A

BOOK

OF

DIAGNOSTIC

MEDICAL

INSTRUMENTATION

(By Viral Sir)

❖ INDEX :

Ch-1 Medical Instrumentation System

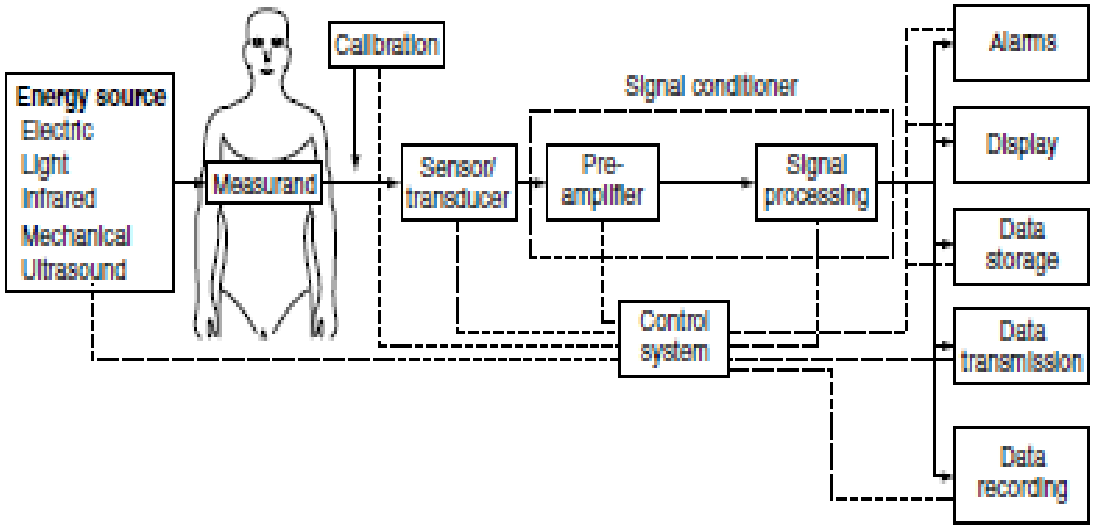


Fig. 1.9 *General block diagram of a medical instrumentation system*

The primary purpose of medical instrumentation is to measure or determine the presence of some physical quantity that may some way assist the medical personnel to make better diagnosis and treatment. Accordingly, many types of instrumentation systems are presently used in hospitals and other medical facilities. The majority of the instruments are electrical or electronic systems, although mechanical systems such as ventilators or spirometers are also employed. Because of the predominantly large number of electronic systems used in medical practice, the concepts explained hereafter are mostly related to electronic medical instruments. Certain characteristic features, which are common to most instrumentation systems, are also applicable to medical instrumentation systems. In the broadest sense, any medical instrument (Fig. 1.9) would comprise of the following four basic functional components:

- Measurand:

The physical quantity or condition that the instrumentation system measures is called the measurand. The source for the measurand is the human body which generates a variety of signals. The measurand may be on the surface of the body (electrocardiogram potential) or it may be blood pressure in the chambers of the heart.

- Transducer/Sensor:

A transducer is a device that converts one form of energy to another. Because of the familiar advantages of electric and electronic methods of measurement, it is the usual practice to convert into electrical quantities all non-electrical phenomenon associated with the measurand with the help of a transducer. For example: a piezo-electric crystal converts mechanical vibrations into an electrical signal and therefore, is a transducer. The primary function of the transducer is to provide a usable output in response to the measurand which may be a specific physical quantity, property or condition. In practice, two or more transducers may be used simultaneously to make measurements of a number of physiological parameters.

Another term 'sensor' is also used in medical instrumentation systems. Basically, a sensor converts a physical measurand to an electrical signal. The sensor should be minimally invasive and interface with the living system with minimum extraction of energy.

- Signal Conditioner:

Converts the output of the transducer into an electrical quantity suitable for operation of the display or recording system. Signal conditioners may vary in complexity from a simple resistance network or impedance matching device to multi-stage amplifiers and other complex electronic circuitry. Signal conditioning usually include functions such as amplification, filtering (analog or digital) analog-to-digital and digital-to-analog conversion or signal transmission circuitry. They help in increasing the sensitivity of instruments by amplification of the original signal or its transduced form.

- Display System:

Provides a visible representation of the quantity as a displacement on a scale, or on the chart of a recorder, or on the screen of a cathode ray tube or in numerical form. Although, most of the displays are in the visual form, other forms of displays such as audible signals from alarm or foetal Doppler ultrasonic signals are also used. In addition of the above, the processed signal after signal conditioning may

be passed on to:Alarm System—with upper and lower adjustable thresholds to indicate when the measurand goes beyond preset limits.
Data Storage—to maintain the data for future reference. It may be a hard copy on a paper or on magnetic or semiconductor memories.

Data Transmission—using standard interface connections so that information obtained may be carried to other parts of an integrated system or to transmit it from one location to another. In most of the medical instrumentation systems, some form of calibration is necessary at regular intervals during their operation. The calibration signal is usually applied to the sensor input or as early in the signal conditioning chain as possible. In many measurements in the medical field, some form of stimulus or energy is given to the patient and the effect it has on the patient is measured. The stimulus may be visual in the form of flash of light or audio tone or direct electrical stimulation of some part of the nervous system. A typical example is that of recording of the evoked response with EEG machine when visual/audible stimulus is given to the subject under test. In some ituations, it is required to have automatic control of the transducer, stimulus or signal conditioning part of the system. This is achieved by using a feedback loop in which part of the output from the signal conditioning or display device is fed back to the input stage. Control and feedback may be automatic or manual. Almost all measuring and recording equipment is now controlled by microprocessors as this makes it possible to design equipment that requires minimal user intervention, calibration and set up procedure.

Measurements on the human body can be made at several levels on the functional systems and sub-systems. For example; it is easiest to make measurements on the human body as a whole due to accessible environment. Examples of measurement made on the human body are recording of electrocardiogram and measurement of temperature. The next level of measurements can be made on the major functional systems of the body such as the cardiovascular system, the pulmonary system and so on. Many of the major systems communicate with each other as well as with external environment. The functional systems can be further sub-divided into sub-systems and organs and still smaller units up to the cellular and molecular level. Measurements in the medical field are made at all these levels with specially designed instruments with appropriate degree of sophistication. Measurements in the medical field can be classified into two types: in vivo and in vitro. In vivo measurement is made on or within the living organism itself, such as measurement of pressure in the chambers of the heart. On the other hand, in vitro measurement is performed outside the body. For example; the measurement of blood glucose level in a sample of blood drawn from the patient represent in vitro measurement.

Ch-2 Electro Cardio Graph(ECG)

The electrocardiograph (ECG) is an instrument, which records the electrical activity of the heart. Electrical signals from the heart characteristically precede the normal mechanical function and monitoring of these signals has great clinical significance. ECG provides valuable information about a wide range of cardiac disorders such as the presence of an inactive part (infarction) or an enlargement (cardiac hypertrophy) of the heart muscle. Electrocardiographs are used in catheterization laboratories, coronary care units and for routine diagnostic applications in cardiology. Although the electric field generated by the heart can be best characterized by vector quantities, it is generally convenient to directly measure only scalar quantities, i.e. a voltage difference of Mv order between the given points of the body. The diagnostically useful frequency range is usually accepted as 0.05 to 150 Hz (Golden et al 1973). The amplifier and writing part should faithfully reproduce signals in this range. A good low frequency response is essential to ensure stability of the baseline. High frequency response is a compromise of several factors like isolation between a useful ECG signal from other signals of biological origin (myographic potentials) and limitations of the direct writing pen recorders due to mass, inertia and friction. The interference of nonbiological origin can be handled by using modern differential amplifiers, which are capable of providing excellent rejection capabilities. CMRR of the order of 100-120 dB with 5 kW unbalance in the leads is a desirable feature of ECG machines. In addition to this, under specially adverse circumstances, it becomes necessary to include a notch filter tuned to 50 Hz to reject hum due to power mains. The instability of the baseline, originating from the changes of the contact impedance, demands the application of the automatic baseline stabilizing circuit. A minimum of two paper speeds is necessary (25 and 50 mm per sec) for ECG recording.

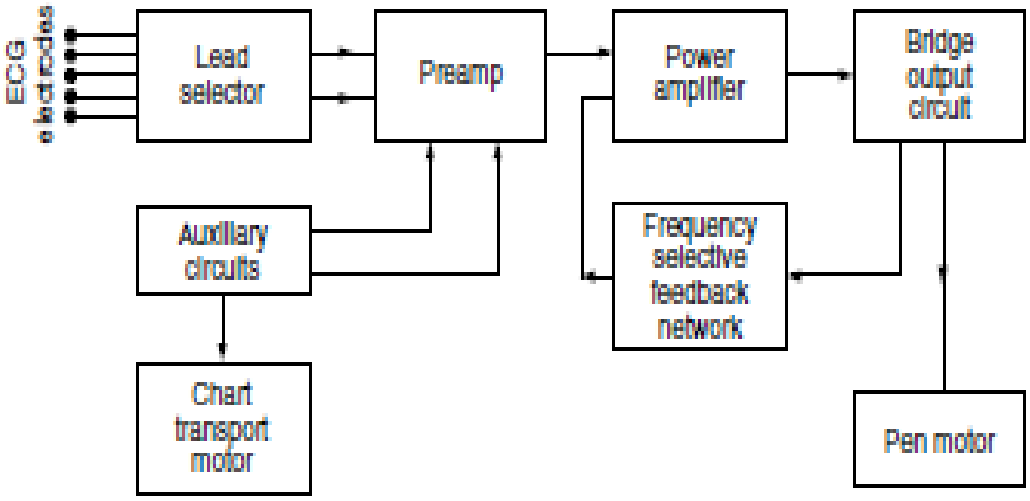


Fig. 5.1 *Block diagram of an ECG machine*

Figure 5.1 shows the block diagram of an electrocardiograph machine. The potentials picked up by the patient electrodes are taken to the lead selector switch. In the lead selector, the electrodes are selected two by two according to the lead program. By means of capacitive coupling, the signal is connected symmetrically to the long-tail pair differential preamplifier. The preamplifier is usually a three or four stage differential amplifier having a sufficiently large negative current feedback, from the end stage to the first stage, which gives a stabilizing effect. The amplified output signal is picked up single-ended and is given to the power amplifier. The power amplifier is generally of the push-pull differentical type. The base of one input transistor of this amplifier is driven by the preamplified unsymmetrical signal. The base of the other transistor is driven by the feedback signal resulting from the pen position and connected via frequency selective network. The output of the power amplifier is single-ended and is fed to the pen motor, which deflects the writing arm on the paper. A direct writing recorder is usually adequate since the ECG signal of interest has limited bandwidth. Frequency selective network is an R-C network, which provides necessary damping of the pen motor and is preset by the manufacturer. The auxiliary circuits provide a 1 mV calibration signal and automatic blocking of the amplifier during a change in the position of the lead switch. It may include a speed control circuit for the chart drive motor.

A ‘stand by’ mode of operation is generally provided on the electrocardiograph. In this mode, the stylus moves in response to input signals, but the paper is stationary. This mode allows the operator to adjust the gain and baseline position controls without wasting paper. Electrocardiograms are almost invariably recorded on graph paper with horizontal and vertical lines at 1 mm intervals with a thicker line at 5 mm intervals. Time measurements and heart rate measurements are made horizontally on the electrocardiogram. For routine work, the paper recording speed is 25 mm/s. Amplitude measurements are made vertically in millivolts. The sensitivity of an electrocardiograph is typically set at 10 mm/mV.

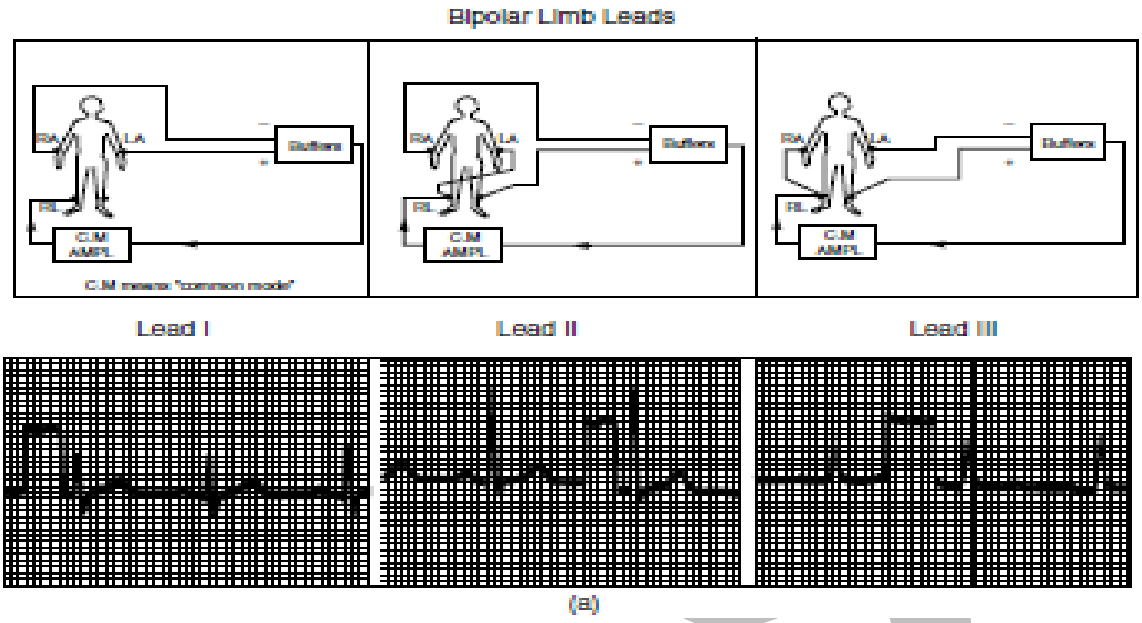
Isolated Preamplifier: It had been traditional for all electrocardiographs to have the right leg (RL) electrode connected to the chassis, and from there to the ground. This provided a ready path for any ground seeking current through the patient and presented an electrical hazard. As the microshock hazard became better understood, particularly when intracardiac catheters are employed, the necessity of isolating the patient from the ground was stressed. The American Heart Association guidelines state that the leakage current should not be greater than 10 microamperes when measured from the patient’s leads to the ground or through the main instrument grounding wire with the ground open or intact. For this, patient leads would have to be isolated from the ground for all line operated units.

Ch-3 ECG Leads

Two electrodes placed over different areas of the heart and connected to the galvanometer will pick up the electrical currents resulting from the potential difference between them. For example, if under one electrode a wave of 1 mV and under the second electrode a wave of 0.2 mV occur at the same time, hen the two electrodes will record the difference between them, i.e. a wave of 0.8 mV. The resulting tracing of voltage difference at any two sites due to electrical activity of the heart is called a "LEAD" (Figs 5.4 (a)-(d)).

- Bipolar Leads:

In bipolar leads, ECG is recorded by using two electrodes such that the final trace corresponds to the difference of electrical potentials existing between them. They are called standard leads and have been universally adopted. They are sometimes also referred to as Einthoven leads (Fig. 5.4(a)). In standard lead I, the electrodes are placed on the right and the left arm (RA and LA). In lead II, the electrodes are placed on the right arm and the left leg and in lead Ill, they are placed on the left arm and the left leg. In all lead connections, the difference of potential measured between two electrodes is always with reference to a third point on the body. This reference point is conventionally taken as the "right leg". The records are, therefore, made by using three electrodes
at a time, the right leg connection being always present. In defining the bipolar leads, Einthoven postulated that at any given instant of the cardiac cycle, the electrical axis of the heart can be represented as a two dimensional vector. The ECG measured from any of the three basic limb leads is a time-variant single-dimensional component of the vector. He proposed that the electric field of the heart could be represented diagrammatically as a triangle, with the heart ideally located at the centre. The triangle, known as the "Einthoven triangle", is shown in Fig. 5.5. The sides of the triangle represent the lines along which the three projections of the ECG vector are measured. It was shown that the instantaneous voltage measuredfrom any one of the three limb lead positions is approximately equal to the algebraic sum of the other two or that the vector sum of the projections on all three lines is equal to zero. In all the bipolar lead positions, QRS of a normal heart is such that the R wave is positive and is greatest in lead II.

