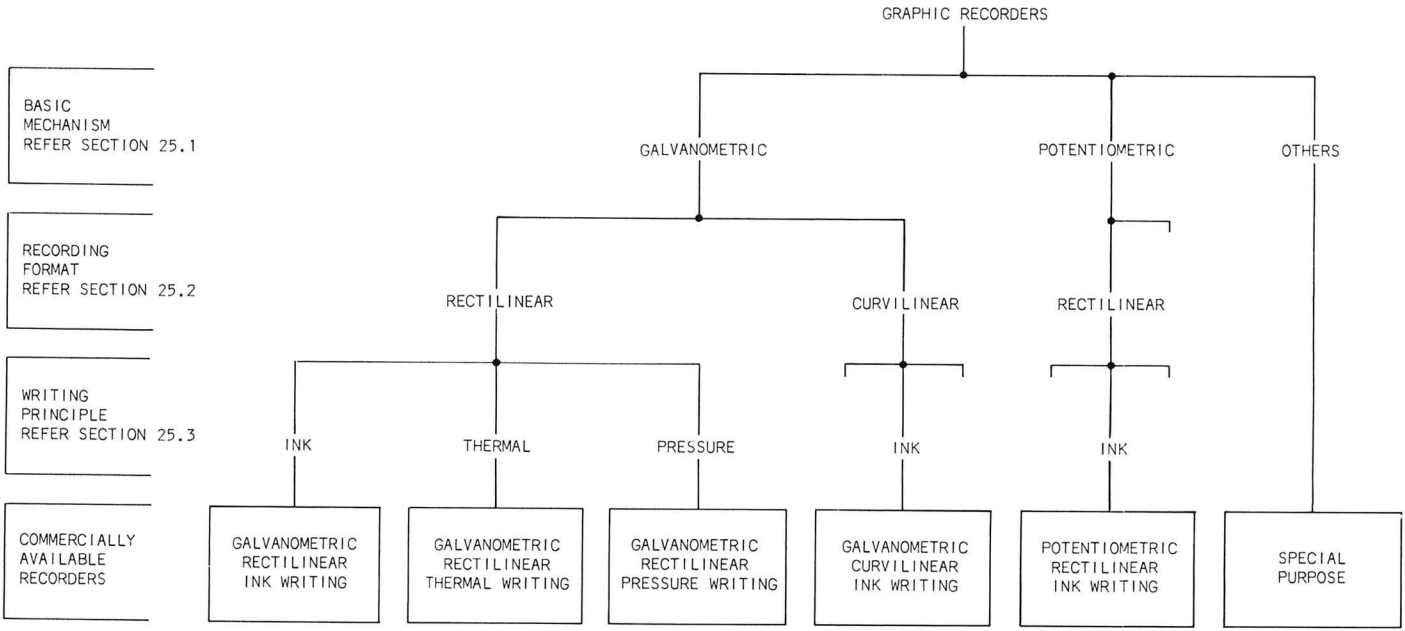
Fig. 25-1. Graphic recorder classifications.

characteristics

Graphic recorders may be broadly classified as to the following three characteristics:

-- Basic Mechanism

The basic mechanism used to convert an input current or voltage into mechanical movement of the recording stylus may use either a galvanometric or a potentiometric principle as discussed in Section 25.1.

-- Recording Format

The geometrical relationship between stylus movement and the chart paper will either result in a curved line or a straight line being transcribed on the chart paper when the stylus is abruptly moved to a new position (referred to as curvilinear or rectilinear recording respectively as discussed in Section 25.2).

-- Writing Principle

A written record is transcribed on the chart paper with either an ink-pen stylus, a heated stylus or a rounded-point stylus (referred to as ink writing, thermal writing and pressure writing respectively as discussed in Section 25.3).

402

INK WRITING

INK PEN
STYLUS

CONVENTIONAL
CHART PAPER WITH
CURVILINEAR GRID

CHART PAPER PULLED
PAST STYLUS BY
DRIVE MECHANISM

CHART PAPER
SUPPLY ROLL
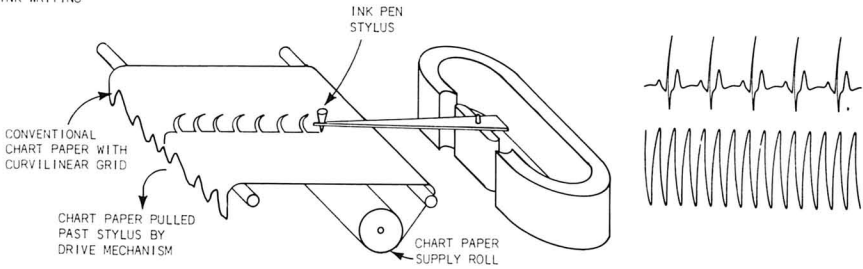
(A) CURVILINEAR RECORDER

THERMAL WRITING

STYLUS ARM

HEATED
STYLUS

HEAT SENSITIVE
CHART PAPER WITH
RECTILINEAR GRID

PAPER PASSES
OVER KNIFE EDGE

INK WRITING

MECHANICAL LINKAGE
TRANSFORMS PIVOTAL
MOVEMENT OF THE COIL
INTO LATERAL MOTION
OF THE PEN TO PRODUCE
RECTILINEAR RECORDING

INK WRITING PSEUDORECTILINEAR

LONG PEN IN CONJUNCTION
WITH LIMITED ARC PRODUCES
CURVILINEAR RECORDINGS
THAT ARE DIFFICULT TO
DIFFERENTIATE FROM
RECTILINEAR RECORDINGS
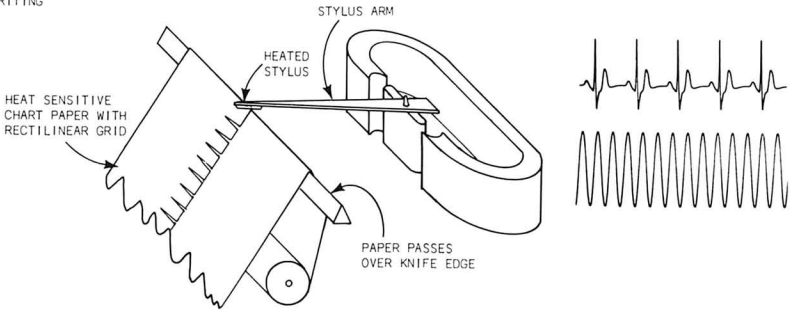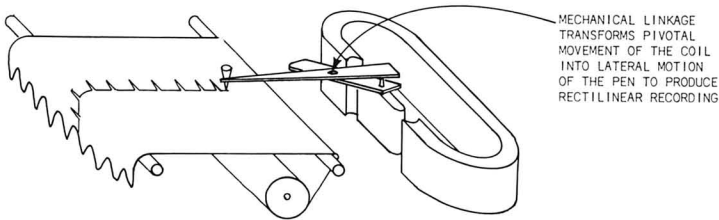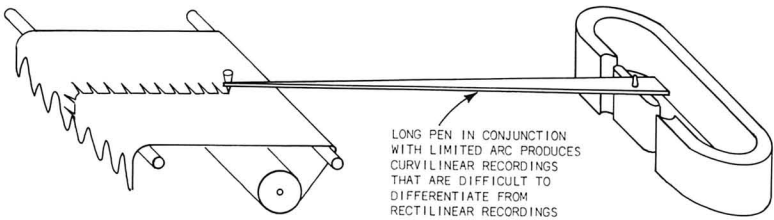
(B) RECTILINEAR RECORDERS

Fig. 25-2. Curvilinear and rectilinear galvanometric recorder
mechanisms.

25.1 BASIC RECORDER MECHANISMS

galvano-
metric
mechanisms

An input current or voltage is converted to
mechanical movement of the stylus by either a
galvanometric or a potentiometric principle.
Galvanometric recorders utilize a moving coil and
magnet assembly with a stylus arm attached to the
moving coil. This assembly is somewhat similar to
the common d'Arsonval galvanometer assembly used in
most conventional panel meters; however, the
assembly used in a graphic recorder must develop
considerably more torque than the assembly used in
a panel meter to overcome forces associated with
stylus pressure and to provide good transient
performance (damping). The various recorder types
shown in Fig. 25-2 all use galvanometric recorder
mechanisms. These mechanisms are normally
constructed so that the angular deflection of the
stylus arm is proportional to the magnitude of
input current, that is, the relationship between
input current and angular deflection is linear.

damping

Damping is particularly important in a galvanometric
recorder. If the moving-coil mechanism is at rest
and a current is suddenly applied to the coil, the
coil, the stylus arm and stylus will commence to
move to a new position. The stylus and arm will
gain momentum during this movement and when the new
equilibrium position is reached this momentum will
cause them to overshoot this position. The process
is then reversed with the stylus and arm oscillating
about the equilibrium position for some time until
finally coming to rest. Damping must be added to a
galvanometric recorder mechanism to overcome these
oscillations, the mechanism being referred to as
critically damped when the stylus and arm assume an
equilibrium position as quickly as possible without
overshooting the position. It is most important
that pen-to-paper friction constitute only a very
small part of the pen system damping, which should
be supplied electrically, otherwise the damping will
be unreliable over a period of time, and will tend
to vary from one side of the chart to the other.

mechanism
frequency
response

Due to finite force developed by a galvanometric recorder mechanism and the finite mass associated with the stylus and arm, the maximum angular acceleration of the system is defined and thus the high frequency response of the system is defined. Most galvanometric recorder mechanisms offer a maximum high frequency response of less than 200 Hz, the high frequency response being directly proportional to the force developed by the galvanometric mechanism and inversely proportional to the moment of inertia of the coil, stylus arm and stylus assembly.

potentio-
metric
mechanism

Potentiometric recorders operate on a servo principle with the position of the stylus arm being detected via a contact mechanism attached to the arm and in contact with a fixed slide-wire as shown in Fig. 25-3. The servo system will move the stylus arm and stylus until the potential between the slide-wire contact and the input voltage is zero. The linearity and accuracy of a potentiometric recorder mechanism is dependent only on the characteristics of the slide wire and associated electronics, that is, the relationship between input voltage and the position of the contact on the slide wire is linear. Potentiometric recorder mechanisms in general provide higher accuracy and better linearity than galvanometric mechanisms.

Damping in a potentiometric recorder mechanism is normally achieved electronically within the servo amplifier. Due to the inherent speed limitations of the servo system and the finite mass and frictional forces associated with the stylus arm, stylus and slide wire, potentiometric recorder mechanisms are inherently low frequency devices and rarely offer a high frequency response in excess of 20 Hz unless stylus deflection is limited to a centimeter or so.

galvano-
metric
versus
potentio-
metric

Since high accuracy and linearity is rarely of paramount importance in biophysical measurements and since a high frequency response of 20 Hz is not adequate for most biophysical recording applications, galvanometric recorders are generally preferred to potentiometric recorders for biophysical applications. Galvanometric recorders are also, in general, less expensive than potentiometric recorders, particularly if multi-channel instruments are considered.
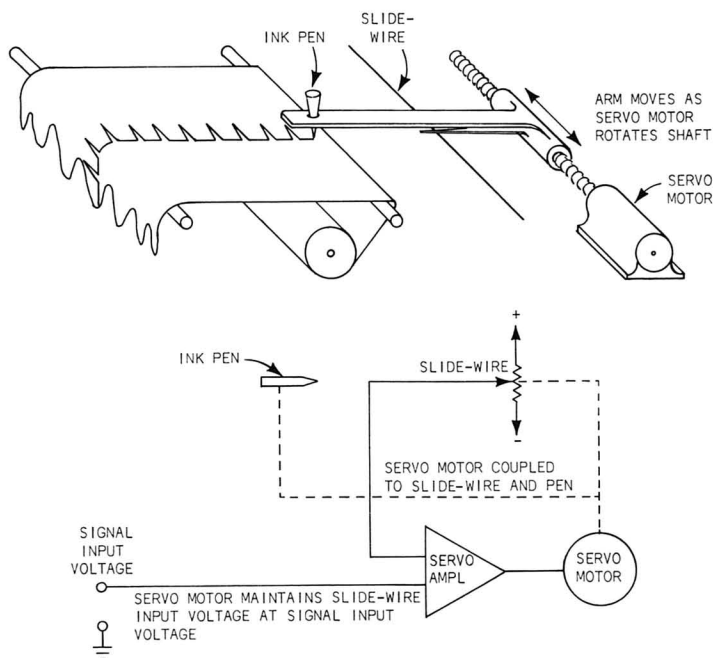
Fig. 25-3. Rectilinear, potentiometric (servo) recorder mechanisms.

25.2 RECORDING FORMATS

curvilinear

The geometrical relationship between stylus movement and the chart paper is either referred to as curvilinear or rectilinear. A pivoted stylus arm will transcribe an arc at its tip if the arm is caused to rotate about its pivot point. A stylus at the tip of the arm will, therefore, transcribe a curved line on stationary chart paper located beneath the stylus. Since the line is curved, the recording format is referred to as curvilinear recording. The basic galvanometric mechanism shown in Fig. 25-2A produces a curvilinear recording. A typical ECG and sinewave recorded with a curvilinear recorder is shown in Fig. 25-2A. Although the chart paper used with curvilinear recorders is marked with a curvilinear grid, the distortion produced by curvilinear recording is, at least, an inconvenience and thus rectilinear recorders are preferred. Curvilinear recorders are, however, considerably less expensive than rectilinear recorders and are extensively used, particularly for less critical applications such as may be encountered in a student laboratory.

Movement of a stylus in a straight line perpendicular to the direction of movement of the chart paper would transcribe a straight line on stationary chart paper located beneath the stylus. Since the line is straight, the recording format is referred to as rectilinear recording. The potentiometric mechanism shown in Fig. 25-3 produces a rectilinear recording. Rectilinear recording may also be achieved with the galvanometric recorder mechanisms shown in Fig. 25-2B.

The thermal writing rectilinear recorder shown in Fig. 25-2B transcribes a curvilinear arc at the end of the stylus arm; however, the contact between the arm and the chart paper effectively moves up and down the stylus arm as the arm rotates about its pivot point, a straight line is, therefore, transcribed onto stationary chart paper. In analyzing the geometry of this technique it is apparent that the movement of the stylus on the chart paper is not directly proportional to the angular deflection of the stylus arm, thus producing a linearity error in the recorder. This linearity error is less than 1% if the total stylus deflection in both directions from its central resting point is limited to half the perpendicular distance between the "knife edge" and the pivot point.

Various recorder manufacturers have designed ingenious mechanical linkages between the stylus arm and chart paper to produce true rectilinear recording from a galvanometric mechanism as shown in the ink writing recorder in Fig. 25-2B. If stylus deflections are small when compared with the length of the stylus arm then the recording obtained, while being a true curvilinear recording, is difficult to differentiate from a rectilinear recording. This technique is extensively used in EEG recorders and is shown diagramatically in the ink-writing, pseudo-rectilinear recorder in Fig. 25-2B.

25.3 WRITING PRINCIPLES

The stylus in a graphic recorder must cause a written record to appear on the chart paper as the stylus moves across the surface of the paper. This written record may be achieved with a pen attached to the stylus arm, a heated stylus attached to the stylus arm, or with a rounded point attached to the stylus arm. The writing techniques are referred to as ink writing, thermal writing and pressure writing, respectively.

ink writing

Ink-writing graphic recorders use ink pens operating on a capillary and siphon principle to record on untreated chart paper. The ink pens need constant attention to prevent blockage; however, the recording paper used is conventional paper, being considerably less expensive than the treated papers used for either thermal writing or pressure writing recorders. Thus, the ink-writing recorder is preferred in applications where a large volume of recording paper is required such as in electroencelphalography.

thermal writing

The thermal writing recorder uses a heated stylus in conjunction with specially coated chart paper. The stylus is heated by electric current flowing through a resistive element at the end of the stylus arm, as the heated stylus travels across the chart paper it melts off a white coating from the chart paper to expose a dark undersurface. While the thermal writing recorder requires little maintenance, the specially treated paper is somewhat expensive making this technique unsuited to applications requiring long term recording.

pressure writing

Pressure writing recorders also utilize special chart paper. Carbon treated paper is transcribed by a rounded stylus at the end of the stylus arm with the pressure of the stylus leaving a black trace on the paper. Pressure writing recorders in general exhibit poor high speed characteristics due to the finite pressure required and, thus, the finite frictional forces involved at the stylus-paper interface.

TECHNI-RITE ELECTRONICS INC.
MODEL TR711

3dB FREQUENCY RESPONSE FROM DC TO 35Hz
FOR 4cm P-P DEFLECTIONS AND FROM DC TO
125Hz FOR 0.8cm P-P DEFLECTIONS.

RECTILINEAR THERMAL WRITING

1mm/s TO 50mm/s

AC POWERED

WEIGHT - 10 POUNDS

A SINGLE-CHANNEL, PORTABLE, HIGH-SPEED RECORDER
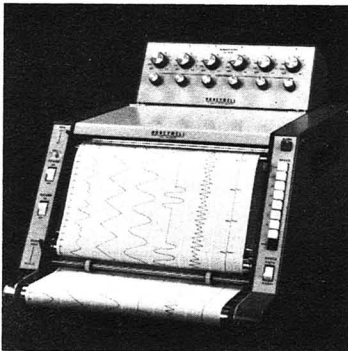
HONEYWELL MODEL 2506 SHOWN WITH
MODEL 125 ATTENUATOR

FLAT FREQUENCY RESPONSE FROM DC TO 30Hz
FOR 4cm P-P DEFLECTIONS AND FROM DC TO
80Hz FOR 1cm P-P.

RECTILINEAR INK WRITING

1mm/min TO 250mm/s

AC POWERED

WEIGHT - 65 POUNDS

A SIX-CHANNEL HIGH-SPEED RECORDER
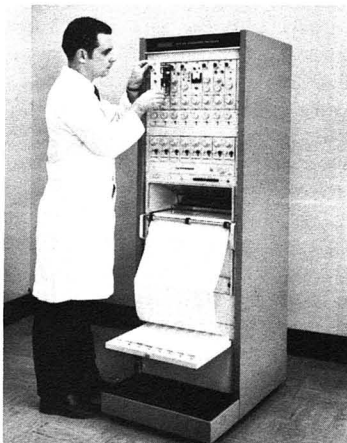
BECKMAN INSTRUMENTS INC.
TYPE S11 DYNOGRAPH *

FLAT FREQUENCY RESPONSE FROM DC TO 42Hz
FOR 4cm P-P DEFLECTION AND FLAT WITHIN
±30% FOR 0.8cm P-P DEFLECTION.

RECTILINEAR PRESSURIZED INK
WRITING

0.1mm/min TO 250mm/s

AC POWERED

RACK MOUNTED

*TRADEMARK BECKMAN INSTRUMENTS, INC.

