A Basic Biomedical Recording System

Recorders provide permanent visual trace or record of an applied electrical signal. The most basic electronic recording system consists of 3 important components, see Figure 1:

1. Electrodes/Transducer
2. Signal conditioner
3. Writing system

The **electrode** picks up the bioelectric potentials whereas the transducer converts the physiological signal to be measured into a usable electrical output.
The **signal conditioner** converts the output of the electrode/transducer into an electrical quantity suitable for operating the writing system.
The **writing system** provides a visible graphic representation of the quantity of the physiological variable of interest.

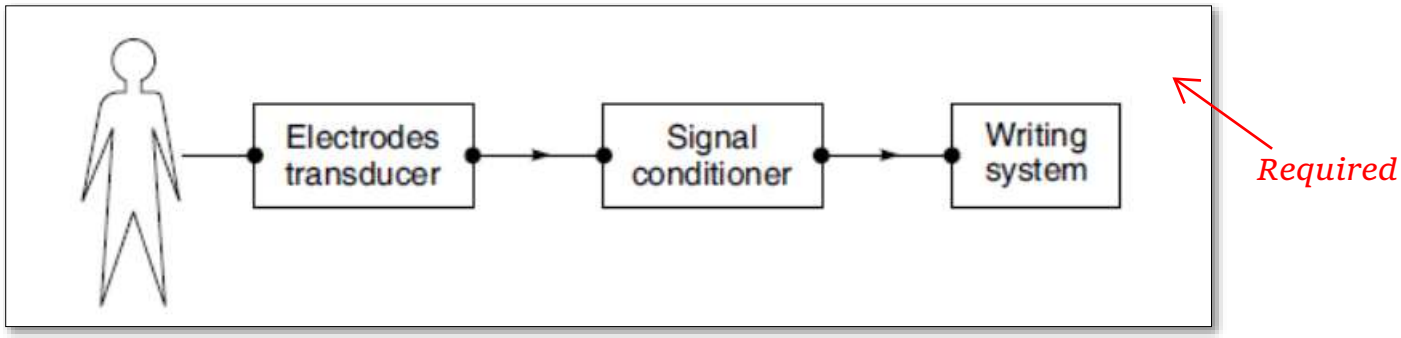

Figure 1 A Basic Electronic Recording System

In medical recorders, the signal conditioner usually consists of a preamplifier and the main amplifier. Both of these amplifiers must satisfy specific operation requirements such as input impedance, gain and frequency response characteristics for an effective reproduction of the input signal. To make the signal from any transducer compatible with the input signal required for the driver amplifier of the display or recording system, it is usual to arrange to normalize the electrical signals produced by each transducer. This is done in the signal conditioner which adjusts its output to a common signal level e.g. one volt. The necessary adjustments of gain and frequency response are provided by the signal conditioners. This means that, it is possible to interchange the signal conditions to record any one of the physical or bioelectric events on the same writing channel. The writing systems which are available in many forms constitute the key portion of the recording instrument. Examples of commonly used writing system include: The Inkjet recorder, Galvanometer type pen recorder, Potentiometric recorder, etc.

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(a, b) represent the electrode-tissue interface. 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.

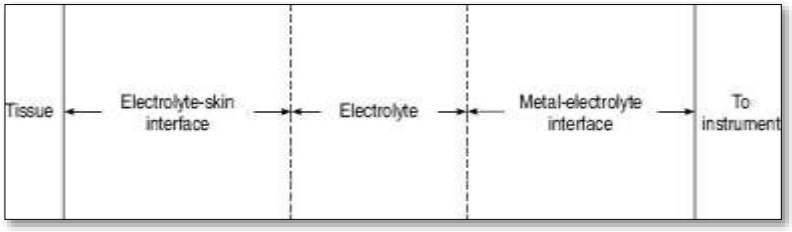

Figure 2 (a) Electrode-tissue interface for surface electrodes used with electrode jelly

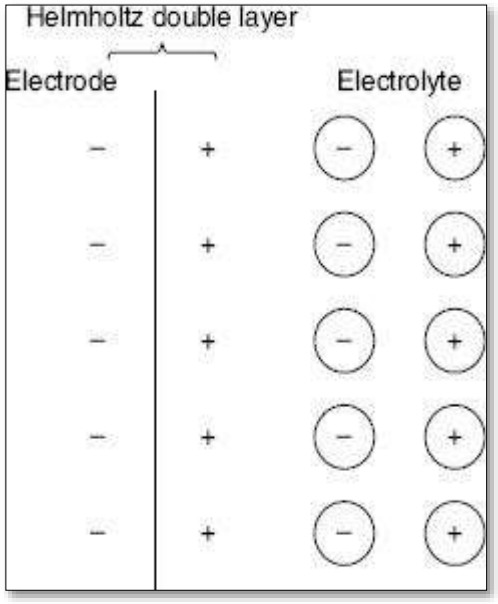

Figure 2 (b) The electrode tissue interface circuit involves transfer of electrons from the metal phase to an ionic carrier in the electrolyte, a charge double layer (capacitance) forms at the interface>

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 [Figure 3]. 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*.

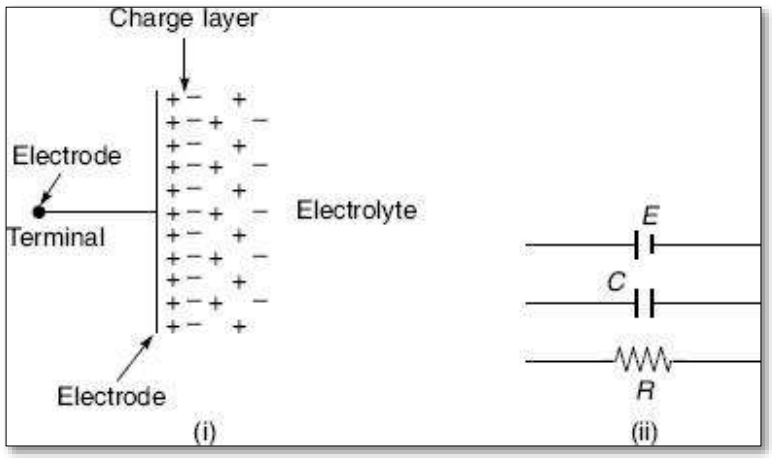

Figure 3 (i) Charge distribution at electrode-electrolyte interface (ii) three components representing the interface

To a first-order approximation, the half-cell potential is equal to the electrode potential of the metal, if the electrodes were used in a chemical measuring application. All electrode potentials are measured with respect to a reference electrode, usually that of hydrogen absorbed on platinum black. This is an inconvenient electrode to make and, therefore, other alternative electrodes which may have fairly stable and repeatable potential (e.g. calomel electrode) are employed. Electrode potentials of some of the commonly used metals in the electrochemical series with respect to hydrogen are given in Table 1.

Table 1 Electrode Potentials of Some Metals with Respect to Hydrogen

Metal	Ionic symbol	Electrode potential
Aluminium	$_{Al}+++$	–1.66 V
Iron	Fe^{++}	–0.44 V
Lead	Pb^{++}	–0.12 V
Hydrogen	H^{+}	0
Copper	$_{C}++$	+0.34 V
Silver	Ag^{+}	+0.80 V
Platinum	Pt^{+}	+1.2 V
Gold	Au^{+}	+1.69 V

Table 1 shows that the electrode potentials are appreciable when dissimilar metals are used. They also exist, though of smaller magnitude, even if electrodes of similar materials are employed. The lowest potential has been observed to be in the silver-silver chloride electrodes. The values of the capacitance and the resistance depend upon many factors which include the current density, temperature, type and concentration of the electrolyte and the type of metal used.

The difference in half-cell potentials that exists between two electrodes is also called *offset potential'*. The differential amplifiers used to measure potentials between two electrodes are generally designed to cancel the electrode offset potential so that only the signals of interest are recorded. The electrode offset potential produced between electrodes may be unstable and unpredictable. The long-term change in this potential appears as baseline drift and short-term changes as noise on the recorded trace. If electrodes are used with ac-coupled amplifiers, the long term drift may be partially rejected by the low frequency characteristics of the amplifier. But it will depend upon the rate of change of electrode offset potential in relation to the ac-coupling time constant in the amplifier. For example, if the electrode offset potential drift rate is 1 mV/s, satisfactory results can only be obtained if the low frequency response of the amplifier is 1 Hz.

