

TECHNIQUES OF PHYSIOLOGICAL MONITORING

VOLUME III — SYSTEMS

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FOREWORD

This study was initiated by the Biophysics Laboratory of the 6570th Aerospace Medical Research Laboratories, Aerospace Medical Division, Wright-Patterson Air Force Base, Ohio. The research was conducted by the RCA Service Company, Camden, New Jersey, under Contract No. AF 33(657)-9252. Mr. Walter L. Becker was principal investigator for RCA Service Company. J. M. Gibson, 1st Lt., USAF Multienvironment Division, was the contract monitor, and the work was performed in support of Project No. 7222, "Biophysics of Flight," Task No. 722203, "Specialized Instrumentation."

This is the third volume of a three-volume handbook. Volume I, prepared under Contract No. AF 33(616)-7750 and published in September 1962 (AD 288905), describes the basic physiological systems and their measurable parameters. Volume II, prepared under Contract No. AF 33(657)-9252 and published in November 1963 (AD 426816), surveys the components used in physiological monitoring systems, primarily those suitable for aerospace applications.

This third volume was written by Mr. Richard Alnutt and Mr. Walter L. Becker, and edited by Mr. Robert E. Barbieri, all of RCA Service Company. Carl Berkley, Scientific Director of the Foundation for Medical Technology, participated as a technical consultant.

The authors gratefully acknowledge the valuable assistance of Mr. Miles A. McLennan, former contract monitor on this program, who guided RCA's effort in the preparation of these three volumes from their inception in 1961 until his retirement early in 1964.

It should be noted that the citation of specific commercial equipments in portions of this volume reflects their appearance in the primary sources or references used, and such citation does not constitute endorsement by the authors, by RCA Service Company, or by the Air Force.

This technical documentary report has been reviewed and is approved.

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ABSTRACT

This volume is a discussion of monitoring systems. While the applications of physiological monitoring are many and varied, the primary concern here is with viability monitoring, the use of a measurement system to obtain factual, quantitative information about the physiological responses of a subject in a stressful environment, in order to plan protective measures which will ensure the safety and functional capability of that subject in such environments. Included are a description of instrumentation required for the measurement of individual physiological parameters, a discussion of related problems in system design, including simultaneous measurement of several parameters, data transmission or telemetry, and the use of magnetic tape recording as a system adjunct. Basic guidelines of system troubleshooting and interference reduction are also included. Section IV contains a brief survey of additional measurement techniques and data handling considerations which, while not state of the art or standard practice, will undoubtedly affect the field of physiological monitoring in the near future.

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*Section I***INTRODUCTION****PURPOSE AND SCOPE OF HANDBOOK**

This is the third and final volume of a three-volume handbook devoted to the monitoring of a physiological subject in an abnormal environment by electronic techniques. The handbook is concerned primarily with the instrumentation used to determine the vital state of normal subjects in varied environments, principally aerospace.

Volume I is devoted primarily to a discussion of the various physiological systems. It explains briefly the functions which underlie those system parameters or responses which can be measured by electronic techniques. The known responses to abnormal, aerospace environments also were surveyed, and system requirements for electronic measurement were briefly discussed.

Volume II discusses the components used in physiological systems. Performance characteristics and capabilities of electrodes and transducers, signal modifiers, and recording and display devices are covered in some detail, including the use of magnetic tape recorders. The requirements and capabilities of data transmission and data processing equipment also are covered.

This third volume is devoted to a discussion of monitoring systems. Section II, specifically, is devoted to the description of instrumentation required for the measurement of individual physiological parameters. As an introduction to that section, this first section includes a review of the general system design factors touched on in Volume I.

Following the individual system descriptions in Section II, Section III discusses related problems in system design, including (1) simultaneous measurement of several parameters, (2) data transmission or telemetry, and (3) the use of magnetic tape recording as a system adjunct. The basic guidelines of system troubleshooting and interference reduction are also included.

Finally, Section IV contains a brief survey of additional measurement techniques and data handling considerations which while not state of the art or standard practice will undoubtedly affect the field of physiological monitoring in the near future.

THE MEASUREMENT OBJECTIVE

While the applications of physiological monitoring are many and varied, the primary concern in this handbook is with viability monitoring: the use of a measurement

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system to obtain factual, quantitative information about the physiological responses of a subject in a stressful environment, in order to plan protective measures which will ensure the safety and functional capability of that subject in such environments.

Such systems are employed in two ways. One is in basic research, to establish quantitatively the effect of certain environments upon the vital state of a subject. The other is functional monitoring, to obtain a record of such responses during tactical missions. The record may simply be historical, or it may be part of an alarm system to protect the subject in the event of abnormal deviations from the expected conditions.

Such measurement is the province of aerospace medical practice. The types of measurement made have grown out of clinical work, and the instrumentation used has been drawn both from clinical practice and from aerospace telemetry.

There are of course numerous differences between the measurement techniques employed in the clinic and those employed in field or aerospace applications. The measurement data required are not necessarily identical, nor are the environmental factors of the measurement situation. And in functional monitoring the measurement must be accomplished with a minimum of encumbering instrumentation attached to the subject.

In clinical practice, measurement by electronic techniques such as those discussed in this handbook is carried out for a number of purposes, including diagnosis and treatment of patients, and pathological studies. While the instrumentation employed is similar to that in the field, it may be either more or less sophisticated. Some of the obvious differences in these applications are:

- Aerospace monitoring is usually continuous, and may be employed for the full length of a mission, running from several hours to several days. Clinical measurement of many parameters frequently is conducted over test periods of only a few minutes duration.
- Aerospace monitoring may entail exposure of the subject (and the instrumentation) to one or more widely varying environmental stresses, including temperature, atmospheric pressure, g forces, vibration, etc.
- Aerospace monitoring normally must allow for a greater degree of subject movement than prevails in the clinic. The instrumentation that is attached to the subject is doubly limited, then, both in the complexity and kind of devices that can be used and in the type of measurements that can be accomplished successfully under such circumstances.

Finally, because the purpose of aerospace monitoring is frequently routine viability checking, rather than diagnosis, the system requirements for the instrumentation may be less stringent than in the clinic. Both the content of the data (specifically,

frequency content) and the accuracy of the measurement may be reduced from the minimum levels that would be acceptable in clinical measurement.

MEASUREMENT SYSTEM COMPONENTS

The physiological monitoring system comprises a series of components which permit the registration of physiological data through a series of intermediate electrical operations. To accomplish such operations electrically, the system input must consist of a sensor which will provide an electrical analog signal -- either by electrical transduction or with electrode pickup of bioelectric signals. Registration of the electrical analog may be in one of several forms. It may be electrical, by magnetic tape recording, to preserve the measurement for subsequent display or processing. It may be a scalar indication, as on a panel meter, or it may be a digital printout of numerical values. Finally, it may be a graphic presentation, such as on an oscilloscope screen, or a graphic record obtained with a pen writer or an optical galvanometer.

Intermediate components accomplish the amplification, filtering, and other shaping of the input signal and, as required, the transmission of the signal by wire or radio linkage. All of these components are linked together as shown in figure 1. Their capabilities and limitations have been discussed in detail in Volume II of this handbook. The following paragraphs afford a brief review of the nature of these components, leading into the more detailed treatment of their use in the subsequent system discussions.

1. Sensors

To accomplish measurement of physiological phenomena electrically, the physiological variable must first be sensed by a pickup device at the input of the system. The pickup device or sensor provides an electrical analog of the physiological variable, so that subsequent operations in the monitoring system can be accomplished electrically (electronically). This device may be (1) an electrode lead system to pick up bioelectrical signals, or (2) an electrical transducer which converts a physiological parameter into a corresponding electrical output.

Bioelectrical signals are sensed directly with surface electrodes. Other physiological variables are evidenced by minute surface displacements, and can be sensed by various force, pressure, or displacement transducers of suitable sensitivity. These include piezoelectric devices, such as microphones, variable resistance devices such as strain gages, variable reactance devices, and differential transformers. Still other variables can be sensed by photoelectric transducers, where the response is evidenced by changes in tissue opacity, or by electrochemical transducers, where biological gases are being monitored. Finally, temperature-sensitive devices such as thermistors are used to monitor body temperature and respiration.

The sensors are attached directly to the subject with output connections to conditioning circuitry. Excitation voltages for measurement circuits, as well as output leads,

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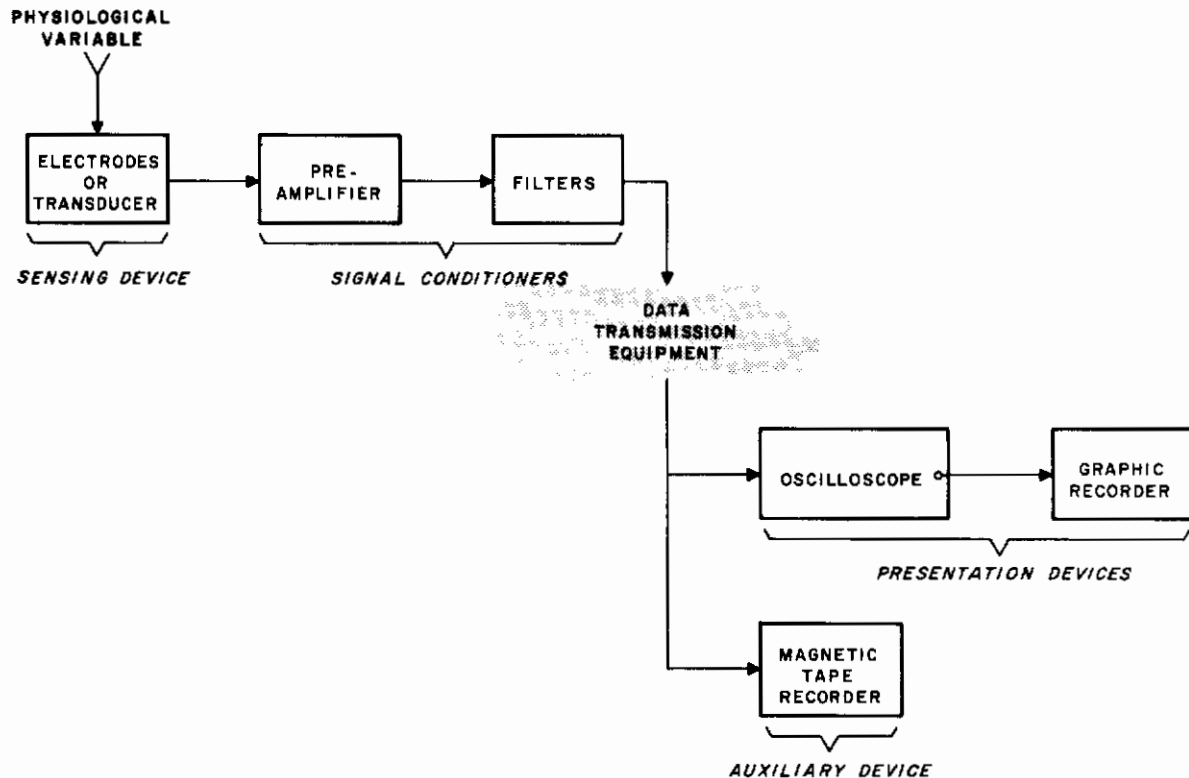


Figure 1. Basic Measuring System Components

also may be connected to the sensors. These sensors must be mounted or attached in such a way as to minimize response to body movement, must be relatively immune to impact and vibration stresses, and must in general operate over the range of environments to which the subject himself will be exposed.

The various sensing devices are described in greater detail in Volume II, Section I. Useful surveys of this critical area are contained in references 25, 32, 43, and 57.

II. Signal Conditioners

Bioelectric potentials, and the voltage outputs of most transducers used in medical electronics, are very low in level, in the microvolt and millivolt range. An immediate requirement then is for sizable amplification, with stages of 500 or 1000 gain factors.

Many signals of interest are slow-varying, near-dc phenomena, and d-c amplification is also required.

In addition, the signal must, in most instances, be modified for a number of reasons. Filters are used to limit the frequency bandwidth to a particular range of interest, or to

MEASUREMENT SYSTEM COMPONENTS

block out certain interfering signals. Filter circuits are employed also to obtain differentiated or integrated outputs. Amplifiers with appropriate impedance characteristics will be employed to match impedances between stages for optimum voltage transfer.

Oscillators are used in many measurements also. They may be employed to provide excitation voltages in certain electrode-type measurements, or they may provide a carrier signal which is modified by the output of a transducer that is included in the oscillator tank circuit. Multivibrator-type circuits are used in some systems to generate pulses, square waves, and ramp functions which in turn are used to control the operation of other system components. Also, since the time constants of these multivibrator circuits and their outputs can be controlled carefully, they make possible the accurate quantization of analog inputs, by measuring signal amplitude as a function of time.

Refer to Volume II, Section II, for more extended treatment of individual modifiers.

III. Presentation Devices

The two basic types of signal presentation, or measurement registration, are scalar and graphic. Digital display and printout are also desirable in many applications. The type of display or recording will vary with the application. Frequently more than one type of device will be used, for multipurpose handling of the measurement.

The most common scalar display device is a moving-coil milliammeter or millivoltmeter. Most special-purpose devices available from medical suppliers, such as GSR meters or cardiometers, include amplifiers, and other signal conditioners in a single package, with a panel meter for on-line (simultaneous) display and monitoring, and an auxiliary output jack for driving a graphic recording device.

Digital display and printout are used in monitoring situations where fast reading by the monitor, in easily read numeric form, is desired. Devices for this purpose accept analog input signals and accomplish the necessary conversion for digital output.

Graphic display is obtained by applying analog input signals to the vertical deflection plates of a cathode-ray tube in an oscilloscope. Permanent records can be made by photographing the face of the cathode-ray tube.

More commonly, when permanent graphic records are desired, oscillographic recorders are employed. For lower frequency (100-150 cps) signals, direct-writing instruments of the galvanometric or potentiometric type are employed. Pens or heated styli trace the input signal variations on scaled chart paper, and movement of the chart paper past the pen at fixed rates provides the time base for the measurements.

When graphic recordings of higher frequencies are needed, oscillographs of the

optical galvanometer type are used. These devices employ low-mass galvanometric elements capable of following frequencies up to several kilocycles. A beam of light deflected by a mirror attached to the moving coil traces the input signal variations of photosensitive paper.

The various recording and display devices are treated in further detail in Volume II, Section III.

IV. Data Transmission Equipment

Data transmission equipment provides the link between the components at the measurement site (sensors and initial signal conditioners) and the control or recording site. The complexity of the data transmission requirement may vary widely. Some measurement situations, over short distances, require no additional components, only direct wire connection between components. Somewhat longer distances may still be covered by wire, but will require either d-c line-driving amplifiers or carrier amplifiers to retain the information on the input signal.

Many aerospace applications will require the use of a radio telemetry link -- an obvious necessity for in-flight monitoring unless on-board recording is all that is desired. Various techniques are used to transmit information at radio frequencies, including amplitude, frequency and pulse modulation of r-f transmitters. The transmission of many channels of information on single radio channels is accomplished by time- and frequency-division multiplexing schemes. All of these techniques have been discussed in detail in Volume II, Section V, and system considerations for physiological telemetry are taken up further in Section III of this volume.

V. Other Components

A. Power Supplies

For the sake of simplicity, power supplies were not indicated on the block diagram of figure 1. With the exception of some electrodes and transducers used to sense the physiological variable, all components of the measuring system require some source of d-c voltages for their operation. These sources may include chemical batteries or photoelectric energy converters, or d-c power supplies operating from a-c power lines. Combinations of battery and power-supply sources are frequently employed.

Many system components contain their own power supplies, each requiring connection to an a-c source. In other systems, several components may draw power from the same supply. Common power supplies are efficient and economical, but they can be a source of system trouble, by permitting unwanted interaction between components through the power supply connections. Refer also to Volume II, pages 136-148.

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B. Magnetic Tape Recorders

Magnetic tape recording provides a means for storing experimental data in the electrical form in which it was originally obtained. Such recording is an intermediate step in the measurement process, and it is used at the same point in a system as the data transmission or telemetry link. (It may, of course, be used in addition to the telemetry link, with magnetic recording taking place either before or after transmission.)

Magnetic tape recording provides compact storage with easy access to the stored information. It permits real-time recording with minimum instrumentation. Subsequent operations, such as graphic recording or display or data processing, can then be accomplished at a later time (and, significantly, as many times as desired) simply by playing back the recording to furnish input signals to the appropriate devices. Thus, while a magnetic tape recorder is not a necessary component for physiological monitoring, it is certainly a desirable one for many applications. Refer to the discussion in Volume II, pages 119-133.

C. Data Processing Equipment

Data processing, while not a part of measurement systems per se, is a related process involving the reduction and interpretation of information obtained with such systems. And the time may come when physiological monitoring and rapid computer processing will serve as integral parts of an overall command system, which could be used, for example to control the operation of a life support system operating in hazardous environments.

A rather extensive introduction to the uses of data processing equipment, including both analog and digital computers, is contained in Volume II, Section VI. Additional data processing considerations are touched on in the final section of this volume.

An ultimate application of this sort will involve on-line computation, whereas most data processing currently done in connection with physiological data does not. In digital computation, for example, analog information from a stored record, such as a chart recording or magnetic tape recording, or tabular numeric information previously extracted from such recordings, must be converted into digital form before it can be read into a computer for processing. There are devices which can perform these operations automatically. It is therefore conceivable that for certain applications, the output of a monitoring system could be applied directly to computer input components, so that computer operations could follow the response of a monitoring system directly.

BASIC SYSTEM REQUIREMENTS

In meeting and solving the problem of physiological measurement, a monitoring

system must satisfy two general conditions: it should have a minimum effect on the phenomenon being measured, and it should respond only to the phenomenon being measured. In terms of specific system parameters, the monitoring system obviously must possess certain response characteristics -- both in terms of frequency and dynamics, or amplitude -- that correspond to the range of variations in the phenomenon being measured. Implicitly, system response, in addition to being adequate in range, must also be reasonably linear, and accurate, if the measurements obtained are to be meaningful analogs of the physiological system under observation. Finally, these criteria of performance must be met by all components of the system, for final system output can be no better than that of any of its components.*

I. Frequency Response

The monitoring system must possess satisfactory response to physiological inputs over the range of frequencies containing the most significant data for the particular measurement. In general, the inputs from the various human body systems are measured and recorded as direct-current variations; the frequency content of these variations ranges from zero to several thousand cycles per second. With a few exceptions (heart sounds, muscle action potentials, etc) significant information on the responses being monitored is contained in the frequencies below 100 cycles per second. Table I summarizes these frequency requirements. These ranges are discussed in detail elsewhere in the handbook, for individual physiological responses.

Generally speaking, the frequency requirements cited above are well within the capability of electrical and electronic instrumentation. Possible problem areas in frequency range and response are discussed below.

A. Transducers

Transducers operate on a physiological input to produce a voltage (or a change in some other electrical characteristic) that is related in value to the input. This input-output relationship is the transfer function of the transducer. Transducers are designed for specific applications so that the transfer function will be constant with frequency (linear) over most of the frequency range of the input.

As can be seen in figure 2, transducer response is dependent upon the relationship of the input frequency to the natural frequency of the transducer. The transducer will have a sharply peaked response (solid lines) at its undamped natural frequency. With an appropriate amount of damping in the transducer system, the

*With certain exceptions. Linear system response, for example, may be obtainable even though one component (say, the transducer) has a nonlinear response, by inserting elsewhere in the system a component (such as an amplifier stage) which possesses an equally nonlinear, but inverse, response.

BASIC SYSTEM REQUIREMENTS

change in the transfer function will be more nearly constant (linear) with changes in the input frequency (dotted lines) (ref. 40).

TABLE I. TYPICAL REQUIREMENTS FOR FREQUENCY RESPONSE IN PHYSIOLOGICAL MONITORING SYSTEMS

Measurement	Spectrum (cps)*
Heart potentials (ECG)	0.1 to 100
Heart potentials (rate measurement)	5 to 25
Heart sounds	16 to 2000
Pressure pulse	0 to 30
Brain potentials	1 to 80
Muscle potentials	10 to 5000
Basal skin resistance	0 to 1
Galvanic skin response	0.01 to 1
Respiration (qualitative)	0.1 to 2
Respiration (true shape)	0 to 10

*Spectrum limits are between the half-power points (3 db down).

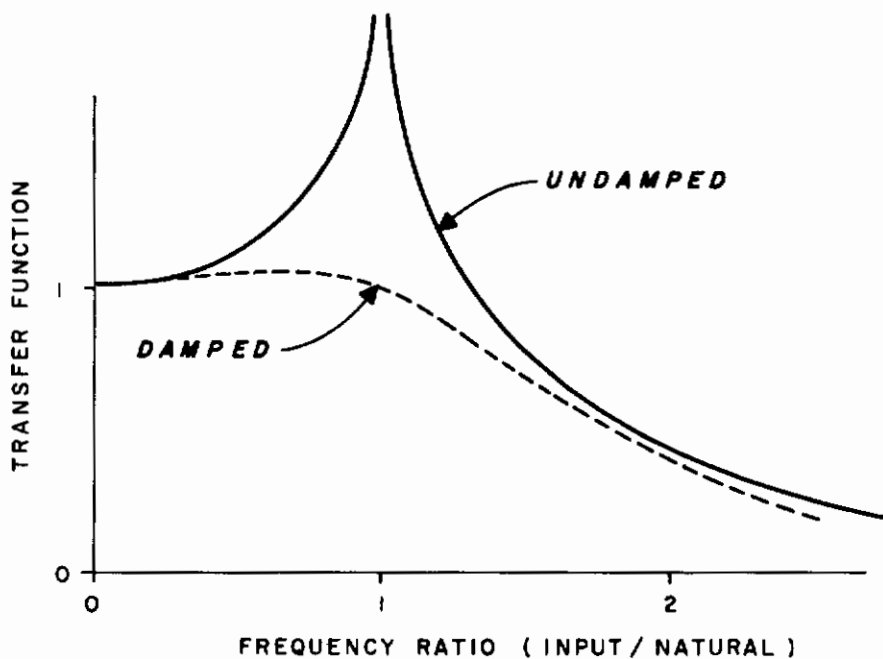


Figure 2. Typical Transducer Frequency Response

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There are instances, of course, where nonlinear transducer response is desirable. An example of this is in the recording of heart sounds, where a microphone with a nonlinear characteristic may be employed to produce a graphic trace which is similar to the logarithmic response of the human ear, as experienced by the clinician accustomed to listening to these sounds directly with a stethoscope.

B. Amplifiers

It is usually possible to design amplifiers and other electronic circuitry which will possess the desirable frequency response characteristics necessary for the measurement at hand. One area of difficulty is in the handling of d-c baselines and slow-varying data signals. It is difficult to design a high-gain d-c amplifier that will be free of baseline drift for extended periods of time, especially within the limits of size, weight, and power consumption that are necessary in aerospace environments.

Chopper amplifiers and carrier amplifiers will provide high gain and more stable operation than direct-coupled amplifiers, but they too possess certain disadvantages. Mechanical choppers may impose a limit upon the high-frequency response of the system. And carrier amplifiers are inherently noisier than the other types, so that there will be some sacrifice in the signal-to-noise ratio.

C. Presentation Devices

Of principal concern in selection of the appropriate presentation device for a given measurement is the high-frequency response desired. Direct-writing instruments -- galvanometric or potentiometric recorders -- have an upper frequency limit of about 100 to 150 cycles per second. For those measurements where higher frequency information is desired, recording must be accomplished by an optical galvanometer, or with an oscilloscopic camera.* For wide-range or high-frequency recording, even the oscilloscope may be unsuitable. This is because the illumination on the face of the cathode-ray tube varies inversely (generally speaking) with the speed of beam travel (sweep and deflection). If both high- and low-frequency phenomena are being displayed, the variations in illumination will limit the effectiveness of photographic recording.

D. Other Frequency Considerations

Certain other frequency criteria must be considered, besides the inherent capability of the measuring system to handle the frequencies of the phenomenon being measured.

*Higher frequency data can also be recorded on direct-writing instruments if the data is first recorded on magnetic tape as an intermediate step. The tape can then be played back at a slower speed to provide a lower frequency input that is within the frequency handling capability of the direct-writing recorder.

1. Telemetry

In a remote monitoring system, with telemetry components employed for transmission of the measured data, overall system performance will be tied to the bandwidth or channel capacity of the telemetry link. A narrow-band channel may place a limit on the frequency content of the transmitted signal; this in turn can permit relaxation of the requirements for frequency response in other components of the system. Conversely, a firm requirement for monitoring several different physiological responses will dictate an increase in the channel bandwidth of the telemetry link to be used in the system.

2. Interference Filtering

In physiological monitoring, there is always a likelihood that simultaneous phenomena will be sensed along with the desired response. Too, in the measurement situation in the field or in aerospace environments, interference from movement-generated artifacts may intrude on the desired signal. When frequency-selective filters are employed to minimize such artifacts, there will be a corresponding reduction in the overall system bandpass, and the frequency response requirements of the system components may be reduced accordingly.

II. Dynamic Range or Response

A physiological monitoring system, and all components in it must be able to respond to the full range of amplitude variations in the phenomenon being measured. Obviously, a transducer with an operating range of 0 to 14 psi cannot be used to monitor pressure variations of 20 or 25 psi.

Equally important, the operating range of the system should not exceed the measured variation by too large a degree, or the resolution and precision of the system will fall short of its potential. As a general rule, a system should be designed to handle the highest expected amplitude signal amplitude variation, with perhaps a 20-percent added capability for overloads.*

Dynamic range considerations affect all components: in transducers the operating range must equal the input variations; in amplifiers the optimum bias or operating level should accommodate maximum variations without overdriving or distortion; in telemetry components, carrier and transmitter units should be set to operate near full modulation; in recording and display devices, sensitivity settings should permit full range response at the highest input level expected.

*At the same time, system components should be able to handle dynamic overloads of perhaps 10 times the measurement range without damage.

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The dynamic range of a system is clearly related to the overall resolution and precision that can be attained with the system. For example, if a 50-psi pressure variation is to be measured with a 0.5 psi resolution factor (1 percent), a transducer with an operating range of 0 to 500 psi should not be used, or the actual resolution will be reduced by a factor of 10.

Similarly, improper handling of the dynamic range in a telemetry link can reduce the effective signal-to-noise ratio, and system precision will be lost. In a data channel of a telemetry link, the noise level is independent of the information level. Therefore, the higher the level of the information signal, the higher the precision. For example, a 5-volt information channel may possess an inherent noise error of 0.1 volt; if the transmitter is set for full modulation by the information signal, a precision of 2 percent can be attained; if the information signal modulates the transmitter only 50 percent, precision falls off to only 4 percent (ref. 54).

III. System Accuracy and Precision

The accuracy of a monitoring system may be considered as the degree of freedom from error in the measurement. Component specifications, in fact, usually express the accuracy of an instrument in terms of error, such as 1 percent of full scale, or 0.1 volt in a 100-volt range. System precision, on the other hand, is a measure of system resolution. A voltage reading of 0.026 volt is more precise than one of 0.03 volt. It may not be more accurate. That depends upon the amount of error present in the system.

Error may be one of three types: systematic, dynamic, and random. Component specifications generally refer to systematic error. Within a given range, for example, a transducer will have a linear output, but only within the accuracy figure quoted. Systematic error can usually be allowed for in calibration (ref. 40).

Dynamic error results from the inability of a component to follow the variations in the phenomenon being measured. It can also result from changes in the environment of the measurement. Dynamic error of the first type, when it is a function of a known system operating characteristic, such as nonlinear dynamic or frequency response, can be compensated for by calibration. Error caused by environmental change may escape detection and definition, unless the environment is being monitored at the same time.

Random error results from many variable factors. In a panel meter, or oscillograph, for example, mechanical factors such as stickiness or electrical factors such as hysteresis may produce accidental errors; a known input value of, say, 1.65 microvolts may in repeated operations be read or recorded as 1.64, 1.67, 1.65 or 1.66 microvolts. In electronic circuits, tube and circuit noise are random in nature. In radio telemetry, atmospheric noise also is random in nature. All these errors are statistical in nature, and they may therefore be accounted for statistically in interpreting the system output.

The degree of error that can be tolerated, or the amount of accuracy that must be

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attained, will naturally vary with the application. In most instrumentation schemes, an accuracy figure of 1 percent of full scale is certainly desirable, and frequently approachable. In aerospace field applications, an accuracy figure of no more than 5 percent may be the best that can be hoped for in terms of the overall limits of extreme environment, equipment cost, and equipment size and weight.

Overall system error is a geometric sum of component errors. Therefore, when there are practical limits on the amount of error reduction that can be effected, attempts to minimize error should be concentrated on the one or two components that have the greatest error. (See also ref. 39 and 72.)

IV. Effects of the Act of Measurement Upon Measured Parameters

Ideally, the techniques employed to observe physiological phenomena should in no way affect the measured parameters. However, since these effects can seldom be eliminated completely, the goal logically becomes that of minimizing the inaccuracies that result from the act of measurement. There are three principal ways in which such inaccuracies may occur.

First, the measuring device may absorb energy from the system under consideration. For example, an ECG amplifier with a low input impedance will decrease the voltage available at the amplifier input terminals, degrading the accuracy and validity of the measurement.

Second, the sensors attached to the body may distort that region of the body or exert unnatural forces upon the system under observation. For example, a temperature probe may be attached so that it irritates underlying tissues. Such distortions will lead to misleading data.

In reality, few measurement errors of this sort can be ascribed completely to a single factor. As an example, consider blood pressure measurement by means of a catheter. Here, the catheter-transducer system absorbs energy from the circulatory system, and the catheter itself obstructs the normal blood flow to some degree. Thus, both energy absorption and distortion of normal conditions are involved in the measurement, and their effects may be so closely related as to be indistinguishable. Similar problems exist in measuring respiration (by either chest-strap or spirometry techniques) in plethysmography and in muscle strength measurement.

The third source of error is much more subtle and far more difficult to overcome, and it is frequently ignored. It is the psychological state of the experimental subject. Many physiological parameters, such as heart rate, vasomotor tone, GSR, and brain potentials, are highly susceptible to psychological influence. And the psychological state of the subject may change markedly, either as a result of the instrumenting procedure or in reaction to the environmental stresses to which he may be subjected. The psychological factor, then, must be recognized and accounted for if meaningful data are to be obtained.

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SYSTEM INTEGRATION

All of the factors discussed above must be borne in mind when the time comes to put together an instrument system for a particular physiological measurement. The planning of such a system begins with the choice of phenomenon to be measured. Then the range of amplitudes and frequencies expected must be determined. This determines the elementary system input, and the system transducer requirement. System output is determined by the type of recording or display that is required (ref. 67).

The components between input and output are logically determined in the same fashion:

- The type of conditioning that the input signal requires to make possible the recording desired.
- The requirements of data transmission.
- The type of power needed to operate all components.

After the initial measurement requirements are set, other considerations follow:

- The environmental stresses at the measuring site, and how they will affect system operation.
- The physical limitations of components, such as size and weight.
- How active the subject will be, and the problems in attaching the transducer to the subject.
- The probable sources of interference, and the steps that will be necessary to minimize this interference.

Finally, the effect of combining the desired components into a system must be considered so that they will work together without reducing, at any point, the desired information content of the signal.

These are the factors that must be considered in planning individual measurement systems such as described in Section II. Most of the problems of component selection associated with adequate output levels and signal handling capabilities are discussed elsewhere in this handbook. Problems affecting integration of components are discussed briefly below.

I. Impedance Matching

Adjacent components in a physiological monitoring system must be matched prop-

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erly, in terms of input and output impedances, if signal transfer is to be accomplished between them without loss or distortion of the signal. Signal transfer entails the transfer of voltage only (disregarding current), or of power (voltage and current).

Power transfer is maximized when the output impedance of a driving component exactly matches the input impedance of the driven component. At the system level, most signal transfer between components is a voltage transfer, with power transfers being needed between circuits of certain components. Thus, a voltage transfer takes place at the input of oscillographic recorder (an input voltage preamplifier), and a transfer takes place between the driving amplifier and the galvanometric mechanism of the recorder. Power transfer depends upon the resistive and reactive impedances present in the input and output circuits. Maximum transfer occurs when the reactances are equal and opposite (capacitive versus inductive). Properly adjusted transformer circuits are frequently employed for interstage impedance matching.

For voltage transfer between components in a system, the impedance matching problem is not so great. The chief requirement for effective voltage transfer is a relatively low impedance at the output of the driving unit, and a relatively high impedance at the input of the driven unit. The actual voltage transferred is a function of the resistance ratio between the output and input circuits. In vacuum-tube circuitry, input impedances of most stages are generally high, and it is not difficult to effect voltage transfer. In transistor circuitry, however, many circuit configurations possess very low input impedances. Maximum voltage transfer may require the insertion of emitter-follower amplifier stages in the system. Emitter-followers have no significant gain (maximum gain is about unity), but they do have high input impedances. They can be used effectively then as input stages to couple signals to voltage amplifier stages with low input impedances (ref. 67).

II. Environmental Factors

The environmental extremes that will be experienced at the site of measurement must be taken into consideration, since they can have considerable effect upon the operation of the monitoring instruments. Electronic circuits, both signal conditioning and telemetry, may be located in the same physical environment as the subject. Equipment specifications with respect to such environments as shock, vibration, temperature, and atmospheric pressure must meet the demands of such aerospace applications as centrifuges, temperature-humidity chambers, vertical accelerators, and of course high-performance aircraft and spacecraft.

Perhaps the greatest problem is in the successful attachment of electrodes and transducers to human subjects who will themselves be moving about, and will be subject to shock and vibration environments. Enormous movement artifacts can be generated in the sensors under such circumstances, seriously distorting or even completely overriding the desired physiological signal.

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Many transducers are affected by environmental phenomena such as pressure, temperature, acceleration, etc. Variations in the environments will produce unwanted signal variations or changes in baseline signal levels. Transducer specifications must define their performance in the presence of such environments.

III. Equipment Size and Weight

Limitations to equipment size and weight are fairly self-evident. In ground measuring environments, there may be no significant limit on the size and weight of components used in a measuring system. In-flight and spaceborne monitoring must give due consideration to the critical limits on payload of the flight vehicle. Large, unminiaturized components using vacuum-tube circuitry may be further limited by the amount of power available in the vehicle to operate the instrumentation. Transistorized equipment is certainly preferable, including miniaturized units with small, self-contained battery power packs.

Miniaturized components also facilitate the use of personal telemetry systems. These systems permit the attachment of signal conditioning circuitry and even small short-range telemetry transmitters directly to the subject. Such attachment minimizes the wire connections that must be made between the subject and external instrumentation, or even eliminates the need completely.

Size, weight, and power limitations in in-flight monitoring will also affect other system decisions, such as the feasibility of on-board recording in place of air-to-ground telemetry.

IV. Equipment Interaction

Artifact and interference in a system can come from within the system as well as without. Internal interference can arise in several ways, as follows:

- Improper grounding, or too many ground points in a system, can create unwanted circuit loops, permitting the signal in one circuit to affect the signal in another.
- Electromagnetic coupling may cause unwanted coupling between circuits, especially in transformer circuits.
- Similarly, capacitive coupling between lines may distort signals.
- Spurious radiation from oscillator circuits, such as those in subcarrier oscillator stages, may be picked up in other circuits by lines which act as antennas, completely drowning out the data signal in that circuit.

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- Circuit performance may vary when two or more components draw power from the same power supply in such a way as to affect the stability of the output from that power supply.

These are important considerations. How to troubleshoot a system for such problems is discussed in Section III of this volume, together with ways of minimizing such interference (including external interference).

Section II

INDIVIDUAL MEASUREMENT SYSTEMS

HEART POTENTIALS

One of the most useful indices of cardiovascular function is the electrocardiogram, or ECG. The ECG is a graphic trace of electrical potentials that are measured at the surface of the body and are related to cardiac activity. Electrocardiography as a clinical technique has well-established norms based on millions of measurements, and it provides useful physiological data regarding heart rate and chronic and acute irregularities in heart rhythm, both as a result of environmental stress and of cardiac disease. It is also a possible index of overall subject alertness and excitement, especially when used in conjunction with other measurements such as blood pressure and respiration.

As measured at the surface of the body, heart potentials have peak amplitudes of 1 to 2 millivolts, and frequency components ranging from 0 to 100 or 200 cycles per second. Since the potentials at different points on the surface vary with the distance and the direction from the heart, they can be recorded simply by placing two electrodes on different parts of the body, amplifying the potential that appears across them, and applying the amplified signal to a graphic recording device. (The recording obtained, which shows changes in potential with time, is called the electrocardiogram; the apparatus employed, which includes sensors, amplifier, and recorder, is called an electrocardiograph.) In practice, of course, measurement is not that simple, since there are other potentials at the surface of the body besides those having their origin in cardiac activity. Chief among these are resting potentials and muscle action potentials.

Resting potentials, which arise from electrochemical action that takes place in all body tissue, may run as high as 150 millivolts. Being essentially d-c phenomena, they are readily removed from the measurement situation by the use of capacitive coupling between the ECG electrodes and the input of the ECG amplifier or preamplifier (ref. 58). Muscle action or electromyographic potentials are caused by the contraction of skeletal muscle. In clinical electrocardiography these potentials are minimized by immobilizing the subject; in aerospace monitoring, however, subject movement must be accepted, and some degree of interference from this source must be expected. Fortunately, electromyographic potentials are highest in amplitude at frequencies above the range of interest of electrocardiography. Consequently, some measure of control can be effected through low-pass filtering.

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The measurement problem is further complicated by the numerous refinements in electrocardiographic technique which must be considered if successful recording is to be accomplished. These refinements involve: (1) the maintenance of good electrical contact at the skin-electrode junction, (2) the selection of points of the body of the subject for the most useful electrocardiographic signal, and (3) the use of signal modifiers to remove electrical artifacts from the electrocardiographic signal. Furthermore, electrocardiographic techniques that are suitable to the clinic must be modified to satisfy the environmental and operational requirements for monitoring in aerospace applications.

I. Electrocardiographic Norms

The electrocardiogram, which has been used for many years in clinical observation and diagnosis, is a readily interpretable record. The form, amplitude, and time content of the electrocardiograph have been well established through literally millions of clinical observations.

Figure 3 depicts the composition of a standard electrocardiogram obtained with lead II electrodes (refer to the discussion of leads on page 24). This electrocardiogram shows the potential changes during one heartbeat. These potential changes are a function of the depolarization and repolarization of muscle tissues concerned in heart action.*

The smooth P wave in figure 3 results from depolarization of the atrium. The sharp spike demarked by Q, R, and S (called the QRS complex) results from depolarization of the ventricle. The smooth T wave corresponds to repolarization of the ventricle. (Repolarization of the atrium takes place during ventricular depolarization and is masked in the electrocardiogram by the QRS complex.) The smaller U wave following T also is associated with ventricular repolarization; it is low in amplitude and does not appear on all electrocardiograms.

The lead II waveform is the standard representation of the electrocardiogram. It is obtained with one electrode attached to the right wrist and the other attached to the left ankle. Table II gives average values for the lead II electrocardiogram of an adult male subject, listing the amplitudes of the various PQRST components and the duration of clinically significant intervals. Electrodes placed at other points on the body will produce electrocardiograms of similar character, but there will be marked differences in the amplitudes (and even the polarities) of the various components of the waveform. Table III lists average electrocardiographic amplitudes for the various lead systems in standard usage (ref. 3).

* Refer to Volume I of this handbook (pages 18-32) or to standard physiology texts (ref. 48, 61, 62) for a discussion of the relationship between electrical and mechanical action in the heart.

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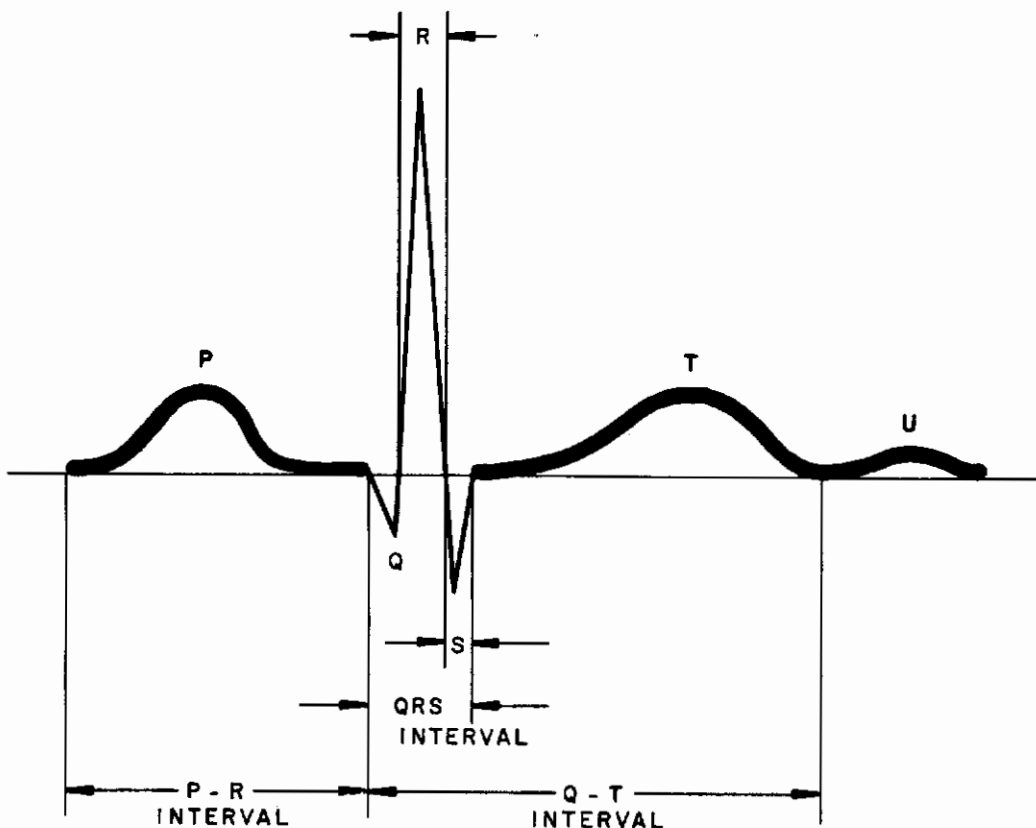


Figure 3. Composition of the Electrocardiogram

TABLE II. CHARACTERISTICS OF THE TYPICAL ELECTROCARDIOGRAM*

Parameter	Duration (milliseconds)	Potential (millivolts)
P wave	90 (70-120)	0.1 (0.02-0.25)
P-R interval	160 (110-210)	
Q	-	0.03 (0-0.19)
R	-	0.98 (0.5-1.72)
S	-	0.01 (0-0.82)
QRS duration	83 (50-100)	
T wave	-	0.29 (0.05-0.63)
Q-T interval	397 (337-433)	
P-R interval - time required for atrial depolarization. QRS duration - time required for ventricular depolarization. Q-T interval - time required for ventricular depolarization and repolarization.		
*Lead II, on an adult male subject.		