AN EIGHT-CHANNEL HIGH-SPEED RECORDER

Fig. 25-4. Typical commercial recorders.

25.4 COMMERCIAL GALVANOMETRIC RECORDERS

Three commercial high speed recorders are shown in Fig. 25-4. All of these recorders are suited to biophysical measurements, providing chart speeds consistent with biophysical signals containing frequencies to 100 Hz or so. The single channel recorder shown in Fig. 25-4 is ideally suited for use in conjunction with the Tektronix Type 410 Physiological Monitor. The Type 410 provides a high level signal output via a rear panel connector which can be directly connected to the single channel recorder. Adequate output is provided by the Type 410 for use with almost any recorder and is directly compatible for use with any of the recorders shown in Fig. 25-4 as all of these recorders include input signal attenuation.

Type 410
recorder
output

25.5 SPECIAL PURPOSE RECORDERS

The preceding discussion has covered the recorder types in common use for biophysical measurements; however, many different types of graphic recorders, such as optical recorders and x-y recorders, are in common use and are occasionally used for biophysical measurements.

low cost
recorder

A low cost, single-channel, low speed recorder is shown in Fig. 25-5. This recorder is limited to low speed applications providing a maximum chart paper speed of 0.5 inches per second. This recorder uses a galvanometric movement that is almost identical to the movement used in most panel meters. The stylus arm is completely free to move and, thus, no frictional forces are involved. Once every second a mechanical lever forces this stylus arm and stylus against pressure-sensitive chart paper, leaving a small impression. Since the chart speed is limited, the written information on the chart paper appears as a series of closely spaced dots. This form of recorder is particularly suited for recording very low frequency physiological signals such as body temperature, heart rate variation, etc. Very slow chart speeds are available to allow 24 hours of information to be recorded on less than 2 inches of chart paper.

RUSTRAK INSTRUMENT CO.
MODEL 88

FOR DC OR LOW FREQUENCY USE ONLY
RESPONSE TIME IS ≈0.8s

RECTILINEAR PRESSURE-SENSITIVE
PAPER WRITING SYSTEM
0.06in/h TO 0.5in/s

AC OR LOW VOLTAGE DC POWERED

WEIGHT - 5 POUNDS

Fig. 25-5. A single-channel low speed, low cost recorder.
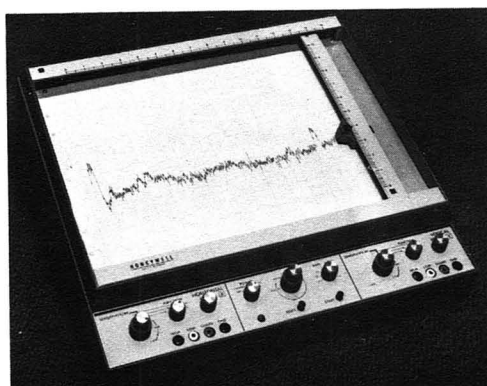
Fig. 25-6. An *x* - *y* recorder.

optical
recorder

Many graphic recorders use optical coupling between
the basic recorder mechanism and the chart paper,
the "chart paper" consisting of photographic film.
High frequency performance to several kHz is
obtainable with these optical recorders; however, in
most instances, if high frequency performance in
excess of that obtainable with conventional graphic
recorders is desired, an oscilloscope and continuous
motion camera is preferable to an optical recorder
for they are far less susceptible to mechanical
shock and vibration and, in general, far more
versatile.

x-y recorder

An x-y recorder is a device used to plot two
variables against one another on chart paper rather
than, as with a conventional graphic recorder,
plotting one variable against time on chart paper.
A single sheet of graph is commonly used as the chart
paper, the stylus moving in both x and y directions
across the paper. A typical x-y recorder is shown
in Fig. 25-6. The transient response performance
of x-y recorders limits them to low frequency
applications. Except where extreme accuracy is
required, an oscilloscope and conventional
oscilloscope camera is normally preferred to the
x-y recorder as the combination offers superior
transient performance and greater versatility.

26

MAGNETIC TAPE RECORDERS

A magnetic tape recorder is an analog storage device. *Electrical* signals fed into a magnetic tape recorder are stored on the magnetic tape with the same *electrical* signals available from the magnetic tape recorder when required. Although the oscilloscope camera and the graphic recorder are information storage devices, once information has been recorded by these devices it is extremely difficult to regenerate the information in its original electrical format. The magnetic tape recorder is the only convenient device that permits recording of data in such a manner as to allow it to be reproduced at some later time in its original
"holding"
device

soft/hard
copy
electrical format, thus, making it a "holding" device as well as a recording device. A magnetic tape recording may be referred to as a "soft copy" as it is capable of being reconverted to the original electrical variable. The oscilloscope camera and graphic recorder produce "hard copy" which is not capable of convenient reconversion.

The magnetic tape recorder allows biophysical signals to be recorded in real time during a physiological experiment and then allows the experimental results to be reproduced at some later time. This allows those performing the experiment to concentrate on obtaining data during the experiment and then to concentrate on analyzing the data after the experiment has been completed. Some monitoring of the recorded signal must be used, however, as it is frequently too late to correct some obvious fault in the recording system unless signals are monitored while they are being stored on magnetic tape.
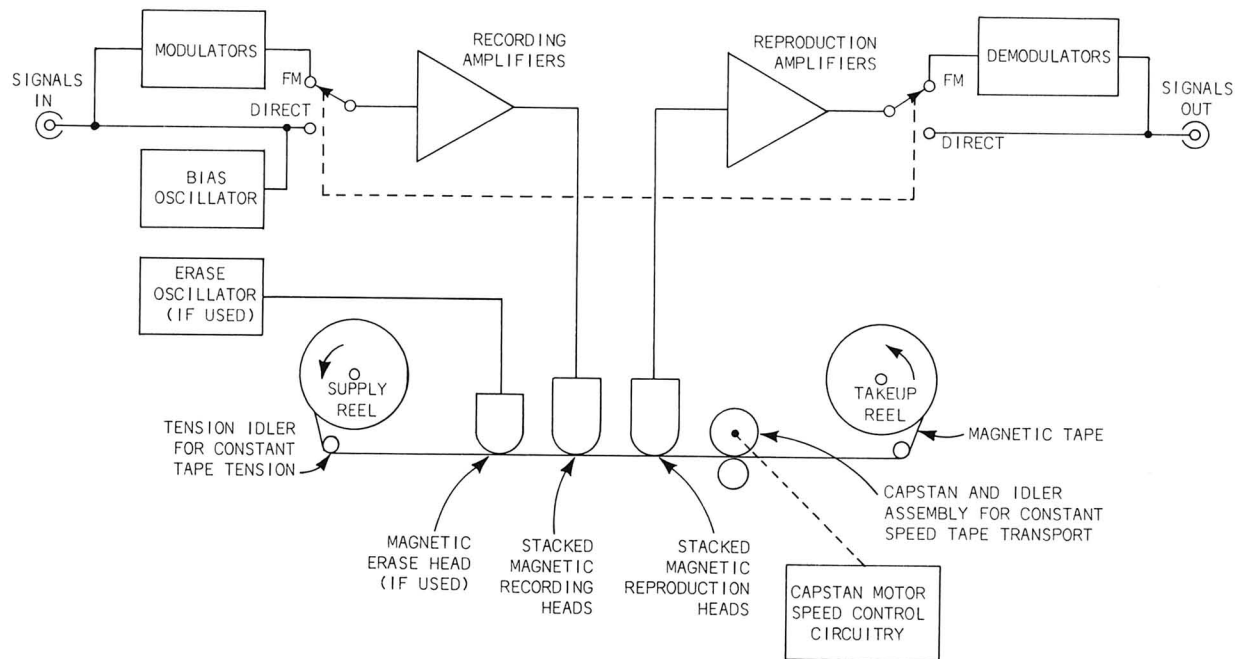
Fig. 26-1. Basic components of an instrumentation tape recorder.

Since most magnetic tape recorders have the capability of recording information at one speed and then reproducing it at a different speed, time expansion or compression of the information can be

record slow, playback fast

accomplished. By recording at a slow speed and reproducing at a faster speed, many hours of experimental data can be reproduced in just a few minutes, thus taking full advantage of the capabilities of other instruments in the measuring system such as the high speed capability of data processing equipment or the high frequency

record fast, playback slow

capability of a spectrum analyzer. By recording information at a high speed and reproducing it at a slower speed, high frequency information is reproduced as low frequency information, thus allowing more time for detailed analysis of the information and allowing the subsequent use of low frequency instrumentation such as a graphic recorder.

Two basic recording techniques are used in instrumentation magnetic tape recorders suitable for biophysical applications -- direct recording and indirect frequency-modulation (FM) recording. These two techniques are covered separately in Sections 26.1 and 26.2.

26.1 DIRECT MAGNETIC TAPE RECORDING

The following discussion assumes that the reader is familiar with magnetic tape recorder principles as they relate to the unsophisticated audio tape recorder commonly used for home entertainment. These recorders all use the direct recording technique.

Fig. 26-1 shows the basic components of an instrumentation tape recorder and shows switching in both the recording circuitry and the reproduction circuitry for switching between DIRECT or FM. The tape recorder is operating in the direct recording mode with the switch in DIRECT. Referring to Fig. 26-1, in the direct mode the input signal is algebraically added to a high frequency bias signal of about 100 kHz and then fed directly into a

recording-circuit description

recording amplifier which provides the recording current necessary to drive the magnetic recording head. The recording head converts the current into a varying magnetic flux which changes the residual magnetism on the magnetic tape as it moves past the recording head. This magnetism on the tape, in turn,

416

produces a change in the output voltage from the
magnetic reproduction head. The voltage produced
at the reproduction head is then amplified in a
reproduction amplifier to provide a high level
output signal.

Direct magnetic tape recording has two severe
disadvantages -- poor low frequency response and
poor output level stability. The poor low frequency
response is due to the magnetic characteristics of
the recording and reproduction heads as the
efficiency of the heads decreases as frequency drops
below a few hundred hertz. For biophysical
applications it can be stated that direct magnetic
tape recording is unsatisfactory for signals below
100 Hz.

"direct"
mode
limitations

If a constant-level, audio-frequency signal is
recorded on a home entertainment audio tape recorder,
the level of the output signal will fluctuate by
30% or more primarily due to variations in the
magnetic coating on the tape and, also, to variations
encountered in the contact between the tape and the
recording and reproduction heads. This 30%
variation, while being quite acceptable for home
entertainment recorders, is unsatisfactory for
biophysical measurements. By the use of high quality
magnetic tape and careful design of the tape
transport system, this amplitude stability can be
held within 10% or so. This 10% output level
variation exists on most instrumentation tape
recorders operating in the direct mode and is a
severe limitation in many applications. This
limitation is completely overcome by recording via
the indirect FM technique and this technique also
extends low frequency response to DC.

26.2 INDIRECT FM MAGNETIC TAPE RECORDING

Referring to Fig. 26-1 with the recorder operating in the FM mode as shown, the input signal is applied to a modulator where it is used to frequency modulate a carrier frequency, this modulated carrier is then fed to the recording amplifier which provides the recording current necessary to drive the magnetic recording head. The subsequent voltage from the reproduction head is then amplified and demodulated to provide an output signal. Indirect FM tape recording permits frequencies down to and including DC to be recorded.

"FM" mode
limitations

The high frequency response of any magnetic tape system is limited by the characteristics of the recording and reproduction heads, by the signal-to-noise ratio that is acceptable, and to some extent by the characteristics of the magnetic recording tape used. Since the high frequency response of the magnetic tape system is limited and thereby limiting the FM carrier frequency, the high frequency response of the indirect FM system is further limited due to the maximum deviation limits of the FM modulator and demodulator (normally about 20% of the FM carrier frequency).

418

DIRECT OR FM RECORD/REPRODUCE SYSTEMS

SIX TAPE TRANSPORT SPEEDS
    FROM 15/16in/s TO 30in/s

USES EITHER 1/4in WIDE OR 1/2in WIDE
    MAGNETIC TAPE

SEVEN DATA CHANNELS AND AN ADDITIONAL
VOICE CHANNEL FOR VOICE ANNOTATIONS

DIRECT RECORD/REPRODUCE SYSTEM:
    100Hz TO 150kHz AT 30in/s
    50Hz TO 5kHz AT 15/16in/s

FM RECORD/REPRODUCE SYSTEM:
    DC TO 10kHz AT 30in/s
    DC TO 312Hz AT 15/16in/s

PORTABLE CASE OR RACK MOUNTED
    WEIGHT - 80 POUNDS

Fig. 26-2. The Model PR-500 seven-channel instrumentation
recorder (Ampex Corporation).

sidebands
limit
bandwidth

Referring to the instrumentation recorder shown in
Fig. 26-2, the specifications for the high
frequency response of the direct record/reproduce
system is only 5 kHz at a tape speed of 15/16 inches
per second. The maximum frequency component in the
modulated FM carrier frequency must, therefore, be
below 5 kHz. Since 8 sidebands are required on each
side of an FM carrier frequency for almost distortion
free FM reproduction, the high frequency response of
the recorder in the FM record/reproduce mode is,
thus, limited to *one-sixteenth of the high frequency
response* of the direct record/reproduce system or
only 312 Hz at a tape speed of 15/16 inches per
second. As the tape speed of the system is
increased, the high frequency performance of the
recording and reproduction heads is increased, the
FM modulation frequency can be increased, and,
therefore, the high frequency performance in the FM
record/reproduce mode is proportionally increased.

In the indirect FM recording mode, amplitude
stability between the signal applied to the recording
head and the signal received from the reproduction
head need not be considered because all instability
will either be removed in a limiting stage in the
reproduction amplifier or will be of no consequence
due to the insensitivity of the demodulator to
amplitude modulated signals.

26.3 TAPE TRANSPORT MECHANISM

The basic components of the tape transport mechanism used in an instrumentation tape recorder is shown in Fig. 26-1. The tape supply reel may supply either 1/4- or 1/2-inch wide tape from a 10 1/2-inch diameter reel which supplies 3600 feet of normal thickness tape. A rotating capstan and idler assembly pull this tape past the magnetic heads at a constant speed; the tape is then collected on a takeup reel. Tension idler assemblies on both sides of the capstan are used to provide constant tension on the tape.

tape speeds

The speed at which the magnetic tape is drawn past the magnetic heads is controlled by the speed of the capstan motor which drives the capstan and idler assembly. Most instrumentation tape recorders offer a speed range from 15/16 in/s to 30 in/s with four intermediate speeds. Some instrumentation recorders operate up to speeds of 240 in/s.

tape speed stability

Tape speed stability is an important consideration in instrumentation recorders. Relative tape speed instability is referred to as "wow" if occurring at less than about 2 Hz and "flutter" if occurring at greater than about 2 Hz. With the recorder operating in the direct mode, any instability in the tape speed will appear as a change in frequency of the output signal. With the recorder operating in the FM mode, any instability in the tape speed will appear as an amplitude variation in the output signal. Control circuitry within instrumentation tape recorders maintains a constant tape speed by using some form of servo mechanism; the feedback to the servo is derived via a tachometer attached to the capstan or via the output from a constant frequency source that has been recorded onto one channel of a multi-channel recorder.

26.4 OTHER MAGNETIC TAPE RECORDER CONSIDERATIONS

erasure

If a magnetic tape has been used previously to record information, this information must be erased from the tape before the tape can be reused for recording. Erasure can be achieved with a bulk tape eraser that exposes a complete reel of tape to an intense low frequency magnetic field. Erasure may also be accomplished via a magnetic erase head on the tape recorder; this head exposes the tape to an intense high frequency magnetic field before the tape reaches the magnetic recording head. Many instrumentation tape recorders, particularly multi-channel recorders, do not incorporate erasure as it is generally more convenient and less expensive to use a bulk eraser.

track width

The previous discussion assumes only one channel of information being recorded on the tape recorder. Most commercial instrumentation tape recorders offer more than one channel of information; the recorder shown in Fig. 26-2 offers seven channels for information plus an additional channel for voice annotations. Obviously, if seven tracks plus a voice track are to be stacked onto a 1/4-inch wide tape, the width of each track must be extremely narrow. The instrumentation recorder shown in Fig. 26-2 uses track widths of 0.024 inch.

The tape transport used in an instrumentation tape recorder must be capable of quickly winding tape in the forward direction to allow a desired segment of tape to be reached quickly. Similarly, the transport must be capable of quickly rewinding the tape back onto the supply reel. The braking system incorporated into the tape transport mechanism must be capable of quickly stopping tape movement when a stop command is received.

noise   The maximum achievable signal to noise ratio in the
best quality tape recorders, in either the direct
or FM modes, is about 44 dB, it is thus necessary to
record at as high a signal level as the system will
permit and to use some filtering during playback to
reduce the noise as much as possible. The signal
to noise ratio in the FM mode may be expected to
approach 44 dB, however in the direct mode the signal
to noise ratio will rarely exceed 30 dB.
Instrumentation grade magnetic tape should be used
for physiological recording, in contrast to lower
quality entertainment grade tape. The use of
instrumentation grade tape will insure that the
maximum signal to noise ratio of the system is
obtained, it will insure that the maximum amplitude
stability in the direct mode is obtained and it will
insure that there is no "information dropout" due to
"dropouts" on the tape. Careful maintenance is
particularly important with magnetic tape recorders
as the signal to noise ratio of the recorder will be
drastically reduced should the record and/or playback
heads become worn, dirty and/or magnetized.
Manufacturers maintenance instructions usually
include head cleaning and demagnetizing procedures.

27

DATA TRANSMISSION AND PROCESSING

Transmission of data between various elements in a biophysical measuring system is normally accomplished directly with shielded cable. Shielded cable is, however, unsuited for many data transmission requirements, particularly for transmission over long distances or if it is desired for one item of instrumentation to be completely isolated both physically and electrically from other items in the system. Data transmission via a radio link, referred to as telemetry, is used in these applications. Data transmission becomes particularly important if the physiological signal is to undergo data processing in a computer as, in most instances, the computer is located remotely from the data source.

27.1 DATA TRANSMISSION VIA SHIELDED CABLE

Data transmission via shielded cable is the simplest form of data transmission; the signal is transmitted in its original analog voltage form. For faithful transmission, the transmission medium must be capable of passing the physiological signal over its maximum conceivable amplitude and bandwidth range. In a direct—wire unterminated transmission link

cable losses   using shielded cable, losses in signal amplitude are caused by cable reactance and are in direct proportion to cable length. For most physiological data transmission applications, cable reactance imposes a practical maximum cable length of about 1,000 feet when the cable is driven by normal instrumentation providing signals from an output impedance of perhaps 100 ohms.