Also, the absolute value of the electrode offset potential is rarely significant except when it may exceed the maximum dc differential offset of the amplifier. In such a case, the trace may go out of the monitor screen or the pen in a recording instrument shifts to the extreme end of the chart paper, and then it will not be possible to bring them back. Silver-silver chloride electrodes have been found to give almost noise free characteristics. They are also found to be acceptable from the point of view of long-term drift. Electrodes made of stainless steel are generally not acceptable for high sensitivity physiological recordings. This is because stainless steel electrodes in contact with a saline electrolyte produce a potential difference of 10 mV between the electrodes, whereas this value is 2.5 mV for silver-silver chloride electrodes. Some representative values of potential between electrodes in electrolytes are given in Table 2.3. Staewen (1982) discusses various aspects concerning dc offset voltage standard for pregelled ECG disposable electrode.

Table 2 Potential between Electrodes in Electrolytes (Geddes and Baker, 1975)

Electrode metal	Electrolyte	Potential difference between electrodes
Stainless steel	Saline	10 mV
Silver	Saline	94 mV
Silver-silver chloride	Saline	2.5 mV
Silver-silver chloride(11 mm disc)	ECG paste	0.47 mV
Silver-silver chloride (sponge)	ECG paste	0.2 mV

Warburg (1899) in his pioneering studies discovered that a single electrode/electrolyte interface can be represented by a series capacitance *C* and resistance *R* as shown in Fig. 2.8(a). However, *C* and *R* are unlike real capacitors and resistors because their values are frequency and current density dependent. Often, these components are called the polarization capacitance and resistance. Warburg found that, for low current density, the reactance *X* of *C* (1/2 πf*C*) equals *R*; both varied almost inversely as

$$R = X = k/\sqrt{f}$$

the square root of frequency, i.e., where *k* is a constant. The consequence of this relationship is that the phase angle θ is constant at π/4 for all frequencies. However, only a limited number of studies have tested the accuracy of the Warburg model (Ragheb and Geddes, 1990).

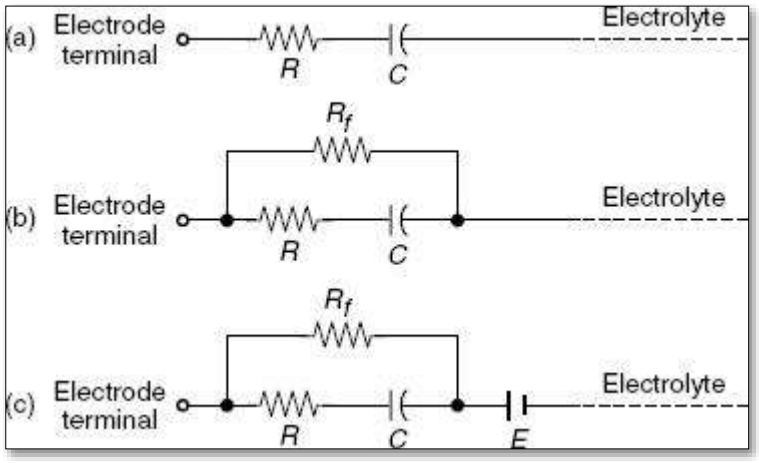

Figure 4 (a)Warburg equivalent for an electrode-electrolyte interface (b) Addition of the Faradic leakage resistance Rf to account for the direct current properties (c) Half-cell potential E of the electrode electrolyte interface

It has been observed that the Warburg series *RC* equivalent does not adequately represent the behaviour of an electrode/electrolyte interface as this equivalent does not truly account for the very low-frequency behaviour of the interface. It is well known that such an interface can pass direct current. Therefore, a resistance R_f placed in parallel with the Warburg equivalent is more appropriate. Figure 4, shows this equivalent circuit in which Rf represents the Faradic leakage resistance. The value of Rf is high in the low-frequency region and is dependent on current density, increasing with a increase in current density.

To complete the equivalent circuit of an electrode/electrolyte interface, it is necessary to add the half-cell potentialE. This is the potential developed at the electrode/electrolyte interface. The value of E depends on the species of metal and the type of electrolyte, its concentration and temperature. Figure 4 illustrates the complete equivalent circuit of a single electrode/electrolyte 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.

When bioelectric events are recorded, interference signals are produced by the potential differences of metal-electrolyte and the electrolyte-skin interface. Normally, these potential differences are connected in opposition during the recording procedure, and in the case of a truly reversible and uniform electrode pair, their difference would be nil. However, in practice, a difference of potential—may be extremely small—is found to exist between electrodes produced even under conditions of utmost care during manufacture. Also, some of the elements in the equivalent circuit are time-dependent and are bound to show slow variations with time.

The main reason for this rate of change is due to a relative displacement affecting chiefly the potential of the metal-electrolyte transition. Other factors responsible for variations of potential difference with time can possibly be temperature variations, relative displacement of the components in the system and changes in the electrolyte concentration, etc. (Odman and Oberg, 1982). If ac signals are to be recorded, the potential difference between the two electrodes will not interfere with the useful signals, provided that the contact potential difference between the electrodes is constant. However, if the rate of change with time of the contact potential falls within the frequency spectrum of the signal under test, an error will be produced. The problem of difference of contact potentials becomes serious in case dc signals such as EOG are to be recorded. Any variation in the contact potential would greatly alter the character of the signal to be recorded which may itself be of extremely low amplitude— of the order of a few microvolts.

Based on the above mentioned considerations, it is possible to construct the circuit in which a pair of electrodes is placed in electrolytic contact with a subject. 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 (Figure. 5), one is due to the bioelectric event (E_b) and the other two are non-physiologic and 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_t 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.

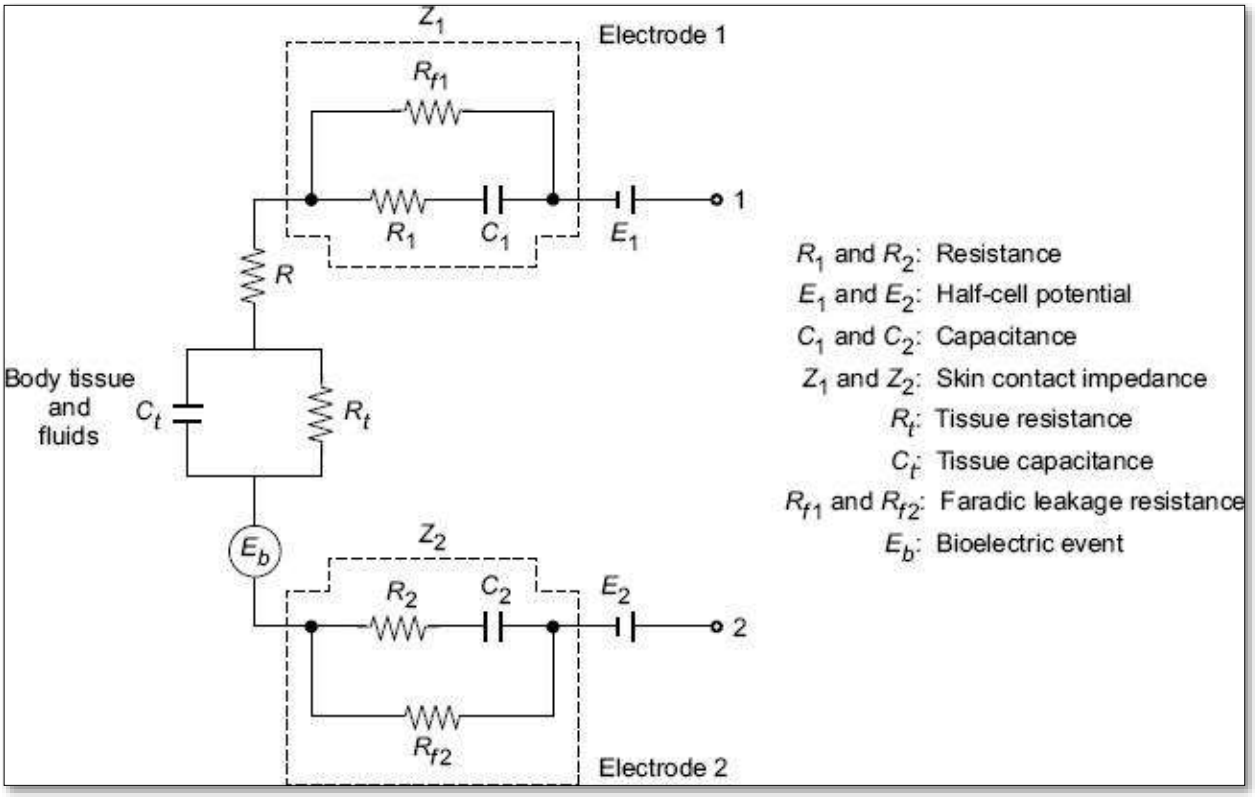

Figure 5 Equivalent circuit for a pair of electrodes (1, 2) on a subject represented by R, R t, Ct. Embedded in the subject is a bioelectric generator Eb (after Tacker & Geddes, 1996)

Polarization

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.