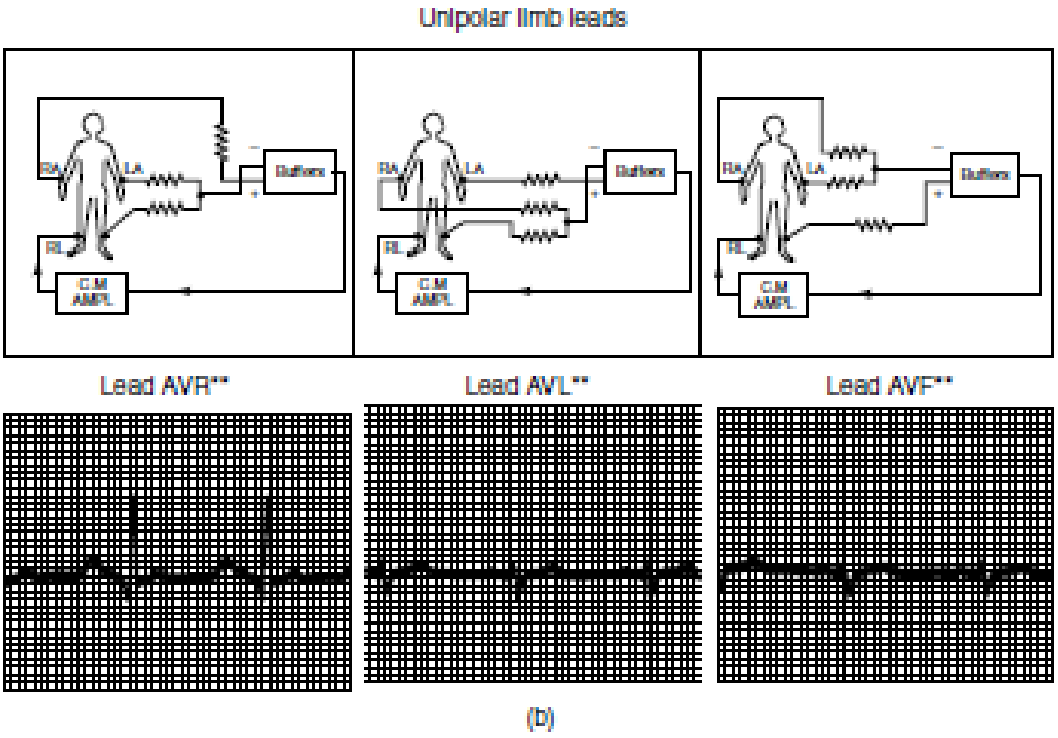


(a)

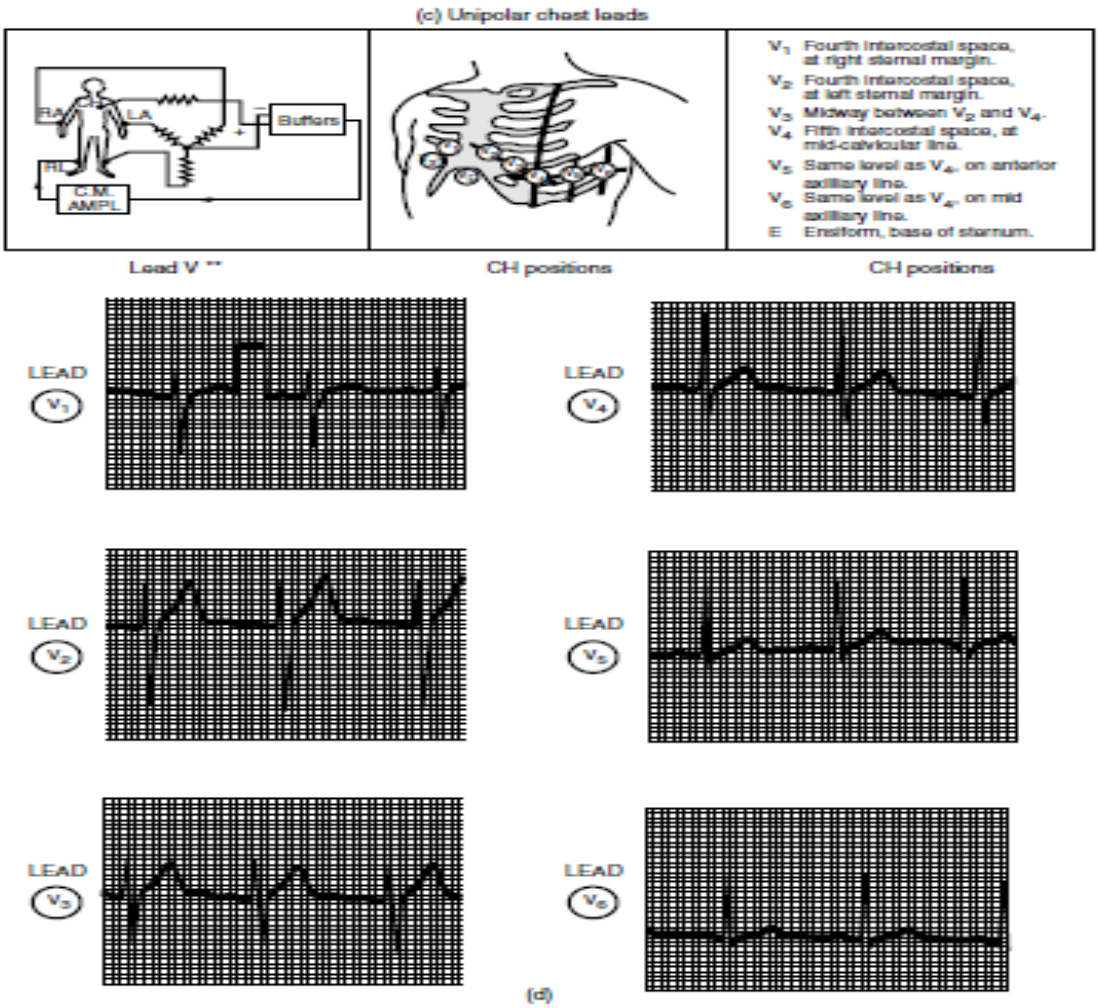
➢ Unipolar Leads (V Leads):

The standard leads record the difference in electrical potential between two points on the body produced by the heart's action. Quite often, this voltage will show smaller changes than either of the potentials and so better sensitivity an be obtained if the potential of a single electrode is recorded. Moreover, if the electrode is placed on the chest close to the heart, higher potentials can be detected than normally available at the limbs. This lead to the development of unipolar leads introduced by Wilson in 1894. In this arrangement, the electrocardiogram is recorded between a single exploratory electrode and the central terminal, which has a potential corresponding to the centre of the body. In practice, the reference electrode or central terminal is obtained by a combination of several electrodes tied together at one point. Two types of unipolar leads are employed which are: (i) limb leads, and (ii) precordial leads. (i) Limb leads In unipolar limb leads (Fig. 5.4(b)), two of the limb leads are tied together and recorded with respect to the third limb. In the lead identified as AVR, the right arm is recorded with respect to a reference established by joining the left arm and left leg electrodes. In the AVL lead, the left arm is recorded with respect to the common junction of the right arm and left leg. In the AVF lead, the left leg is recorded with respect to the two arm electrodes tied together. They are also called augmented leads or 'averaging leads'. The resistances inserted between the electrodes-machine connections are known as 'averaging resistances'.(ii) Precordial leads The second type of unipolar lead is a precordial lead. It employs an exploring electrode to record the potential of the heart action on the chest at six different positions.

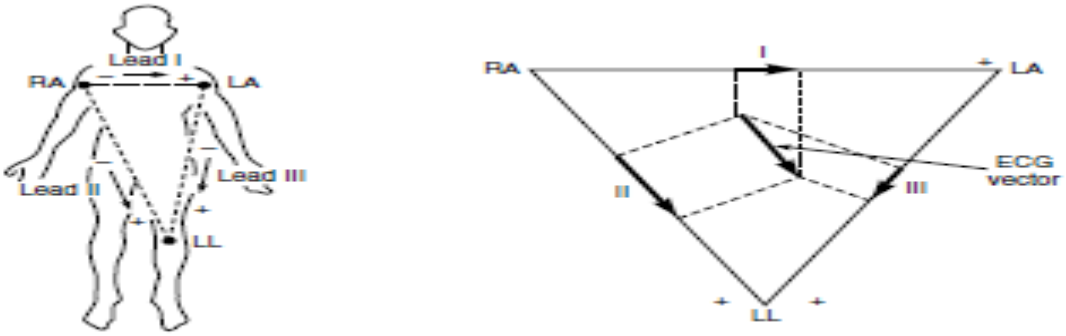
These leads are designated by the capital letter 'V' followed by a subscript numeral, which represents the position of the electrode on the pericardium. The positions of the chest leads are shown in Fig. 5.4(c).



(b)



➤ **Fig. 5.4** *Types of lead connections with typical ECG waveforms (c) position of the chest lead in unipolar precordial lead recording (d) C leads (Courtesy: Hewlett Packard, USA)*



➤ **Fig. 5.5** *The Einthoven triangle for defining ECG leads*

Ch-4 Vectorcardiograph

Vectorcardiography is the technique of analyzing the electrical activity of the heart by obtaining ECG's along three axes at right angles to one another and displaying any two of these ECGs as a vector display on an X-Y oscilloscope. The display is known as a vectorcardiogram (VCG). In contrast, the electrocardiogram which displays the electrical potential in any one single axis, the vectorcardiogram displays the same electrical events simultaneously in two perpendicular axes. This gives a vectorial representation of the distribution of electrical potentials generated by the heart, and produces loop type patterns (Fig. 5.8) on the CRT screen. Usually a photograph is taken of each cardiac cycle. From such pictures, the magnitude and orientation of the P, Q, R, S and T vector loops are determined.

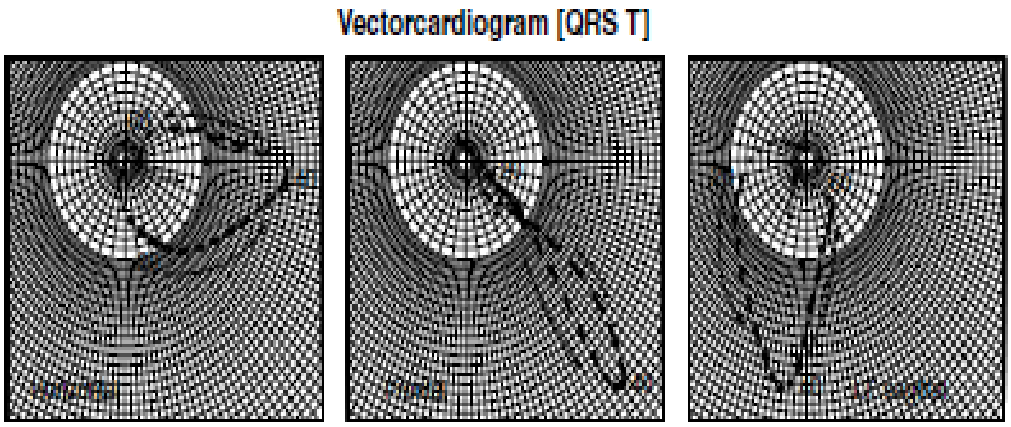


Fig. 5.8 *Typical normal loop patterns recorded in three planes on a direct writing vectorcardiograph*

VCG illustrates the phase differences between the voltages and also the various leads from which it is derived. The major information that it provides is the direction of depolarization and repolarization of the atria and the ventricles. Each vectorcardiogram exhibits three loops, showing the vector orientation of the P wave, the QRSaxis and the T wave. Because of the high amplitude associated with QRS, loops from the QRS complex predominate. An increase in horizontal and

vertical deflection sensitivities is normally required to adequately display the loops resulting from the P wave and T wave. Bourne (1974) describes circuit details of an automated vector ECG recording system.

The VCG has been demonstrated to be superior to the standard 12-lead scalar electrocardiogram in the recognition of undetected atrial and ventricular hypertrophy, sensitivity in identification ofmyocardial infarction and capability for diagnosis of multiple infarctions in the presence of fascicular and bundle branch blocks (Benchimol and Desser, 1975).

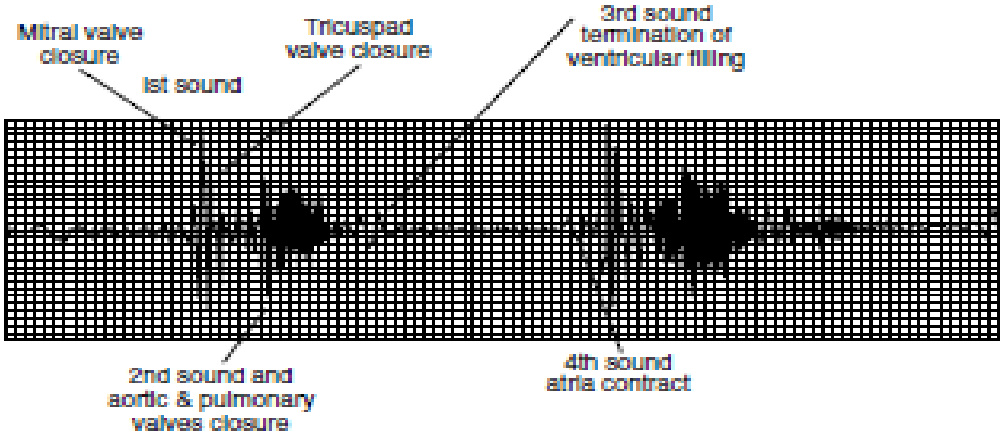
Ch-5 Phonocardiograph

The phonocardiograph is an instrument used for recording the sounds connected with the pumping action of the heart. These sounds provide an indication of the heart rate and its rhythmicity. They also give useful information regarding effectiveness of blood pumping and valve action.

Heart sounds are diagnostically useful. Sounds produced by healthy hearts are remarkably identical and abnormal sounds always corelate to specific physical abnormalities. From the beginning till today, the principal instrument used for the clinical detection of heart sounds is the acoustical stethoscope. An improvement over the acoustal stethoscope, which usually has low fidelity, is the electronic stethoscope consisting of a microphone, an amplifier and a head set. Electronic stethoscopes can detect heart sounds which are too low in intensity or too high in frequency to be heard in a purely acoustal instrument. The phonocardiographs provide a recording of the waveforms of the heart sounds. These waveforms are diagnostically more important and revealing than the sounds themselves.

- Origin of Heart Sound:

The sounds are produced by the mechanical events that occur during the heart cycle. These sounds can be from the movement of the heart wall, closure of walls and turbulence and leakage of blood flow. A typical recording of these sounds is illustrated in Fig. 5.9. The first sound, which corresponds to the R wave of the ECG, is longer in duration, lower in frequency, and greater in intensity than the second sound. The sound is produced principally by closure of the valves between the upper and lower chambers of the heart, i.e. it occurs at the termination of the atria contraction and at the onset of the ventricular contraction. The closure of the mitral and tricuspid valve contributes largely to the first sound. The frequencies of these sounds are generally in the range of 30 to 100 Hz and the duration is between 50 to 100 ms. The second sound is higher in pitch than the first, with frequencies above 100 Hz and the duration between 25 to 50 ms. This sound is produced by the slight back flow of blood into the heart before the valves close and then by the closure of the valves in the arteries leading out of the ventricles. This means that it occurs at the closure of aortic and the pulmonic valves.



➤ **Fig. 5.9** *Basic heart sounds in a typical phonocardiogram recording*

The heart also produces third and fourth sounds but they are much lower in intensity and are normally inaudible. The third sound is produced by the inflow of blood to the ventricles and the fourth sound is produced by the contraction of the atria. These sounds are called diastolic sounds and are generally inaudible in the normal adult but are commonly heard among children.

- Microphones for phonocardiography:

Two types of microphones are commonly in use for recording phonocardiograms. They are the contact microphone and the air coupled microphone. They are further categorized into crystal type or dynamic type based on their principle of operation. The crystal microphone contains a wafer of piezo-electric material, which generates potentials when subjected to mechanical stresses due to heart sounds. They are smaller in size and more sensitive than the dynamic microphone. The dynamic type microphone consists of a moving coil having a fixed magnetic core inside it. The coil moves with the heart sounds and produces a voltage because of its interaction with the magnetic flux. The phonocardiogram depends extensively on the technical design of the microphone, since it does not transform the acoustic oscillations into electrical potential uniformly for all frequencies. Therefore, the heart sound recordings made with a microphone are valid only for that particular type of microphone. As a consequence, microphones of various types cannot, as a rule, be interchanged. A new acoustic sensor, which enhances the audibility of heart sounds and enables recording of quantitative acoustic spectral data is described by Kassal et al, 1994. This device is a polymerbased adherent differential-output sensor, which is only 1.0 mm thick. The device is compliant and can be applied to the skin with gel and two-sided adhesive material, and can conform to the contour of the patient's body. The device can be used for phonocardiography, lung sounds and the detection of Korotkoff sounds.

The device is not a microphone and does not detect acoustic pressure, rather it actually discriminates against it. Instead, the sensor detects the motion of the skin that results from acoustic energy incident upon it from within the soft tissue. Its principle sensing components is PVDF (poly-vinylidene fluoride), which is a piezo-electric polymer. It produces charges of equal magnitude and opposite polarity on opposite surfaces when a mechanical strain is imposed on the material. The voltage generated in the sensor due to the flexing motion forms the basis of electronic stethoscopes, and real time digital acoustic spectral analysis of heart sounds.

Ch-6 Electroencephalograph

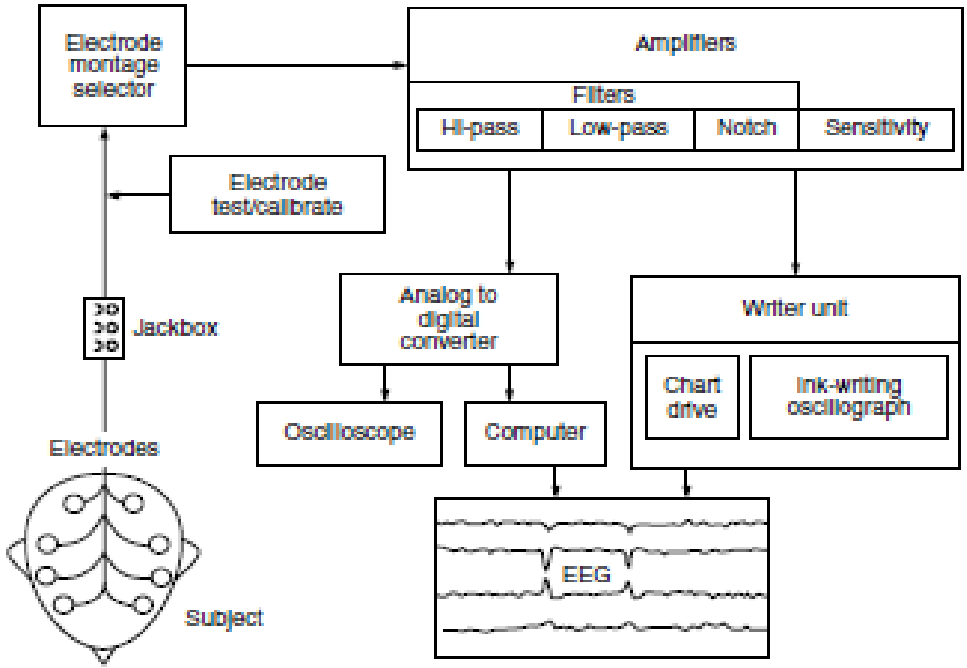
Electroencephalograph is an instrument for recording the electrical activity of the brain, by suitably placing surface electrodes on the scalp. EEG, describing the general function of the brain activity, is the superimposed wave of neuron potentials operating in a non-synchronized manner in the physical sense. Its stochastic nature originates just from this, and the prominent signal groups can be empirically connected to diagnostic conclusions. Monitoring the electroencephalogram has proven to be an effective method of diagnosing many neurological illnesses and diseases, such as epilepsy, tumour, cerebrovascular lesions, ischemia and problems associated with trauma. It is also effectively used in the operating room to facilitate anaesthetics and to establish the integrity of the anaesthetized patient's nervous system. This has become possible with the advent of small, computer-based EEG analyzers. Consequently, routine EEG monitoring in the operating room and intensive care units is becoming popular. Several types of electrodes may be used to record EEG. These include: Peel and Stick electrodes, Silver plated cup electrodes and Needle electrodes.