Within the bandwidth of interest in biophysical
measurements (DC to 30,000 Hz) the capacitance from
the shielded cable inner conductor to the shield is
the most important single factor limiting the high
frequency performance of the transmission system.
The bandwidth of a shielded cable transmission
system, when driven from a voltage source, is
given by:

$$f_{3dB} = \frac{.16 \times 10^{12}}{RCl}$$

R is the output resistance of the driving source in
ohms, C is the nominal capacitance of the shielded
cable in picofarads per foot and l is the length of
the cable in feet. Instrumentation providing an
output signal via a vacuum-tube cathode-follower
circuit provides a typical driving source impedance
of ≤1,000 ohms; instrumentation providing an output
signal via a transistor emitter-follower circuit
provides a typical driving source impedance of

determining
cable system
bandwidth

≤100 ohms. Most shielded cable has a capacitance
of between 20 and 40 pF/ft; the common 1/4-inch
diameter shielded cable used in many physiological
laboratories exhibits a typical capacitance of 30
pF/ft. Relating this information to the previous
formula, a 1,000-foot length of 30 pF/ft shielded
cable, when driven from a source impedance of 100
ohms, provides a transmission system bandwidth of
50 kHz. Practically, the system bandwidth will be
somewhat less than this as the driving amplifier
does not "see" only the capacitance of the line,
but its inductance and resistance as well. At
50 kHz, a 1,000-foot line, allowing for propagation
velocity in the cable, is a significant fraction of
a wavelength long.

Although unterminated direct wire transmission
links using shielded cable can be used to about
1,000 feet it is recommended that terminated
systems be used for analog data transmission in
excess of 50 feet. Terminated systems require the

shielded cable to be driven from a source of its own characteristic impedance and be terminated in its characteristic impedance. To avoid ground loops, it is customary, and usually essential, to transmit from a single-ended driver having the line characteristic impedance, and to receive by a differential amplifier between line and shield having a high common mode impedance, and a differential input impedance equal to the characteristic impedance of the line.

The overall bandwidth of a biophysical measurement system is given by:

$$f = \frac{1}{\sqrt{\frac{1}{f1^2} + \frac{1}{f2^2} + \frac{1}{f3^2} + \ldots\text{etc.}}}$$

where $f1$, $f2$, $f3$, etc. are the bandwidths (upper -3 dB frequency limits) of the individual components in the system including the transmission components between individual instruments.

AM techniques Wire transmission can be used over very long distances by employing an amplitude modulated carrier frequency with the information to be transmitted being used to amplitude modulate a carrier frequency. This carrier frequency must be higher than the highest frequency component in the modulating signal by a factor of at least five to allow low-distortion demodulation at the receiver. Any amplitude modulated transmission system is only as good as the ability of the system to faithfully reproduce amplitude variations. System attenuation, system noise, system nonlinearity and/or interference from other sources produce errors in the information transmitted. For this reason, amplitude modulated data transmission techniques are limited to wire transmission rather than to radio link transmission.

27.2 DATA TRANSMISSION VIA A TELEMETRY LINK

Telemetry links used in biophysical measurements
range from the short range systems used in behavioral
studies laboratories to the extremely long range
systems used in the aerospace industry. The
principal difference between the short range and
long range systems is the power output capability of
the transmitter and the sensitivity of the receiver.

short
range
applications

Short range telemetry systems are used in behavioral
studies to completely isolate a subject or laboratory
animal from a recording system. A small telemetry
transmitter may be attached to the subject to
transmit, for example, the subject's ECG; a telemetry
receiver may be located only a few feet away to allow
this ECG signal to be processed by a physiological
measurement system. This allows complete subject
freedom of movement and thus allows for more natural
subject behavior. As a telemetry transmitter is
normally a completely isolated battery operated
device, telemetry offers a degree of safety
unattainable with conventional instrumentation due
to the complete elimination of direct electrical
connections between the subject and the
instrumentation.

single-
channel
telemetry

Short range telemetry systems usually use FM
techniques and may transmit in the FM broadcast
band (88 to 108 MHz). They may be simple systems
providing only one data channel or more complex
systems providing multichannel capability by using
subcarrier modulation. A typical single-channel
telemetry link is shown diagramatically in Fig. 27-1.
The single-channel battery-powered FM transmitter
shown would normally be one cubic inch or so in
volume providing several hundred hours of operation
from one or two miniature mercury batteries. The
receiver must provide DC coupling from the
discriminator to give the system a DC signal
transmission capability, and, to avoid DC drift,
both transmitter and receiver should be crystal
controlled.
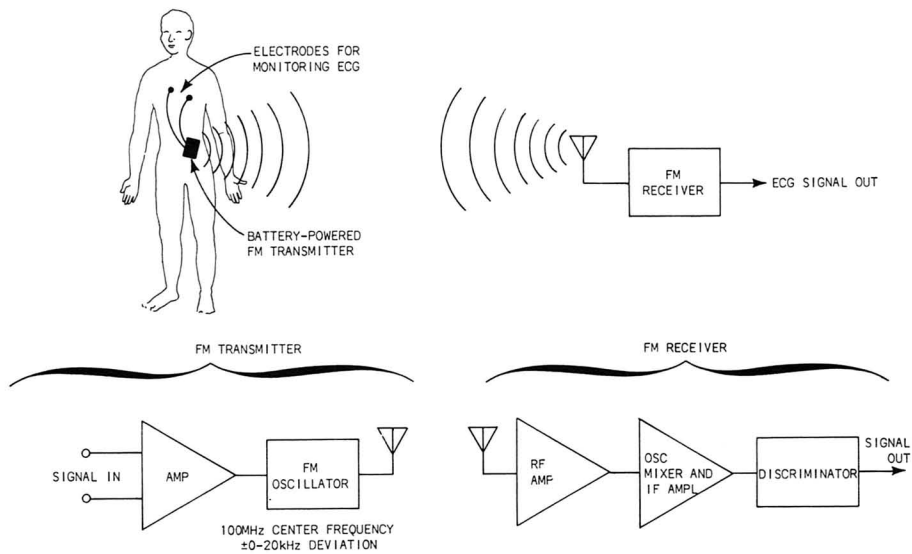
Fig. 27-1. A typical single-channel telemetry link.

FM/FM TRANSMITTER

100MHz CENTER FREQUENCY
±0-20kHz DURATION

SUBCARRIER OSCILLATORS

1 3kHz CENTER FREQUENCY
   ±0-225Hz DEVIATION

2 3.9kHz CENTER FREQUENCY
   ±0-293Hz DEVIATION

3 5.4kHz COMPUTER FREQUENCY
   ±0-405Hz DEVIATION

IRIG PROPORTIONAL SUBCARRIER BANDS (Hz)

BAND	LOWER LIMIT	CENTER FREQ	UPPER LIMIT	BAND WIDTH	NOMINAL INTELL. FREQ
1	370	400	430	60	6
2	518	560	602	84	8
3	675	730	785	110	11
4	888	960	1,032	144	14
5	1,202	1,300	1,398	196	20
6	1,572	1,700	1,828	256	25
7	2,127	2,300	2,473	346	35
8	2,775	3,000	3,225	450	45
9	3,607	3,900	4,193	586	60
10	4,995	5,400	5,805	810	80
11	6,799	7,350	7,901	1,102	110
12	9,712	10,500	11,288	1,576	160
13	13,412	14,500	15,588	2,176	220
14	20,350	22,000	23,650	3,300	330
15	27,750	30,000	32,250	4,500	450
16	37,000	40,000	43,000	6,000	600
17	48,560	52,500	56,440	7,880	790
18	64,750	70,000	75,250	10,500	1,050
19	86,025	93,000	99,975	13,950	1,400
20	114,700	124,000	133,300	18,600	1,900
21	152,625	165,000	177,375	24,750	2,500
A	18,700	22,000	25,300	6,600	660
B	25,500	30,000	34,500	9,000	900
C	34,000	40,000	46,000	12,000	1,200
D	44,620	52,500	60,380	15,760	1,600
E	59,500	70,000	80,500	21,000	2,100
F	79,050	93,000	106,950	27,900	2,800
G	105,400	124,000	142,600	37,200	3,700
H	140,250	165,000	189,750	49,500	5,000

FM/FM RECEIVER

Fig. 27-2. A typical multichannel telemetry link — frequency division multiplexed.
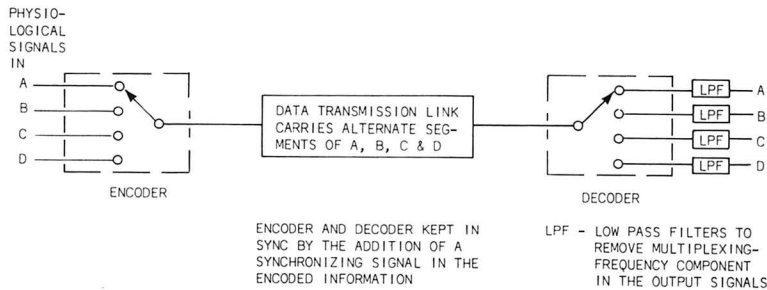
Fig. 27-3. Principle of time division multiplexing.

A typical multichannel telemetry link is shown
diagramatically in Fig. 27-2. The input signal is
used to frequency modulate subcarrier oscillators
operating below 100 kHz; the frequency modulated
outputs from these three subcarrier oscillators are
mixed and used to frequency modulate a high frequency
FM/FM system   oscillator. This system is referred to as an FM/FM
system as both the subcarrier oscillators and the
main oscillator are frequency modulated. A list of
IRIG FM/FM subcarrier frequency bands is shown in
Fig. 27-2. The maximum signal that can be
transmitted via the telemetry link is known as the
nominal intelligence frequency and is directly
proportional to the bandwidth of the subcarrier
channel used. It can be seen from Fig. 27-2 that,
at a subchannel bandwidth of 24.75 kHz, the nominal
intelligence frequency is only 2.5 kHz. Many
multichannel telemetry systems use wider channel
bandwidths and thus require correspondingly greater
channel separations. Subcarrier bands 1 through 21
shown in Fig. 27-2 have relatively narrow
bandwidths; however, subcarrier bands A through H
have wider bandwidths to allow higher frequency
information to be transmitted.

27.3 TIME DIVISION MULTIPLEXING

Multiplexing refers to the combining of several
channels of information to allow this information to
be transmitted via one data transmission link. The
FM/FM telemetry system discussed previously is an
example of frequency division multiplexing as the
signals are all transmitted simultaneously using
various *frequency* bands to separate the signals.
frequency   Whereas frequency division multiplexing is true
division   *simultaneous transmission* of separate channels of
versus   data, time division multiplexing consists of
time   *sequential transmission* of separate channels of data.
division   This multiplexing technique is best illustrated by
the operation of a rotating commutator in which the
several input signals are switched sequentially onto
a common output channel. The multiplexed output
may be transmitted directly over wire or used to
modulate a high frequency carrier. At the
receiving end the signal is separated back into
individual channels by a decoder that is synchronized
with the transmitting encoder. This technique is
illustrated in Fig. 27-3. The frequency at which
the encoder switches between channels should be
substantially higher than the maximum information-
frequency content of either of the channels.

Although FM/FM multiplexing and time division
multiplexing are commonly used for biophysical
applications, many other information transmission
systems are used for special applications. These
transmission systems are not unique to the
biophysical sciences and any text on radio
communications should provide a good reference as
to other possible data transmission systems.

27.4 DATA PROCESSING WITH AN ANALOG COMPUTER

The analog computer is a highly versatile signal
processor. Analog computers consist of precision
electronic modules such as operational amplifiers,
voltage sources, voltage dividers, multipliers and
other special devices. The analog computer is able
to analyze nonelectronic systems by representing
these systems in electronic units. Basic system
quantities are normally represented by varying
voltage or current; system constants are represented

principle

by either discrete resistors or by voltage dividers.
Once a physical quantity has been converted to an
electronic analog, the analog computer processes
these electronic quantities by using various
operational amplifier configurations for
amplification, attenuation, integration, summation
and numerous other functions.

A sample problem and its solution via an analog
computer are shown in Fig. 27-4. The equation shown
(Fig. 27-4) represents the behavior of a blood flow
system when subjected to a step change in pressure.
If this step change in pressure (p) is represented
by a voltage step and the desired pressure in some
other part of the flow system (y) is represented
as an output voltage from the analog computer, then
the differential equation relating p to y can be
solved by the circuit shown. This circuit only uses
five operational amplifiers and two precision
attenuators; however, typical problems solved by an
analog computer may involve the use of many more
components. Commercial analog computers may vary in
size from units offering only six operational
amplifiers to extremely large installations offering
several hundred operational amplifiers. The
mechanical arrangement of operational amplifiers and
other modules in analog computers is normally
arranged so that interconnection between various
modules can be achieved by a patching network.

PROBLEM - A PHYSIOLOGICAL BLOOD-FLOW ANALYSIS

A STEP PRESSURE CHANGE TO A FLOW SYSTEM (p) CAUSES A DAMPED SINUSOIDAL PRESSURE CHANGE (y) TO BE PROPAGATED THROUGHOUT THE SYSTEM.   SYSTEM ANALYSIS YIELDS THE FOLLOWING SECOND ORDER DIFFERENTIAL EQUATION WITH SYSTEM CONSTANTS REPRESENTED BY C1, C2 AND C3.

$$C2\, \frac{d^2y}{dt^2} + C1\, \frac{dy}{dt} + C3\, y = p$$

SOLUTION - SOLVE THE ABOVE EQUATION WITH OPERATIONAL AMPLIFIERS AND CALIBRATED DIVIDERS.

SINCE THE EQUATION REPRESENTS A DYNAMIC SYSTEM, SOLUTION OF THE ABOVE WOULD YIELD A COMPLEX SERIES OF TERMS TO REPRESENT THE OUTPUT OF THE SYSTEM.   THE ANALYSIS CAN BEST BE ACCOMPLISHED WITH AN ANALOG COMPUTER. THE ABOVE EQUATION CAN BE REWRITTEN:

$$\frac{p}{C2} - \frac{C1}{C2}\frac{dy}{dt} - \frac{C3}{C2}\, y = \frac{d^2y}{dt^2}$$
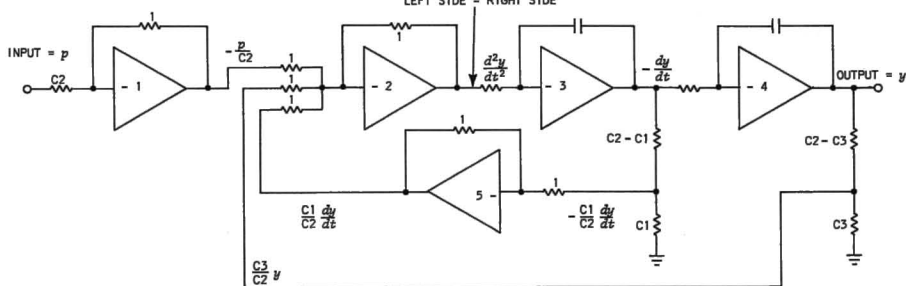
Fig. 27-4. Physiological system analysis with an analog computer.

27.5 DATA PROCESSING WITH A DIGITAL COMPUTER

time
sharing

With the advent of digital computer time sharing techniques, the use of the digital computer is now economically feasible for many biophysical measurement applications. While the central processing unit of a digital computer can normally only process information from one source at any one time, it can often process this information in a fraction of a second and then begin processing information from another source. This is referred to as time sharing. In a time shared system, due to the speed of the computer, the user is often unaware that the central processing unit is sharing its time between him and many other users.

digital-
analog
conversion

Data must be presented to, and is received from, a digital computer in a digital format. If analog data is to be processed with a digital computer, it must, therefore, be converted via an analog to digital converter. Conversely, if the data received from the digital computer is required in an analog format, it must be converted to this format in a digital to analog converter. If a computer is located remotely from the analog data gathering site, it is preferable to transmit and receive data in analog form. If the data is digitized on site, it must be either transmitted in parallel to the computer (one cable per bit), which is very expensive, or serially, which is also expensive and slow as well. It is quite feasible for ECG or EEG data, but impractical for direct brain recording or action potentials.

real time
processing

Since the time scale of analog data is usually important, this time scale must be maintained when processing analog data via a digital computer. It is thus necessary that the computer process this data in real time as shown in Fig. 27-5A. This precludes the direct use of a time shared computing system unless the priorities in the time shared system are arranged so that processing of this analog information takes absolute priority over any other processing. In this case, the time shared computer system would process this data in real time. For this reason, many self contained small computers are used in this application.

Fig. 27-5. Analog data to and from a digital computer.

Although the time scale of analog data is usually important and must be maintained during a data processing procedure, it is possible for a time shared computer system to indirectly process this data in other than real time and then to reconstruct the data in real time after a finite delay through the use of storage devices. Referring to Fig. 27-5B, analog data is converted to digital data via an analog to digital converter; this digital data is then stored in a digital storage device (normally located at the computer site). The time shared computing system then accepts data from the digital storage device on a time shared basis with transfer being controlled by the computer. Digital data from the computer will then be produced in segments as the computing system finds time to process stored data. These segments of digital data can then be converted to analog data and stored or they may be stored in a digital storage device before converting. All of the analog data can then be withdrawn from the storage device to preserve the time scale of the data.

preserving
time scale

TEKTRONIX TYPE T4002 GRAPHIC COMPUTER
TERMINAL WITH ANALOG INPUT CAPABILITY

11In STORAGE
DISPLAY CRT

PHYSIOLOGICAL SIGNAL
TO AUXILIARY
INPUT INTERFACE

ACOUSTIC
COUPLER
OR MODEM

TELEPHONE
LINE TO
COMPUTER

KEYBOARD ALLOWS COMMANDS
TO BE SENT TO THE COMPUTER
ALONG WITH PHYSIOLOGICAL
SIGNAL IN DIGITAL FORMAT

Fig. 27-6. A computer terminal used for physiological signal analysis
and processing via a remote computer.

Type T4002   The Tektronix Type T4002 Graphic Computer Terminal
              shown in Fig. 27-6, when interfaced with a custom
              analog input module, is ideally suited to analog
              input data processing with a time shared computing
              system. The terminal, when used in conjunction with
              an acoustic coupler or "modem" (MOdulator-
              DEModulator) and data communications service,
              provides the necessary electronics to implement the
              time shared computing system discussed previously
              and shown diagramatically in Fig. 27-5. The storage
              function is achieved by using the storage cathode-ray
              tube of the terminal. The communication line provided
              by the data communication service should have
              sufficient bandwidth (baud rate capability) to allow
              data transmission in real time. The ECG signal
              typically requires a 400 baud line.

27.6 DATA PROCESSING APPLICATIONS AND SOFTWARE

              Data processing applications in biophysical
              measurements require application software (computer
              programs) to control the computer in performing
              specific tasks. This software is continuously being
              developed by computer manufacturers, computer users
              and consulting firms and most computer manufacturers
              have a "library" of software programs available to
              their customers.