Although polarizable electrodes are becoming less common, they are still in use. They usually employ stainless steel and are used for resting ECGs or other situations where there is small likelihood that the electrodes would be exposed to a large pulse of energy (such as a defibrillation discharge) in which case they would retain a residual charge, become polarized, and will no longer transmit the relatively small bioelectric signals, thus becoming useless.

Non-polarizing electrodes on the other hand, are designed to rapidly dissipate any charge imbalance induced by powerful electrical discharges such as a defibrillation procedure. Rapid depolarization enables the immediate reappearance of bioelectric signals on the monitor after defibrillation. For this reason, non-polarizing electrodes have become the electrodes of choice for monitoring in the intensive care units and stress testing procedures. Historically, these electrodes employ a conducting metal with a silver/silver-chloride (Ag/AgCl) surface in contact with the conducting gel. The choice of metals for electrodes is not determined only by their susceptibility to polarization, but other factors such as mechanical properties, skin irritation or skin staining, etc. have also to be taken into consideration. A detailed comprehensive review of electrodes for measurement of bioelectronic events is given by Geddes and Baker (1975).

The bioelectrical events are usually recorded by means of metallic electrodes placed on the surface of the body. The electrical activity generated by various muscles and nerves within the body is conducted to the electrode sites through the body tissues, reaches the electrodes through the skin electrode transition and is then conducted by direct wire connection to the input circuit of the recording machine. The impedance at the electrode-skin junction comes in the overall circuitry of the recording machine and, therefore, has significant effect on the final record. Skin electrode impedance is known as the contact impedance and is of a value much greater than the electrical impedance of the body tissue as measured beneath the skin. The outer horny layer of the skin is responsible for the bulk of the skin contact impedance and, therefore, a careful skin preparation is essential in order to obtain best results.

Skin Contact Impedance

Measurement of Skin Contact Impedance:

A convenient method to measure the contact impedance at any individual electrode is shown in Figure 6, This method has been suggested by Miller (1969). The three electrodes, A, B and C, have contact impedance respectively of Z_a, Z_b and Z_c. An oscillator provides a constant current in the frequency range of 0.1–100 Hz through the 47 kΩ series resistor. By suitably positioning the switch, a sensitive oscilloscope can be used to monitor either the voltage dropped across the 1 kΩ resistor or the voltage dropped across Z_b. The voltage drop across Z_b 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 kΩ resistor it is possible to calculate the circuit current and thus to obtain a value for Z_c. Using this technique, the skin contact impedance of the following types of electrodes were measured by Hill and Khandpur (1969).

A. Plastic cup self-adhesive electrodes (Boter *et al.*, 1966)

B. Metal plate limb electrodes used with conducting jelly

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

D. Dry multi-point limb electrodes (Lewes, 1966)

E. Dry multi-point suction chest electrodes

F. Self-adhesive multi-point chest electrodes used with conducting jelly

G. Self-adhesive gauze electrodes

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

Representative plots of contact impedance versus frequency are shown in Figure7.

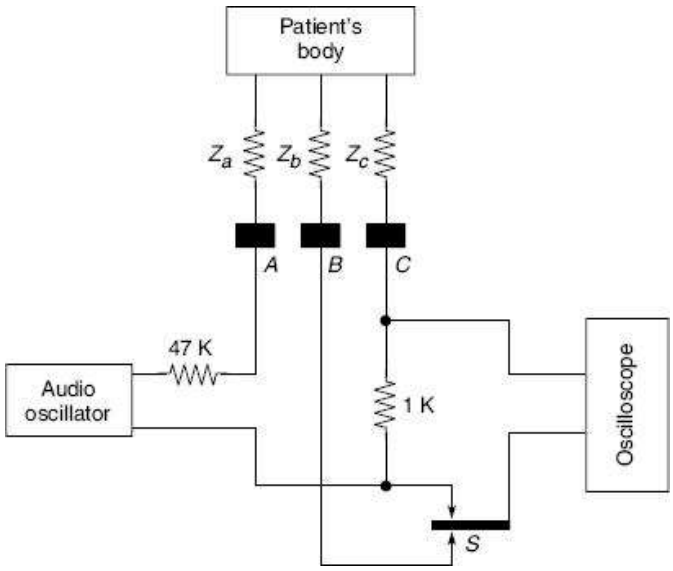

Figure 6 Arrangement for measurement of electrode skin-contact impedance for surface electrodes

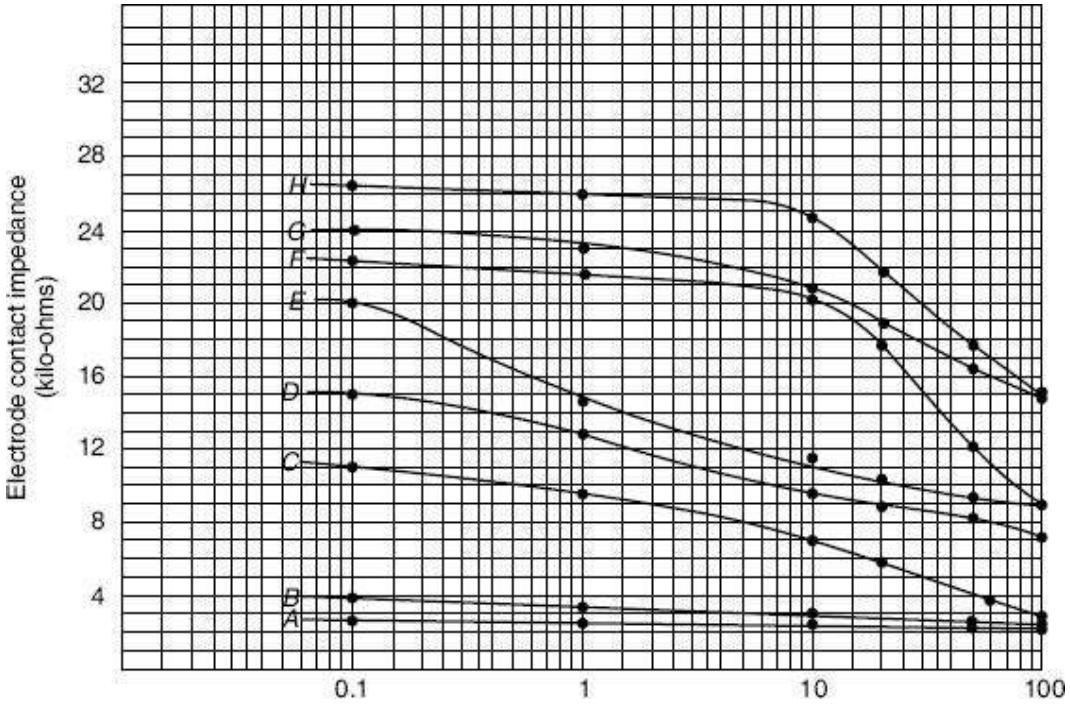

Figure 11 Electrode skin-contact impedance versus signal frequency for different types of electrodes (after Hill and Khandpur, 1969)

Usually the contact impedance in respect of surface electrodes used for recording of ECG is measured (Grimnes, 1983) at 10– 20 Hz because most of the energy content of the ECG is concentrated below 30 Hz. Geddes and Baker (1968) used a synchronous rectifier with a phase sensitive detector to continuously measure the resistive and reactive components of the impedance.

Motion Artefacts

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 longer 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. However, silver-silver chloride electrodes do not cause much problem since silver chloride is almost insoluble in a chloride containing solution. Therefore, when these electrodes are used, the skin irritation is mostly caused by the paste and or the adhesive. Most commercial pastes produce about the same irritation when used on unprepared skin. They cause itching due to restricted perspiration, and reddening of the skin directly under the electrodes appears in 2–4 days. Thakor and Webster (1985) studied the sources of artefacts, means of reducing them using skin preparations, the electrode designs and their placement on the chest for long-term ambulatory ECG.