EEG electrodes are smaller in size than ECG electrodes. They may be applied separately to the scalp or may be mounted in special bands, which can be placed on the patient's head. In either case, electrode jelly or paste is used to improve the electrical contact. If the electrodes are intended to be used under the skin of the scalp, needle electrodes are used. They offer the advantage of reducing movement artefacts. EEG electrodes give high skin contact impedance as compared to ECG electrodes. Good electrode impedance should be generally below 5 kilohms. Impedance between a pair of electrodes must also be balanced or the difference between them should be less than 2 kilohms. EEG preamplifiers are generally designed to have a very high value of input impedance to take care of high electrode impedance. EEG may be recorded by picking up the voltage difference between an active electrode on the scalp with respect to a reference electrode on the ear lobe or any other part of the body. This type of recording is called 'monopolar' recording. However, 'bipolar' recording is more popular wherein the voltage difference between two scalp electrodes is recorded. Such recordings are done with multi-channel electroencephalographs. EEG signals picked up by the surface electrodes are usually small as compared with the ECG signals. They may be several hundred microvolts, but 50 microvolts peak-to-peak

is the most typical. The brain waves, unlike the electrical activity of the heart, do not represent the same pattern over and over again. Therefore, brain recordings are made over a much longer interval of time in order to be able to detect any kind of abnormalities.

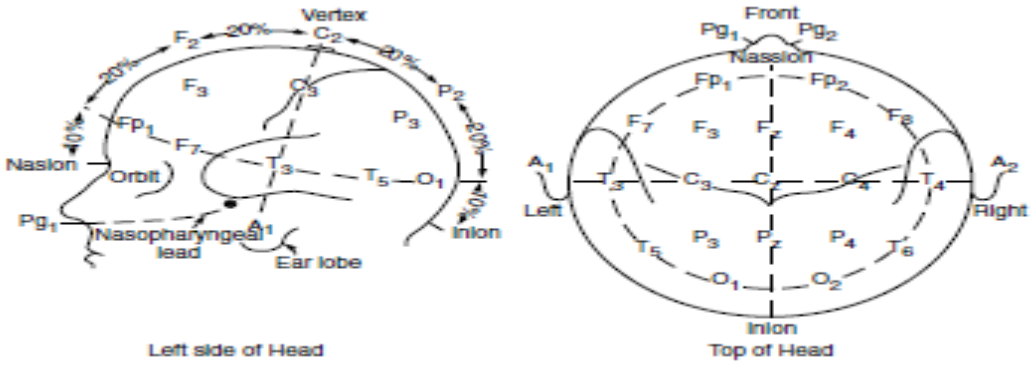
- Block Diagram of Electroencephalogram:

Montages: A pattern of electrodes on the head and the channels they are connected to is called a montage. Montages are always symmetrical. The reference electrode is generally placed on a nonactive site such as the forehead or earlobe. EEG electrodes are arranged on the scalp according to a standard known as the 10/20 system, adopted by the American EEG Society (Barlow et al, 1974). Traditionally, there are 21 electrode locations in the 10/20 system. This system involves placement of electrodes at distances of 10% and 20% of measured coronal, sagittal and circumferential arcs between landmarks on the cranium (Fig. 5.12). Electrodes are identified according to their position on the head: Fpfor frontal-polar, F for frontal, C for central, P for parietal, T for temporal and Ofor occipital. Odd numbers refer to electrodes on the left side of the head and even numbers represent those on the right while Z denotes midline electrodes. One electrode is labelled isoground and placed at a relatively neutral site on the head, usually the midline forehead. A new montage convention has recently been introduced in which electrodes are spaced at 5% distances along the cranium. These electrodes are called closely spaced electrodes and have their own naming convention (Fig. 5.13).

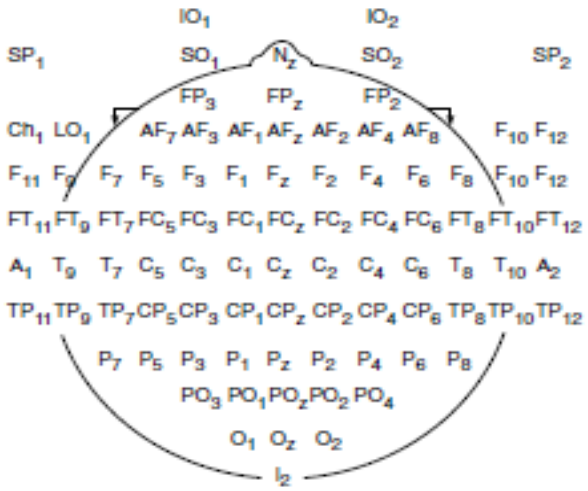


➤ Fig. 5.11 *Schematic diagram of an EEG machine (after Isley et al, 1998)*

Electrode Montage Selector: EEG signals are transmitted from the electrodes to the head box, which is labelled according to the 10-20 system, and then to the montage selector. The montage selector on analog EEG machine is a large panel containing switches that allow the user to select which electrode pair will have signals subtracted from each other to create an array of channels of output called a montage. Each channel is created in the form of the input from one electrode minus the input from a second electrode. Montages are either bipolar (made by the subtraction of signals from adjacent electrode pairs) or referential (made by subtracting the potential of a common reference electrode from each electrode on the head). In order to minimize noise, a separate reference is often chosen for each side of the head e.g. the ipsilateral ear. Bipolar and referential montages contain the same basic information that is transformable into either format by simple substration as long as all the electrodes, including reference, are included in both montages and linked to one common reference. Many modern digital EEG machines record information referentially, allowing easy conversion to several different bipolar montages. The advantage of recording EEG in several montages is that each montage displays different spatial characteristics of the same data.



➤ **Fig. 5.12** *10-20 system of placement of electrodes*



➤ **Fig. 5.13** *Pictorial representation of closely spaced electrodes*

Preamplifier: Every channel has an individual, multistage, ac coupled, very sensitive amplifier with differential input and adjustable gain in a wide range. Its frequency response can be selected by single-stage passive filters. A calibrating signal is used for controlling and documenting the sensitivity of the amplifier channels. This supplies a voltage step of adequate amplitude to the input of the channels. A typical value of the calibration signal is 50 uV/cm. The preamplifier used in electroencephalographs must have high gain and low noise characteristics because the EEG potentials are small in amplitude. In addition, the amplifier must have very high common-mode rejection to minimise stray interference signals from power lines and other electrical equipments. The amplifier must be free from drift so as to prevent the slow movement of the recording pen from its centre position as a result of changes in temperature, etc. EEG amplifiers must have high gain in the presence of unbalanced source resistances and dc skin potentials at least up to 100 mV. Noise performance is crucial in EEG work because skin electrodes couple brain waves of only a few microvolts to the amplifier. Each individual EEG signal should be preferably amplified at the bedside. Therefore, a specially designed connector box, which can be mounted near the patient, is generally employed with EEG machines. This ensures the avoidance of cable or switching artefacts. The use of electrode amplifiers at the site also eliminates undesirable cross-talk effects of the individual electrode potentials. The connector box also carries a circuit arrangement for measuring the skin contact impedance of electrodes with ac. Thus, poor electrode-to-skin contacts above a predetermined level can be easily spotted out.

Sensitivity Control: The overall sensitivity of an EEG machine is the gain of the amplifier multiplied by the sensitivity of the writer. Thus, if the writer sensitivity is 1 cm/V, the amplifier must have an overall gain of 20,000 for a 50 mV signal. The various stages are capacitor coupled. An EEG machine has two types of gain

controls. One is continuously variable and it is used to equalize the sensitivities of all channels. The other control operates in steps and is meant to increase or reduce the sensitivity of a channel by known amounts. This control is usually calibrated in decibels. The gain of amplifiers is normally set so that signals of about 200 mV deflect the pens over their full linear range. Artefacts, several times greater than this, can cause excessive deflections of the pen by charging the coupling capacitors to large voltages. This will make the system unusable over a period depending upon the value of the coupling capacitors. To overcome this problem, most modern EEG machines have de-blocking circuits similar to those used in ECG machines.

Filters: Just like in an ECG when recorded by surface electrodes, an EEG may also contain muscle artefacts due to contraction of the scalp and neck muscles, which overlie the brain and skull. The artefacts are large and sharp, in contrast to the ECG, causing great difficulty in both clinical and automated EEG interpretation. The most effective way to eliminate muscle artefact is to advise the subject to relax, but it is not always successful. These artefacts are generally removed using lowpass filters. This filter on an EEG machine has several selectable positions, which are usually labelled in terms of a time constant. A typical set of time constant values for the low-frequency control are 0.03, 0.1, 0.3 and 1.0 s. These time constants correspond to 3 dB points at frequencies of 5.3, 1.6, 0.53 and 0.16 Hz. The upper cut-off frequency can be controlled by the high frequency filter. Several values can be selected, typical of them being 15, 30, 70 and 300 Hz. Some EEG machines have a notch filter sharply tuned at 50 Hz so as to eliminate mains frequency interference. These however have the undesirable property of 'ringing' i.e. they produce a damped oscillatory response to a square wave calibration waveform or a muscle potential. The use of notch filters should preferably be restricted to exceptional circumstances when all other methods of eliminating interference have been found to be ineffective. The high frequency response of an EEG machine will be the resultant of the response of the amplifier and the writing part. However, the figure mentioned on the high frequency filter control of most EEG machines generally refers to the amplifier. The typical frequency range of standard EEG machines is from 0.1 Hz to 70 Hz, though newer machines allow the detection and filtering of frequencies up to several hundred Hertz. This may be of importance in some intracranial recordings.

Noise: EEG amplifiers are selected for minimum noise level, which is expressed in terms of an equivalent input voltage. Two microvolts is often stated as the acceptable figure for EEG recording. Noise contains components at all frequencies and because of this, the recorded noise increases with the bandwidth of the system. It is therefore important to restrict the bandwidth to that required for faithful reproduction of the signal. Noise level should be specified as peak-to-peak value as it is seen on the record rather than rms value, which could be misleading.

Writing Part: The writing part of an EEG machine is usually of the ink type direct writing recorder. The best types of pen motors used in EEG machines have a frequency response of about 90 Hz. Most of the machines have a response lower than this, and some of them have it even as low as 45 Hz. The ink jet recording system, which gives a response up to 1000 Hz, is useful for some special applications.

Paper Drive: This is provided by a synchronous motor. An accurate and stable paper drive mechanism is necessary and it is normal practice to have several paper speeds available for selection. Speeds of 15, 30 and 60 mm/s are essential. Some machines also provide speed values outside this range. A time scale is usually registered on the record by one or two time marker pens, which make a mark once per second. Timing pulses are preferably generated independently of the paper drive mechanism in order to avoid difference in timing marks due to changes in paper speed.

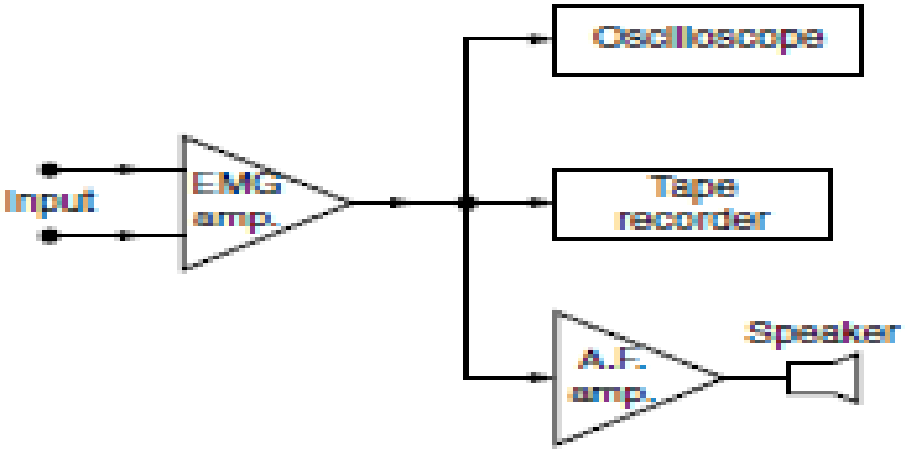
Channels:An electroencephalogram is recorded simultaneously from an array of many electrodes. The record can be made from bipolar or monopolar leads. The electrodes are connected to separate amplifiers and writing systems. Commercial EEG machines have up to 32 channels, although 8 or 16 channels are more common.

Microprocessors are now employed in most of the commercially available EEG machines. These machines permit customer programmable montage selection; for example, up to eight electrode combinations can be selected with a keyboard switch. In fact, any desired combination of electrodes can be selected with push buttons and can be memorized. These machines also include a video monitor screen to display the selected pattern (montage) as well as the position of scalp sites with electrode-to-skin contact. Individual channel control settings for gain and filter positions can be displayed on the video monitor for immediate review. Therefore, a setting can be changed by a simple push button operation while looking at the display. Modern EEG machines are mostly PC based, with a pentium processor, 16-MB RAM, atleast a 2 GB hard disk, cache memory and a 4 GB DAT tape drive. The system can store up to 40 hours of EEG. The EEG is displayed on a 43 cm colour monitor with a resolution of 1280 \ 1024 pixels. The user interface is through an ASCII keyboard and the output is available in the hard copy form through a laser printer.

Ch-7 Electromyography

Electromyograph is an instrument used for recording the electrical activity of the muscles to determine whether the muscle is contracting or not; or for displaying on the CRO and loudspeaker the action potentials spontaneously present in a muscle or those induced by voluntary contractions as a means of detecting the nature and location of motor unit lesions; or for recording the electrical activity evoked in a muscle by the stimulation of its nerve. The instrument is useful for making a study of several aspects of neuromuscular function, neuromuscular condition, extent of nerve lesion, reflex responses, etc.

EMG measurements are also important for the myoelectric control of prosthetic devices (artificial limbs). This use involves picking up EMG signals from the muscles at the terminated nerve endings of the remaining limb and using the signals to activate a mechanical arm. This is the most demanding requirement from an EMG since on it depends the working of the prosthetic device.



➢ Fig. 5.15 *Block diagram of a typical set-up for EMG recording*

EMG is usually recorded by using surface electrodes or more often by using needle electrodes, which are inserted directly into the muscle. The surface electrodes may be disposable, adhesive types or the ones which can be used repeatedly. A ground electrode is necessary for providing a common reference for measurement. These electrodes pick up the potentials produced by the contracting muscle fibres. The signal can then be amplified and displayed on the screen of a cathode ray tube. It is also applied to an audioamplifier connected to a loudspeaker. A trained EMG interpreter can diagnose various muscular disorders by listening to the sounds produced when the muscle potentials are fed to the loudspeaker. The block diagram (Fig. 5.15) shows a typical set-up for EMG recordings. The oscilloscope displays EMG waveforms. The tape recorder is included in the system to facilitate playback and study of the EMG sound waveforms at a later convenient time. The waveform can also be photographed from the CRT screen by using a synchronized camera. The amplitude of the EMG signals depends upon various factors, e.g. the type and placement of electrodes used and the degree of muscular exertions. The needle electrode in contact with a single muscle fibre will pick up spike type voltages whereas a surface electrode picks up many overlapping spikes and therefore produces an average voltage effect. A typical EMG signal ranges from 0.1 to 0.5 mV. They may contain frequency components extending up to 10 kHz. Such high frequency signals cannot be recorded on the conventional pen recorders and therefore, they are usually displayed on the CRT screen.

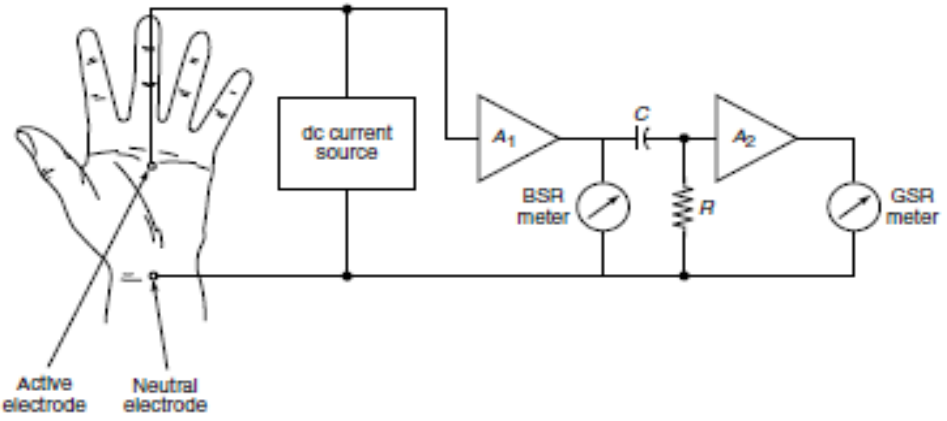
Ch-8 Biofeedback Instrumentation

Feedback is a common engineering term and refers to its function to control a process. When this concept is applied to biological processes within the body, it is known as biofeedback. Here again, biofeedback is a means for gaining control of the body processes to create a specially required psychological state so as to increase relaxation, relieve pain and develop healthier and more comfortable life patterns. The technique involves the measurement of a variable produced by the body process and compares it with a reference value. Based on the difference between the measured and reference value, action is taken to bring the variable to the reference value. It may be noted that biofeedback is not a treatment. Rather, biofeedback training is an educational process for learning specialized mind/body skills. Through practice, one learns to recognize physiological responses and to control them rather than having them control us. The objective of biofeedback training is to gain self-regulatory skills which help to adjust the activity in various systems to optimal levels.