              The digital computer is particularly suited to signal
              averaging. Digital Equipment Corporation has a
signal   signal averaging program for their PDP-8/I computer
averaging   that also permits prestimulus averaging and
              computation of statistical information such as
              standard deviation and trends. The Digital Equipment
              Corporation's PDP-8/I computer may be used in
              conjunction with the Tektronix Type 611 Storage
              Display Unit and an analog to digital converter as
              a complete signal averaging instrument.

ECG
analysis

Computers are now extensively used for analysis of ECG waveforms. The information storage and waveform comparison capability of the computer allow it to compare an ECG signal with other ECG signals previously analyzed to determine if the signal is within acceptable statistical limits. The Control Data Corporation provides software for electrocardiogram analysis to use with their Model 1704 Computer; the complete package of computer software and peripheral equipment is designated the "1700 Computer Electrocardiogram Analysis System." This system identifies, measures and analyzes waveforms of 12-lead electrocardiograms by performing a pattern recognition analysis on the electrocardiogram and provides a printout of the results for diagnostic use.

Computers are ideally suited to simulation of physiological systems, thus, in many cases, allowing an analysis of physiological systems within a computer rather than by performing actual physiological measurements.

28

INTENSIVE CARE CONCEPTS

intensive
care units

Most diseases of the heart and of the circulatory system, referred to as cardiovascular diseases, strike without warning and prompt treatment is required if death is to be averted. Such treatment is best provided in a specialized area of a hospital referred to as an "intensive care unit." These specialized hospital units provide constant observation of the subject, constant monitoring of the subject's physiological condition and provide immediate emergency treatment should it be required. Coronary intensive care units are used for treatment of diseases of the heart such as the myocardial infarction or "heart attack." Stroke intensive care units are used for treatment of diseases of the circulatory system such as the stroke. Pulmonary intensive care units are used for treatment of respiratory diseases.

An intensive care unit may consist of one or more subject-monitoring sites, referred to as "beds" as each site is, in fact, a bed. Electronic instrumentation at each subject-monitoring site monitors various physiological signals from one subject and may activate alarms should these physiological signals be above or below predetermined limits. The complete intensive care unit consists of not only the necessary monitoring equipment but also the necessary trained personnel and emergency equipment to allow immediate treatment of cardiac malfunctions.

Intensive care units are usually constructed to
suit a particular hospital's requirements. Widely
varying approaches are taken as to the physiological
functions to be monitored and the type of monitoring
equipment required and, in general, no two intensive
care units are alike. Intensive care
instrumentation is continuously being developed to
accomplish more advanced physiological monitoring
techniques. The following discussion on intensive
care concepts is intended to present some of the
physiological functions that *may* be monitored and
some of the typical instrumentation that *may* be
used to monitor these functions. The discussion is
conceptual in nature and in no way attempts to
survey all current intensive care applications and
instrumentation.

28.1 PHYSIOLOGICAL FUNCTIONS TO BE MONITORED DURING INTENSIVE CARE

Since subjects in coronary intensive care units are
suffering from cardiovascular diseases, all
physiological functions associated with the heart
and the circulatory system should be monitored. In
monitoring these physiological functions, it should
be borne in mind that the subject is in a recovery
situation, thus, the physiological monitoring devices
should not hinder the recovery process and should be
as unobjectionable as possible to the subject.

The principal physiological signal monitored in an
intensive care unit is often the electrocardiogram.
The electrocardiogram is usually monitored in the
lead II configuration with two active electrodes
placed approximately 12 inches apart along the
maximum potential axis of the subject's heart. A
third electrode (ground) should be located elsewhere
ECG   on the chest. This electrocardiogram monitoring
monitoring   configuration is referred to as a three-lead chest
cluster. Tektronix produces a patient cable for
use with their Type 410 Physiological Monitor that
is specifically intended for monitoring during
intensive care. The electrodes used for ECG
monitoring during intensive care must be suited for
long term monitoring applications. The Tektronix
silver/silver-chloride electrode system provided with
the Type 410 Physiological Monitor is ideally suited

to this application as the electrode paste supplied produces no subject discomfort or skin irritation and the relatively large amount of paste required between the subject and the electrode prevents the paste from drying out due to evaporation and skin absorption.

blood
pressure

The second physiological parameter often of prime importance in intensive care monitoring is blood pressure. Blood pressure can be and often is monitored using an intra-arterial catheter and transducer; however, the catheter results in considerable subject discomfort and many intensive care units prefer to monitor blood pressure by some alternate method.

Korotkoff
system

Blood pressure can be monitored using the automatic cuff pump and Korotkoff microphone blood-pressure measurement system shown in Fig. 8-11. Although this system is occasionally used in intensive care units, it also possesses the disadvantage of being somewhat uncomfortable to the subject (bruises), and more importantly, being a sampling technique, it does not provide a continuous record of the subject's blood pressure. Thus, if for some reason the subject's blood pressure were to suddenly drop, this system may take a minute or so to detect this pressure drop.

plethysmo-
graph

Blood pressure monitoring with a plethysmograph offers the least discomfort to the subject; however, it provides only a relative indication of the well being of the circulatory system rather than providing absolute values for diastolic and systolic pressure.

Although considerable controversy still exists as to the best method for routine blood pressure measurement in an intensive care unit, each of these techniques possesses attributes and a single intensive care unit may employ one or more of these techniques and, indeed, all three may be available if required. Although diastolic and systolic arterial pressure are commonly monitored, mean arterial pressure and venous pressure are also monitored in some instances.

respiration

body
temperature

It is often desirable to monitor the subject's
respiratory activity during intensive care; this
may be accomplished with a thermistor pneumograph
placed in the subject's nostril. It is often also
desirable to monitor body temperature in intensive
care subjects via a rectal or armpit thermistor
probe.

pacemaker

Monitoring of the physiological signals referred to
previously necessitates numerous electrodes, etc.
being placed on the subject. In addition, it is
often desirable to have cardiac pacemaker electrodes
applied to the subject's chest. Although these
electrodes are not used during routine intensive
care, they should be connected to a cardiac
pacemaker for immediate emergency use if required.

28.2 INTENSIVE CARE INSTRUMENTATION

A conceivable intensive care instrumentation system
is shown in Figs. 28-1, 28-2 and 28-3. Fig. 28-1
depicts the total instrumentation layout in a four-
bed intensive care unit. Each of the four beds
includes separate subject-monitoring instrumentation
providing an indication of the subject's
physiological condition as indicated in Fig. 28-2.
Signals from each of these four instrumentation
modules are also connected to a central nurse's
station to permit monitoring by the nurse on duty,
to allow selective recording of the ECG, and to
allow the ECG signal and/or audio-visual information
to be transmitted throughout the hospital via a
closed-circuit television link. The television
camera and closed circuit link may be regarded as
"luxury items," most other features shown are
essential.

ELECTROCARDIOGRAM WAVEFORM AND/OR ALPHANUMERICS
DISPLAYED ON LOW-COST COMMERCIAL TELEVISION RECEIVERS
LOCATED THROUGHOUT THE HOSPITAL. ALLOWS AUDIO-VISUAL
COMMUNICATION FROM THE NURSE'S STATION TO MEDICAL
STAFF THROUGHOUT THE HOSPITAL.

VIA
CLOSED-CIRCUIT
TV SYSTEM

CENTRAL NURSE'S STATION

NURSE OVERSEES THE SYSTEM: OBSERVES INFORMATION
FROM SUBJECT MONITORS AND TRANSMITS INFORMATION
RELATING TO POSSIBLE PROBLEMS TO THE TELEVISION
RECEIVERS.

INSTRUMENTATION CONSOLE

VIA SHIELDED CABLES

INTENSIVE CARE WARD — FOUR BED

VIA
CLOSED-CIRCUIT
TV SYSTEM

REMOTE-CONTROL TV CAMERA
MOUNTED ABOVE SUBJECTS

SUBJECT-MONITORING
INSTRUMENTATION

SUBJECT REQUIRING INTENSIVE CARE

Fig. 28-1. Instrumentation in an intensive care unit.
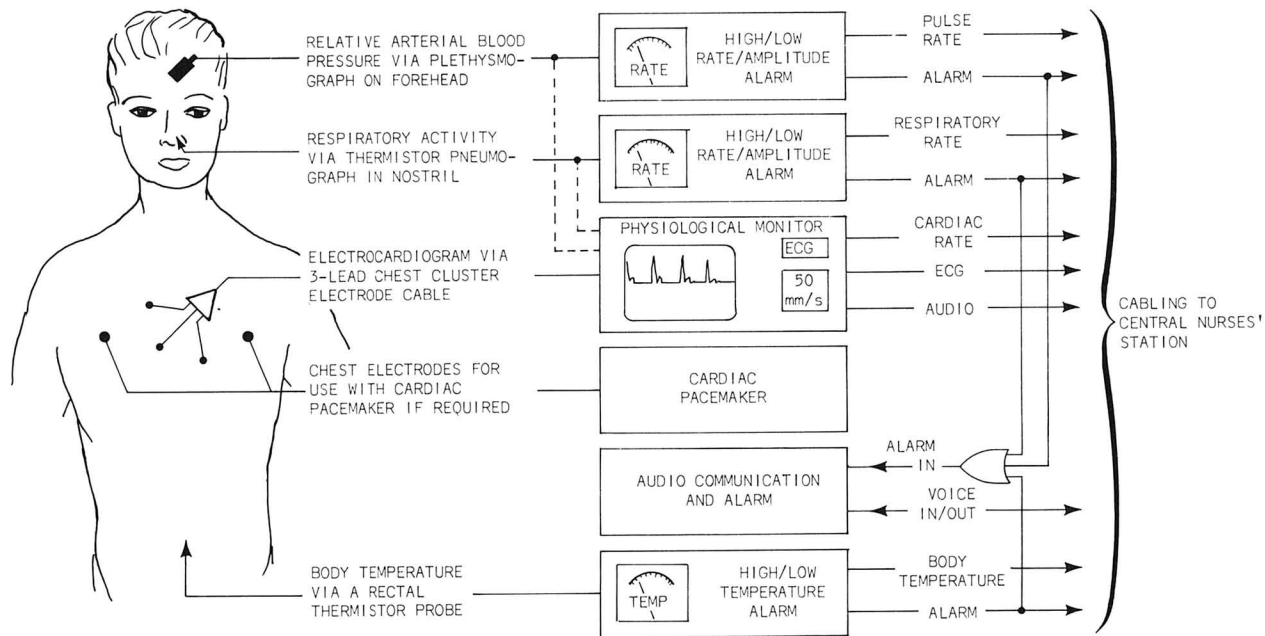
Fig. 28-2. Subject monitoring instrumentation for one intensive care "bed."

Referring to the subject-monitoring instrumentation located beside each intensive care bed as shown in Fig. 28-2, relative arterial blood pressure, respiratory activity, the electrocardiogram and body temperature are monitored. Relative arterial blood

plethysmo-
graph

pressure is monitored via a plethysmograph on either the subject's forehead, his nasal septum or the lobe of his ear. Finger plethysmographs are rarely used in intensive care due to their susceptibility to subject movement. The instrumentation associated with the plethysmograph may provide an indication of cardiac rate; it may also provide an alarm signal should the cardiac rate go above or below preset limits or should the amplitude of signal produced by the plethysmograph fall below a predetermined level (indicating a loss in blood pressure or flow).

Respiratory activity is monitored via a thermistor pneumograph in the nostril with the associated instrumentation providing an indication of respiratory rate as well as providing an alarm if this rate falls outside predetermined limits or if the signal level produced by the thermistor

pneumo-
graph

pneumograph is reduced below some predetermined amplitude (indicating a loss of respiratory activity).

Type 410
Monitor

The electrocardiogram is monitored by a Tektronix Type 410 Physiological Monitor using a Tektronix three-lead chest-cluster electrode cable. The physiological monitor provides an ECG output from a low impedance source as well as an audio output to audibly indicate the cardiac rate or to indicate a loss of cardiac activity.

thermistor
probe

Body temperature is monitored via a rectal thermistor probe. The associated instrumentation indicating body temperature may also contain an alarm system which will be activated if the body temperature should fall outside predetermined limits.

Instrumentation located beside each intensive care bed should, preferably, be away from the subject's range of vision as its presence can be somewhat disconcerting to some subjects. The intensive care ward may also contain a closed-circuit television system to allow one or more subjects to be viewed

television
monitoring

via a television camera. This television camera may continuously scan the subjects in the intensive care ward or its position may be controlled from the central nurse's station.

additional
equipment

In addition to physiological and visual monitoring of the subject, a cardiac pacemaker module and audio communication and alarm module are also included with the instrumentation at each intensive care bed. The cardiac pacemaker provides variable-amplitude, variable-rate pulses for cardiac pacemaking should it be required. The audio communication and alarm panel provides audio-visual alarm indication of abnormalities in blood pressure, cardiac rate, respiratory activity or body temperature and provides audio communication between the intensive care bed and the nurse's station.

central
nurse's
station

An intensive care unit central nurse's station is shown in Fig. 28-3. Multiconnector cable connects the output from the four subject-monitoring sites located beside each intensive care bed to the central nurse's station. Each subject's ECG is continuously displayed via a four channel oscilloscope and is continuously recorded on a memory loop tape recorder. This tape recorder contains the previous one-minute ECG history for each subject by recording the ECG on a tape loop "one minute" in length. While some central stations duplicate all physiological indicators at the central station, this is normally unnecessary and, in the interest of simplicity, it is preferable to only provide one set of indicators for relative blood pressure, respiratory activity and body temperature. These indicators can be manually switched between the four beds or the switching may be activated by the alarm system with the monitors being automatically switched to the bed providing the alarm signal. When an alarm is received at the central nurse's station, it may also be used to connect the appropriate ECG signal to a scan converter and ECG chart recorder and to start the chart recorder. In this way a permanent record is achieved on the chart recorder beginning one minute prior to the alarm being sounded and information is displayed on the scan converter for transmission via the hospital closed-circuit TV system to other medical personnel involved. The scan converter and closed-circuit TV system may also incorporate alphanumeric input to allow alphanumeric data relating to the intensive care subject to be displayed on television receivers located throughout the hospital.
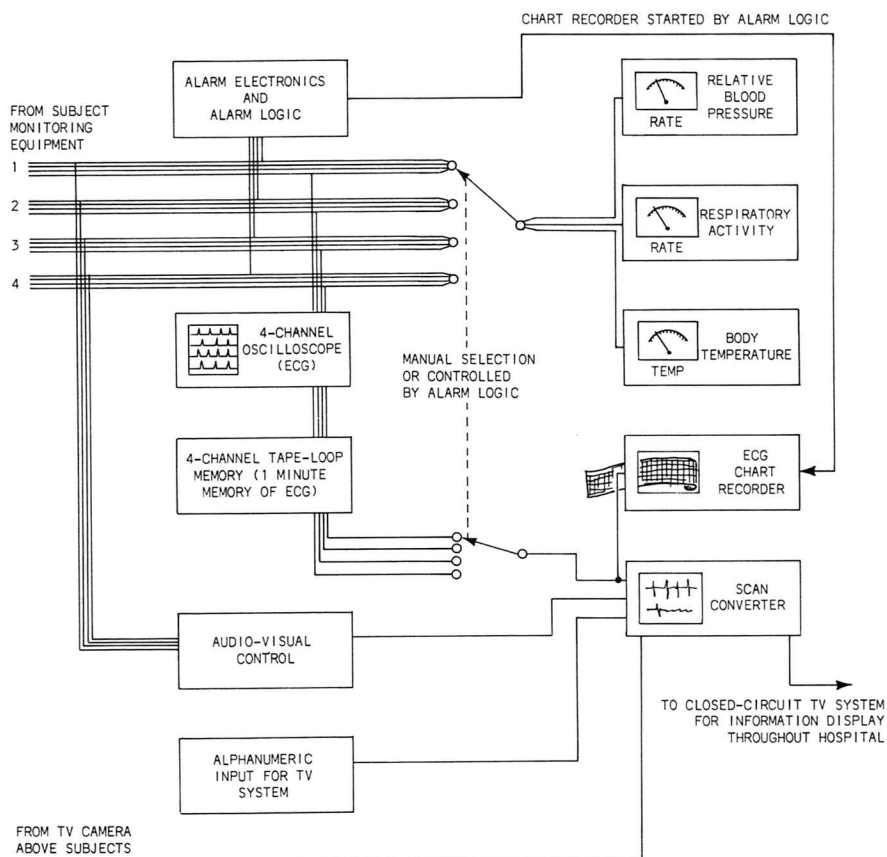
Fig. 28-3. Central nurses' station instrumentation for 4 intensive care beds.

computers
in
intensive
care

Increasing use is being made of digital computers in intensive care units. These computers provide storage of the subject's physiological data and can continuously interrogate this data to determine if it is within predetermined limits. The computer can perform multivariant analysis of the subject's ECG and can compare his ECG with previous records to indicate changes in the ECG which may be clinically significant. The advent of the time shared computer system allows one computer to simultaneously perform many tasks throughout a hospital, thus making the computer economically feasible for many hospitals.

SECTION IV

APPENDIX

Chapter 29 describes custom instrumentation required
to perform the measurements discussed in Section II.
While most of these custom items are simple and
should more correctly be referred to as adapters,
they are necessary for the correct operation of the
measurement systems and are not generally available
commercially.

Chapter 30 gives formal definitions for the biomedical
terminology used throughout this book. It is hoped
that this chapter contains the definitions of any
terms used that would not normally be regarded as
common terminology by the electronics engineer.

29

CUSTOM INSTRUMENTATION

This chapter provides circuits and construction
details for various items of custom instrumentation
that are required to implement some of the
biophysical measurements discussed in Sections II
and III. Details of these items are presented here
since similar items do not appear to be available
commercially. These items have been constructed and
used satisfactorily for the measurement technique
for which they were intended. Normal tolerance
variations in components may, however, necessitate
minor circuit changes in some instances. It is
anticipated that the items presented will be
constructed by competent electronic technicians or
electronic engineers who are able to evaluate the
performance of the device and modify this performance
if necessary.

Although the Tektronix, Inc. part number is provided
for all the parts necessary to construct these items,
most of the parts are common electronic components
and are available through local suppliers. We wish
to encourage purchase from these local suppliers
whenever possible. If necessary, however, parts may
be purchased in small quantities through Tektronix
field offices located throughout the United States
or through Tektronix subsidiaries or distributors in
other countries. These field offices, subsidiaries
and distributors are listed in the current Tektronix
catalog. When purchasing these parts from Tektronix,
please quote the Tektronix part numbers for the
individual components shown with the circuits. While
all parts were available from Tektronix at the time
of preparation of this text, Tektronix makes no
guarantee as to the continued availability of these
parts.

