MODULE -2

BIOELECTRIC SIGNALS AND ELECTRODES

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2.1: Sources of Biomedical Signals

Biomedical signals are those signals (phenomenon that conveys information) which are used primarily for extracting information on a biological system under investigation.

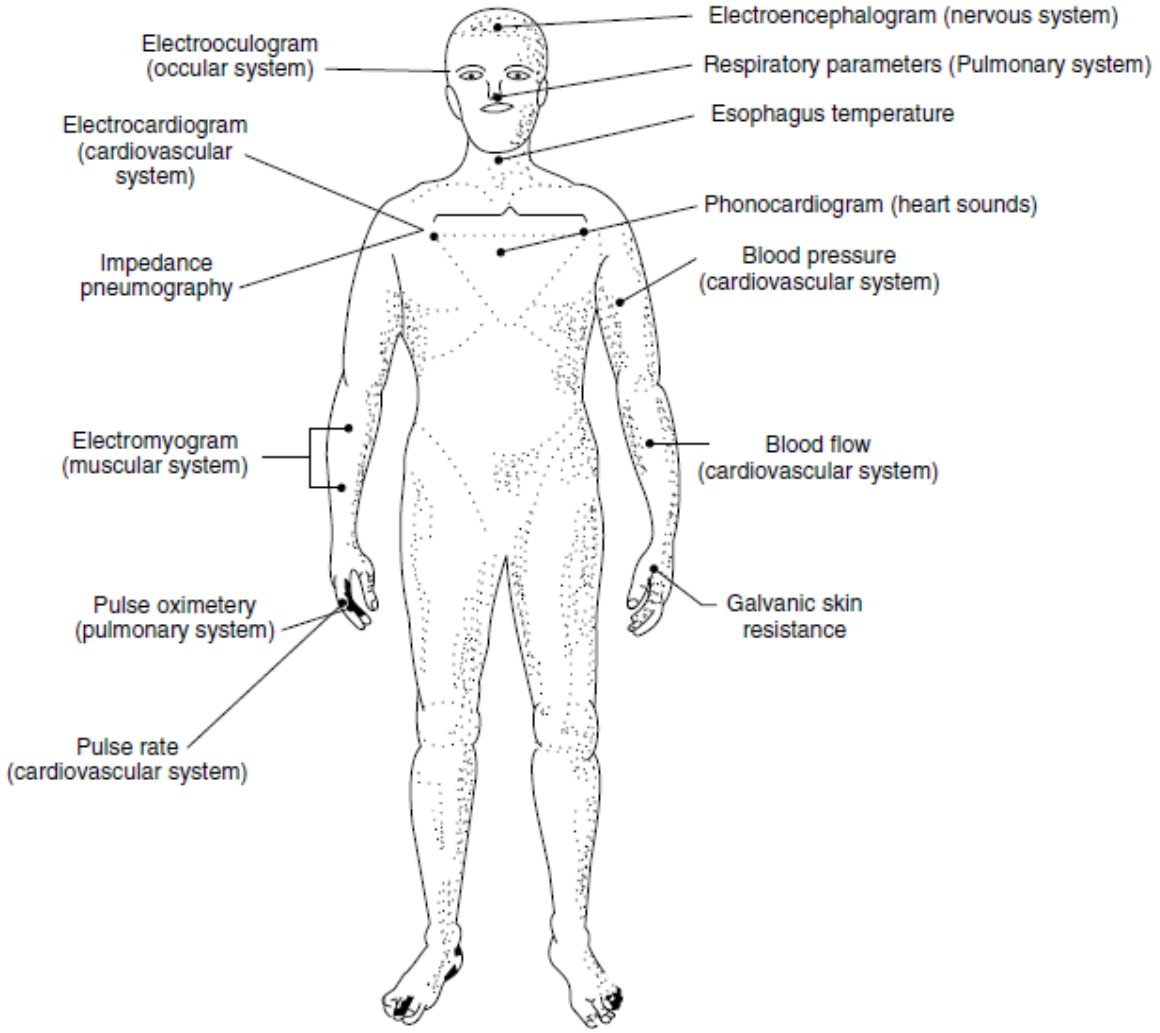

Fig. 1.8 ***Sources of biomedical signals***

The process of extracting information could be as simple as feeling the pulse of a person on the wrist or as complex as analyzing the structure of internal soft tissues by an ultrasound scanner. Biomedical signals originate from a variety of sources (Fig. 1.8) such as:

Bioelectric Signals: These are unique to the biomedical systems. They are generated by nerve cells and muscle cells. Their basic source is the cell membrane potential which under certain conditions may be excited to generate an action potential. The electric field generated by the action of many cells constitutes the bio-electric signal. The most common examples of bioelectric signals are the ECG (electrocardiographic) and EEG (electroencephalographic) signals.

Bioacoustic Signals: The measurement of acoustic signals created by many biomedical phenomena provides information about the underlying phenomena. The examples of such signals are: flow of blood in the heart, through the heart's valves and flow of air through the upper and lower airways and in the lungs which generate typical acoustic signal.

Biomechanical Signals: These signals originate from some mechanical function of the biological system. They include all types of motion and displacement signals, pressure and flow signals etc. The movement of the chest wall in accordance with the respiratory activity is an example of this type of signal.

Biochemical Signals: The signals which are obtained as a result of chemical measurements from the living tissue or from samples analyzed in the laboratory. The examples are measurement of partial pressure of carbon-dioxide (pCO2), partial pressure of oxygen (pO2) and concentration of various ions in the blood.

Biomagnetic Signals: Extremely weak magnetic fields are produced by various organs such as the brain, heart and lungs. The measurement of these signals provides information which is not available in other types of bio-signals such bio-electric signals. A typical example is that of magneto-encephalograph signal from the brain.

Bio-optical Signals: These signals are generated as result of optical functions of the biological systems, occurring either naturally or induced by the measurement process. For example, blood oxygenation may be estimated by measuring the transmitted/back scattered light from a tissue at different wavelengths.

Bio-impedance Signals: The impedance of the tissue is a source of important information concerning its composition, blood distribution and blood volume etc. The measurement of galvanic skin resistance is a typical example of this type of signal. The bio-impedance signal is also obtained by injecting sinusoidal current in the tissue and measuring the voltage drop generated by the tissue impedance. The measurement of respiration rate based on bio-impedance technique is an example of this type of signals.

2.2: Origin of Bioelectric Signals

The association of electricity with medical science dates back to the 18th century when Galvani demonstrated that most of the physiological processes were accompanied with electrical changes. This discovery formed the basis of the explanation of the action of living

tissues in terms of bioelectric potentials. It is now well established that the human body, which is composed of living tissues, can be considered as a power station generating multiple electrical signals with two internal sources, namely muscles and nerves.

Normal muscular contraction is associated with the migration of ions which generates potential differences measurable with suitably placed electrodes. For example, the heart and the brain produce characteristic patterns of voltage variations which when recorded and analyzed are useful in both clinical practice and research. Potential differences are also generated by the electrochemical changes accompanied with the conduction of signals along the nerves to or from the brain. These signals are of the order of a few microvolts and give rise to a complicated pattern of electrical activity when recorded. The fact that the activity of the living tissues is due to the potential changes in them suggested the use of external electricity for the diagnosis of certain diseases affecting muscles and nerves, for the augmentation or replacement of a deficient natural activity or for the restoration of a palsied muscle.

Bioelectric potentials are generated at a cellular level and the source of these potentials is ionic in nature. A cell consists of an ionic conductor separated from the outside environment by a semipermeable membrane which acts as a selective ionic filter to the ions. This means that some ions can pass through the membrane freely where as others cannot do so. All living matter is composed of cells of different types. Human cells may vary from 1 micron to 100 microns in diameter, from 1 mm to 1 m in length, and have a typical membrane thickness of 0.01 micron (Peter Strong, 1973). Surrounding the cells of the body are body fluids, which are ionic and which provide a conducting medium for electric potentials. The principal ions involved with the phenomena of producing cell potentials are sodium (Na^+), potassium (K^+) and chloride ($Cl^–$). The membrane of excitable cells readily permits the entry of K^+ and $Cl^–$ but impedes the flow of Na^+ even though there may be a very high concentration gradiant of sodium across the cell membrane. This results in the concentration of the sodium ion more on the outside of the cell membrane than on the inside. Since sodium is a positive ion, in its resting state, a cell has a negative charge along the inner surface of its membrane and a positive charge along the outer portion.

The unequal charge distribution is a result of certain electrochemical reactions and processes occurring within the living cell and the potential measured is called the resting potential. The

cell in such a condition is said to be polarized. A decrease in this resting membrane potential difference is called depolarization..