SILVER-SILVER CHLORIDE ELECTRODES

One of the important desirable characteristics of the electrodes designed to pick up signals from biological objects is that they should not polarize. This means that the electrode potential must not vary considerably even when current is passed through them. Electrodes made of silver-silver chloride have been found to yield acceptable standards of performance. By properly preparing and selecting the electrodes, pairs have been produced with potential differences between them of only fractions of a millivolt (Feder, 1963). Standing voltage of not more than 0.1 mV with a drift over 30 min. of about 0.5 mV was achieved in properly selected silver-silver chloride electrodes by Venables and Sayer (1963). Silver-silver chloride electrodes are also nontoxic and are preferred over other electrodes like zinc-zinc sulphate, which also produce low offset potential characteristics, but are highly toxic to exposed tissues. Silver-silver chloride electrodes meet the demands of medical practice with their highly reproducible parameters and superior properties with regard to long-term stability.

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 chloride 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.

$$2Na^+ + 2H_20 + 2\,electrons \rightarrow 2NaOH + H_2$$

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.

Geddes *et al*. (1969) investigated the effect of the chloride deposit on the impedance-frequency characteristics of the silver-silver chloride electrodes. They demonstrated that the impedance was different for different layers of chloride and that there is an optimum chloriding, which gives the lowest impedance.

They concluded that the lowest electrode-electrolyte impedance in the frequency range of 10 Hz to 10 kHz was found to occur with a chloride deposit ranging between 100 and 500 mAs/cm^2 of electrode area. To achieve this deposit by manipulation of current and time, the minimum constant chloriding current density should be 5 mA/cm^2 of electrode area.

Higher values may be used with a corresponding reduction in time to achieve the 100–500 mAs/cm^2 chloride deposit. With this chloride deposit, the electrode electrolyte impedance was found to be resistive. The use of a chloride deposit in excess of this range did not alter the resistive nature of the electrode-electrolyte impedance although it increased its magnitude.

Cole and Kishimoto (1962), however found that the chloride deposit for achieving the lowest impedance is 2000 mAs/cm^2. Geddes (1972) confirmed that an optimal coating of silver chloride applied to a silver electrode minimizes the electrical impedance. This is supported by Getzel and Webster (1976) who concluded that silver chloride may be applied to cleaned silver electrodes in the amount of 1050–1350 mA s/cm^2 in order to reduce the impedance of the electrodes. However, to further reduce the impedance of the electrodes, they should be coated with at least 2000 mAs/cm^2 of silver chloride followed by immersion in a photographic developer for 3 minutes. A second layer of silver chloride, however, did not result in any further reduction in impedance. Grubbs and Worley (1983) obtained a lower and more stable impedance electrode by placing a heavier initial chloride coat on an etched silver electrode, and then electrolytically removing a portion of that coat

Types of Bioelectrodes

Introduction

Bioelectric events have to be picked up from the surface of the body before they can be put to the amplifier for subsequent record or display. This is done by use of electrodes. The potentials produced at different points are measured by placing electrodes at various points on the body. They carry the currents produced due to potential differences to instrumentation amplifiers, where the signals are amplified and further processed by signal processing systems.
Bioelectrodes can be classified as:

- **Surface electrodes:** These electrodes pick up potentials from the surface of the tissue.
- **Deep seated electrodes:** These electrodes are inserted inside a live tissue or cell.

When a measurement is made outside the body by placing surface electrodes it is called in vitro measurement. When measurement is made by inserting a needle electrode inside the tissue, it is called in vivo measurement.

Key Properties of Bioelectrodes

Bioelectrodes should possess the following properties:

- They should be good conductors
- They should have low impedance
- They should not polarize when a current flows through them
- They should establish a good contact with the body and not cause motion
- Potentials generated at the metal electrolyte (jelly) surface should be low.
- They should not cause itching, swelling or discomfort to the patient for example the metal should not be toxic
- They should be mechanically rugged
- They should be chemically inert
- They should be easy clean

Figure 1 shows how the electrodes make contact with the skin surface

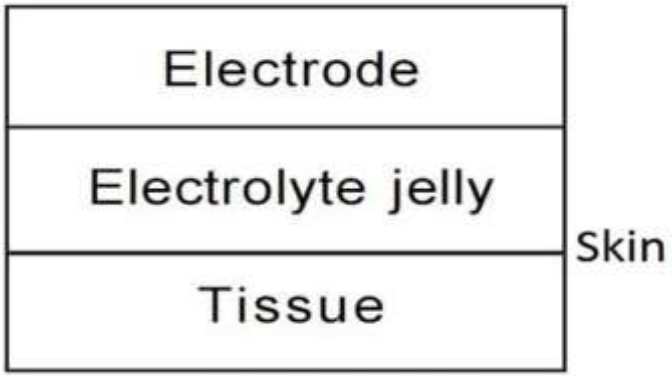

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Figure 1 Electrolyte-skin interface

Electrode Metal and Electrolyte Interface

The metal of the electrode has a tendency to discharge ions into the electrolyte. This voltage is called half-cell potential or offset potential. In certain metals and electrolytes, this potential is significant and in most metals and electrolytes it is significant, see Figure 2

Electrode metal	*Electrolyte*	*Potential difference between electrodes*
Ag	Saline	9.4 mV
Ag–AgCl	Saline	2.5 mV
Ag–AgCl	ECG paste	0.47 mV
Sponge Ag–AgCl	ECG paste	0.2 mV
Stainless Steel	Saline	10 mV

Figure 2. Potential difference between electrode and electrolyte

Materials used for Electrodes

The materials used to make Electrodes include:

- Aluminium (Al)
- Copper (Cu)
- Silver (Ag)
- Gold (Au)
- Platinum (Pt)

When we compare these metals to Hydrogen electrode, each metal has some potential as shown in the below figure:

Al^{+++}	– 1.606 V
Fe^{++}	– 0.44 V
Pb^{++}	– 0.12 V
Cu^{++}	+ 0.34 V
Ag^{+}	+ 0.8 V
Pt^{+}	+ 1.2 V
Au^{+}	+ 1.69 V

Figure 3. Hydrogen vs. other metals potentials

The Skin Contact Impedance

The outer layer of the skin has large impedance which is much greater than the electrical impedance of body tissue beneath the skin. The outer layer skin is responsible for the bulk of the skin contact impedance and therefore, a careful skin preparation is essential in order to obtain best results.

Types of Electrodes used in Biomedical Measurements

The three basic types of biopotential electrodes used in biomedical measurements are:

1. Microelectrodes
2. Skin surface electrodes
3. Needle electrodes

Microelectrodes

These electrodes are designed to measure bioelectric potentials near or within the cell. These electrodes are much smaller in cross-sectional area as compared to the size of the cell in which they are to be inserted so that their penetration should not damage the cell. A cell is rarely larger than 500 microns, therefore these microelectrodes should have tip dimension of about 5 microns and the tip should be strong enough to penetrate the cell without damage.

Microelectrodes can be of two types:

1. Metal
2. Micropipette

METAL MICROELECTRODES

Metal microelectrodes are formed from a fine needle of a suitable metal down to a fine tip. Then the needle is coated almost to the tip with an insulating material. These electrodes are used to measure the biopotential from the metal-electrolyte interface i.e. the electrode potential is developed across the metal-electrolyte interface which is proportional to the exchange of ions between the metal and the electrolyte of the body.

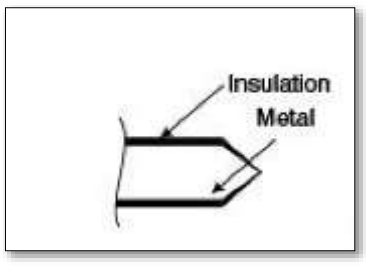
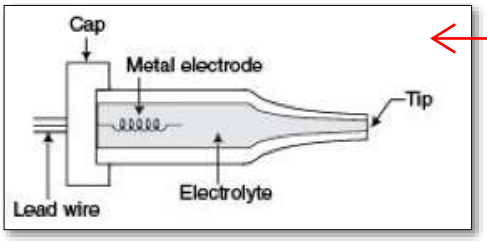


Figure 4.Metal microelectrode

MICROPIPETTE MICROELECTRODE

The **Micropipette microelectrode** is a microcapillary made of glass which is filled with an electrolyte as shown in the figure below:



Required

Figure 5 Micropipette microelectrode

The metal microelectrodes are used in direct contact with the cell and they have lower resistance. However, these electrodes tend to develop unstable electrode offset potentials. The micropipette microelectrodes have a dual interface; one interface is formed by metal wire in contact with the electrolyte solution filled in the micropipette while the other interface is formed between the electrolyte inside the micropipette and the fluids inside or immediately outside the cell.