The items described in Sections 29.1 through 29.5 are intended for use in conjunction with the Tektronix Type 410 Physiological Monitor.

The items described in Sections 29.6 through 29.9 are intended for use with the Tektronix Type 3A8 Operational Amplifier plug-in unit. These items may also be used in conjunction with the Tektronix Type O Operational Amplifier plug-in unit. It is also probable that these circuits may be suited for use with other operational amplifier modules.

The items described in Sections 29.10 through 29.15 are independent items of instrumentation and are not intended for use with any specific instrument.

29.1 TYPE 410 MODIFICATION FOR FETAL ECG USE

Monitoring of the fetal ECG requires the use of a physiological monitor with an amplifier having a low frequency 3 dB point of >1 Hz. The standard 410 Physiological Monitor has a low frequency response of <0.1 Hz and is thus unsuited for monitoring the fetal ECG. The EEG function of the Type 410 Physiological Monitor may be modified to allow it to be used to record the fetal ECG. This modification consists of modifying the low frequency response of the Vertical Amplifier in the EEG position from <0.1 Hz to >1 Hz. See Fig. 29-1.

The low frequency response of the amplifier is changed by decreasing the time constant of the AC coupling network from 1 microfarad/2.2 megohm to 1 microfarad/75 kilohm by adding 75 kilohm resistors when the 410 is operated in the EEG mode. This also reduces the amplifier's gain by a factor of 2. Since no spare switch positions are available on the 410 INPUT SELECTOR switch, switching of the 75 kilohm resistors must be accomplished using transistors. The 410 horizontal and audio circuit produces a positive logic level when the 410 is operated in the EEG mode. This level is used to freerun the sweep and to disable the audio circuit, however, it may also be used in this modification to turn on the switching transistors Q1A and Q1B.

TO JUNCTION
D134, D135

R1
75kΩ

C1
.047µF

Q1A

R3
360kΩ

R4
360kΩ

D1

TO HORIZONTAL
AND AUDIO CIRCUIT
BOARD. PIN H.

R5
360kΩ

Q1B

R2
75kΩ

C2
.047µF
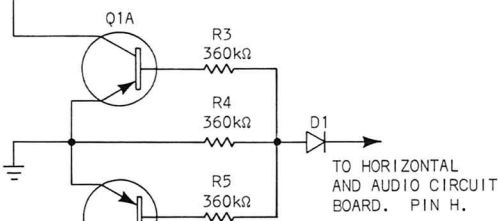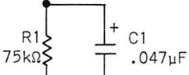
TO JUNCTION
D235, D236
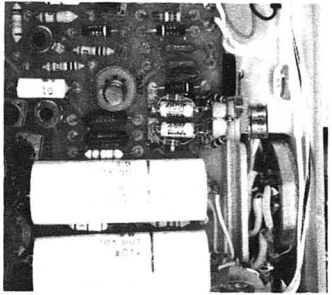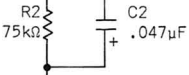
ADDITIONAL COMPONENTS
LOCATED WITHIN THE 410.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.047µF 35V CAPACITOR	290-0282-00
C2	.047µF 35V CAPACITOR	290-0282-00
D1	CD6538 DIODE	152-0185-00
Q1	DUAL PNP TRANSISTOR	151-0261-00
R1	75kΩ 1/8W RESISTOR	317-0753-00
R2	75kΩ 1/8W RESISTOR	317-0753-00
R3	360kΩ 1/8W RESISTOR	317-0364-00
R4	360kΩ 1/8W RESISTOR	317-0364-00
R5	360kΩ 1/8W RESISTOR	317-0364-00
--	DUAL TRANSISTOR SOCKET	136-0235-00
--	8 INCH INSULATED WIRE	--

TO HORIZONTAL AND
AUDIO CIRCUIT BOARD
PIN H

CIRCUIT
GROUND

CIRCUIT BOARD
JUNCTION OF
D236, D235, R234,
C234, D260, Q137B

CIRCUIT BOARD
JUNCTION OF
D135, D134, R134,
C134, D160, Q137A

R2
C2
R5
D1
TAB
R4
DUAL
TRANSISTOR
SOCKET
Q1A,B
C1
R1
R3

EXISTING
COMPONENTS
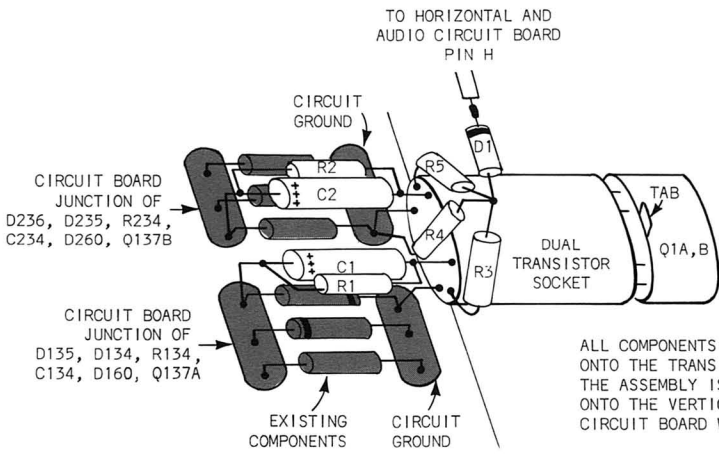
CIRCUIT
GROUND

ALL COMPONENTS ARE MOUNTED
ONTO THE TRANSISTOR SOCKET.
THE ASSEMBLY IS THEN WIRED
ONTO THE VERTICAL AMPLIFIER
CIRCUIT BOARD WITHIN THE 410.

Fig. 29-1. Type 410 modification for fetal ECG use.
EEG position now 5 mm/50 µV, 1 Hz — 100 Hz.

	FOR 50µV/cm SENSITIVITY	FOR 500µV/cm SENSITIVITY	FOR 5mV/cm SENSITIVITY
	EEG +	ECG LA	AUX +
	EEG −	ECG RA	AUX −

QTY	DESCRIPTION	TEKTRONIX PN
2	BNC CONNECTORS	131-0352-01
3	2-PIN PLUGS (P1,P2,P3)	134-0089-00
3 (ft)	SHIELDED CABLE	175-0284-00
1	BOX w/HARDWARE *	013-0073-00

*CONSTRUCTION NOTE:

- REMOVE 3 BANANA PLUGS FROM BOX
- ENLARGE THE 2 OUTER VACATED HOLES TO 5/15 INCH
- TAP ENLARGED HOLES 3/8 INCH × 32 TAP

Fig. 29-2. A BNC input adapter for the Type 410 Monitor.

CKT NO	DESCRIPTION	TEKTRONIX PN
R1	1.47kΩ RESISTOR	321-0209-00
R2	5kΩ POTENTIOMETER	311-0171-00
R3	3kΩ RESISTOR	318-0099-00
R4	3kΩ RESISTOR	318-0099-00
2 ea	BNC CONNECTORS	131-0352-01
6 ea (ft)	2 CONDUCTOR SHIELDED CABLE	175-0152-01
1 ea	BOX w/HARDWARE *	013-0073-00
1 ea	7-PIN CONNECTOR (P1)	131-0551-01
1 ea	WASHER, LOCKING	210-0046-00
1 ea	1/4 INCH NUT	210-0583-00

*CONSTRUCTION NOTE:

- REMOVE 3 BANANA PLUGS FROM BOX
- ENLARGE THE 2 OUTER VACATED HOLES TO 5/16 INCH
- TAP ENLARGED HOLES 3/8 INCH × 32 TAP
- FILE SLOT IN BOX FOR CABLE CLEARANCE
- ENLARGE THE CENTER VACATED HOLE TO 1/4 INCH

Fig. 29-3. A signal output adapter for the Type 410 Monitor that
provides a differential signal output from the Type 410
Monitor at 100 mV for each cm deflection on the 410 CRT.

29.2   A BNC INPUT ADAPTER FOR THE TYPE 410 MONITOR

Refer to Fig. 29-2. This adapter requires little
explanation, being entirely passive. The adapter
simply provides BNC to "410 System" adaption to
allow conventional BNC accessories, etc., to be used
to connect physiological signals to the Type 410
Monitor.

Referring to the markings located on the Type 410
patient cable, the adapter may be connected to the
AUX+ and AUX- positions to allow the 410 to be used
in the auxiliary mode, providing a sensitivity of
5 mV per centimeter. With the adapter connected to
the ECG LA and ECG RA positions, the 410 must be
operated in the ECG lead I mode and will provide a
sensitivity of 500 μV per centimeter. With the
adapter connected to the EEG+ and EEG- positions,
the Type 410 must be operated in the EEG mode and
will provide a sensitivity of 50 μV per centimeter.

29.3   A SIGNAL OUTPUT ADAPTER FOR THE TYPE 410 MONITOR

Refer to Fig. 29-3. This adapter provides a
vertical signal output from the Type 410 Vertical
Amplifier via conventional BNC connectors. A
differential output of 100 mV for each centimeter of
vertical deflection on the Type 410 CRT is provided.
The adapter should, preferably, be used with
auxiliary equipment having a differential input,
however satisfactory results may be obtained when
using only one output from the adapter in
conjunction with equipment having only a "single
ended" input. The sensitivity in the single ended
mode will be 50 mV for each centimeter of vertical
deflection and care should be taken to avoid
excessive common mode signals at the 410 input.

The Type 410 provides a differential vertical signal
out at the rear of the instrument of approximately
65 μA for each centimeter of CRT deflection. A
differential output voltage is produced across a
load on this current source consisting of R1, R2,
R3 and R4. R2 varies this load to allow the output
voltage to be varied. If alternative output voltages
are required the values of R3 and R4 should be
changed accordingly.

29.4 A RESISTIVE TRANSDUCER ADAPTER FOR THE TYPE 410 MONITOR

Refer to Fig. 29-4. This adapter allows resistive transducers to be used with the Type 410 Monitor operating in the auxiliary mode. A change in the resistive transducer's nominal resistance of less than 0.1% produces 1 centimeter of vertical deflection on the Type 410 CRT. The frequency response of the system is from 0.1 Hz to 250 Hz.

R1, R2, Q1 and Q2 provide a constant current source for the resistive transducer, thus the voltage across the transducer is proportional to the transducer's resistance. Capacitors C1 and C2 couple changes in this voltage to the vertical amplifier of the Type 410 Monitor.

Almost any resistive transducer having a nominal resistance below 10 kΩ may be used, however the adapter is not intended for use with bridge type resistive transducers. The adapter was specifically constructed for use with the thermistor pneumograph described in Section 29.5.

TRANSDUCER SUPPLY CURRENT	VALUE OF R2	TEKTRONIX PN FOR R2	OPTIMUM TRANSDUCER NOMINAL RESISTANCE	SENSITIVITY CHANGE IN RESISTANCE (δR) FOR 1 DIV VERTICAL DEFLECTION ON 410 CRT
$I \approx$ 1mA	680Ω	317-0681-00	10kΩ – 5kΩ	$\delta R \approx$ 5 OHMS/DIV
2mA	330Ω	317-0331-00	5kΩ – 2kΩ	2.5 OHMS/DIV
5mA	130Ω	321-0108-00	2kΩ – 1kΩ	1 OHM/DIV
10mA	62Ω	317-0620-00	LESS THAN 1kΩ	0.5 OHM/DIV

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	1μF 35V CAPACITOR	290-0308-00
C2	1μF 35V CAPACITOR	290-0308-00
P1	7-PIN PLUG	134-0090-00
Q1	SILICON NPN	151-0206-00
Q2	SILICON NPN	151-0206-00
R1	100kΩ 1/8W RESISTOR	317-0104-00
R2	* 1/8W RESISTOR	*
R3	RESISTIVE TRANSDUCER	* --
QTY – 4 ft	SHIELDED CABLE	175-0284-00

*SEE CHART FOR VALUE AND PN

ALL COMPONENTS ARE MOUNTED ONTO THE PINS OF THE 7-PIN PLUG. THE PLUG COVER IS THEN FITTED LEAVING THE SHIELDED CABLE PROJECTING FOR CONNECTION TO A RESISTIVE TRANSDUCER.
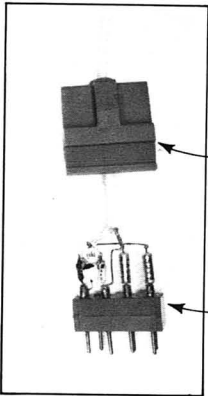
COVER, PART OF P1, SNAPS OVER PLUG AND COMPONENTS

P1

SHIELDED CABLE TO RESISTIVE TRANSDUCER

Fig. 29-4. A resistive transducer adapter for the Type 410 Monitor.
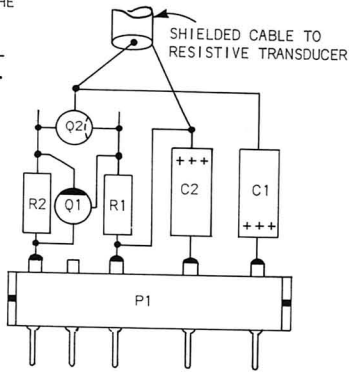
- SOLDER THERMISTOR TO CABLE
- SLIDE HOLDER OVER JUNCTION
- FIX HOLDER IN PLACE WITH CEMENT
- SOLDER CABLE TO TRANSDUCER ADAPTER
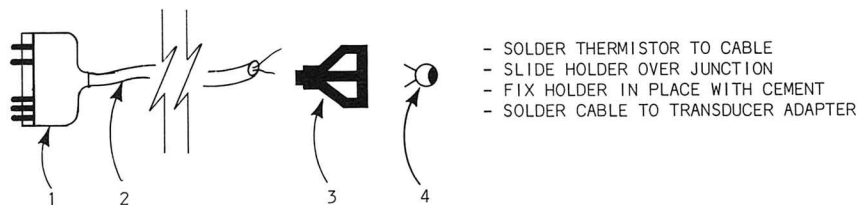
ITEM	DESCRIPTION	QTY	TEKTRONIX PN
1	RESISTIVE TRANSDUCER INTERFACE ADAPTER - 5mA.	1	REFER FIG 29-4 FOR DETAILS
2	PVC-COVERED COAXIAL CABLE	4 (ft)	175-0284-00
3	HOLDER - LARGE 7/16 INCH DIA	1	352-0077-00
	OR SMALL 1/4 INCH DIA		352-0078-00
4	1kΩ THERMISTOR	1	307-0127-01

THERMISTOR PNEUMOGRAPH
w/LARGE HOLDER
(FOR ADULTS)
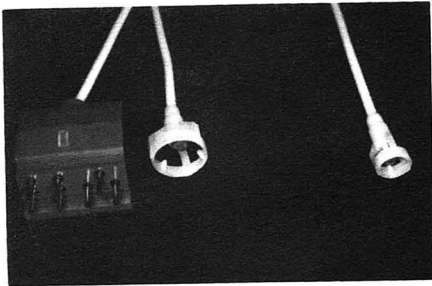
OPTIONAL SMALL
HOLDER
(FOR CHILDREN)

THERMISTOR PNEUMOGRAPH
FITTED TO SUBJECT
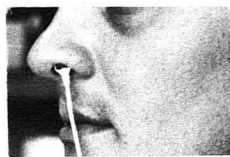
Fig. 29-5. A thermistor pneumograph for the Type 410 Monitor.

29.5 A THERMISTOR PNEUMOGRAPH FOR THE TYPE 410 MONITOR

Refer to Fig. 29-5. This pneumograph allows a
subject's respiration to be monitored by monitoring
the temperature difference between inhaled and
exhaled air with a thermistor.

A thermistor having a nominal resistance of 1 kΩ is
used in conjunction with suitable mechanical holders
and the resistive transducer adapter described in
Section 29.4. Either a large holder (for adults)
or a small holder (for children) may be used to hold
the pneumograph in place at the subject's nostril.

29.6 AN INPUT NEUTRALIZING ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-6. This adapter provides the extremely high input impedance necessary to allow amplification of signals derived from microelectrodes. The input resistance into this adapter is greater than 10^9 ohms, the input capacitance is less than 1 picofarad and the adapter has a gain of 2X and a low output impedance to allow it to be used in conjunction with conventional instrumentation. The adapter uses both operational amplifier units available in the Tektronix Type 3A8 Operational Amplifier plug-in unit.

Operational amplifier #1 is operating as a non-inverting amplifier having a gain of 2 determined by R2, R3 and R4. R1, C1 and C2 provide frequency response compensation for these gain setting resistors. Positive feedback is produced via R7, C3 and C4 to a current summing point at the junction of R5 and R6. This positive feedback current is adjusted so as to equal the sum of the input current to the operational amplifier via R5 plus the current through R6. Thus the current from the input must be zero. The output from operational amplifier #1 provides the output from the adapter and also provides an input to operational amplifier #2 operating as a unity gain follower. The components associated with the positive input of operational amplifier #2 form a 2X attenuator, thus the output from operational amplifier #2 is half of the output of operational amplifier #1, that is, equal to the input to operational amplifier #1. The output from operational amplifier #2 is used to drive an ungrounded shield surrounding the input to the adapter. This provides input "guarding" to effectively reduce the capacitance of the input cable.

REFER ALSO TO 3A8 INSTRUCTION MANUAL.
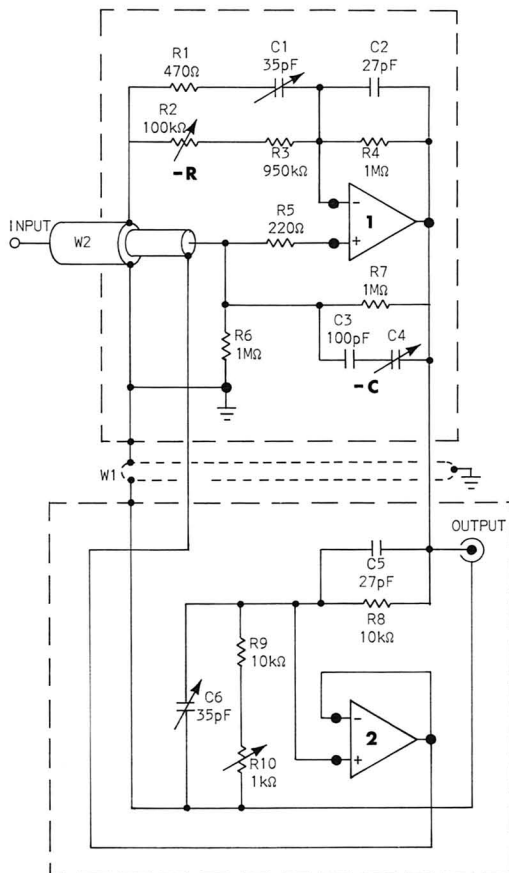
$R_i > 10^9 \Omega$ SET 3A8 OPERATIONAL
$C_i < 1pF$ AMPLIFIERS Z_i AND Z_f
        SWITCHES TO EXT,
GAIN 2X   GRID SELECTION TO + GRID.
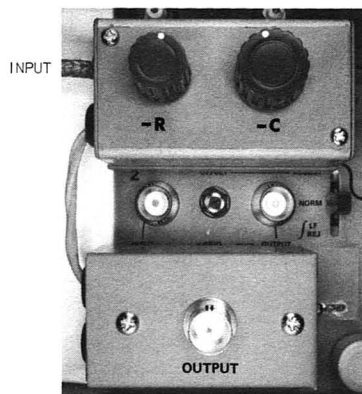
SET C1 FOR OPTIMUM TRANSIENT
RESPONSE WITH INPUT CONNECTED
DIRECTLY TO 40mV CALIBRATOR.