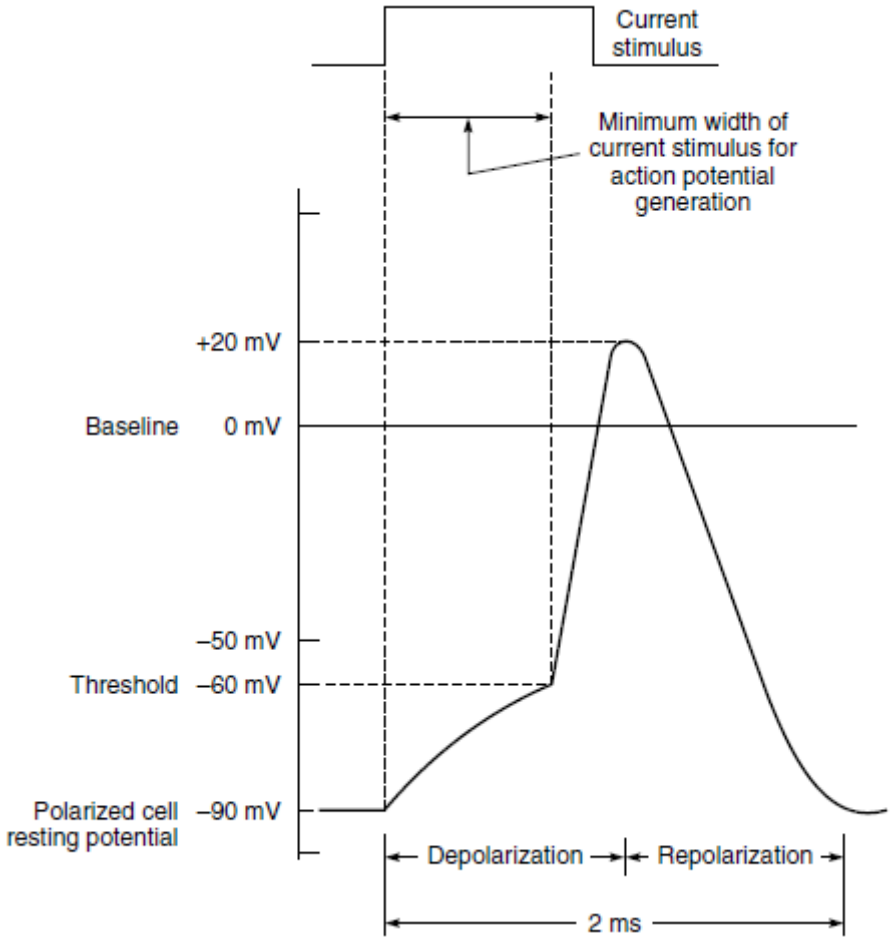

Fig. 2.1 ***A typical cell potential waveform***

The distribution of positively charged ions on the outer surface and negatively charged ionsinside the cell membrane results in the difference of potential across it and the cell becomes,in effect, a tiny biological battery. Experiments have shown that the internal resting potentialwithin a cell is approximately –90 mV with reference to the outside of the cell. When the cellis excited or stimulated, the outer side of the cell membrane becomes momentarily negativewith respect to the interior. This process is called depolarization and the cell potential changes to approximately +20 mV. Repolarization then takes place a short time later when the cell regains its normal state in which the inside of the membrane is again negative with respect to the outside. Repolarization is necessary in order to re-establish the resting potential. This discharging and recharging of the cell produces the voltage waveforms

which can be recorded by suitable methods using microelectrodes. A typical cell potential waveform so recorded is shown in Fig. 2.1

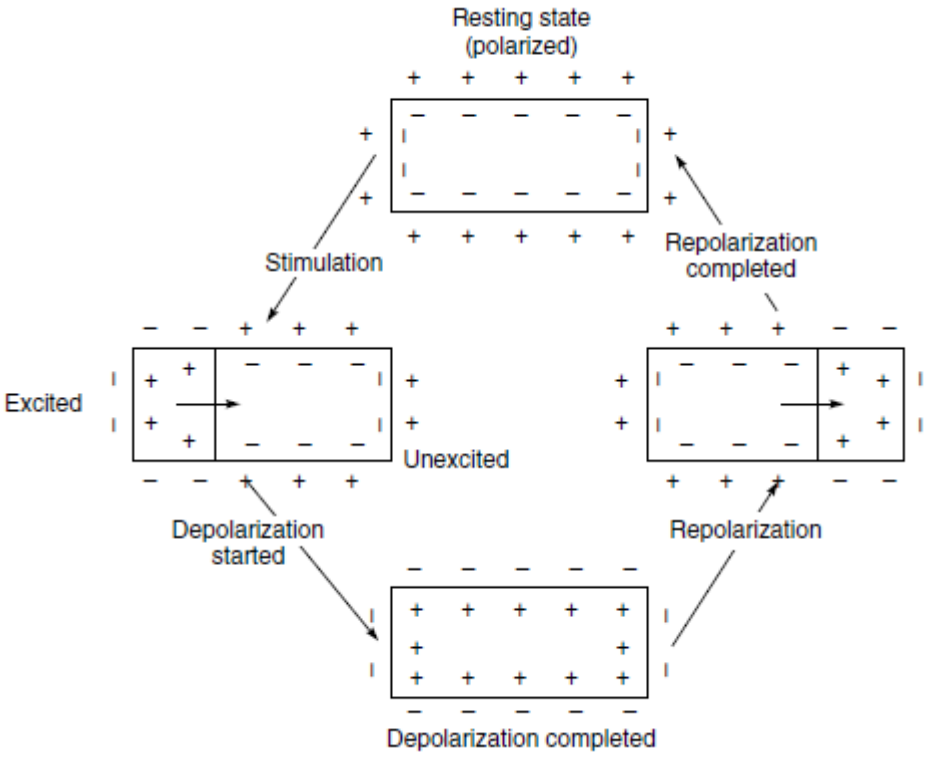

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Fig. 2.2 Electrical activity associated with one contraction in a muscle

The wave of excitation while propagating in the muscle causes its contraction. The contraction wave always follows the excitation wave because of its lower velocity. This phenomenon is found with the skeletal muscles, the heart muscle and the smooth muscles. In its turn, every contraction (movement) of a muscle results in the production of an electric voltage. This voltage occurs in the muscle in such a way that the moving muscle section is always negative with respect to its surroundings. These voltages are called action potentials because they are generated by the action of the muscles. After complete contraction, repolarization takes place resulting in the relaxation of the muscle and its returning to the original state. Figure 2.2 shows electrical activity associated with one contraction in a muscle

The currents involved in bioelectricity are unlike the currents involved in electronics. Bioelectric currents are due to positive and negative ion movement within a conductive fluid. The ions possess finite mass and encounter resistance to movement within the fluid for they have limited speeds.

The cell action potential, therefore, shows a finite rise time and fall time. It may be noted that a cell may be caused to depolarize and then repolarize by subjecting the cell membrane to an ionic current. However, unless a stimulus above a certain minimum value is applied, the cell will not be depolarized and no action potential is generated. This value is known as the stimulus threshold. After a cell is stimulated, a finite period of time is required for the cell to return to its pre-stimulus state. This is because the energy associated with the action potential is developed from metabolic processes within the cell which take time for completion. This period is known as refractory period.

• Table 2.1 *Bioelectric Signals*

Parameter	*Primary signal characteristics*	*Type of Electrode*
Electrocardiography (ECG)	Frequency range: 0.05 to 120 Hz Signal amplitude: 0.1 to 5 μV Typical signal: 1 μV	Skin electrodes
Electroencephalography (EEG)	Frequency range: 0.1 to 100 Hz Signal amplitude: 2 to 200 μV Typical signal: 50 μV	Scalp electrodes
Electromyography (EMG)	Frequency range: 5 to 2000 Hz Signal amplitude: 0.1 to 5 μV	Needle electrodes
Electroretinography (ERG)	Frequency range: dc to 20 Hz Signal amplitude: 0.5 μV to 1 μV Typical signal: 0.5 μV	Contact electrodes
Electro-oculography (EOG)	Frequency range: dc to 100 Hz Signal amplitude: 10 to 3500 μV Typical signal: 0.5 μV	Contact electrodes

The bioelectric signals of clinical interest, which are often recorded, are produced by the coordinated activity of large groups of cells. In this type of synchronized excitation of many cells, the charges tend to migrate through the body fluids towards the still unexcited cell areas. Such charge migration constitutes an electric current and hence sets up potential differences between various portions of the body, including its outer surface. Such potential differences can be conveniently picked up by placing conducting plates (electrodes) at any two points on the surface of the body and measured with the help of a sensitive instrument. These potentials are highly significant for diagnosis and therapy. The primary characteristics of typical bioelectric signals are given in Table 2.1.

2.3 : Electrocardiogram (ECG)

The recording of the electrical activity associated with the functioning of the heart is known as electrocardiogram. ECG is a quasi-periodical, rhythmically repeating signal synchronized

by the function of the heart, which acts as a generator of bioelectric events. This generated signal can be described by means of a simple electric dipole (pole consisting of a positive and negative pair of charge). The dipole generates a field vector, changing nearly periodically in time and space and its effects are measured on the surface. The waveforms thus recorded have been standardized in terms of amplitude and phase relationships and any deviation from this would reflect the presence of an abnormality. Therefore, it is important to understand the electrical activity and the associated mechanical sequences performed by the heart in providing the driving force for the circulation of blood. The heart has its own system for generating and conducting action potentials through a complex change of ionic concentration across the cell membrane. Located in the top right atrium near the entry of the vena cava, are a group of cells known as the sino-atrial node (SA node) that initiate the heart activity and act as the primary pace maker of the heart (Fig. 2.3). The SA node is 25 to 30 mm in length and 2 to 5 mm thick. It generates impulses at the normal rate of the heart,