TABLE III. AMPLITUDES OF ELECTROCARDIOGRAPHIC WAVES FOR VARIOUS LEAD CONFIGURATIONS

Lead	P	Q	R	S	T
I	0.74 (0.15-1.15)	0.3 (0-1.6)	5.3 (0.7-11.3)	1.0 (0-3.6)	2.2 (0.6-4.2)
II	0.98 (0-1.90)	0.3 (0-1.8)	7.1 (1.8-16.8)	1.2 (0-4.9)	2.6 (0.6-5.5)
III	0.42 (-0.73 to 1.30)	0.4 (0-2.8)	3.8 (0.3-13.1)	1.2 (0-5.5)	0.5 (-0.6 to 3.1)
aVR	0.81 (-1.79 to -0.10)	2.1 (0-9.0)	1.0 (0-3.3)	3.5 (0-14.7)	-2.4 (-5.4 to -0.3)
aVL	0.71 (-0.85 to 1.40)	0.3 (0-2.2)	2.6 (0-7.5)	1.3 (0-9.0)	0.9 (-1.6 to 2.7)
aVF	0.77 (-0.60 to 1.60)	0.3 (0-1.9)	5.3 (0.2-14.8)	1.0 (0-7.1)	1.7 (-0.4 to 4.6)
V ₁	0.61 (-0.80 to 1.80)	0	3.3 (0-9.4)	8.4 (0.8-21.3)	0.9 (-2.0 to 12.2)
V ₂	0.63 (0.15-1.55)	0	5.4 (0.4-15.2)	13.0 (1.9-27.4)	3.6 (-1.4 to 14.4)
V ₃	0.65 (0-1.80)	0.01 (0-0.5)	7.9 (0.6-22.4)	9.6 (0.9-22.2)	4.4 (0-16.0)
V ₄	0.64 (0.10-2.30)	0.1 (0-1.6)	11.9 (1.8-32.0)	6.3 (0.2-20.9)	4.6 (0.5 to 13.1)
V ₅	0.60 (0-2.40)	0.5 (0-2.1)	11.4 (4.2-24.2)	2.6 (0-9.7)	3.5 (0-9.6)
V ₆	0.56 (0-1.90)	0.4 (0-2.7)	9.4 (2.5-26.0)	1.3 (0-8.4)	2.8 (0-6.7)

Note: These values are for an adult subject, 30 to 40 years of age.
The values are in millimeters deflection; read 0.1 millivolt per millimeter.
Mean values are followed by limits, in parentheses.

Ref. 3

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II. Special Monitoring Requirements

The problem in electrocardiographic monitoring in aerospace medical applications is to obtain recordings which provide significant information on the general physiological state under limitations imposed by the experimental situation (ref. 60). Like all of the monitoring systems discussed in this handbook, the electrocardiographic system should be adaptable to measurement situations with stringent limits on size, weight, and power consumption. It should not interfere with the functioning of the subject, and it should provide stable operation for extended periods without adjustment.

Factors that affect the quality of ECG recording and which may have to be tolerated or provided for in the system design are:

- Motion of the subject.
- Vibration from airframe, centrifuge, etc.
- Physical interference with body sensors caused by flight suits, etc.
- Long-term measurements (from several hours to over 24 hours).
- A high degree of subject perspiration.
- Telemetry bandwidth limitations.
- High g acceleration loads.

Most of these factors affect the design and placement of the electrodes.

III. Electrodes

A. Requirements

The basic requirements in the design and selection of electrodes are good electrical contact and low impedance. For electrocardiographic measurement beyond the controlled conditions in the laboratory, additional requirements are imposed by the need for long-term attachment and for subject freedom during measurement. Under these conditions, the following factors are significant:

- The electrode material should be nonpolarizing.
- The electrode should be small and low in mass.
- Electrode-skin contact should be liquid rather than dry.
- The electrode should cause little or no skin irritation.

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B. Characteristics

Electrocardiography outside the laboratory, while not standardized, is a proved technique. Numerous users have successfully made long-term ECG measurements under various conditions of environmental stress and subject movement. The various electrode techniques employed are summarized below.

1. Configuration

Electrodes are formed as flat discs, circular wire meshes, or cups (with preformed "pockets" for restraining an electrolyte). They are small and light, with electrode-skin junctions as small as 1/2 inch square, as compared with clinical plate electrodes measuring roughly 2-1/2 x 1-1/2 inches.

2. Material

The electrode element is generally stainless steel or Monel metal, silver, or silver/silver chloride composition. Acceptably low amounts of polarization can be maintained with any of these materials.

3. Physical Attachment

Some users attach the electrode directly to the body by lapping the edges of the electrode with a quick-drying, hard cement such as Duco. Adhesive tape and adhesive cork combinations are more commonly employed. Simple elastic straps about the chest also are used, but at the cost of some additional subject discomfort.

4. Electrical Contact

An electrolyte is normally applied between the electrode and the skin. It is a paste or a gel with a binding ingredient, such as bentonite clay, to keep it moist and viscous for extended periods (48 hours or more). Use of electrolyte assures good electrical contact and a low junction resistance; more significantly, the liquid nature of the contact assures a constant junction resistance by isolating movement. A restricting factor in the use of electrolyte has been the irritation and damage to tissue it can cause, especially during long-term measurement; this restriction is reduced in proportion to the decrease in the size of the electrodes employed.

5. Notes on Lead Impedance

Variations in the previously discussed factors affect the impedance of the skin-electrode junction. Indeed, many of the variations have been devised with the express purpose of minimizing electrode impedance. Certain of the smaller electrode configurations, however, were designed primarily to overcome movement artifact or to reduce discomfort in long-term applications. The higher impedance inherent in a

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smaller skin-electrode junction has not been a problem where the associated electronics have a correspondingly high input impedance; as long as the impedance ratio is high, good voltage transfer is obtained for recording (ref. 60).

When electrode and amplifier impedance combinations are allowed to go too high, however, reactive pathways are introduced into the system, permitting pick-up of artifacts from adjacent electrical or electronic systems. Although there are methods for removing such artifacts, it may be more desirable to use electrode and amplifier circuits with lower impedances. Electrodes of the fluid or floating type generally provide the required low electrode impedance. An electrode of the fluid type was used in the NASA Mercury program (ref. 76). It consisted of a metal cup in an inert holder, with a layer of modified bentonite electrolyte between the cup and the skin of the subject. The resistance of this electrode at application was only 1500 to 2500 ohms. After 12 hours, the resistance had risen to only 3000 ohms, and after 24 hours to only 3700 ohms (on the average). Other electrodes, held directly against the skin with adhesives of various kinds, have displayed inherently higher resistances (50,000 ohms and up) at application, and these resistances have risen, in some cases, more than 300 percent over a 24-hour period.

IV. Leads

Historically, the correlation of cardiac activity with surface potentials dates back to 1856. Quantitative recording awaited the development of suitable instrumentation. In 1887, Ludwig and Waller successfully measured heart potentials at the chest using a capillary electroscope. Einthoven, in 1903, established electrocardiography as an exact technique with successful application of the string galvanometer.* He originated the technique of measuring heart potentials at the extremities, and established the recording conventions governing the graphic representation of these potentials. His system of electrodes, called limb leads, has remained a standard point of reference for most subsequent clinical electrocardiography. (See ref. 13 and 62.)

Einthoven's limb leads are termed bipolar; that is, they register the relative difference in potential between two points on the body. Other systems of leads have been developed to record the potential at various points relative to some reference point: these are called unipolar leads. Still other leads have been employed to obtain recordings of greater amplitude, to better emphasize certain components of the heart's electrical activity, or to avoid the myographic potentials from skeletal muscle masses. The various lead types are discussed below. Note that in these discussions oriented toward the medical practice, the term "lead" is used differently than in standard electrical or electronic usage, where the terms "lead" and "electrode" are practically

* With the development and perfection of electronic amplifying components, Einthoven's string galvanometer has been largely replaced by more sophisticated instrumentation. However, it has remained the standard despite the fact that it is more difficult to use and it requires photographic development of the record.

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synonymous. Electrical measurement presupposes the use of at least two electrodes to complete a measurement circuit. In electrocardiography usage, the term "lead" actually designates a system of two or more conducting paths with terminal electrodes.

A. Standard Limb Leads

Standard limb leads, using the Einthoven technique, measure differences in potential at the extremities of the body. Three electrodes are used, one on each wrist, and one on the left ankle (see figure 4). Different potentials will be measured between any two of the electrodes, so that the three electrodes form leads I, II, and III, as follows:

1. Lead I. The electrocardiograph is connected above the wrists, between the right (R) and left (L) arms. The polarity of the connections is chosen (following Einthoven's conventions) so that when L is positive relative to R, there is an upward deflection of the P and R segments of the electrocardiogram.

2. Lead II. The electrocardiograph is connected between R and above the left ankle (F). Polarity of the connections is such that there is an upward deflection of the P and R segments of the electrocardiogram when F is positive relative to R.

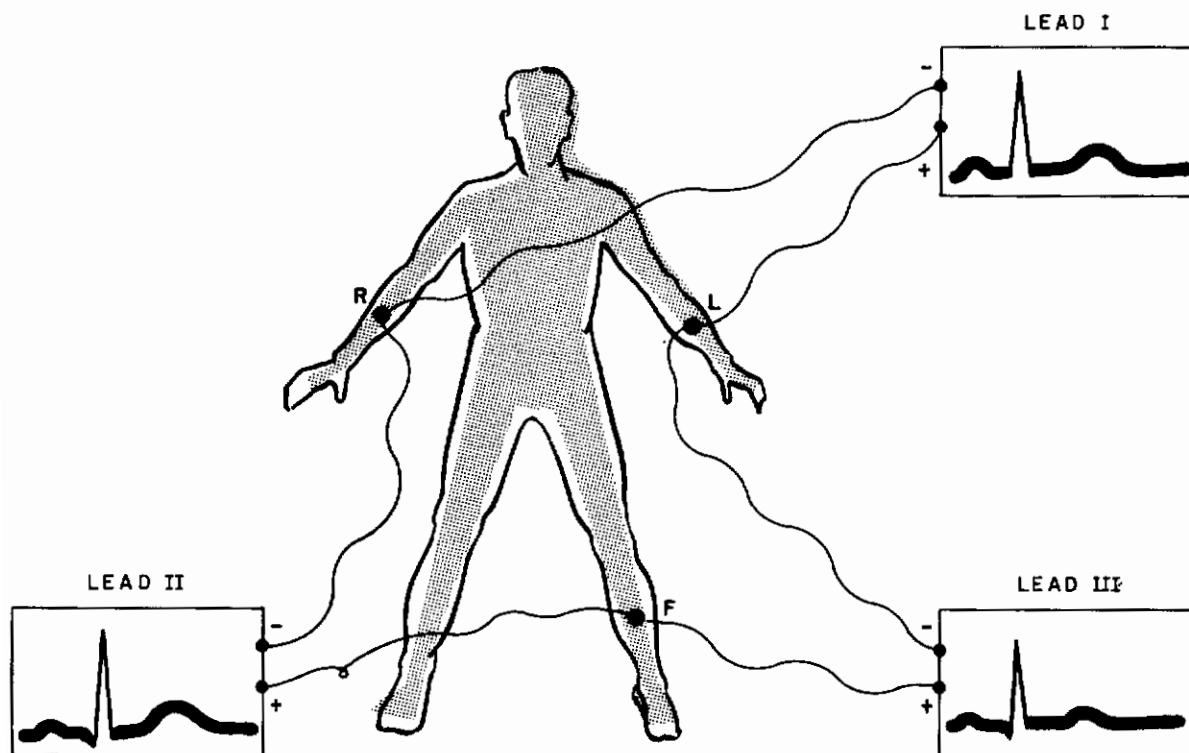


Figure 4. Standard Limb Leads

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3. Lead III. The electrocardiograph is connected between L and F. Polarity is such that there is an upward deflection of the P and R segments of the electrocardiogram when F is positive relative to L.

As mentioned previously, limb leads I, II, and III do not measure an absolute voltage at any of the three sites of electrode placement, but rather the difference in potential that exists between any two. Thus the voltages in each lead may be designated as follows:

$$\text{Lead I} - V_I = V_L - V_R$$

$$\text{Lead II} - V_{II} = V_F - V_R$$

$$\text{Lead III} - V_{III} = V_F - V_L$$

By simple algebraic reduction, it can be seen then that

$$V_{II} = V_{III} + V_I$$

This elementary relationship, known as Einthoven's law, makes clear that when any two of the standard lead potentials are measured, the third is obtained. If the three standard leads are recorded simultaneously, the relationship is self-evident. The same rule applies for recording the various leads serially; however, there may be appreciable changes in the state of the subject and his accompanying cardiac activity, even in the brief time between measurements, and allowance must be made for these changes. The relationship is pointed out to the reader here for what usefulness he will make of it; ready knowledge of the relative potential values can help in identifying and labeling traces from a series of measurements.

B. Unipolar Leads

To determine the voltage at each of the limbs relative to some fixed reference point as opposed to the difference in voltage between limbs, a system of unipolar limb leads was developed (ref. 77). A so-called indifferent electrode or reference point is produced by forming a central terminal (CT) by connecting each of the limb leads together through 5000-ohm resistors, as shown in figure 5, and connecting this CT to the negative side of the electrocardiograph. An electrode from the positive side of the electrocardiograph, known as the exploring electrode, is then placed at the right arm, the left arm, or the left ankle to measure, respectively, V_R , V_L , or V_F . The indifferent electrode is the reference point for all measurements.

A modification of this lead technique is used to obtain recordings of greater potential amplitude. These augmented unipolar limb leads, devised by E. Goldberger (ref. 27), call for the removal of the CT connection to the point where the exploring electrode is placed (see note on figure 5). Modern electrocardiographs generally provide for this change in lead wiring through a built-in selector switch; it is the standard

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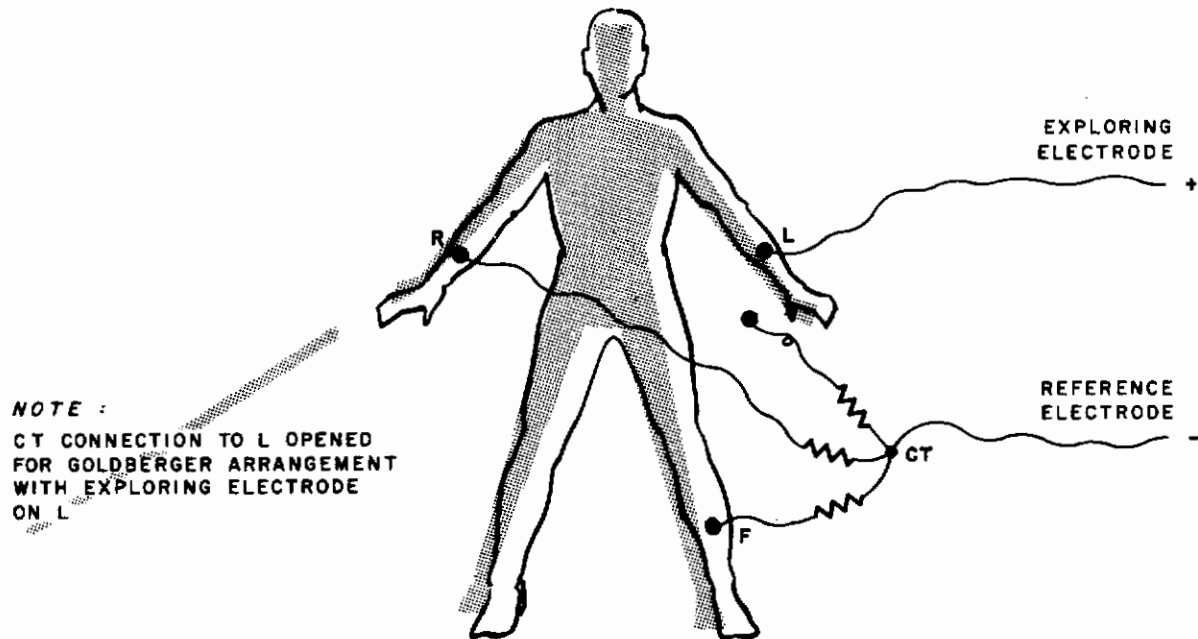


Figure 5. Unipolar Limb Leads

unipolar limb lead wired into most electrocardiographs. Polarity of the connections is as follows:

1. Lead aVR. The exploring (positive) electrode is placed at R, and the negative side of the electrocardiograph is connected to both L and F through resistors.
2. Lead aVL. The exploring electrode is placed at L, with R and F tied to the negative side of the electrocardiograph through resistors.
3. Lead aVF. The exploring electrode is placed at F, with R and L connected to the negative side of the electrocardiograph through resistors.

C. Precordial Leads

Another major group of electrodes commonly used in clinical practice is precordial leads, so called because the exploring electrode is placed at various points on the precordia, that area of the chest overlying the heart. Precordial voltages are designated by the point on the chest at which the electrode is placed. Six of these points, shown in figure 6, have been standardized as follows:

- V_1 - at the fourth intercostal space (ICS) at the right sternal border.
- V_2 - at the fourth ICS at the left sternal border.

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- V_3 - at a point midway between V_2 and V_4 .
- V_4 - fifth ICS at the left midclavicular line.
- V_5 - left anterior axillary line at height of V_4 .
- V_6 - left midaxillary line at height of V_4 .

All of the above points are established with reference to an indifferent electrode connected to the CT*, with polarity such that when the precordial point is relatively positive, there is an upward deflection on the electrocardiogram. (See ref. 3 and 13.)

D. Leads for Active Subjects

The lead configurations discussed thus far are those developed in clinical practice. Under such conditions, rather restrictive measurement conditions prevail: the subject is not moving (in fact he is usually resting on his back to ensure a "normal" sample of heart activity); and the measurements are short term. The electrodes are required to yield good results for only a short period of time.

Obviously, when monitoring relatively active subjects in aerospace environments, lead configurations must be used that are less critical than those used in clinical measurements. For one thing, the placement of electrodes at the extremities is virtually prohibited because of subject movements, restricting electrode placement to the torso. Locating electrodes on the torso must be done carefully. The usual criteria are to find a site over bone and away from large muscle masses. There are two lead configurations, the axillary and the sternal, which generally satisfy these requirements. (See ref. 60 and 76.)

* Measurements also are made with so-called single precordial leads. For these measurements, no central terminal is employed. The exploring electrode is placed over one of the six designated points on the chest, as indicated above, and the second indifferent electrode is placed at either the left arm (L), right arm (R), or left leg (F). To record with single precordial leads, using an electrocardiograph that is wired for standard leads I, II, or III, care must be taken to maintain polarity, so that upward deflections are obtained when the exploring electrode is positive.

The convention is as follows:

For lead I (R and L) electrodes, R is connected to the left leg, and L is used as the exploring electrode.

For lead II (R and F) electrodes, R is connected to the left leg, and F is used as the exploring electrode.

For lead III (L and F) electrodes, L is connected to the left leg, and F is used as the exploring electrode.

HEART POTENTIALS

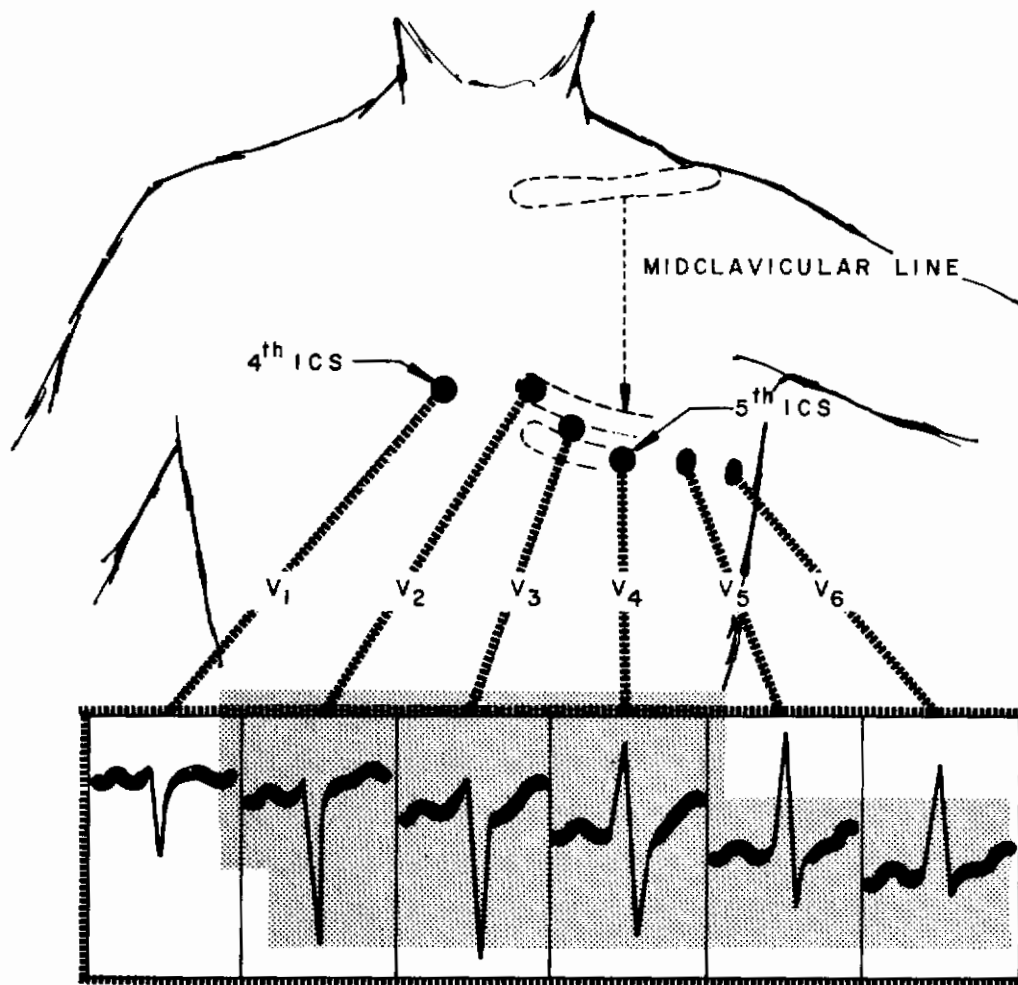


Figure 6. Unipolar Precordial Leads and Typical Waveshapes at Each

1. Axillary Leads

Axillary leads perhaps come closest to approximating the voltage relationships obtained with leads placed on extremities. The equivalent R and L electrodes are placed on the chest on the midaxillary line. Potentials obtained here are related closely to the potentials at the shoulder, which are essentially the same as those normally sensed by electrodes placed out on the arms. The equivalent F electrode is generally placed on the left side of the trunk, at the base of the rib cage. Electrocardiograms obtained with axillary leads resemble very closely those obtained with standard leads I and II. The resemblance to electrocardiograms obtained with lead I is particularly good with the two electrodes placed on the axillary line at about the level of the heart. The voltage obtained is not quite as high as that obtained with a sternal lead (see figure 7), but it can be increased by moving the positive electrode (left side of chest) forward, closer to the heart. The resemblance to the standard lead cardiogram

Contrails
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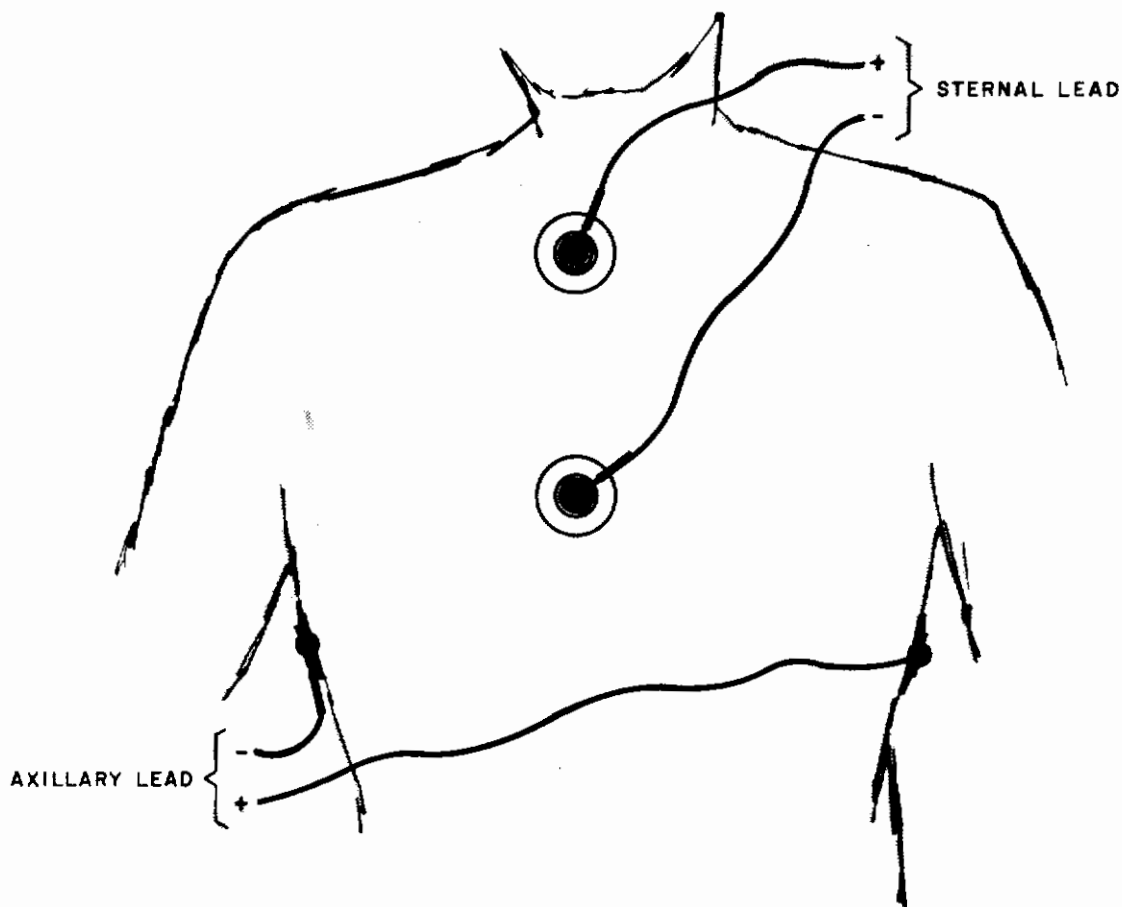


Figure 7. Axillary and Sternal Leads

suffers, however, and there will be significant changes in the electrocardiogram for very slight changes in position of the positive electrode.

Axillary leads are moderately free from movement artifacts. They are necessarily placed over muscle tissue, and raising or swinging of the arms can introduce large electromyographic signals into the ECG. Where wide arm movements are restricted (for example, when the subject is wearing a pressure suit), axillary lead systems can produce useful electrocardiograms.

Figure 7 shows an axillary lead system used in NASA's Project Mercury (ref. 76). Resembling a modified lead II, the equivalent R electrode is placed on the rib cage at the third intercostal space, and the equivalent F electrode is placed at the base of the rib cage. (The axillary F electrode is sometimes placed near the left groin, and if a ground electrode is used, it is placed near the right groin.)

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2. Sternal Leads

Sternal leads offer several advantages over axillary leads in aerospace monitoring. Because sternal leads are placed nearly directly over bone, they are relatively free from motion artifacts. It has been reported that if the sternal configuration has been applied carefully, the electrocardiogram is undisturbed even while the subject's chest is being pounded (ref. 73). Further, the attachment over bone insures longer lasting skin-electrode contact and freedom from baseline shift during subject activity (as seen in other lead arrangements). Sternal leads also produce readings with much higher voltage amplitudes than other configurations with larger, more distinct P and T waves, because of their proximity to the heart. These higher readings can be a disadvantage in that only slight shifts in electrode placement or in the position of the heart produce markedly different readings on the same subject. As a result, correlation of airborne readings with baseline data is difficult. It would be preferable if baseline data were accumulated using the same electrode positions as the actual field experiments (ref. 60).

Figure 7 shows the placement of leads on the sternum. The upper electrode is placed on the manubrium, and the lower on the xiphoid process. Polarity is normally established for an upward deflection on the electrocardiogram when the upper electrode is positive relative to the lower.

V. Signal Conditioning

A. Amplifiers

The principal characteristics of interest in electrocardiographic amplifiers are (1) high gain, (2) high input impedance, and (3) differential input, with high common mode rejection.

1. Gain

Cardiac potentials measured at the surface have amplitudes of 1 to 2 millivolts, peak to peak. The first amplifier or preamplifier generally should provide a minimum gain of 500, so that at least 0.5 volt, peak to peak, is delivered at the output. The output voltage required will vary depending upon the particular application.

2. Input Impedance

The ECG amplifier should have a high input impedance, relative to the output impedance of the related electrodes, for a maximum voltage transfer. A typical value is about 100,000 ohms, measured from either side of a balanced differential amplifier to ground and an equal or higher value when measured between the two sides.

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3. Differential Input

In electrocardiography, electrical signals are measured between two points on the body of the subject, rather than between the subject and some common point or ground. A differential rather than a single-ended amplifier is therefore most generally employed; the output of this amplifier is a function of the difference between the voltages at the two input leads. One of the chief advantages of such an amplifier is that a common voltage present between both inputs, and a ground point will not be amplified since it produces no difference voltage. Thus, artifacts present at the input, such as 60 cps hum picked up by both leads, will cancel out. Amplifiers with common mode rejection on the order of 1000 to 1 (50 to 60 db) are commonly available.

4. Capacity Coupling

The electrodes in a lead system are normally coupled to their amplifiers through capacitors, or RC coupling networks. The time constant in such a network is chosen long enough to pass the lowest frequency of interest in the electrocardiogram (0.1 or 0.2 cycle per second), while still presenting an essentially open circuit to unwanted d-c potentials (such as body resting potentials, which can be large enough in amplitude to completely override the electrocardiographic potentials).

5. Bandwidth

The normal specification for bandwidth in ECG amplifiers is from approximately 0.1 cycle per second to a few hundred cycles. However, oscillographic recorders often limit the upper frequency response to 100 cycles per second or less. Further, low-pass filters with corner frequencies of no more than 100 cycles per second may be employed purposely to remove high frequency interference from the electrocardiogram. It can be seen that extremely high frequency transmission is not a critical factor in ECG systems and is often undesirable.

6. Output Voltage

Usual ECG amplifiers or preamplifiers normally provide an output of 1.0 to 5.0 volts rms although this may vary with application.

7. Output Impedance

The optimum output impedance of an ECG amplifier depends upon its application. If it is to feed another amplifier, and maximum voltage transfer is desired, it is generally enough to specify that its impedance is low as compared to the input impedance of the following stage. If the amplifier is to supply appreciable power to a pen or indicating device, its output impedance should equal that of the device it is driving.

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8. Stability

The amplifier characteristics should remain constant over the time period of measurement. Since the chief factor affecting stability is temperature variation, amplifier stability is frequently specified both as gain stability versus time and over an ambient temperature range (e.g., $\pm 1\%$ from 0°C to 50°C).

9. Note on Carrier Operation

One of the drawbacks in amplifier operation at extended low frequencies, is susceptibility to transient blocking when straight RC coupling is employed. Blocking occurs when the amplifier output is driven to saturation by sudden high-amplitude transients. With the long time constant required for low-frequency operation, this condition may be sustained for as long as several seconds. One method of dealing with this is to use a carrier amplifier rather than a standard differential amplifier (ref. 45). The input signal is used to modulate a higher frequency carrier, and it is this higher frequency modulated signal which is amplified. The higher frequency amplifier, with its shorter time constant, will be free from blocking effects. Modulator design for such an amplifier must be optimized to reduce the inherently high noise level in modulator circuits.

B. Filters

The generally accepted frequency spectrum in clinical ECG recording is from 0.1 to 100 cycles per second. In aerospace applications, several artifacts will be unavoidably present at these frequencies. Several users have demonstrated that the bandwidth of the electrocardiogram can be reduced, by cutoff filtering at both the low and high ends, to attenuate unwanted signals without destroying the pertinent electrocardiographic data. (See ref. 60 and 71.) The goal is reduction rather than elimination of artifact, so that the record obtained is still interpretable. If too much filtering is employed, in an attempt to completely remove artifact, significant information may be removed as well; about all that may be preserved is the rudimentary heart rate indication.

1. Low-Pass Filtering

With certain leads, and with certain kinds of subject movement during recording, muscle action potentials are a frequent source of interference. Most of the energy associated with these potentials falls within a range from 100 to several thousand cycles. By eliminating some of the higher frequencies, most such myographic tracings can be removed, or at least reduced to the point where the electrocardiogram is interpretable. Low-pass filters with a cutoff frequency of 100 cps, and an attenuation of 12 to 18 db per octave above that point are generally accepted as having no significant effect on cardiographic data.

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Lowering the cutoff frequency to 50 cps may also be desirable to remove 60 cps hum, as well as myographic interference. However, with some subjects, the amplitudes of certain ECG components, such as the Q and S waves, will be reduced. Filtering out frequencies above 20 or 25 cps will cause even greater changes in the ECG components. The ECG may be made readable, at least in terms of rate, in the presence of strong myographic interference; but there will be marked distortions in the amplitude and duration of the QRS complex, the S wave, the S-T segment, and other components. Some of these changes may be attributed to the phase shift that is introduced by filter circuits. Just how much of such distortion can be tolerated depends on the intended use of the electrocardiogram so obtained.

2. High-Pass Filtering

Electrocardiograms utilizing frequency response down to dc will exhibit some degree of baseline shift due to electrode polarization or changes in the resistance at the skin-electrode junction. The degree of shift will vary with the type of electrode used, the amount of subject movement experienced during recording, and the length of time of the recording. This shift can be minimized by high-pass filtering.

High-pass filters with a cutoff frequency of 0.1 or 0.2 cps, and an attenuation of 6 to 12 db per octave, will eliminate the very low frequency baseline shift without destroying useful ECG information. Increasing the cutoff frequency, upwards toward 1.0 cps, will distort such ECG components as the S-T segment and the T wave, seriously affecting the interpretability of the electrocardiogram.

VI. Notes on Operation

A. Recording

The electrocardiogram is by definition a graphic trace of heart potentials. Typically, the electrocardiogram is produced directly during measurement on an oscillograph. It can also be displayed on an oscilloscope (and recorded there with a camera), or recorded electrically on magnetic tape for subsequent processing, display, or graphic recording.

Graphic recording of the heart potentials was established as a precise technique by Einthoven, in his work with the string galvanometer. Following Einthoven's practice, an input of 1 millivolt from the electrodes is represented by a pen deflection of 1 centimeter on the electrocardiogram. (Refer to figure 8.) Standard electrocardiograph chart paper is ruled in 1 millimeter blocks, with a heavier line every 5 millimeters.

Standard chart speed for ECG recording is 25 millimeters per second, so 1 block along the horizontal axis represents 0.04 second, or 40 milliseconds, and 5 blocks along the horizontal axis represents 0.20 second, or 200 milliseconds. Where

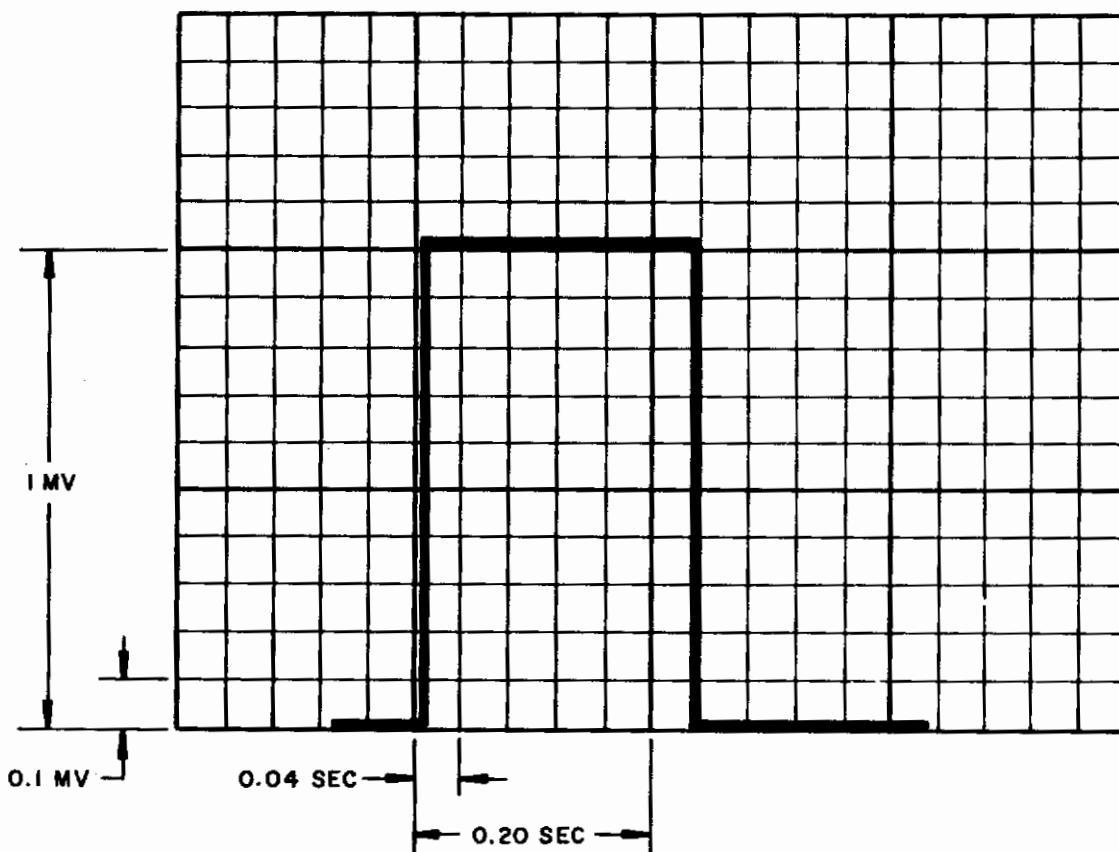


Figure 8. ECG Recording Conventions

an expanded record is desired for detailed analysis, the chart speed may be increased. Similarly, on a multichannel recorder, the sensitivity of a given channel may be increased so that, for a given input, there will be a greater deflection of the recording pen, permitting full-width recording of a single channel on chart paper designed for two, four, or more channels.

Of course, the actual voltage driving the pen motor in an electrocardiograph is much higher (by a factor of 1000 or more) than the voltage sensed on the body with electrodes. To standardize the recording, a 1-millivolt signal from a controlled source is applied to the input of the electrocardiograph, and the gain and sensitivity adjusted until a 1-centimeter deflection is obtained. Most commercial recorders contain such a calibration voltage source.

B. Balance Control

Most differential amplifiers contain a balance control to permit adjustment of the common mode rejection. It consists of a potentiometer control for setting the relative gain in each half of the amplifier. For initial adjustment the two input terminals

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are connected together and a relatively high-level signal (about 100 millivolts, for example) is applied between both inputs and ground. The balance control is then adjusted for minimum deflection of the recording pen.

C. Shock Hazards

In any measurement involving the connection of a subject through electrodes to electronic circuitry, care should be taken to avoid exposing the subject to electrical shock and burns. One step that can be taken is to employ transistor circuits in the first stage amplifiers associated with the electrodes, to isolate the subject from high-voltage circuits. Care should also be taken in establishing ground leads, especially when a ground electrode is used with lead systems, to avoid ground return paths, through the lead system, paralleling high-voltage ground returns.

With the best wiring techniques, however, there is always a danger of shocks through inadvertent shorts between the leads and following circuitry. Each electrode lead should therefore be fused. A 5-milliampere fuse will generally limit current in the lead to nondestructive levels. For further protection, primarily from high-voltage transients, each electrode lead should be shunted to ground through voltage-limiting diodes. A typical fuse-diode system is shown in figure 9 (ref. 26).

HEART SOUNDS

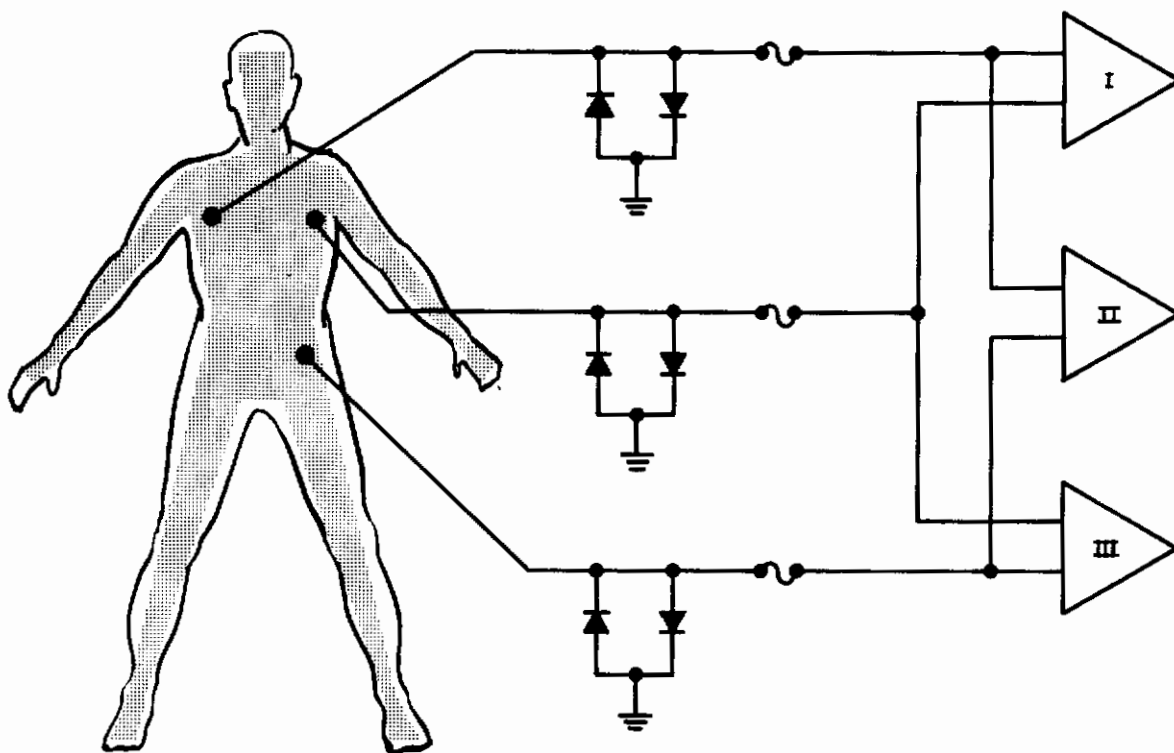
The pattern of sounds produced by the various mechanical events of the cardiac cycle is one of the easiest to monitor. As precordial vibrations, these sounds can be detected with the unaided ear placed on the surface of the chest. A simple mechanical aid, the stethoscope, has been in use since the early 19th century. Phonocardiography is the current technique of recording these sounds directly to obtain a graphic record. The chief limitation of this technique in aerospace applications is inherent in the measurement itself: the sound-conducting structure of the torso transmits high-level ambient noise to the transducer with the same facility that it transmits the physiological sounds. Hence, measurements in certain vibratory or sound-pressure environments are not technically feasible, except with drastic filtering to reduce artifacts.