Many different physiological processes have been evaluated for possible control by biofeedback methods. However, the following four neural functions are commonly employed:

· Emotions or Electrodermal Activity (Galvanic skin response measurements)
· Muscle tension or EMG (Electromyograph measurements)
· Temperature/sympathetic pattern (Thermistor readings)
· Pulse (Heart rate monitoring)

Electrodermal activity is measured in two ways: BSR (basal skin response) and GSR (galvanic skin response) is a measure of the average activity of the sweat glands and is a measure of the phasic activity (the high and low points) of these glands. BSR gives the baseline value of the skin resistance where as GSR is due to the activity of the sweat glands. The GSR is measured most conveniently at the palms of the hand, where the body has the highest concentration of sweat glands. The measurement is made using a dc current source. Silver-silver electrodes are used to measure and record the BSR and GSR. Figure 5.18 shows the arrangement for measuring these parameters. The BSR output is connected to an RC network with a time constant of 3 to 5 seconds which enables the measurement of GSR as a change of the skin resistance.



➤ **Fig. 5.18** *Block diagram for measurement and record of Basal Skin Resistance (BSR) and Galvanic Skin Response (GSR)*

Biofeedback instrumentation for the measurement of EMG, temperature and pulse/heart rate is not different from other instruments used for the measurement of physiological variables. Transducers and amplifiers are employed to measure the variable that is to be controlled by the feedback process. The magnitude of the measured variable or changes in the magnitude are converted into a suitable visual or auditory stimulus that is presented to the subject. Based on the stimulus, the subject learns to control the abnormal conditions. Reports have appeared in literature regarding applications of biofeedback to control migraine headaches, to slow down heart rate, etc. Biofeedback techniques have been greatly refined and computerized biofeedback training and psychological computer-assisted guidance programs in the privacy of one's home are now a reality.

Ch-9 Heart Rate Measurement

Heart rate is derived by the amplification of the ECG signal and by measuring either the average or instantaneous time intervals between two successive R peaks. Techniques used to calculate heart rate include: · Average calculation This is the oldest and most popular technique. An average rate (beats/min) is calculated by counting the number of pulses in a given time. The average method of calculation does not show changes in the time between beats and thus does not represent the true picture of the heart's response to exercise, stress and environment. Beat-to-beat calculation This is done by measuring the time (T), in seconds, between two consecutive pulses, and converting this time into beats/min., using the formula beats/ min. = 60/T. This technique accurately represents the true picture of the heart rate. Combination of beat-to-beat calculation with averaging This is based on a four or six beats average. The advantage of this technique over the averaging techniques is its similarity with the beat-to-beat monitoring system. The heart rate meters, which are a part of the patient monitoring systems, are usually of the average reading type. They work on the basis of converting each R wave of the ECG into a pulse of fixed amplitude and duration and then determining the average current from these pulses. They incorporate specially designed frequency to a voltage converter circuit to display the average heart rate in terms of beats per minute. Instantaneous heart rate facilitates detection of arrhythmias and permits the timely observation of incipient cardiac emergencies. Calculation of heart rate from a patient's ECG is based upon the reliable detection of the QRScomplex (Thakor, et al 1983). Most of the instruments are, however, quite sensitive to the muscle noise (artefact) generated by patient movement. This noise often causes a false high rate that may exceed the high rate alarm. A method to reduce false alarm is by using a QRS matched filter, as suggested by Hanna (1980). This filter is a fifteen sample finiteimpulse-response-filter whose impulse response shape approximates the shape of a normal QRS complex. The filter, therefore, would have maximum absolute output when similarly shaped waveforms are input. The output from other parts of the ECG waveform, like a T wave, will produce reduced output.

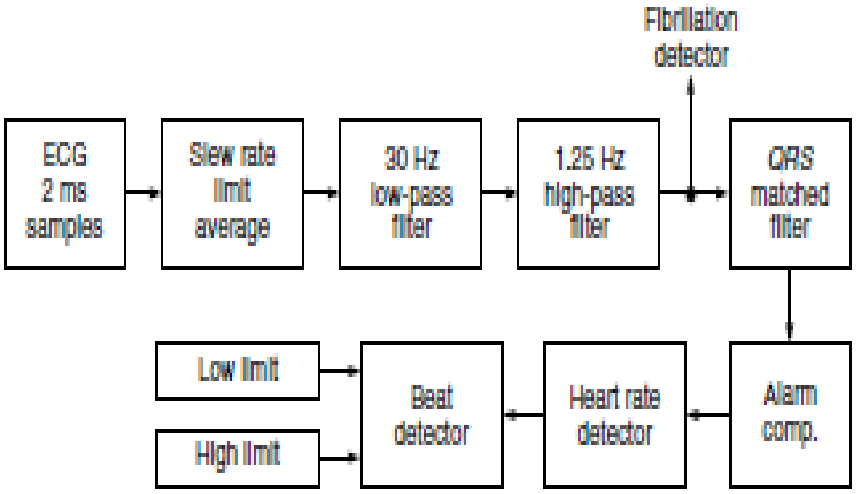


Fig. 6.13 *Block diagram of the cardiotachometer based on matched QRS filter (redrawn after Hanna, 1980; by permission of Hewlett Packard, USA)*

Figure 6.13 is a block diagram of the scheme. The ECG is sampled every 2 ms. Fast transition and high amplitude components are attenuated by a slew rate limiter which reduces the amplitude of pacemaker artefacts and the probability of counting these artefacts as beats. Two adjacent 2 ms samples are averaged and the result is a train of 4 ms samples. In order to remove unnecessary high frequency components of the signal, a 30 Hz, infinite-impulse-response, Butterworth filter is employed. This produces 8 ms samples in the process. Any dc offset with the signal is removed by a 1.25 Hz high-pass filter. The clamped and filtered ECG waveform is finally passed through a QRSmatched filter. The beat detector recognizes QRScomplexes in the processed ECG waveform value that has occurred since the last heart beat. If this value exceeds a threshold value, a heart beat is counted. The beat interval averaged over several beats is used to calculate the heart rate for display, alarm limit comparison, trending and recorder annotation. The threshold in this arrangement gets automatically adjusted depending upon the value of the QRSwave amplitude and the interval between the QRScomplexes. Following each beat, an inhibitory period of 200 ms is introduced during which no heart beat is detected. This reduces the possibility of the T wave from getting counted. The inhibitory period is also kept varied as an inverse function of the high rate limit, with lower high rate limits giving longer inhibitory periods.

Based on the power spectra estimation of the QRScomplex, Thakor et al (1984 b) have suggested that a bandpass filter with a centre frequency of 17 Hz and a Qof five, yields the best signal to noise ratio. Such a simple filter should be useful in the design of heart rate meters, arrhythmia monitors and implantable pacemakers.

The subject of reliable detection of R-wave continues to be of great interest for the researchers. Besides the hardware approach, a number of software based approaches have been reported in literature. Since the ultimate aim of detecting the R-wave is to automate the interpretation of ECG and detect arrhythmias, they are best covered in the succeeding chapter.

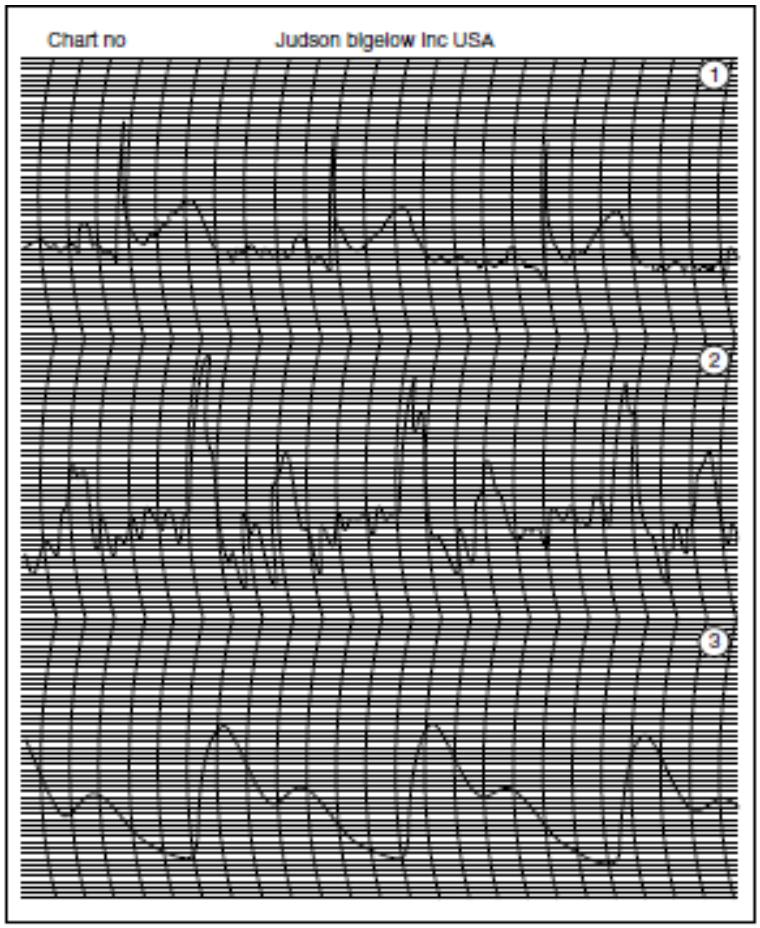
Ch-10 Pulse Rate Measurement

Each time the heart muscle contracts, blood is ejected from the ventricles and a pulse of pressure is transmitted through the circulatory system. This pressure pulse when travelling through the vessels, causes vessel-wall displacement, which is measurable at various points of the peripheral circulatory system. The pulse can be felt by placing the finger tip over the radial artery in the wrist or some other location where an artery seems just below the skin. The timing and wave shape of the pressure pulse are diagnostically important as they provide valuable information.

The pulse pressure and waveform are indicators for blood pressure and flow. Instruments used to detect the arterial pulse and pulse pressure waveforms in the extremities are called plethysmographs. Most plethysmograph techniques respond to a change in the volume of blood as a measure of blood pressure. The pulse gives a measure of pulse wave velocity and can be recorded and compared with the ECG signal (Fig. 6.14). The pulse wave travels at 5 to 15 m/s, depending on the size and rigidity of the arterial walls. The larger and more rigid the artery walls, the greater the velocity. The velocity is 10–15 times faster than blood flow, and is relatively independent of it.

The methods used for the detection of volume (pulse) changes due to blood flow are:

- Electrical impedance changes
- Strain gauge or microphone (mechanical)
- Optical changes (changes in density)

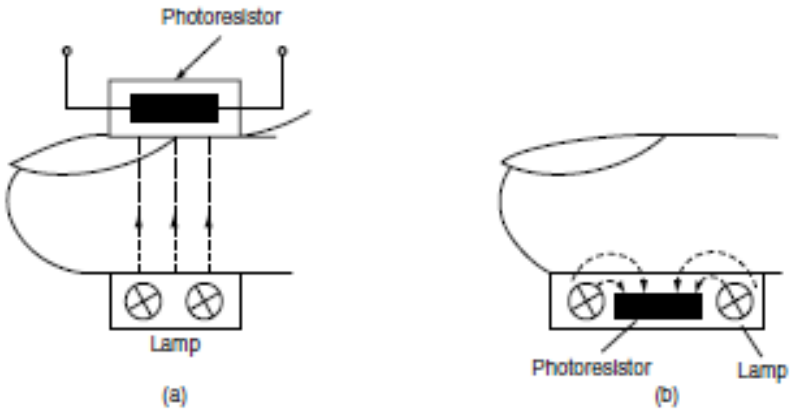


➤ **Fig. 6.14** *Pulse pick up, showing time relationship with electrocardiogram (i) ECG (ii) crystal microphone pulse pick-up (iii) photoelectric pulse pick-up*

An electric impedance method measures the impedance change between two electrodes caused by the change in blood volume between them. The change in impedance (0.1 ohm) may be small as compared to the total impedance (several hundred ohms). The impedance is measured by applying an alternating current between electrodes attached to the body. An alternating signal (10–100 kHz) is used (rather than dc) in order to prevent polarization of the electrodes. The mechanical method involves the use of a strain gauge connected to a rubber-band placed around a limb or finger. Expansion in the band due to change in blood volume causes a change in resistance of the strain gauge. In another technique, a sensitive crystal microphone is placed on the skin's surface to pick up the pulsation. The most commonly used method to measure pulsatile blood volume changes is by the photoelectric method. Two methods are common: Reflectance method and transmittance method. In the transmittance method (Fig. 6.15(a)) a light-emitting diode (LED) and photoresistor are mounted in an enclosure that fits over the tip of the patient's finger. Light is transmitted through the finger tip of

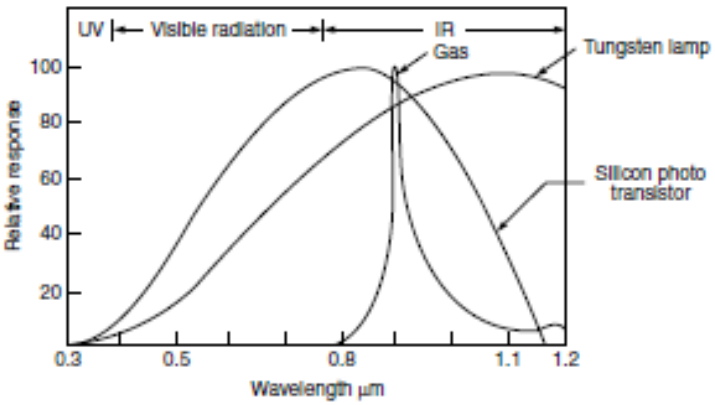
the subject's finger and the resistance of the photoresistor is etermined by the amount of light reaching it. With each contraction of the heart, blood is forced to the extremities and the amount of blood in the finger increases. It alters the optical density with the result that the light transmission through the finger reduces and the resistance of the photoresistor increases accordingly. The photoresistor is connected as part of a voltage divider circuit and produces a voltage that varies with the amount of blood in the finger. This voltage that closely follows the pressure pulse and its waveshape can be displayed on an oscilloscope or recorded on a strip-chart recorder.

The arrangement used in the reflectance method of photoelectric plethysmography is shown in Fig. 6.15(b). The photoresistor, in this case, is placed adjacent to the exciter lamp. Part of the light rays emitted by the LED is reflected and scattered from the skin and the tissues and falls on the photoresistor. The quantity of light reflected is determined by the blood saturation of the capillaries and, therefore, the voltage drop across the photoresistor, connected as a voltage divider, will vary in proportion to the volume changes of the blood vessels.

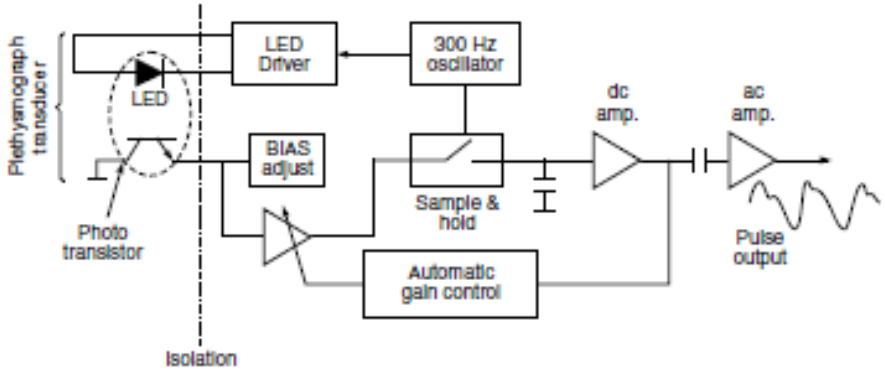


➤ **Fig. 6.15** *Arrangement of photoresistor and lamp in a finger probe for pulse pick-up: (a) transmission method, (b) reflectance method*

The LED phototransistor-photoplethysmograph transducer (Lee et al, 1975) consists of a Ga-As infrared emitting diode and a phototransistor in a compact package measuring 6.25 \ 4.5 \ 4.75 mm. The peak spectral emission of the LED is at 0.94 mm with a 0.707 peak bandwidth of 0.04 mm. The phototransistor is sensitive to radiation between 0.4 and 1.1 mm (Fig. 6.16). For pulse rate measurement, a photoelectric transducer suitable for use on the finger or ear lobe is used. The signal from the photocell is amplified and filtered (0.5 to 5 Hz passband) and the time interval between two successive pulses is measured. The measuring range is 0–250 bpm. Careful placement and application of the device is essential in order to prevent movement artefacts due to mechanical distortion of the skin.



➤ **Fig. 6.16** *Relative spectral response for silicon phototransistor and the radiant spectral distribution of a tungsten lamp and a gallium-aresenide lamp (after Lee et al. 1975; reproducd by permission of IEEE Trans. Biomed. Eng.)*



➤ **Fig. 6.17** *Block diagram for processing plythysmographic signal*

Figure 6.17 shows the block diagram for processing the plethysmographic signal detected from a photoelectric transducer. The circuit consists of two parts, a LED oscillator and driver, which produce 300 Hz, 50 ms infrared light pulses to the finger probe attached to the patient, and a phototransistor that picks up the attenuated light. The electrical signal obtained from the phototransistor is amplified and its peak value is sampled and filtered. An automatic gain control circuit adjusts the amplifier gain to yield a constant average pulse height at the output. The ac component with a frequency in the heart rate range (0.8–5 Hz), is further amplified to output the plethysmographic pulse rate form. This signal is transmitted across the isolation barrier, demodulated, low-pass filtered and transmitted to the analog multiplexer resident on the CPU board.

A Piezo-electric crystal can also be used to detect the pulse wave at certain places of the peripheral system where considerable displacement of the tissue layer above the artery is involved. The arrangement consists of a piezo-electric crystal

clamped in a hermetically sealed capsule subject to displacement stresses. The displacement can be transmitted to the crystal through a soft rubber diaphragm. The crystal can be connected to an ECG recorder for recording the pressure pulse waveform.