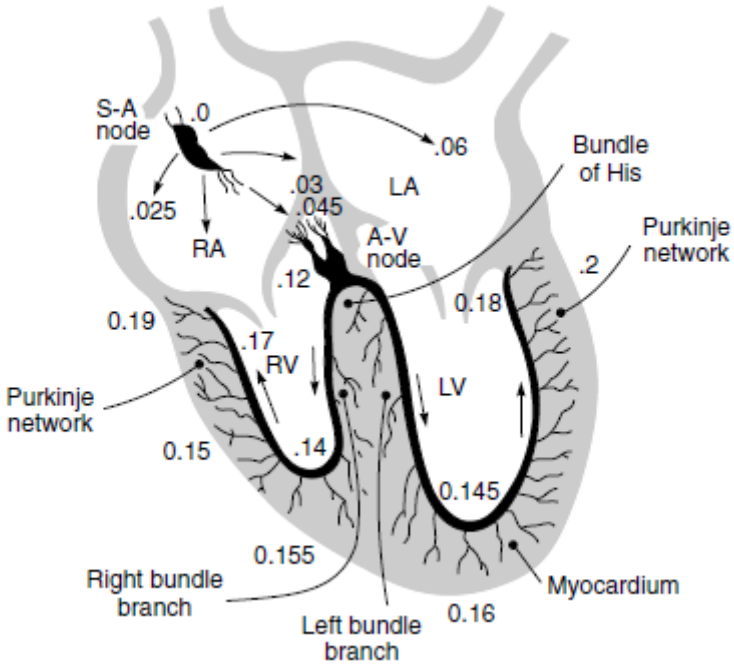


Fig. 2.3 ***The position of the sino-atrial node in the heart from where the impulse responsible for the electrical activity of the heart originates. The arrow shows the path of the impulse.***
Note: The numbers like 0.18, 0.145, 0.15, 0.2 ... etc. indicate the time taken for the impulse to travel from the S-A node to various parts of the heart

about 72 beats per minute at rest. Because the body acts as a purely resistive medium, the potential field generated by the SA node extends to the other parts of the heart. The wave propagates through the right and left atria at a velocity of about 1 m/s. About 0.1 s are required for the excitation of the atria to be completed. The action potential contracts the

atrial muscle and the impulse spreads through the atrial wall about 0.04s to the AV (atrio-ventricular) node. This node is located in the lower part of the wall between the two atria.
The AV node delays the spread of excitation for about 0.12 s, due to the presence of a fibrous barrier of non-excitable cells that effectively prevent its propagation from continuing beyond the limits of the atria. Then, a special conduction system, known as the bundle of His (pronounced as hiss) carries the action potential to the ventricles. The atria and ventricles are thus functionally linked only by the AV node and the conduction system. The AV node delay ensures that the atria complete their contraction before there is any ventricular contraction. The impulse leaves the AV node via the bundle of His. The fibres in this bundle, known as Purkinje fibres, after a short distance split into two branches to initiate action potentials simultaneously in the two ventricles.

Conduction velocity in the Purkinje fibres is about 1.5 to 2.5 m/s. Since the direction of the impulse propagating in the bundle of His is from the apex of the heart, ventricular contraction begins at the apex and proceeds upward through the ventricular walls. This results in the contraction of the ventricles producing a squeezing action which forces the blood out of the ventricles into the arterial system. Figure 2.3 shows the time for action potential to propagate to various areas of the heart.
The normal wave pattern of the electrocardiogram is shown in Fig. 2.4. The *PR* and *PQ i*nterval, measured from the beginning of the *P* wave to the onset of the *R* or *Q* wave respectively, marks the time which an impulse leaving the SA node takes to reach the ventricles. The *PR*interval normally lies between 0.12 to 0.2 s. The *QRS*interval, which represents the time taken by the heart impulse to travel first through the interventricular system and then through the free walls of the ventricles, normally varies from 0.05 to 0.10s.

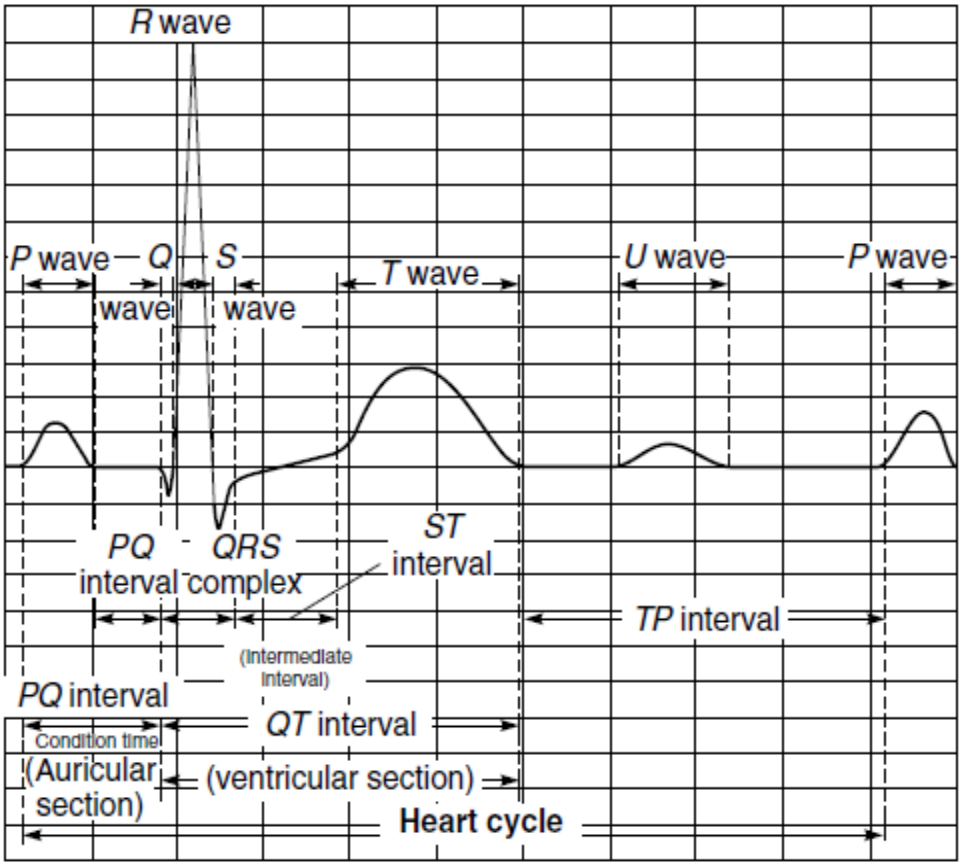


Fig. 2.4 ***Normal wave pattern of an ECG waveform recorded in the standard lead position***

The*T* wave represents repolarization of both ventricles. The *QT* interval, therefore, is the period for one complete ventricular contraction (systole). Ventricular diastole, starting from the end of the *T* wave extends to the beginning of the next *Q*wave. Typical amplitude of *QRS*is 1 mV for a normal human heart, when recorded in lead 1 position.

2.4 : Electroencephalogram (EEG)

The brain generates rhythmical potentials which originate in the individual neurons of the brain. These potentials get summated as millions of cell discharge synchronously and appear as a surface waveform, the recording of which is known as the electroencephalogram (Fig. 2.5). The neurons, like the other cells of the body, are electrically polarized at rest. The interior of the neuron is at a potential of about –70 mV relative to the exterior. When a neuron is exposed to a stimulus above a certain threshold, a nerve impulse, seen as a change in membrane potential, is generated which spreads in the cell resulting in the depolarization of the cell. Shortly afterwards, repolarization occurs.

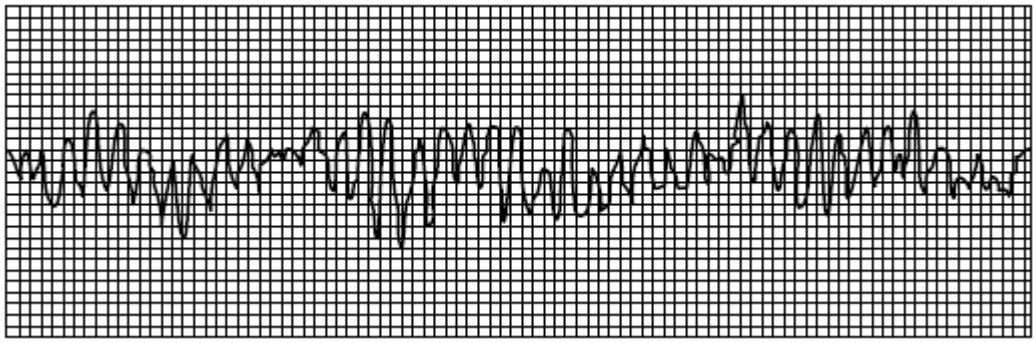


Fig. 2.5 ***Typical EEG signal waveform***

The EEG signal can be picked up with electrodes either from the scalp or directly from the cerebral cortex. The peak-to-peak amplitude of the waves that can be picked up from the scalp is normally 100 μV or less while that on the exposed brain, is about 1 mV. The frequency varies greatly with different behavioural states. The normal EEG frequency content ranges from 0.5 to 50 Hz. The nature of the wave varies over the different parts of the scalp. The variations in EEG signals both in terms of amplitude and frequency are of diagnostic value. Frequency information is particularly significant since the basic frequency of the EEG range is classified into the following five bands for purposes of EEG analysis:

Delta (d) 0.5 – 4 Hz

Theta (q) 4 – 8 Hz

Alpha (a) 8 – 13 Hz

Beta (b) 13 – 22 Hz

Gamma (g) 22 – 30 Hz

The alpha rhythm is one of the principal components of the EEG and is an indicator of the state of 'alertness' of the brain. It serves as an indicator of the depth of anaesthesia in the operating room. The frequency of the EEG seems to be affected by the mental activity of a person. The wide variation among individuals and the lack of repeatability in a given person from one occasion to another makes the analysis a difficult proposition. However, certain characteristic EEG waveforms can be conveniently related to gross abnormalities like epileptic seizures and sleep disorders. Besides the importance of the frequency content of the EEG pattern, phase relationships between similar EEG patterns from different parts of the brain are also being studied with great interest in order to obtain additional knowledge regarding the functioning of the brain. Another important measurement is the recording of 'evoked response', which indicates the disturbance in the EEG pattern resulting from

external stimuli. The stimuli could be a flash of light or a click of sound. Since the responses to the stimuli are repeatable, the evoked response can be distinguished from the rest of the EEG activity by averaging techniques to obtain useful information about the functioning of particular parts of the brain.

2.5: Electromyogram (EMG)

The contraction of the skeletal muscle results in the generation of action potentials in the individual muscle fibres, a record of which is known as electromyogram.