I. Components of the Phonocardiogram

There are several sources of the sounds that accompany the cardiac cycle: these include the hydraulic pressure transients associated with sudden acceleration or deceleration of blood flow; valve closure; vibration of the heart chamber walls and valves; and turbulence of blood flow (detected as heart murmurs).

There are four principal sounds. The first two are quite pronounced; the latter two are seldom heard in direct auscultation, but are present in most phonocardiograms. These are:

HEART SOUNDS



NOTE :
ALL DIODES SILICON TYPE

Figure 9. Typical Fuse-Diode System for Shock Protection

1. Onset of ventricular contraction, including the closing of the atrio-ventricular (mitral and tricuspid) valves.
2. End of ventricular systole, including the closure of the aortic and pulmonary valves, and the opening snap of mitral or tricuspid valves.
3. Filling of the ventricles.
4. Atrial contraction, with the final movement of blood from the atrium into the distended ventricles.

The characteristics of the four sounds are listed in table IV. Refer to standard text (ref. 42 and 63) for further details.

Figure 10 shows the sounds as they appear on a typical phonocardiogram and relates them to the electrocardiogram (ref. 3). Note the extracardiac sound that appears between the first and second sounds; while its source is uncertain, it appears to be related to respiration and to changes in posture.

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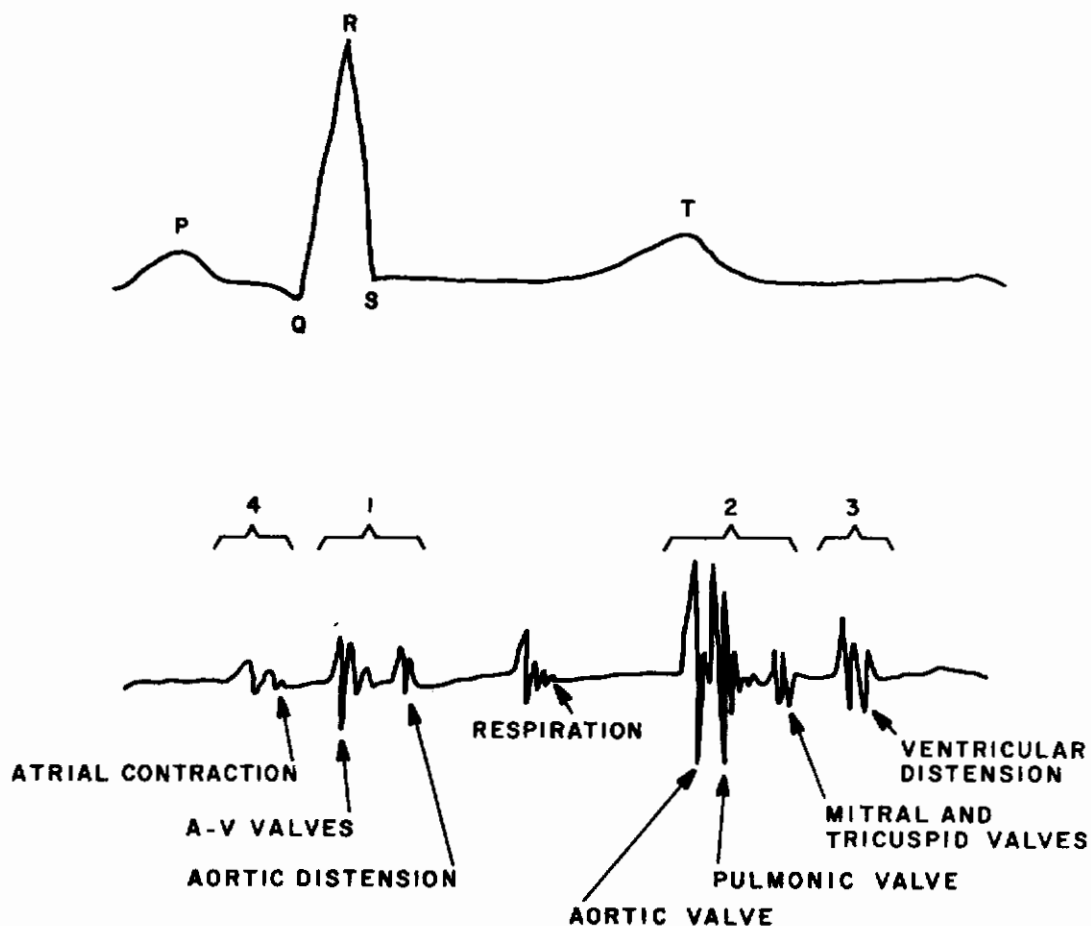


Figure 10. Components of the Phonocardiogram and Their Relationship to the Electrocardiogram

TABLE IV. MAJOR HEART SOUNDS

Sound	No. of Vibrations (cycles)	Total Duration (milliseconds)	Period (milliseconds)	Calculated Frequency (cycles per second)
1	5-11	60-160	9-30	45 (33-111)
2	3-8	40-110	-	50 (31-111)
3	1-3	20-90	20-30	33 (32-50)
4	1.5-3	40-60	12-38	12-84

HEART SOUNDS

II. Microphones

The stethoscope was the first device used as an aid in the detection and monitoring of heart sounds. A purely acoustic device, it does not actively amplify the sound, but channels and directs it more effectively from the chest to the ear. The stethoscope also aids by distorting the sound so that the frequencies produced by the sound more nearly match the frequency response of the ear, which falls off logarithmically both at the low and high ends of the audible spectrum (about 20 and 15,000 cycles, respectively). Stethoscopes of the diaphragm type attenuate low frequencies, while those of the bell type attenuate high frequencies.

With the development of electronic instrumentation, transducers (microphones) have been employed to amplify the sounds of the heart. Microphones are used in stethoscopes for improved auscultation. They are essential in phonocardiography to provide the electrical input to a graphic recording system.

A. Attachment

Some microphones are designed to be attached directly to the subject; that is, one element (usually the diaphragm) is pressed directly against the chest. This type corresponds to diaphragm-type stethoscopes, and if the contact pressure required is high, the microphones tend to attenuate low-frequency chest vibrations.

Other types of microphones are attached indirectly; a cup or bell-shaped housing is used to create a closed air volume between the chest and the diaphragm. The air pressure against the diaphragm then varies with the excursions of the skin beneath the housing. With this type of attachment, resonance of the closed air chamber must be avoided. A minimum volume of 10 cubic centimeters is recommended, and the air chamber should have a diameter-to-height ratio of 3:1 (ref. 49).

B. Microphone Characteristics

Except for the carbon or carbon-pile type microphone, which is too non-linear in response for successful recording, most microphones can be used for detecting heart sounds. Piezoelectric and capacitive microphones produce linear tracings proportional to the displacement of the chest wall. Dynamic or moving coil microphones produce outputs that are proportional to the velocity of the displacement. A desirable byproduct of the operation of the latter type is that the response increases with the frequency of the sounds; therefore, built-in signal equalization or low-frequency attenuation is provided, which results in balanced response over the frequency range of interest.

Other characteristics of microphones that are important when measuring the cardiac cycle are as follows:

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- Output. Most microphone outputs can be applied to any high-gain amplifier or preamplifier that accepts millivolt inputs.
- Sensitivity. Sensitivity is expressed in terms of output below 1 volt per microbar of sound pressure. For example, a microphone with a rated sensitivity of -60 db generates a voltage that is 60 db below a reference level of 1 volt per dyne per square centimeter. Microphones are calibrated accordingly in terms of a constant pressure applied uniformly to the diaphragm.
- Frequency Response. Most microphones used in phonocardiography have a frequency response from 10 or 20 cps to 1000 or 2000 cps. The full-range response is desired in some phonocardiographic registration, but a high-frequency cutoff at about 500 cps is employed frequently to minimize ambient noise. In either case, the use of many standard ECG amplifiers is prohibited because, while otherwise adequate, they are designed for a linear response to no more than 100 or 200 cps.
- Dynamic Range. Microphones should have a useful dynamic range of about 80 db to handle all sound data from the high intensity low-frequency waves (about 10 cps) to the threshold response at about 1000 cps.
- Impedance. About 100, 200, or 400 ohms.

III. Signal Conditioning

The phonocardiogram obtained with the microphone is modified principally by selective filtering to remove high-frequency noise components or to equalize response in studying the full frequency range of interest. Unless real time monitoring is essential to the application, the phonocardiogram should be recorded directly on magnetic tape to preserve all the frequency data possible, and subsequent playbacks should be filtered for more detailed study and analysis of the data. (See ref. 49.)

Since the frequency spectrum of the phonocardiogram is in the audio range, magnetic recording on magnetic tape by the direct technique (AM) usually is satisfactory. A good quality audio instrument probably serves as well as an instrumentation recorder. AM recording is preferable because of its inherently better signal-to-noise ratio. However, there may be some distortion at the low-frequency end, where the wavelength may exceed the width of the gap in the recording head. Indirect (FM) recording on an instrumentation recorder avoids any such distortion.

When a full analysis of the phonocardiogram is desired, a single graph or trace obtained by a microphone does not properly cover the whole sound spectrum, since sound intensity decreases with frequency at about 12 db per octave. A more useful

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representation is obtained by dividing the spectrum into several channels, using band-pass filters, and providing attenuation or equalization as required to produce multi-channel recordings at a standardized amplitude for each channel or band.* Figure 11 illustrates this concept. The heavy vertical arrows represent the amount of amplification or attenuation needed to equalize each channel.

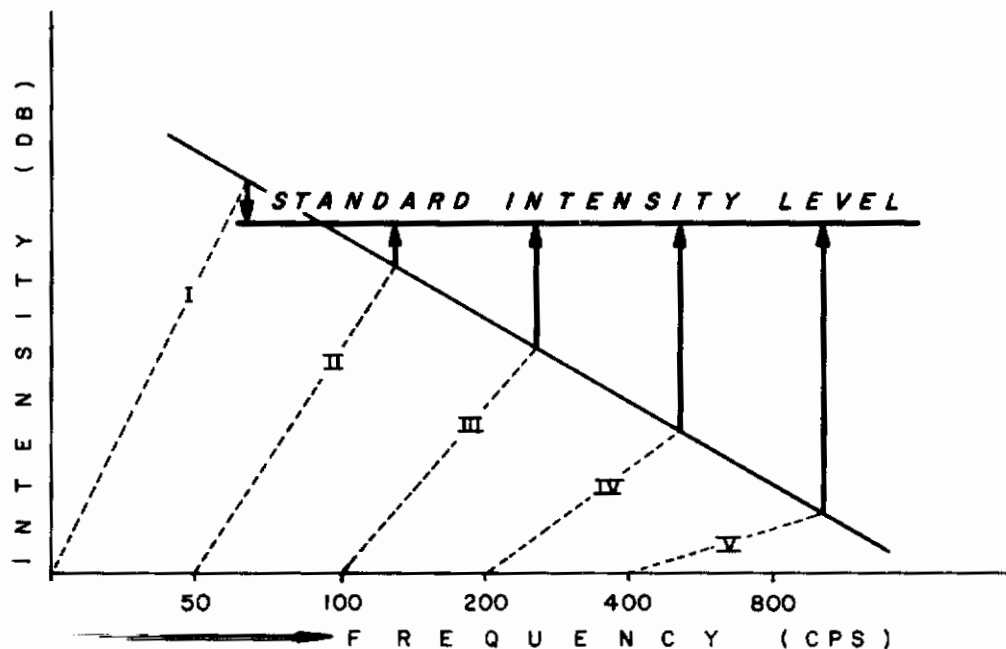


Figure 11. Five-Band Division of Frequencies for Signal Equalization

IV. Notes on Operation

A. Display and Recording

Heart sounds may be displayed readily on a cathode-ray tube. With its fast response, an oscilloscope CRT can easily handle the wide frequency range. However, for optimum results, or for good photographic recording of the CRT display, the intensity and focus controls must be adjusted to compensate for the relatively weak beam intensity at higher frequencies.

* The sound spectrogram is another more sophisticated technique developed by Bell Telephone Laboratories to give a better visual equivalent of the sounds normally heard in auscultation. The spectrogram yields a frequency-versus-time display on an oscillograph. While it "looks" more like the stethoscope "sounds," some authorities feel that the conventional amplitude-versus-time trace can be interpreted more readily by the clinician or physiologist.

INDIVIDUAL MEASUREMENT SYSTEMS

For full-range recording, the optical galvanometer is probably a better choice than the CRT. (Direct-writing instruments, with a maximum frequency response of about 200 cps, would only be able to handle a portion of the phonocardiographic data.*) For phonocardiographic registration, the chart speed should be 50 to 75 millimeters per second (compared to the standard 25 millimeters per second for the ECG).

B. Location of the Microphone

Several areas of the chest (and back) are suitable for phonocardiographic registration, depending upon the sound characteristics that are of interest. Four of the more commonly used locations are the following (see figure 12 and ref. 49):

- Aortic area. At the second intercostal space (ICS) on the right sternal margin.
- Pulmonary area. At the second ICS on the left sternal margin.
- Left lower sternal border. At the fourth ICS on the left sternal margin.
- Apex beat. At the fifth ICS on the midclavicular line.

C. Correlation With Other Data

Accurate interpretation of the phonocardiogram usually depends upon its correlation with other physiological data that have been recorded at the same time. Simultaneous registration of the ECG is almost mandatory for this purpose. Arterial (carotid) pulse data also may be useful in pinpointing aortic valve closure. A respiration rate tracing sometimes is useful with respect to the splitting of the second heart sound that is seen on some phonocardiograms. (In the clinic, respirations are often marked manually on the phonocardiogram by the recording technician.)

HEART RATE (AND PULSE RATE)

The heart rate, giving the number of beats or pulses per unit time, is a valuable index of overall cardiovascular function. Although rate information may be derived readily from recordings of other measurements (e.g., the QRS complex of the ECG and the pulses on a pulse pressure waveform), a cardiometer or cardiograph is used to obtain rate information without manually processing large amounts of recorded data (by counting peaks for a given amount of time).

* The Russians have cited a technique (ref. 5) of integrated phonocardiography which, while retaining most significant cardiac data, can be accomplished with a relatively narrow-band channel.

HEART RATE (AND PULSE RATE)

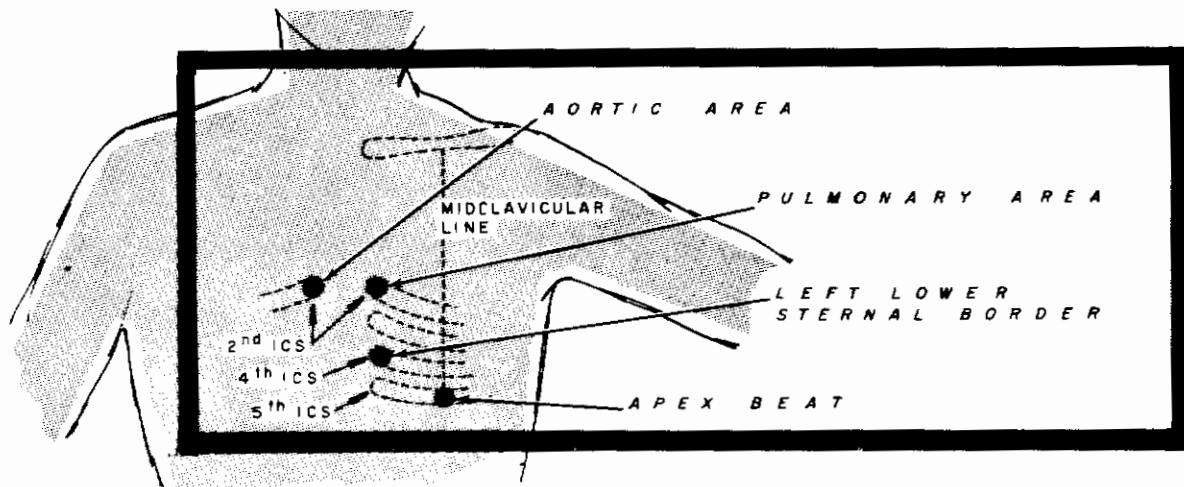


Figure 12. Location of the Phonocardiographic Microphone

The cardi tachometer consists essentially of a trigger circuit, and an integrating network or a digital counter to convert the analog input to a rate indication. The input is taken from ECG leads, or from a pressure or plethysmographic pickup. The output, depending upon the circuit details, will be proportional to either the heart rate itself or to its reciprocal, the interval between successive peaks. The display or recording may be discontinuous, yielding an instantaneous beat-by-beat rate indication, or it may be continuous, yielding a mean rate indication over a controlled period of time.

I. Sensors

The heart rate input may be derived from any of the sensors discussed under plethysmography (page 50) or electrocardiography (page 22). The selection of a particular sensor depends upon the needs of the measurement situation. For example, if heart rate rather than ECG data are of primary interest, a sternal ECG lead yielding strong R peaks would be preferred to a modified limb lead, which might be preferred for certain kinds of ECG data.

II. Signal Conditioning

Typically, a cardi tachometer consists of an input sensor, a differential input pre-amplifier, the tachometer unit proper, and a meter display or a high-impedance graphic recording device. There are numerous techniques for conditioning the signal in the cardi tachometer unit, depending upon the type of display or recording that is desired. (See ref. 37.)

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A. Continuous Cardiometer

A continuous or integrating cardiometer produces a d-c output that is linearly proportional to the integrated rate of the input pulses. Figure 13 shows a circuit used for this type of measurement. The input from a preamplifier is capacitor-coupled to T1, an emitter follower, and then is fed to a monostable flip-flop, T2-T3. The flip-flop produces pulses of constant duration, which are applied through T4 to an integrating filter network. From the filter network, the pulses are applied to emitter-follower T5, from which the output is taken. The d-c voltage across the filter network is the integral of the current flowing into it; therefore, the voltage at the output of T5 is proportional to the rate of the pulses from T2-T3.

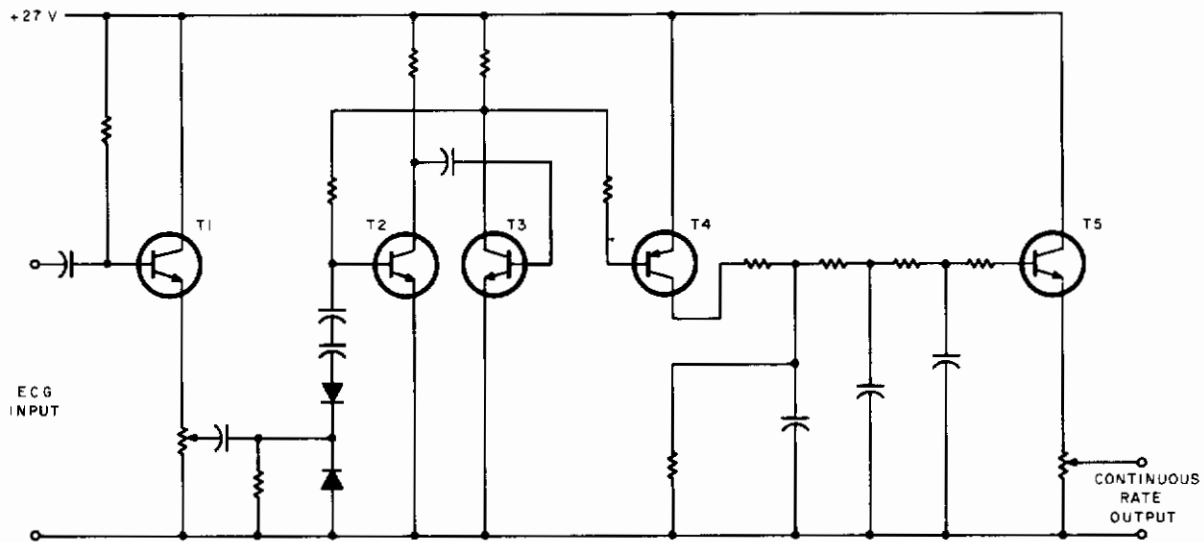


Figure 13. Circuit for an Integrating Cardiometer

B. Instantaneous Cardiometer

An instantaneous or discontinuous cardiometer produces a discrete output at each beat of the monitored signal. In most configurations, this type of output is proportional to the interval between beats or the reciprocal of the rate itself. The output is essentially hyperbolic with respect to rate; calibration is required to achieve linear recording or display.

Figure 14 shows an output circuit used for instantaneous measurements. (The input stages are similar to the integrating cardiometer shown in figure 13.) Here, the output from the flip-flop stage is used to close relay SR-1 momentarily. Between pulses, capacitor C1 is charging through resistor R1. During the momentary closing of the relay, capacitor C2 charges to the voltage on C1. When the relay reopens, C1 discharges through T5 and then recharges through R1. The output, determined by the

HEART RATE (AND PULSE RATE)

amount of charge taken by C2, is proportional to the time between pulses or the charge time on C1.

Figure 15 depicts an integrated and an instantaneous cardiogram. The integrated waveform lags the instantaneous trace, since its response to a step change is necessarily delayed by several pulses, but it gives a better indication of the trend of rate variations.

C. Digital Readout

Another form of instantaneous cardiogram uses counting circuitry to provide a beat-by-beat digital output for digital display or printout. A block diagram of this type is shown in figure 16. Other types simply total the number of beats for a given sampling period (1 minute, for example, as cited in ref. 38).

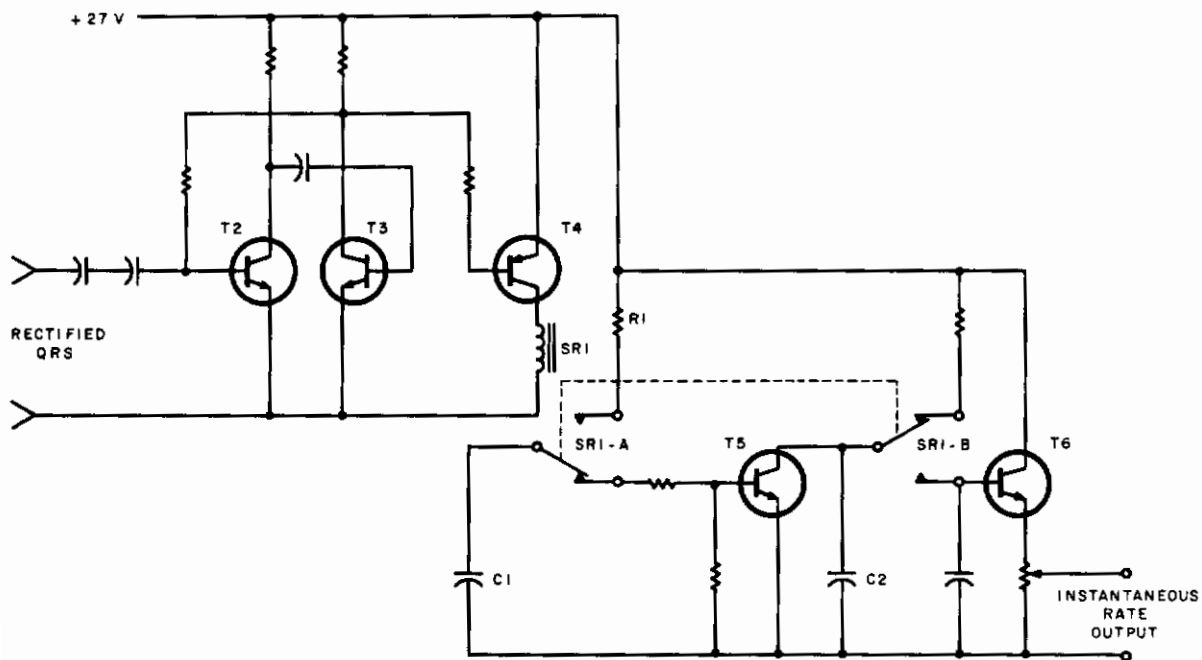


Figure 14. Output Stages for an Instantaneous Cardiometer

III. Notes on Operation

A. Calibration

Fundamentally, the method of calibrating the cardiometer is straightforward. A known input from a slow pulse generator circuit (30-240 pulses per minute) is applied to the input, and the sensitivity is adjusted (either in the cardiometer unit proper or in the recorder) for a corresponding indication. The chief difficulty is

INDIVIDUAL MEASUREMENT SYSTEMS

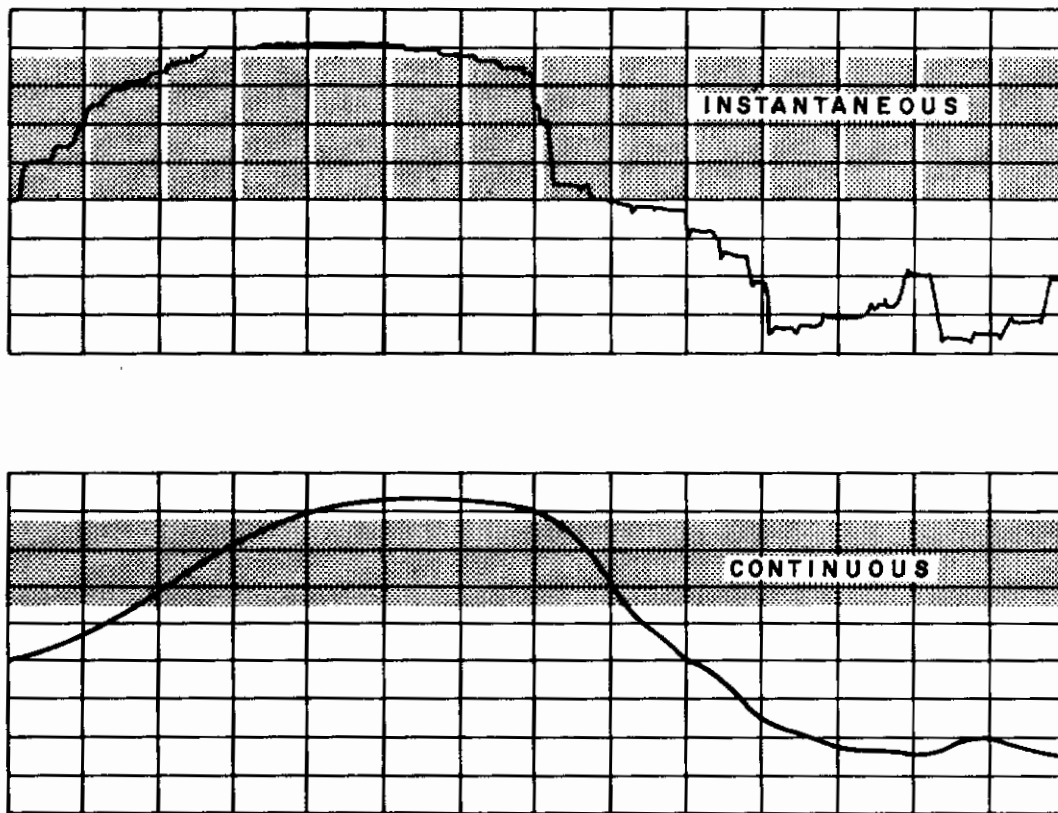


Figure 15. Instantaneous and Continuous Cardiograms

in the inherent nonlinearity of those tachometer circuits which are time or interval sensitive rather than rate sensitive. To provide linear scalar indications or traces, non-linear circuit elements must be included in the tachometer and the scale must be compressed.

B. Recording

The heart rate signal can obviously be recorded effectively at the same chart speed (25 millimeters per second) used for the ECG. When rate information alone is being recorded, however, chart speeds as low as 10 or even 5 millimeters per second may suffice, particularly if a compressed indication of rate trend is desired.

There are several variations of recording technique employed in the commercially available instantaneous cardiographs. Some follow the mean rate variation in an analog fashion, as shown in figure 15. Others give a beat-by-beat deflection with the stylus, returning to a baseline after each beat. In this type, the amount of deflection is proportional to the measured interval between beats, so the envelope of the tracing shows the rate trend as a function of interval; if the recording is to be calibrated in beats per minute, the amplitude scale will be inverted.

BALLISTOCARDIOGRAPHY

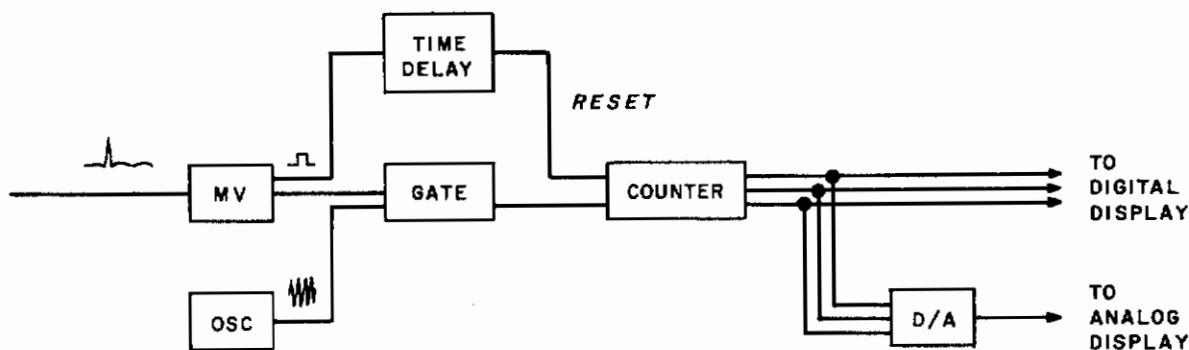


Figure 16. Block Diagram of a Digital Cardiometer

BALLISTOCARDIOGRAPHY

Ballistocardiography is another technique for studying cardiac activity. Basically, it consists of the graphic registration of extremely small ballistic reactions of the body caused by heart contractions and by the movement of blood in the large vessels. It gives a picture of the force, regularity, and velocity at which blood is ejected from the ventricles into the large arterial bodies, and a picture of the filling of the heart during diastole.

Heretofore, ballistocardiography has been chiefly a laboratory method. The principal apparatus (the high- and low-frequency seismic tables of Starr and Nickerson) have been too unwieldy for widespread use. With the development of practicable direct ballistocardiography with simple, portable apparatus, general clinical application is becoming possible. (See ref. 17.)

While not yet a technique suitable for continuous monitoring applications, ballistocardiography by direct methods will find more and more application in aerospace medical practice. Already a standard part of flight candidate medical examinations, ballistocardiography recently has been used to test the effects of pressure breathing, pressure suits, and weightlessness on cardiac activity.

1. Components of the Ballistocardiogram

A fundamental ballistocardiographic (BCG) tracing is shown in figure 17. The characteristic pulses represent headward and footward movements accompanying one cycle of cardiovascular action (headward movements cause positive deflections; footward movements cause negative deflections). At the start of systole, there is a small headward movement, H. At or just after ejection, as blood goes up in the aorta, there is a footward movement, I. Then, as blood turns down in the aortic arch, there is a headward recoil of the body, J. A footward movement, K, accompanies deceleration of the downward flow in the aorta. The smaller following diastolic movements, L, M, and N, are not as clearly defined as to origin.

INDIVIDUAL MEASUREMENT SYSTEMS

Despite the use of many different measurement techniques, there is general agreement on the main H-I-J components of the BCG waveform, and the BCG has become a reliable clinical index. Since it is a sensitive technique, the BCG is useful in diagnosing conditions with subtle effects that do not appear initially in other cardiac indices such as the ECG (ref. 9).

II. BCG Transducers

Ballistocardiograms can be obtained with different types of transducers. Displacement measurements are possible with piezoelectric transducers; dynamic or moving coil transducers yield velocity tracings; and acceleration tracings can be obtained from accelerometers, such as loaded piezoelectric devices, or by differentiating the output of velocity transducers. Some users have suggested that valuable measurements can be obtained in terms of jerk, the time differential of acceleration.

A typical transducer is the dynamic or moving-coil type used in the Dock type of direct ballistocardiograph. In the Dock method, two coils are mounted on a bar, which is firmly attached to the shins of a reclining subject. A permanent magnet is mounted on a stand (not attached to the legs) that fits in a space between the two coils. The legs of the subject are supported at the heels, as shown in figure 18. With the shin bar firmly attached to the shins, slight body movements cause the coils in the bar to cut the lines of force of the magnet, producing the output voltage. Standard parameters for direct measurement of this sort are as follows (ref. 4):

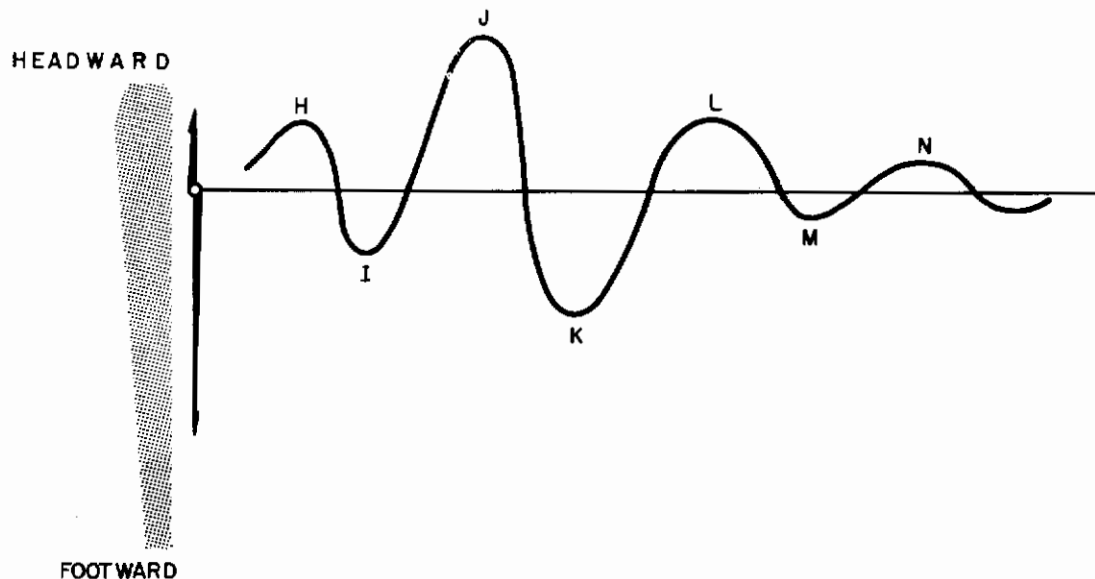


Figure 17. A Typical Ballistocardiogram

BALLISTOCARDIOGRAPHY

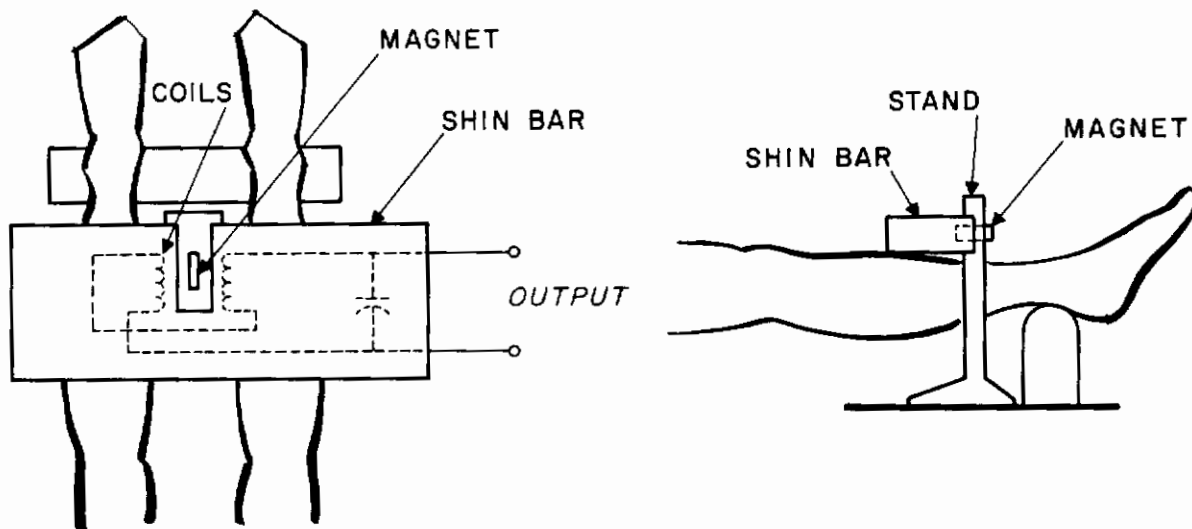


Figure 18. Dock-Type Direct Ballistocardiograph

- Frequency of normal body vibrations: 4 cycles per second.
- Displacement: 0.006 centimeter.
- Velocity: 0.1 centimeter per second.
- Acceleration: 2 centimeters per second².

III. Signal Conditioning

A. Filters

Filter networks are used in BCG circuits for two basic purposes: one is to remove vibrational artifacts above the frequencies of interest; the other is to operate upon the transduced output to obtain recordings of the parameter of interest -- either displacement, velocity, or acceleration (ref. 9). Depending upon the type of transducer used, a combination of filters can be employed which, by either integrating or differentiating the signal, will yield the wanted measurement. For example, if displacement is given by the transducer, differentiate once for velocity or twice for acceleration; if velocity is given, integrate for displacement or differentiate for acceleration; if acceleration is given integrate once for velocity or twice for displacement; and for jerk, differentiate displacement three times, velocity twice, or acceleration once.

INDIVIDUAL MEASUREMENT SYSTEMS

B. Amplifiers

A suitable BCG amplifier should be linear in the region of 1 to 30 cycles per second. Overall sensitivity, including the recording device used, should yield a 10-millimeter deflection with a 1-millivolt signal.

IV. Recorders

Once transduced and properly conditioned, the BCG signal can be recorded conveniently on any device suitable for ECG or EEG recording. A multichannel recorder should be used if available, since correlate ECG data should be recorded at the same time, as well as other cardiovascular indices of interest.

PLETHYSMOGRAPHY

An instrument called the plethysmograph is used to obtain a measurement of blood flow indirectly without exposing or puncturing the blood vessels. The plethysmograph senses changes in volume in a body region (limb, finger, etc) as a result of the flow of blood through that region. There are four basic plethysmographic techniques: photoelectric, impedance, segmental, and pneumatic.

Presently, the plethysmograph is more suited for use in the clinic rather than in a hostile environment. Instantaneous volume readings can be rendered meaningless by small body movements, quite aside from the artifacts generated by moving the transducer itself. Of the four plethysmographic techniques, the photoelectric is probably least susceptible to movements; unfortunately, the photoelectric type is one of the more difficult to calibrate in terms of exact volume values.

In general, the use of plethysmographic techniques in the field is fine for sensing arterial pulses or for recording pulse pressure waveforms, but blood volume or blood pressure indications so obtained are more relative than quantitative.

I. Photoelectric Plethysmograph

The photoelectric plethysmograph detects changes in transparency in a body region as a result of blood flow through that region, and produces an output related to the volume of blood flowing. Present devices usually house the light source and the photo-detector side by side in a small unit, which is placed over the area of interest. Most commercially available photoelectric plethysmographs are designed for clipping to or slipping over the fingertip; the transducer also can be attached with elastic straps or adhesives directly over a major artery, such as the temporal, brachial, or radial.

Figure 19 shows a typical circuit arrangement for the photoelectric plethysmograph (ref. 29, 30). The transducer may be either a photoconductor or a photovoltaic cell. The values of R1, R2, and R3 in the bridge are chosen to provide an output suit-

PLETHYSMOGRAPHY

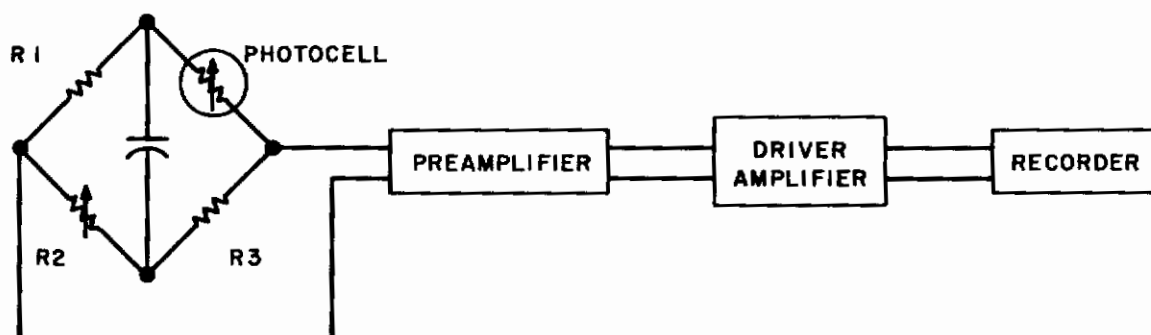


Figure 19. A Photoelectric Plethysmograph

able for transmission or recording. R2 serves additionally as a calibrating potentiometer. The bridge is dc-coupled to the preamplifier; typically, it consists of at least one push-pull gain stage (20X), followed by a low-impedance emitter follower output.

The output of the preamplifier is dc-coupled (as shown) to a driver amplifier and recorder. Alternately, the preamplified signal may be fed to a voltage amplifier stage for line or r-f transmission.

The use of a photoconductor in such a circuit has the advantage of very small size and a low impedance when the unit is illuminated. However, photoconductors have a high temperature coefficient, and they should not be used if large changes in temperature are part of the measurement environment.

II. Impedance Plethysmograph

The impedance plethysmograph is another device for obtaining flow measurement data by means of a calibrated blood volume pulse. The volume pulse is obtained by impressing a low-level r-f signal across a portion of the body and measuring the changes in this signal as it is affected by changes in body impedance. The impedance changes are proportional to the volume of blood in the region at the instant of measurement. Blood flow may be estimated in several ways; i.e., by multiplying the area under the volume pulse by the heart rate. (See ref. 22.)

This measurement routinely is made at the finger. A pair of metal ring electrodes is placed on the finger, usually one on either side of the second knuckle. A low-level current of no more than a milliampere is applied to the electrodes from an r-f oscillator operating at about 175 kilocycles; the r-f output is applied to a tuned amplifier; and impedance changes between the electrodes modulate the input to the amplifier. The amplifier output then is demodulated, providing the d-c analog of the volumetric changes.

Like the other plethysmographic techniques, the impedance plethysmograph is not

INDIVIDUAL MEASUREMENT SYSTEMS

instrumented readily for aerospace applications. Subject movements invalidate volumetric data easily, calibration and subsequent data processing are tedious, and stable electrode placement on the finger is difficult. For further information on the impedance plethysmograph, refer to standard texts (ref. 55 and 56). Refer also to the discussion of the impedance pneumograph (page 88), a measurement technique which is, in part, an outgrowth of the recurrence of respiration artifacts in plethysmographic measurements.