The micropipette microelectrodes tend to develop stable electrode offset potentials and thus preferred where steady-state potentials measurements are required however metal microelectrodes have advantages too such as:

- Lower impedance
- Infinite shelf life
- Repeatable and reproducible performance
- Easy cleaning and maintenance.

Body Surface Electrodes

These electrodes are designed to measure ECG, EEG, and EMG potentials from the surface of the skin; hence these electrodes are least traumatic. Body Surface electrodes are designed and used to measure bioelectric potentials from the surface of the body. They are available in many forms and sizes. The larger electrodes are usually used for sensing of ECG potentials as these measurements do not depend on the specific localization of the electrodes. However, for sensing of EEG and EMG potentials, smaller electrodes are used as sensing for them depends upon the location of electrode or measurement. Metal plate and suction cup type electrodes are body surface electrodes but they have a common problem or the possibility of slippage or movement. These electrodes are sensitive to movements hence produce incorrect measurements on shifting. To avoid this problem, the floating electrodes are used. The principle of the floating electrode is to eliminate the movement artifacts (false signals) by avoiding any direct contact of the metal electrode with the skin. The contact between the metal electrode and skin is maintained by the electrolyte paste or jelly.

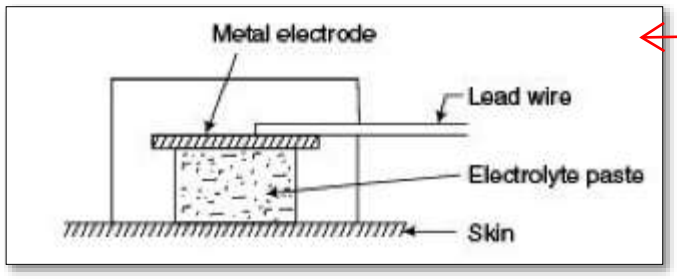


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Figure 6 Floating type body surface electrodes

TYPES OF SURFACE ELECTRODES

Examples of surface electrodes include:

1. Limb electrode
2. Disposal electrode
3. Pre-gelled electrode
4. Circular electrode.

More details on Surface electrodes are shown in the figure below, From above, Figure 7 (a), shows a rectangular electrode held together in position by elastic rubber straps. Figure (b), shows a disposal electrode with adhesive to hold the electrode in position. Figure (c) exhibits disposable pre-jelled electrodes. Figure (d) shows a circular electrode, with a rubber bulb which creates a partial vacuum so that the electrode is held in a position and no adhesive is used.

Needle Electrodes

They are generally made of stainless steel. These electrodes are designed to penetrate the skin surface of the body to some depth to record EEG potentials of a region of the brain or EMG potentials of a muscle. These electrodes have to be sharp and small like subdermal needles which help them to easily penetrate the scalp for measuring the EEG potentials. They are required to penetrate up to some surface at certain depth of the skin which is parallel to the surface of the brain or muscle.

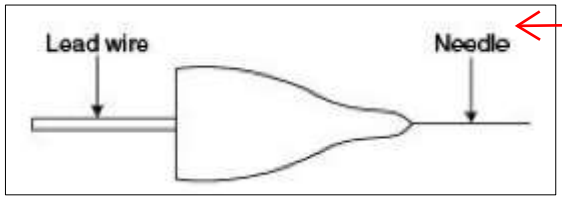

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Figure 8 Needle electrode for EEG

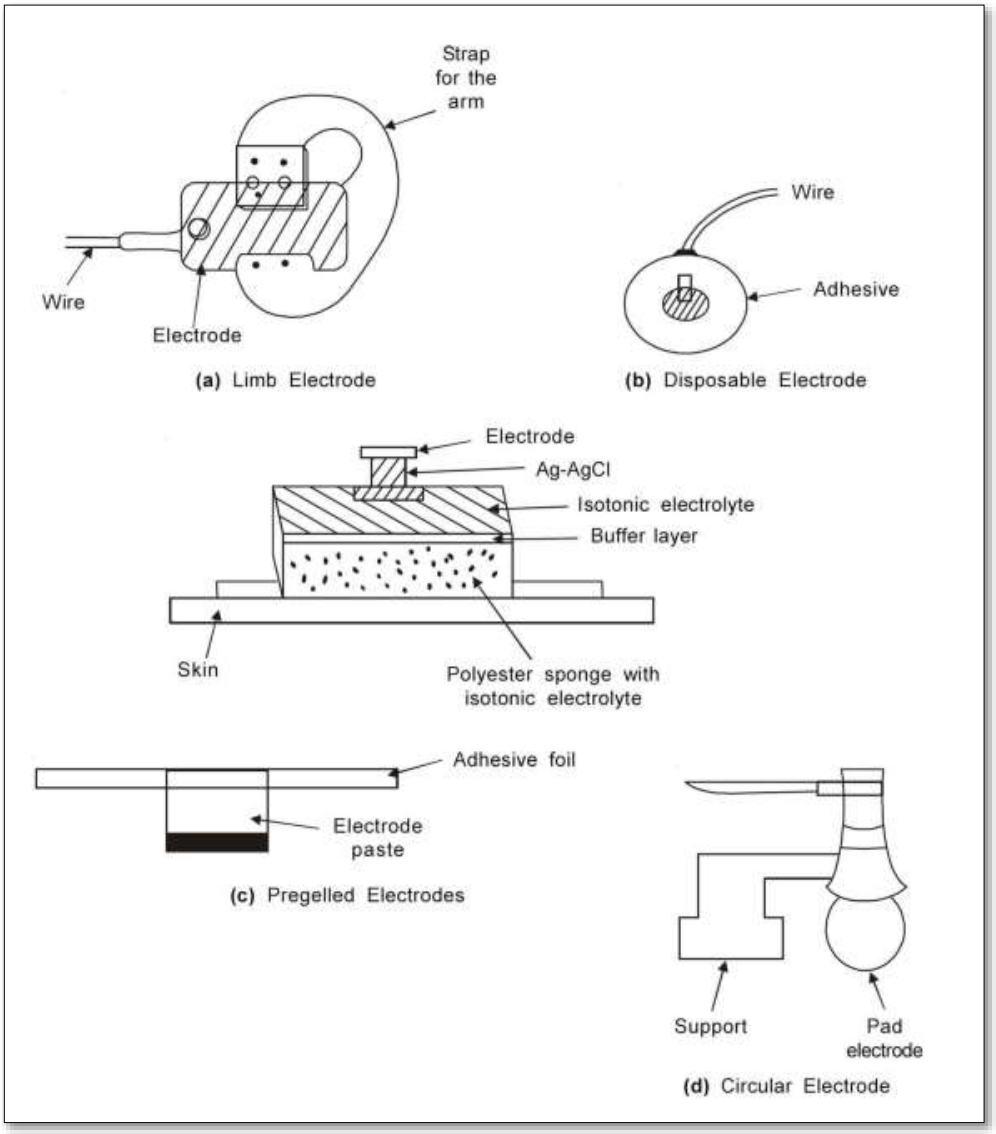

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Figure 7 Types of Surface Electrodes used in Biomedical Instrumentation.

ELECTRODES FOR ECG

Limb Electrodes

The most common type of electrodes routinely used for recording ECG are rectangular or circular surface electrodes Figure 9. 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 kΩ 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 reusable and last several years.

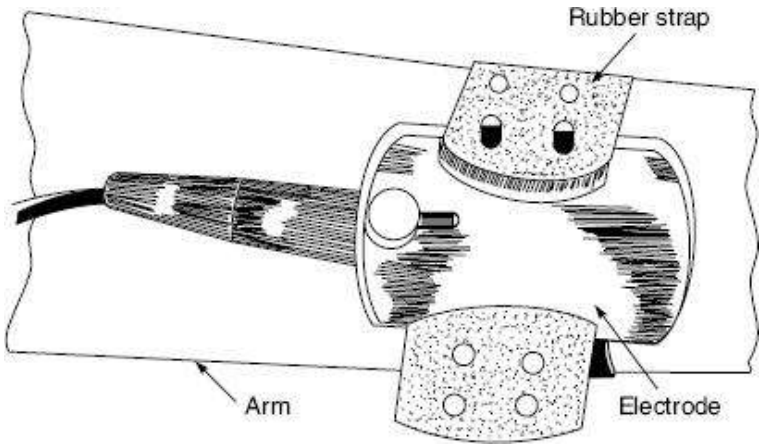

Figure 9 ECG plate electrode. The electrode is usually fastened to the arm or leg with a perforated rubber strap which keeps it in position during ECG recording

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.