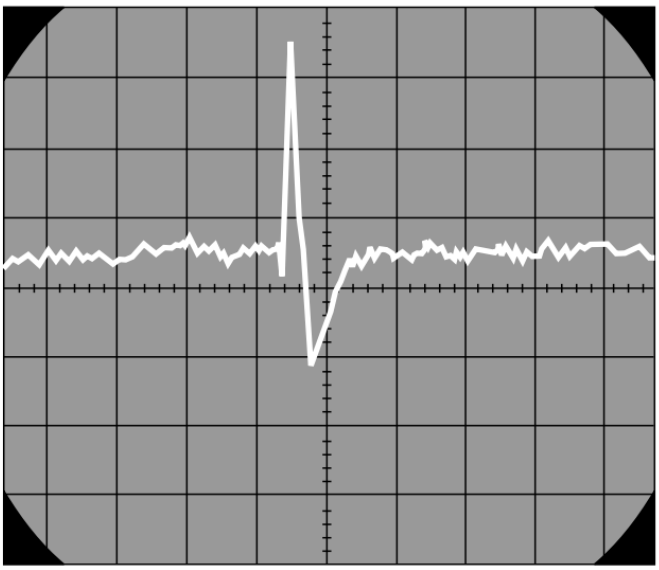
The activity is similar to that observed in the cardiac muscle, but in the skeletal muscle, repolarization takes place much more rapidly, the action potential lasting only a few milliseconds.

Since most EMG measurements are made to obtain an indication of the amount of activity of a given muscle, or a group of muscles, rather than of an individual muscle fibre, the EMG pattern is usually a summation of the individual action potentials from the fibres constituting the muscle or muscles being studied.

The electrical activity of the underlying muscle mass can be observed by means of surface electrodes on the skin.

However, it is usually preferred to record the action potentials from individual motor units for better diagnostic information using needle electrodes. In voluntary contraction of the skeletal muscle, the muscle potentials range from 50 mV to 5 mV and the duration from 2 to 15 ms. The values vary with the anatomic position of the muscle and the size and location of the electrode. In a relaxed muscle, there are normally no action potentials. A

typical EMG signal is shown in Fig. 2.6.



➤ **Fig. 2.6** *Waveshape of a typical EMG signal*

2.6: Electrooculogram (EOG)

Electro-oculography is the recording of the bio-potentials generated by the movement of the eye ball. The EOG potentials are picked up by small surface electrodes placed on the skin near the eye.

One pair of electrodes is placed above and below the eye to pick up voltages corresponding to vertical movements of the eye ball. Another pair of electrodes is positioned to the left and right of the eye to measure horizontal movement.

The recording pen is centred on the recording paper, corresponding to the voltage changes accompanying it. EOG has applications mostly for research and is not widely used for clinical purposes.

2.7: Electroretinogram (ERG)

It is found that an electrical potential exists between the cornea and the back of the eye. This potential changes when the eye is illuminated.

The process of recording the change in potential when light falls on the eye is called electroretinography.

ERG potentials can be recorded with a pair of electrodes. One of the electrodes is mounted on a contact lens and is in direct contact with the cornea.

The other electrode is placed on the skin adjacent to the outer corner of the eye. A reference electrode may be placed on the forehead.

A general purpose direct writing recorder may be used for recording electroretinograms.

The magnitude of the ERG voltage depends upon the intensity and duration of the light falling on the eye. It may be typically about 500 mV.

2.8: Recording Electrodes–Electrode-tissue interface,

Bioelectric events have to be picked up from the surface of the body before they can be put into the amplifier for subsequent record or display. This is done by using electrodes.

Electrodes make a transfer from the ionic conduction in the tissue to the electronic conduction which is necessary for making measurements.

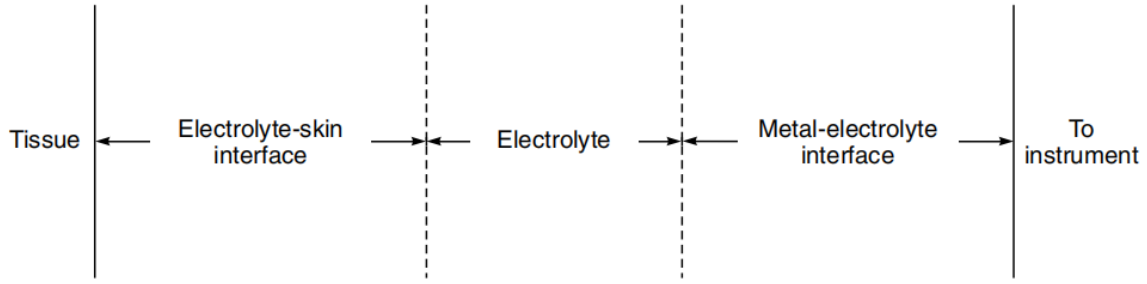
Two types of electrodes are used in practice-surface electrodes and the deep-seated electrodes. The surface electrodes pick up the potential difference from the tissue surface when placed over it without damaging the live tissue.

2.8.1 ELECTRODE TISSUE INTERFACE

The most commonly used electrodes in patient monitoring and related studies are surface electrodes. The notable examples are when they are used for recording ECG, EEG and respiratory activity by impedance pneumography.

In order to avoid movement artefacts and to obtain a clearly established contact (low contact impedance) an electrolyte or electrode paste is usually employed as an interface between the

electrode and the surface of the source of the event. Figure 2.7 (a, b) represent the electrode-tissue interface.

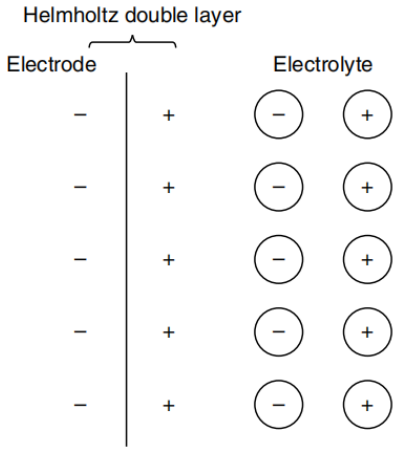


➤ **Fig. 2.7(a)** *Electrode-tissue interface for surface electrodes used with electrode jelly*

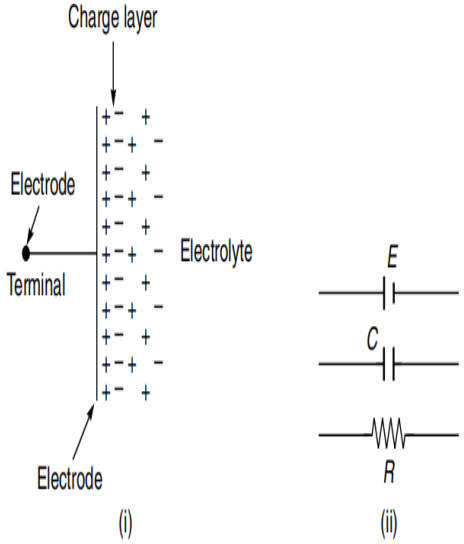
The characteristic of a surface electrode composed of a metal electrode and attached to the surface of the body through an electrolyte (electrode jelly) are dependent upon the conditions at the metal-electrolyte interface, the electrolyte-skin interface and the quality of the electrolyte.

Metal-Electrolyte Interface:

At the metal-electrolyte transition, there is a tendency for each electrode to discharge ions into the solution and for ions in the electrolyte to combine with each electrode. The net result is the creation of a charge gradient (difference of potential) at each electrode, the spatial arrangement of which is called the electrical double layer (Fig. 2.7(c)). The double layer is known to be present in the region immediately adjacent to the electrode and can be represented, in its simplest form, as two parallel sheets of charge of opposite sign separated by a thin film of dielectric. Therefore, the metal-electrolyte interface appears to consist of a voltage source in series with a parallel combination of a capacitance and reaction resistance. The voltage developed is called the half-cell potential.



▸ **Fig. 2.7(b)** *Electrode tissue interface circuit involves transfer of electrons from the metal phase to an ionic carrier in the electrolyte, a charge double layer (capacitance) forms at the interface*



▸ **Fig. 2.7(c)** *(i) Charge distribution at electrode-electrolyte interface (ii) Three components representing the interface*

Electrolyte-Skin Interface:

An approximation of the electrolyte-skin interface can be had by assuming that the skin acts as a diaphragm arranged between two solutions (electrolyte and body fluids) of different concentrations containing the same ions, which is bound to give potential differences.

The simplest equivalent representation could then be described as a voltage source in series with a parallel combination of a capacitance and resistance. The capacitance represents the charge developed at the phase boundary whereas the resistance depends upon the conditions associated with ion-migration along the phase boundaries and inside the diaphragm.

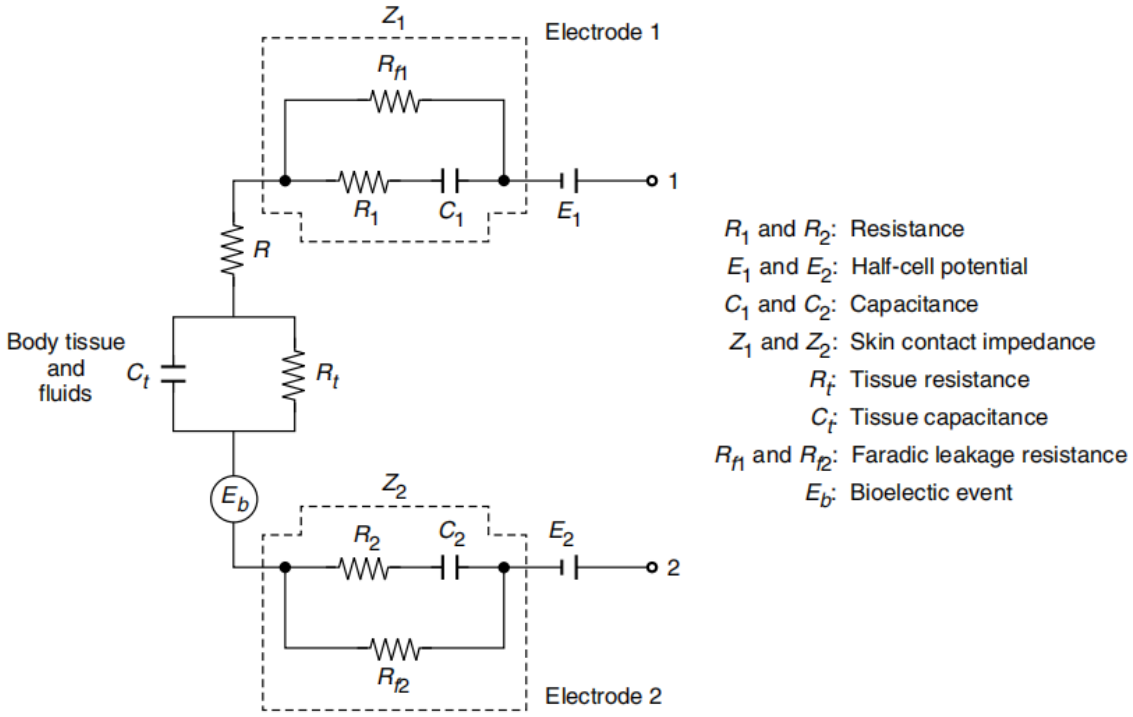
The above discussion shows that there is a possibility of the presence of voltages of Non physiological origin. These voltages are called contact potentials.