III. Segmental and Pneumatic Plethysmographs

The segmental plethysmograph measures volumetric changes in a digit or limb by means of a strain gage, usually a mercury-in-rubber type strain gage. Volume is measured as a function of changes in girth resulting from changes in blood flow in the measured member. The strain gage is placed in a bridge circuit for measurement. (Refer to Volume II, pages 23-24, for an illustration of the technique of matching the output of the mercury strain gage to a conventional strain gage preamplifier and direct-writing recorder.)

While the segmental plethysmograph can yield data on a large number of volumetric parameters (e.g., vascular capacity, blood flow, effects of g forces), its usefulness as a source of quantitative data in a monitoring application is limited (ref. 29, 30).

The pneumatic plethysmograph is perhaps even less adaptable than the above-mentioned instruments to use outside the clinic or laboratory. True volumetric readings are obtained with a pressure plethysmograph transducer, which senses the change in pressure in a closed air chamber as a result of the pulsatile change in volume of a digit inserted into that chamber. Such a measurement can be extremely accurate when it is taken on a subject with a properly sensitive device in rigidly controlled conditions. The stressful environments and subject movement anticipated in most aerospace applications preclude its use there, but it can be used if the subject is quiet during the period of measurement.

BLOOD PRESSURE

The pressure of the blood in the great arteries is one of the most significant indexes of cardiovascular function. Blood pressure is determined by many interacting factors, including blood viscosity and the elasticity and diameter of blood vessels. Most important, as a measure of subject viability, blood pressure is related to the volume of blood in the vascular system and to the cardiac output. Like the electrocardiogram, the measurement of blood pressure is a standard clinical technique with well established norms, and as such it is generally considered one of the more valuable and necessary parameters for an aerospace monitoring system.

The pressure-measuring systems devised for monitoring systems are necessarily indirect systems, since the only direct method of measuring pressure is by arterial

puncture and the use of intra-arterial catheters. The following pages describe the various indirect measuring systems that have been employed with any degree of success. The principal type used is the autosphygmomanometer, which is the field extension of the familiar technique of auscultation used in the clinic. (Refer also to ref. 65.)

I. The Measurement Technique

Two pressure measurements are of significance: the systolic and the diastolic. The systolic blood pressure is the maximum pressure that occurs with the ejection of blood into the aorta following ventricular contraction. Diastolic pressure occurs during ventricular diastole, with the recoil of the blood in the arteries. (Refer to Volume I, pages 18-31.) In indirect measurements, the air pressure in an occluding cuff is monitored during a controlled cycle of inflation and deflation, and the systolic and diastolic points are determined by the coincidence of pulse indications from a transducer placed over the occluded artery. These pulse indications serve the same function as the Korotkov sounds in standard auscultation.

The usual site for the cuff and pulse transducer is over the brachial artery on the upper arm. At this location, with the arm down, the pressures recorded are the same as those at the heart, being at the same level.

In a similar technique, an occluding cuff is applied to a finger, and the pressure readings are correlated with volume pulses obtained from a plethysmographic transducer attached to the fingertip. In this type of measurement, compensation must be made for the lower pressure readings which are obtained at the extremities.

Because of the blood volume-blood pressure relationship, attempts have been made to obtain indirect blood pressure data using the plethysmographic pickup alone, without an occluding cuff. The pulse volume tracing obtained by such means is similar to the pulse pressure waveform obtained by direct intra-arterial measurement. By calibrating the pulse volume waveform against baseline pressure data, an index of corresponding blood pressure can be obtained; but the exactness of quantitative data so obtained is questionable.

II. A Basic Autosphygmomanometer

There are several approaches to automating the auscultatory blood pressure measurement technique, but they all have certain components and methods in common. A typical device is that developed by the Air Force's School of Aerospace Medicine, shown diagrammatically in figure 20. Hundreds of successful inflight measurements have been made with this system in high-performance aircraft; the pressure levels and transducer outputs are relatively immune to high g stresses and to artifacts from subject movement, ambient noise, and vibration. (See ref. 74.)

The basic components in this system are the occluding cuff, a source of air, and a

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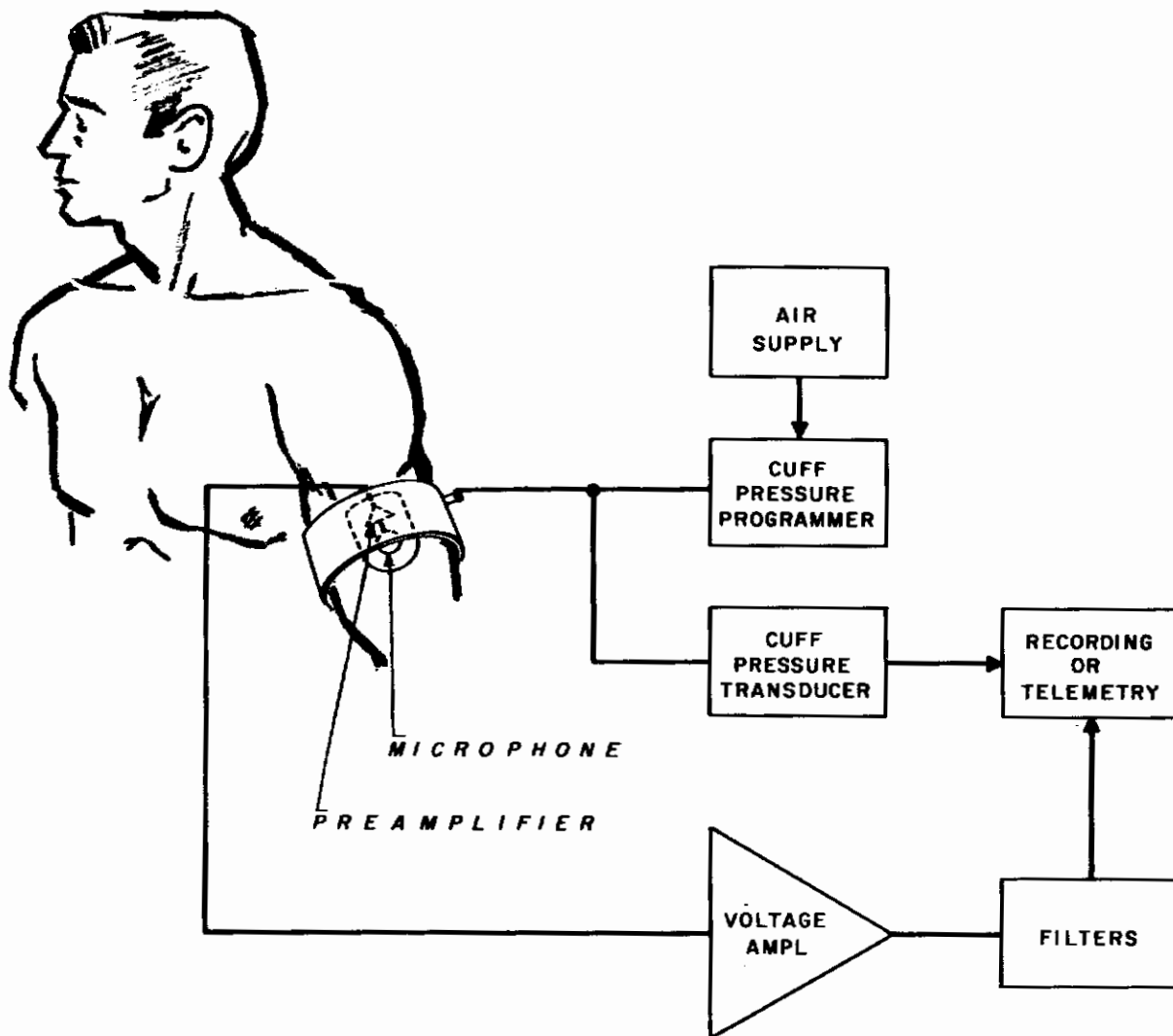


Figure 20. An Autosphygmomanometer for Blood Pressure Measurement

programmer to control the pressure in the cuff; a transducer to measure the cuff pressure; a microphone to detect arterial pulses; a microphone preamplifier and voltage amplifier; and bandpass filters to condition the amplifier microphone output. The microphone and transducer outputs then are applied to a telemetry system or to recording and display components.

The cuff pressure is recorded as the regulator cycles; the cuff is inflated rapidly to a pressure well above anticipated systolic pressures and then allowed to bleed off to a point below diastolic pressure. Simultaneously, the microphone output is recorded on separate channels. (The microphone output is split by bandpass filters into two or more frequency bands to assure a readable pulse output even in the presence of noise artifact.) As the cuff pressure falls below the level necessary for full occlusion, the first

BLOOD PRESSURE

pulse is registered, which marks the systolic point. Pulses continue to register until the cuff pressure is too low to cause any change in the pulse sound; the last pulse to appear marks the diastolic point. Figure 21 shows a typical recording of this type.

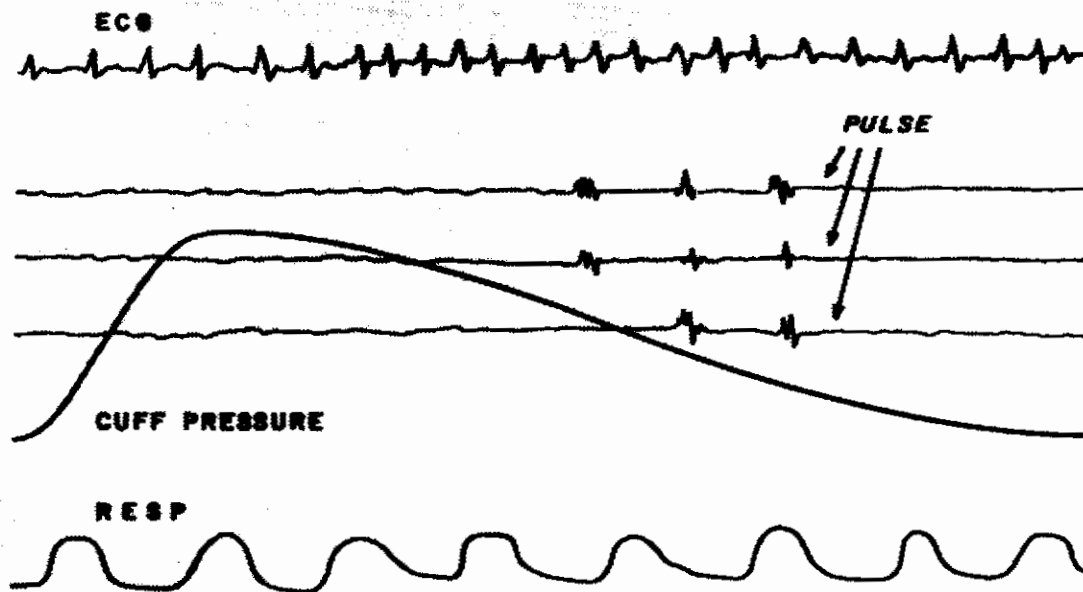


Figure 21. Multichannel Recording Showing Simultaneous Trace of Cuff Pressure and Pulses

The following paragraphs discuss the various components of this system in more detail.

A. Programmer

The programmer used in the system shown in figure 20 is an electromechanical bypass-type pressure regulator. Using low-pressure nitrogen available within the airframe, the programmer delivers a sawtooth pressure profile, controlled by a cam on a timing motor. This unit is designed for 1-minute sampling cycles. It inflates the cuff from 0 to 225 mm Hg in 3 seconds, leaks pressure at a linear rate to about 20 mm Hg in 25 seconds, dumps pressure from 20 to 0 mm Hg in 2 seconds, holds at zero pressure for 30 seconds, and recycles. Pressure levels are undisturbed by sudden changes in cuff volume caused by arm movement; gas that is bypassed is vented to the cabin or suit so that the cuff pressure is always read with respect to ambient pressure.

B. Cuff Pressure Sensor

The pressure transducer used in the School of Aerospace Medicine system is a differential type. A Sanborn diaphragm-type strain gage transducer is used in ground

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checks, providing a millivolt output.* For inflight measurements, a Daystrom-Wiancko transducer, Model P2-3136 is employed.** This is a pressure transducer of the Bourdon tube type, using a variable reluctance bridge circuit. It operates from 28-volt aircraft power, and delivers a high-level, 0-5 volt output into a 50-kilohm load.

C. Microphone

The pulse sensor in the system is a Gulton Model MP-202 piezoelectric contact microphone.*** Contact microphones of this type are essential in high-level ambient sound environments that exist in high performance aircraft. This particular model has the standard mounting disc removed, and it is mounted in an arm pad fabricated of silicon rubber, designed to be slipped between the occluding cuff and the upper arm. The microphone button protrudes from the inner surface of the arm pad in order to sense the arterial pulse properly. Nominal distance is 1 millimeter above the pad, but this may vary, depending upon the contact pressure required for optimum output with a particular subject.

D. Microphone Preamplifier

The piezoelectric microphone presents a high source impedance of about 10 megohms. A two-transistor preamplifier was designed to load the microphone properly (input impedance of 50 megohms) and retain good low-frequency response. The output impedance of the preamplifier is 10,000 ohms. The preamplifier and the small mercury cells required for its operation are potted with the microphone in the silicon rubber arm pad. Figure 22 shows the circuit for the preamplifier.

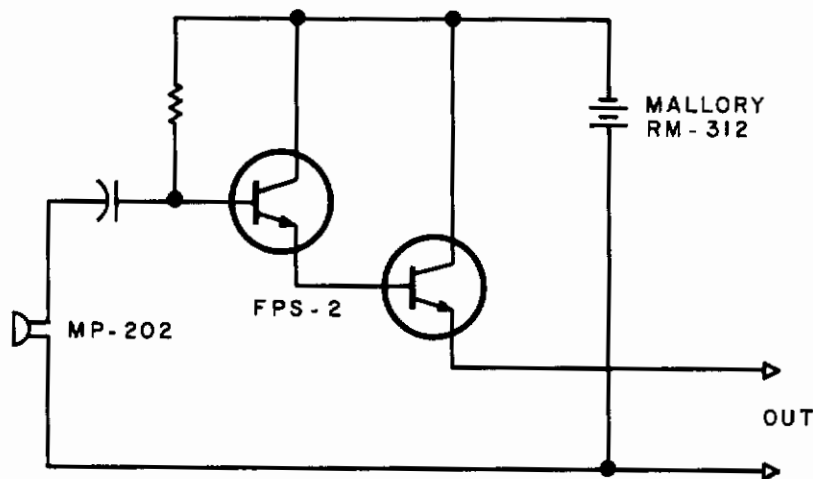


Figure 22. Microphone Preamplifier Used in Autosphygmomanometer (Fig. 20)

* The Sanborn Co., Waltham, Mass.

** Daystrom-Wiancko Engineering Co., Pasadena, Calif.

***Gulton Industries, Inc., Metuchin, N.J.

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E. Voltage Amplifier

The voltage amplifier used in the system is a Taber Model 196G.* It operates from the 28-volt aircraft power, and provides a gain of 200 to 1000. The input, as shown in figure 23, is shunted by a 100-kilohm resistor; this completes the ground return for the batteries in the preamplifier (figure 22). Therefore, when the unit is not in use, the connection between the arm pad and amplifier can be opened, and there is no drain on the preamplifier batteries.

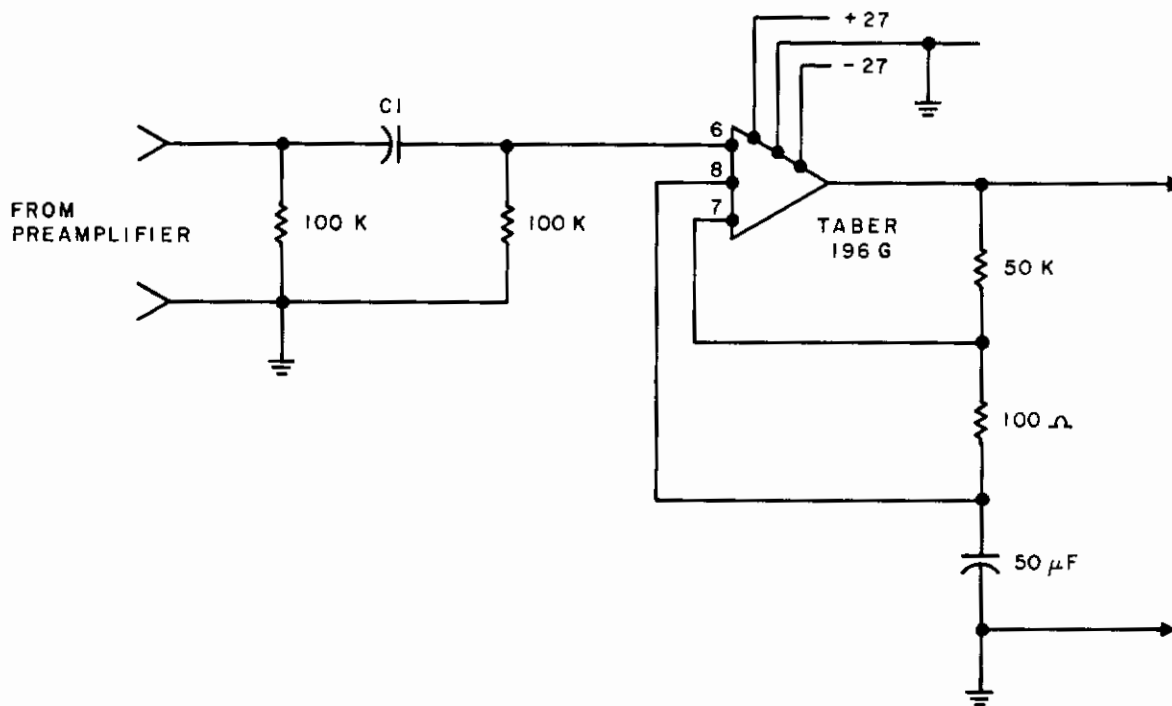


Figure 23. Voltage Amplifier Connections Used in Autosphygmomanometer (Fig. 20)

F. Filters

Narrow-bandpass filters are used to counter the effects of high-level ambient noise. Two or more filters are used, each centered at different, harmonically unrelated frequencies in the range between 50 and 250 cps. At least one channel will deliver recognizable pulses, even with ambient noise present (up to 125 db). The frequencies below 50 cps are removed to avoid ambiguity in the reading of the diastolic point; frequencies above 250 cps are not necessary for good registration, and are beyond the effective response of most recorders anyway. Figure 24 shows a typical three-channel filter configuration.

* Taber Instrument Corp., Tonawanda, N. Y.

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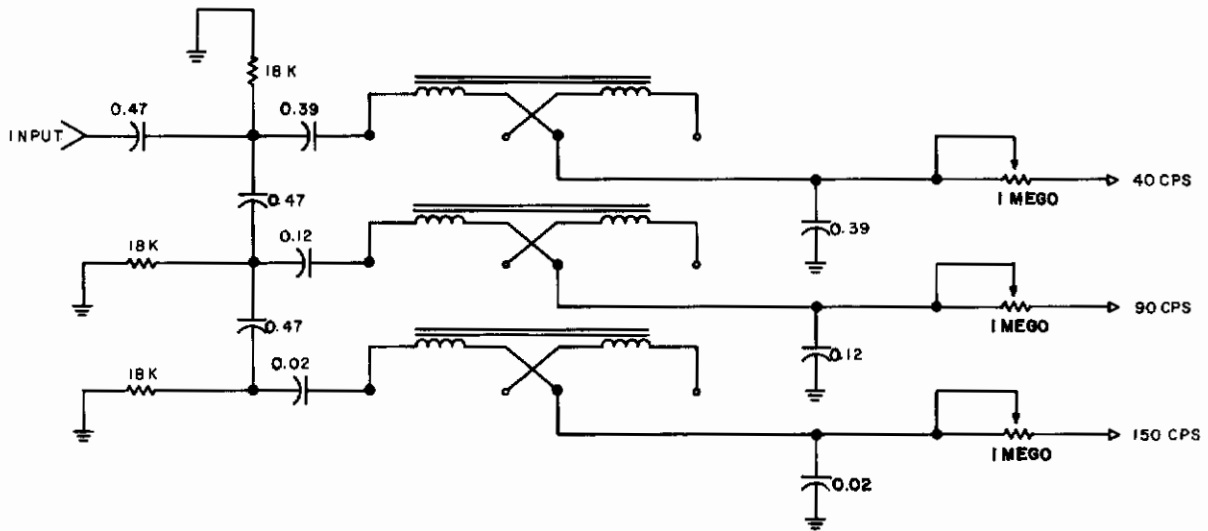


Figure 24. Microphone Filter Network Used in Autosphygmomanometer (Fig. 20)

III. Other Measurement Techniques

The foregoing description of the School of Aerospace Medicine measurement system is typical of the needs of an automatic blood pressure monitor of the auscultatory type. Similar systems have been devised with variations in detail to suit individual applications. Some of these are noted below.

A. Other Methods of Pulse Registration

A system has been developed at the Aero Medical Laboratory at Wright-Patterson Air Force Base in which cuff pressure is monitored and recorded as described previously, but several different techniques of monitoring the arterial pulse are employed (ref. 80). Initially, a water-filled plastic bag was inserted beneath the cuff, and pulses were detected with a pressure transducer. Occasional leakage of the fluid system in some applications prompted the replacement of this type of pulse sensor with a strain gage, which is mounted on a flexible metal plate and placed directly over the brachial artery. With a strain gage in the system, a d-c or chopper amplifier is needed to amplify the pulse signal. Another approach to the problem was to use an extremely sensitive transducer to monitor the cuff pressure variations so that arterial pulsations were sensed directly by the cuff monitor and indicated directly on the cuff pressure tracing. Such a system is of course equally sensitive to movement artifacts.

A system designed at NASA's Ames Research Center is quite similar to the School of Aerospace Medicine device. The chief difference between the two is that the output of the microphone in the NASA system, after being amplified and filtered

BLOOD PRESSURE

(bandpass from 15 to 200 cps), is added directly to the output of the cuff pressure transducer for single-channel transmission and recording. The pulses appear on the recording as blips on the sawtooth cuff pressure waveform. (See ref. 64.)

B. Coincidence Gating

An accessory to the basic autosphygmomanometer system has been designed by the School of Aerospace Medicine for the purpose of obtaining digital readout of blood pressure data (ref. 38). (Refer to figure 25.) In this system, the output of the pressure transducer is read out intermittently to obtain a pressure reading only at the time of arterial pulsation. To discriminate against artifacts from ambient noise or motion, the microphone output is passed through two bandpass filters immediately after preamplification, and coincident pulses are obtained at harmonically unrelated frequencies. The signal from each filter is applied to an impedance-matching emitter follower, a diode envelope detector, and a trigger circuit (one-shot multivibrator). Both triggers then are passed to an AND gate. The pulse from the gate is used to gate the output of the cuff pressure transducer into a digital voltmeter (or into recording or telemetry circuits).

An autosphygmomanometer made by Systems Research* includes an additional refinement. This system uses logic and timing circuits to program cuff inflation and deflation, and to read out the cuff pressure only at the systolic and diastolic points after redundancy and coincidence checks have been performed on the pulse-sensing components. Figure 26 shows the components in this system.

Two pulse sensors, the microphone and the thermistor, provide a check against artifacts; both must be activated simultaneously before the AND gate passes the pulse indication. The thermistor senses arterial pulses indirectly; arterial pulsations change the pressure in the cuff minutely, and with each pulse, a small portion of air is vented from the cuff across the thermistor, cooling it and producing a pulse indication.

A further check against artifacts, and a means of assuring a more accurate pressure indication, is obtained by a timing and logic circuit which requires second, confirming pulse indications at pressure points of interest. This system is illustrated in the description of the measurement sequence below.

1. Diastolic Pressure Measurement

The cuff is inflated rapidly to a point just below diastolic pressure, and then slowly inflated (2 mm Hg per second) until a pulse is obtained from the sensor AND gate. At this point, inflation stops and a 1-1/2-second timer is activated. If a second pulse passes the sensor gate during this interval, the cuff pressure is read out; if no pulse is received, the cuff deflates and the measurement is repeated.

*Systems Research Laboratories, Inc., Dayton, Ohio

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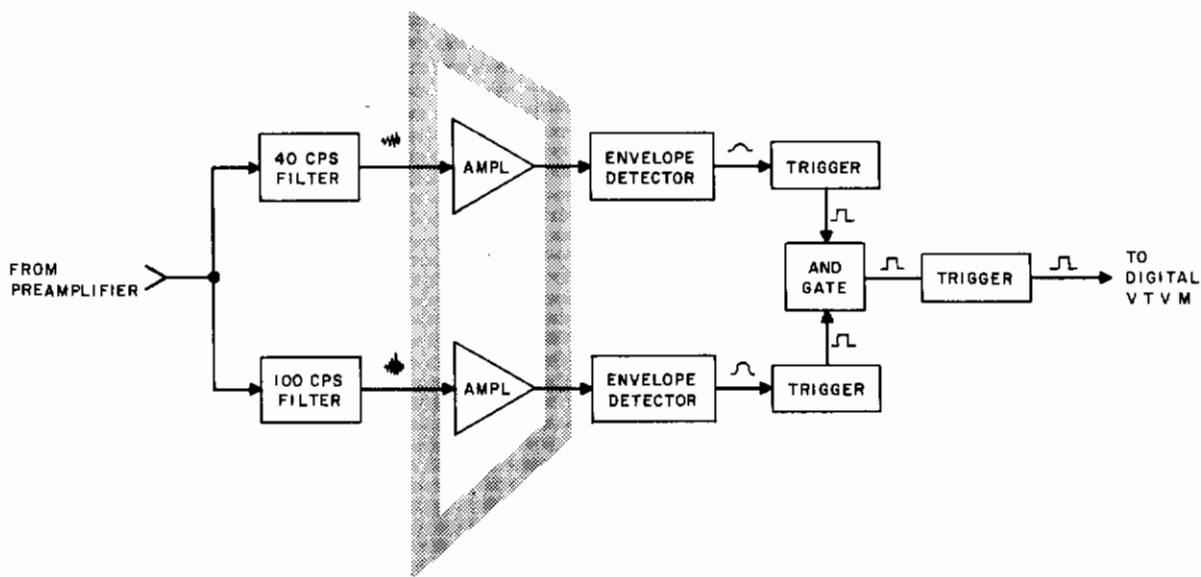


Figure 25. Circuit for Digital Readout of Autosphygmomanometer

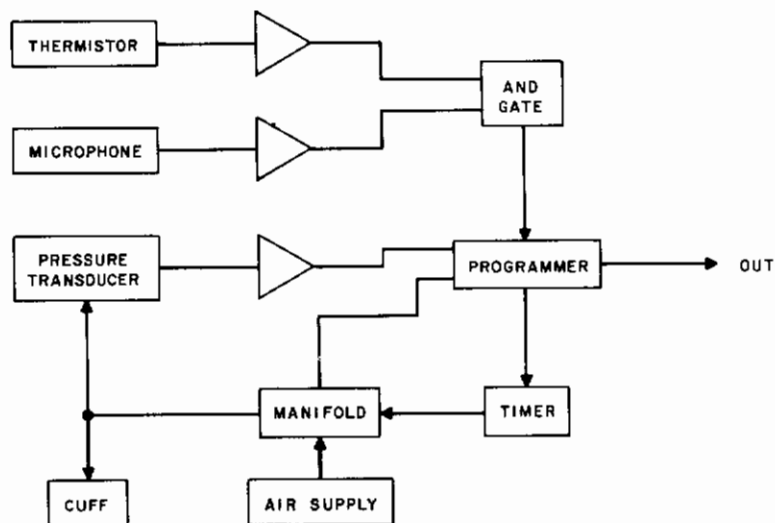


Figure 26. Configuration of a Coincidence-Gated Autosphygmomanometer

PULSE WAVE VELOCITY

2. Occlusion

After diastolic pressure is read out, the cuff is inflated rapidly to a pressure just above systolic, to occlude the artery. The timer is activated, and the cuff inflates slowly. If a pulse is received, the timer resets and inflation continues until no pulse is received during the 1-1/2-second interval.

3. Systolic Pressure Measurement

When occlusion has been verified, the timer is deactivated and the cuff deflates slowly until a pulse is received from the sensor gate. At this point, deflation stops and the timer is reactivated. If a second pulse occurs during the 1-1/2-second interval, the cuff pressure is read out; if no pulse is received, the pressure is returned to above systolic, and the measurement is repeated.

The system normally obtains diastolic and systolic readout in the time required for 7 to 10 heartbeats. Consequently, whole or partial arterial occlusion occurs for about 6 seconds in each cycle, and venous occlusion for about 15 seconds in each cycle. The entire system, as packaged by Systems Research, fits in a light-weight backpack measuring 11 x 9 x 3 inches. With the addition of a transistorized FM transmitter and a small 28-volt power pack, short-range telemetry of blood pressure data from an ambulatory or working subject is possible.

PULSE WAVE VELOCITY

Pulse wave velocity refers to the velocity of propagation of the pulse pressure wave through the arterial tree. While this parameter is of clinical interest in its own right, recently it has been investigated as a means of obtaining an indirect index of overall cardiac function, and particularly of blood pressure. Blood pressure can be measured successfully with the automatic sphygmomanometers described previously, but only on an intermittent basis and not continuously. Techniques of measuring pulse wave velocity, while not yet perfected, do permit continuous monitoring, which provides information on the overall tone of the vascular system (arterial distensibility); this information in turn can be related to the systolic blood pressure.

1. The Pulse Wave Phenomenon

When the ventricles contract, forcing a volume of blood into the aorta, two quite different events occur: one is the flow of blood through the arterial tree, at a velocity of 0.2 to 0.6 meter per second in the larger arteries and at a much slower velocity in the smaller vessels; the other is an expansion of the vessel walls, occurring first in the aorta and then travelling out along the peripheral blood vessels as a result of a pressure wave transmitted in the blood. This is the pulse wave that may be detected at any convenient point on the surface over the arteries. The velocity of the pulse wave varies from roughly 5 to 15 meters per second (see table V).

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TABLE V. AVERAGE PULSE WAVE VELOCITIES BETWEEN VARIOUS ARTERIAL POINTS

Recording Sites		Velocity (Meters/Second)
From	To	
Aorta	Brachial artery	7.2 (5.4 - 8.5)
Aorta	Radial artery	8.8 (6.6 - 13.8)
Aorta	Femoral artery	6.3 (4.7 - 10.4)
Brachial artery	Radial artery	12.5 (9.3 - 22.5)

The pulse wave velocity is related directly to the rigidity of the arterial walls. An increase in internal (blood) pressure increases the rigidity of the walls; it follows that there is a corresponding, proportional increase in the pulse wave velocity. For example, for an increase of diastolic blood pressure from 100 to 150 mm Hg, the velocity of the pulse wave increases from a mean of about 9 to about 16 meters per second. Figure 27 shows this relationship.

Pulse wave velocities have been determined in the laboratory by several complex techniques. One of these is to apply a known blood pressure reading (obtained with an intra-arterial catheter and pressure transducer) to an analog computer which is set up to solve a general equation for fluid dynamics (ref. 36). For a continuous monitoring

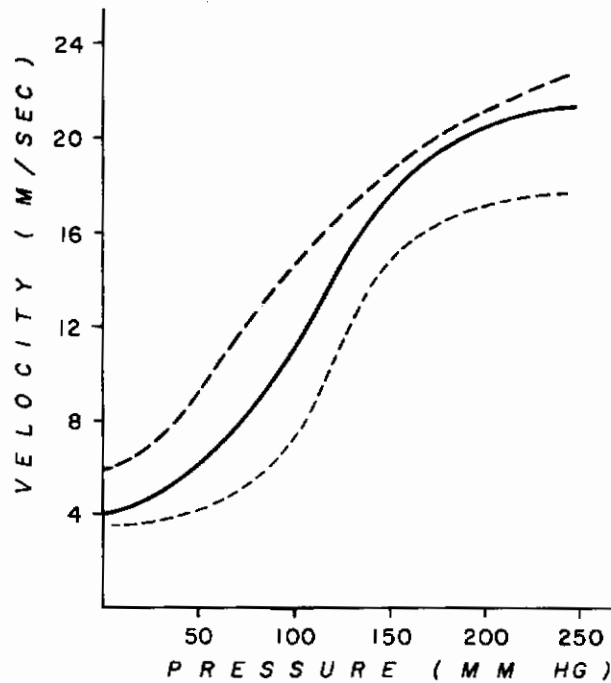


Figure 27. The Relationship of Pulse Wave Velocity to Diastolic Pressure

Contrails

PULSE WAVE VELOCITY

system, such blood pressure data are not available, but the measurement of velocity can be determined conveniently as a time interval measurement.

One method for measuring the time interval is to sense the pulse pressure wave at two sites on the body separated by a known distance, and measure the difference in arrival time of the pulse wave between the proximal and the distal site (ref. 21). The arrival time is always computed at the intercept of the diastolic point of the wave and the ascending anacrotic limb (figure 28) at the start of systole, since this portion of the waveform is most clearly defined on all recordings. Hence, the corresponding pressure at the time of measurement is diastolic. Another time interval measuring technique is to measure the elapsed time between the peak of the R wave on the ECG and the arrival of the pulse pressure wave at some point along the arterial tree.

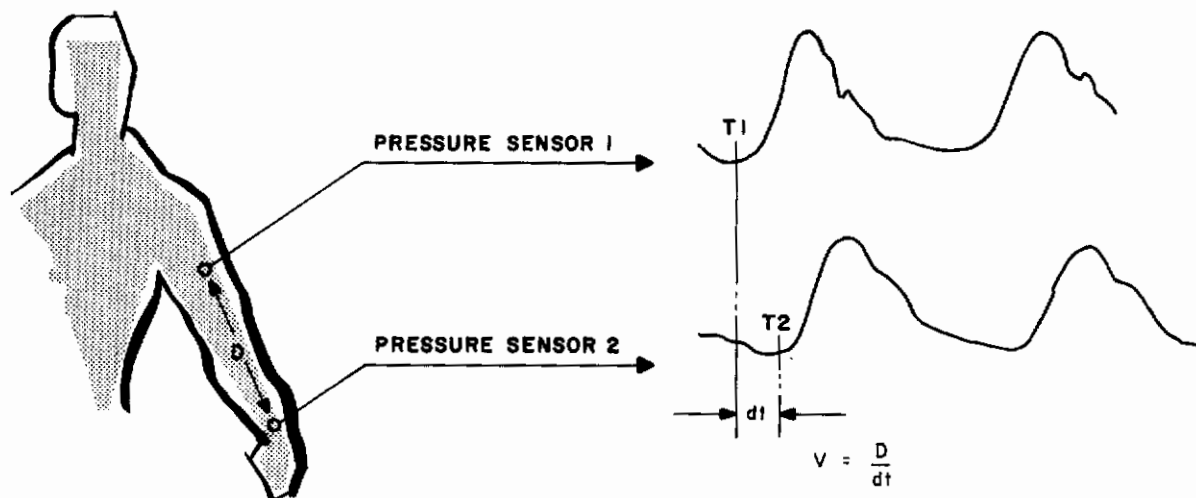


Figure 28. An Indirect Technique for Measuring Pulse Wave Velocity

II. Instrumentation

Pulse wave velocity data can be obtained by the manual processing of the recorded traces of two simultaneous pulse pressure waveforms, as indicated above, or by the manual processing of the simultaneous pulse-pressure and ECG waveforms. To obtain velocity data automatically, the sensor inputs must be processed in circuitry similar to that used in the cardiograph. This processing yields time interval data which can be calibrated in terms of velocity or, with sufficient empiric data, in millimeters of mercury (diastolic pressure).

To obtain quantitative pressure data, the relationship between pressure and velocity must be formulated. The expression,

$$P = 12.2 V - 130$$

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for example, holds true for the linear portion of the curve in figure 27 (between 50 and 150 mm Hg) when appropriate empiric corrections are applied. To instrument the pressure measurement in terms of a time interval (dt) measurement, two pulse pressure sensors are located on the subject a known distance, D , apart. The above expression is then modified as follows:

$$P = 12.2 D/dt - 130$$

The output of the appropriate time interval or tachometer circuitry, therefore, can be calibrated directly in mm Hg. The block diagram of figure 29 shows this arrangement. The pulse pressure wave arriving at pressure sensor 1 starts the digital counter. The arrival of the wave at pressure sensor 2 stops it.

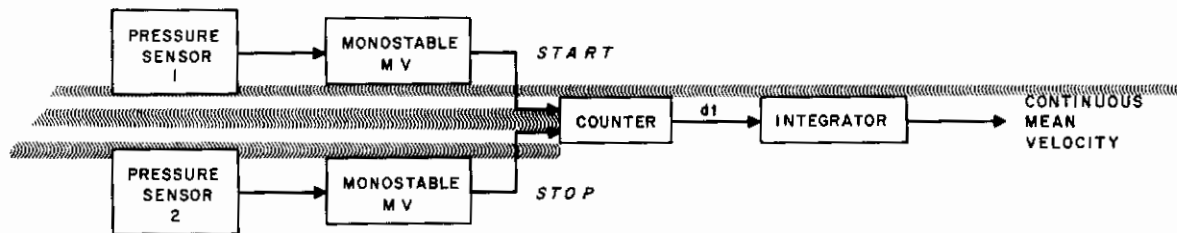


Figure 29. Components for Automatic Registration of Pulse Wave Velocity

Any suitable pulse detector may be used. The piezoelectric microphone described on page 56 is typical. If the R wave of the ECG is being used to trigger the time interval count, the ECG lead system should be one that provides a strong, well-defined QRS complex. Trigger circuits, integrators, counters, and recorders should be selected to provide the type of record of wave velocity desired (instantaneous, continuous, or digital) in accordance with the considerations on pages 44 to 46.

BRAIN POTENTIALS

The recording of potentials emanating from the human brain offers one of the few objective measurements of the activity of the central nervous system in man. A vast body of data has accumulated in the last generation which has been useful in the study and treatment of organic disorders, such as brain lesions and tumors, and diseases such as epilepsy. These data also have been used to draw correlates of the electrical or biochemical activity in the central nervous system showing a relationship between human brain potentials and psychophysiologic or behavioral responses. It is in this latter use that the study of the electroencephalogram has been of interest in aerospace medicine, because it affords a measurable index of the state of consciousness or alertness of a subject exposed to the stresses of aerospace environments.*

* Refer also to Volume I, pages 8-12.

As measured on the scalp, potentials from the brain usually are about 50 microvolts in amplitude; quite often they are less and rarely do they exceed 200 microvolts. Amplitude variations generally are nonperiodic, complex, and of low frequency (largely in the range of 1 to 60 cycles per second, but with some frequency components of several hundred cycles per second). When measured directly on the cortex of the brain rather than on the scalp, potentials are as much as 100 times greater in amplitude (up to 50 millivolts), since in this case there would not be signal loss through the thick bone of the skull. Such measurements, properly called electrocorticograms, are otherwise identical with the electroencephalogram in such characteristics as frequency and phase.

The electrocorticogram is by its nature restricted to the clinic in application, and even there it is rare, because of the difficulty and hazard in penetrating the skull. The electroencephalogram is the accepted and most easily instrumented method of monitoring brain potentials.

The electroencephalogram is obtained by placing an array of electrodes on the scalp, connected as independent bipolar pairs or as a variety of unipolar leads, using a central terminal and an exploring electrode much the same as is done in electrocardiography. There are, understandably, several similarities between electrocardiographic and electroencephalographic techniques. In both techniques, the recording of potentials requires the use of capacitor-coupled electrodes and differential input stages for the successful isolation of unwanted potentials. Both brain and heart potentials require considerable amplification; the brain potentials measured at the scalp, however, are about 1/20th the amplitude of the heart potentials measured on the body, so input EEG circuitry must be correspondingly more sensitive.

There are differences in brain and heart potentials which dictate differences in ECG and EEG systems. For one thing, the electroencephalogram is not repetitive as is the electrocardiogram; therefore, recordings must be obtained over greater periods of time to obtain meaningful data. In addition, the complex of potentials on the scalp because of the complex structure and function of the brain requires many more electrodes for meaningful recording and interpretation. As many as 18 electrodes may be switched together in various combinations, with simultaneous recording on as many as eight channels.

I. Components of the Electroencephalogram

The electroencephalogram is much more complex than the electrocardiogram, and not at all easy to typify. In general, biological sources of electrical potentials do not generate pure sinusoidal waves. But the electrocardiogram, reacting to the rhythmic beating of the heart, displays a regular, consistent pattern, marked by a monophasic P and T waves and a diphasic QRS complex (figure 3). Its components are measurable (tables II and III), and interpretation of these measurements follows long-established clinical procedures.

INDIVIDUAL MEASUREMENT SYSTEMS

The electroencephalogram is not so easily interpreted. Most EEG recordings do contain two basic rhythms (the alpha and beta rhythms discussed later). But some sources of EEG potentials have not as yet been identified, and the potentials therefore cannot be definitely attributed to specific physiologic or psychophysiologic functions in the central nervous system (ref. 1). EEG tracings nevertheless do contain data of proven clinical usefulness, and they may be described in terms of the individual characteristics of their waveforms.

A. Frequency

The electroencephalogram appears as a complex, nonperiodic waveform. It contains many frequencies with shifting phase relations and varying amplitudes. When subjected to frequency analysis, the electroencephalogram reveals a fairly broad spread of energies, between the frequencies of 0 and 100 cycles per second, with peaks in certain frequency regions. When analyzed visually, the EEG tracing appears to contain the following distinct frequency bands:

- | | |
|--------------------------------------|----------------|
| • Slow or delta rhythms | 0.5 to 3.5 cps |
| • Intermediate slow or theta rhythms | 4 to 7 cps |
| • Alpha rhythms | 8 to 13 cps |
| • Intermediate fast frequencies | 14 to 17 cps |
| • Fast or beta rhythms | 18 to 30 cps |
| • Very fast frequencies | 30 cps and up |

Note that these designations are largely a convenience; a given wave component may be designated as an alpha wave, or a 10-cps wave, because it is measurably about 0.1 second in duration. It may appear transiently in the overall tracing only once every second, and not necessarily as 10 cycles in a one-second interval (ref. 51).

B. Amplitude

While the average potential difference between electrodes in an EEG lead system is about 50 microvolts, EEG signals of interest range from as low as 2 microvolts to as high as 200 microvolts. Electronic circuits must be particularly noise-free to permit the successful recording of such low-amplitude potentials (ref. 10).

As a general rule, the slower waves of the EEG have the higher amplitudes. Common exceptions to this rule include fast spike potentials and the low-voltage slow waves of early sleep (ref. 51).

C. Phase

Components of the electroencephalogram may be characterized as either di-phasic (swinging both positive and negative with respect to a reference level) or mono-phasic (the excursions from a reference level are either all positive or all negative). The phase characteristic for a given lead or channel may change during a recording over a sufficiently long period of time.