There is another variation of the finger plethysmograph in which an air-coupled piezo-electric transducer is employed. As the volume of blood in the finger varies during the cardiac cycle, slight changes occur in the size of the finger. These changes can be transmitted as pressure variations in the air column inside the plastic tubing. A piezo-electric transducer at the end of the tube converts the pressure changes to a corresponding electrical signal. This signal can then be amplified and displayed. Similarly, a semiconductor strain gauge can be used to detect the displacement of the vessel wall due to a pulse wave. Monitoring the peripheral pulse is more useful and dependable than monitoring the heart rate derived from ECG in the case of a heart block because it can immediately indicate the cessation of blood circulation in the limb terminals. Moreover, a photoelectric pick-up transducer is much easier to apply than the three ECG electrodes. The amplitude of the plethysmographic signal obtained is also quite large as compared to the ECG signal and therefore, gives better signal-tonoise ratio. However, the technique is severely subject to motion artefacts.

Ch-11 Blood Pressure Measurement

Blood pressure is the most often measured and the most intensively studied parameter in medical and physiological practice. The determination of only its maximum and minimum levels during each cardiac cycle supplemented by information about other physiological parameters is an invaluable diagnostic aid to assess the vascular condition and certain other aspects of cardiac performance. Pressure measurements are a vital indication in the successful treatment and management of critically ill patients in an intensive cardiac care or of patients undergoing cardiac catheterization. The tremendous research and development for an automatic blood pressure monitor has resulted in several methods but only very few have been commercialized due to certain practical difficulties.

Blood is pumped by the left side of the heart into the aorta, which supplies it to the arterial circuit. Due to the load resistance of the arterioles and precapillaries, it loses most of its pressure and returns to the heart at a low pressure via highly distensible veins. The right side of the heart pumps it to the pulmonary circuit, which operates at a lower pressure. The heart supplies blood to both circuits as simultaneous intermittent flow pulses of variable rate and volume. The maximum pressure reached during cardiac ejection is called systolic pressure and the minimum pressure occurring at the end of a ventricular relaxation is termed as diastolic pressure. The mean arterial pressure over one cardiac cycle is approximated by adding one-third of the pulse pressure (difference between systolic and diastolic values) to the diastolic pressure. All blood pressure measurements are made with reference to the atmospheric pressure.

The most frequently monitored pressures, which have clinical usefulness in medium and longterm patient monitoring, are the arterial pressure and the venous pressure. There are two basic methods for measuring blood pressure–direct and indirect.

- **Direct Method:-**

The direct method of pressure measurement is used when the highest degree of absolute accuracy, dynamic response and continuous monitoring is required. The method is also used to measure the pressure in deep regions inaccessible by indirect means. For direct measurement, a catheter or a needle type probe is inserted through a vein or artery to the area of interest. Two types of probes can be used. One type is the catheter tip probe in which the sensor is mounted on the tip of the probe and the pressures exerted on it are converted to the proportional electrical signals. The other is the fluid-filled catheter type, which transmits the pressure exerted on its fluid-filled column to an external transducer. This transducer converts the exerted pressure to electrical signals. The electrical signals can then be amplified and displayed or recorded. Catheter tip probes provide the maximum dynamic response and avoid acceleration artefacts whereas the fluid-filled catheter type systems require careful adjustment of the catheter dimensions to obtain an optimum dynamic response. Measurement of blood pressure by the direct method, though an invasive technique, gives not only the systolic, diastolic and mean pressures, but also a visualization of the pulse contour and such information as stroke volume, duration of systole, ejection time and other variables. Once an arterial catheter is in place, it is also convenient for drawing blood samples to determine the cardiac output (by dye dilution curve method), blood gases and other chemistries. Problems of catheter insertion have largely been eliminated and complications have been minimized. This has been due to the development of a simple percutaneous cannulation technique; a continuous flush system that causes minimal signal distortion and simple, stable electronics which the paramedical staff can easily operate.

A typical set-up of a fluid-filled system for measuring blood pressure shown in Fig. 6.19. Before inserting the catheters into the blood vessel it is important that the fluid-filled system should be thoroughly flushed. In practice a steady flow of sterile saline is passed through the catheter to prevent blood clotting in it. As air bubbles dampen the frequency response of the system, it should be ensured that the system is free from them. Figure 6.20 shows a simplified circuit diagram commonly used for processing the electrical signals received from the pressure transducer for the measurement of arterial pressure. The transducer is excited with a 5 V dc excitation. The electrical signals corresponding to the arterial pressure are amplified in an operational amplifier or a carrier amplifier. The modern preamplifier for processing pressure signals are of the isolated type and therefore comprise of floating and grounded circuits similar to ECG amplifiers. The excitation for the transducer comes from an amplitude controlled bridge oscillator

through an isolating transformer, which provides an interconnection between the floating and grounded circuits. An additional secondary winding in the transformer is used to obtain isolated power supply for the floating circuits. The input stage is a differential circuit, which amplifies pressure change, which is sensed in the patient connected circuit. The gain of the amplifier can be adjusted depending upon the sensitivity of the transducer. After RF filtering, the signal is transformer-coupled to a synchronized demodulator for removing the carrier frequency from the pressure signal. For the measurement of systolic pressure, a conventional peak reading type voltmeter is used. When a positive going pressure pulse appears at A, diode D3 conducts and charges C3 to the peak value of the input signal, which corresponds to the systolic value. Time constant R3C3 is chosen in such a way that it gives a steady output to the indicating meter.

The value of diastolic pressure is derived in an indirect way. A clamping circuit consisting of C1and D1 is used to develop a voltage equal to the peak-to-peak value of the pulse pressure. This voltage appears across R1. Diode D2 would then conduct and charge capacitor C2 to the peak value of the pulse signal. The diastolic pressure is indicated by a second meter M2 which shows thedifference between the peak systolic minus the peak-to-peak pulse pressure signal. The mean arterial pressure can also be read by using a smoothing circuit when required. Central venous pressure (CVP) measurements made with needle cannulation techniques prove extremely useful in the management of acute circulatory failure and in the maintenance of blood volume in difficult fluid balance problems. Simple water manometers are still the most common measuring device in use, although highly sensitive pressure transducers are preferred when accurate measurements are required. However, the transducers cannot be conveniently mounted at the catheter tip and small positional changes cause large errors in venous pressure. Infusing intravenous fluids while measuring pressure through the same catheter is another problem encountered in these measurements. Central venous pressure is usually measured from a catheter located in the superior vena cava. The CVP reflects the pressure of the right atrium and is sometimes referred to as right atrial pressure. The catheter can even be located in the right atrium. Major peripheral veins used as entry sites for CVP monitoring are the brachial, subclavian and jugular veins.Catheters used for CVP monitoring are usually 25 to 30 cm long. Long catheters, is they remain in place over extended periods of time are susceptible to the formation of fibrin sheaths along their outer surfaces. Besides this, air can be aspirated into a catheter that is situated in an area of low pressure (as compared to the atmospheric pressure), resulting in thrombo-embolic complications. A continuous infusion of heparin solution will reduce this tendency. Also, it should be ensured that there is no possibility of air intake. Development of the Swan-Ganz catheter- a balloon tipped, flexible catheter that can be flow-

directed from a peripheral vein into the pulmonary artery, has made routine clinical monitoring of pulmonary artery pressure possible (Swan and Ganz, 1970).

Information about pulmonary artery wedge pressure or end diastolic pressure in the pulmonary artery gives a good indication of the left atrial pressure. This is a very valuable parameter in predicting and treating left ventricular failure in myocardial infarction in patients undergoing cardiac surgery. Clinical experience has demonstrated the difficulty in maintaining a high-quality arterial pulse waveform during direct measurements of blood pressure. Minute leaks in the stopcocks permit a small quantity of blood to enter the catheter where it clots. Even with a highly leak proof system, clots still form at the catheter tip due to the small volume of blood which may enter as a result of gauge volume displacement (0.04 mm3 per 100 mmHg) and any volume displacement of minute entrapped air bubbles. This type of clotting at the catheter tip can be avoided by using the continuous flush system. Pressure transducers presently available incorporate a continuous flush arrangement. The source of fluid for the flushing system (Fig. 6.21) is a plastic bag (600 ml), which is filled with normal saline and kept at a pressure of 300 mmHg. The high pressure fluid then flows through a Millipore filter (0.22m) which is essential to prevent clogging of the fine bore resistance element and which also serves to filter any bacteria found in the solution. Continuous flush is achieved by using a large resistive element to convert the pressure source to a flow source. With a 0.05 mm diameter glass tubing 1 cm long, flow across the element with 300 mmHg pressure is about 3 ml/h. It is found that large flow rates can cause significant error when using the small diameter catheter. Flow rates of 3 ml/h for adults and 0.5 ml/h for children have been found to be adequate. To initially fill the transducer and catheter, a fast flush feature is needed. This is done by using a rubber valve in the system which when operated permits a fast flush, fills the transducer, and purges the air bubbles from the flush system. Venous pressure measurement can be made by using a strain gauge transducer and a similar electronic signal processing circuitry. The transducers should be of higher sensitivity to give more accurate results at lower pressures. Since the blood pressure is always referred to as the atmospheric pressure at the height of the heart, a correction must be applied while making venous pressure measurements to compensate for the difference of level between the heart and the site of measurement. A correction of 7.8 mmHg is applied for every 10 cm. The site of measurement is below the height of the heart.

- **Indirect Method:-**

The classical method of making an indirect measurement of blood pressure is by the use of a cuff over the limb containing the artery. This technique was introduced by Riva-Rocci for the determination of systolic and diastolic pressures. Initially, the pressure in the cuff is raised to a level well above the systolic pressure so that the flow of blood is completely terminated. Pressure in the cuff is then released at a particular rate. When it reaches a level, which is below the systolic pressure, a brief flow occurs. If the cuff pressure is allowed to fall further, just below the diastolic pressure value, the flow becomes normal and uninterrupted. The problem here finally reduces to determining the exact instant at which the artery just opensand when it is fully opened. The method given by Korotkoff and based on the sounds produced by flow changes is the one normally used in the conventional sphygmomanometers. The sounds first appear (Fig. 6.26) when the cuff pressure falls to just below the systolic pressure. They are produced by the brief turbulent flow terminated by a sharp collapse of the vessel and persist as the cuff pressure continues to fall. The sounds disappear or change in character at just below diastolic pressure when the flow is no longer interrupted. These sounds are picked up by using a microphone placed over an artery distal to the cuff. The sphygmomanometric technique is an ausculatory method; it depends upon the operator recognizing the occurrence and disappearance of the Korotkoff sounds with variations in cuff pressure.

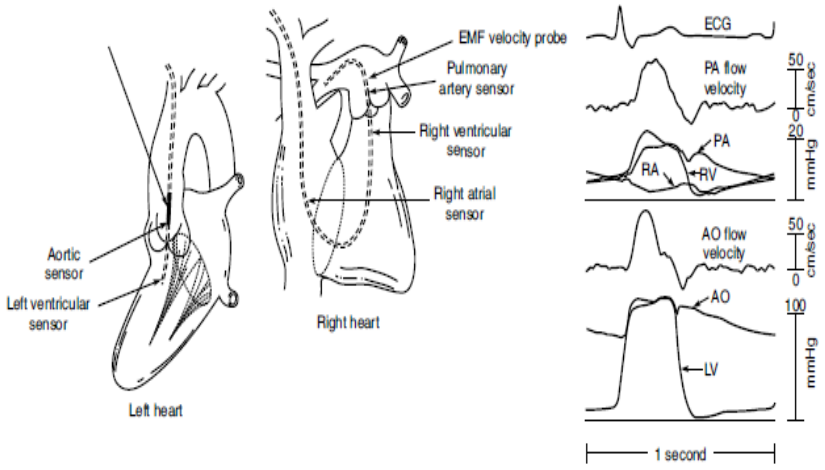


Fig. 6.25 *Simultaneous recording of pulmonary artery and aortic flow velocity signals with high fidelity pressure waveforms using mikro-tip transducers (Courtesy: Miller Instruments Inc. USA)*

A number of automated blood pressure measuring instruments have been designed which make use of the Riva-Rocci method. They operate in a manner analogous to that employed by a human operator, but differ in the method of detecting the pulsations of blood flow at the systolic and diastolic levels. Frequency bands that best discriminate the Korotkoff sounds at systole and diastole from the sou ds immediately preceding these events must be defined for achieving a high degree of reliability in the automatic electronic blood pressure instruments. Golden et al (1974) carried out a special analysis of seven Korotkoff sounds centred about the systolic and diastolic ausculatory events and found that a maximum increase in amplitude at the systolic transition occurred in the 18‑26 Hz band. Similarly, a maximum decrease in spectral energy of diastolic Korotkoff sounds, at ausculatory cessation, was observed within a 40‑60 Hz passband.

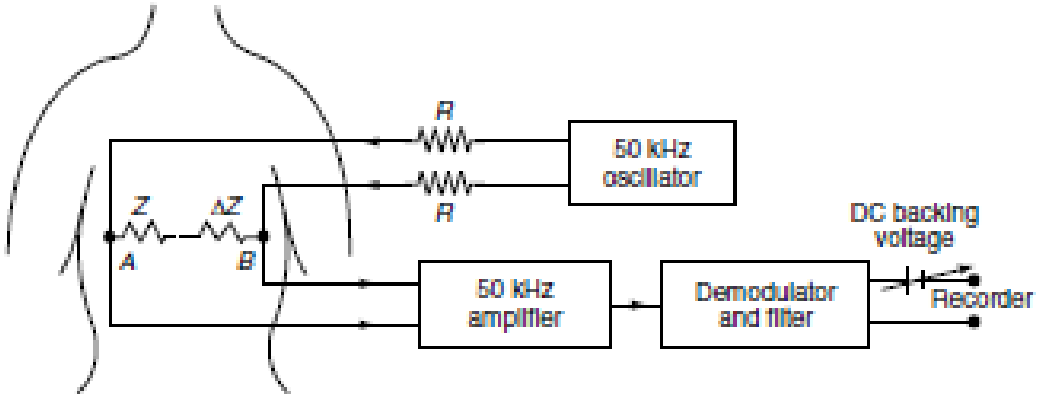
Ch-12 Respiration Rate Measurement

The primary functions of the respiratory system are to supply oxygen and remove carbon dioxide from the tissues. The action of breathing is controlled by a muscular action causing the volume of the lung to increase and decrease to effect a precise and sensitive control of the tension of carbon dioxide in the arterial blood. Under normal circumstances, this is rhythmic action with the result that the respiration rate provides a fairly good idea about the relative respiratory activity. Several techniques have been developed for the measurement of the respiration rate. The choice of a particular method depends mostly upon the ease of application of the transducer and their acceptance by the subject under test. Some of the commonly used methods for the measurement of respiration rate are explained below.

- Impedance pneumography Method:-

This is an indirect technique for the measurement of respiration rate. Using externally applied electrodes on the thorax, the impedance pneumograph measures rate through the relationship between respiratory depth and thoracic impedance change. The technique avoids encumbering the subject with masks, tubes, flowmeters or spirometers, does not impede respiration and has minimal effect on the psychological state of the subject. Impedance method for measuring respiration rate consists in passing a high frequency current through the appropriately placed electrodes on the surface of the body (Fig. 6.36) and detecting the modulated signal. The signal is modulated by changes in the body impedance, accompanying the respiratory cycle. The electrode used for impedance pneumograph are of the self-adhesive type. Contact with the skin is made through the electrode cream layer for minimizing motion artefacts. The electrodes, when the skin is properly prepared, offer an impedance of 150 to 200 W. The change in impedance corresponding to each respiratory cycle is of the order of 1% of the base impedance. The two electrode impedance pneumograph is convenient for use with quiet subjects. Movement artefacts are produced due to changes in the electrode contact impedance, in case the subject ismoving. These artefacts can be significantly reduced by using a four electrode impedance pneumograph. In this case, the output from the oscillator is applied to the two outer electrodes. By doing so, the main oscillator current does not flow through the contact impedance of the measuring electrodes. This system is useful for monitoring restless subjects such as

babies. To avoid the stimulation of sensory receptors, nerves and muscle, currents higher in frequency than 5 kHz must be used for the measurement of physiological events by impedance. Frequencies lower than 5 kHz are particularly hazardous since ventricular fibrillation may be produced with substantial current flow. The use of higher frequencies not only provides the protection sought in the avoidance of tissue stimulation, but also provides the safe use of currents of magnitude, which could be lethal if the frequencies were lower. Electrical impedance changes associated with physiological activity have been studied extensively. Some of the physiological quantities which have been measured and recorded by the impedance method include respiration, blood flow, stroke volume, autonomic nervous system activity, muscle contraction, eye movement, endocrine activity and activity of the brain cells. The impedance-based method of measuring respiration rate is commonly employed in patient monitoring systems. The electrodes used for this purpose are the same as those used for ECG measurement. The dynamic measuring range of the amplifier is 0.1 to 3.0 W with a frequency response of 0.2 to 3.0 Hz corresponding to respiratory rate of 12 to 180 per minute. The amplifier operates within an impedance window established by the static impedance level (approx. 3 k ohms) and its output produces a respiratory waveform from which respiratory rate is derived.