Welsh Cup Electrode

Welsh Cup Electrode or suction electrode Figure 10 is a metallic cup shaped electrode which is used for recording ECG from various positions on the chest. It 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. However, they are now being gradually replaced with disposal electrodes, as they are liable to infection due to inadequate cleaning procedures.

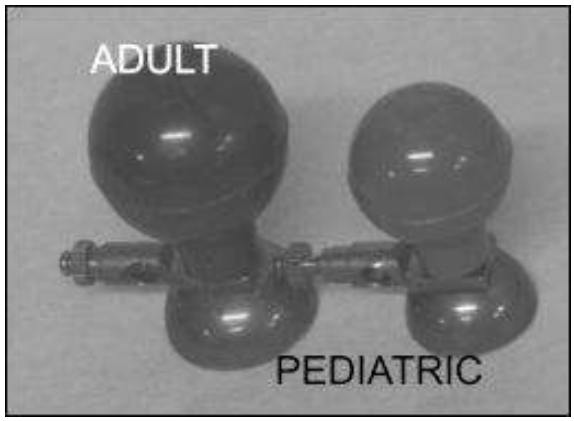

Figure 10 Suction cup electrode.

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, Kahn (1965) and Boter *et al.* (1966). 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 (Figure 11) 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. Patten *et al.* (1966) have described spray-on chest electrodes where a conducting spot is developed on the skin by spraying a film of conducting adhesive (mixture of Duco cement, silver powder and acetone). 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 kΩ. Completely flexible ECG electrodes for the long-term monitoring of ECG during space flight are reported by Sandler*et al.* (1973). These electrodes were made of silver-impregnated silastic rubber and were found to be comfortable to wear. They were also evaluated for use during exercise or prolonged monitoring as may be necessary in an intensive care or coronary care unit.

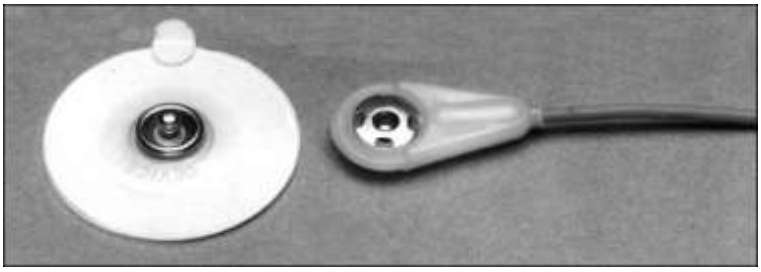

Figure 11 Light-weight floating electrode with press stud for long term monitoring of ECG.

Pregelled Disposable Electrodes

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. (a) illustrates the principle of a pre-gelled electrode while Figure. 12(b) shows a cross-section of such an electrode. The main design feature of these electrodes which that helps in reducing the possibility of artefacts, drift and baseline wandering is the provision of a high-absorbency 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 13 shows a typical pregelled electrode.

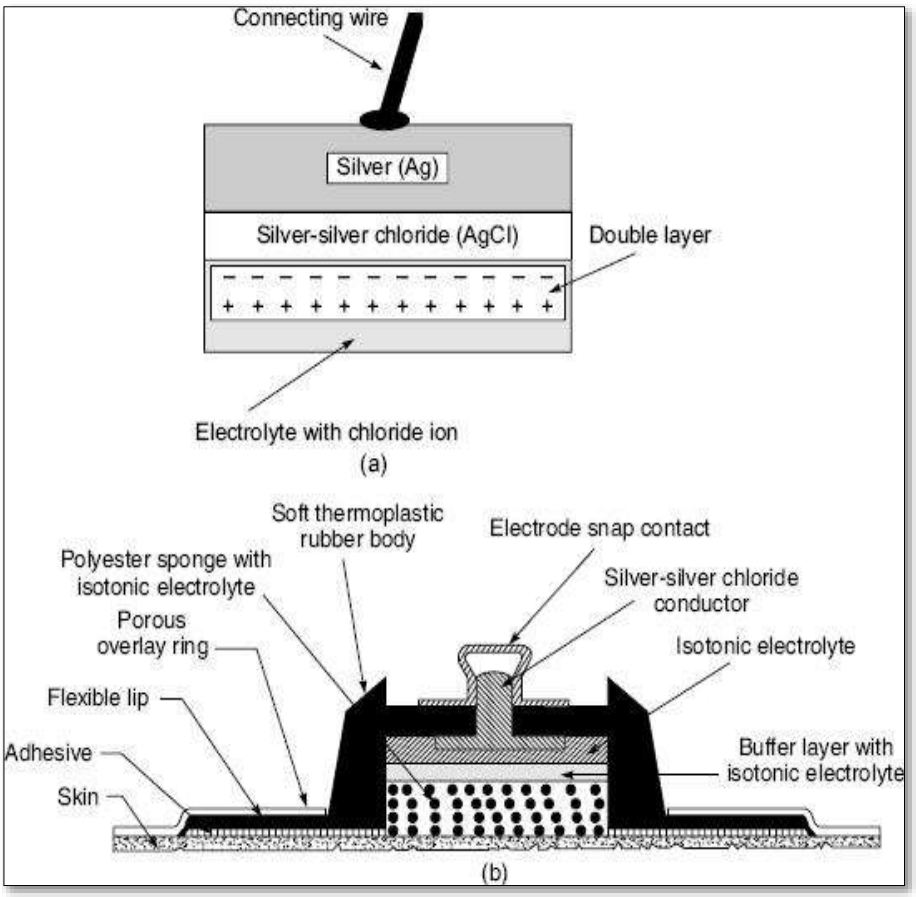

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Figure 12 (a) Principle of pre-gelled ECG electrode made of silver-silver chloride. The electrode has electrolyte layers that are made of a firm gel which has adhesive properties. The firm gel minimizes the disturbance of the charge double layer. (b) Cross-section of a typical pre-gelled electrode

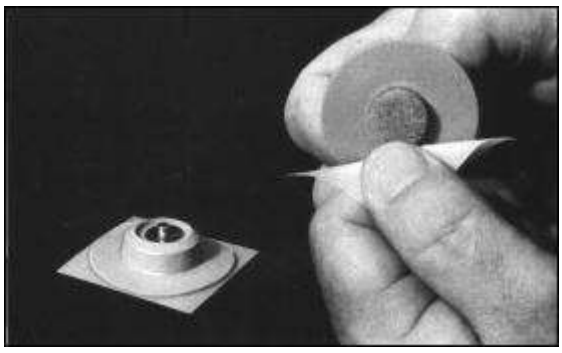

Figure 13 Disposable pre-gelled ECG electrode. A porous tape overlaying ring placed over the electrode resists perspiration and ensures positive placement under stress conditions (Courtesy: Del Mar Avionics, U.S.A.)

Various manufacturers offer common features in pre-gelled electrode construction. The lead wire's female connector "snaps" on, allowing a convenient snap-on pull-off connection with a 360 rotation providing mechanical and electrical connection. The plastic eyelet or sensor has a diameter of 0.5–1.5 cm and is electroplated with silver up to a thickness of 10 mm. The surface of the Ag layer is partially converted to AgCl. The tape is made from one of the adhesive coated occlusive foams made from a plastic, such as polyethylene, or a porous backing, such as non-woven cloth. Tapes used for first aid dressings are suitable. The electrode diameters range from 4–6 cm. Some advantages of these electrodes as compared to plates and Welsh bulbs are : as there is no risk of infection which is possible with reusable electrodes, their smaller size makes them less prone to detachment and also less time is required per ECG procedure.
Skin-electrode adhesion is an important performance criterion. Partial electrode lift would cause a gel dry out and intermittent contact artefacts, while premature electrode fall-off would increase monitoring costs. The adhesives used to secure electrodes to the skin are usually pressure sensitive adhesives which implies that a force must be applied to achieve adhesion. The adhesive should have good bonding capability; internal strength so that upon removal, objectionable "stringy" residue will not remain on the skin; good temperature stability; immunity to oxidation from air pollutants, saline and other common solutions; resistance to water, isopropyl alcohol, saline and other common solutions used in hospitals and low potential for skin irritation. Trimby (1976) describes an alternative design to reduce motion artefacts generated by mechanical shocks. Ideally, the thin layer (critical layer) of electrolyte just under the metal piece should be cushioned from all mechanical shocks. Figure 14 shows the construction of fluid column electrodes (HP 14245A, HP 14248A) in which the critical layer is protected by a semi-rigid collar. These and several other commercially available electrodes are surrounded on the first few tenths of a millimetre by a plastic collar. However, even with this mechanical stabilization, a strong force in the vertical direction will be transmitted up the electrolyte column and will disturb the interface area. Motion artefact will still be seen in the trace. The rigid collar serves to minimize interference from minor pulls on the lead wire.