D. Wave Groupings

The overall pattern of an EEG tracing may be characterized as follows:

- Regular - rhythmic or repetitive.
- Irregular - arrhythmic, with waves of dissimilar durations during a 1-second interval.
- Paroxysmal - high-amplitude bursts above background voltage level, possibly including large changes in frequency.

E. Signal Localization

While their sources and mechanics are not fully understood, brain potentials are known to originate in neuronal tissue. The spike potentials of individual neurons travel over the surface of the cortex (as well as through the deeper layers) so that the potential at any point on the scalp (as under an electrode) represents an average of the potential changes in a large number of neurons. The potential changes or EEG patterns vary from one region of the brain to another, which is the reason for the large number of electrodes and channels used in EEG recording.

1. Regions of the Brain

Recording channels and their associated electrodes are identified by the region of the brain whose potentials they sense. The eight principal regions comprise the four major lobes -- frontal, temporal, parietal, and occipital -- in the right and left hemispheres, respectively. Individual electrode locations within a region are then designated with an arbitrary system of numbers, as shown in figure 30. This drawing also shows the relatively indifferent points on the head that also are used in monopolar recording: the right and left ear lobes, the nose, the occipital protuberance, and the vertex. (See ref. 1.)

The interpretation of EEG tracings usually requires a comparison of data received from symmetrically placed electrodes (e.g., right frontal vs left frontal), with the presence or absence of symmetry in the two tracings being of clinical significance.

INDIVIDUAL MEASUREMENT SYSTEMS

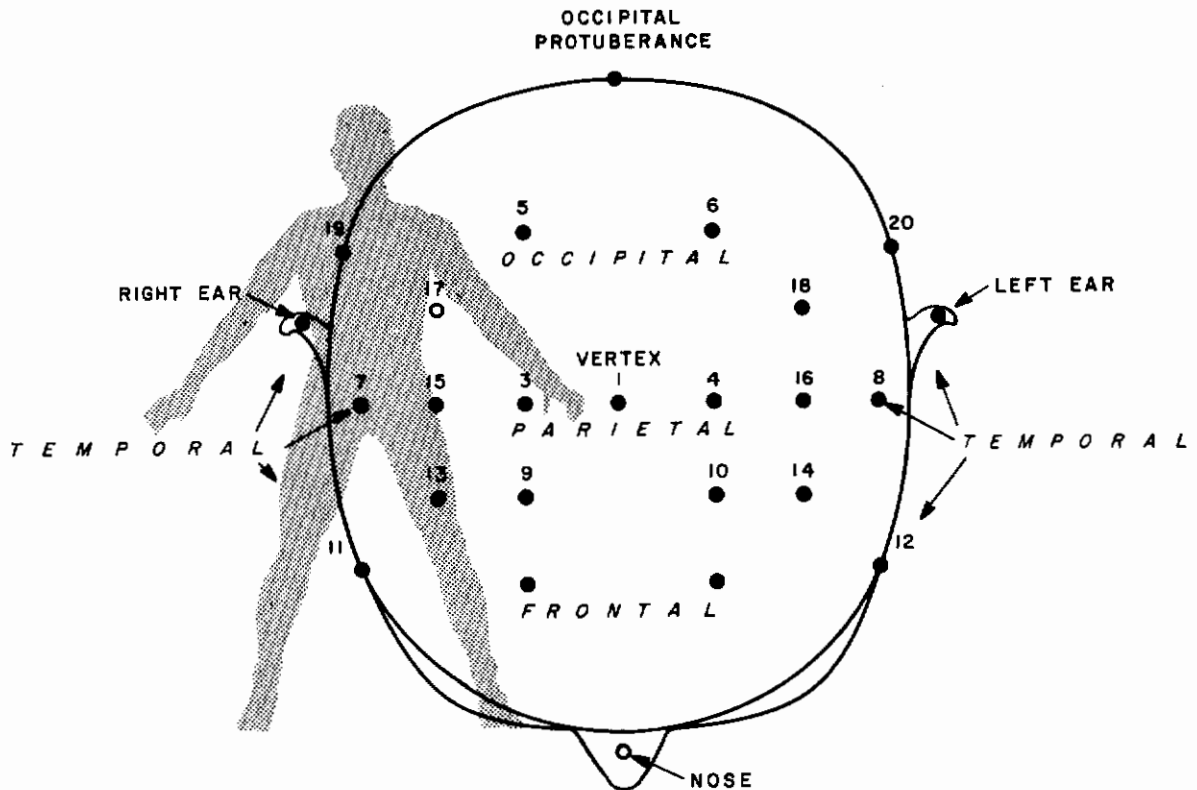


Figure 30. Location of Electrodes on the Scalp

2. Associated Electrical Activity

a. Delta and Theta Rhythms

A small amount of delta and theta activity, if symmetrical and of low voltage, is not considered abnormal. Delta activity is generally sensed in the frontal leads.

b. Alpha Rhythms

While alpha rhythms may not appear in all EEG recordings, they are considered the normal rhythm in a relaxed, awake adult. Alpha activity is found mostly in the occipital and parietal leads. Alpha rhythms offer an index of subject alertness, since they disappear with the initial opening of the eyes, and remain suppressed during periods of alert attention (concentration on mental problems or on visual scanning).

BRAIN POTENTIALS

c. Beta Rhythms

Beta rhythms are most commonly the strongest on frontal leads. As a general rule in all areas, fast waves of low voltage are considered normal, and high-voltage activity is considered abnormal; the latter may result from barbiturates or other drugs). Very fast spikes or sharp waves are always considered abnormal.

II. Aerospace Medical Applications

The instrumentation and techniques for EEG recording that are described herein draw heavily upon fairly standardized clinical practices. There are no standard techniques for comparable recording on active subjects, in stressful environments. However, much of the data that is desired in the clinic is pathological in nature; its significance is not measured in terms of varying response to environmental stress.

In aerospace medicine, on the other hand, relatively less significant data are sought, and recordings of the complexity and quality required in the clinic may not be required. For instance, as a simple index of alertness or consciousness, an airborne EEG recording may be desired to provide (1) a measure of the subject's reaction to high g stresses, or (2) a warning of hypoxia. Potentials of interest are concentrated largely in slow wave activity, low on the EEG spectrum (say, 3 to 30 cycles per second). (See ref. 31.)

This greatly simplifies the bandwidth and performance requirements for the telemetering of individual channels of data. Further, since indications of generalized activity rather than comparative area activity will often suffice, the complex clinical electrode arrays will probably not be required. Successful measurement with such limited but significant objectives may be conducted using only four, two, or even one pair of electrodes.

In general, it should be noted that in field applications involving subject activity, the EEG as a general physiologic index is of limited utility. The low signal level and the required electrode arrangements make it subject to extraneous signals and artifacts.

III. Electrodes

A. Requirements

As in other physiological monitoring systems using electrodes for sensing potentials on the human body, electroencephalographic measurement requires electrodes providing good electrical contact and low impedance. The electrodes should be small and easy to attach, causing a minimum of subject irritation or discomfort. They also should be firmly attached and nonpolarizing for long-term measurements with moderate amounts of subject movement or activity.

INDIVIDUAL MEASUREMENT SYSTEMS

B. Characteristics

Considering electrical requirements alone, needle electrodes are the most suitable. They offer the best electrical contact with minimum resistance, since they bypass (by penetrating) the high-resistance dermal tissue on the surface of the scalp. However, there are obvious objections to their use, especially in aerospace environments, so most electroencephalography is done with some type of surface-attached electrode. Characteristics may be summarized as follows:

1. Configuration

The usual configuration is that of a small metal disc or cup, about 3 to 6 millimeters in diameter (less than 1/4 inch), and no more than 1 or 2 millimeters thick.

2. Material

The metals used are stainless steel, silver, or silver/silver chloride. The latter is preferred because of its freedom from polarization. Suitable electrodes may be formed from flattened drops of solder.

3. Attachment

Electrodes usually are attached to the scalp with collodion or a hard cement such as Duco.

4. Electrolyte

An electrolytic paste or gel normally is used to ensure good electrical contact and a low junction resistance. It is rubbed into the scalp at the point where the electrode will be attached. Many cup electrodes have a small hole through the back permitting the insertion of paste or gel (with a syringe) after the electrode is attached.

5. Impedance

Most commercial electroencephalographs are designed for a low electrode-pair resistance of about 3000 ohms. If the scalp is cleaned and the electrolyte and electrodes applied carefully, suitable resistances below this figure can be attained.

IV. Electrode Connections

There are two basic methods of connecting electrodes for recording the electroencephalogram: the monopolar and the bipolar (ref. 66).

BRAIN POTENTIALS

A. Monopolar Electrodes

In monopolar electrode connections, an active electrode is placed over the cortical region of interest, and an indifferent electrode is placed at some point relatively distant from the cortical region. An indifferent electrode is formed commonly by attaching one electrode to the right ear lobe, attaching another electrode to the left ear lobe, and connecting the two electrodes electrically. Electroencephalographs are wired generally so that any active electrode of interest can be paired with an indifferent electrode by a simple switch connection. While many points on the cortical area may be of interest, active monopolar electrodes usually are placed as follows:

- Frontal: high on the forehead over each eye (points 1 and 2 of figure 30)
- Parietal: on the midline of the parietal region, above the external ear (points 3 and 4 of figure 30)
- Occipital: above and to either side of the occipital protuberance (points 5 and 6 of figure 30)
- Temporal: halfway between the external ear and the parietal electrode (points 7 and 8 of figure 30)

The lobe of the ear does not provide a truly indifferent electrode, since some brain potential is sensed at this location. In some applications, an indifferent point is obtained by averaging techniques, such as those used in electrocardiography; all electrodes on the scalp, except the one considered active, are tied together, and this combination serves as the indifferent electrode.

Another method for obtaining indifferent electrodes is to place the electrodes far out on the body, away from the head. These electrodes pick up extraneous potentials, such as those from the heart or skeletal muscle; although these potentials distort the record, they are easily distinguished from brain potentials.

B. Bipolar Electrodes

In bipolar electrode connections, two active electrodes are placed over the cortical regions, recording the difference in potential between the two regions. Clinically meaningful data usually are obtained by the simultaneous recording of several electrode pairs placed symmetrically over the right and left hemispheres. For example (refer to figure 30), the recording obtained from points 1 and 9 is compared to one obtained from points 2 and 10, but not with one obtained from points 4 and 6. Essentially monopolar connections, such as between point 1 and the right ear, or point 2 and the left ear, are compared in the same fashion.

INDIVIDUAL MEASUREMENT SYSTEMS

The main application of bipolar connections is clinical; they are most useful in localizing electrical activity in connection with brain disorders such as lesions or tumors. Some workers in aerospace medicine have reported success with nonsymmetrical bipolar connections; they have recorded potential differences between the left parietal and right occipital regions, and between the right parietal and left occipital regions, in studying the responses of certain subjects to airborne stresses, such as the high g forces of acrobatic maneuvers (ref. 50).

V. Signal Conditioning

The requirements for recording the electroencephalogram are essentially the same as those for recording the ECG and other physiological potentials. High-gain, high-impedance input stages are used, and the leads are capacitor-coupled to isolate d-c artifact. The differences that exist in ECG circuits are required to maintain stable, linear, noise-free response to inputs of extremely low amplitude and extended low frequencies.

A. System Characteristics

1. Gain

EEG potentials from the electrodes range from as low as 2 microvolts to as high as 200 microvolts. The first-stage amplifier or preamplifier associated with each electrode pair should provide a minimum gain of 1000. Additional amplifier stages then raise the signal to a level suitable for data transmission or recording.

In some cases adequate recording can be obtained by cascading two specially designed low-noise type ECG amplifiers with gains of 500 or 1000 each. Several manufacturers also supply EEG preamplifiers with gains of 50 for use with ECG circuitry.

2. Bandwidth

As in other systems, the frequency response of an EEG system is limited first by the capabilities of the recording system, and second by the high- and low-pass filters employed to reduce artifacts. While most recordings cover a bandwidth of 0.1 or 0.2 to 100 cps, the following frequency ranges often are used:

- For accurate, noise-free recording of extended low frequencies, chopper amplifiers with flat response from dc to 150 cps.
- For removal of baseline shift and other low-frequency artifacts, variable low cutoff frequencies of 0.35, 0.7, and 1.5 cps.
- For personal telemetry where interest is centered on slow-wave activity (such as alpha rhythms), miniaturized amplifiers with a low-frequency cutoff of 2 cps.

- For removal of EMG and other high-frequency artifacts, variable high cutoff frequencies such as 30, 40, 50, or 70 cps.

3. Sensitivity

The maximum sensitivity of an EEG system is normally 1 millimeter deflection for 1 microvolt input. For optimum response (full-scale deflection) to the wide range of amplitudes (2 to 200 microvolts) that may be present in a given recording, most electroencephalograms incorporate variable attenuators to provide a range of sensitivities, such as 1, 2, 5, 10, 20, 50, and 100 microvolts per millimeter deflection.

VI. Notes on Operation

A. Recording

For recording an electroencephalogram, the amplified potentials from the brain are applied to an oscillographic recorder (an integral part of the clinical electroencephalograph). Recording normally is done at a chart speed of 3 centimeters per second, but most oscillographs used for EEG recording permit half-speed (1.5 centimeters) and double-speed (6 centimeters) recording, and even faster speeds (up to 25 centimeters) are used for study of expanded slow-wave recordings.

Standard clinical multichannel recorders with 4, 6, 8, or 16 channels allow about 1 inch per channel (about 25 millimeters full-scale deflection). The recorder customarily is calibrated before each recording or series of recordings by applying a known calibration voltage to each channel simultaneously, and adjusting the gain on each channel for identical deflections. Most commercial electroencephalographs contain calibration voltage sources, from which voltages from 10 to 1000 microvolts may be selected.

When several bipolar electrode pairs are being recorded simultaneously (standard clinical practice), symmetry and polarity of the electrode pairs must be maintained. Figure 31 illustrates this point. If, for example, the right frontal-parietal EEG (points 1-3) is being compared to the left (points 2-4), reversed lead connections must be avoided (i.e., points 1-3 compared with points 4-2) to prevent the introduction of an artificial and clinically misleading phase reversal on the recording (ref. 51).

B. Balance Control

As is typical using ECG and other differential input recording techniques, a balancing potentiometer should be provided so that the input stage can be adjusted for minimum output when a test signal is supplied common to the inputs with respect to ground. When a chopper-type amplifier is employed, this provision is unnecessary, since the input is coupled through a transformer and is completely isolated from ground.

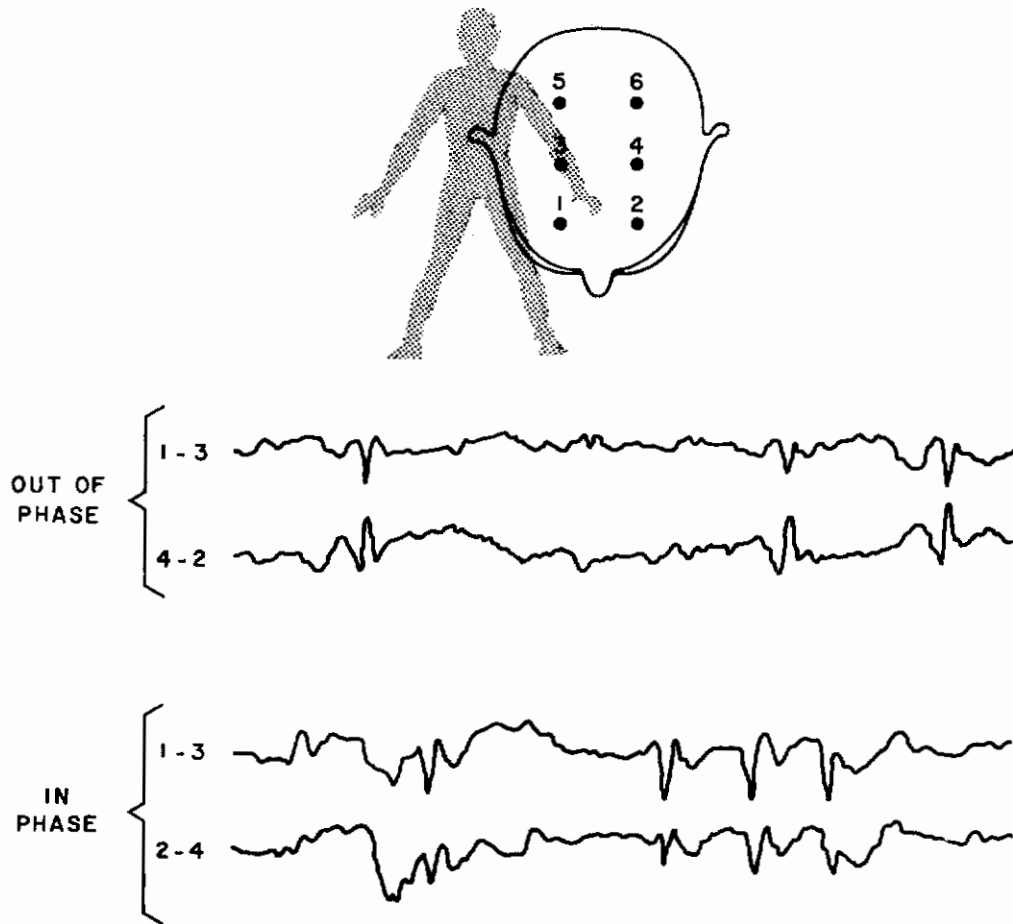


Figure 31. Artificial Phase Reversal Caused by Improper Electrode Polarity

C. Shock Hazard

In any measurement involving the coupling of a subject and circuits through electrodes, the individual electrode leads should be fused to protect the subject from shock hazards. Junction boxes with this provision are available for most systems that are built around off-the-shelf components.

D. Artifacts

The EEG recording is susceptible to several sources of external interference, such as 60-cycle hum and radio-frequency interference (discussed in Section III). Subject movement and improper application of electrodes also can interrupt and distort the recording, and there are several sources of electrical interference on the body of the subject which can be sensed by the EEG electrodes (ref. 66).

MUSCLE ACTION POTENTIALS

1. Electromyograms

Electromyographic potentials emanate from the muscles of the scalp, neck, and jaws, which are under continuous tonic stimulation. These potentials may be minimized by the use of low-pass filters.

2. Skin Resistance

Slow changes in skin resistance may vary the voltage sensed between electrode pairs and can produce a slow shift in the baseline of the EEG recording. These effects are minimized by the use of high-pass filters.

Sweating, particularly in the frontal regions, can also produce baseline shift by contributing to electrode polarization. Nonpolarizing electrodes should be used and, if sweating is profuse, the affected scalp area should be sponged dry and cooled periodically during the recording.

3. Eye Movement

The retinal metabolism normally generates a steady potential of about 1 millivolt, which is present across the surface of the head; being steady, it does not normally appear on the electroencephalogram. With the closing of the eye, however, or with eyelid flutter or movement of the eyeball, noticeable shifts in potential occur.

Potentials from blinking and tremor are relatively high and are easy to distinguish on an EEG recording. The artifacts produced by rhythmic eye movement (nystagmus), however, are quite similar to the standard brain wave. They sometimes can be distinguished clearly by slight reorientation of the affected electrodes. If this doesn't prove feasible, the artifacts can be compensated for by making a separate, simultaneous recording of the eye potentials alone, and then comparing this recording with the electroencephalogram for identification of these eye movement artifacts. (Refer to the discussion of electro-oculography on page 80).

MUSCLE ACTION POTENTIALS

The action of skeletal muscle (contraction and relaxation) is controlled by the central nervous system. Contraction of muscle fibre is triggered by electrical discharges from a motor neuron. Electromyography is the technique of measuring and recording the potentials in the muscle fibres which are associated with muscular contraction.

In clinical practice, electromyography has been used in the study of neuromuscular disorders. Using needle electrodes inserted into the muscle tissue of interest, characteristic electrical data can be obtained from very small regions of muscle, down to single motor units (a single motor neuron and its associated muscle fibres).

INDIVIDUAL MEASUREMENT SYSTEMS

For continuous monitoring of action potentials, needle electrodes cannot be used and surface electrodes, similar to those used in EEG, are employed. With surface electrodes, localization of muscle action potentials is impossible, and only gross potentials over relatively large muscle blocks can be obtained. Such measurements can be useful, however, and surface electromyography is used to assess gross body movement patterns and to evaluate certain physiological states such as fatigue. Correlated with other data such as the EEG and the GSR, the surface EMG can contribute to the drawing of psychophysiological profiles, which are useful in stress studies and in monitoring for subject viability.

I. The Electromyographic Waveform

The action potential from a single motor unit is typically a single sharp diphasic or triphasic spike which swings about the baseline or resting potential of unstimulated muscle. The surface electromyogram consists of a series of such spiked discharges (figure 32) from the many motor units which underlie the surface electrode contact area. A typical waveform accompanying muscle flexion contains from just a few to 25 or more polyphasic spikes, depending upon the size of the electrode and the degree of synchronism of the underlying muscle fibres. These spikes are intermittent, with an apparently Gaussian distribution (ref. 66).

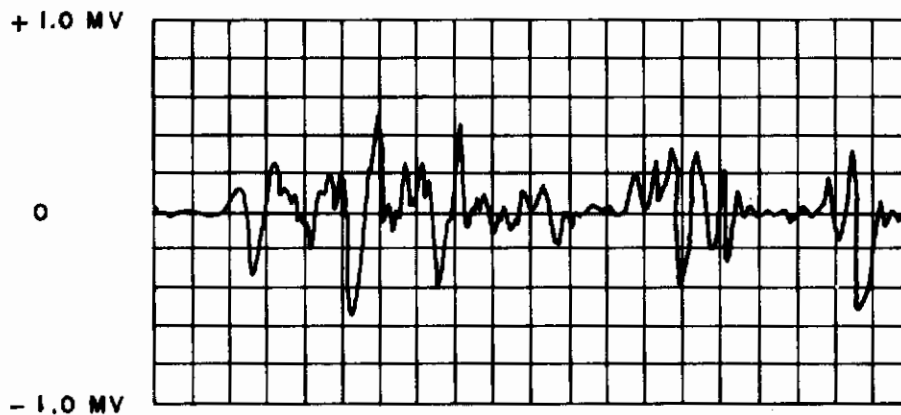


Figure 32. A Typical Electromyogram

The duration of the spikes is short (roughly between 5 and 15 microseconds), and they contain frequency components of 20 to 1000 cycles per second. Spike amplitudes may vary from as low as 25 to as high as 5000 microvolts; on the average, however, the variation is from 100 to 1000 microvolts (0.1 to 1.0 millivolts).

II. Electrodes

All the EMG parameters are affected by the area and quality of contact of the surface electrodes. Small metal disc or cup electrodes of the EEG type are commonly

used. Good attachment techniques with electrode paste or jelly are required to minimize contact resistance, so that low-level signals are not attenuated.

The movement of electrodes is a prime cause of artifacts during EMG recording. The electrodes are necessarily placed over active muscle, and they are subjected to considerable movement during exercise. Every effort must be made to minimize the breaks in contact that can occur. One approach that promises success is to fabricate electrodes of extremely low mass material. Small disc electrodes with a mass under 2 grams, made of foil-backed Mylar tape or aluminum foil, are attached to the skin with a ring of adhesive around the perimeter, which entraps the electrode paste used and prevents drying (ref. 69).

As in most other electrode techniques, ungrounded bipolar leads are preferred, measuring only the potential difference existing between the two measurement sites. Interference potentials present at both electrodes are blocked. The electrodes should be located so as to avoid an ECG artifact. Even when placed on the same limb, a pair of EMG electrodes may act as an ECG lead. Placement perpendicular to the long axis of the leg or arm reduces the likelihood of ECG artifacts.

Similarly, electrodes placed on the head, neck, or scalp may pick up EEG signals which, above 20 cps, are difficult to distinguish from the EMG. Electrodes placed on the forehead should have a nonvertical alignment to avoid picking up the corneal-retinal potential. Also, placement over a pulsating artery should be avoided, since a rhythmic artifact synchronous with the heartbeat may be detected. (See also ref. 16.)

III. Signal Conditioning

Routine requirements for low-level, high-impedance electrode sensing are applicable. EEG preamplifiers and amplifiers may be used with success, since they possess suitable high gain and a high input impedance. An expanded bandpass may be desired if an EEG amplifier is used that was modified so that the instrument could be used with recording devices of low-frequency response.

Filters may be used to remove noise from the EMG with only limited success, since there is usually an overlap in frequencies of interest. However, high-pass filters are used to remove the lower order EEG artifacts, and the slow shifts in baseline skin resistance. Filter networks are also employed to integrate the EMG, providing an output whose amplitude (envelope) is proportional to the repetition rate of EMG spike activity, a measure that is useful in some applications (such as evaluating muscle tone and pathology).

IV. Recording

EMG frequencies of interest are generally too high to be recorded properly with direct-writing instruments. Galvanometric recorders may be used or the EMG display

INDIVIDUAL MEASUREMENT SYSTEMS

on an oscilloscope may be photographed. If magnetic tape is used for on-line recording, then subsequent graphic recording is possible with direct-writing instruments, with the tape being played back at a slower speed to lower the frequencies of the components of the EMG waveform. A typical system configuration, with simultaneous oscilloscope display and magnetic tape recording, is shown in figure 33.

OCULAR POTENTIALS

Two measurements made by electrode techniques attempt to shed light on the visual process by correlating certain visual phenomena with observable accompanying electrical phenomena. One of these is the electroretinogram, which is a measure of potential differences between the cornea and the retina of the eye; the other is the electro-oculogram, which measures potential changes in extra-ocular muscles as a result of eye movement.

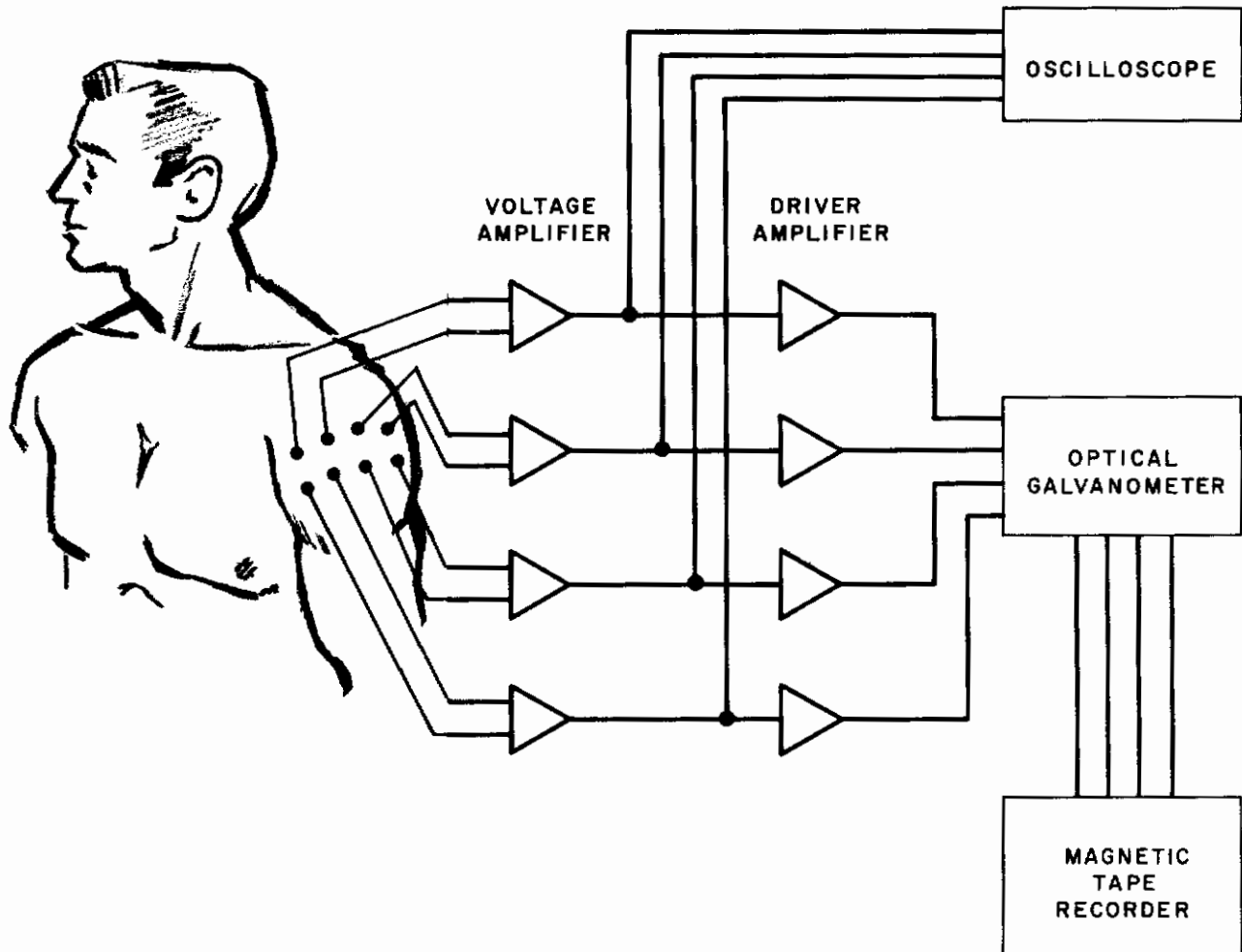


Figure 33. A Four-Channel Electromyograph With Multiple Outputs

OCULAR POTENTIALS

I. The Electroretinogram

The electroretinogram measures changes in potential between the cornea and the retina resulting from responses to light stimuli. While the measurement is useful in aerospace medicine for such applications as dark-adaptation studies, its application to continuous monitoring is doubtful because the sensing of corneal potentials requires the use of a contacting electrode (a plastic cup or contact lens) over the eye of the subject. (The reference electrode is placed on the forehead, where the potential is assumed to be the same as that at the retina or fundus.) Also, its usefulness in an aerospace environment is not presently known. Therefore, the characteristics of the electroretinogram are summarized only briefly in the following paragraphs.

The cornea is always at a positive potential with respect to the retina. The potential difference between them may be sensed with suitable electrodes, and variations in this waveform (which seldom exceed one half a millivolt) are registered when the eye is stimulated with light. The characteristic waveform of this response is shown in figure 34. The waveform is generally considered to be the resultant of the three components shown in broken lines. When light is applied to the dark-adapted eye, there is a small drop in positivity of the cornea (the a-wave), followed by a sharp positive spike (the b-wave), and then a long slow positive surge (the c-wave). After strong, prolonged light is extinguished (light-adaptation), there is a smaller positive rise (the d-wave) before the corneal potential drops off.

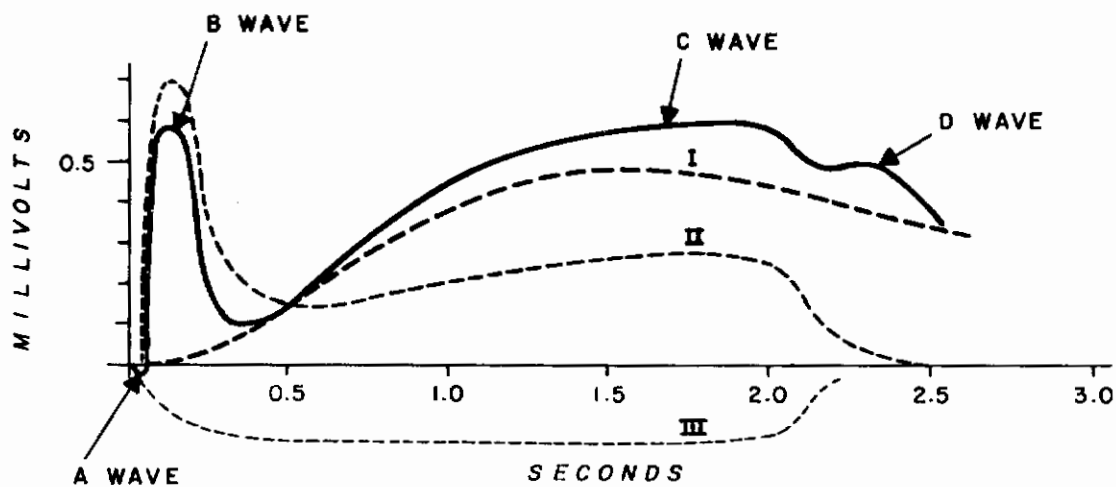


Figure 34. The Electroretinogram

The relationship of these responses to visual mechanisms is complex, involving to varying degrees optic nerve discharges and rod and cone activity in the eye. The I component, for instance, disappears with light adaptation, which is suggestive of rod or scotopic vision. Component II, which produces the positive b-wave, is regarded by some as a reaction primarily of the rods. Component III, which is negative, is considered an inhibitory process, often thought of as the process which "wipes" the retina clean of previous stimuli.

INDIVIDUAL MEASUREMENT SYSTEMS

All of the above described factors show marked differences with individual retinal characteristics; with the frequency, duration, and intensity of light; and with the physiological state of adaptation. (See ref. 10 and 34.)

II. The Electro-oculogram

The electro-oculogram measures changes in the potential found on the skin surfaces as a result of movement (rotation) of the eyeball within its socket. Electrode pairs placed horizontally at the corners of the eyes detect horizontal movements, while electrode pairs placed above and below the eyes will detect vertical movements (ref. 51).

Actually, the electrode pairs are sensing the steady potential -- about 1 millivolt -- that exists between the cornea and the retina. When the eyes are fixed straight ahead, the electrodes detect a steady baseline potential; with movement of the eyes, the potential across the electrodes changes, and, if the electrodes are connected to a recorder, there is a corresponding deflection on the recording. The deflection is either positive or negative, depending on the polarity of the connections, and the amplitude is proportional to the amount of movement. Figure 35 shows this relationship diagrammatically, and also illustrates the approximate electrode locations.

Simple movement data so obtained provide relative indications of such psychophysiological states as alertness and mental activity. Movement data also can show the

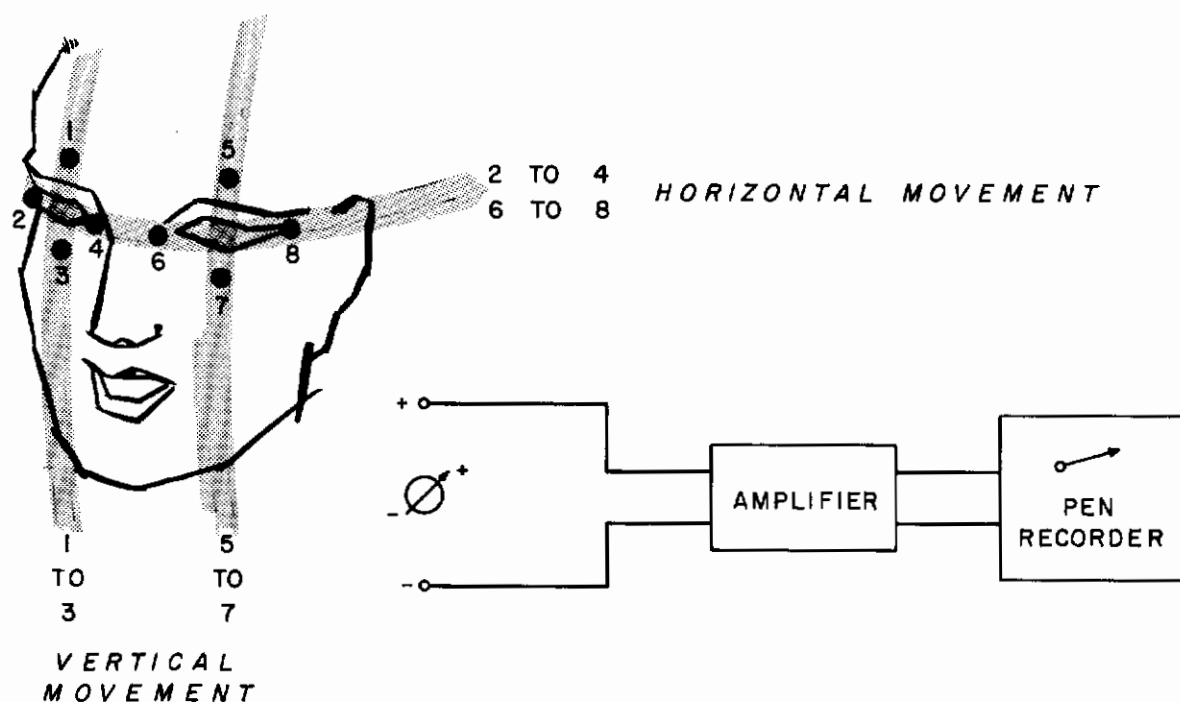


Figure 35. Method of Obtaining the Electro-oculogram

GALVANIC SKIN RESPONSE

occurrence of nystagmus, which the Russians have reported to be a possible effect of the weightless condition (ref. 2). If baseline data are obtained on the subject and compared against a calibration signal amplitude, quantitative data as to the amplitude of eye movement, in degrees, also are available.

A. The Measured Potential

The electro-oculographic (EOG) potential depends upon the highly variable phenomenon of eye movement. Whether the signal is processed as an a-c or d-c signal depends largely on the application. If relative movement or nystagmus is being monitored, the signal is treated as an a-c input; if a constant indication of eye position is wanted, the signal is processed as a d-c input. The amplitude of the EOG may be as small as 0.05 millivolt and as much as 3.5 millivolts for measurable rotations of up to 70 degrees.*

B. Instrumentation

Electrodes for EOG measurement, usually small disc or cup electrodes of the type used for EEG work, are fabricated of either stainless steel or silver/silver chloride. They are attached with collodion or similar adhesive, using electrode paste. Un-grounded bipolar leads are employed.

When measuring the EOG as an a-c phenomenon, capacitor-coupled amplifier stages, such as those used for the ECG or the EEG, may be used. If the latter are employed, the channel gain should be reduced or attenuators should be inserted in the input because of the relatively high signal level. When measured as a d-c phenomenon, of course, a d-c or chopper-type amplifier must be used.

GALVANIC SKIN RESPONSE

The galvanic skin response (GSR) is used in many research projects concerned with psychophysiological effects. The GSR and the electroencephalogram are the two most important electrical indices of nervous activity in man. Basically, the GSR is a relatively slow change in the resistance of the skin measured between two points. The change is triggered by both internal and external stimuli, and it can be detected in several ways, chiefly by passing a small current through the skin and measuring the changes in resistance that accompany the stimuli.** Responses to specific (identifiable) stimuli correlate with the emotional state of the subject (anxiety, fear, etc). The lie detector is a familiar application.

*The electro-oculogram is sometimes obtained in conjunction with the myogram. The oculogram gives directional data, and the integrated myogram gives force data (ref. 12).

**Actually, passing a current through the skin creates polarization potentials in the skin which behave like a complex impedance. Under the constant current conditions of most GSR measurements, these potentials may be measured in terms of equivalent resistances.

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Responses to nonspecific stimuli are smaller, but they have distinctive patterns similar to the specific responses and are readily distinguishable from random fluctuations or artifacts. The frequency of nonspecific responses is low during sleep, and it increases with subject alertness, so they are valuable in studying or monitoring states of consciousness.

There are three basic techniques for measuring the GSR: measuring resistance with a direct current impressed across the electrodes; measuring impedance with alternating current; and measuring skin potentials with no excitation current (ref. 70). The d-c measurement is the most common, and is discussed in the following paragraphs.

The basic requirements for skin resistance measurements are fairly simple. A minimal circuit arrangement that could be used is shown in figure 36 (ref. 35). The system shown provides a linear scale with resistance proportional to pen deflection, so long as skin resistance is small compared to the 10-megohm resistor. Calibration is possible with a helipot or decade resistors across the electrodes.

Measurements taken with this system would show the baseline skin resistance and variations of resistance, or GSR (responses to stimuli). The variations are usually in the range of tens of kilohms, but the baseline resistance itself may vary from the kilohm to the megohm range. In measuring the GSR, it is customary either to use a differential input or a capacitor-coupled stage at some point before transmission or recording; in this way the slowly shifting (essentially dc) baseline is blocked and only the more rapid responses are passed. An RC stage with a time constant of about 5 seconds usually suffices for this purpose.

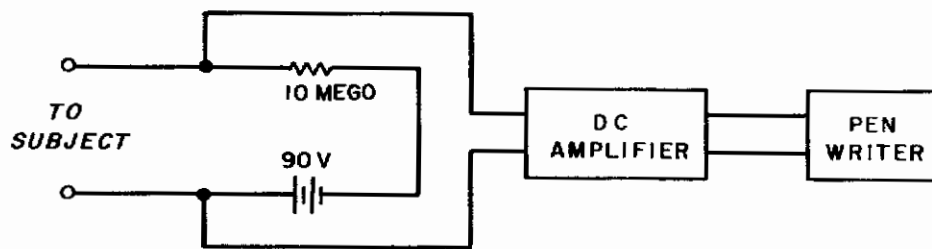


Figure 36. Basic GSR Circuitry

1. Electrodes for GSR Measurement

Among other factors, the GSR is related to the activity of sweat glands, and the most clearly defined responses can be measured with electrodes placed on the palms of the hands or the soles of the feet. Use of the hands as an electrode site has obvious objections in many aerospace applications, so the electrodes are usually placed against the soles of the feet.

GALVANIC SKIN RESPONSE

The type of electrodes used depends upon the requirements of the particular conditioning circuitry in the system. Electrodes must be selected with suitable base resistance, which depends upon the size of the electrode-skin contact, the material, and whether or not an electrolyte is used. Some users feel that dry electrodes are unsuitable because of variations in the contact area (subject movement or loosening of the attachment). Others object equally to the wet electrodes because of the resistance changes that accompany drying of the electrolyte after application. Much success has been reported with electrodes consisting of soft metal plates of zinc or lead; but it has been reported also that chlorided silver is preferable, since there is evidence that zinc electrodes augment the GSR (ref. 44 and 59).

The preferred electrode lead configuration is an ungrounded, two-electrode arrangement (bipolar). A floating configuration of this type prevents ground loop interference if the system is being used in conjunction with other electrode systems such as the ECG. Similarly, this arrangement isolates the subject and protects him from accidental shock. If a grounded electrode configuration is used, a fuse-diode arrangement similar to that described for the ECG electrode, should be placed in the ungrounded lead to protect the subject (refer to page 36).

II. Signal Conditioning

A. Baseline Skin Resistance Measurement

A small, transistorized, resistance-measuring circuit for measuring GSR in a personal telemetry system is shown in figure 37 (ref. 44). The d-c voltage to the electrodes is supplied from an oscillator (20 to 100 kc) through a full-wave rectifier. Changes in resistance at the electrodes vary the load on the oscillator. The output is taken from the primary winding of the oscillator transformer and passed through a voltage-doubling rectifier and integrator to remove the oscillator frequency. Adding a capacitively coupled amplifier to the output would block the baseline data and permit readout of the GSR. However, a fast change in baseline resistance might block the amplifier for several seconds.