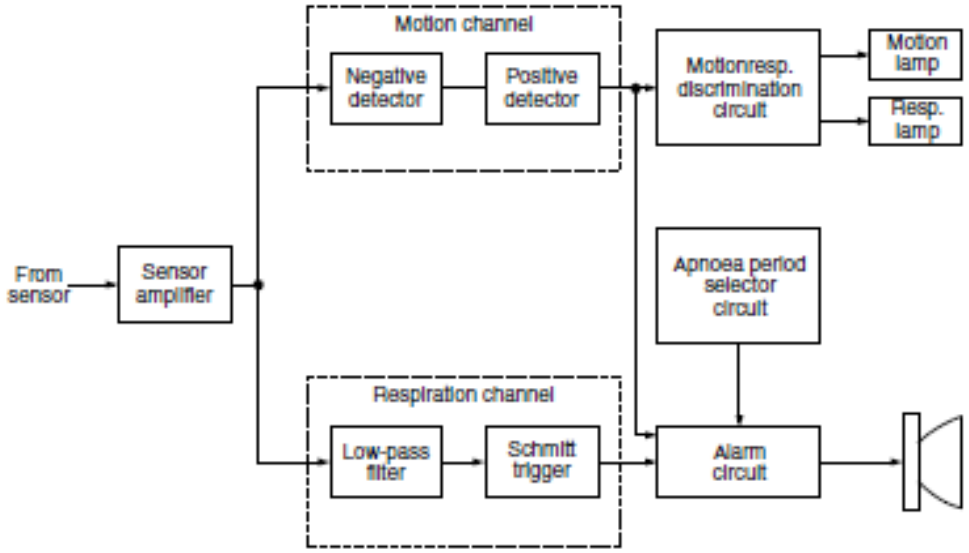
➤ Fig. 6.36 *Principle of impedance pneumograph (two electrode method)*

- Apnoea Monitor:-

Apnoea is the cessation of breathing which may precede the arrest of the heart and circulation in several clinical situations such as head injury, drug overdose, anaesthetic complications and obstructive respiratory diseases. Apnoea may also occur in premature babies during the first weeks of life because of their immature nervous system. If apnoea persists for a prolonged period, brain function can be severely damaged. Therefore, apnoeic patients require close and constant observation of their respiratory activity. Apnoea monitors are particularly useful

for monitoring the respiratory activity of premature infantsSeveral contactless methods are available for monitoring the respiration of infants. The most successful apnoea monitors to-date have been the mattress monitors. These instruments rely for their operation on the fact that the process of breathing redistributes an infant's weight and this is detected by some form of a pressure sensitive pad or mattress on which the infant is nursed. The mattress, in its simplest form, is a multi-compartment air bed, and in this case the weight redistribution forces air to flow from one compartment to another. The air flow, is detected by the cooling effect it produces on a heated thermistor bead. Though the technique is simple, the main disadvantage with the air mattress is the short-term sensitivity variation and the double peaking effect when inspiration or expiration produce separate cooling of the thermistor. Alternatively, a capacitance type pressure sensor in the form of a thin square pad is usually placed under or slightly above the infant's head. Respiratory movements produce regular pressure changes on the pad and these alter the capacitance between the electrode plates incorporated in the pad. This capacitance change is measured by applying a 200 kHz signal across the electrodes and by detecting the current flow with a phase-sensitive amplifier. Two types of electrodes can be used: (i) 70 mm plates, 350 mm apart in a plastic tube which is placed alongside the body; (ii) 250 mm long, 60 mm diameter cylinders placed one on either side of the body. This system is much too sensitive to people moving nearby and thus an electrically screened incubator is essential for the infant. Impedance pneumography is another practical method to monitor the breathing of the patient. The technique also enables the simultaneous monitoring of the heart rate and respiration. The heart rate is known to drop during apnoea. Monitoring the heart rate and respiration thus gives an extra measure of security. Electrodes measure the effort to breath and not the actual ventilation (Kulkarni, 1991). Impedance pneumography has certain inherent disadvantages. One is that the placement of the electrodes is very critical and the other is cardiovascular artefact. This results from the detection of movement between the electrodes because of the cardiovascular system, rather than due to respiration.

Apnoea monitors need to be designed to reject this artefact. Silvola (1989) describes a new non-invasive piezo-electric transducer for the recording of respiration, heart rate and body movements using the PVDF (polyvinylidenefluoride) polymer film. The transducer consists of an area of about 1000 cm2 PVDF film (length 40–50 cm, width 20–30 cm) with a thickness of 40 mm. The PVDF elements are placed directly on the bed mattress without being fixed on the skin. The recordings can be performed when the subject is lying on their back, stomach or on their side.



➤ **Fig. 6.38** *Block diagram of apnoea monitor (Courtesy: B-D Electrodyne, USA)*

Apnoea monitors are generally designed to give audio-visual signals under apnoeic conditions when no respiration occurs within a selectable period of 10, 20 or 30 s. The apnoea monitors are basically motion detectors and are thus subject to other motion artefacts also which could give false readings. The instruments must, therefore, provide means of elimination of these error sources. Figure 6.38 shows a block diagram of an apnoea monitor. It basically consists of an input amplifier circuit, motion and respiration channels, a motion/respiration discrimination circuit, and an alarm circuit. The input circuit consists of a high input impedance amplifier which couples the input signal from the sensor pad to the logic circuits. The sensor may be a strain gauge transducer embedded in the mattress. The output of the amplifier is adjusted to zero volts with offset adjustment provided in the amplifier. The amplified signal goes to motion and respiration channels connected in parallel. The motion channel discriminates between motion and respirationas a function of frequency. In the case of motion signals, high level signals above a fixed threshold are detected from the sensor. In the respiration channel, a low-pass filter is incorporated. Low frequency signals below 1.5 Hz (respiration) cause the output of the Schmidt trigger circuit to pulse at the respiration rate. Higher frequency signals, above 1.5 Hz (motion), cause the output of the trigger to go positive. Absence of the signal (apnoea) causes the output of the Schmidt trigger to go negative. The outputs of the motion and the respiration signals are combined in a comparator circuit, which compares the polarities of the motion and respiration channel signals to indicate respiration. The presence of respiration is indicated by a flashing lamp. The output of the discrimination detector also goes to an apnoea

period selector circuit, a low frequency alarm oscillator and driver, a tone oscillator and audio amplifier connected to a speaker. Audible alarm is given at a frequency of 800–1000 Hz, which is pulsed at 2 Hz. An alternative method of detecting apnoea is based on electromagnetic induction. It consists in passing a high frequency alternating current through a transmitting coil and creating an alternating magnetic field. The transmitting coil is placed at some distance from the infant. The receiving coil is applied on to the abdomen of the infant. The alternating magnetic field induces an emf in the receiving coil. The movement of the abdominal wall with the infant's respiration results in inducing an amplitude-modulated signal in the receiving coil. If this amplitude-modulated signal is demodulated, the modulation frequency corresponding to the respiration frequency can be recovered. Another contactless method for monitoring the breathing activity of premature babies is by the use of microwave energy. It operates by directing low intensity microwave energy (10 GHz) at the individual to be monitored and detecting this energy, after its reflection from the moving surface. The difference between the transmitted and received microwave frequencies (Doppler-shift) provide a signal voltage, which can be amplified, and used as an indicator of the continuance of respiratory activity.

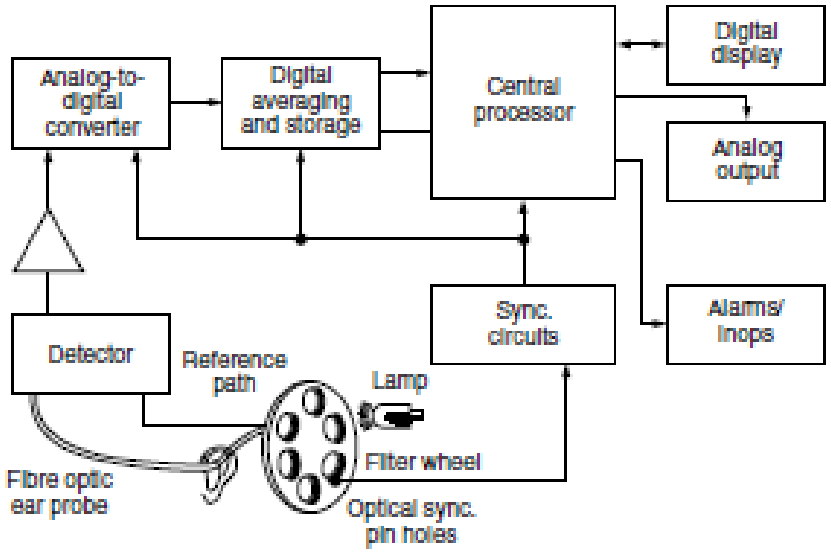
Ch-13 Oxymeters

Oximetry refers to the determination of the percentage of oxygen saturation of the circulating arterial blood. By definition:

$$\text{Oxygen saturation} = \frac{[HbO_2]}{[HbO_2]+[Hb]}$$

In clinical practice, percentage of oxygen saturation in the blood is of great importance. This saturation being a bio-constant, is an indications of the performance of the most important cardio-respiratory functions. It is maintained at a fairly constant value to within a few percent in an healthy organism. The main application areas of oximetry are the diagnosis of cardiac and vascular anomalies; the treatment of post-operative anoxia and the treatment of anoxia resulting from pulmonary affections. Also, a major concern during anaesthesia is the prevention of tissue hypoxia, necessitating immediate and direct information about the level of tissue oxygenation. Oximetry is now considered a standard of care in anaesthesiology and has significantly reduced anaesthesia-related cardiac deaths.

- **Ear Oxymeter:-**



➤ Fig. 10.5 *Block diagram of ear oximeter Model 47201A H.P. (Courtesy: Hewlett Packard, USA)*

Figure 10.5 explains the basic operation of the instrument. The light source is a tungsten-iodine lamp that has a high output in the spectrum of interest. A lens system collimates the light beam and directs it through thin-film interference filters that provide wavelength selection. These filtersare mounted in the periphery of a wheel rotating at 1300 rpm and thus cut the light beam sequentially. The filtered light beam then enters a fibre optic bundle that carries it to the ear. Another fibre optic bundle carries the light passing through the ear back to a detector in the instrument. A second light path is developed with a beam splitter in the path of the collimated light beam near the source. This path also passes through the filter wheel and then through a fibre optic bundle directly to the photodetector. So, the detector receives two light pulses for each wavelength. The processor takes the ratio of two pulses as the measured value; so readings are compensated for any changes in the spectral characteristics of the light source and optical system. The current developed at the photodetector is only 0.5 nA or less during a light pulse. This is amplified in a high gain amplifier and then converted to a 16-bit digital form by an A-D converter synchronized with the wheel rotation. The 16-bit words are given to a digital signal averager that performs two functions. First, it averages out the noise content of the signal with a time constant of 1.6 s and secondly it serves as a buffer to hold information till it is required for computation. Computation of percent oxygen saturation is accomplished by a 24-bit algorithmic-state machine. It uses serial processing with the program stored in ROM and the necessary coefficients of the equations stored on a field programmable ROM. The computation circuits also derive the quantity of total haemoglobin seen within the field of view of the earpiece. If this quantity is low, the instrument displays an 'Off Ear' indication. From the computational section, data is transferred in pulse-decimal form to the output circuit board where it is converted to BCD for the front panel digital display. The patient related part consists of arterializing blood flow in the pinna by a brisk 15 s rub. Application of the probe to the ear results in a suitable display in about 30s. A built-in heater regulated to 41oC maintains arterialization. Restandardization is not required when the instrument is to be used on other patients. Measurements at eight wavelengths provide a great deal of information, which makes it possible to account for eight unknowns. This is sufficient to take into consideration the patient to patient variables and account for the various forms of haemoglobin. The procedure is simple, requiring only the storage of initial light intensities at each of the eight wavelengths. However, it is still necessary to arterialize blood flow by warming the ear, and a large ear probe incorporating fibre optics is necessary to make the system work.

- **Pulse Oxymeter:-**

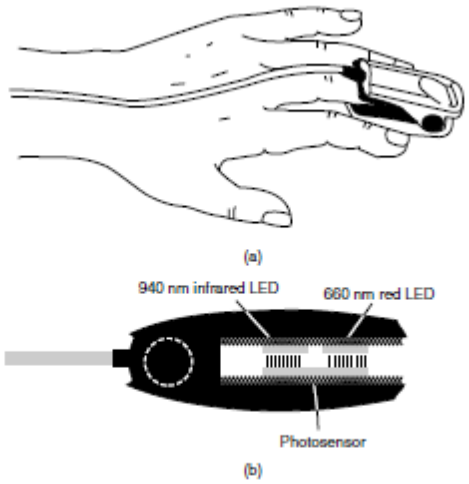
Pulse oximetry is based on the concept that arterial oxygen saturation determinations can be made using two wavelengths, provided the measurements are made on the pulsatile part of the waveform.The two wavelengths assume that only two absorbers are present; namely oxyhaemoglobin (HbO2) and reduced haemoglobin (Hb). These observations, proven by clinical experience, are based on the following:

(i) Light passing through the ear or finger will be absorbed by skin pigments, tissue, cartilage,bone, arterial blood, venous blood.

(ii) The absorbances are additive and obey the Beer-Lambert law: A = -log T = log lo/I = e D C where Io and I are incident and transmitted light intensities, e is the extinction coefficient,D is the depth of the absorbing layer and C is concentration.

(iii) Most of the absorbances are fixed and do not change with time. Even blood in the capillaries and veins under steady state metabolic circumstances is constant in composition and flow, at least over short periods of time.

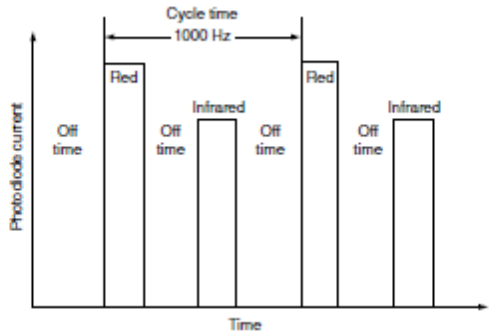
(iv) Only the blood flow in the arteries and arterioles is pulsatile.



➤ Fig. 10.6 (a) *A typical finger tip pulse oximeter probe in use*
(b) *Components of a pulse oximeter probe*

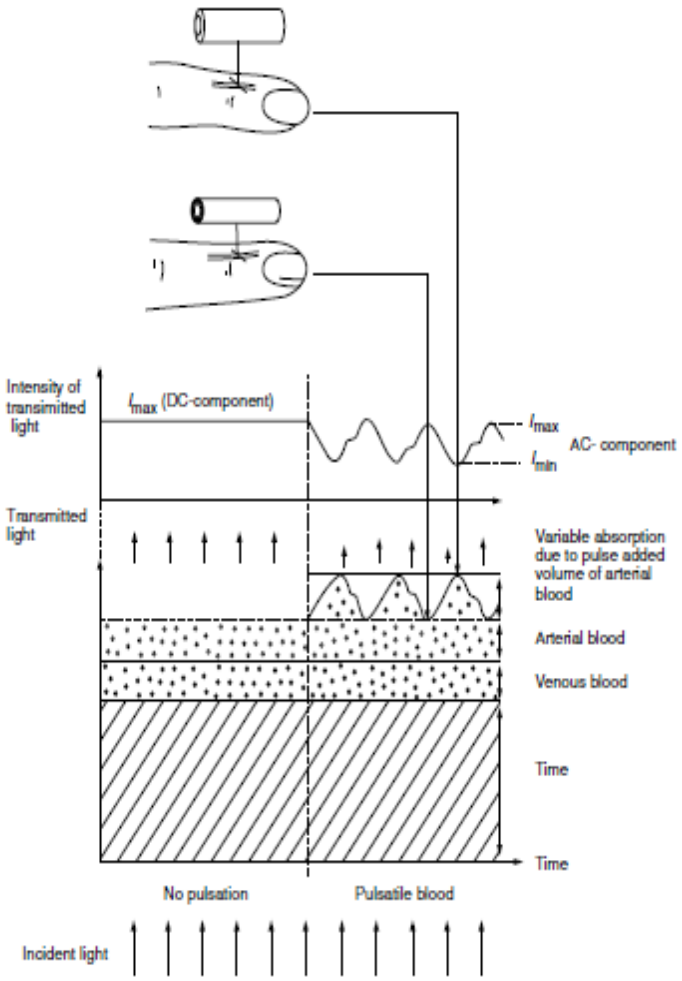
Therefore, only measuring the changing signal, measures only the absorbance due to arterial blood and makes possible the determination of arterial oxygen saturation (SaO2). This is uninfluenced by all the other absorbers which are simply part of the constant background signal. Figure 10.6 (a) shows a typical finger tip oximeter probe in use whereas Fig. 10.6(b) shows the construction of a typical pulse oximeter probe. This has two LEDs (light emitting diodes), one that transmits

infrared light at a wavelength of approximately 940 nm and the other transmitting light at approximately 660 nm. The absorption of these select wavelengths of light through living tissues is significantly different for oxygenated haemoglobin (HbO2) and reduced haemoglobin (Hb). The absorption of these selected wavelengths of light passing through living tissue is measured with a photosensor.



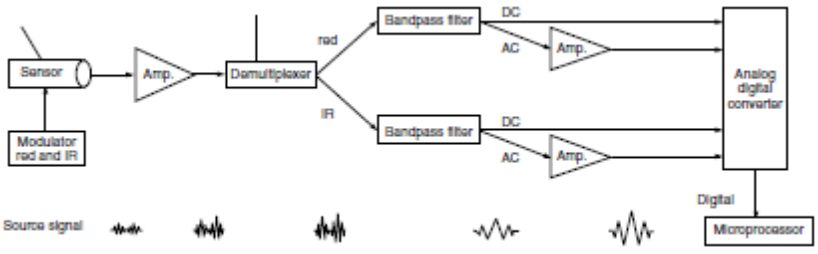
➤ **Fig. 10.7** *Typical pulsing of red and infrared light emitting diodes by a pulse oximeter*

The red and infrared LEDs within the probe are driven in different ways, depending on the manufacturer. Most probes have a single photodetector (PIN-diode), so the light sources are generally sequenced on and off. A typical pulsing scheme of the LEDs is shown in Fig. 10.7. Tocompensate for ambient light during the time when both LEDs are off, the light level is measured and then subtracted from each light channel between cycles. This minimizes the effects due to ambient conditions which may vary during monitoring. Depending on the make and model of pulse oximeters, the drive currents of LEDs, pulse widths, off and on cycles between pulses and cycle times can all vary (Ackerman and Weith, 1995).