The electrical equivalent circuit of the surface electrode suggests that the voltage presented to the measuring instrument from the electrode consists of two main components. One is the contact potential and the other is the biological signal of interest.

The contact potential depends upon several factors and may produce an interference signal which exceeds several times the useful signal. The contact potential is found to be a function of the type of skin, skin preparation and composition of the electrolyte.

The electrodes are used to measure a bioelectric event and are connected to a differential amplifier. Three potentials are found to exist in this circuit (Fig. 2.9), one is due to the bioelectric event (*Eb*) and the other two are non-physiologicand represent the half-cell potentials (*E*1 and *E*2) of the electrodes. *Z*1 and *Z*2 are the skin contact impedances of these electrodes and *R* is the tissue resistance or resistance of the bioelectric generator.

This circuit shows that the impedance of the electrodes would be high in the low frequency region and it would decrease with increasing frequency. It is further clear that in the measurement of a bioelectric signal, it is essential to minimize potential drops across the electrode impedance. This is achieved by making the skin-contact impedance as low as possible and making the input impedance of the measuring device as high as possible.



➤ **Fig. 2.9** *Equivalent circuit for a pair of electrodes (1,2) on a subject represented by R R_t, C_t. Embedded in the subject is a bioelectric generator E_b (after Tacker and Geddes, 1996)*

POLARISATION

If a low voltage is applied to two electrodes placed in a solution, the electrical double layers are disturbed. Depending on the metals constituting the electrodes, a steady flow of current may or may not take place.

In some metal/liquid interfaces, the electrical double layer gets temporarily disturbed by the externally applied voltage, and therefore, a very small current flows after the first surge, thus indicating a high resistance. This type of electrode will not permit the measurement of steady or slowly varying potentials in the tissues.

They are said to, be polarized or nonreversible. Thus, the phenomenon of polarization affects the electro-chemical double layer on the electrode surface and manifests itself in changing the value of the impedance and voltage source representing the transition layer.

Parsons (1964) stated that electrodes in which no net transfer of charge takes place across the metal-electrolyte interface can be termed as perfectly polarized. Those in which unhindered exchange of charge is possible are called non-polarizable or reversible electrodes. The ionic double layer in metals of these electrodes is such that they allow considerable current to flow when a small voltage is applied, thus offering a low resistance.

SKIN CONTACT IMPEDANCE :

Measurement of Skin Contact Impedance: A convenient method to measure the contact impedance at any individual electrode is shown in Fig. 2.11.

The three electrodes, A, B and C, have contact impedance respectively of Za, Zb andZc. An oscillator provides a constant current in the frequency range of 0.1–100 Hz through the 47 kW series resistor.

By suitably positioning the switch, a sensitive oscilloscope can be used to monitor either the voltage dropped across the 1 kW resistor or the voltage dropped across Zb.

The voltage drop across Zb can be neglected since the input impedance of the oscilloscope used with an input probe is usually high. From the voltage dropped across the 1 kW resistor it is possible to calculate the circuit current and thus to obtain a value for Zc.

Using this technique, the skin contact impedance of the following types of electrodes were measured by Hill and Khandpur (1969).

• Plastic cup self-adhesive electrodes (Boter et al, 1966)

• Metal plate limb electrodes used with conducting jelly

• Metal plate electrodes used with conducting plastic (Jenkner, 1967)

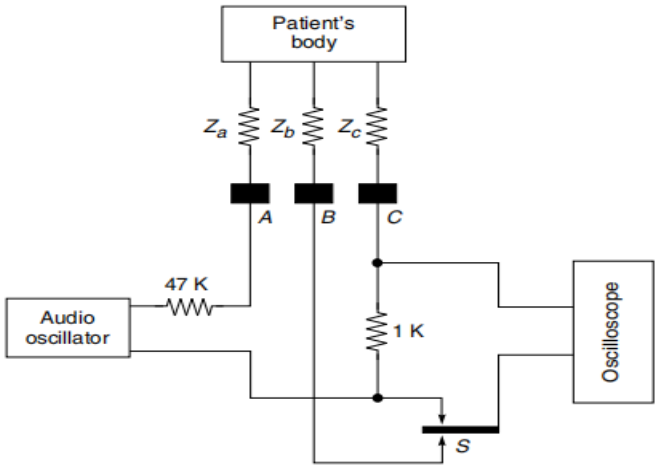
• Dry multi-point limb electrodes (Lewes, 1966)

• Dry multi-point suction chest electrodes

• Self-adhesive multi-point chest electrodes used with conducting jelly

• Self-adhesive gauze electrodes

• Self-adhesive dry multi-point chest electrodes (Lewes and Hill, 1967)



➤ **Fig. 2.11** *Arrangement for measurement electrode skin-contact impedance for surface electrodes*

Motion Artifacts

Motion artefact is a problem in biopotential measurements. The problem is greatest in cardiac stress laboratories where the exercise ECG is recorded. The problem is also serious in coronary care units where patients are monitored for relatively long periods.

Motion of the subject under measurement creates artefacts which may even mask the desired signal or cause an abrupt shift in the baseline. These artefacts may result in a display being

unreadable, a recording instrument exceeding its range, a computer yielding incorrect output or a false alarm being triggered by the monitoring device.

Tam and Webster (1977) concluded that the skin-electrolytic paste interface is the major source of motion artefact. When a metal electrode contacts an electrolytic paste, a half cell potential is generated at the electrode-paste interface. Kahn (1965) demonstrated that when polarizable metal-plate electrodes are used, the electrode-paste interface can be a source of motion artefact.

When the paste is agitated, the half-cell potential varies because of the altered metallic ion gradient at the interface. He recorded a 1 mV offset potential change from a silver-silver chloride electrode exposed to a flowing stream of saline solution, as contrasted to 30 mV change for some silver electrodes.

Motion artefact is reduced to a negligible magnitude by skin abrasion. However, when the skin is abraided, it is more susceptible to irritants. The possible sources for skin irritation include the electrode, the paste and the adhesive. When large currents flow through metallic electrodes, migration of some ions into the skin can cause irritation.

2.9 Silver-Silver Chloride electrodes

Production of Silver-Silver Chloride Electrodes: Silver-silver chloride electrodes are normally prepared by electrolysis. Two silver discs are suspended in a saline solution. The positive pole of a dc supply is connected to the disc to be chlorided and the negative pole goes to the other disc. A current at the rate of 1 mA/cm^2 of surface area is passed through the electrode for several minutes. A layer of silver chloride is thus deposited on the surface of the anode. The chemical changes that take place at the anode and cathode respectively are:

$$NaCl = Na^+ + Cl^-$$
$$Cl^- + Ag^+ \rightarrow AgCl$$

The positively charged sodium ions generate hydrogen when they reach the cathode surface.

$$2Na^+ + 2H_2O + 2 \text{ electrons} \rightarrow 2NaOH + H_2$$

To prepare silver-silver chloride electrodes of good quality, only pure silver should be used and the saline solution should be made from analar grade sodium chloride. Before chloriding, silver must be cleaned—preferably by the electrolytic method.

2.10 ELECTRODES FOR ECG

LIMB ELECTRODE:

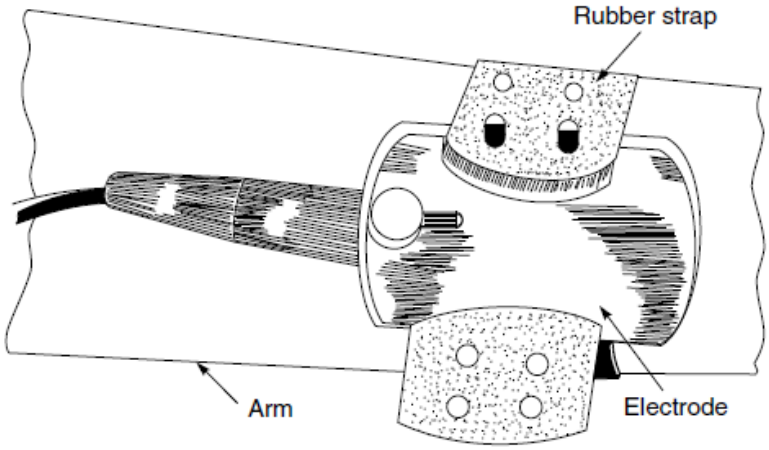
The most common type of electrodes routinely used for recording ECG are rectangular or circular surface electrodes (Fig. 2.13).

The material used is german silver, nickel silver or nickel plated steel. They are applied to the surface of the body with electrode jelly.

The typical value of the contact impedance of these electrodes, which are of normal size, is nearly 2 to 5 kW when measured at 10 Hz.

The electrodes are held in position by elastic straps. They are also called limb electrodes as they are most suitable for application on the four limbs of the body.