The output of this circuit is nearly logarithmic; the output voltage varies roughly as the square of the resistance change. However, for small variations, the response is essentially linear with a small percentage of error.

B. True Differential Input GSR Measurement

One drawback of the circuit shown in figure 37 is that the amplitude of the GSR depends upon the amplitude of the baseline resistance. An improved circuit configuration measures the absolute value of the GSR independent of the baseline voltage. Here, the oscillator supplies a constant voltage (clipped by a Zener diode), and a transistor supplies a constant current to the electrodes through a transformer. The voltage drop across the electrodes is reflected into the transformer, causing a voltage drop

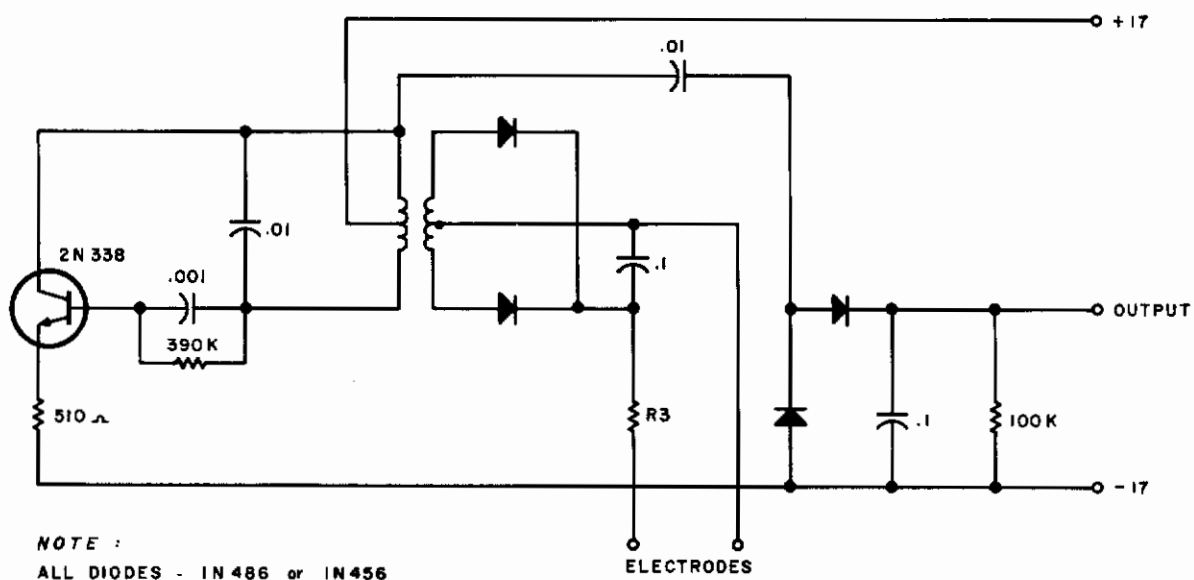


Figure 37. A Circuit for Baseline Skin Resistance Measurement

linearly proportional to the drop across the electrodes. The signal on the transformer is amplified and rectified with a voltage-doubling diode circuit (ref. 44).

C. Range Centering

The recording or display of skin resistance measurements is usually done on two channels (one for baseline resistance and one for GSR proper). The baseline measurement is much less sensitive because of the tremendous range of baseline variations (the amplitude of the scale is compressed). The GSR (or fine resistance) measurement is much more sensitive, with the scale expanded to measure small variations above or below a center point.

An electronic reset circuit is sometimes employed to prevent the fine resistance measurement from exceeding the selected scale range. A reset circuit, designed for the two signal conditioners described above, is shown in the block diagram in figure 38 (ref. 44). The baseline resistance output, taken from the input signal conditioner, also is applied through a capacitor, C, to a chopper amplifier, which provides the fine resistance or GSR output.

The two Schmitt triggers (ST) are set to the high and low limits of the desired range. If the GSR output drops below the lower limit, ST-1 is activated; if it exceeds the upper limit, ST-2 is activated. Either will trigger the electronic switch, discharging capacitor C1. The sudden voltage drop on the input or the chopper returns the GSR output to the center value and resets the Schmitt triggers.

GALVANIC SKIN RESPONSE

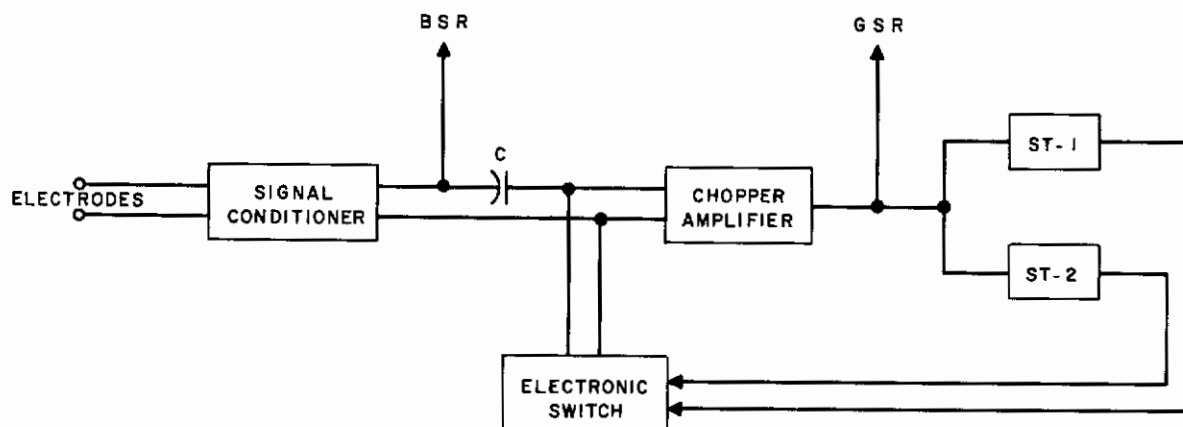


Figure 38. Components for Linear GSR Measurement

The base resistance output varies from 0 to 10 volts (for 0 to 100 kilohms resistance). The fine resistance output is centered at 5.5 volts, and may vary between 3 and 8 volts. For some recording systems, the value of the fine resistance output will have to be lowered. The sensitivity may be adjusted for 500, 1000, 2000, or 4000 ohms at full deviation (2.5 volts). Figure 39 shows a typical tracing obtained with this system.

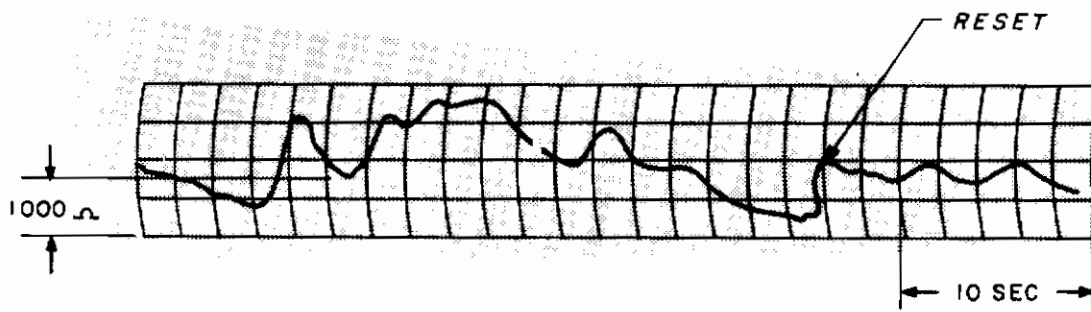


Figure 39. A Linear GSR Waveform

D. Skin Conductance Measurement

While most GSR measurements are by definition resistance measurements, the GSR also can be expressed as changes in skin conductance. Among other things, the amplitude response of the GSR is more linear with changes in tissue permeability, and quantitative measurements are considered more exact when expressed in micromhos. For conductance measurements, a constant voltage that does not vary with skin resistance changes must be supplied to the electrodes. The tissue resistance monitor

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developed by Airborne Instrument Laboratory* is of this type. It supplies a low-frequency (5 cps) voltage to the electrodes. A recording milliammeter calibrated in micromhos is used to record the response (ref. 28 and 59).

III. Notes on Recording

The sample tracing shown in figure 39 was made on a Brush** recorder at a chart speed of 5 millimeters per second (ref. 44). The GSR is commonly recorded at this speed or at the standard rate of 15 millimeters per second, so that specific and non-specific responses can be studied (ref. 35). For long-term measurements or baseline skin resistance, a much slower recording speed is desirable, so that overall trends in baseline shift are apparent. Chart speeds of 1 millimeter per minute or less are used for this purpose.

RESPIRATION RATE AND DEPTH

There are several proven instrumentation techniques for obtaining accurate respiration rate information, even in inaccessible or stressful environments. In such environments, however, it is difficult to obtain reliable quantitative data, which makes the measurement of respiration depth difficult by these techniques. The three most commonly used techniques, to be discussed below, are:

1. The chest strap. It measures the changes in girth that accompany respiration by means of a displacement transducer mounted on a flexible strap.
2. The temperature monitor. It senses changes in temperature in a thermistor or thermocouple as a function of the passage of air from the mouth or nostrils during respiration.
3. The impedance pneumograph. It senses the impedance changes in the chest that accompany respiration by impressing a current across two axillary electrodes.

If the monitoring situation permits, certain other, standard laboratory techniques may be used, such as flowmeters, spirometers, and gas collection and analysis equipment. (The latter does not lend itself to continuous monitoring, however.)

I. Chest Excursion

A chest strap with a dimensional transducer to sense the excursions of the chest during breathing is used commonly for measuring respiration. The strap is placed about the chest, somewhere between the nipple line and the base of the sternum. The

*Airborne Instruments Laboratory, Cutler-Hammer, Inc., Mineola, N.Y.

**Brush Instruments Div., Clearite Corp., Cleveland, Ohio

RESPIRATION RATE AND DEPTH

transducer may be a linear or a rotary potentiometer, a resistance wire strain gage, or an electrolytic strain gage. The transducer normally is placed in a bridge circuit to obtain a d-c output proportional to the resistance change produced by the chest displacement. Strain gages made of small rubber tubes filled with mercury (Whitney gages) perform better than most other types. According to a series of experiments carried out by Dr. Albert Ax, the combination of two Whitney gages (one around the nipple line and the other around the abdomen) gives better correlation to a spirometer measurement than the impedance pneumograph. The transducers are also lightweight and very comfortable for the subject.

This method can yield satisfactory rate information when the bridge current and the strap tension are properly adjusted; recognizable d-c pulses are produced which are suitable for analog recording or for integrating or counting in a pneumotachometer circuit.

There are, however, several objections to this type of measurement. Postural changes and movement may produce artifacts. Also, in long term measurements, the belt may be quite uncomfortable. A subject who breathes normally initially with the chest strap in place may in time unconsciously change his breathing rhythm to avoid working against the strap, with the result that chest excursions are reduced and a useful signal may be lost. While many users have found it perfectly satisfactory in a wide variety of applications, none has found it free of artifacts or capable of calibration for recording quantitative (minute volume) information.

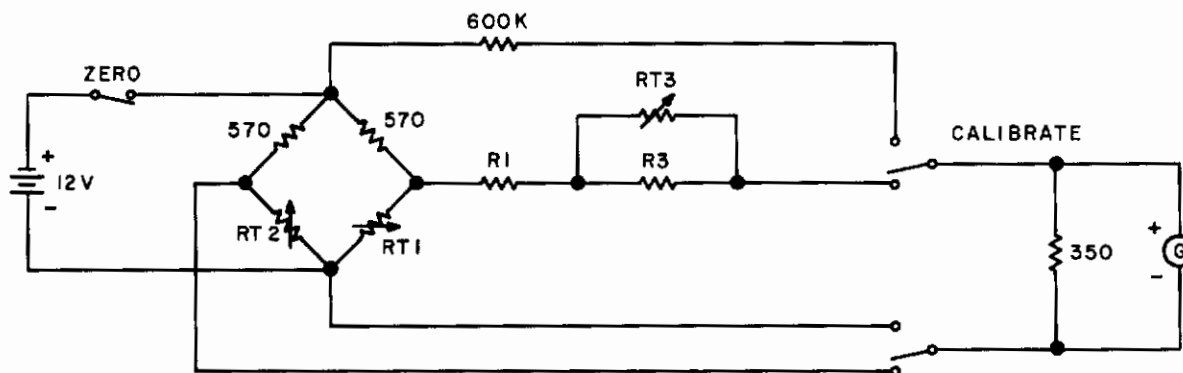
II. Temperature

Another method of monitoring respiration rate is with the use of a thermocouple or thermistor to sense the flow of air in an air line, or in the vicinity of the subject's mouth or nostrils. In the latter type, the transducer may be mounted on the outlet of a face mask or on the microphone of a full face mask. The housing should be designed to channel air from both mouth and nostrils. Even so, movement of the subject's head may direct the expirations away from the transducer.

Thermocouples generate a small voltage when warmed by the passage of expired air. This voltage may then be applied to an amplifier for processing. Thermistors operate differently in that they are generally heated to several degrees above ambient, and their significant resistance change results from cooling by the passage of expired air.

Thermistors are inserted in bridge measuring circuits to obtain the necessary output voltage variation. The bridge is balanced with a second thermistor in the opposite leg, which is not exposed to changes in temperature. If the respiration is being measured as a function of oxygen flow in an air line, a third thermistor may be used to monitor the base temperature of the oxygen supply, compensating for any bridge imbalance created that is not a function of respiration.

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LEGEND :

- RT 1 - SENSOR
- RT 2 - REFERENCE
- RT 3 - TEMPERATURE COMPENSATION

Figure 40. A Thermistor Bridge Circuit for Respiration Rate Measurement

Figure 40 shows a circuit that is used for such a measurement (ref. 26). RT1 is the respiration sensing thermistor in the flow line, and RT2 is the balancing thermistor. RT3 responds to and corrects for changes in temperature in the flow line, without responding to flow itself. The 12-volt excitation heats RT1 to about 500°F. The bridge output is suitable to drive a galvanometric recorder directly.

Quantification of respiration rate data obtained with the thermistor is difficult at best. The head-mounted thermistor is subject to artifacts when head movements deflect the stream of expired air. Furthermore, the voltage amplitude changes produced by air flow do not correlate directly with the tidal volume, and the resistance change of the thermistor is logarithmic, not linear, with temperature. There have been attempts to make the signal response linear by certain circuit operations; for example, the thermistor output is applied to a function generator with an inverse characteristic, and its output is then calibrated as linear with respect to air flow.

III. Impedance Changes

A device called the impedance pneumograph can be used for the measurement of respiration. In certain individuals under favorable circumstances, it shows good correlation with respiration volume. However, the impedance pneumograph, like the transducers employed in the first two techniques discussed on page 86, are difficult to calibrate for linear depth indications, and they are generally subject to gross artifact. Using a simple electrode technique, it permits accurate measurement of rate information with minimum subject instrumentation (the data can in fact be taken from electrodes which are being used to monitor the ECG). (See ref. 23 and 47.)

RESPIRATION RATE AND DEPTH

The technique consists of applying a high-frequency low-level carrier signal through the thorax. This carrier is modulated by transthoracic impedance changes accompanying respiration, and the modulated signal is amplified and demodulated to derive a d-c voltage proportional to the rate and depth of respiration. With baseline resistance through the thorax of several hundred ohms, resistance changes of 1 percent or only 1 ohm can be measured accurately.

The circuitry required consists of a suitable oscillator to generate the carrier, a reliable electrode lead arrangement, and amplifying and demodulating circuits that provide an output adequate to drive a recording or telemetry stage without further d-c amplification.

Figure 41 shows a typical circuit (ref. 23). The oscillator operates at 20 kilocycles and develops a 3-volt open circuit voltage on the secondary of an isolating transformer. The transformer output is applied through impedance-matching resistors to the electrodes. With a baseline resistance in the chest, the voltage drop across the electrodes will be about 0.15 volt, and no more than 0.3 milliampere will flow through the subject.

The voltage drop across the electrodes is transformer-coupled to a bandpass amplifier and to a bridge rectifier-demodulator, which provides a d-c signal for subsequent telemetry or recording. As a function of transthoracic impedance, the voltage drop across the electrodes represents the level of respiration. The minimum signal following expiration is the baseline, and, above this, the voltage is proportional to the depth of the breath.*

A large coupling capacitor can be added to the output circuit so that respiration pulses only will be passed. This capacitor will remove the d-c level (the respiration baseline), which is subject to drift because of changes in electrode contact impedance. A convenient time constant is 5 seconds, but shorter time constants may be used if extensive subject movements are anticipated.

A. Electrodes

With impedance changes as small as 1 ohm producing the pneumographic signal of interest, the importance of good electrode techniques is obvious. The factors discussed for in-flight electrocardiographic measurements (page 28) apply here as well. Firm attachment and low mass are important to ensure a constant contact resistance and to minimize artifacts due to movements. Electrode resistance, and attachment, are more critical in impedance pneumography than in any other measurement.

*It should be noted, however, that the exact relationship of this measurement to respiratory mechanics is not known. Marked deflections of the impedance pneumogram have been obtained during Valsalva maneuvers (forced expiration against a closed glottis) indicating that other factors besides tidal volume contribute to impedance changes (ref. 47).

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The nature of the pneumographic measurement dictates that electrodes be placed in the axillary region, one on each side of the chest. There are variations in response with small changes in electrode placement. It has been determined that on most subjects, with the electrodes on the mid or anterior axillary line, greatest response is obtained in the region of the sixth intercostal space, and it falls off above or below that region. Furthermore, there are marked (and sometimes extreme) differences in response between subjects of different body build and with different skin thicknesses. The most well defined signals have been obtained from persons of light build and movement artifact is more pronounced when the skin fold under the electrodes is thicker (ref. 47).

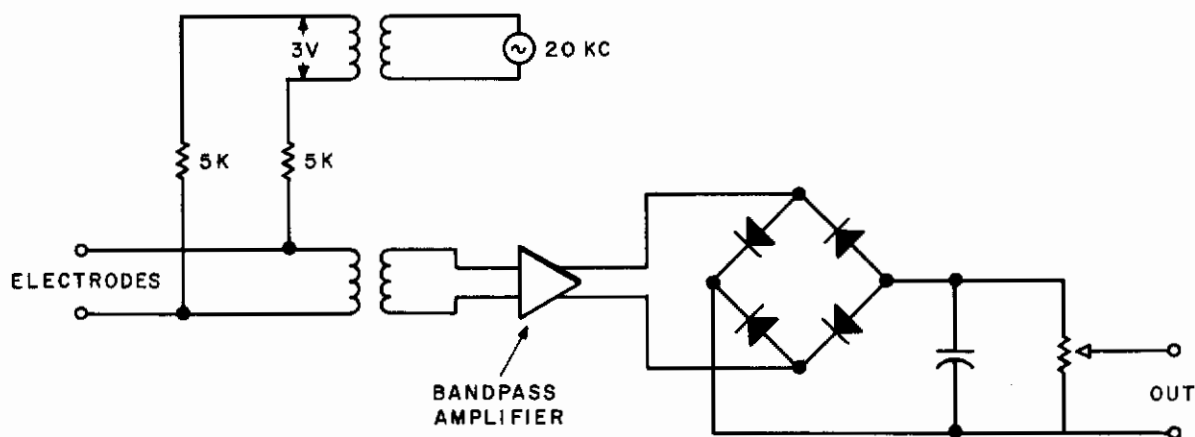


Figure 41. An Impedance Pneumograph

B. Oscillator and Demodulator

Figure 42 shows the circuit diagram for a complete impedance pneumograph that was developed for simultaneous use with ECG electrodes (ref. 47). The electrodes are connected directly to the secondary winding of the oscillator transformer. (A 0.1-microfarad capacitor isolates the oscillator circuit from low-frequency cardiac voltages.) The voltage drop across the electrodes presents a changing load to the oscillator during breathing, which effectively modulates the oscillator signal on the primary windings of the transformer. This signal is applied to a voltage-doubling diode rectifier, and the d-c variations are fed to a two-stage amplifier, with the output taken from the second-stage collector.

With the careful selection of resistance values for R1 and R2, an output of 4.5 volts is possible, which will vary a full volt for a 1-percent variation in transthoracic resistance.

PARTIAL PRESSURE OF RESPIRATORY GASES

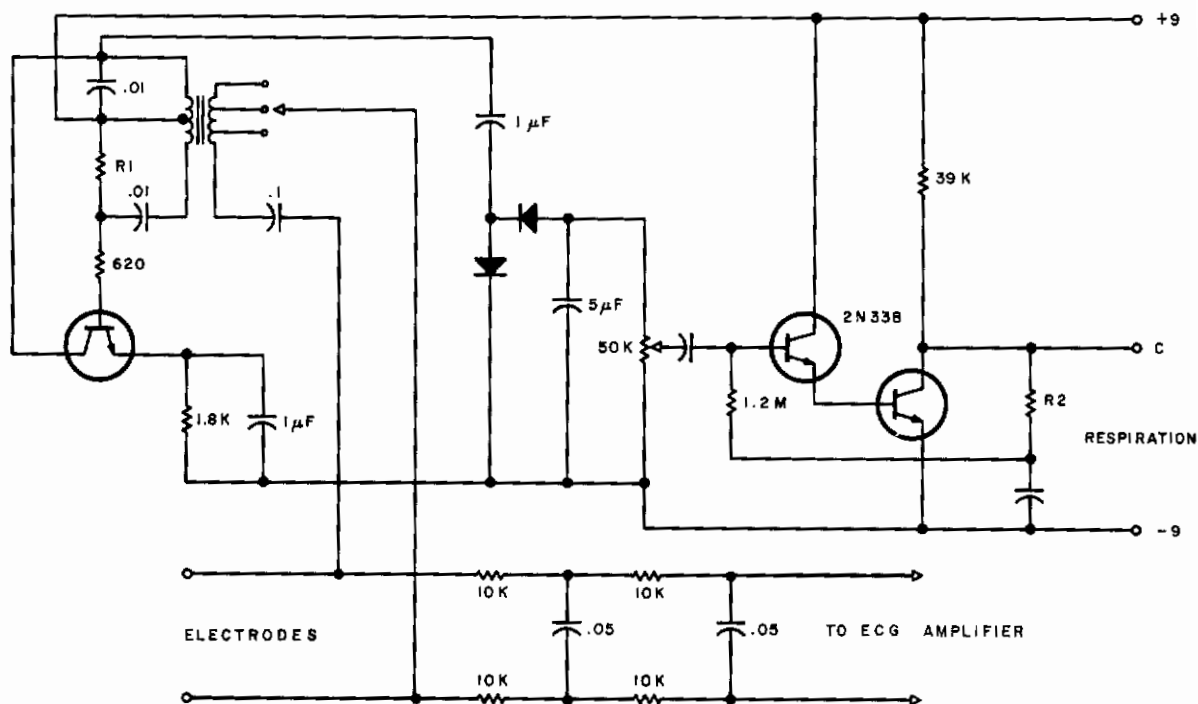


Figure 42. An Impedance Pneumograph for Quantitative Measurement

C. Calibration

Figure 43 compares a recording from the circuit described above with a corresponding recording made at the same time with a conventional spirometer (ref. 47). There is a good correlation in both amplitude and frequency response, with no noticeable phase shift. If calibration runs are made prior to a test measurement, the amplitude variations of the impedance pneumogram can be scaled for quantitative volume indications. With good electrode techniques and reliable circuitry, the impedance pneumograph should permit tidal and minute volume determinations, as well as respiration rate.

PARTIAL PRESSURE OF RESPIRATORY GASES

A comprehensive system for monitoring the respiration of a subject in stressful environments should include the measurement of oxygen partial pressures as an index of oxygen consumption, and of carbon dioxide partial pressures for indications of abnormal carbon dioxide retention. Exact quantitative data on these partial pressures are best obtained with closed-air respiratory systems, using laboratory-type gas analyzers, but these instruments are too bulky for most aerospace applications, and do not always lend themselves to continuous monitoring.

Two techniques are used for the continuous measurement of oxygen partial pressure in a monitoring application; both have been investigated for possible use as hypoxia

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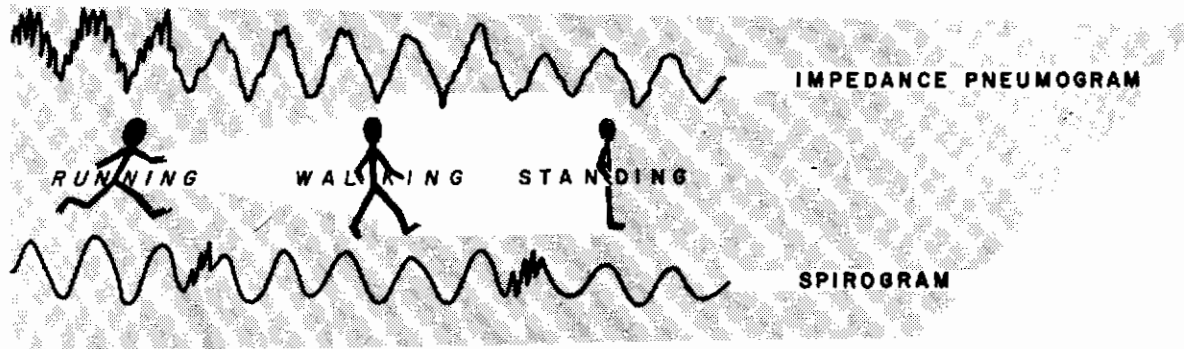


Figure 43. Comparison of Simultaneous Recordings With the Impedance Pneumograph and a Spirometer

warning devices. One of these measures the oxygen transport function by examining the blood photometrically with an oximeter to determine the saturation of oxygenated hemoglobin. The degree of saturation can be related to the partial pressure of oxygen in blood. The second technique, only recently applied to the continuous monitoring of oxygen, monitors the respiratory partial pressure with an oxygen-sensitive polarographic cell.

Polarographic techniques, also used for the continuous measurement of carbon dioxide partial pressures, have been limited primarily to environmental gas measurement, such as suit and cabin pressures. The response time of the carbon dioxide cell is too slow (several minutes) to follow the respiratory function.

I. Polarographic Measurement of Oxygen Partial Pressure

Most measurements of oxygen partial pressure are made with a variation of the polarographic cell, called the Clark electrode (described in Volume II). The Clark electrode is polarized by a small voltage. It depolarizes in the presence of ambient oxygen, causing a current flow that is proportional to the amount of oxygen present. The current is measured with a series microammeter by potentiometric methods, or with a servo device such as a d-c feedback amplifier.

The system described below was developed at the School of Aerospace Medicine as a potential hypoxia warning device. The system uses a transducer which, while similar, is not a true polarographic cell; it generates its own emf sufficient to reduce oxygen and permit current flow.

A. Transducer

The cell developed at the School of Aviation Medicine* uses a cathode or

*This device is commercially available from Chemtronics, Inc., Houston, Texas

PARTIAL PRESSURE OF RESPIRATORY GASES

sensing electrode of gold, a reference electrode of cadmium, and a chloride electrolyte such as dilute potassium or sodium chloride (ref. 53). The entire surface is encapsulated in a permeable polyethylene membrane which has a high specificity to oxygen. With a potential of about 0.5 volt between the electrodes, oxygen introduced through the membrane collects at the cathode and is reduced, depolarizing the cathode and permitting current to flow.

The characteristics of this cell are as follows:

Output current:	4 milliamperes at 150 mm Hg (20 millivolts across a 5000-ohm compensating thermistor)
Response time:	As fast as 2 seconds to 95% of equilibrium, depending upon membrane thickness
Range:	Zero to about 700 mm Hg (linear)
Sensitivity:	Within 0.5 mm Hg
Accuracy	Within 1% of full scale
Temperature Range:	5 to 45° C (20 to 40° recommended)

This device can be conveniently calibrated by exposure to ambient air. While it does have a positive temperature coefficient (about 5%/°C), this can be compensated for readily by shunting the cell with a thermistor, which has a negative temperature coefficient.

B. Signal Conditioners

The circuitry developed for this device is shown in figure 44. The initial stage of d-c amplification is followed by a 400-cps chopper, a feedback amplifier, a power amplifier, and a bridge rectifier. The output of the rectifier is a varying dc showing the breath-by-breath response of the cell, with amplitudes representing oxygen partial pressures. A typical record obtained with this system is shown in figure 45 (ref. 52).

C. Hypoxia Warning Circuitry

The actual warning circuit is a relay circuit that operates off the output of the rectifier in figure 44. The rectifier is shunted by a relay that controls a warning light. When the output falls below a predetermined level, the relay deenergizes, and the alarm light goes on. When the output rises above the alarm level, the relay is energized and the light goes off.

The optimum level at which relay action should take place has not been determined completely, but the limits are fairly obvious. The critical level for unconsciousness due to hypoxia is 60 mm Hg; at this level the partial pressure of oxygen in

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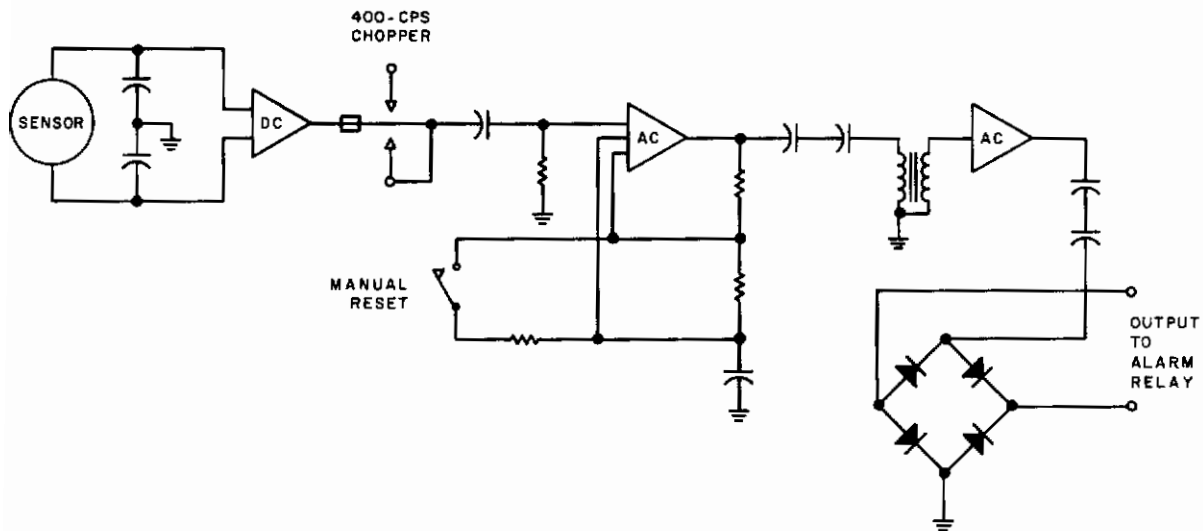


Figure 44. Circuit for a Hypoxia Warning System Using an Electrochemical Transducer

the lungs falls below the venous partial pressure, and oxygen exchange no longer occurs. The normal partial pressure of oxygen (O_2) in expired air is 100 mm Hg. Allowing a buffer zone at either extreme, a setting between 70 and 90 mm Hg is dictated.

A sensitive relay is needed for this type of alarm whose pull-in current is approximately the same as its drop-out current. If the pull-in current is too high, the relay will not re-energize promptly when the pressure level rises back above the alarm point. The alarm indication will continue even though the pressure is well above the danger point. To overcome this possibility, a manual reset switch can be incorporated in the warning system, as shown in figure 44. The relay is connected into the feedback

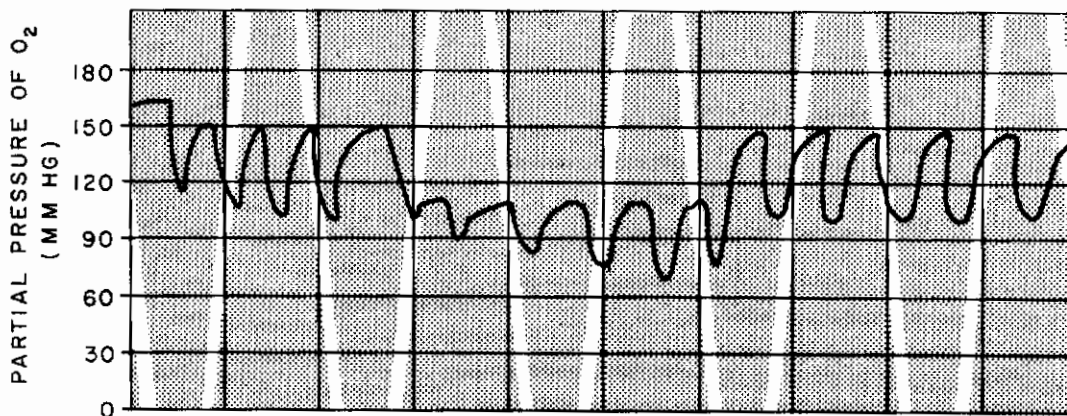


Figure 45. Breath-by-Breath Output Obtained With Electrochemical Oxygen Sensor

circuit. When it is closed, the feedback is momentarily decreased, causing the amplifier gain to increase just enough to reset the relay.

II. Oxygen Saturation of Blood

The oxygen saturation of blood can be determined by measuring the percentage of hemoglobin in blood in oxygenated form with a photometric device called the oximeter. The oximeter employs two separate photocells which respond to red and infrared light that pass through the lobe of the ear. The responses of the two cells vary because of the absorption of the red light by the oxygenated hemoglobin in the blood in the ear. When the outputs of the two cells are connected to elements of a bridge measuring circuit, a difference voltage results that is proportional to the percentage of oxygen saturation in the blood (ref. 79).*

A. Transducer

A standard transducer for this measurement is the oximeter earpiece made by Waters (Model XE-60A)**. The unit provides continuous monitoring of the relative saturation of the blood in the ear, using two iron-selenium photocells. A pressure diaphragm is incorporated to occlude the ear to obtain reference readings through the bloodless ear. When used with suitable signal conditioning circuitry, this instrument measures absolute percentages accurate to within 2 percent (standard mean deviation) in the 90 to 100 percent saturation range, and to within 5 percent at the 75 percent range.

B. Signal Conditioning

The output of the oximeter transducer is nonlinear, and measuring and amplifying circuitry used with it must be logarithmic to provide output voltages that are directly proportional. Waters makes a unit which incorporates amplification and galvanometric display in a single unit. With proper calibration and manual conversion of the galvanometer display, full scale accuracy of 2 percent can be obtained. Ensco*** makes a logarithmic amplifier (model OAS-1B) for use with the Waters earpiece that displays the percentage of saturation on the galvanometer directly. It also provides a 0.25-volt output to drive a recording device.

TEMPERATURE

Body temperature is one of the most valuable indexes of the overall physiological state and of reaction to specific physiological stress. Its measurement is a vital part of

*Refer also to Volume II, pages 41-42.

**The Waters Corporation, Rochester, Minn.

***Ensco, Inc., Salt Lake City, Utah.

INDIVIDUAL MEASUREMENT SYSTEMS

continuous monitoring from certain stressful environments, and it is generally included in most general physiological monitoring systems.

Normal body temperature measured orally is about 98.6° F (rectal temperature is about a degree higher, and axillary temperature is about a degree lower). Deviations of more than 2 degrees from these figures indicate unusual ambient conditions or serious change in the physiological state. For monitoring, the temperature span of interest is roughly 95° F to 105° F.

As discussed in previous volumes, three devices have been used for monitoring temperature: bimetallic thermocouples, calibrated resistance thermometers, and thermistors. Of the three, the thermistor is by far the most sensitive. A semiconductor device, the thermistor has a high negative coefficient with temperature. Typical units undergo a 4 percent change in resistance per ° C change in temperature (2.2 percent per ° F). Used in a bridge circuit with a meter of only moderate sensitivity, the thermistor can easily detect temperature changes as small as 0.1° F (ref. 11).

Early thermistors were not used extensively because their characteristic resistance changed as they aged. Current manufacturing techniques have overcome this limitation, and they are now available with a resistance characteristic that permits easy calibration and even interchanging of individual probes used with the same thermometer unit. (The thermometer unit consists of the measuring and display circuits that read out the thermistor resistance change in degrees of temperature.)

Thermistors are available in small ceramic-like beads for probing small surfaces or orifices, or in flat discs of various sizes for surface temperature measurements. Some probe types are small enough to fit in a 22-gauge hypodermic needle; the surface temperature units are about 1/2 inch in diameter.

Body temperature is measured at several points on the surface of the body or several inches deep in the rectum. The rectal probe affords the most stable body temperature measurement, if the depth of insertion can be maintained, but its use is a source of discomfort to most subjects, and therefore is avoided when possible.

Measurement of body temperature at the surface introduces several problems. Body temperature is not constant at all points, and several sensors may have to be attached to the body to obtain mean or average values of temperature. The axilla probably affords the single most stable surface temperature measurement. Also, the method of attachment is somewhat critical. If attached too loosely, the thermistor responds to surrounding ambient air temperature; if attached too tightly, the temperature readings are affected by local skin irritations.

A typical system used for temperature measurements with a thermistor is shown in figure 46 (ref. 26). The system uses a Yellow Springs* thermistor (Model 409) to

*Yellow Springs Instrument Co., Inc., Yellow Springs, Ohio.

TEMPERATURE

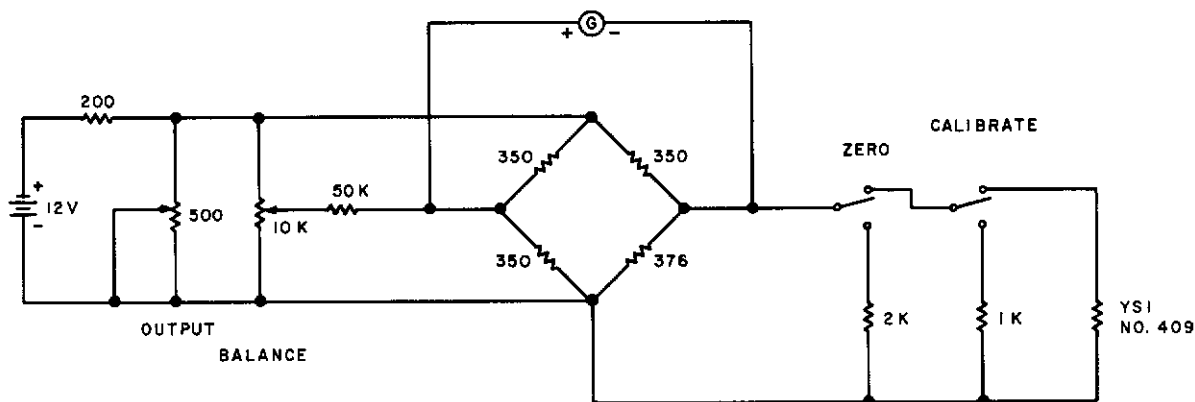


Figure 46. A Thermistor Bridge Circuit for Body Temperature Measurement

measure skin temperature in the range of 80° F to 120° F. Over this range, the resistance of the thermistor changes from approximately 2100 to 840 ohms. The 350-ohm bridge circuit is calibrated by the substitution of 1000- and 2000-ohm precision resistors for the sensor. The 500-ohm potentiometer adjusts the output of the bridge, and the 10,000-ohm potentiometer balances the bridge when the temperature range is shifted. The output of the bridge is fed to an optical galvanometer (CEC* Model 7-341), which is a multichannel instrument that can record other channels of physiological and environmental data.

In other applications, the bridge output might be applied simply to a sensitive milliammeter circuit for direct display or to a low-frequency telemetry channel. The bridge output requires direct coupling to signal conditioning circuitry, since the signal derived from the measurement is by its nature a d-c, baseline phenomenon. A high signal level avoids drift in d-c preamplification.

If the temperature sensor is not exposed to widely changing environmental temperatures, the normal temperature deviation expected is small, and low-frequency telemetry channels can be employed. Furthermore, if time-division multiplexing is used, the duty time of the telemetry channel can be very low, since it is a slowly changing variable.

*Consolidated Electrodynamics Corp., Pasadena, Calif.

Section III

OTHER SYSTEM CONSIDERATIONS

Certain aspects of monitoring system design and operation must be considered when individual system components are used in practical applications. While individual physiological parameters may be monitored or measured in certain situations with few problems, usually, however, two or more physiological parameters must be measured simultaneously. Further, data transmission may be required to accomplish monitoring from an inaccessible location or stressful environment.

The problems of instrumenting multichannel systems, including such difficulties as mutual interference between data channels, are discussed in this section. Also treated are the use of a data transmission or telemetry link, the requirements for such a linkage, and the limitations it may impose upon the overall system performance. In addition, the capabilities and limitations of magnetic tape recording, either as an adjunct to or a substitute for a telemetry link, are reviewed. Finally, the likely sources of interference that may hinder system operation are considered, and guidelines for troubleshooting a monitoring system are established.

MULTIPLE CHANNEL OPERATIONS

As indicated above, most monitoring systems are designed for applications in which several physiological variables (plus related information) are to be monitored and recorded at the same time. Certain inputs plainly require correlated physiological data for complete interpretation, and, up to a point, the use of certain complex and expensive system components, such as a radio-frequency telemetry link or a multichannel oscillographic recorder, becomes justified economically when many channels of information must be handled simultaneously.

Instrumenting such multichannel systems requires considerable planning; simply connecting the several individual monitoring system components together may result in many problems. Interaction between channels may result in recordings that are distorted or filled with artifacts, the information capacity of data transmission channels may be overtaxed, and there may be problems in the simultaneous registration of the several data inputs.

The problem of multiple data channels has already been touched upon in the discussion of individual measurement systems in Section II. Blood pressure monitoring by means of the auscultatory sphygmomanometer (page 52), for example, entails the simultaneous recording of (1) the occluding pressure in an arm cuff and (2) coincident arterial

MULTIPLE CHANNEL OPERATIONS

pulses. In most such applications, two separate data channels must be provided in the monitoring system, one for each input. In the more automatic versions of this type of measurement, the signals conveying arterial pulse information are used to trigger instantaneous readout of cuff pressure signals; thus, beyond the initial signal-conditioning stages, only one data channel is required for the transmission and recording of cuff pressure data.

There are systems in which a single input data signal, with appropriate signal conditioning, can be used to monitor two discrete physiological variables. For example, a single data channel for recording an electrocardiogram can furnish, through counting circuitry, the automatic registration of heart rate information. Also, the output from a single channel for monitoring basal skin resistance data can be modified by conditioning circuitry to remove the baseline data and provide a separate output for the galvanic skin response (ref. 44). Similarly, a single electrode lead (two wires) for the electrocardiogram, with appropriate excitation circuitry, can monitor impedance variations that are an analog of the respiratory cycle (the impedance pneumogram, page 88, and ref. 24 and 47). In this instance, a single sensor, the ECG electrode lead, is coupled to the two data channels through frequency-sensitive circuit elements of the appropriate impedance to prevent the signal variations in one channel appearing in the other and interfering with the signal variations there.