➤ Fig. 10.8 *Infrared light absorption in the finger*

The output of the photodiode has a raw signal represented in Fig. 10.8. There will be one signal that represents the absorption of red light (660 nm) and one that represents infrared light (940 nm). The ac signal is due to the pulsing of arterial blood while the dc signal is due to all the non-pulsing absorbers in the tissue. Oxygen saturation is estimated from the ratio (R) of pulse-added red absorbance at 660 nm to the pulse-added infrared absorbances at 940 nm.



➤ Fig. 10.9 *Analog signal processing by pulse oximetry*

Figure 10.9 shows the analog signal processing technique used in pulse oximeters. To simplify the diagram, the circuitry required to drive the LEDs in the sensor are omitted, and only the analog signal processing blocks between the sensor and the digital processing circuitry are shown. The signal from the sensor is a current. The first amplifier stage is a current to voltage converter. The voltage signal then goes through the following circuits: amplifiers to further amplify the signal; noise filters to remove different kinds of interference, a demultiplexer to separate the interleaved red and infrared signals; bandpass filters to separate the low frequency (dc) component from the pulsatile, higher frequency (ac) component; and an analog-digital converter to convert the continuously varying signal to a digital representation. An advancement over the analog signal processing arrangement has been described by Reuss (2000) which eliminates analog circuitry for signal processing and replaces it with a digital signal processing the microprocessor. The output from the sensor is directly given to a high dynamic range analog-to-digital convertor followed by a microprocessor which supports the required digital signal processing. This technique offers the advantages of less circuitry, higher reliability, smaller size and lower costAn accuracy of 1% or better has been reported for the saturation range of above 80% for most transmission type pulse oximeters. Usually, the accuracy is less at lower saturation because of non-linear effects of absorption. The pulse oximeter offers the following advantages:

· It removed the requirement of arterializing blood flow. No heating or rubbing is necessary. The measurement requires that pulsatile activity should be present, but the level is not critical. · Since a change in signal is measured, it is not necessary to store any initial light intensity values, simplifying operational procedures.

· The instrument can be empirically calibrated. Subject variability (skin pigmentation, thickness, tissue, sensor location, etc.) has no significant influence on the measurement.

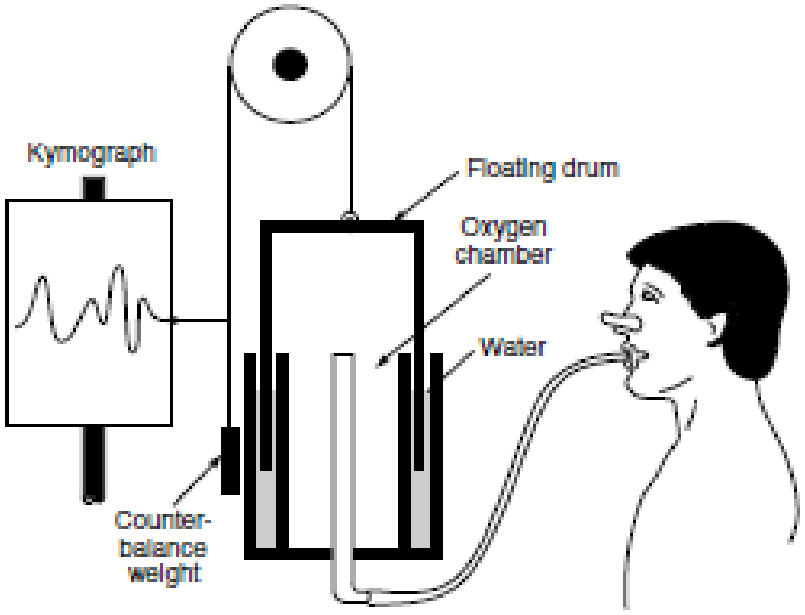
· True arterial saturation is measured because the pulsatile signal comes from the arterial blood.

A limitation of the pulse oximeter is that ambient lights have been shown to interfere with the measurement. Therefore, covering the cuff with an opaque material is necessary to prevent such interference. Motion artifact is also a potential problem. This is because the information containing pulse activity is in the same frequency range as motion artifact.

Ch-14 Spirometers

The instrument used to measure lung capacity and volume is called a spirometer. Basically, the record obtained from this device is called a spirogram. Spirometers are calibrated containers that collect gas and make measurements of lung volume or capacity that can be expired. By adding a time base, flow-dependent quantities can be measured. The addition of gas analzsers makes the spirometer a complete pulmonary function testing laboratory.

- **Basic Spirometer:-**



➤ Fig. 13.3 *Basic water sealed spirometer*

Most of the respiratory measurements can be adequately carried out by the classic water-sealed spirometer (Fig. 13.3). This consists of an upright, water filled cylinder containing an inverted counter weighted bell. Breathing into the bell changes the volume of gases trapped inside, and thechange in volume is translated into vertical motion, which is recorded on the moving drum of a Kymograph. The excursion of the bell will be proportional to the tidal volume. For most purposes,

the bell has a capacity of the order of 6–8 l. Unless a special light weight bell is provided, the normal spirometer is only capable of responding fully to slow respiratory rates and not to rapid breathing, sometimes encountered after anaesthesia. Also, the frequency response of a spirometer must be adequate for the measurement of the forced expiratory volume. The instrument should have no hysteresis, i.e. the same volume should be reached whether the spirometer is being filled or being emptied to that volume. As the water-sealed spirometer includes moving masses in the form of the bell and counterweights, this leads to the usual problems of inertia and possible oscillation of the bell. This can lead to an over-estimation of the expiratory volume. A suggested compensation is by the use of a spirometer bell having a large diameter and which fits closely over the central core of the spirometer, so that the area of water covered by the bell is small in relation to that of the water tank. If the spirometer is used for time-dependent parameters, then it must also have a fast response time, with a flat frequency response up to 12 Hz. This requirement applies not only to the spirometer, but also to the recorder used in conjunction with the recording device. The spirometer is a mechanical integrator, since the input is air flow and the output is volume displacement. An electrical signal proportional to volume displacement can be obtained by using a linear potentiometer connected to the pulley portion of the spirometer. The spirometer is a heavily damped device so that small changes in inspired and expired air volumes are not recorded. The spirometers can be fitted with a linear motion potentiometer, which directly converts spirometer volume changes into an electrical signal. The signal may be used to feed a flow-volume differentiator for the evaluation and recording of data. The response usually is • } 1% to 2 Hz and • } 10% to 10 Hz. Tests made using the spirometer are not analytical. Also, they are not completely objective because the results are dependent on the cooperation of the patient and the coaching efforts of a good respiratory technician.There have been efforts to develop electronic spirometers which could provide greater information- delivering and time-saving capabilities. Also, there have been efforts to obtain more definitive diagnostic information than spirometry alone can provide. Calculating results manually from the graph of the mechanical volume spirometer requires considerable time. Transducers have been designed to transform the movement of the bell, bellows or piston of volume spirometers into electrical signals. These are then used to compute the numerical results electronically. The popularity and low cost of personal computers have made them an attractive method of automating both volume and flow spirometers. An accurate spirometer connected to a personal computer with a good software programme has the potential of allowing untrained personnel to obtain accurate result.

- **Wedge Spirometer:-**

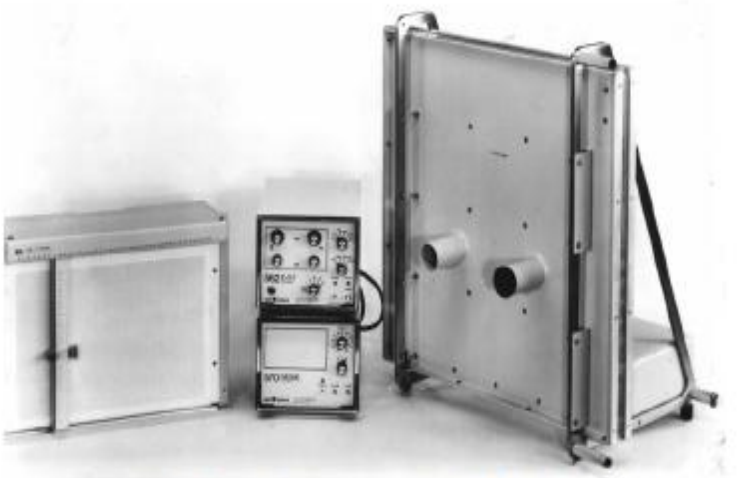


Fig. 13.4 *Wedge spirometer (Courtesy: Med. Science, USA)*

A wedge spirometer (Fig. 13.4) consists of two square pans, parallel to each other and hinged along one edge. The first pan is permanently attached to the wedge casting stand and contains a pair of 5 cm inlet tubes. The other pan swings freely along its hinge with respect to the fixed pan. A space existing between the two pans is sealed airtight with vinyl bellows. The bellows is extremely flexible in the direction of pan motion but it offers high resistance to 'ballooning' or inward and outward expansion from the spirometer. As a result, when a pressure gradient exists between the interior of the wedge and the atmosphere, there will only be a negligible distortion of the bellows.

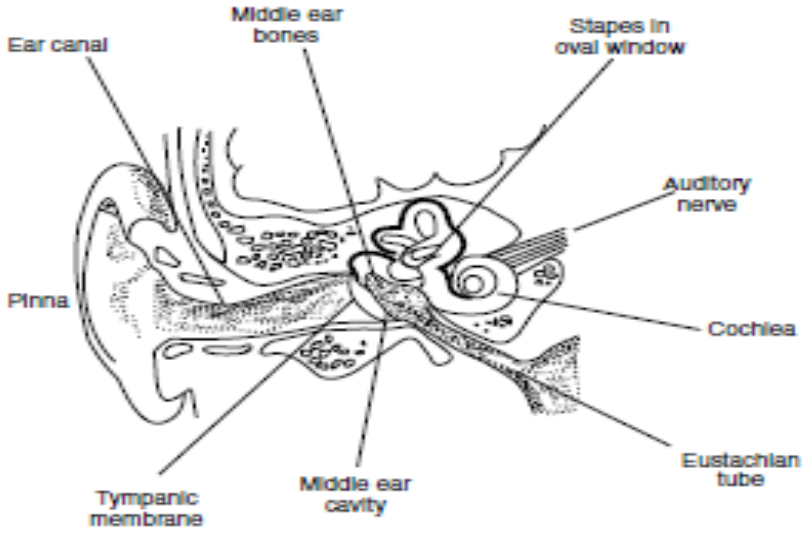
As gas enters or leaves the wedge, the moving pan will change position in compensation for this change in volume. The construction of the wedge is such that the moving pan will respond to very slight changes in volume. Under normal conditions, the pressure gradient that exists between the wedge and the atmosphere amounts to only a fraction of a millimeter of water. Volume and flow signals for the wedge are obtained independently from two linear transducers. The transducers are attached to the fixed frame and are coupled to the edge of the moving pan. One transducer produces a dc signal proportional to displacement (volume), while the other has a dc output proportional to velocity (flow). The transducer outputs are connected to an electronics unit, which contains the power supply, an amplifier, and the built-in calibration networks. A pointer attached to the moving pan and a

scale affixed to the frame, combine to provide a mechanical read out for determining the approximate volume position of the spirometer. When open to the atmosphere and standing upright, the wedge will empty itself due to the force of gravity acting on the moving member. An adjustable tilt mechanism provides the means for changing the resting point of the moving pan to any desired volume point. An adjustable magnetic stop insures a more highly defined resting position. Neither the tilt nor the magnetic stop has any noticeable effect on the moving pan position once it is connected to a closed system. This is primarily due to the large surface area of the pans, which serves to convert small pressures into large forces. Thus, the relatively small forces due to gravity and the magnetic stop are overcome by a negligible rise in pressure in the patient' s lungs. When gravitational return of the moving pan to the resting position is deemed undesirable, the wedge may be turned on its side so that at any point, the pan will be in a state of equilibrium. The wedge may be calibrated with a selector switch, which determines the magnitude of the calibration signal. The volume may be calibrated with a signal corresponding to either 0.5 ml or 5 l. The flow calibration signals for each particular wedge are adjusted, using special fixtures. A volume of one litre is introduced at a certain point and a flow rate of 1 l/s is introduced at another point, with the calibration signals then being adjusted to produce equal signals. As on conventional spirometers, all standard pulmonary function tests may be performed on the wedge. X-Y recorders featuring high acceleration slew rates may be used in recording flow/ volume loops.

Ch-15 Hearing Mechanism

Sound waves are longitudinal waves in which the motion of each particle of the medium in which the wave is travelling, moves backward and forward along a line in the direction in which the wave is propagated. The human aural system reacts to these oscillating pressure changes and transmits them to the brain through a series of steps. Figure 17.1 shows the anatomy of the human ear. The outer ear consists of the pinna or auricle, together with the ear canal, the external auditory meatus, which is a convoluted tube, about 1 cm3 in volume and terminates at its inner end in thetympanic membrane. The pinna scatters acoustic waves so that some of the scattered energy enters the auditory canal and pushes against the tympanic membrane during a wave of compression. The distance membrane moves is a function of the force and velocity with which the air molecules strike it and is, therefore, related to the loudness of sound. The tympanic membrane separates the ear canal from the middle ear cavity. The middle ear is exposed to atmospheric pressure only through the eustachian tube, which connects it to the pharynx and nose or mouth. The sound energy from the tympanic membrane is transmitted through the cavity of the middle ear, to the receptor cells in the inner ear, which are surrounded by fluid. Thus, the major function of the middle ear is to transfer movements of the air in the outer ear to the fluid-filled chambers of the inner ear. A chain of three small, middle ear bones couple the tympanic membrane to a membrane covered opening, called the oval window. The total force on the oval window is the same as that on the tympanic membrane. The size of the window being very small, it experiences much greater force per unit area. One of the bones, called the stapes, rests upon the lower end of the cochlea and passes the vibrations directly into the fluid within. The inner ear or cochlea is a fluid-filled coiled passage in the temporal bone. It is almost completely divided lengthwise by the basilar membrane. Most of the pressure wave received by the cochlea is transmitted to this membrane, which is deflected into the scale tympani. The membrane has different resonating properties along its length, responding to high frequencies at the stapes end and to low frequencies at its upper end. The membrane contains the sensitive receptor cells, which transform sound energy or pressure waves into action potentials. The nerve impulses thus initiated are propagated along the acoustic nerve fibres to the brain with a speed of 100 m/s. The pattern of nerve impulses arriving in the brain is associated with the subjectively experienced sound, which has attributes of loudness, pitch and timbre (quality). The appreciation of sound is mainly a cerebral function. However, the

recognition of notes is partly a function of the cochlea. Therefore, if it is defective, the individual may not hear certain tones. Hearing is affected by anything which interferes with the conduction of sound waves to the cochlea, such as a perforated tympanic membrane (ear drum), disease of the middle ear or disease of the cochlea itself or its connection in the central nervous system.The sounds reaching the ear are characterized by loudness (intensity), which depends upon the amplitude of the waves; by pitch, which depends upon their frequency; and by quality, which results from the combination and interaction of the waves. The human ear responds to vibrations ranging from 20 to 20,000 Hz. The waves of speech and many other common sounds are not of single frequency but are complex waves made up of several frequencies of vibration. The number of sound frequencies in addition to the fundamental tone, i.e. the degree of purity of the sound wave is related to the quality or timbre of the sound. The human ear can in fact, distinguish some 400,000 different sounds.



➤ Fig. 17.1 *Anatomy of ear*

- **<u>Air and Bone Conduction:-</u>**

Air conduction, by definition, is the transmission of sound through the external and middle ear to the internal ear. Bone conduction, on the other hand, refers to transmission of sound to the internal ear mediated by mechanical vibration of the cranial bones and soft tissues. The most importantdiagnostic differential from the standpoint of the functional hearing tests is the relationship between air and bone conduction acuity. Clinical observation has shown that hard-of-hearing patients with middle ear disease usually have normal hearing by bone

conduction, whereas patients with inner ear involvement have decreased or diminished bone- onduction.

It has been concluded from clinical observations that an approximate 60 dB loss is the maximal air conduction impairment to be anticipated with middle ear defect. Therefore, if the air conduction loss in a patient with apparently typical middle ear pathology exceeds 60 dB, it is likely that inner ear impairment is superimposed on the middle ear lesion. Figure 17.2 shows that upto a frequency of about 14–16 kHz, the thresholds for air conduction and bone conduction have the same shape. At this frequency, they both show an abrupt fall in sensitivity with a slope of 50 dB/octave. The start of this slope defines the 'end point' of the ear. For air conducted signals, the fall in sensitivity continues, so that for instance at 25 kHz, 5 W of acoustic power (equivalent to about 500 W of electrical power) is needed to produce a hearing response. On the other hand, for bone conducted signals, there is a change in slope again, at about 2 kHz above the end point. From then on up to 200 kHz, the threshold sensitivity falls at a rate of 10 to 15 decibels per octave. So, in the ultrasonic region, a bone conducted signal of less than one electrical watt is audible. There is a rapid drop in impedance of the middle ear at high frequencies and very little of the acoustical energy fed to the ear by air conduction is transmitted to the cochlea. But bone-conducted sound by-passes the middle ear, as is shown in Fig. 17.3. This to some extent explains the different threshold shapes at high frequencies.