The size of the limb electrodes is usually 3 \ 5 cm and they are generally made of german silver, an alloy of zinc, copper and nickel. They are reusuable and last several years.



➤ **Fig. 2.13** *ECG plate electrode. The electrode is usually fastened to the arm or leg with a perforated rubber strap which keeps it in position during ECG recording*

Limb electrodes are generally preferred for use during surgery because the patient's limbs are relatively immobile. Moreover, chest electrodes cannot be used as they would interfere with the surgery.

Limb electrodes are not suitable for use in long-term patient monitoring because the long flowing leads are inconvenient to the patient. Also, the electromyographic voltages generated by the activity of the limb muscles makes them unsuitable for use when monitoring conscious and semi-conscious patients.

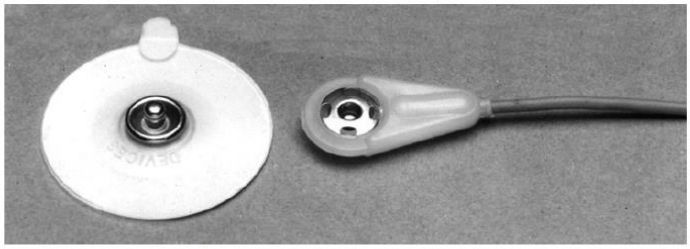
Suction-cup electrode is commonly used to record the unipolar chest leads. It has a high contact impedance as only the rim of the electrode is in contact with the skin. The electrode is

popular for its practicality, being easily attachable to fleshy parts of the body. Electrode jelly forms the vacuum seal.

FLOATING ELECTRODES:

Limb electrodes generally suffer from what is known as motion artefacts caused due to the relative motion at the interface between the metal electrode and the adjacent layer of electrode jelly.

The interface can be stabilized by the use of floating electrodes in which the metal electrode does not make direct contact with the skin. The electrode (Fig. 2.14) consists of a light-weight metalled screen or plate held away from the subject by a flat washer which is connected to the skin. Floating electrodes can be recharged, i.e. the jelly in the electrodes can be replenished if desired.



➤ Fig. 2.14 *Light weight floating electrode with press stud for long-term monitoring of ECG*

Connection with the instrument is established with silver-plated copper wires fixed in the conducting adhesive. The type of electrodes are extremely light-weight and do not make use of electrode jelly.

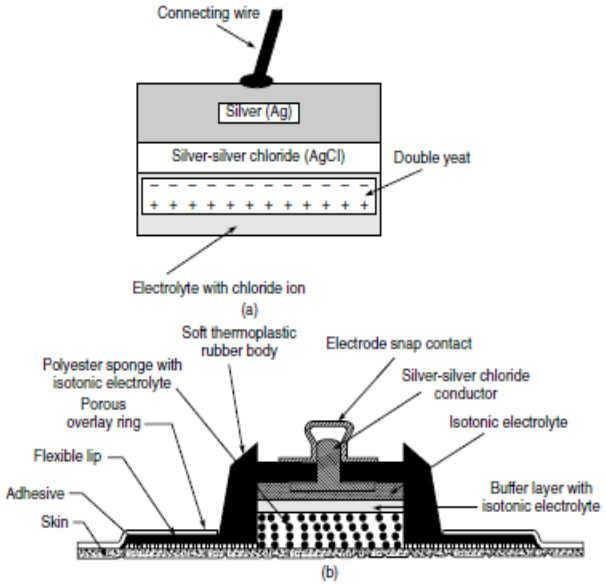
This makes them ideal for use in monitoring the ECG of exercising subjects and aeroplane pilots as they give rise to minimal motion artefacts. The contact impedance shown by these electrodes is of the order of 50 kW.

PREGELLED DISPOSABLE ELECTRODE

Electrodes which are employed in stress testing or long term monitoring, present additional problems because of the severe stresses, perspiration and major body movement encountered in such studies.

Both design considerations and application techniques of electrodes used in electrocardiography are necessary to prevent random noise on the baseline, baseline wandering and skin contact over extended periods causing a loss of signal.

To overcome problems due to prolonged application, special disposable electrodes have been developed.



➤ **Fig. 2.15** (a) *Principle of pre-gelled ECG electrode made of silver-silver chloride. The electrode has electrolyte layers that are made of a firm gel which has adhesive properties. The firm gel minimizes the disturbance of the charge double layer*
(b) *Cross-section of a typical pre-gelled electrode*

Figure 2.15(a) illustrates the principle of a pregelled electrode while Fig. 2.15(b) shows a cross-section of such an electrode. The main design feature of these electrodes which helps in reducing the possibility of artefacts, drift and baseline wandering is the provision of a high-absorbancy buffer layer with isotonic electrolyte.

This layer absorbs the effects of movement of the electrode in relationship to the skin, and attempts to maintain the polarization associated with the half-cell potential constant.

Since perspiration is the most common cause of electrode displacement, the use of an additional porous overlay disc resists perspiration and ensures secure placement of the electrode on the skin even under stress conditions. Figure 2.16 show a typical pregelled electrode.

PASTELESS ELECTRODES

ECG monitoring electrodes, in a majority of the cases, are metal plates applied to the skin after cleaning and application of a coupling-electrolyte in the form of an electrode paste or jelly.

Such preliminary preparation can be sometimes irritating and time consuming. Also, it is often not done satisfactorily, resulting in problems like poor quality signals and baseline drift, etc. Another disadvantage of using electrode jelly is that during long-term monitoring there is likely to be patient-skin reactions as the electrode-skin interface dries out in a matter of a few hours.

The electrodes need to be periodically removed for jelly replenishments, thus causing further discomfort due to repetitive skin preparation. In addition, bacterial and fungal growth can take place under electrodes worn over long periods. Also, in conductive electrodes, shifts in electrode position at the electrode-skin interface appear as baseline drift, particularly when the subject moves.

Therefore, any attempt of using a dry electrode that may dispense with the practice of skin preparation would look attractive.

Capacitive Electrodes: A metal plate electrode in direct contact with the skin though makes a very high resistive contact and has a considerable capacitive contact too with the skin.

By using a very high input impedance amplifier, it is possible to record a signal through the tissue electrode capacitance. Lopez and Richardson (1969) describe the construction of electrodes which can be capacitively coupled to the subject.

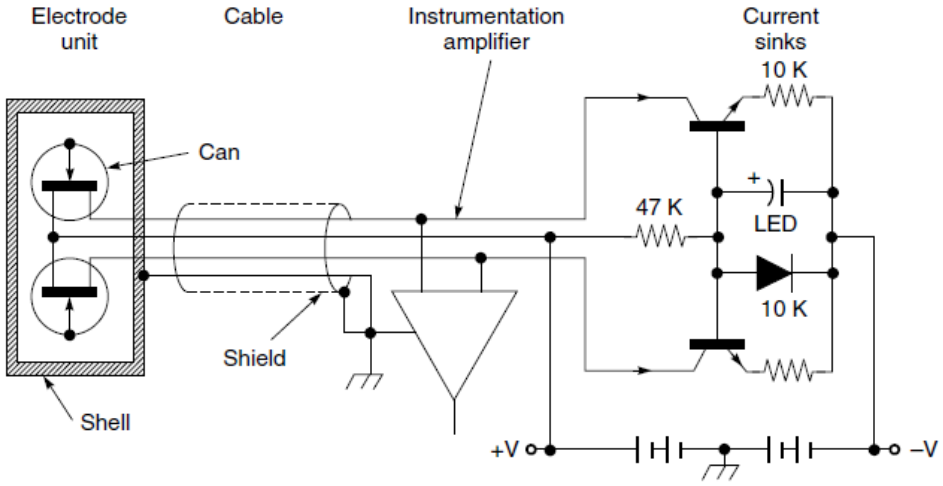
The electrode consists of an aluminium plate which is anodized on the surface to be placed in contact with the skin. The ohmic resistance of the anodized electrode is about 1 to 30 GW (1000–30,000 MW). Two such electrodes are applied to the subject.

Luca *et al* (1979) designed an electrode and amplifier as an integrated unit, so that the assembly could be used in the front end of the commonly used biomedical recorders. The

arrangement (Fig. 2.18) basically comprises a metal shell which performs a dual function as a housing for the electrode and as the ground contact.

The shell is made of highly pure titanium metal measuring 30 \ 1 5 \ 7 mm. Two FETs are cemented with epoxy glue in the middle of the shell, their centers spaced 10 mm apart. The recording surfaces are formed by the cases of the two FETs.

The cans have a diameter of 4.5 mm and are made of stainless steel. The rectangular border of the shell acts as the ground contact and the remainder of the shell forms a shield against interfering radiation.

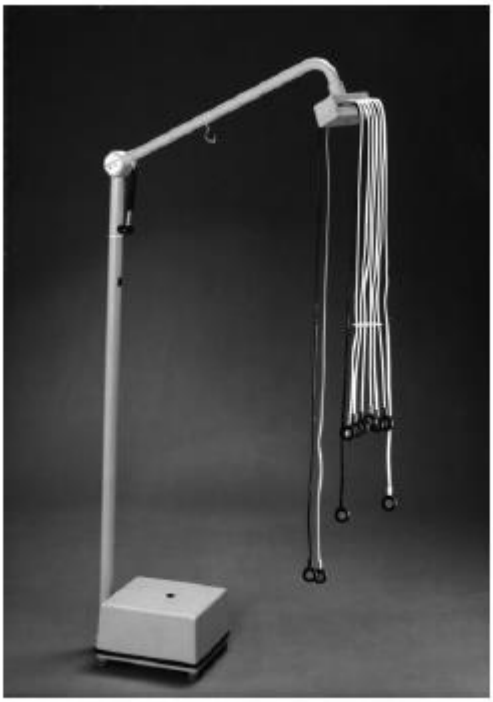


➤ **Fig. 2.18** *Schematic diagram of integrated electrode and amplifier arrangement for pasteless operation (after Luca et al, 1979; reproduced by permission of Med. and Biol. Eng. and Comp.)*

The source leads of the two FETs are connected to the differential inputs of instrumentation Amplifier. The amplifier (Analog Devices 521) has a high ac input impedance (> 100 MW).