SET C6 AND R10 FOR OPTIMUM TRAN-
SIENT RESPONSE WITH INPUT CONNEC-
TED TO 40mV CALIBRATOR VIA 10MΩ
SERIES RESISTANCE.

R2 AND C4 ARE FRONT PANEL
ADJUSTMENTS.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	9-35pF CAPACITOR	281-0092-00
C2	27pF CAPACITOR	283-0076-00
C3	100pF CAPACITOR	283-0060-00
C4	VAR CAPACITOR (-C)	-- *
C5	27pF CAPACITOR	283-0076-00
C6	9-35pF CAPACITOR	281-0092-00
R1	470Ω 1/4W 5% RESISTOR	315-0471-00
R2	100kΩ POTENTIOMETER (-R)	311-0467-00
R3	950kΩ 1/8W 1% RESISTOR	318-0095-00
R4	1MΩ 1/8W 1% RESISTOR	321-0481-00
R5	220Ω 1/4W 5% RESISTOR	315-0221-00
R6	1MΩ 1/8W 1% RESISTOR	321-0481-00
R7	1MΩ 1/8W 1% RESISTOR	321-0481-00
R8	10kΩ 1/8W 1% RESISTOR	321-0289-00
R9	10kΩ 1/8W 1% RESISTOR	321-0289-00
R10	1kΩ POTENTIOMETER	311-0635-00
W1	1 ft 2-CONDUCTOR SHIELDED CABLE	175-0072-00
W2	1 ft 1-CONDUCTOR SHIELDED CABLE	175-0284-00
	1 ft SHIELDING BRAID WIRE	176-0047-00

MECHANICAL		
QTY	DESCRIPTION	TEKTRONIX PN
1	ADAPTER-TERMINAL ASSY	013-0048-01
1	COMPENSATION ADAPTER *	013-0081-00
1	1/4 INCH GROMMET	348-0020-00
3	5/16 INCH GROMMETS	348-0003-00
1	CONNECTOR BNC	131-0106-01
1	KNOB FOR R2	366-0153-00
1	NUT FOR R2	210-0583-00
1	WASHER-LOCKING FOR R2	210-0046-00
1	LUG-SOLDER FOR R2	210-0223-00

*C4 AND ASSOCIATED HARDWARE ARE SUPPLIED
AS PART OF THE COMPENSATING ADAPTER. THE
TEKTRONIX PN FOR C4 IS 281-0090-00.

Fig. 29-6. An input neutralizing adapter for the Type 3A8 Operational Amplifier.

29.7 AN ABSOLUTE VALUE ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-7. This adapter provides full wave rectification of input signals producing positive output signals for both positive and negative input signals. The adapter has an absolute gain of 1.

With a negative going input, diode D1 conducts and the operational amplifier operates as an inverting amplifier with unity gain determined by R1, R2, R3 and the output impedance of the input source. With a positive going input, diode D2 conducts and the operational amplifier operates as a unity gain, noninverting amplifier, as "Z_i" at the operational amplifier negative input is effectively infinite. R4 ensures that the adapter has the same input resistance for both positive and negative signals, thus maintaining a constant potential on C1. Diodes D1 and D2 effectively reduce the input voltage level by approximately 0.3 volts.

INPUT
VIA 3A8

OUTPUT
VIA 3A8

C2
15pF

C1
1μ

D1

R1
100kΩ

R2
200kΩ

D2

R3
2.2MΩ

R4
200kΩ
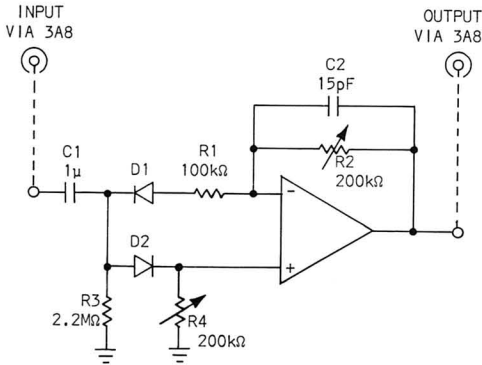
ADJUST R4 FOR 50% DUTY FACTOR WITH SINEWAVE
ADJUST R2 FOR SYMMETRY WITH SINEWAVE

INPUT   OUTPUT

R4

R2
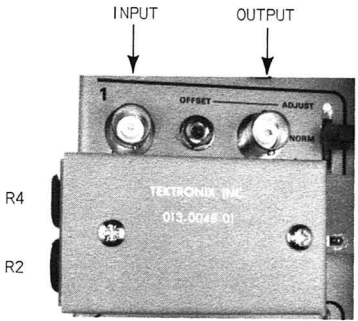
CKT NO	DESCRIPTION	TEKTRONIX PN
C1	1μF 100V CAPACITOR	285-0815-00
C2	15pF CAPACITOR	281-0509-00
D1 D2	MATCHED PAIR DIODES	152-0110-00
R1	100kΩ 1/4W RESISTOR	315-0104-00
R2	200kΩ POTENTIOMETER	311-0660-00
R3	2.2MΩ 1/4W RESISTOR	316-0225-00
R4	200kΩ POTENTIOMETER	311-0660-00

MECHANICAL
  QTY

| 1 | ADAPTER TERMINAL ASSY | 013-0048-01 |
| 2 | 5/16 INCH GROMMETS | 348-0003-00 |

PERFORMANCE AT 500Hz

10V P-P INPUT   OUTPUT = |INPUT|

1V P-P INPUT   OUTPUT = 0.75|INPUT|

0.1V P-P INPUT
OUTPUT = 0.25|INPUT| + 5mV
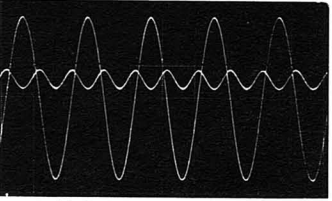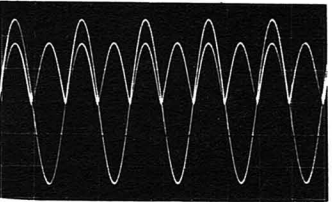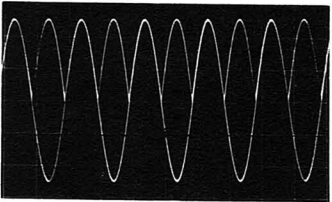
Fig. 29-7. An absolute value adapter for the Type 3A8 Operational Amplifier.

462

REED RELAY (S1) OPENS AND CLOSES
IN LESS THAN 1ms

POSITIVE GATING PULSE OPENS S1

(FROM TEKTRONIX 561B
OR 564B OSCILLOSCOPE
CALIBRATOR)

S1

CAL

13mA

Q1

R1
33kΩ

D1

L1

+GATE
FROM 3B4

GATE
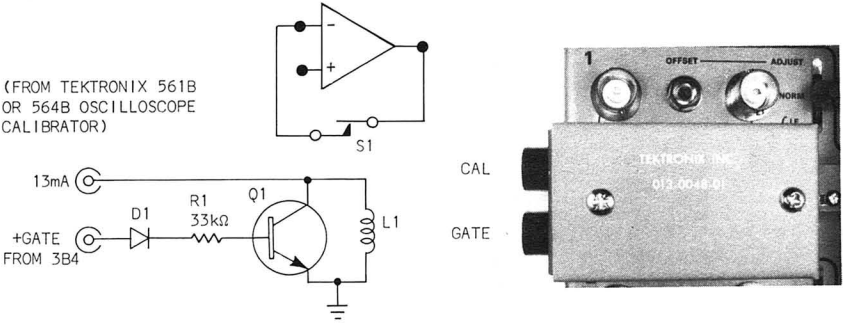
CKT NO	DESCRIPTION	TEKTRONIX PN
D1	DIODE - SILICON	152-0185-00
L1	REED DRIVE COIL	108-0442-00
Q1	2N1893 TRANSISTOR	151-0096-00
R1	33kΩ 1/4W RESISTOR	315-0333-00
S1	REED SWITCH	260-0877-00

MECHANICAL
QTY

| 1 | ADAPTER TERMINAL ASSY 013-0048-01 |

PERFORMANCE AS GATED INTEGRATOR - .5ms/div

INTEGRATOR - $Z_i = 0.1M\Omega$, $Z_f = 0.1\mu F$
INTEGRATOR GATED "ON" BY 3B4 FOR 2.3ms
INPUT TO INTEGRATOR 1V DC

3B4
GATING PULSE

10V/cm

INTEGRATOR
OUTPUT

0.1V/cm

"ON" DELAY
0.3ms

"OFF" DELAY
0.3ms
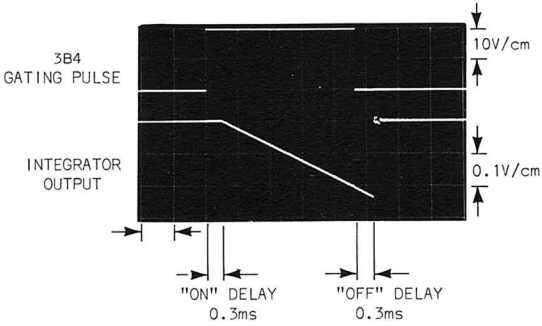
Fig. 29-8. A low-speed gating adapter for the Type 3A8 Operational Amplifier.

29.8 A LOW SPEED GATING ADAPTER FOR THE TYPE 3A8 OPERATIONAL
AMPLIFIER

Refer to Fig. 29-8. This adapter provides a reed
switch between the negative input and output of an
operational amplifier for various gating
applications, particularly for gated integration
applications. The reed drive coil, L1, is powered
from the 13 mA available when the calibrator in a
Tektronix Type 561B or 564B Oscilloscope is operated
in the 40 volt DC position. With no input signal,
this 13 mA flows in L1, causing S1 to close. When
a positive level is applied to the base of Q1, Q1
conducts, diverting the 13 mA to Q1 and thus causing
S1 to open. R1 is chosen as 33 kΩ to allow the
adapter to be powered from the 20 volt plus gate
signal available from the Type 3B4 Time Base unit.
This adapter may be used in conjunction with lower
voltage gating signals by correspondingly reducing
the value of R1. A delay of 0.3 ms is produced
between the application of the gating pulse and the
activation of S1.

INPUT
(VIA 3A8)

→| t |←

NUMBER OF STEPS
VARIES FROM 4
TO >500 WITH
C_f = 0.1µF AND
INPUT PULSES
OF 25 VOLTS

C1
.03µ

R1
5kΩ

D2

D1
T12G

D3

D4

C_f

R2
1MΩ

D7
T12G

C2
.001µ

R3
10kΩ

D6

Q1
2N1308

OUTPUT
(VIA 3A8)

→| t |←

+15V

0V

R5
10kΩ

ERASE PULSE
IF REQUIRED

R4
8.2k

D8
2.2mA

C3
.01µ

SPRING RETURN
ACTION FROM
RESET TO HOLD

RESET

HOLD

SELF RESET

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.03µF 100V CAPACITOR	285-0702-00
C2	.001µF 200V CAPACITOR	283-0067-00
C3	.01µF 200V CAPACITOR	283-0005-00
Cf	SELECTED BY TYPE 3A8 Zf CONTROL (0.1µF)	
D1	DIODE T12G	152-0008-00
D2-D6	5 EACH SI DIODE CD61165	5 EACH 152-0245-00
D7	DIODE T12G	152-0008-00
D8	2.2mA TUNNEL DIODE TD-2	152-0402-00
R1	5kΩ MIN POT	311-0310-00
R2	1MΩ 1/4W RESISTOR	315-0105-00
R3	10kΩ 1/4W RESISTOR	315-0103-00
R4	8.2kΩ 1/4W RESISTOR	315-0822-00
R5	10kΩ 1/4W RESISTOR	315-0103-00
Q1	2N1308 TRANSISTOR	151-0072-00
MECHANICAL QTY		
1	ADAPTER TERMINAL ASSY	013-0048-01
1	3-POSITION SWITCH	260-0614-00
1	KNOB FOR R1	366-0189-00
2	WASHERS FOR R1 AND SWITCH	210-0905-00
2	NUTS FOR R1 AND SWITCH	210-0562-00

STEP OUTPUT WAVEFORM →

OUTPUT VOLTAGE IS
PROPORTIONAL TO R4

EXCELLENT STEP LINEARITY
FOR SWEEP STEPPING OR
SCANNING APPLICATIONS →

ERASE PULSE OUT CAN BE USED TO
ERASE A STORAGE OSCILLOSCOPE WHEN
THE STEP RESETS TO ZERO

Fig. 29-9. A resetting step generator adapter for the Type 3A8
Operational Amplifier.

29.9 A RESETTING STEP GENERATOR ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-9. This adapter produces a stairstep waveform, the stepping commands being received from an external input. This external input may typically be a gating pulse derived from the time base of an oscilloscope. The output stairstep may also be used to erase a storage oscilloscope during step reset.

The circuit shown provides a step ramp of 15 volts peak, with the number of steps continuously variable from 4 to 500 or more. The required step command input is a 20 to 25 volt gate waveform from a source impedance of 5,000 ohms or less. Component selection allows for the use of nearly any step command waveform greater than about 1.5 volts. Linearity appears to be better than ±3%.

A negative charge is applied to D2 from the input, the charge is then transferred to C_f by the operational amplifier to provide a voltage step at the output. The charge is the charge in C1 developed during the "on" time of the input pulse. The charge is transferred to C_f during the off time of the input pulse. The amount of voltage developed across C_f for each step is dependent on the input pulse amplitude, the setting of R1 and the ratio $C1/C_f$.

The positive going stairstep at the output provides both collector voltage for Q1 and current for the tunnel diode D8 which is normally in the low voltage state. When the stairstep reaches the amplitude at which the current through R4 and D8 reaches 2.2 mA, the tunnel diode switches to its high voltage state, turning on Q1. The collector of Q1 snaps sharply negative. This negative going step is coupled to the +grid of the operational amplifier via C2 and D5, causing the output to fall negative. The negative transition at the output is also coupled via C2 and D5 to the positive grid, completing a regenerative loop and forcing the output to continue falling even when D8 has reverted to its low voltage state and Q1 is turned off. D8 and Q1, then, only trigger the reset; the "work" is done by the operational amplifier.

D3 prevents the falling output from driving the
-grid negative and cancelling the regenerative action;
D4 catches the output when it reaches ground
potential.   D6 prevents the collector of Q1 from
being driven negative.

The negative transition drives the junction of C2
and R3 to approximately -15 volts, it then recovers
to ground potential via the C2-R2 time constant.
Until the +grid recovers to 0 volts and D5
disconnects, further input pulses to the -grid are
shunted to ground via D3 and do not charge the
stairstep.   This provides a natural holdoff action.
The circuit comes out of holdoff with a sharp
regenerated step to 0 V.

29.10   A SELF-CONTAINED RESETTING STAIRSTEP GENERATOR

Refer to Fig. 29-10.   This stairstep generator
provides a 16 step ramp of 1 volt peak, the step
command input requiring a negative going transition
of greater than 2 volts in amplitude to initiate the
stepping action.   This input may typically be the
gating pulse derived from the time base of an
oscilloscope.   The output stairstep may also be
used to erase a storage oscilloscope during step
reset or to erase one half of the screen of a split
screen oscilloscope after 8 steps have been
completed.

U1 consists of 4 toggling flip-flops with the "1"
output of each flip-flop driving the clock input of
the adjacent flip-flop.   R1, R2, R3, R4 and R5 form
a digital to analog converter to allow the digital
levels produced by the 4 flip-flops to be transformed
to analog steps.   R5 may be reduced in value to
provide less ramp output.   A decrease in R5 will
also improve the linearity of the circuit.   The
linearity of the circuit may also be improved by
selecting U1 for optimum linearity.
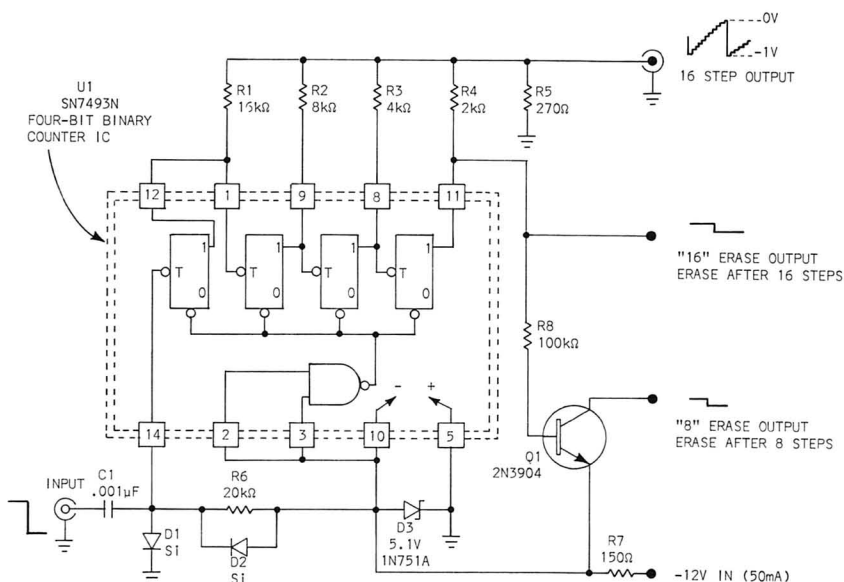
STEP GENERATOR WILL STEP ONCE FOR EACH
NEGATIVE TRANSITION. MINIMUM AMPLITUDE 2V.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.001µF 500V CAPACITOR	283-0078-00
D1	SILICON DIODE	152-0185-00
D2	SILICON DIODE	152-0185-00
D3	1N751A ZENER DIODE	152-0279-00
Q1	2N3904 TRANSISTOR	151-0190-00
R1	16.2kΩ ±1% RESISTOR	321-0309-00
R2	7.96kΩ ±1% RESISTOR	321-0638-00
R3	4.02kΩ ±1% RESISTOR	321-0251-00
R4	2.0kΩ ±1% RESISTOR	321-0222-00
R5	270Ω ±5% RESISTOR	315-0271-00
R6	20kΩ ±5% RESISTOR	315-0203-00
R7	150Ω 1/2W RESISTOR	301-0151-00
R8	100kΩ ±5% RESISTOR	315-0104-00
U1	INTEGRATED CKT T1	
	SN7493N	156-0032-00
MECHANICAL		
QTY		
1	BOX - TERMINAL	202-0054-00
1	CONNECTOR - BNC MALE	131-0428-00
1	CONNECTOR - BNC FEMALE	131-0106-00
3	CONNECTOR - TERMINAL JACK	131-0251-00
1	SOCKET - INTEGRATED CKT	136-0269-00
1	COVER - BOTTOM	200-0252-00
2	SETSCREW 4-40	213-0048-00
2	SCREW, THREAD-FORMING	213-0141-00

WITH R5 AT 270Ω THE OUTPUT
IS ≈ 1V BUT IS SLIGHTLY
NONLINEAR.