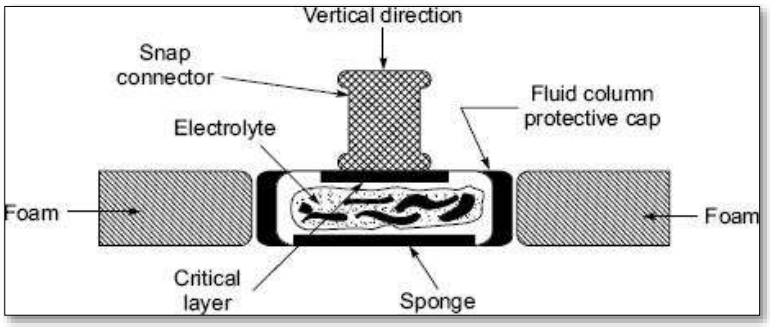

Figure 14 Construction of fluid column electrode (After Trimby, 1976; Courtesy: Hewlett Packard, U.S.A.)

Pre-gelled ECG surface electrodes are manufactured with 0.3–1.5 g of electrolyte paste (gel, cream, or jelly) in contact with the sensor which forms a conducting bridge with the skin. A high

value of [Cl^-] common in pastes renders the electrode more non-polarizable and decreases the electrolyte skin impedance. However, it must not be high enough to cause skin irritation. Ideally, gels should possess the following properties:

- ❖ Stay moist for the intended shelf life and during use. This is controlled by including a humectant in the gel.
- ❖ Prevent micro-organism and mould growth. Generally, the gels contain a bactericide/fungicide and may be disinfected using gamma radiation.
- ❖ Provide low electrolyte skin impedance by having ionic salts and surfactants.
- ❖ Cause minimum skin irritation, for which gels should have a pH range of 3.5–9.

Electrode-skin contact impedance with respect to pregelled disposable ECG electrodes was measured by Klingler *et al.* (1979 b) using a 10 Hz ac current source. It was found that the skin on the average contributed a resistance of 564 Ω; to the equivalent electrode impedance if mildly abraided, but contributed 54.7 kΩ if applied to clean, dry skin. They further found that over 90% of electrodes on abraided skin will have less than 5 kΩ impedance imbalance. The curve for clean, dry skin shows a very significant imbalance. In fact 20% of the electrodes will exceed 15 kΩ imbalance.

ECG electrodes are used in conjunction with cardiac monitors or ECG recorders which invariably have dc-input bias currents. Klingler *et al.* (1979 b) studied the effects of these small dc currents on the offset potentials of disposable ECG electrodes. They found that after periods ranging from a few minutes to several days, the electrodes tested (four brands of silver-silver chloride and two brands of stainless steel) exhibited offset potentials exceeding 200 mV after subjection to dc bias currents over 200 nA. All silver-silver chloride electrodes were able to withstand bias currents of 200 nA, with minimal changes in offset for periods up to seven days. On the other hand, the stainless steel electrodes exhibited large offset potentials within minutes after subjection to bias currents of only 100 nA. Based on this study, a 200 nA limit on the dc-input bias current for cardiac monitors is suggested.

The modern ECG monitors are generally provided with inputs protected against defibrillator overloads. The high defibrillating currents are harmlessly bypassed through neon or diode breakdown circuits. However, this unidirectional current passes through and tends to polarize the electrodes. Usually, the standards on ECG monitors require that the trace be readable within 5 s after three or fewer defibrillator discharges. This implies that the electrode polarization voltage must return to below 300 mV within a few seconds after application of the defibrillating voltages. Schoenberg *et al.* (1979) developed a standard test method for evaluating defibrillation recovery characteristics of disposable ECG electrodes.

ECG pregelled electrodes can be characterized electrically by tests developed by the Association for the Advancement of Medical Instrumentation (AAMI), USA to establish a reasonable safety and efficacy level in clinical use of electrodes. In abridged form the standards are as follows:

1. *Direct-current offset voltage:* A pair of electrodes connected gel-to-gel, after 1 min stabilization period must exhibit offset voltage no greater than 100 mV.

2. *Combined offset instability and internal noise:* A pair of electrodes connected gel-to-gel shall after 1 min. stabilization shall generate a voltage no greater than 150 mV in the passband.

3. *Alternating-current impedance:* The average value of 10 Hz for at least 12 electrode pairs connected at a level of impressed current not exceeding 100 mA peak to peak shall not exceed 2 kΩ. None of the individual pair impedances shall exceed 3 kΩ.

4. *Defibrillation overload recovery:* The absolute value of polarization potential of a pair of electrodes connected gel-to-gel shall not exceed 100 mV, 5s after each of four capacitor discharges. The capacitor should be 10 mF charged to 200 V and discharged through the electrode pair with 100 Ω; in series.

5. *Bias current tolerance:* The observed dc voltage offset change across an electrode pair connected gel-to-gel shall not exceed 100 mV when subjected to a continuous 200 mA dc current over the period recommended by the manufacturer for the clinical use of the electrodes. In no case shall this period be less than 8 hours.

Pasteless Electrode

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 (Stevens, 1963). 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 GΩ (1000– 30,000 MΩ). Two such electrodes are applied to the subject without any preparation of the skin and the output of the source followers is connected to a conventional electrocardiograph. Wolfson and Neuman (1969) also designed a capacitively coupling electrode and used a high input impedance amplifier having a MOSFET in the input stage arranged in a source-follower configuration. The capacitances encountered in such type of electrodes range from about 5000 to 20,000 pf/cm^2 of the electrode area (Geddes, 1972).

Conrad (1990) illustrates the construction of a capacitive electrode which is formed from conductive silicon wafer, oxygen diffused into one surface produces a silicon dioxide layer, which serves as the dielectric. A high performace FET operational amplifier in unity gain configuration acts as an impedance changer to permit use with systems designed for paste type electrodes. The insulated, capacitively coupled electrode is used on unprepared skin, which acts as one plate of the capacitor, the substrate acts as the other plate.

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 (Figure 15) 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 × 15 × 7 mm. Two FETs are cemented with epoxy glue in the middle of the shell, their centres 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. The source leads of the two FETs are connected to the differential inputs of an instrumentation amplifier. The amplifier (Analog Devices 521) has a high ac input impedance (> 100 MΩ).

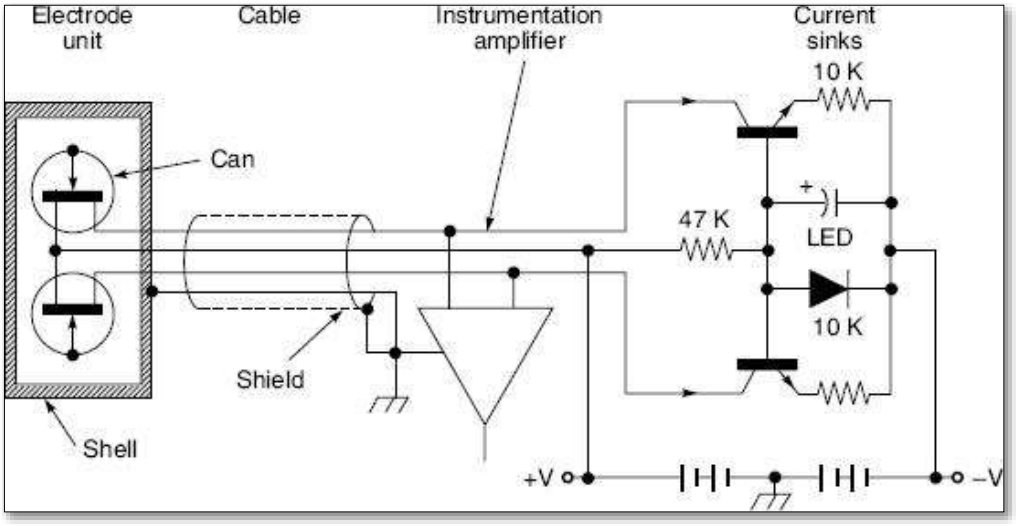

Figure 15 Schematic diagram of integrated electrode and amplifier arrangement for pasteless operation (After Luca et al. 1979; Reproduced by permission of Med. and Biol. Eng. and Comput.)