Air-Jet ECG Electrodes*:* Wohnhas (1991) describes a novel air-jet electrode which employs Bernoulli technology to achieve constant and secure electrode contact resulting in quality tracings while minimizing artefact and maximizing baseline stability.

Air-jet electrodes (Fig. 2.19) are Ag-AgCl electrodes encased within a contoured medical silicon Cup bounded by a skin-engaging rim. The contact area (pill) is anchored to a layer of synthetic, sintered carbon by a titanium screw. The miniature silver venturi air-jet bisects the sintered layer of synthetic carbon.



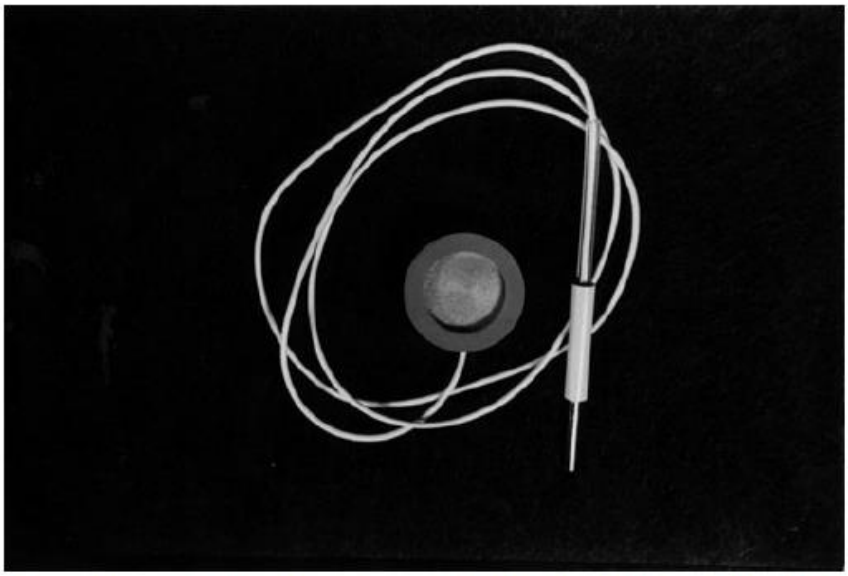
➤ **Fig. 2.19** *Air-jet electrodes (Courtesy: M/s Medi-Globe, USA)*

2.11 ELECTRODES FOR EEG

Among the most commonly used electrodes for EEG (electroencephalogram) recording are the chlorided silver discs (Fig. 2.20) having approximately 6–8 mm diameters. Contact with the scalp is made via an electrolytic paste through a washer of soft felt.

They have ac resistance varying from 3–20 kW. Small needle electrodes are sometimes used for carrying out special EEG studies when they are inserted subcutaneously. Silver ball or pellet electrodes covered with a small cloth pad are useful when electrical activity is to be recorded from the exposed cortex, but they have high dc resistances.

Hector (1968) describes a pad electrode (Fig. 2.21(a)) which is made from a silver rod belled out at the end and padded with a sponge, or a similar material, contained in gauze. It is screwed into an insulated mount and held in place on the head with a rubber cap.



➤ **Fig. 2.20** *EEG electrode which can be applied to the surface of the skin by an adhesive tape (Courtesy: In Vivo Metrics, USA)*

To hold three such electrodes, an adjustable tripod mount is employed. Another type of EEG electrode consists of multiple fine chlorided silver wires (Fig. 2.21(b)) fixed in a rigid plastic cup.

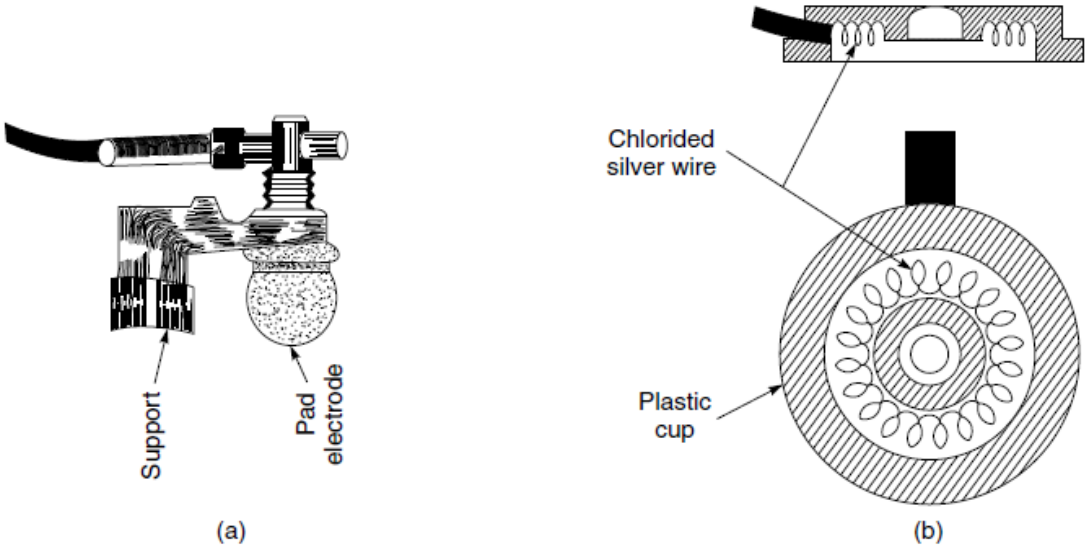
The plastic cup is fixed to the scalp with an adhesive. It is filled with jelly through a hole in the top. In this electrode, contact with the tissue is made via an electrolyte bridge so that jelly in contact with the electrode metal is not disturbed by scalp movement.

To avoid metal junctions which may get corroded with electrolyte, the silver wires are used as the output lead. The large surface area and excess of silver chloride favour stability.

2.12 ELECTRODES OF EMG

Electrodes for electromyographic work are usually of the needle type (Fig. 2.22(a)). Needle electrodes are used in clinical electromyography, neurography and other electrophysiological investigations of the muscle tissues underneath the skin and in the deeper tissues.

The material of the needle electrode is generally stainless steel. In spite of the fact that stainless steel is unfavorable electrode material from the point of view of noise, it is preferred in EMG work due to its mechanical solidity and low price.



➤ **Fig. 2.21** (a) *Pad electrode for recording EEG (after Hector, 1968)*
(b) *EEG electrode consisting of chlorided silver wire in plastic cup. Jelly is inserted from the hole kept for the purpose*

Needle electrodes are designed to be fully autoclavable and in any case they should be thoroughly sterilized before use. Needle electrodes come in various forms. The monopolar needle electrode usually consists of a Teflon coated stainless steel wire which is bare only at the tip. It is found that after the needle has been used a number of times, the Teflon coating will recede, increasing the tip area.

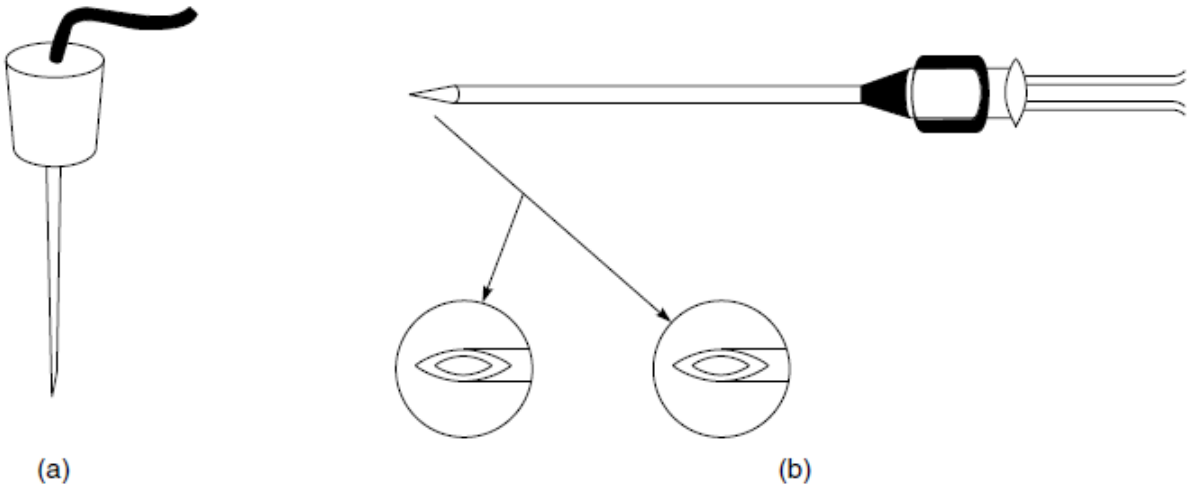
The needle must be discarded when this occurs. Bipolar (double coaxial) needle electrodes contain two insulated wires within a metal cannula. The two wires are bared at the tip and provide the contacts to the patient.

The cannula acts as the ground. Bipolar electrodes are electrically symmetrical and have no polarity sense. A concentric (coaxial) core needle electrode contains both the active and reference electrode within the same structure.

It consists of an insulated wire contained within a hypodermic needle (Fig. 2.22(b)). The inner wire is exposed at the tip and this forms one electrode. The concentric needle is very convenient to use and has very stable electrical characteristics.

Care should be taken to maintain the surface electrode in good condition in order to avoid artefacts. Concentric needle electrodes are made by moulding a fine platinum wire into a hypodermic needle having an outside diameter less than 0.6 mm.