1. Sources of Interchannel Interference

A multichannel system generally will use multiple sensing devices, with separate data channels to handle and record the output from each sensing device. The chief problem in such a system is the complete separation of channel information; interference between channels may occur at almost any point in the system. It may occur at the source or point of signal pickup from the body of the subject, particularly when electrodes are used to sense different physiological variables. It may occur in signal conditioning circuitry, or in data transmission circuitry.

Interfering signals or artifacts may be coupled between channels directly (through common ground or power supply connections or common electrode leads), capacitively, or inductively. Potential sources of interference are multiple electrode leads, exposed leads or lines in adjacent circuits, unshielded oscillator or transformer circuitry, transmitting antennas, and a-c power supplies. There are various techniques and devices used for eliminating interference; for example, differential input circuits are used to block out extraneous voltages at the physiological source, grounding connections are well planned, circuit elements are bypassed to conduct some kinds of interference to ground, and shielding is used to intercept electromagnetic or electrostatic interference.

Some of these remedial techniques have been described for individual measurements in Section II. Refer also to the general discussion of interference and its elimination further on in this section.

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II. Problems in Multiple Electrode Pickup

A sensor attached to the body of a subject may respond to some variable other than the one desired or to more than one variable, especially when electrodes in close proximity are used to sense bioelectric phenomena. (These undesired responses also are possible with physical transducers: a displacement transducer used to monitor respiration may pick up a pressure pulse, a phonocardiographic microphone may pick up a respiratory component, etc.)

This problem with electrode techniques is not peculiar to multichannel monitoring. Diverse unrelated voltages are present on the surface of the body at all times. Even in single-channel monitoring, then, electrodes and signal conditioning components must discriminate among these diverse voltages, so that only the parameter of interest is recorded by the system. As described for specific variables in Section II, three basic techniques are used to block out extraneous voltage signals: careful electrode placement, frequency-selective filtering, and differential input connection.

There are times when none of these techniques is completely effective. Electro-oculographic potentials interfere with or appear on the electroencephalogram, particularly when recording from the frontal regions of the brain. Filters cannot be used because they would remove EEG as well as EOG data. The only recourse here is to record the EOG on a separate channel at the same time (using a separate lead system for this purpose), so that the artifact on the wanted channel can be identified readily. In some applications, artifacts may be just as difficult to remove but they are easily identifiable; therefore, they can be detected and compensated for in subsequent analyses of records. For example, a steady pattern of spikes on an EEG is recognized as the QRS complex of the ECG, or a rhythmic artifact superimposed on the EMG is identified as a pressure pulse resulting from the location of one electrode over an artery.

III. Transmission Factors

If multiple channels of physiological data must be obtained from remote or inaccessible locations, there are two factors that must be considered when choosing a data transmission link: First, the data channels must be combined (multiplexed) for transmission over a single data link, which involves the study of the various methods of modulation and multiplexing available. Second, the bandwidth of the telemetry link, which varies with the type of multiplexing employed, must be determined so as to establish the best combination of bandwidths of the channels feeding into and out of the data lines.

These subjects have been treated in detail in Volume II, Section V, and they are reviewed below under TELEMETRY SYSTEMS considerations.

IV. Multichannel Display and Recording

When several channels of physiological data are being monitored at the same time, it is usually desirable to provide simultaneous registration of the various channels on a single record. Recording or display can be accomplished with the necessary number of individual, single-channel devices, but a multichannel device affords greater convenience in comparing data on different channels against the same time base.

For simple meter or dial display, of course, a bank of instruments, one per channel, necessarily must be provided. But for graphic presentation numerous multichannel techniques are available. Display of several channels on an oscilloscope can be accomplished with a multi-gun cathode-ray tube, or with a single-gun cathode-ray tube equipped with an electronic sampling switch and a long-persistence screen.

Graphic recording devices are readily available for simultaneous recording of two, four, eight, or more channels of information, all on the same time base. The normal arrangement is to allot a certain portion of the strip chart to each channel, but with optical galvanometers, the recording of each channel can make use of the full width of the strip chart. Recorders powered by pen motors are also available with such full-width recording, but only by offsetting each of the styli so that they do not interfere one with the other; this makes interpretation of the record somewhat difficult because of the displacement of each channel along the time base.

Multiple-channel systems can also be recorded on magnetic tape. On subsequent playback, one or all of the channels of recorded data can then be applied to the graphic presentation devices described above.

TELEMETRY SYSTEMS

When the remoteness or inaccessibility of a monitoring location makes it necessary to employ a telemetry link to transmit the information from the sensing devices to the presentation devices, equipment should be chosen that is an uncomplicated and inexpensive as possible while still providing a minimum of distortion or reduction of the desired information. The first step in establishing a telemetry link is to determine the actual transmission requirement. The following factors should be considered: (1) transmission range, (2) transmission environment, at the measurement site and over the link, both physically and as a source of signal error, (3) number of data channels to be accommodated, and (4) bandwidth of the individual channels, which is the chief criterion for the information content of the signals.

The appropriate hardware must be selected that possesses the operating principles and capabilities that will satisfy the requirement. A choice must be made between wire and radio links. If any but a simple direct-wire link is used, appropriate methods for multiplexing several channels and for modulating a carrier also must be chosen.

OTHER SYSTEM CONSIDERATIONS

All of these aspects of data transmission have been discussed in detail in Volume II of this handbook. The next few paragraphs review certain factors that are pertinent to the overall systems discussion in this final volume.

I. Wire Versus Radio-Frequency Telemetry

The emphasis here and elsewhere in this handbook is on radio-frequency telemetry. The criterion for wire-link telemetry is fairly simple: transmission must be over a ground-to-ground hookup and preferably over a short span (perhaps no more than a few hundred feet), permitting direct-wire connection. Wire-carrier systems certainly are possible for longer ground-to-ground links, but modulating and multiplexing components are necessary for a successful system. Once these are involved, it usually is preferable to go another step and provide the greater flexibility of an r-f link.

Wire systems have numerous applications in aerospace medical work. They can be used to monitor subjects in such stressful environments as a centrifuge or vertical acceleration tower. One obvious advantage is the greater bandwidths that are possible, and, in addition, continuous data can be obtained from remote locations which correlate more precisely (compared to an r-f link) with baseline measurements obtained with short, direct-wire hookups in the laboratory. (Some information is lost in any system where the data signal must be processed for sampling, modulation, multiplexing, demultiplexing, and demodulation.)

There often are applications for short-range telemetry that can be served by either wire or radio links. With limited performance requirements that can be satisfied by either type of hookup, the choice depends largely upon equipment availability, cost, power requirements, and ultimately, the personal choice of the users, who will be guided by their own equipment experience and the degree of success or failure they have had with such instrumentation in the past. Frequently, an application can be served adequately by a direct-wire link, but a small personal telemetry r-f system is chosen instead, simply because of the extra comfort and degree of freedom of movement afforded the subject (freedom from wire connections between the subject's body and external instrumentation).

Monitoring a subject in a high-performance aircraft or spacecraft from the ground obviously necessitates the use of a wireless, radio-frequency system. The decisions to be made in this application are the selection of appropriate methods for modulation and multiplexing to satisfy the transmission requirement.

II. Modulation Techniques

In radio telemetry, some variation of carrier-frequency operation is employed to convey the intelligence from the transmitting to the receiving site. The high radio-frequency signal generated by the transmitter must be processed so that its variations will contain the information in the relatively low-frequency signals coming from the

TELEMETRY SYSTEMS

sensing and signal conditioning components which precede the transmitter. The two basic transmission techniques are (1) magnitude domain and (2) time domain transmission.

Magnitude domain transmission is, simply speaking, amplitude modulation. The desired information is contained in the amplitude variations of the radio-frequency carrier. Amplitude modulation is relatively easy to instrument and it has the advantage of retaining the information in analog form. Unfortunately, almost all forms of interference and artifacts that can affect a carrier signal act directly upon signal amplitude; therefore, the potential loss of intelligence in the system is greatest with magnitude-domain transmission. Signal attenuation, system noise, system nonlinearity, and interference of various kinds (power lines, static electricity, etc) cause unwanted variations in signal amplitude. Amplitude modulation then is seldom employed in the transmission of precise quantitative data, except in intermediate stages before the actual transmission.

Time-domain transmission uses modulating techniques that convey the desired information as variations in the time characteristics rather than amplitude variations of the radio-frequency carrier. The variations may be changes in the instantaneous frequency or phase of a sinusoidal carrier, or they may be changes in the characteristics of a pulsed carrier signal.

Frequency modulation, or FM, is the simplest form of time-domain encoding. Successive cycles of a sinusoidal carrier signal are made to vary in period or frequency in accordance with the variations in amplitude of the information signal. Like amplitude modulation, this technique retains the information in continuous, analog form.

Pulse modulation techniques convey the desired information in discrete or discontinuous analog form. The information signal is sampled at periodic intervals, and pulses are generated at those intervals which convey, through some characteristic of the pulse, the amplitude of the information signal at the interval. The following are the basic types of pulse modulation (described in detail in Volume II, Section V):

- **Pulse Amplitude Modulation.** This is a special case of pulsed magnitude domain transmission; the intelligence is transmitted as a function of pulse amplitude, and as such is subject to the same limitations as continuous amplitude modulation.
- **Pulse Duration Modulation.** The duration or width of the pulse is made to vary in proportion to the amplitude of the data signal at the time of pulse generation.
- **Pulse Position Modulation.** The time of occurrence of the pulse, with reference to some fixed time index, is made to vary in accordance with the amplitude of the data signal.

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- **Pulse Code Modulation.** A train of fixed uniform pulses is keyed on or off in accordance with a binary code that represents the digitized value of the data signal amplitude.

Since carrier modulation or carrier keying essentially is a switching action, relatively simple, error-free circuit elements may be used. Systems become complex in pulse-type modulation, however, because correct timing must be maintained between transmitter and receiver, particularly when high sampling rates are used to accommodate a multiplicity of wideband channels on a single pulse-modulated carrier.

III. Multiplexing Techniques

Typically, a radio-frequency telemetry link transmits information from several data channels simultaneously. The several inputs are combined into one signal, which is used in the transmitter to modulate the carrier. At the receiver, the combined signal is separated into individual channels, from which the desired information is extracted and passed on to multichannel display and recording devices. The two principal types of multiplexing are time division and frequency division.

In a time-division multiplexed system, several channels of pulsed data signals are interlaced to form a single pulse train which then is used to modulate a carrier signal generated by a transmitter. Multiplexing and modulation of the analog input signals essentially are accomplished simultaneously, since the sampling component (the commutator) samples each input channel in a sequential, cyclic fashion.

Frequency-division multiplexing uses an intermediate step called subcarrier modulation. The input signal for each channel modulates a subcarrier signal (usually in the range of 400 to 70,000 cycles per second). The modulated subcarriers then are mixed (added) to form a composite signal, which is used to modulate the final r-f carrier.

Various combinations of these techniques may be used, especially in applications like aircraft or missile testing that may require many separate information channels. In time-division multiplexing, a technique called subcommutation may be used to combine many signals for optimum use of channel bandwidth. Several low-frequency channels are commutated, and the resultant signal is applied as one input to a final commutator for combination with several more high-frequency channels.

Similarly, time- and frequency-division techniques may be combined. Several channels of low-frequency information may be commutated, and the resultant signal applied as a single input to modulate one of many subcarrier oscillators to produce a composite modulating signal for an FM/FM system.

IV. Radio-Frequency Component Considerations

The selection of radio-frequency components depends upon many factors, all of

Contrails

MAGNETIC TAPE RECORDING

which must be considered in order to satisfy the transmission requirement with the proper combination of components.

R-f transmitters, for example, are selected on the basis of the transmission range requirements, the transmitting frequency, and the type of modulation being used. Also involved are the type of antenna and the sensitivity of the receiver that will be used, and practical criteria such as cost, size, and power consumption. For example, powerful transmitters are needed to overcome the signal attenuation of long-distance r-f links, but the stronger signal may be unnecessary if sensitive receivers or high-gain receiving antennas are used.

When a choice is to be made between FM transmitters and pulse-modulated transmitters, the selection may be determined simply on the basis of ready equipment availability. For some applications, pulse systems may be preferred to FM/FM systems because of objectionable baseline shifts in the latter caused by subcarrier oscillator drifting. Pulse-code modulation is probably the most error-free for the transmission of precise quantitative data, but it may not be chosen because of the added complexity of the necessary analog-to-digital conversion components. Systems employing pulse-amplitude and pulse-duration modulation, on the other hand, are generally less complex than FM/FM systems.

V. Radio-Frequency Interference

One factor that must be considered in instrumenting for r-f telemetry of physiological data is the possibility of radio-frequency interference destroying or distorting the output signals. In an airborne system or a small personal telemetry system, the transmitting antenna is fairly close to the sensing circuitry, and improperly shielded electrode leads can act as an antenna, picking up radiation from the nearby transmitting antenna. R-f signals as high as 30 volts may be induced by this means at the system input, completely destroying the usefulness of the particular data channel. Care must be exercised in packaging and wiring such compact systems to avoid this possibility. (See also the discussion of electromagnetic and radiative interference further on in this section.)

MAGNETIC TAPE RECORDING

Magnetic tape recording is one of the most useful adjuncts to a physiological monitoring system. Located in the overall measurement system at either the transmitting or receiving ends of the telemetry link, magnetic tape recording provides a means for holding experimental data in compact, readily available form for a wide variety of purposes. Among its capabilities are the following:

- Recordings retain signal information in the original electrical form as obtained from the sensing devices.

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- Recordings are available for immediate playback.
- Recordings may be played back as often as desired, giving great flexibility to data reduction and analysis programs.
- The frequency range of the tape may extend from dc to the megacycle range.
- With appropriate multiplexing, literally hundreds of channels may be recorded simultaneously on the same tape.
- The tape transport mechanism makes it possible to alter the time base of the recording during playback; therefore, the data can be expanded or compressed along the time axis to suit analysis requirements.

Precision instrumentation tape recorders are available for many applications. Large console units are employed to record the output at the receiving end of telemetry systems. Smaller units can be placed on board flight vehicles for inflight monitoring, where they either substitute for or supplement air-to-ground telemetry. Smaller, compact units, of the type used for audio recording, can be attached to or carried by the subject, providing a complete, self-contained monitoring system.

These smaller units, although quite suitable for many applications, have their limitations. Their chief drawback results from their mechanical components: it is difficult to engineer small and light tape transport mechanisms with precise tape speed control. As a result, recordings are subject to wow and flutter. One way to overcome this difficulty is to set aside one channel on the recording for flutter compensation.

A typical example of flutter compensating is that which is possible in indirect (FM) recording. The individual data signals are applied to subcarrier oscillators, whose outputs are combined and used to modulate the carrier oscillator which drives the recording head. One channel (one subcarrier oscillator) is left unmodulated, with the undeviated subcarrier frequency being applied to the multiplexer along with the data channels. On playback, the d-c output in this channel should, after discrimination, be zero; any variation appearing in the recording of that channel can be attributed to tape speed variations alone. During playback, then, this signal can be inverted and added to the output signal from each of the data channels, effectively cancelling out the signal components causing the wow and flutter.

There are three basic techniques for recording data on magnetic tape. One is direct or amplitude modulation, as is commonly employed in audio recording. While effective in many applications, direct recording has inadequate low-frequency response, prohibiting its use for recording certain physiological parameters. Much more often used are indirect (or FM subcarrier) and digital (or pulse) techniques. These latter two are adaptable to use with the type of modulating and multiplexing circuitry employed in the multichannel telemetry components of a physiological monitoring system.

SYSTEM MAINTENANCE AND OPERATING FACTORS

SYSTEM MAINTENANCE AND OPERATING FACTORS

Correct utilization of an instrumentation system for physiological monitoring requires attention to the prevention and correction of all system troubles. Improper system operation results basically from two things: malfunctions in the equipment itself, and the introduction of interfering signals in the data channels of the system. The following paragraphs describe general preventive and corrective maintenance considerations for good system operation. The sources of electrical interference, and means for suppressing interference are also discussed in some detail.

I. Equipment Maintenance

There are two objectives in performing maintenance on instrumentation systems: (1) to have it operate at peak efficiency and (2) to minimize equipment down time (out of operation). These objectives are best realized if maintenance is considered when an instrument, major component, or system is designed rather than after difficulties develop. Good maintenance practice includes the prevention as well as the correction of trouble.

A. Preventive Maintenance

The primary function of preventive maintenance is to forestall the failure of equipment while it is in operation with systematic checks and inspections. A regularly scheduled preventive maintenance routine should be established for inspecting and testing the equipment. In general, the equipment should be inspected for signs of overheating, dirt and corrosion, and loose connections. Indicating instruments, such as meters and oscilloscopes, should be observed for normal indications and proper functioning of controls. Preventive maintenance also includes adjustments necessary for normal equipment operation, battery checks, and checks for obvious abnormal operation.

B. Corrective Maintenance

In most cases, the trouble causing a failure or abnormal condition in equipment is revealed by cursory examination to be simple, obvious, and easily corrected. Cables and wires are checked to ensure that they are connected securely; indicator lamps, meters, fuses, and circuit breakers are observed for indications of abnormalities; controls and switches are checked for proper settings; and batteries are tested for indications of weakness. If the source of trouble cannot be found by a general examination, a systematic and orderly troubleshooting procedure is necessary.

1. Troubleshooting Procedures

A logical troubleshooting procedure is one that will localize and then isolate the trouble without the need for backtracking and "easter-egging" (the futile hunt for a trouble without a reasonable, systematic procedure). The number of checks

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and the time consumed are reduced to a minimum. Generally, a troubleshooting procedure should consist of the following. First, the trouble is localized to a particular unit of the equipment, either visually or with test equipment. When the faulty unit is located, the trouble is localized to a particular stage or circuit, and the defective part is isolated. As various checks are made during the troubleshooting process, the information should be correlated so that the next check can logically be determined.

2. Types of Malfunctions

Troubleshooting procedures are usually predicated on the search for the cause of a particular malfunction. The most significant indication of malfunction is a faulty output, such as no output, inadequate output, nonlinear output, extraneous output, and zero level drift.

An indication of no output is usually the easiest type of trouble to diagnose and correct. Troubles in the primary power circuits are often responsible for this type of failure. Some of the sources of power failure are: a-c power lines; power plugs; a-c, d-c, and filament power switches; power rectifiers; fuses; and batteries. Other likely causes include defective tubes, transistors, and transducers, and faults in the d-c power distribution system.

Inadequate signal output most likely is caused by weak tubes or low d-c voltages to tube and transistor elements. Low voltages in a-c supplies are caused by weak rectifiers, leaky or open filter capacitors, voltage-dropping resistors that have increased in value, and leaky or shorted bypass capacitors. In battery-operated circuits, the voltage output of a battery may have dropped below minimum tolerance under load conditions.

The nonlinear performance of an instrumentation system is indicated by erroneous output signals. The output signals cannot be related to the input signals at all levels of amplitude. A careful stage-by-stage check of the calibration of the instrumentation system should be made to locate the source of trouble. If the trouble cannot be corrected by the alignment and adjustment of controls, the operating voltages to the tubes or transistors in the circuits may be improper and should be checked carefully. A faulty transducer also may cause a nonlinear response in the output of an instrumentation system.

An extraneous oscillatory signal appearing at the output of the system (a display or recording device) usually indicates the equipment is passing signals introduced by the a-c power supply. This type of signal appears as a regular oscillation on the trace, and its frequency may be determined by studying the output. Signals at 60 cps (the most commonly found) are generally caused by pickup of power line frequencies in high-gain, high impedance stages. (For a discussion of causes and method of reducing power line interference, refer to page 112.) Signals at 120 cycles generally indicate a faulty filter capacitor in a full wave power supply.

SYSTEM MAINTENANCE AND OPERATING FACTORS

Other signals, of a similar appearance, although they may occur at almost any frequency, may be the result of undesirable oscillation within the system itself. The cause is usually unwanted coupling between high-level and low-level stages. Generally, faulty components or poor design or layout will be found to be the cause. The remedy lies in thorough systematic troubleshooting techniques.

Zero level drift of the output of a display device may be caused by temperature sensitivity of a component part, an unstable power supply, a bad tube, or a weak battery, or by misadjustment of a balancing control for temperature or power supply changes. A transducer or capacitor, particularly a temperature-compensating type, may be unusually sensitive to normal changes of temperature or may be subjected to abnormal temperature changes created by an overheated part such as a power resistor. Faulty or inadequate regulation of power supplies may cause instability in the voltages distributed to the tubes and circuits. A bad tube or weak battery also may be responsible for zero level drift in an indicating instrument.

C. Precautions in Testing Transistor Circuits

When connecting power supplies or test equipment to transistor circuits, polarities must be correct. Transistors and other similar semiconductor devices can be damaged or destroyed by the application of voltages of wrong polarity. In particular, be careful when using an ohmmeter. The ohmmeter contains batteries which may exceed the voltage ratings of some transistors. Also, in the low resistance ranges, the ohmmeter may be capable of delivering currents in excess of those which the transistor is capable of withstanding. A third problem exists because the polarity of ohmmeters may or may not be the same as the polarity labels on the meter panel. Thus any ohmmeter must be checked with another instrument in order to determine its polarity before it is utilized to measure resistances in transistor circuits.

High transient currents or voltages can damage semiconductors while troubleshooting in several ways. When applying a-c power test equipment or a soldering iron, care should be taken to ensure that power line leakage current is not excessive. Use of an isolation transformer is a good precaution to employ with all a-c operated test equipment and soldering irons, unless it has been determined that the equipment contains a transformer in its power supply or shows no current leakage. When using test equipment, it is good practice to connect a common ground between the test equipment and the equipment under test.

Transistors may also be damaged when an excessively high pulse is applied from test equipment. The safest procedure is to start with a signal level below the rating given for the circuit under test and then apply the required signal levels. Relatively high current transients can occur when test equipment is connected to a circuit where low-impedance paths exist.

Further, all power should be off when connections are loosened or moved, or

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assemblies or parts changed. Stray inductance at such times often causes inductive kickback. This is prevented by ascertaining that all parts of the circuit are secure before starting the test or energizing the equipment, that all power is off when any unit in the equipment is changed, and that all possible capacitor charges are removed from parts and test equipment before they are used with the equipment.

D. Repair and Replacement

1. Electron Tubes

Electron tubes should be replaced only upon failure or when it is evident that the tube is not properly performing its function in the circuit. The indiscriminate replacement of good tubes by new ones should not be resorted to as a method of troubleshooting because it may result in the need for recalibration or readjustment of critical circuits. A faulty tube can be detected by a good tube tester or by functional tests in the equipment.

2. Electronic Parts

When faulty parts are replaced in electronic equipment, the replacement part must have the required specifications for the circuit. Resistors should be of the same ohmic value, wattage rating, and tolerance as the original resistor in the circuit. Capacitors should be the same type, value, voltage rating, temperature coefficient, and tolerance as the faulty one being replaced.

The replacement part should be installed securely in the position of the original part. Leads should be short, direct, and soldered securely. When several leads are disconnected to replace a part such as a transformer, the leads should be tagged or otherwise clearly identified to ensure proper reconnection when the new part is installed.

3. Transistors

Transistors and other semiconductor devices are easily damaged by heat. Connections should not be soldered to the terminal of a transistor unless it is provided with long pigtail leads. When soldering, use a low-heat iron (less than 40 watts). Connect a heat sink such as a pair of long-nose pliers to the wire between the point being soldered and the transistor. One method of checking the efficiency of the heat sink is to put a small piece of beeswax between the semiconductor and the heat sink. When the wax melts, the temperature limit has been reached and the heat source should be removed immediately.

II. Electrical Interference

Electrical interference is any electrical disturbance or signal which causes an

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undesirable response or malfunction in electronic equipment. Any visible, audible, or otherwise measurable response is considered undesirable if not produced by a desired signal. Physiological monitoring equipment is particularly susceptible to interference (artifacts) because of the extremely low-level signals being monitored.

The following pages review fundamental considerations in the prevention and treatment of interference. References 18, 67, and 78 contain useful discussions in this area.

A. Interference Sources

There are two general forms of interference: random noise and signal. The term "random noise" is used for disturbances that are completely without regularity in their detailed properties. Characteristics common to random noise are random amplitudes, random phases, and lack of periodicity. The energy of random noise is spread fairly uniformly throughout the total or a large segment of the frequency spectrum, depending on the source. Interfering signals, on the other hand, are usually periodic and show some regularity, although their waveshapes and phases may be subject to some fluctuation. Like random noise their energy is often spread over a wide frequency spectrum. Unlike random noise, the distribution of signal energy usually is nonuniform, showing wide variations with definite maxima and minima.

Interference may be classified according to the way in which it is generated: it may be man-made, internal, and atmospheric. Atmospheric and internally generated sources of interference generate random noise, and man-made sources generate signal interference.

1. Man-Made Noise Sources

Man-made noise sources of interference include all electrical and electromechanical devices manufactured and operated by man. (This type of interference, except for radio transmitters, is broadband, covering a wide range of frequencies. The interference from radio transmitters usually is found on a specific or narrow bands of frequencies.)

a. Rotating Machinery

Interference is generated in rotating machinery by the transients from brush bounce, surface irregularities, arcing, static discharge, and stray magnetic fields. Any rotating machinery with sliding contacts such as brushes may be regarded as a potential source of electrical interference. All d-c and a-c/d-c motors, because they have brushes, are possible offenders, and some a-c motors, such as are contained in electric drills, stirring motors, circulating pumps, and centrifuge motors, have brushes and also may be sources of interference. Large a-c induction motors are usually designed to prevent interference, but small clock and fan motors are seldom shielded and may be interference sources.

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b. Fluorescent Lights

Interference may be caused by the normal ionization of fluorescent lamps or by defective starter and ballast units. If the interference is not the result of a faulty part in the lamp, installing a special filter at the lamp may be helpful. If the interference persists, the lamp may be moved, or shielded, or replaced by an incandescent lamp.

c. Relays and Switches

Relays and switches contain contacts that open and close during normal operation. If the voltage across the contacts is high enough, arcing may occur as the contacts are actuated. In addition, mechanical bouncing of the contacts may be a factor. In either case, transients are developed that can cause interference. For example, thermostatically controlled heating or cooling devices have relays and switches which may prove troublesome. A filter, consisting of a capacitor and resistor in series across the contacts, is often effective in suppressing the arcing and consequently the interference. (See figure 47.)

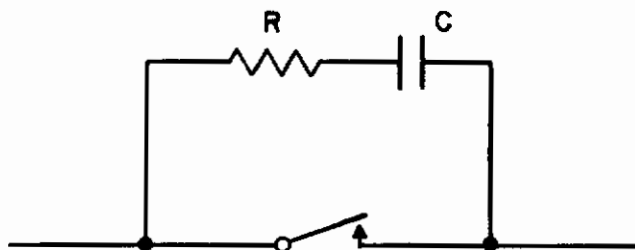


Figure 47. Filter for Suppression of Switching Transients

d. Industrial, Scientific, and Medical Apparatus

Apparatus that generates r-f energy such as diathermy machines, X-ray machines, and induction heaters, can be serious sources of interference. This type of interference can be radiated or conducted over long distances and may be difficult to locate. This type of interference is reduced most effectively at the source by filter traps or shielding.

e. Power Lines

The network of power lines installed in buildings, in addition to being a source of interference with its 60-cycle current, is a common medium by which interference is transmitted from other sources. The electromagnetic fields set up by the alternating current may induce an objectionable 60-cycle disturbance in the monitoring equipment. The coupling of interference from other sources by power lines into

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sensitive equipment can be reduced to acceptable limits by installing the power lines in conduit and by using filters in the line at the interference source and at the equipment. Flexible a-c line cords should be made of shielded cable with an outer insulating covering.

f. Pulsed Radar Transmitters

The radar modulator pulse is made up of a wide range of harmonics of the basic pulse-recurrence frequency, and interference occurs at an audio rate which can be heard as a tone at this frequency. If the interference is coming directly from the moving antenna structure, the signal level may vary in intensity, depending upon the rpm of the radar antenna. This type of interference is particularly troublesome to radio receivers.

g. Ignition Systems

The normal arcing in spark plugs and contact-breaker points of electrical ignition systems used in gasoline engines may interfere with reception in radio receivers. Ignition noise is limited to the frequency range of about 15 megacycles to 500 megacycles and to a distance of about 500 feet. Ignition noise can be suppressed at the source by shielding and filtering the entire ignition system. However, interference from ignition noise is best avoided by using FM radio systems rather than AM, since ignition noise is of an amplitude nature.

h. Radio Transmitters

Interference from radio transmitters usually occurs on discrete frequencies within the passband of the receiver. The interference may originate as cochannel, adjacent-channel, or harmonic signals. Tuned bandpass or low-pass filters will eliminate unwanted radiation from the interfering transmitter, except for that caused by the fundamental frequency. Tuned bandpass or high-pass filters at the receiver input are used to reduce the pickup of interfering radio signals. It is possible also for the local oscillator of a nearby superhetrodyne receiver to generate interfering signals.

2. Internal Noise

Internal noise is an inherent random noise generated within the system by undesirable and inherent electron motion in resistive circuit elements and tubes. Although the level of internal noise is low, the signal voltages at the input of physiological amplifiers and telemetry receivers are of such small values that the internal noise level becomes significant. In fact, this type of noise often is the limiting factor in the ability of these equipments to handle weak signals. Since internal noise is a function of equipment design, this problem can be dealt with best by using equipment with special low-noise circuits and components.

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a. Resistance Noise (Gaussian or White)

The noise generated within resistive elements is caused chiefly by the thermal agitation of the molecules of a conductor. This, in turn, sets up a random electron motion that creates minute noise currents in the conducting material. The common carbon resistor is particularly noisy and unsuitable for low-level signal amplifiers. Special low noise resistors such as the high-stability cracked carbon resistor and the wire-wound or metallized resistor are available for this application.

b. Tube and Transistor Noise

The noise generated within tubes and transistors is caused generally by the effects of random fluctuations of electrons on the various elements. The most important component of tube noise is called shot-effect noise. This noise is caused by random fluctuations of plate current and, when amplified and made audible, sounds like a shower of shot falling on a metal surface. The effects of tube noise are minimized by selecting tubes for low-signal level circuits with inherent low noise characteristics and by operating these tubes at low voltages. Triode tubes, in general, have better noise characteristics than pentodes; and, among the triodes, certain design configurations are better than others for low noise applications. There also are some tubes that are particularly sensitive to mechanical shock and vibration and create an effect known as microphonic noise. The electrodes of a microphonic tube are set in motion by mechanical vibrations, altering the position of the tube elements in relation to each other. This, in turn, varies the electrical characteristics of the tube accordingly. Microphonics are minimized by careful tube selection and shock and vibration mounting of the tubes or the chassis.

3. Atmospheric Noise

Atmospheric noise is external noise that originates in the atmosphere and outer space. The most well-known form of natural noise is atmospheric noise produced by lightning discharges during thunderstorms. Interference from thunderstorms is greatest at a frequency of about 10 kilocycles and diminishes with increases in frequency. Above 60 megacycles it is negligible, except for local disturbances.

Another type of natural noise in the same frequency range as atmospheric noise is precipitation static, which results from static discharges at a receiving antenna produced by particles of rain, snow, sleet, or dust blown against the antenna. The level of precipitation static is dependent on the weather, and it is most severe in polar regions.

Above 60 megacycles, the dominant noise factor is cosmic noise from interstellar space. This too diminishes with frequency until it is negligible at about 1000 megacycles. At frequencies above 1000 megacycles, the principal noise factor

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is that caused by the absorption of energy from radio waves by oxygen and water vapor in the atmosphere, decreasing the signal level.

Interference from atmospheric noise is minimized by the selection of suitable operating frequencies, by the use of directional antennas, and by the use of the minimum bandwidth required for the transmission of data.

B. Transmission of Interference

Interference is transmitted from its source by capacitive coupling, inductive coupling, conduction, or direct radiation.

1. Capacitive Coupling

Capacitive or electrostatic coupling occurs when one circuit is linked with another by mutual capacitance. (Refer to figure 48.) The interference generator acts as one plate of the capacitor and some element of the monitoring equipment acts as the other plate. Capacitive coupling is reduced by placing a conducting shield at ground potential between the plates of the capacitor. In practice, the shield is expanded to form a cage that encloses the whole input circuit where this type of coupling usually occurs.

2. Inductive Coupling

Inductive or electromagnetic coupling occurs when a conductor is present in the electromagnetic field set up by interference. (See figure 49.) The interference generator acts as the primary of a transformer and some element of the

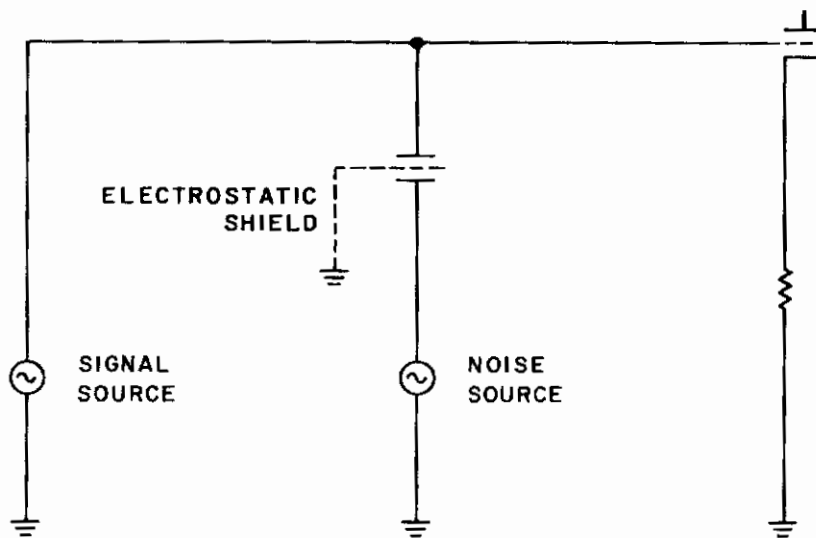


Figure 48. Capacitive Coupling of Interference

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monitoring equipment acts as the secondary. Inductive coupling is the most difficult to isolate and eliminate. Reduction of the interference at its source is most effective, but not always possible. The circuit picking up the interference may be enclosed in multiple grounded magnetic shields, but complete decoupling by the use of shields is extremely difficult.

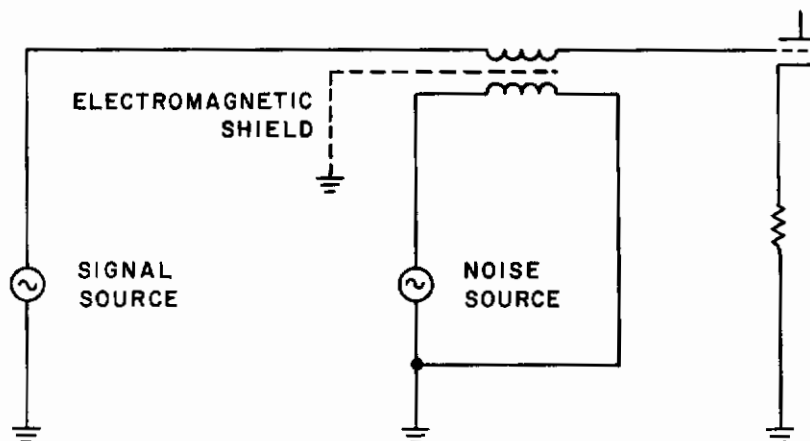


Figure 49. Inductive Coupling of Interference

3. Conduction

Conduction is the transfer of noise energy along a conductor from the interference source. The return path may be another metallic lead, a mutual capacitance, or a common metallic structure. The most common method of transmitting interference by conduction is through power and control cables. Such cables may provide a direct metallic connection between an interference source and a receiver, or they may have interfering voltages induced in them which then are conducted to the receiver. An example of interference by conduction is shown in figure 50. This type of interference is eliminated by the use of filters, preferably at the noise source.

Coupling interference by conduction also occurs as a result of the ground loop, which is the name given to the loop formed when various parts of an equipment are connected to different ground points and there is a resistance between them. Although the resistance may be small, usually less than one ohm, an appreciable level of signal is developed as the interfering signal current flows through it. The various parts of a circuit, equipment, or system should be grounded at the same point or, if different ground points are used, there should be no resistance between them (i.e., resistance should be as low as possible). An example of a ground loop and how it can be corrected is shown in figure 51.

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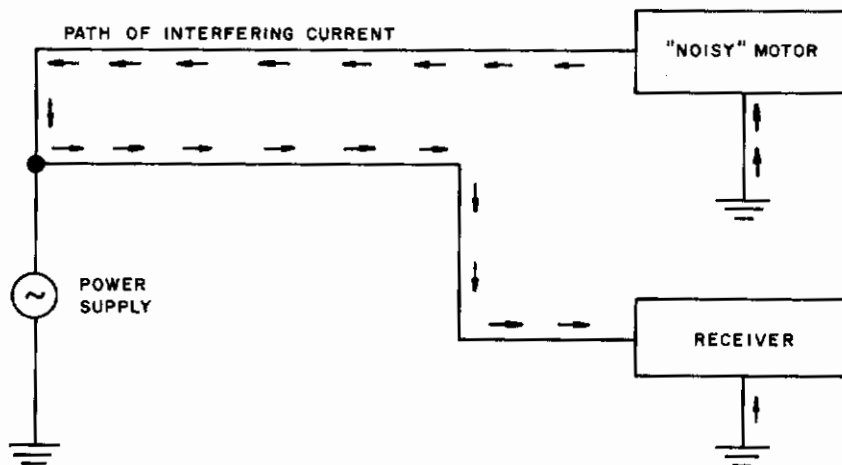


Figure 50. Interference Coupling by Conduction

4. Direct Radiation

Radiation is the phenomenon by which electromagnetic waves spread out in space from a source according to the laws of wave propagation. Interference by radiation occurs when a radio receiver detects radio signals other than the desired signal. Interference may enter a system through the antenna or by susceptible circuit components. (See figure 52.) Radiated interference is reduced by shielding at the source or at the receiver.

C. Suppression of Interference

There are five methods of preventing interference from reaching the monitoring equipment: location, orientation, shielding, grounding, and filtering.

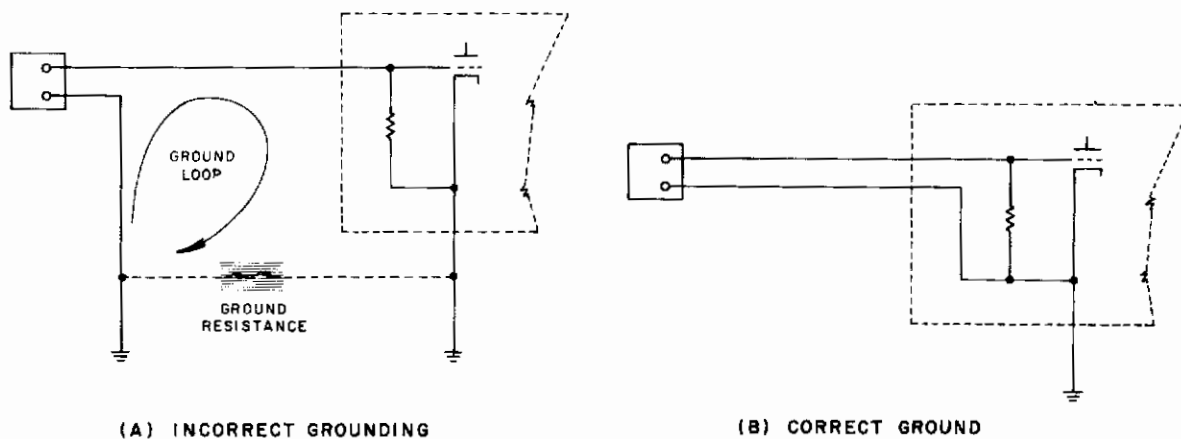


Figure 51. Ground Loops

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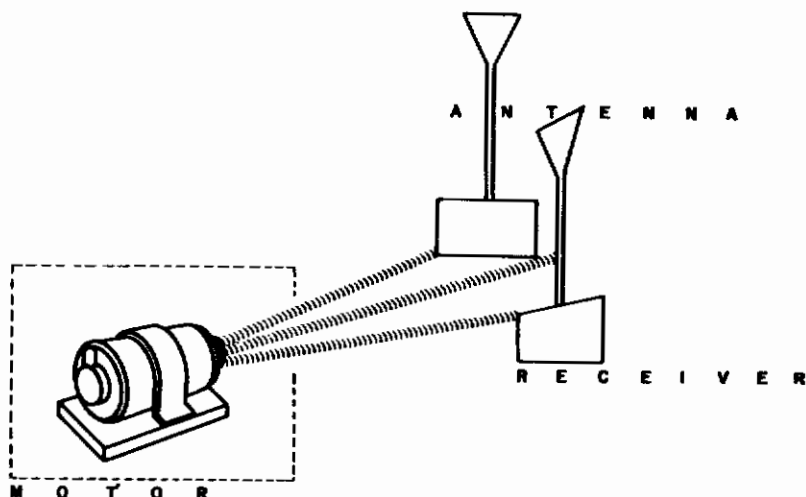


Figure 52. Radiative Coupling of Interference

1. Location

In the location method, the interfering equipment and lead wires likely to carry interfering currents are mounted or placed as far away as possible from the equipment and all power, control, input, and output cables connected to the equipment. In addition, equipment can be placed to take advantage of the natural shielding of metallic structures.

2. Orientation

Conductors sensitive to interference that must be located close to conductors carrying interference currents should be oriented to avoid paralleling the interference conductors and should cross them as nearly as possible at right angles. With proper orientation, the inductive coupling between circuits can be reduced to zero, since the number of magnetic flux linkages is minimum between wires perpendicular to each other. The optimum orientation is best obtained by actual experiment.

3. Shielding

A shield is a conducting sheet or covering used to attenuate interfering signals or noise or prevent them from reaching sensitive circuits. All practical shields are made of metals of high conductivity, and either they are used to confine the interfering energy largely within a limited area or they prevent any appreciable interference energy from entering a specified area. The first method is preferred since detectable interference cannot be radiated to any interference-sensitive equipment. The second method provides protection for interference-sensitive items placed in an area where interference is present and not easily eliminated. (See figure 53.)