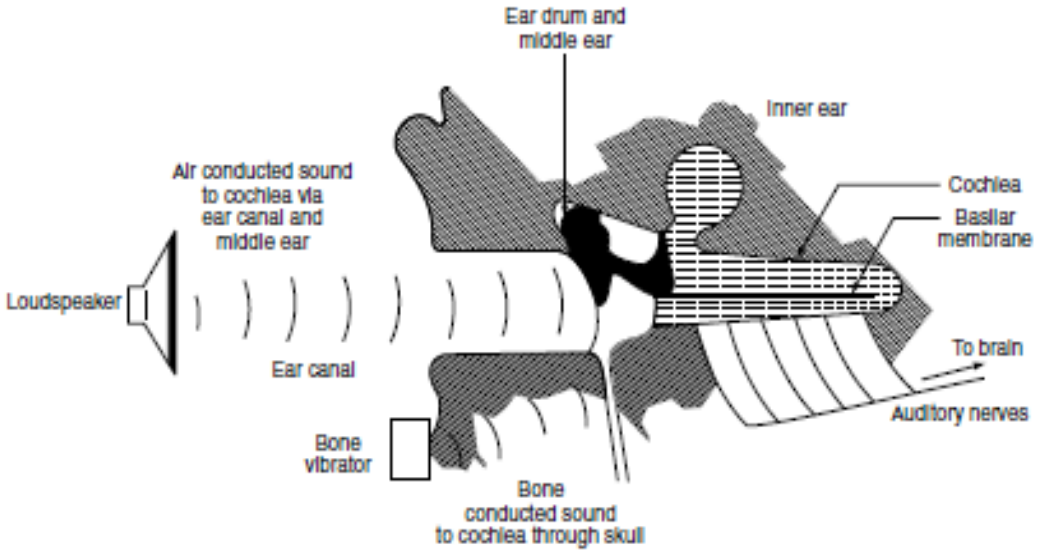


Fig. 17.3 *Mechanism of air and bone conduction*

Ch-16 Audiometers

An audiometer is a specialized equipment, which is used for the identification of hearing loss in individuals, and the quantitative determination of the degree and nature of such a loss. It is essentially an oscillator driving a pair of headphones and is calibrated in terms of frequency and acoustic output. Both frequency and output are adjustable over the audio range. The instrument is also provided with a calibrated noise source and bone-conductor vibrator. Audiometers may be divided into two main groups on the basis of the type of stimulus they provide to elicit auditory response: pure-tone audiometers and speech audiometers. A pure-tone audiometer is used primarily to obtain air-conduction and bone-conduction thresholds of hearing.

These thresholds are helpful in the diagnosis of hearing loss. Pure-tone screening tests are employed extensively in industrial and school hearing conservation programmes. Speech audiometers are normally used to determine speech reception thresholds for diagnostic purposes and to assess and evaluate the performance of hearing aids. Screening audiometers are used to separate two groups of people. One that can hear as well as or better than a particular standard and the other that cannot hear so well. Applications of these instruments are found in industry, chools and military service. An important application of audiometers is in industry. They help to assess the hearing function of personnel at different stages of their detection of changes in auditory acuity, identify noise susceptible persons and evaluate the effectiveness of ear protectors and noise control measures. In conventional pure-tone audiometry, head phones are worn by the subject and a set of responses is obtained for air-conducted sounds directed to each ear in turn. A bone conductor vibrator can then be attached to the head at the centre forehead position to see whether the hearing threshold improves. If it does, then the disorder is most likely wholly or partly conductive in origin. To avoid stimulation of the ear not under test with the vibrator, it can be temporarily made deaf by introducing a suitable masking noise in the non-test ear via an earphone. A narrow-band noise centred on the pure-tone test frequency or a wide-band white noise is used for this purpose.The problem of how to recognize the need for masking and then applying the correct intensity poses a considerable difficulty.

- **Requirement of Audiometers:-**

Modern audiometers are solid-state instruments covering a frequency range from approximately 100 to 10,000 Hz. Some instruments produce this range in discrete octave or semi-octave steps or intervals, while others provide for continuously variable frequency over their designed range. The frequency must remain sensibly constant at a value within 1–3% of the indicated value. Where automatic recording facilities include a continuous sweep frequency, the rate of change is normally kept as one octave per minute. If an automatic recording audiometer provides fixed frequencies, then a minimum period of 30 s must be allowed at each frequency. The test frequencies should have sufficient purity of tone or approximation to the deal sine wave form to ensure response only to the desired fundamental frequency. The maximum harmonic distortion in pure-tone air conduction audiometry is specified as 2% for the second and third harmonic and much less at higher order harmonics. The total harmonic distortion should not be more than 3%. The intensity range of most audiometers starts from approximately 15 dB above normal to 95 dB below normal over a frequency range from approximately 500 to 4000 Hz. The intensity range is somewhat less for frequencies below 500 Hz and above 4000 Hz. This is partly because of certain instrumental limitations imposed by the earphone or vibrator and partly due to the desire to avoid the threshold of feeling from stimulation at the lower frequency levels. The threshold of feeling is the sensation of pain or tickle in the ear, which results from sound pressures and limits the maximal sound intensity that can be tolerated by the ear. The intensity level at which the threshold of feeling is stimulated varies with frequency. For example, the threshold of feeling is stimulated at an intensity level approximately 120 dB above the normal threshold of audibility from about 500to 4000 Hz, but at 64 Hz the threshold of feeling is stimulated by sound pressures approximately 65 dB above the normal threshold value. The attenuation dials on the audiometers provide variable intensity or volume controls. They are calibrated in decibels usually in discrete steps, which differ by 5 dB in intensity from step to step. Auditory acuity for each frequency is thus measured in dB above or below the normal hearingzero dB reference level for that frequency. This level is the minimal intensity at which each given frequency can be perceived by the normal ear in a noise free environment and is experimentally determined by averaging the results of measurement on a large number of normal individuals between 18 and 25 years of age. Audiometers usually have two channels with single pure-tone generators. The first channel has

pure-tone or speech output while the second channel has nominal masking. The pure-tone and speech can be switched to both channels for special tests. Channel two can have either wide or narrow-band masking. Each channel has an accurate independent attenuator output and the transducers are switched to each attenuator as required. In the recent years, numerous audiometric products incorporating microprocessors have been introduced in the market. Such equipment offers greater convenience in calibration, test signal presentation and versatility. Automated data collection and storage are also useful features included in such equipment. It may however be noted that the audiometric measurement principles as described in the following text are not altered with the use of microprocessors and digital technology. It is extremely important in audiometry to ensure that only the testing signal reaches the ear. Therefore, all testing must be done in a noise-free environment. Since environmental noise is difficult to control, the noise-free conditions are achieved by performing the audiometric testing in a sound isolating enclosure. Such enclosures help to attenuate all frequencies within the sensory range below the threshold of hearing of normal ears. By using double wall construction and appropriate sound absorbing material, it is common to achieve 25 dB attenuation at 125 Hz and 60 dB attenuation at frequencies between 1000 and 8000 Hz.

- **Pure tone audiometer:-**

A wave in air, which involves only one frequency of vibration, is known as pure-tone. Pure-tone audiometry is used in routine tests and, therefore, it is the most widely used technique for determining hearing loss. Pure-tone audiometers usually generate test tones in octave steps from 125 to 8000 Hz, the signal intensity ranging from -10 dB to +100 dB. Pure-tone audiometry has several advantages, which makes it specifically suitable for making threshold sensitivity easurements. A pure-tone is the simplest type of auditory stimulus. It can be specified accurately in terms of frequency and intensity. These parameters can be controlled with a high degree of precision. Speech audiometry normally allows measurements to be made within the frequency range of 300-3000 Hz. Some patients may have impaired high frequency response due to high intensity level occupational noise at 4000 or 6000 Hz. Pure-tone measurements at these frequencies prove to be a more sensitive indicator of the effect of such noise on the ear than speech tests. Changes in threshold sensitivity associated with various middle ear surgical procedures can be monitored more accurately with pure-tone than speech tests. A pure-tone audiometer basically consists of an LC oscillator in which the inductance and tuning capacitance are of close tolerances for having a precise control on the frequency of oscillations. The oscillator is coupled to an output current amplifier stage to produce the required power levels. The attenuators used in these

instruments are of the ladder type, of nominal 10 W impedance. The signals are presented acoustically to the ear by an earphone or small loudspeaker.

- **SPEECH AUDIOMETER:-**

Besides tonal audiometry, it is sometimes necessary to carry out tests with spoken voices. These tests are particularly important before prescribing hearing-aids and in determining the deterioration of speech understanding of patients. Specially designed speech audiometers are used for this purpose. They incorporate a good quality tape recorder, which can play recorded speech. A double band tape recorder is preferred to interface the two channel audiometer units. Masking noise is supplied by the noise generator. The two channels supply the two head-phones or the two loud speakers which are of 25 W each. The tape recorder has a capacity for recording a limitless variety of test material and a consistency of speech input, which cannot be obtained for live-voice audiometry in relation to testretest repeatability. Another advantage of the tape recorded material is that the test words and sentences can be selected to cater for the widely differing needs of age, intelligence, dialect and language. In speech audiometers, live-voice facilities are incorporated primarily for communication purposes as the inherent unreliability of live-voice speech tests may lead to serious errors. The microphone amplifier used for this purpose is a simple two stage amplifier. The frequency response characteristics of a live-voice channel should be such that with the microphone in a free sound field having a constant sound pressure level, the sound pressure level developed by the earphone of the audiometer in the artificial ear at frequencies in the range 250 to 4000 Hz does not differ from that at 1000 Hz by more than 110 dB. Also, it shall not rise at any frequency outside this band by more than 15 dB, relative to the level at 1000 Hz.

CH-17 Hearing Aid

Hearing loss has many forms. The most common is related to the body aging process and to longterm cumulative exposure of the ear to sound energy. As one grows older, it becomes more difficult to hear. The ear becomes less sensitive to sound, less precise as a sound analyzer and less effective as a speech processor. Loss of hearing differs greatly in different individuals. Changes in the ear occur gradually over time. However, by the time the changes are manifested, it is stimated that approximately 30 to 50 percent or more of the sensory cells in the inner ear have suffered irreparable structural damage or are missing (Engebretson, 1994). Under these conditions, the only choice available for hearing-impaired individuals is to wear a hearing aid. Hearing impairment is caused by either loss in sensitivity (loss in perceived loudness), or loss in the ability to discriminate different speech sounds or both. Loss of loudness may be due to either increased mechanical impedance between the outer ear and the inner ear or by the reduced sensitivity of the sensory organ of hearing. Loss of the discrimination ability is basically associated with damage to the sensory organ, although, other neural structures at higher levels may also be involved. The modern hearing aid became possible with the invention of the transistor, which has enabled to develop small, power-efficient amplifier circuits that could be packed in a form that fits behind or in the ear. Even though the primary function of an hearing aid is to compensate for the loss of sensitivity of the impaired ear, in practice, it is not this simple. The ear behaves differently for soft sounds near the hearing threshold than it does for loud sounds. Therefore, a frequency response that restores normal hearing thresholds for soft sounds will not, in general be appropriate for louder sounds. Furthermore, even when speech sounds are made audible for the hearing-impaired listener, it does not follow that he/she will be able to understand speech. Hearing-impaired listeners experience more difficulty in understanding speech in background noise than normalhearing listeners.

- **Conventional Hearing Aid:-**

Modern hearing aids have evolved from single-transistor amplifiers to modern multi-channel designs containing hundreds and even thousands of transistors. A typical design is shown in Fig. 17.6. The basic functional parts include a microphone and associated preamplifier, an automatic gain control circuit (AGC), a set of active filters, a mixer and power amplifier, an output transducer or receiver. The total circuitry works on a battery. The use of multiple channels in this design provides different compression characteristics for different frequency ranges. Typically, the crossover frequencies of the channels and the compression characteristics can be adjusted with potentiometers. Most of the latest hearing aids are electronically programmable. The programmable parameters are downloaded from a computer-based system and stored in digital registers. The register outputs are used to switch resistor networks that control various analog circuitry. The active filters are adjusted to generally provide for low-frequency attenuation of up to 30–40 dB relative to the high-frequency response. This is because most hearing aid wearers require high frequency gain.

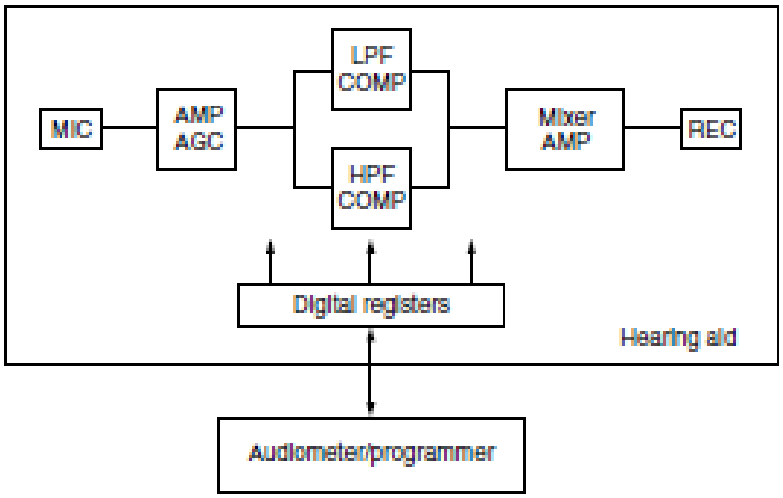


Fig. 17.6 *Conventional analog type hearing aid*

The transducer in a hearing aid, which is a microphone, can be realized in an integrated form with a field-effect transistor preamplifier (Fig. 17.7). The preamplifier is housed in the metallic, microphone case to shield its input from extraneous noise. On the other hand, the receiver is an electromagnetic device, which drives a miniature diaphragm to produce acoustic output. The acoustic output is routed to the ear-mould through a flexible tubing whose frequency

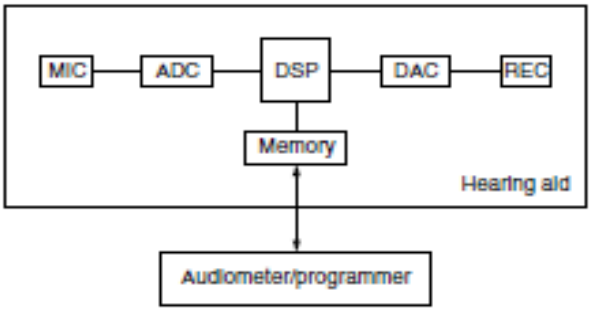
response can be altered to boost the high-frequency response. This is done by tapering its inside diameter from the ear mould back to the receiver port end.
All the electronics circuitry is packaged in a housing, which can be designed for fitting to the ear in any one of the following ways:
1. Placing all the components in a pocket-sized enclosure or box which is connected to the output transducer worn in the ear. The box can be carried in the shirt pocket or carried with a belt around the waist. With the availability of miniature-sized aids, this approach is no longer employed.
2. The components are packaged in a curved module, which is designed to fit comfortably behind the ear.
3. The most popular design is in which the total package can be put inside the outer ear.

Much depends on the performance of the filters for further reduction of the size and improvement in the working of the hearing aids. The dynamic range of an operational amplifier, which is the basic building block of an electronic filter, decreases as the three halves power of feature size. Since dynamic range of analog hearing aids is already marginally acceptable, it appears that further reduction in size to achieve increased processing complexity is not practical. The potential for greater dynamic range, with less power consumption and greater complexity in hearing aid design is feasible only with digital processing technologies.

- **Digital Hearing Aid:-**

A typical digital hearing aid is illustrated in Fig.17.8. The major parts are the microphone, an analog-to-digital converter (ADC), the digital signal processor (DSP), the digital-to-analog converter (DAC), the receiver and a two port memory. Essentially, sound waves picked up by the microphone and transformed into electrical signals are converted into digital form by an A-D converter. A typical microphone will have an internal noise of 20 dB SPL (sound pressure level) when referred to the input and maximum undistorted output corresponding to a signal of about 90 dB SPL. Allowing some margin for peak performance, the total dynamic range required of the ADC is 80 dB. This requirement can be achieved with a 14 bit A-D converter. The DSP is a fixed (wired-program) digital processing device containing an array of adders, multipliers and registers which provide the undamental operations necessary for implementing various digital algorithms. In a general-purpose DSP, considerable power is consumed in executing the programme instructions. Since power consumption is a major consideration in the design of hearing aids, the wired-program approach is followed. The DSP is associated with a two-port memory, which is used to store processing parameters that can be down loaded from the external programmer to the hearing aid while it

is adjusted for the intended user. The dynamic range requirements of the DAC are more sever. Some hearing impaired listeners have almost normal sensitivity at low frequencies but significantly elevated thresholds at high frequencies. Since the conversion noise generated by the DAC has a uniform spectrum and is a function of the overall output signal level, high-level high-frequency sounds can create lowfrequency noise and distortion that falls above the threshold at low frequencies.



▸ **Fig. 17.8** *Block dragram of a digital hearing aid (after Engebreston 1994)*

The digital hearing aids are implemented with CMOS technology, with a feature size of 1 mm or less and with an estimated power consumption of 20 mW. An estimated 10,000 CMOS inverters are required to implement 400,000 multiply-add operations for filtering, compression functions and other processing requirements.The digital hearing aids promise to provide capabilities of superior signal processing, ease of fitting and stable long-term performance. However, they are still under development. It has often been seen that a person buys a hearing aid but does not use it because it does not help very much. The basic reason is that the impaired ear has its capacity to process speech and hearing aids are simply sound amplifiers that do not compensate for the loss of processing power. It needs to be emphasized that today's hearing aids are at an early stage of development and need to reach a highly refined stage before they can find wide spread and useful applications. The potential areas of improvement include shaping the frequency response to invert the patient's hearing loss, enhancing the signal-to-noise ratio with adaptive filtering, reducing acoustic feedback and compressing/expanding signals with minimum distortion.

Question Bank

1) Explain generalized block diagram of a diagnostic medical instrumentation system.
2) Classify medical instruments based on different working principles.
3) Describe working principle of electrocardiograph and explain its block diagram.
4) Classify and explain bipolar and unipolar leads used for ECG measurements.
5) Explain block diagram of vector cardiograph.
6) Explain phonocardiograph in detail.
7) Define 10-20 electrode placement method used for EEG.
8) Explain EEG with its working principle.
9) Explain EMG with suitable diagram.
10) Explain biofeedback instrumentation with neat diagram.
11) Explain average heart rate meter.
12) Explain instantaneous heart rate meter.
13) Describe techniques of BP measurements.
14) Draw and explain apnoea monitor.
15) Explain pulse oxymeter with suitable diagram.
16) Explain ear oxymeter with suitable diagrams.
17) Draw and explain basic spirometer.
18) Draw and explain wedge spirometer.
19) What is air and bone conduction? Explain.
20) Explain types of audiometers.
21) Explain hearing aid with suitable diagram.

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