WITH R5 AT 27Ω (TEKTRONIX
PN 315-0270-00) THE OUTPUT
IS REDUCED TO ≈ 0.1V, HOW-
EVER, LINEARITY IS IMPROVED.

R8, Q1 AND "8" ERASE OUTPUT
ONLY REQUIRED WITH SPLIT-
SCREEN STORAGE OSCILLO-
SCOPES SUCH AS TYPE 564B.
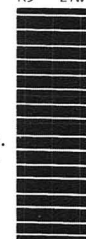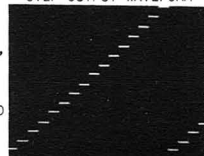
IF ONLY 8 STEPS REQUIRED,
DISCONNECT [1] FROM [12] AND
CONNECT IT TO [14] .

LINEARITY MAY BE IMPROVED
BY SELECTING U1.
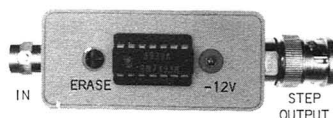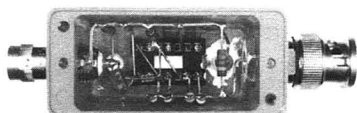
STEP OUTPUT WAVEFORM

Fig. 29-10. A self-contained resetting stairstep generator.

ONE OF THE
EIGHT ELECTRODES
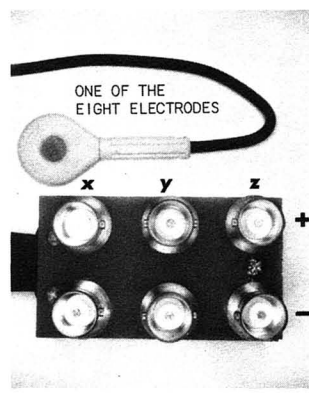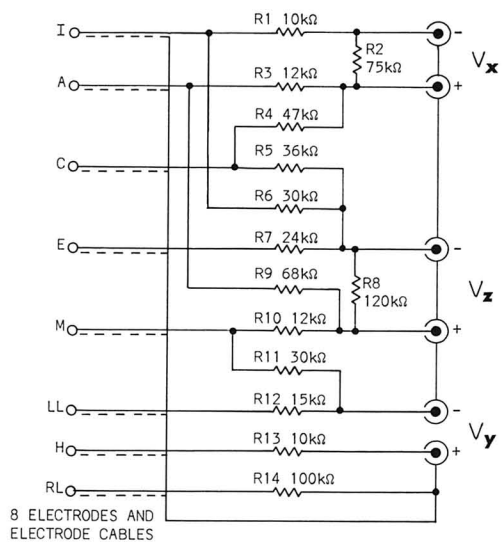
CKT NO	DESCRIPTION	TEKTRONIX PN
R1	10kΩ 1/8W RESISTOR	317-0103-00
R2	75kΩ 1/8W RESISTOR	317-0753-00
R3	12kΩ 1/8W RESISTOR	317-0123-00
R4	47kΩ 1/8W RESISTOR	317-0473-00
R5	36kΩ 1/8W RESISTOR	317-0363-00
R6	30kΩ 1/8W RESISTOR	317-0303-00
R7	24kΩ 1/8W RESISTOR	317-0243-00
R8	120kΩ 1/8W RESISTOR	317-0124-00
R9	68kΩ 1/8W RESISTOR	317-0683-00
R10	12kΩ 1/8W RESISTOR	317-0123-00
R11	30kΩ 1/8W RESISTOR	317-0303-00
R12	15kΩ 1/8W RESISTOR	317-0153-00
R13	10kΩ 1/8W RESISTOR	317-0103-00
R14	100kΩ 1/8W RESISTOR	317-0104-00

MECHANICAL
QTY

1	ELECTRODE WITH 4ft CABLE	012-0121-21
A	ELECTRODE WITH 4ft CABLE	012-0121-21
C	ELECTRODE WITH 4ft CABLE	012-0121-21
E	ELECTRODE WITH 4ft CABLE	012-0121-21
M	ELECTRODE WITH 4ft CABLE	012-0121-21
LL	ELECTRODE WITH 6ft CABLE	012-0121-22
H	ELECTRODE WITH 4ft CABLE	012-0121-21
RL	ELECTRODE WITH 6ft CABLE	012-0121-25
6 ea	BNC CONNECTORS	131-0352-01
1 ea	ADAPTER-TERMINAL ASSY	013-0048-01*
1 ea	1/2 INCH GROMMET	348-0005-00

*CONSTRUCTION NOTE:

- REMOVE 6 BANANA PLUGS FROM TERMINAL ASSY
- ENLARGE VACATED HOLES TO 3/8 INCH
- EPOXY 6 BNC CONNECTORS INTO HOLES
- REMOVE 2 BANANA SOCKETS FROM TERMINAL ASSY
- ENLARGE AND JOIN VACATED HOLES FOR 1/2 INCH GROMMET
- REMOVE PLUGS FROM THE ENDS OF THE 8 ELECTRODES

Fig. 29-11. A "Frank" network for vectorcardiographic use.

29.11 A FRANK NETWORK FOR VECTORCARDIOGRAPHIC USE

Refer to Fig. 29-11. This network provides the
necessary electrode signal interconnections and
attenuations for vectorcardiographic use using
the Frank system. Eight silver/silver-chloride
electrodes are applied to the subject and the three
orthogonal vectorcardiographic signals, x, y, and z,
are available as differential signals via three pairs
of BNC connectors.

29.12   A CURRENT LIMITING ADAPTER FOR PROTECTION FROM ELECTRIC SHOCK

Refer to Fig. 29-12. This adapter limits current between the input and output or between the output and input to less than 300 µA peak, thus allowing conventional electronic instrumentation to be used in conjunction with human subjects.

Current limiting is achieved by field effect diodes D1 and D2 which limit current in either direction to 300 µA within the voltage rating of the diodes (100 volts). Neons B1 and B2 ensure that these diodes are not subjected to voltages that exceed their voltage rating. R1 and R2 provide some current limiting to protect B1 and B2 should high-amplitude, high-power-capability signals be applied to either the input or to the output. Potentiometer R3 allows the dynamic impedance of two such adapters to be balanced to achieve optimum common mode rejection ratio in a differential amplifier.
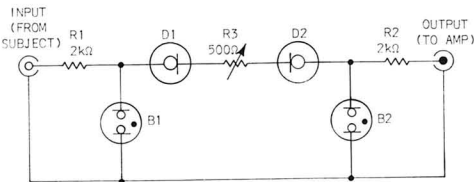
NOTE: TWO OF THESE ADAPTERS ARE REQUIRED FOR USE WITH DIFFERENTIAL AMPLIFIERS.

BANDWIDTH - DC to 500kHz
MAXIMUM SIGNAL INPUT - 50V PEAK
ADJUST R3 FOR COMMON MODE BALANCE WHEN USING 2 ADAPTERS WITH A DIFFERENTIAL AMPLIFIER. CMRR OF 3A9 WITH 2 ADAPTERS IS 100,000:1.

CKT NO	DESCRIPTION	TEKTRONIX PN
B1	NEON BULB 5AH-B	150-0067-00
B2	NEON BULB 5AH-B	150-0067-00
D1	FET DIODE	152-0328-00
D2	FET DIODE	152-0328-00
R1	RESISTOR 2kΩ 1/2W	323-0222-00
R2	RESISTOR 2kΩ 1/2W	323-0222-00
R3	POTENTIOMETER 500Ω	311-0634-00
--	GROMMET FOR R3	348-0003-00
--	ACCESSORY HOUSING	011-0081-00
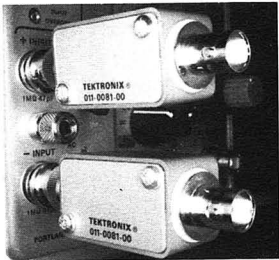
TWO ADAPTERS IN USE WITH A TEKTRONIX TYPE 3A9 DIFFERENTIAL AMPLIFIER

LIMITS CURRENT FLOW TO THE SUBJECT FROM THE AMPLIFIER TO LESS THEN 300µA

Fig. 29-12. A current-limiting adapter for protection from electric shock.

CLOCKWISE

INPUT

R_i = 1MΩ
MAX INPUT 50V

R1
330Ω

R2
5kΩ

C1
.01µF

C2
.0033µF

OUTPUT

TO OSCILLOSCOPE
OR AMPLIFIER
WITH R_i = 1MΩ

C_i < .001µF

S1

OPENS AT FULL
CLOCKWISE ROTATION
OF R2
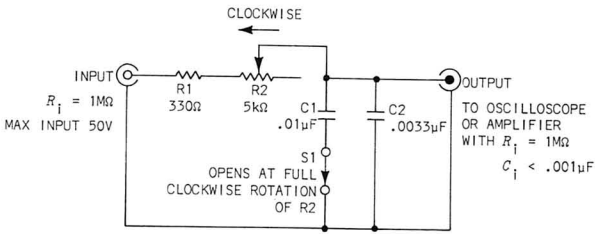
R2 VARIES 3dB CUTOFF FREQUENCY BETWEEN
2.3kHz AND 35kHz. WITH R2 IN FULL
CLOCKWISE POSITION, S1 IS OPEN AND 3dB
CUTOFF IS ABOVE 100kHz.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.01µF 50V ±10% CAPACITOR	283-0155-00
C2	.0033µF 100V ±5% CAPACITOR	283-0051-00
R1	330Ω 1/4W ±5% RESISTOR	315-0331-00
R2	5kΩ POTENTIOMETER	311-0656-00
S1	SWITCH ON R2	--
MECHANICAL QTY		
1	ACCESSORY HOUSING	011-0081-00
1	KNOB – FOR R2	366-0189-00
1	NUT – FOR R2	210-0583-00
1	WASHER-FLAT – FOR R2	210-0905-00
1	WASHER-LOCKING – FOR R2	210-0046-00
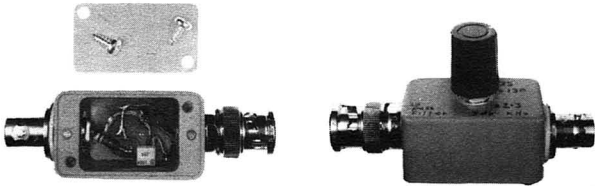
Fig. 29-13. A low-pass filter for physiological signal processing.

29.13 A LO-PASS FILTER FOR PHYSIOLOGICAL SIGNAL PROCESSING

Refer to Fig. 29-13. This filter consists of a
single RC network to allow the frequency response
of instrumentation to be limited in order to
correspondingly limit noise. When used in
conjunction with a conventional oscilloscope vertical
amplifier, and with R2 fully clockwise, the input
impedance is 1 MΩ in parallel with .0033 µF providing
a frequency response in excess of 100 kHz. As R2 is
rotated counterclockwise, the frequency response is
limited between 35 kHz and 2.3 kHz depending on the
position of R2.

INPUT
40V, 1kHz
SQUAREWAVE
FROM TEKTRONIX
561B OR 564B
OSCILLOSCOPE

D1
R1 33kΩ
Q1
R2 1kΩ
R4 3.3kΩ
C1 5.6μ
D2
D3
R3 220Ω
C2 .056μ
D4
R5 18Ω
C3 1μ
OUTPUT

20mV/DIV
0.2s/DIV

E_o = 50mV
t_r = 0.1ms
t_f = 0.2ms
Z_o = 18Ω/1μF
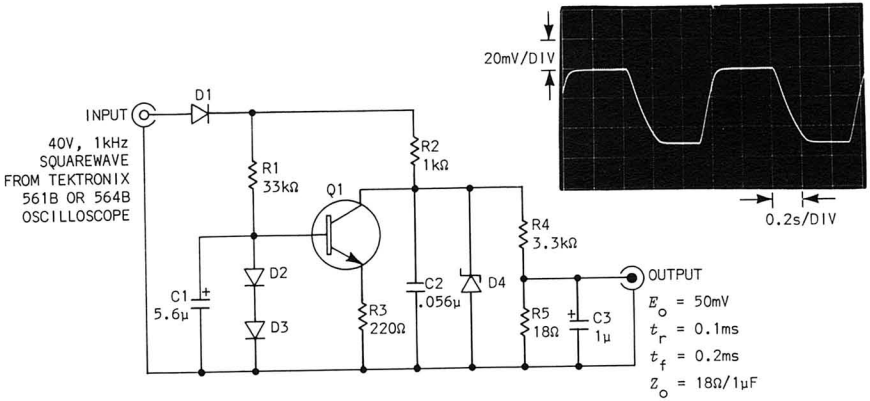
CKT NO	DESCRIPTION	TEKTRONIX PN
C1	5.6μF TANTALUM CAPACITOR	290-0247-00
C2	.056μF ±5% CAPACITOR	285-0684-00
C3	1μF ±10% TANTALUM CAPACITOR	290-0183-00
D1	SILICON DIODE	152-0185-00
D2	SILICON DIODE	152-0185-00
D3	SILICON DIODE	152-0185-00
D4	10V ±5% ZENER DIODE	152-0149-00
Q1	2N3904 TRANSISTOR	151-0190-01
R1	33kΩ 5% 1/4W RESISTOR	315-0333-00
R2	1kΩ 5% 1/4W RESISTOR	315-0102-00
R3	220Ω 5% 1/4W RESISTOR	315-0221-00
R4	3.3kΩ 5% 1/4W RESISTOR	315-0332-00
R5	18Ω 5% 1/4W RESISTOR	315-0180-00

MECHANICAL QTY		
1	BOX - TERMINAL	202-0054-00
1	CONNECTOR - BNC MALE	131-0428-00
1	CONNECTOR - BNC FEMALE	131-0106-00
1	COVER - BOTTOM	200-0252-00
2	SETSCREW, 4-40	213-0048-00
2	SCREW, THREAD-FORMING	213-0141-00

INPUT   OUTPUT
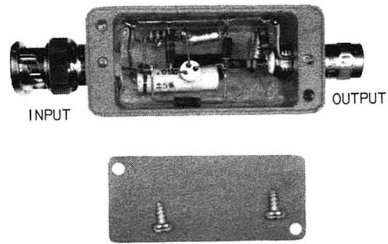
Fig. 29-14. A pulse-shaping circuit simulating the "action potential."

29.14 A PULSE SHAPING CIRCUIT SIMULATING THE ACTION POTENTIAL

Refer to Fig. 29-14. This device provides pulse shaping on the 1 kHz calibrating squarewave produced by Tektronix 561B or 564B Oscilloscopes. The output amplitude is 50 mV, the output risetime 0.1 ms and the output falltime 0.2 ms. The energy of the output pulse is almost all contained in the frequency spectrum from DC to 10 kHz.

When the signal produced by the oscilloscope calibrator is at 40 volts, C2 is charged in a pseudo-constant current mode via R2, providing a risetime of 0.1 ms. When the charge on C2 reaches 10 volts, D4 conducts limiting further voltage increase. This 10 volt level is attenuated by R4 and R5 to provide a 50 mV output. When the signal produced by the oscilloscope calibrator is at zero volts, a charge is maintained on C1, allowing Q1 to conduct in a pseudo-constant current mode to discharge C2 at a constant rate, providing a falltime of 0.2 ms.

29.15 A CONSTANT CURRENT PULSE SOURCE FOR GSR AND OTHER USES

Refer to Fig. 29-15. This pulse generator requires a 40 volt DC power supply which may be obtained from a Tektronix 561B or 564B Oscilloscope calibrator. Shockley diode D2 is functioning as an oscillator, the frequency of oscillation being determined by R1 and C1. The oscillations produced at the anode of D2 turn Q1 on and off via D1 and R2, thus providing a squarewave at the collector of Q1 which is attenuated via R3 and R4. A constant current is produced by this attenuated squarewave via R5.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	4.7µF TANTALUM CAPACITOR	290-0187-00
D1	33V ZENER DIODE	152-0241-00
D2	SHOCKLEY DIODE	152-0136-00
Q1	2N4250 TRANSISTOR	151-0219-00
R1	100kΩ 1/4W RESISTOR	315-0104-00
R2	22kΩ 1/4W RESISTOR	315-0223-00
R3	100kΩ 1/4W RESISTOR	315-0104-00
R4	470kΩ 1/4W RESISTOR	315-0474-00
R5	2.2MΩ 1/4W RESISTOR	315-0225-00
--	ACCESSORY HOUSING	011-0081-00
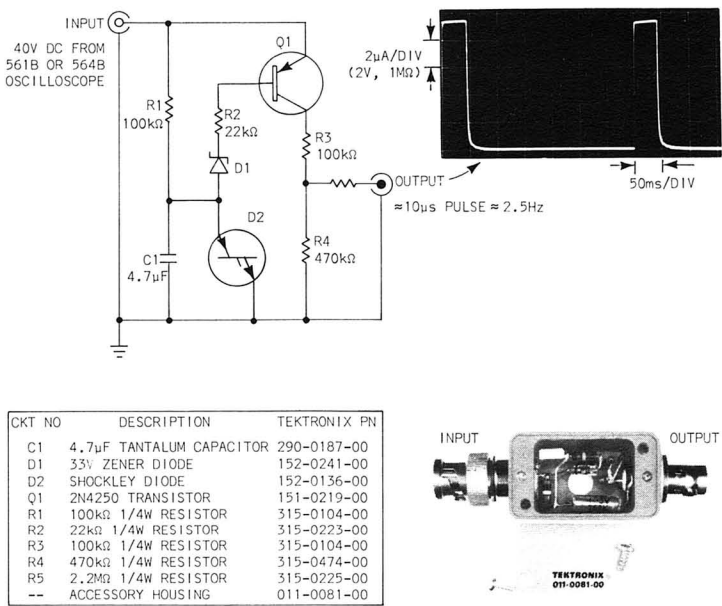
Fig. 29-15. A constant-current, pulse source for GSR and other uses.