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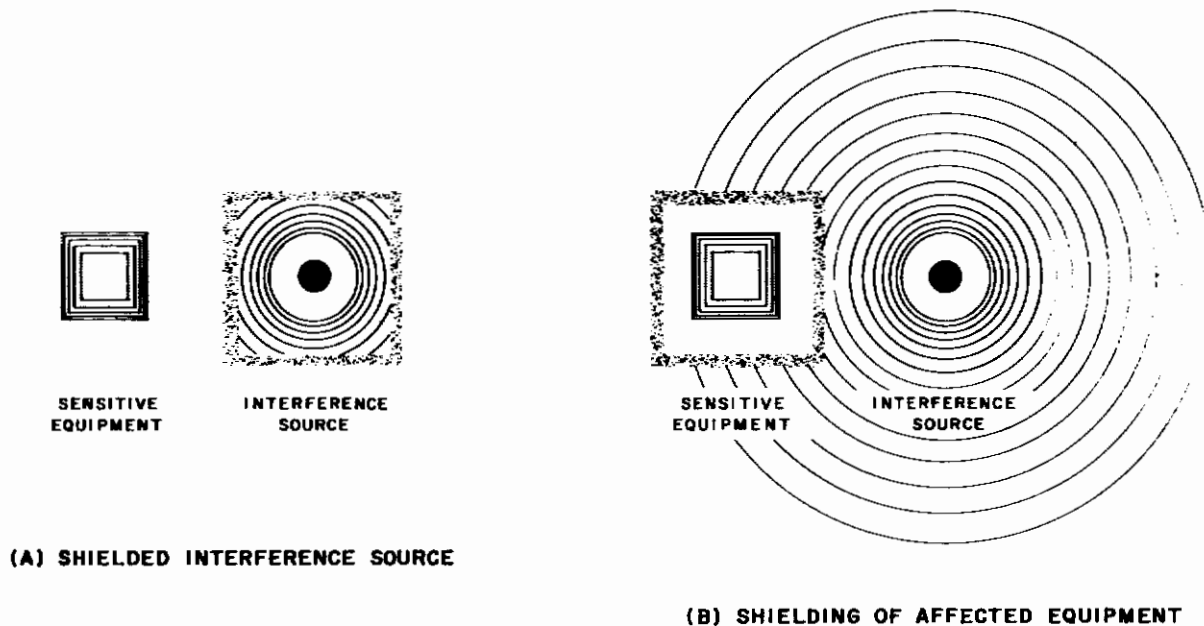


Figure 53. Two Methods of Interference Shielding

Constructing an effective shield presents two problems: one is the prevention of electromagnetic energy passing through the shielding wall, and the other is the prevention of leakage at the seams and joints. The attenuation of electromagnetic energy passing through the shielding wall is a function of thickness of wall material. Providing a shield wall thick enough to be effective usually is not difficult, since the mechanical strength required by the shield normally provides sufficient thickness to prevent penetrations. In many cases the shield does not have to be solid to be effective. Metal screening or perforated metal can be used for relatively low frequencies.

Preventing leakage at the seams and joints, however, can be a problem. When several parts of a complete shield must be joined, the number of joints should be kept to a minimum. Any joints that are present in a shield must have continuous metal-to-metal contact along the complete length of the joint. Continuous contact is possible by using mating members stiff enough to prevent distortion and a sufficient number of screws or bolts to insure high pressure at all points. When continuous contact along a joint is difficult to achieve, a conductive gasket can be used, which consists of a core covered with foil or mesh. When a shield joint is bonded, the mating members must be cleaned thoroughly. If not, the shield will be ineffective.

A shield designed to enclose an equipment or component completely must have openings for power control and output leads, maintenance and servicing, and proper ventilation. The interconnecting wires are filtered and enclosed in cable shield. Maintenance and servicing access holes can be closed with semi-permanent seals. Ventilation openings can be covered with a screen that makes continuous

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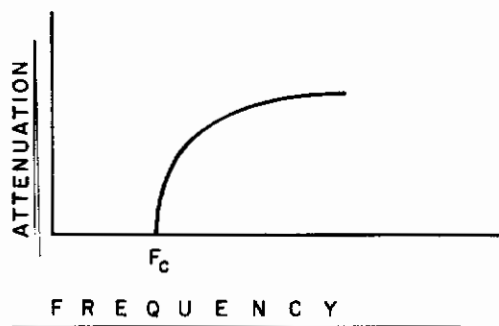
contact around the edges of the opening. The screen must be adequately bonded, with a material of equivalent reluctance, preferably by brazing or soldering.

4. Filters

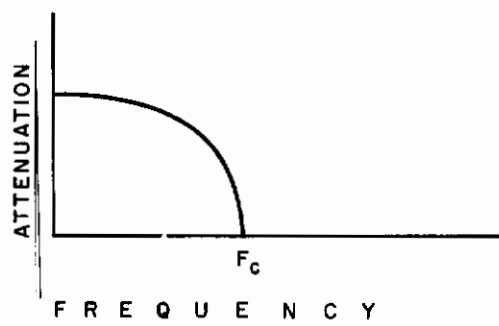
A filter is a device that passes or attenuates a band of frequencies. This band may be either wide or quite narrow. It is used for the suppression of interference when the frequency spectra of the desired signal and the interfering signal differ. Filters may be designed for both radio and audio frequencies and are installed directly in the transmission line or cable where the interference is to be suppressed.

There are four general types of filters: low pass, high pass, bandpass, and band elimination. These are illustrated in figure 54.

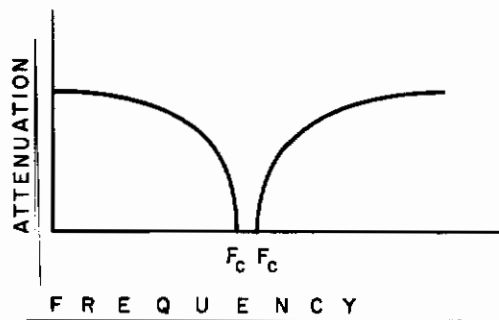
The low-pass filter passes all frequencies from zero (dc) to its cutoff frequency, rejecting or attenuating all higher frequencies. The high-pass filter attenuates



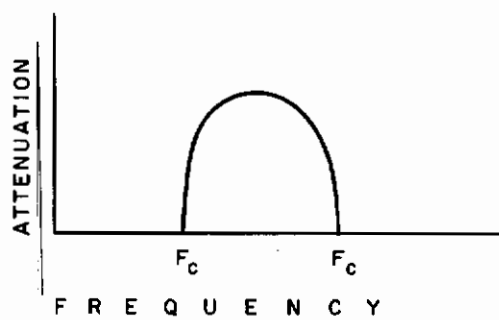
(A) LOW PASS



(B) HIGH PASS



(C) BANDPASS



(D) BAND ELIMINATION

Figure 54. Types of Filters

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all frequencies from zero to its cutoff frequency and passes all frequencies above. The bandpass filter has two cutoff frequencies and passes all frequencies within its specified range. The band-elimination filter has two cutoff frequencies and attenuates all frequencies within its specified range.

Filters are constructed of capacitive, inductive, and resistive elements or combinations thereof. The specific configuration is dependent on the particular application or requirement. The most common example of a simple filter is a capacitor connected between a signal point and ground, effectively bypassing unwanted higher frequencies. To reject unwanted higher frequencies with an inductor, it is placed in series with the line. If more effective filtering or special passband characteristics are desired, complex filters containing combinations of circuit elements may be used.

In addition to these passive filter designs, there are currently in use a group of devices termed active filters. Basically, the filtering elements are the same as those in the passive networks, i.e., resistors, capacitors, and inductors, but in addition, active circuit elements are employed to produce isolation, impedance transformation, and feedback loops which enhance the desirable filter properties. Operation of active filters is generally similar to that of passive filter networks.

Except for the simple capacitor filter, filters are usually inserted in a line so that all energy carried by the line passes through the filter. The filter therefore must be capable of performing its function without disturbing the operation of the line in which it is inserted. Some considerations in the selection or design of filters are listed as follows:

- a. Voltage and current ratings.
- b. Frequencies to be attenuated.
- c. Amount of attenuation required.
- d. Effects of filter on desired signal.

Filters should be installed as close as possible to the source of interference or to the circuit which is to be protected. They should be well shielded and grounded, and the lead length and ground returns should be as short as possible.

5. Bonding and Grounding

Bonding is a method of establishing a low-impedance electrical path for interference currents between metallic parts and equipment. Grounding, on the other hand, refers solely to the electrical connection of metallic parts and structures to earth potential to insure that all connected points will be at the same earth potential. The techniques involved in bonding and grounding are similar. The methods employed and the materials used are often the same.

In general, thin (0.010-inch), solid-strap-type conductors should be used for bonding. However, where excessive vibration is present, a braided strap or stranded conductor is used to avoid breakage from crystallization. Braided or stranded wire is not recommended generally because the individual strands of wire tend to corrode or break, creating conditions that could cause additional interference. Conductors should be as short as possible without mechanical strain. Joined surfaces must be clean and free of dirt or paint. Dissimilar metals should be avoided wherever possible.

When two units are connected through a bond strap, the length of the strap has some value of inductance. The capacitance between bonded members is in parallel with the inductance of the strap, and the impedance will be very high at the resonant frequency of this combination of inductance and capacitance. The resonant frequency should be placed beyond the frequency range of interest by keeping the inductance low through the use of short straps with a width-to-length ratio of at least 1 to 5 and not exceeding the maximum thickness allowable. Jumpers should be as direct as possible and, when practicable, should not exceed three inches in length. Sheet-metal-type screws should not be used. Connections should be soldered or bolted with tooth-type lockwashers to insure proper metal-to-metal contact.

Each electronic system should have a common system ground connection to earth. The resistance from this system ground point to all interconnected units of the electronic system should not exceed 0.5 ohm. The ground point should be, preferably, a direct earth connection. Since this is not always possible, electrical conduits, water pipes, and steam pipes often are used. If tap water is in contact in any way with the input circuit, the cold water pipe must be used as the ground. In some cases, there may be a large potential difference between the water system and the ground wire of the electrical system. In this case, the two systems may have to be electrically connected at the transformer or circuit breakers of the power system. Occasionally, an auxiliary ground system may be necessary if the power circuit ground wire is carrying a heavier load than that for which it was designed. The three-wire conductors that are now commonly used to furnish safety grounds for electrical equipment may be sources of ground loops, particularly if a cold water pipe is used as the reference ground.

Section IV

FUTURE TRENDS

GENERAL

The remaining pages of this handbook discuss future developments which may affect the field of physiological monitoring. Like most engineering disciplines, the instrumentation for physiological measurements is not static, but exists in a state of flux with constant growth and change. Measurements that are commonplace today may well be supplanted by others which, although presently unfeasible, may become possible because of a breakthrough in technology.

Most developments will not be the dramatic type that open whole new lines of investigation. Gradual improvements in equipments and techniques will contribute slowly to the overall state of the physiological monitoring art. These inevitable improvements should be considered when evaluating the limitations of available equipment in present-day measurement situations. The trends of new developments are discussed briefly in the following paragraphs.

I. Refinements in Equipment

Generally, equipment can be improved in three ways: it can be miniaturized, its accuracy can be increased, and its reliability can be improved. There has been steady improvement in the techniques of miniaturization and a correspondingly steady decrease in the size of components. As a result, systems have become less complex (by virtue of the ease with which multiple system functions can be packaged into single units), and requirements for power to operate these systems have been reduced. Miniaturization is especially desirable for sensors, signal conditioners, and telemetry components. Among the obvious advantages are the following:

- More instrumentation can be provided in spacecraft without increase in payload.
- Increased capability and wider use of wireless monitoring from an active subject are possible.
- Miniature computers will permit more data processing at the source end of a system. Among other advantages, more efficient use of the telemetry bandwidth becomes possible in remote monitoring applications.

Improvements in the accuracy and reliability of measurements are also foreseeable, because designers and manufacturers are constantly tightening specifications and

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improving component performance. Improved linearity, dynamic response, and temperature stability (among other parameters) can all contribute to consistency of experimental results.

II. New Sensing and Measuring Techniques

Among the more direct contributions to monitoring technology will be the application to continuous monitoring or measurement techniques which are presently experimental in nature or limited presently to clinical applications. Among the possible developments are the following:

Radiation monitoring. Monitoring of incident radiation will become increasingly important in support of space exploration. Clinical techniques of radiology and radiography may be applicable to continuous monitoring as radiation dosage is reduced.

Ultrasonic techniques. Ultrasonic blood flowmeters and ultrasonic methods for measuring and observing internal organs may be extended to remote monitoring situations.

Magnetic techniques. The use of paramagnetic resonance and nuclear magnetic resonance techniques to measure blood flow in intact limbs and organs may be extended to continuous monitoring applications.

It may be possible to sense physiological variables at (at least) short distances from the subject without actually attaching sensors to the body of the subject. Non-contacting transducer applications may be severely limited in scope, but they demonstrate trends in instrumentation.

Finally, there probably will be developments in applications of the radio pill (endoradiosonde). This method of measuring from within a subject such parameters as temperature, pressure, gaseous concentrations, etc, and accomplishing the wireless transmission of measured data, undoubtedly will be improved greatly as sophisticated circuitry is miniaturized, and as continuing experiments permit more successful correlation and interpretation of the data obtained.

III. Improved Techniques in Data Processing and Transmission

Improvements in data transmission (telemetry) should include (1) an increase in the transmission range of small personal telemetry systems and (2) an increase in the bandwidth of telemetry links, and/or an improvement in the utilization of existing bandwidths through more sophisticated signal modifying, coding, and modulation techniques.

Data processing, especially when employed at the transmitting end of a monitoring system, will be useful in improving system bandwidth capabilities. More importantly,

NEW SENSING AND MEASURING TECHNIQUES

perhaps, data processing will increase the utility of monitoring systems by the multiple, high-speed operations it can perform on the data obtained by the system. By the high-speed reduction, correlation, and interpretation of data made possible with digital computers, useful measurement of many parameters not presently feasible will be commonplace. For example, it will be possible to measure quantitatively many complex, interrelated functions such as psychological or behavioral responses that are evidenced by measurable physiological variables. Also, environmental control and life support systems can be controlled directly and automatically by computer operations using inputs from physiological monitoring systems.

NEW SENSING AND MEASURING TECHNIQUES

I. Ultrasonics

Ultrasound or ultrasonic vibrations are not new in the field of medicine, but they have not yet been employed successfully in continuous monitoring applications. Ultrasound, which may be defined as vibrations in excess of 20,000 cycles per second (commonly in the megacycle range in current applications), is used in therapy, in surgery, in diagnosis, and in biological measurements. The ultrasonic techniques used in therapy and surgery are active; that is, ultrasonic vibrations are used to alter or change biological tissue. Physiological monitoring is concerned primarily with passive techniques, in which ultrasonic vibrations may interact with but not modify biological tissue as a means of obtaining information on the structure and function of that tissue. (See also ref. 15 and 20.)

Mainly, two types of interaction are involved: one is the absorption of ultrasound by gross tissue structures, and the other is the reflection of ultrasound by a discontinuity in internal structure. The first phenomenon, the absorption of ultrasound by gross tissue structures, can be used in basic physiological research for charting the acoustic propagation characteristics of normal (and pathological) tissues. This information can be used for examination of gross anatomy and organ dynamics in living organisms. With sufficient data, internal malfunctions can be diagnosed, and deep body tissue can be altered surgically by focusing ultrasonic techniques without substantially affecting intervening tissue structures.

The second type of interaction, the reflection of ultrasound by a discontinuity in internal structure, makes possible a technique for visualizing internal structures that is comparable to X-ray methods. The difference in absorption characteristic between hard and soft tissue makes X-ray visualization possible. X-ray techniques, however, cannot be used to visualize or differentiate between different types of soft tissue, since all soft tissue has essentially the same X-ray density. By ultrasonic means, soft tissue structures can be distinguished, because they are not acoustically homogenous. When ultrasound is beamed into a region of the body, some ultrasonic energy is reflected from the interfaces between soft tissue structures. With appropriate scanning and

display circuitry, three-dimensional, dynamic displays can be obtained which permit the study of function as well as structure (ref. 8).

Ultrasonic scanning is presently used in ophthalmology, in the localization and diagnosis of tumors, and in blood flow measurement. It may have application, with further development, to continuous monitoring situations. One technique would be an extension of a continuous-wave reflection device developed by the Japanese which is a form of phonocardiography. Sound waves are beamed from the chest wall through the opening between a pair of ribs to the heart. The receiving equipment responds to the frequency shift produced by motion of the heart wall in the direction of propagation of the sound waves. Used in conjunction with the ECG and the standard phonocardiograph, this technique can provide detailed information on parameters such as valve timing which would not otherwise be obtainable.

The use of pulsed rather than continuous ultrasound has advantages in that significant measurements can be made in terms of transit time (a function of attenuation over distance) rather than phase shift, which can be difficult to measure when very small shifts are involved. Pulsed ultrasound techniques are not limited to visualization applications. Pulsed vibrations are employed in ultrasonic blood flowmeters. In this clinical application, an ultrasonic transducer unit is attached to a blood vessel through surgery. While such procedures are not suitable for remote, aerospace applications, this measurement technique is of interest because of its future possibilities.

In the ultrasonic flowmeter, two transducers are employed (typically, barium titanate piezoelectric crystals). They are placed at opposite ends of the transducer housing (fitted to the blood vessel) and serve alternately as transmitter and receiver. The transit time of ultrasound between the transducers is measured first upstream and then downstream, yielding a difference figure that represents the blood flow velocity. Assuming a constant cross-sectional area in the blood vessel, this difference figure can be calibrated in units of blood flow (ref. 19).

II. Radiation Monitoring

Radiation monitoring techniques are of interest in aerospace medicine: one technique is the monitoring of environmental or incident radiation; another is the use of radiation detectors to monitor the movement of radioactive tracer elements through a subject's body. Neither are, strictly speaking, physiological monitoring techniques, nor are they always suitable for continuous monitoring applications. But they are both closely related to physiological studies in aerospace environments.

With manned space operations advancing to high orbits and ultimately to lunar probes and exploration, the monitoring of space radiation becomes increasingly important. A recent Air Force* procurement for the design and construction of a space

*Aeromedical Field Laboratory, Holloman AFB, New Mexico.

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radiation monitoring system was keyed specifically to the energy ranges of ionizing radiation that are biologically significant (i.e., damaging to human tissue). Figure 55 is a block diagram of this system (ref. 14).

In this system (refer to the block diagram) two solid state detectors supply pulse outputs which are applied to logic and pulse height analysis circuits. The logic circuitry identifies the pulses as gamma radiation, alpha particles, neutrons, electrons, protons, or heavy nuclei, and further categorizes them with arbitrarily selected energy levels. The pulses then are fed into appropriate sections of a binary storage and gating unit. The outputs of the storage unit (18 categories of radiation plus dose information)

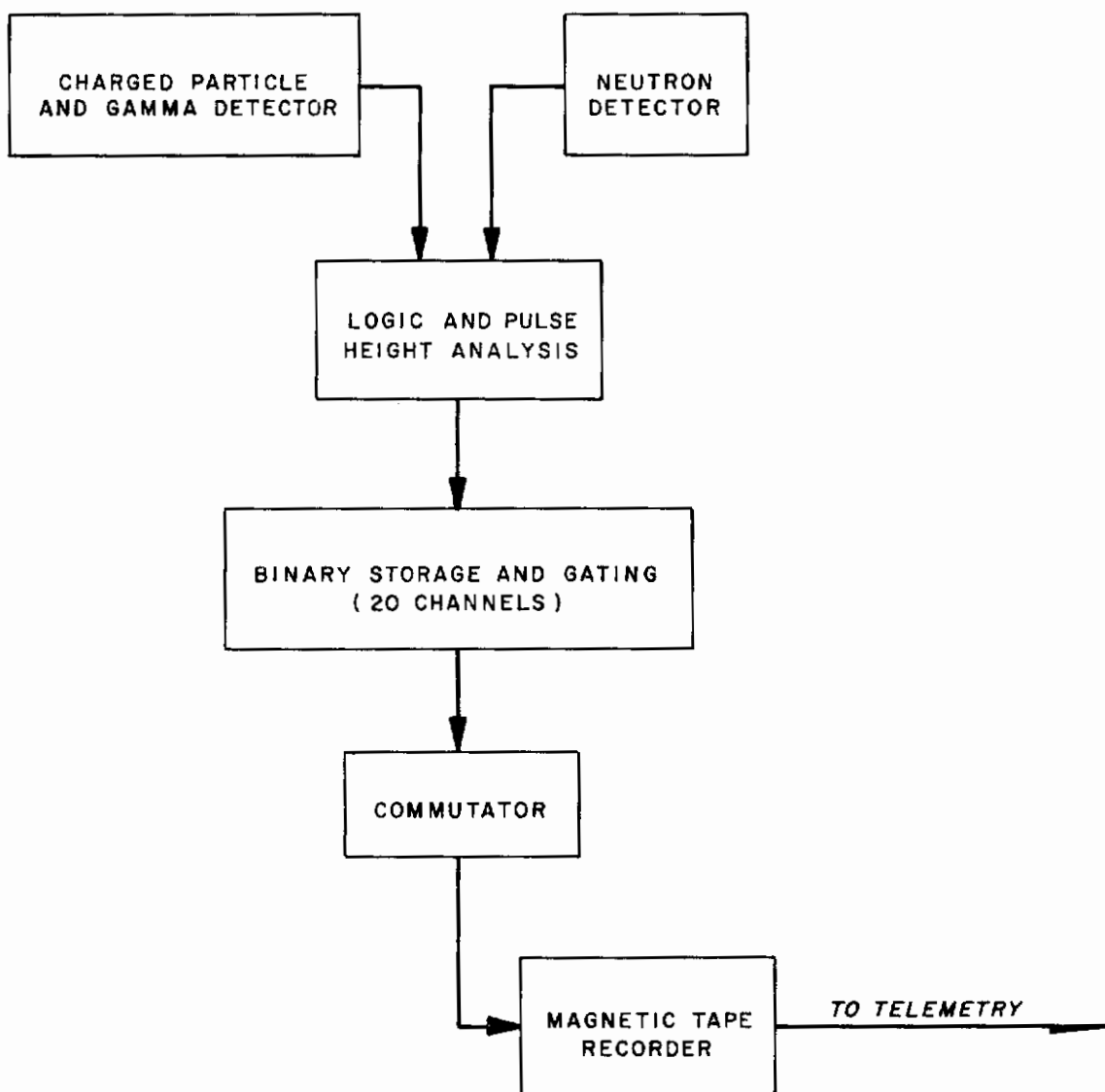


Figure 55. System for Monitoring of Space Radiation

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are commutated and fed to a magnetic tape recorder for subsequent telemetering to a ground station. A record-to-playback speed ratio of 1:18 will permit the transmission of large quantities of stored and recorded data in rapid bursts during those periods of ideal conditions for communications.

The system includes an ingenious method of identifying radiation by solid state detectors. The gamma and charged particle detector, for example, is essentially a coincidence telescope consisting of ceramic magnets; a brass collimator; a lead shield; a lead glass light pipe; and three sensing elements, consisting of a p-n junction detector, an iodide (CsI) scintillator, and a plastic scintillator. The presence or absence of coincidence pulses from these sensing elements determines the identification of the incident radiation and, in part, the level of the radiation. This system permits the measurement of ionizing radiation of biological significance with a degree of discrimination and precision previously unobtainable.

The use of radioactive tracer elements is basically a clinical practice that needs refinement before it can be applied in aerospace environments. In clinical practice a tracer element, such as radioactive iodine, is prepared and introduced into some body system of the subject (orally, intravenously, etc). Suitable radiation detectors, such as Geiger or scintillation counters, measure the amounts of tracer element deposited in certain parts of the body or follow the course of the element through the body system. The clinical radiological equipment, including scintillation crystals, photomultipliers, and metering and scanning circuits, will have to be made compatible with other miniaturized bioinstrumentation; in addition, some means of maintaining orientation between detecting equipment and the subject's body will have to be provided for continuous monitoring applications. These developments will be accelerated by need. For example, tracer techniques may offer one means of studying metabolic processes under conditions of sustained weightlessness (ref. 73).

III. Magnetic Monitoring Techniques

With the present state of the art, there are few magnetic monitoring techniques suitable for aerospace medical application. However, there has been research in the use of this technique for the measurement of at least two phenomena, blood flow and oxygen partial pressures. The feasibility of continuous magnetic monitoring is a distinct possibility.

The National Institute of Health has developed a flowmeter for experimental, clinical use to measure mass blood flow in an intact limb, employing nuclear magnetic resonance techniques (ref. 41). This type of measurement depends upon the magnetic susceptibility of the hydrogen protons in the blood. The protons, which normally are randomly oriented, will absorb energy from a magnetic field, become resonant, and then align themselves. If measured bursts of r-f energy are passed through a limb, the amount of r-f energy absorbed will be a function of the number of protons passing through the limb in a given time and, by extension, a measurement of the regional blood flow.

NEW SENSING AND MEASURING TECHNIQUES

Paramagnetic techniques also have been applied to the measurement of oxygen partial pressures. The concept here is that the magnetic susceptibility of oxygen, while minute, is about one hundred times greater than that of other respiratory gases; therefore, a measurable indication of the interaction of a magnetic field and respiratory oxygen can be obtained. A typical method for instrumenting this measurement might be to couple a paramagnetic body, on a piezoelectric crystal, to an external magnetic field, with the degree of coupling dependent upon the oxygen partial pressure (ref. 75).

IV. Endoradiosondes

The endoradiosonde is a tiny measuring device which can be swallowed or implanted in a subject to obtain physiological data from within the body. It can be used to monitor variables in inaccessible locations that previously could be measured only through the use of such objectionable devices as catheters or esophageal tubes. The endoradiosonde couples data from the internal site to external monitoring equipment without wires or connections of any kind. Thus, the subject under observation is left in a relatively normal physiological state.

The term endoradiosonde (or radio pill) has come into usage because many of the ingested devices used for monitoring include a self-contained radio transmitter. Actually, many measurements of this type are possible by the ingestion of a capsule containing only passive circuit elements (and a suitable transducer), and the transmission of data occurs by the interaction of the circuit elements with an externally supplied r-f field.

While the endoradiosonde can accomplish certain measurements not otherwise possible, the technique is still in its infancy. There are numerous limitations. Range is short and subject mobility is sometimes prohibited. There is a scarcity of quantitative data, and there is sometimes a lack of correlation between measurements obtained by endoradiosondes and those obtained by classical clinical techniques. (See also ref. 33 and 46.)

A. Transducers

Measurements with an endoradiosonde are made most effectively by using a transducer technique that will produce a change in the reactance of a tuned circuit. A temperature-sensitive capacitor is an obvious example, or there are numerous direct (and indirect) methods of displacing a core or coil in a variable inductor. In the active-type endoradiosonde, the reactance change is converted directly into a frequency shift in a simple FM transmitter.

B. Active Transmission

An active endoradiosonde contains a transducer, modulator, transmitter, and

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power supply. In a typical pressure-sensitive radio pill, the active circuitry consists of an oscillator circuit containing a transistor, capacitors, an r-f coil, and a ferrite disc core. The ferrite core is attached to a rubber or plastic diaphragm, which is displaced by pressure changes in the surrounding fluid or tissue. Power normally is supplied by a small battery within the capsule. The battery will determine the active life of the capsule, which may vary in mercury cells from 8 hours to 3 weeks. Experiments have been conducted using a nickel-cadmium cell that could be recharged by an external oscillator. (It is also possible to eliminate the battery altogether, and supply power for transmission by exciting a coil in the endoradiosonde with r-f energy.

Endoradiosonde transmitters generally operate from 0.5 megacycle to 10 megacycles, with the optimum carrier frequency depending upon the specific type of monitoring system used. FM is the most commonly used type of modulation, but other types (except for AM) may be employed. Transmission is possible using a magnetic dipole within the capsule; in some instances, the body of the subject has been employed as an electric dipole for transmitting data.

Reception is possible with ordinary FM receivers, modified to the particular transmission frequency and modulation technique. Antenna techniques may be the chief problem; with the small signal available from the capsule, and the changes in orientation that the capsule may undergo, multiple and omnidirectional antenna configurations may be necessary.

C. Passive Transmission

The passive endoradiosonde consists only of a resonant circuit and a suitable transducer which changes the reactance of the circuit. Data transmission can be accomplished in two ways with the passive circuit: by frequency sensitive absorption and by re-emission. In the first method, the resonant frequency of the circuit can be sensed by grid-dip metering techniques. Using the re-emission technique, the circuit can be excited by short pulses of r-f energy from an external oscillator, producing a ringing frequency which is re-emitted between pulses.

D. Measurements

1. Pressure

To date the most widespread use of the endoradiosonde has been in the measurement of gastro-intestinal pressure. Transduction is accomplished directly, as indicated above, by attaching a moving part (such as a core piece) of an inductive element to the flexible diaphragm of a small pressure vessel.

2. Temperature

Next to pressure, temperature is the most commonly measured parameter.

A simple transducing technique, a temperature-sensitive capacitor in a tuning circuit, will vary with temperature directly, but sensitivity here is limited to rather large temperature variations, which are uncommon in the human subject. A somewhat more sensitive response can be obtained by less direct methods: for example, small temperature variations can be made to produce, with the use of low-boiling-point liquids in a sealed chamber, large pressure variations, which then can be sensed with greater resolution by the pressure-measuring technique described above.

3. Other Variables

Several other variables have been monitored or measured by endoradiosonde techniques. Certain of these, such as pulse and respiration rates and ECG potentials, depend upon the careful localization of the capsule, requiring that the less desirable technique of implantation be used rather than simple swallowing. Certain chemical detection also has been possible on such variables as chloride ion concentration, and partial pressure of hydrogen and oxygen. These latter measurements usually are accomplished by transducer techniques yielding displacement or dimensional changes, which can be sensed by pressure-type diaphragms. Radiation detection also has been suggested as being possible, using an endoradiosonde with a modified transmitter circuit containing a radiation-sensitive semiconductor.

E. Problems in Using Endoradiosondes

Several difficulties involved in the use of endoradiosondes have been suggested previously; the following are others that are worthy of consideration:

1. Localization

The capsule can be positioned precisely by implantation (instead of swallowing), but there are obvious psychological objections. Indirect monitoring of position by fluoroscopy or X-rays yields only relative information, and places other practical limits on the measurement situation.

2. Oscillator Frequency Drift

Oscillators will drift in frequency for several reasons: the transistors used may be temperature sensitive or, in the pressure-measuring devices, unwanted variations may be introduced by leakage at the diaphragm or by changes in atmospheric pressure.

3. Signal Strength

Signal strength depends upon the battery voltage and the orientation of the capsule with the receiving (or exciting) antenna. As a result, the transmitted signal normally is quite weak, which limits the effective operating range considerably.

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4. Delayed Passage

There may be some difficulty in using an endoradiosonde to measure gastrointestinal processes accurately. There is evidence that capsules do not pass through the system at the same rate as food. The capsule is apparently sensed as a "foreign body" in the alimentary canal, and the normal functions which it is intended to monitor are accordingly disrupted.

V. Noncontacting Transducers

One of the ever-present problems in physiological measurement is the effect of the act of measurement upon the measured parameter (discussed on page 13). All sensors will to some extent load the system being measured. Careful electrode and transducer techniques and the miniaturization of sensor components can minimize, but never eliminate this problem. It is interesting to speculate to what extent physiological measurements may be carried out with no transducer in actual contact with the subject (ref. 7).

The limitations to such applications probably are more apparent than the possibilities. The distance over which surface-detectable variables could be detected would be quite short -- no more than a few feet, perhaps. Also, unless the subject were completely immobilized, most measurements probably would require scanning systems of prohibitive sophistication and resolution.

There are surface variables, however, that can be detected, without the transducer contacting the subject. The following are measurements that may have promise:

- Profile changes. One of several scanning techniques, such as the spot edge follower, might be employed to detect surface indications of such variables as respiration, heart rate, and blood pressure.
- Temperature. Passive infrared detection of radiated energy might be made quantitative, over the small range of human variation, permitting absolute temperature measurements or, with scanning techniques, monitoring of temperature distribution over the body of a subject.
- Surface water. The presence of perspiration on the skin might be measured by some technique such as the reflection of microwaves at the absorption frequency of water. As an indication of sweat gland activity, such a measurement might furnish an index of psychological state, similar to the galvanic skin reflex.
- Light reflection. Reflection oximetry techniques might be extended to obtain pulse variations at a distance.

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- **Electromagnetic induction.** An induction coil near or around a subject might detect circulating currents caused by heart potentials, yielding a remote ECG pickup. The magnitude of the signals would be quite small, and range probably would be limited to inches from body surfaces.

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With the ultimate development of new sensing techniques, a whole new gamut of data will be available for the evaluation of human physiological responses to stressful environments. However, the data-handling capability of the transmission link in a remote monitoring situation may be overtaxed, and facilities for interpretation of data at the output may be inadequate for any kind of real-time response to the demands of the measurement situation.

Many monitoring systems employ the standard technique of continuously recording amplitude-versus-time variations for each of many variables throughout the course of a given experiment. The volume of recorded data thus produced is impressively large, but in many applications most of these data are useless. The small amount of significant change data that ultimately may be extracted indicates a waste of transmission system bandwidth and transmission time, a waste of graphic write-out capacity, and a waste of time and energy on the part of the operator who reduces these data manually.

There have been improvements in the display, recording, and processing of physiological measurements, but more are needed. Digital displays now present data on many variables in easy-reading format. Relatively simple differentiating and integrating circuits perform preliminary operations on other variables before transmission or recording. Also, the measurement of some parameters is being instrumented to provide "alarm" type indications when certain predetermined stress thresholds are exceeded.

But all too often, data evaluation still consists of the classical examination of traces by a clinician, drawing on his knowledge, insight, and experience, and perhaps burdened with the necessity of simultaneously correlating multiple physiological and environmental variables, all of which may be interacting. Such techniques obviously will not satisfy the requirement of monitoring systems whose primary purpose is the protection of human life. There could be a potentially dangerous time lag between the real-time acquisition and display of data and the evaluation of the display for "alarm" level indications and appropriate command responses.

Several improvements may solve these problems in monitoring systems of the future. One is the improvement in the bandwidth capability of data-transmission systems, or the more efficient use of bandwidth capability by more sophisticated coding and modulating techniques (discussed in Section V of Volume II). Also, preliminary processing should be augmented before the transmission of data to reduce, by selection or elimination, the amount of data that has to be transmitted, displayed, and evaluated. Such

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processing may take the form of increased signal conditioning, wider use of logic circuitry in initiating or withholding transmissions, and wider use of pretransmission computer operations for the same purpose.

Ultimately, computer operations will be employed for decisions and for the initiation of command responses in monitoring systems used for the protection of human operators (ref. 6). The computers would supply feedback to environmental control and vehicular control systems, making possible rapid, automatic corrections in overly stressful situations.

With respect to this type of development, a recent procurement of the National Aeronautics and Space Administration is of interest. NASA has contracted* for the development of a complete monitoring package called PIAPACS (Psychophysiological Information, Acquisition, Processing, and Control System). This system is intended for the support of advanced manned space flight vehicles by sensing, correlating, predicting, and displaying all significant physiological psychological (or behavioral), and environmental parameters related to protecting a human subject in a stressful environment.

NASA hopes with PIAPACS to advance the state of the art by making possible hitherto unfeasible measurements, and by establishing meaningful norms and limits on many interrelated parameters. The system also will use an on-board microminiature digital computer which can, from real-time measurement data, determine when certain parameters are approaching a predetermined stressful limit, and send command information to environmental control equipment to maintain the subject below stress levels.

As a final note in this handbook, the possible future application of physiological monitoring techniques in such related areas as bionics should be mentioned. A recent communication (ref. 68) has described a servo boost system which, operating from myoelectric signal inputs, can provide mechanical assistance in positioning an operator's hand in a high-g transverse acceleration field. Experiences with fly-by-wire operations in Project Mercury have demonstrated that manual control of space vehicles is sometimes necessary. Yet high-g conditions, which sometimes occur during space flight, may make such control difficult if not impossible. The servo boost system is intended to overcome g forces.

The system is shown schematically in figure 56. The operator's arm is supported in a splint that can be positioned up and down by a uniplanar power drive. (Movement of the splint is hindered by springs which simulate the restraining force of the acceleration field.) EMG electrodes are positioned on the arm over those muscles that would be used to perform the desired movement. The output from the EMG signal conditioners

* Lear Siegler, Inc., Santa Monica, Calif., in response to NASA Proposal PR-3175, has obtained the contract for this development.

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is fed to a control logic computer which, when it receives the correct EMG input, supplies on-off signals to the power drive to position the splint.

The chief problem in this project was to obtain clear, unequivocal control inputs from the EMG pickups. Once characteristic myoelectric patterns were determined, transforms were performed on the raw signals, then control logics were written relating the myoelectric signals to the desired servo response.

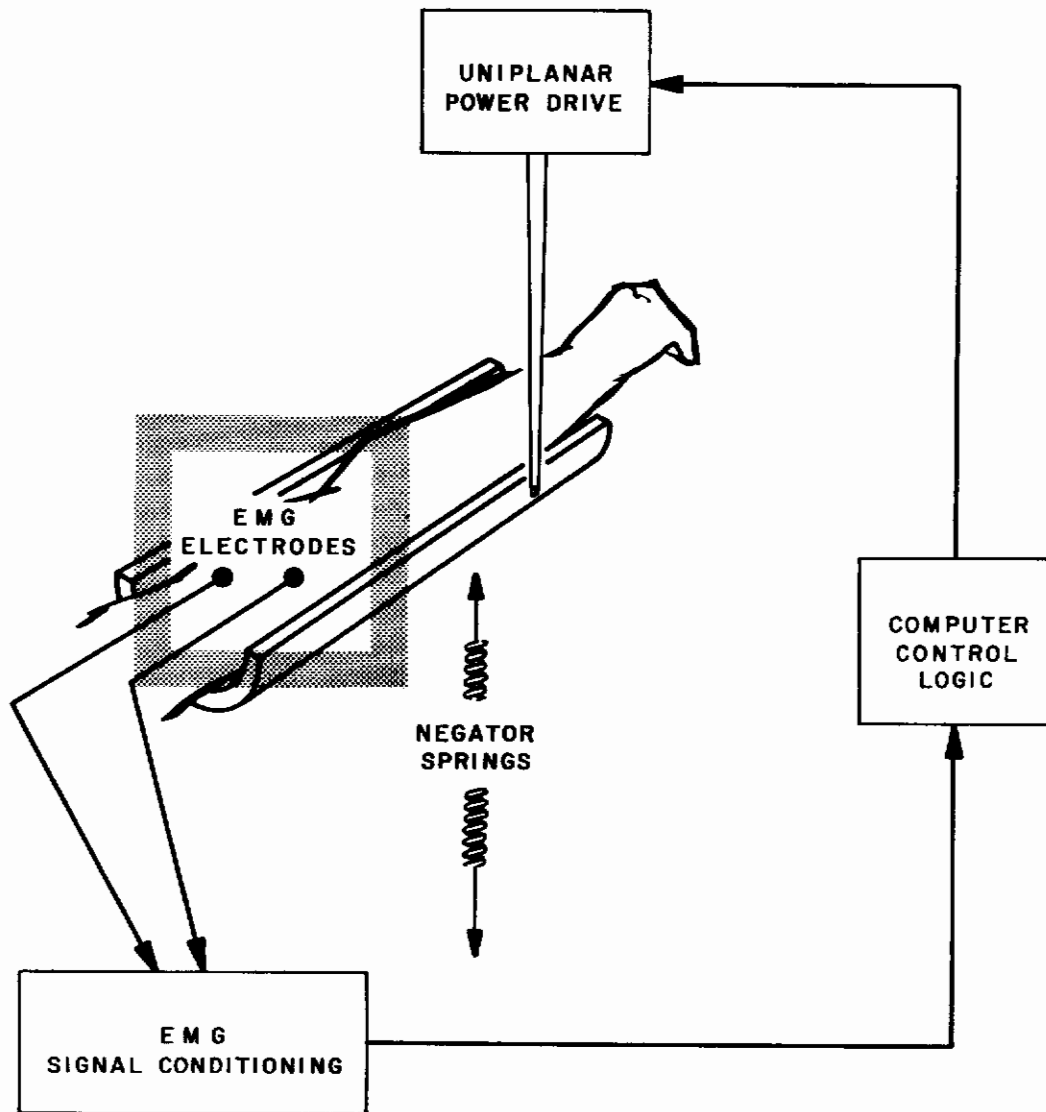


Figure 56. EMG - Servo Control System

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DOCUMENT CONTROL DATA - R&D		
(Security classification of title, body of abstract and indexing annotation must be entered when the overall report is classified)		
1. ORIGINATING ACTIVITY (Corporate author) RCA Service Company Cherry Hill Camden 8, New Jersey	2a. REPORT SECURITY CLASSIFICATION UNCLASSIFIED	
	2b. GROUP N/A	
3. REPORT TITLE TECHNIQUES OF PHYSIOLOGICAL MONITORING VOLUME III. SYSTEMS		
4. DESCRIPTIVE NOTES (Type of report and inclusive dates) State-of-the-art survey, in preparation July 1963 - August 1964		
5. AUTHOR(S) (Last name, first name, initial) Alnutt, Richard W. Becker, Walter C. Barbieri, Robert E., Editor		
6. REPORT DATE October 1964	7a. TOTAL NO OF PAGES 158	7b. NO. OF REFS 80
8a. CONTRACT OR GRANT NO. AF 33(657)-9252 b. PROJECT NO 7222 c. Task No. 722203 d.	9a. ORIGINATOR'S REPORT NUMBER(S) 9b. OTHER REPORT NO(S) (Any other numbers that may be assigned this report) AMRL-TDR-62-98 (III)	
10. AVAILABILITY/LIMITATION NOTICES Qualified requesters may obtain copies of this report from DDC. Available, for sale to the public, from the Office of Technical Services, U.S. Department of Commerce, Washington, D. C. 20230.		
11. SUPPLEMENTARY NOTES	12. SPONSORING MILITARY ACTIVITY Aerospace Medical Research Laboratories, Aerospace Medical Division, Air Force Systems Command, Wright-Patterson AFB, Ohio	
13. ABSTRACT This volume is a discussion of monitoring systems. While the applications of physiological monitoring are many and varied, the primary concern here is with viability monitoring, the use of a measurement system to obtain factual, quantitative information about the physiological responses of a subject in a stressful environment, in order to plan protective measures which will ensure the safety and functional capability of that subject in such environments. Included are a description of instrumentation required for the measurement of individual physiological parameters, a discussion of related problems in system design, including simultaneous measurement of several parameters, data transmission or telemetry, and the use of magnetic tape recording as a system adjunct. Basic guidelines of system troubleshooting and interference reduction are also included. Section IV contains a brief survey of additional measurement techniques and data handling considerations which, while not state of the art or standard practice, will undoubtedly affect the field of physiological monitoring in the near future.		

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1 JAN 64
AF-WP-B-AUG 64 400

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14.	KEY WORDS	LINK A		LINK B		LINK C	
		ROLE	WT	ROLE	WT	ROLE	WT
	<p>Medical Electronics, Monitoring Medical Research, Instrumentation Instrumentation, Biomedical Monitoring, Physiological</p> <p>Space Flight, Medical Aspects Spacecraft, Instrumentation</p> <p>Cardiovascular System, Measurement Respiratory System, Measurement Central Nervous System, Measurement Body Temperature, Measurement</p>						
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