One end of the needle is bevelled to expose the end of the wire and to provide easy penetration when the needle is inserted. The surface area of the exposed tip of the wire may be less than 0.0005 mm^2



➤ **Fig. 2.22** (a) *Needle type EMG electrode*
(b) *Hypodermic needle type EMG electrode*

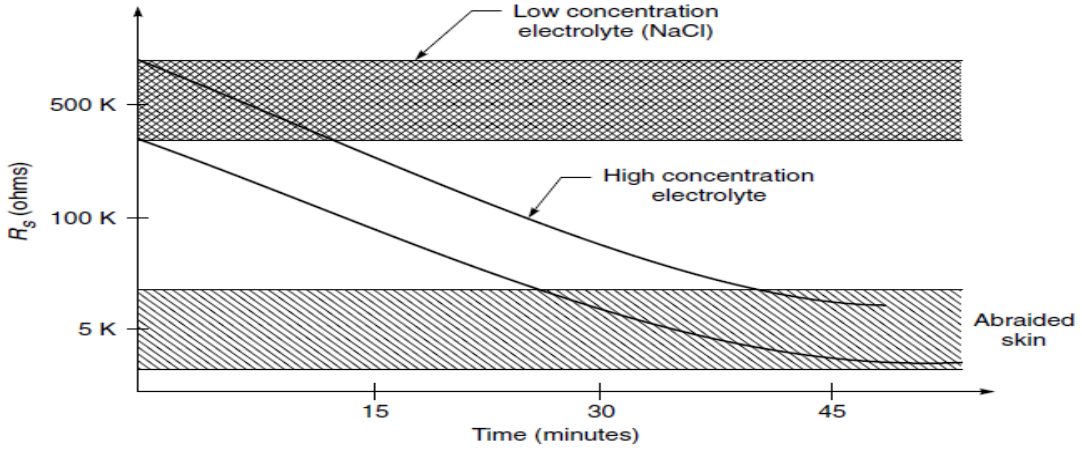
2.13 ELECTRICAL CONDUCTIVITY OF ELECTRODE JELLIES AND CREAMS

Conducting creams and jellies have for long been used to facilitate a more intimate contact between the subject's skin and the recording electrodes. The outer horny layer of the skin is responsible for the bulk of the skin contact impedance, and for this reason careful skin preparation is essential in order to obtain the best results. The recording site should first be cleaned with an ether-meth mixture.

In addition to having good electrical conductivity, the electrode jelly must have a particular chloride ion concentration (about 1%) close to the physiological chloride concentration.

This is primarily important for long-term monitoring because it should not produce a harmful diffusion between the jelly and the body. It is to be particularly ensured that the jelly chosen is of a bland nature and does not contain soap or phenol which can produce a marked irritation of the skin after a few hours.

The electrical conductivity of different makes of electrode cream can be measured (Hill and Khandpur, 1969) by means of the Schering ac bridge circuit. The cream is placed in a Perspex conductivity cell of known dimensions and the resistive component of the cell impedance is measured at 10 Hz, the conductivity being calculated from the cell dimensions.



➤ **Fig. 2.24** *Variation of contact impedance with electrolyte concentration and time (re-drawn after Trimby, 1976; Courtesy: Hewlett Packard, USA)*

The contact impedance of the skin depends upon the type of electrolyte used and the time (Trimby, 1976). Figure 2.24 shows the effect of these parameters. A low concentration sodium chloride electrolyte has 0.5% NaCI and a high concentration electrolyte has a

concentration in the range of 5 to 10% NaCI. The impedance is found to fall rapidly to 40% between 7 to 30 min.

Stabilization occurs at about 30 to 45 min. An interesting observation from this figure is that while pre-rubbing the skin will lower the initial impedance value, the final value after using a high concentration electrolyte becomes nearly the same.

Electrode jelly can be replaced in certain cases by using a conducting plastic as an interface between the electrode and the surface of the body.

2.14 MICROELECTRODES

To study the electrical activity of individual cells, microelectrodes are employed. This type of electrode is small enough with respect to the size of the cell in which it is inserted so that penetration by the electrode does not damage the cell.

The size of an intracellular microelectrode is dictated by the size of the cell and the ability of its enveloping membrane to tolerate penetration by the microelectrode tip. Single-living cells are rarely larger than 0.5 mm (500 microns) and are usually less than one-tenth of this size.

Typical microelectrodes have tip dimensions ranging from 0.5 to 5 microns. The tips of these electrodes have to be sufficiently strong to be introduced through layers of tissues without breaking.

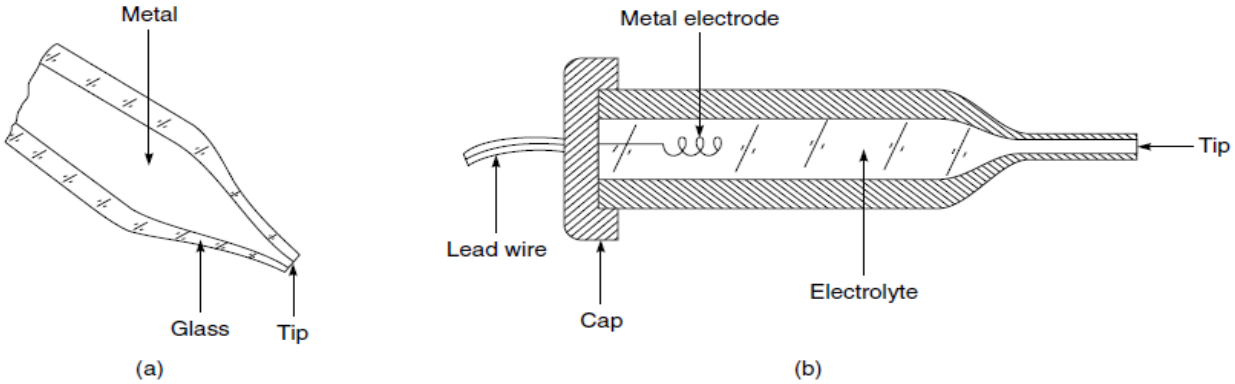
Two types of microelectrodes are generally used: metallic (Fig. 2.25(a)) and glass microcapillaries (Fig. 2.25(b)). Metallic electrodes are formed from a fine needle of a suitable metal drawn to a fine tip. On the other hand, glass electrodes are drawn from Pyrex glass of special grade.

These microcapillaries are usually filled with an electrolyte. The metal microelectrodes are used in direct contact with the biological tissue and, therefore, have a lower resistance. However, they polarize with smaller amplifier input currents.

Hence, they tend to develop unstable electrode offset potentials and are therefore not preferred for steady state potential measurements. On the other hand, in case of glass microelectrodes, improved stability can be obtained by properly choosing the metal and the electrolyte so that the small current passing through their junction may not be able to modify the electrical properties of the electrodes.

Also, the glass microelectrode has a substantial current carrying capacity because of the large surface contact area between the metal and the electrolyte.

The microelectrodes have a very high impedance as compared to conventional electrodes used for recording ECG, EEG, etc. The high impedance of a metal microelectrode is due to the characteristics of the small area metal-electrolyte interface.



➤ **Fig. 2.25** (a) *Microelectrodes—metal microelectrodes*
(b) *Microelectrodes—micropipette or micro capillaries electrode*

Similarly, a micropipet tip is filled with an electrolyte which substitutes an electrolytic conductor of small cross-sectional area, which gives a micropipet its high resistance. Because of high impedance of microelectrodes, amplifiers with extremely high input impedances are required to avoid loading the circuit and to minimize the effects of small changes in interface impedance.

Glass Capillary Electrode

Several methods exist for producing microelectrodes of wide variety and shapes. For drawing electrodes of uniform and accurate diameter, it is essential to maintain constant timing, temperature, strength and direction of pull. These factors are difficult to control when the electrodes are drawn manually.

The mechanical method employs gravitational force for extension and the electrodes which are drawn in one or more stages can readily produce capillary tubes between 3−30 mm diameter, but great difficulty is encountered in producing electrodes of less than 1 mm.

The most commonly used method for making small tip micropipet consists of the circumferential application of heat to a small area of glass tubing which is placed under some

initial tension. When the glass softens, the tension is increased very rapidly and the heat is turned off.

Proper timing, controlled adjustment of the amount of heat as well as the initial and final tensions and cooling result in the production of microcapillaries with controlled dimensions.

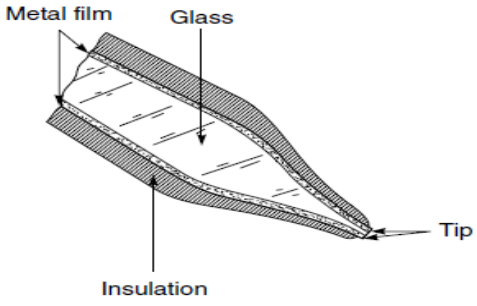
Metal Microelectrode

Metal electrodes with very fine tips used for recording from single cells have the advantage over glass micropipetes of being relatively robust. Steel microelectrodes can be made from ordinary darning needles but preferably they should be of good stainless steel wire.

They can be easily made up to 10 mm diameter but great care has to be taken for diameters as small as 1 µm. Thesevery small tips are not very satisfactory as they are extremely brittle and have very high input impedance.

Hubel (1957) described a method to make tungsten microelectrodes with a tip diameter of 0.4 µm. He used electropointing technique which consists in etching a metal rod while the metal rod is slowly withdrawn from the etching solution, thus forming a tapered tip on the end of the rod. The etched metal is then dipped into an insulating solution for placing insulation on all but the tip.

Figure 2.26 shows the cross-section of a metal microelectrode. In this electrode, a thin film of Precious metal is bonded to the outside of a drawn glass microelectrode. This arrangement offers lower impedance than the microcapillary electrode, infinite shelf life and reproduciable performance, with ease of cleaning and maintenance. The metal—electrolyte interface is between the metal film and the electrolyte of the cell.



➤ **Fig. 2.26** *Microelectrode—thin metal film coated micropipette*