30

DEFINITIONS

This chapter provides definitions for the
biomedical terminology used throughout this book.
These definitions have been taken from either
Webster's International Dictionary (Second Edition
Unabridged 1959), *Webster's Seventh New Collegiate
Dictionary* (1967) or *Dorland's Illustrated Medical
Dictionary* (24th Edition, 1965). The appropriate
reference is given at the end of each definition.
The Websters' are published by G. & C. Merriam
Company, Springfield, Massachusetts, USA and the
Dorland's is published by W.B. Saunders Company,
Philadelphia, Pennsylvania, USA and London, U.K.

A

accretion - The process of growth or enlargement.
(Webster)

alveolus - An air cell of the lungs. (Webster)

amnion - A thin membrane forming a closed sac
surrounding the embryos of reptiles, birds and
mammals. It contains a thin, watery fluid, the
amniotic fluid, in which the embryo is immersed.
(Webster)

amniotic - See amnion.

angstrom - Abbreviated Å. One ten-billionth of a
meter. 1 Å = 10^{-8} cm. (Webster)

anterior - Situated before or toward the front.
(Webster)

aorta - The great trunk artery that carries blood
from the heart to be distributed by branch arteries
through the body. (Webster)

aortic - Pertaining to the aorta. (Webster)

arborizations - Formation of or into a form resembling a tree in properties, growth, structure, or appearance, or such a form and arrangement. (Webster)

arrhythmia - An alteration in rhythm of the heartbeat either in time or force. (Webster)

arteriole - One of the small terminal twigs of an artery that ends in capillaries. (Webster)

artifacts - Any artificial product; any structure or feature that is not natural. (Dorland)

atria - Plural of atrium.

atrio-ventricular - Located between an atrium and ventricle of the heart. (Webster)

atrium - An anatomical cavity or passage; especially the main chamber of an auricle of the heart or the entire auricle. (Webster)

auricle - The chamber or either of the chambers of the heart that receives blood from the veins and forces it into the ventricle or ventricles. (Webster)

autonomic - Acting independently of volition; relating to, affecting, or controlled by the autonomic nervous system. (Webster)

axon - A usually long and single nerve-cell process that, as a rule, conducts impulses away from the cell body. (Webster)

B

bioelectric - See bioelectricity.

bioelectricity - The electrical phenomena which appear in living tissues. (Dorland)

biophysical - Pertaining to the branch of knowledge concerned with the application of physical principles and methods to biological problems. (Webster)

brachial - Relating to the arm or a comparable process. (Webster)

bradycardia - A slow heart rate. (Webster)

bronchi - Plural of bronchus.

bronchus - Either of two primary divisions of the trachea that lead respectively into the right and the left lung; broadly - bronchial tube. (Webster)

bundle of His - A small band of cardiac muscle fibers transmitting the wave of depolarization from the auricles to the ventricles during cardiac contraction.

C   *capillaries* - Any of the smallest vessels of the blood-vascular system connecting arterioles with venules and forming networks throughout the body. (Webster)

cardiac - Pertaining to the heart. (Dorland)

cardiology - The study of the heart and its action and diseases. (Webster)

cardiovascular - Relating to the heart and blood vessels. (Webster)

catheter - A tubular medical device for inserting into canals, vessels, passageways, or body cavities usually to permit injection or withdrawal of fluids or to keep a passage open. (Webster)

cell - A small usually microscopic mass of protoplasm bounded externally by a semipermeable membrane, usually including one or more nuclei and various nonliving products, capable alone or interacting with other cells of performing all the fundamental functions of life, and forming the least structural aggregate of living matter capable of functioning as an independent unit. (Webster)

cephalic - Directed towards or situated on or in or near the head. (Webster)

cerebellum - A large dorsally projecting part of the brain, especially concerned with the coordination of muscles and the maintenance of bodily equilibrium. (Webster)

cerebrum - An enlarged anterior or upper part of the brain. (Webster)

chronaxy - The time required for the excitation of a nervous element by a definite stimulus; the minimum time at which a current just double the rheobase will excite contraction. (Dorland)

cochlea - A division of the labyrinth of the ear of higher vertebrates that is usually coiled like a snail shell and is the seat of the hearing organ. (Webster)

cornea - The transparent part of the coat of the eyeball that covers the iris and pupil and admits light to the interior. (Webster)

cortex - The outer or superficial part of an organ or body structure; especially the outer layer of gray matter of the cerebrum and cerebellum. (Webster)

cortical - Of, relating to, or consisting of the cortex. (Webster)

cranium - The part that encloses the brain. (Webster)

curare - A dried aqueous extract especially of a vine used in medicine to produce muscular relaxation. (Webster)

cytoplasm - The protoplasm of a cell exclusive of that of the nucleus. (Dorland)

D   *defibrillation* - The stoppage of fibrillation of the heart. (Dorland)

defibrillator - An apparatus used to counteract fibrillation (very rapid irregular contractions of the muscle fibers of the heart) by application of electric impulses to the heart. (Dorland)

dendrite - Any of the usual branching protoplasmic processes that conduct impulses toward the body of a nerve cell. (Webster)

depolarize - To cause to become partially or wholly unpolarized. (Webster)

diastole - A rhythmically recurrent expansion especially the dilatation of the cavities of the heart during which they fill with blood. (Webster)

diastolic - See diastole.

dicrotic - Having a double beat; being or relating to the second expansion of the artery that occurs during the diastole of the heart. (Webster)

dorsal - Relating to, situated near or on the back. Especially of an animal or of one of its parts. (Webster)

E

ECG - Abbreviation for electrocardiogram. (Dorland)

ectopic - Located away from normal position. (Dorland)

EEG - Abbreviation for electroencephalogram. (Dorland)

electrocardiogram - The tracing made by an electrocardiograph. (Webster)

electrocardiograph - See electrocardiography.

electrocardiography - The making of graphic records of the electric currents emanating from heart muscle, as a method for studying the action of the heart muscle. (Dorland)

electrode - A conductor used to establish electrical current with a nonmetallic part of a circuit. (Webster)

electrodermal - See electrodermography.

electrodermography - The recording of the electrical resistance of the skin, which varies with the amount of sweating, and constitutes a sensitive index to the activity of the autonomic nervous system. (Dorland)

electroencephalogram - The tracing of brain waves made by an electroencephalograph. (Webster)

electroencephalograph - See electroencephalography.

electroencephalography - Recording of electric
currents developed in the brain by means of
electrodes applied to the scalp, to the surface
of the brain, or placed within the substance of
the brain. (Dorland)

electrogastrogram - The graphic record obtained by
the synchronous recording of the electrical and
mechanical activity of the stomach. (Dorland)

electrolyte - A nonmetallic electric conductor in
which current is carried by the movement of ions.
(Webster)

electromyogram - The tracing of muscular action
potentials by an electromyograph.

electromyograph - See electromyography.

electromyography - The recording of the changes in
electric potential of muscle. (Dorland)

electrophysiology - The science of physiology in its
relations to electricity; the study of the
electric reactions of the body in health.
(Dorland)

embolus - An abnormal particle (as an air bubble)
circulating in the blood. (Webster)

embryo - A human or animal offspring prior to
emergence from the womb or egg; hence, a beginning
or undeveloped stage of anything. (Webster)

EMG - Abbreviation for electromyography. (Dorland)

epilepsy - Any of various disorders marked by
disturbed electrical rhythms of the central nervous
system and typically manifested by convulsive
attacks usually with clouding or consciousness.
(Webster)

extracellular - Situated or occurring outside a cell
or the cells of the body. (Webster)

extracorporeal - Situated or occurring outside the
body. (Dorland)

F

fibula - The outer and smaller of the two bones
of the leg. (Dorland)

fibular - Pertaining to the fibula. (Dorland)

fluoroscopy - Process of using an instrument to observe the internal structure of an opaque object (as the living body) by means of X-rays. (Webster)

frontal plane - Section drawing, etc. parallel to the main axis of the body, and at right angles to the sagittal plane. (Webster)

G

galvanic - Of, relating to, or producing a direct current of electricity. (Webster)

H

hemisphere - Half of any spherical or roughly spherical structure or organ, as demarcated by dividing it into approximately equal portions. (Dorland)

homogeneity - The quality or state of being homogeneous. (Webster)

homogeneous - Of uniform structure or composition throughout. (Webster)

I

infarct - An area of necrosis in a tissue or organ resulting from obstruction of the local circulation by a thrombus or embolus. (Webster)

infarction - See infarct.

inhomogeneities - See inhomogeneity.

inhomogeneity - Something which is not homogeneous. (Webster)

intracellular - Being or occurring within a protoplasmic cell. (Webster)

ion - An atom or group of atoms that carries a positive or negative electric charge as a result of having lost or gained one or more electrons. (Webster)

iris - The circular pigmented membrane behind the cornea of the eye. (Dorland)

iso-electric - Uniformly electric throughout, or having the same electric potential, and hence giving off no current. (Dorland)

isothermal - See isothermic.

isothermic - Having the same temperature. (Dorland)

isotropic - Exhibiting properties with the same
    values when measured along axes in all directions.
    (Webster)

isotropy - See isotropic.

L

latency - A state of seeming inactivity, such as
    that occurring between the instant of stimulation
    and the beginning of response. (Dorland)

lobe - A somewhat rounded projection of division of
    a bodily organ or part. (Webster)

lumen - (1) the cavity of a tubular organ (the
    lumen of a blood vessel). (2) the bore of a
    tube (as of an organ). (Webster)

M

manometer - An instrument for measuring the pressure
    of gases and vapors: pressure gauge. (Webster)

membrane - A thin layer of tissue which covers a
    surface or divides a space or organ. (Dorland)

metabolism - The sum of all the physical and chemical
    processes by which living organized substance is
    produced and maintained. (Dorland)

micron - A unit of length to one thousandth of a
    millimeter. $1\mu = 10^{-3}$ mm $= 10^{-6}$ meters. (Webster)

mitochondria - Small granules or rod-shaped
    structures found in differential staining in the
    cytoplasm of cells. (Dorland)

mitral stenosis - A narrowing of the left atrio-
    ventricular orifice. (Dorland)

myocardial - See myocardium.

myocardium - The middle muscular layer of the heart
    wall. (Webster)

myograph - An apparatus for recording the effects of
    a muscular contraction. (Dorland)

myographic - See myograph.

N

necrosis - Death of tissue, usually as individual cells, groups of cells, or in small localized areas. (Dorland)

neuron - A nerve cell with its processes, collaterals, and terminations regarded as a structural unit of the nervous system. (Dorland)

neuronal - See neuron.

nomography - A graphic method by which the relation between any number of variables may be represented graphically on a plane surface, such as a piece of paper. (Dorland)

nuclei - Plural of nucleus. (Dorland)

nucleus - A central point, group, or mass about which gathering, concentration or accretion takes place -- essential portion of a cell -- group of nerve cells in the central nervous system. (Webster)

O

occipital - See occiput.

occiput - Of or relating to the back part of the head or skull. (Dorland)

organ - A differentiated structure consisting of cells and tissues and performing some specific function. (Webster)

orthogonal - At right angles to.

orthogonality - See orthogonal.

P

parietal - Of, relating to, or forming the upper posterior wall of the head. (Webster)

permeable - See permeate.

permeate - To pass through the pores or interstices. (Webster)

peroneal - Pertaining to the fibula or to the outer side of the leg. (Dorland)

piezoelectric - See piezoelectricity.

piezoelectricity - Electricity or electric polarity
due to pressure especially in a crystalline
substance. (Webster)

plethysmography - The recording of the changes in
the size of a part as modified by the circulation
of the blood in it. (Dorland)

plastid - Any specialized organ of the cell other
than the nucleus. (Dorland)

pneumatic - Relating to, or using air, wind, or
other gas (a) moved or worked by air pressure
(b) adapted for holding or inflated with
compressed air. (Webster)

pneumograph - An instrument for recording the
thoracic movements or volume change during
respiration. (Webster)

pneumotachygraph - An instrument for recording the
velocity of the respired air. (Dorland)

posterior - The hinder parts of the body. (Webster)

protoplasm - The colloidal complex of protein, other
organic and inorganic substances, and water that
constitutes the living nucleus, cytoplasm, plastids,
and mitochondria of the cell and is regarded as
the only form of matter in which the vital
phenomena are manifested. (Webster)

protoplasmic - See protoplasm.

psychogalvanic - See psychogalvanometer.

psychogalvanometer - A galvanometer for recording
the electrical agitations produced by emotional
stresses. (Dorland)

pulmonary - Relating to, functioning like, or
associated with the lungs. (Webster)

pupil - The contractile aperture in the iris of
the eye. (Webster)

R

radical – A group of atoms that is replaceable by a single atom, that is capable of remaining unchanged during a series of reactions, or that may show a definite transitory existence in the course of a reaction. (Webster)

radioisotope – An isotope which is radioactive, produced artificially from the element or from a stable isotope of the element by the action of neutrons, protons, deuterons, or alpha particles in the chain-reacting pile or in the cyclotron. Radioisotopes are used as tracers or indicators by being added to the stable compound under observation, so that the course of the latter in the body (human or animal) can be detected and followed by the radioactivity thus added to it. The stable element so treated is said to be "labeled" or "tagged." (Dorland)

real-time spectrum analyzer – A spectrum analyzer that performs a continuous analysis of the incoming signal with the time sequence of events preserved between input and output.

retina – The sensory membrane of the eye that receives the image formed by the lens, is the immediate instrument of vision and is connected with the brain by the optic nerve. (Webster)

rheobase – The minimum potential of electric current necessary to produce stimulation. (Dorland)

S

sagittal – Of, relating to, or situated in the median plane of the body or any plane parallel thereto. (Webster)

scalp – That part of the integument of the head which normally is covered with hair. (Dorland)

selenide – A compound of selenium with an element or radical. (Webster)

selenium – A nonmetallic element relating to sulphur and tellurium and resembling them chemically. (Webster)

semipermeable - Partially but not freely or wholly permeable (Webster)

sinoatrial node - A microscopic collection of atypical cardiac muscle fibers which is responsible for initiating each cycle of cardiac contraction.

sinus - A cavity in the substance of the bone of the skull that usually communicates with the nostrils and contains air. (Webster) Also, any irregular cavity, particularly in the circulatory system.

spatial - Pertaining to space. (Dorland)

sphygmomanometer - An instrument for measuring blood pressure and especially arterial blood pressure. (Webster)

spirometer - An instrument for measuring the air entering and leaving the lungs. (Webster)

stereotaxic - Pertaining to or characterized by precise positioning in space. (Dorland)

stimulate - To excite to functional activity. (Dorland)

stroboscope - An instrument for determining speeds of rotation or frequencies of vibration made in the form of a rapidly flashing light that illuminates an object intermittently. (Webster)

stroboscopic - Of, utilizing, or relating to a stroboscope. (Webster)

synapse - The point at which a nervous impulse passes from one neuron to another. (Webster)

synaptic - See synapse.

systemic - Pertaining to or affecting the body as a whole. (Dorland)

systole - The contraction, or period of contraction, of the heart, especially that of the ventricules. It coincides with the interval between the first and second heart sound, during which blood is forced into the aorta and the pulmonary trunk. (Dorland)

systolic - See systole.

T *tachycardia* - Relatively rapid heart action. (Webster)

tetrahedron - A solid body having four faces. (Webster)

thermistor - An electrical resistor made of a material whose resistance varies sharply in a known manner with the temperature. (Webster)

thermocouple - A thermoelectric couple (a union of two conductors -- as bars of dissimilar metals joined at their extremities -- for producing a thermoelectric current) used to measure temperature differences. (Webster)

thermoelectric - Of or relating to phenomena involving relations between the temperature and the electrical condition in a metal or in contacting metals. (Webster)

thoracic - See thorax.

thorax - The part of the body of man and other mammals between the neck and the abdomen. (Webster)

thrombus - A clot of blood formed within a blood vessel and remaining attached to its place of origin. (Webster)

tibia - The inner and usually larger of the two bones of the vertebrate hind limb between the knee and ankle. (Webster)

tibial - See tibia.

tissue - An aggregation of similarly specialized cells united in the performance of a particular function. (Dorland)

torso - The human trunk. (Webster)

trachea - The main trunk of the system of tubes by which air passes to and from the lungs. (Webster)

488

transducer - A device that is actuated by power from one system and supplies power in any other form to a second system. (Webster)

U

ulnar - Pertaining to the inner and larger bone of the forearm, on the side opposite that of the thumb. (Dorland)

utero - (1) A combining form from uterus, (Webster), as *utero-vaginal*.
 (2) The Latin dative of uterus, as *in utero*, in the uterus.

uterus - The hollow muscular organ in female animals which is the abode and the place of nourishment of the embryo and fetus. (Dorland)

V

vasoconstriction - Narrowing of the lumen of blood vessels especially as a result of vasomotor action. (Webster)

vasomotor - Any element or agent that affects the caliber of a blood vessel. (Dorland)

vector - A quantity that has magnitude, direction, and sense and that is commonly represented by a directed line segment whose length represents the magnitude and whose orientation in space represents the direction. (Webster)

ventricle - A chamber of the heart which receives blood from a corresponding atrium and from which blood is forced into the arteries. (Webster)

ventricular - See ventricule.

venule - A small vein; especially one of the minute veins connecting the capillary bed with the larger systemic veins. (Webster)

vertex - The top of the head. (Webster)

viable - Capable of living; especially born alive, with such form and development of organs as to be normally capable of living. (Webster)

viability - See viable.

INDEX

499

REFERENCES TO TEKTRONIX PRODUCTS

BOOKS IN THIS SERIES:

CIRCUIT CONCEPTS

title	part number
Digital Concepts	062-1030-00
Horizontal Amplifier Circuits*	062-1144-00
Oscilloscope Cathode-Ray Tubes*	062-0852-01
Oscilloscope Probe Circuits*	062-1146-00
Oscilloscope Trigger Circuits*	062-1056-00
Power Supply Circuits*	062-0888-01
Sampling Oscilloscope Circuits	062-1172-00
Spectrum Analyzer Circuits	062-1055-00
Storage Cathode-Ray Tubes and Circuits	062-0861-01
Sweep Generator Circuits*	062-1098-01
Television Waveform Processing Circuits	062-0955-00
Vertical Amplifier Circuits*	062-1145-00
*7-book set covering Real-Time Oscilloscopes	062-1180-00

MEASUREMENT CONCEPTS

Automated Testing Systems	062-1106-00
Engine Analysis	062-1074-00
Information Display Concepts	062-1005-00
Probe Measurements	062-1120-00
Semiconductor Devices	062-1009-00
Spectrum Analyzer Measurements	062-1070-00
Television System Measurements	062-1064-00
Time-Domain Reflectometry Measurements	062-1244-00
Transducers Measurements	062-1246-00
Biophysical Measurements	062-1247-00