Among the drawbacks of the dry or pasteless electrodes are high electrode-skin resistance and the strong sensitivity to motion. Whereas a high electrode-skin resistance can be overcome by sufficiently high values of the amplifier input resistance, no universal remedy for eliminating motion artefacts from dry bioelectrodes has so far been suggested. Skin preparation certainly minimises motion artefacts (Burbank and Webster, 1978), but this time consuming manipulation loses the practicability of dry electrodes for routine clinical use and is also less acceptable to the patient.

An important characteristic observed with dry surface electrodes is that with the passage of time, the resistance of a dry electrode placed on dry human skin decreases exponentially. Geddes *et al.* (1973) report that measurements made by them on silver disc electrodes (1.7 cm diameter) showed that for 10 determinations, the average initial resistance was about 1.36 MΩ. After about 20 min. the resistance of the electrode-subject circuit had dropped to about one-tenth of its initial value. The fall in impedance is attributed to the presence of small amounts of perspiration accumulated under the electrode.

A non-contact capacitive electrode for biopotential sensor network is shown in Figure16. The sensor network utilizes a single conductive sheet to establish a common body wide reference line, eliminating the need for an explicit signal ground connection. Each electrode senses the local biopotential with a differential gain of 46dB over a 1-100Hz bandwidth. Signals are digitized directly on board with a 16-bit ADC. The coin sized electrode consumes 285μA from a single 3.3V supply, and interfaces with a serial data bus for daisy-chain integration in body area sensor networks (Chi *et al.,* 2009).

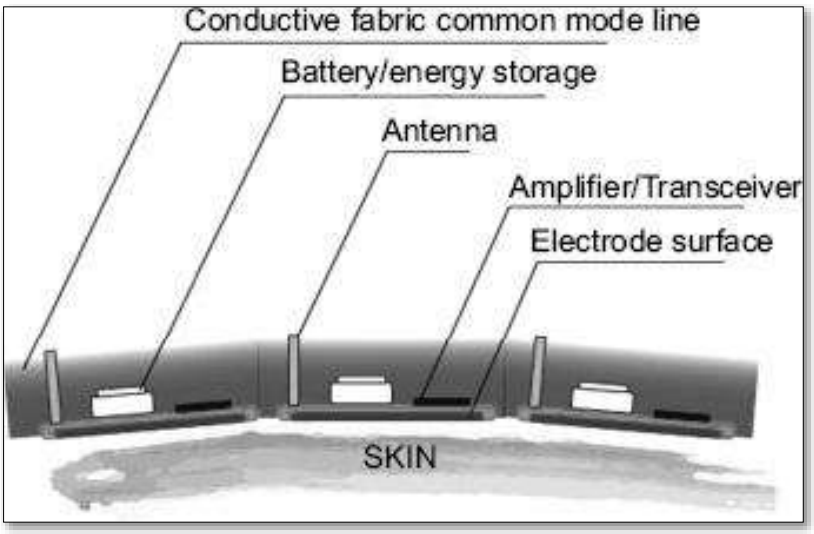

Figure 2.16 Integration of wearable non-contact biopotential sensors in a wireless bio-potential body network embedded in conductive fabric (After Chi et al., 2009)

Each sensor consists of two small round electrically connected standard printed circuit boards the size of a US nickel coin. The upper board contains a 16-bit analog-to-digital converter and voltage reference. Unlike traditional electrodes that output a single analog signal, interfacing is facilitated through two miniature 10-wire ribbon cables on each side which provides power along with the digital clock, control and data lines.

The ADC output from each board is a serial data stream which is shifted in a daisy chain from board to board to the end of the chain which connects to a custom USB data acquisition interface. This connection scheme minimises the amount of cabling required across the sensor network, where the total connection length scales with the number of sensors and the average distance between sensors.

Biopotentials are sensed through a 228 mm^2 copper fill insulated by solder mask on the lower board, which is shielded from external noise by the outer copper ring and a solid metal plane directly above the electrode. The amplifier circuit is placed directly on the top surface of the lower board and output an analog signal which is digitized by the upper board.

Fabric-based Electrodes

Dry electrodes suffer from noise interference. Noise is typically generated from motion artefacts and power line interference. A common solution used to suppress noise in dry electrode signals is a buffer amplifier. A buffer amplifier is essentially an impedance converter, one that converts high-impedance signals to low-impedance signals. Recent research has experimented with integrating buffer amplifiers with electronic textiles to produce fabric-based active electrodes. Fabric-based active electrodes are produced by screen printing a buffer circuit pattern and electrode onto a nonwoven textile. Thereafter, the electrical components are directly attached to the fabric circuit to produce an active electrode.

Merritt *et al.,* 2009 designed and fabricated two versions of the fabric-based active electrode. The first version involved screen printing the circuit onto a nonwoven textile and directly attaching the surface mount device (SMD) components to the textile circuit. The second version explored the possibility of using an interposer, thereby allowing placement of the circuit and its components on a much smaller printed circuit board (PCB). The use of an interposer offers significant advantages in regards to size, cost, manufacturability and resilience.

Direct-Attach Active Electrodes, design involves directly attaching SMD components to a screen-printed electronic textile circuit, which comprises several layers: electrode, circuit, dielectric coating, and conductive adhesive as shown in Figure 17. The electrode layer is located on the reverse side of the fabric, and it is connected to the circuit on the opposite side of the fabric by a via positioned next to the resistor R_1. The conductive lines are printed with a width of 1 mm to reduce the percentage of voids that can hamper line conductivity. Furthermore, the circuit is positioned away from the top of the electrode to allow the electrode to conform to the contours of the body and to reduce the risk of short circuits to the electrode. The conductive adhesive stencil design includes patterns for SMD pads, external pads, and the electrode to R_1 via. The fabrication technique is described in the original article, which also covers the design of interposer active electrodes.

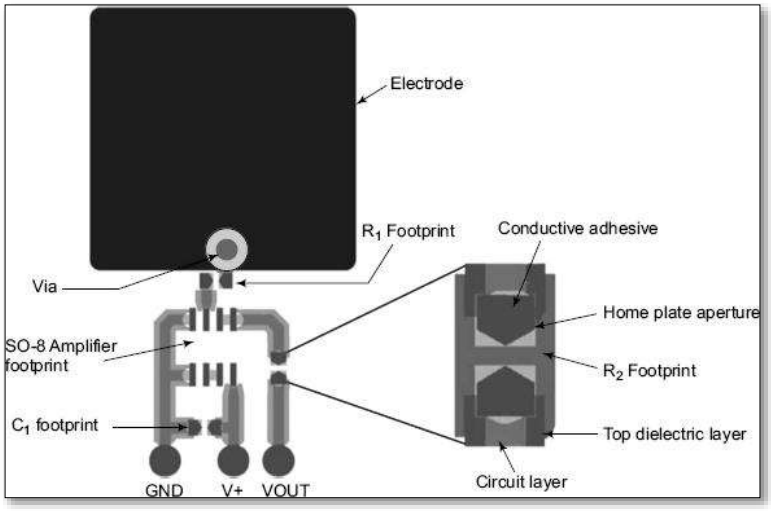

Figure 17 Active electrode screen/stencil design using home plate pattern (After Merrit et al., 2009)

During medical procedures, such as surgery, a patient's vital signs are typically monitored using a web of wires connected to adhesive electrodes. The large number of wires inhibits the medical team's access to the patient while the adhesive electrodes can detach, fail, or be out-of-stock, causing delays in the procedure. In order to combat these problems, Chen *et al.* (2008) have developed the SmartPad: a system that displays a patient's electrocardiogram (ECG) signal without adhesive pads, wires, or active intervention from a clinician. The system automatically selects three electrodes from an array of Cu/Ni- fabric-based electrodes patterned on a thin pad on which the patient lies. The selected electrodes are used to provide a differential 3-lead measurement of the patient's ECG, which is then transmitted wirelessly and displayed on a laptop computer. Figure 18 shows SmartPad electrode array pattern. A flexible conductive Ni/Cu polyester fabric tape from 3M (CN-3190) is used to create the electrode array. This material does not irritate bare skin and is resistant to corrosion. This material is patterned onto a thin foam pad and is connected to interface wires via its own conductive adhesive. The thin foam pad can then be placed in the target environment such as on a stretcher pad or operating table.

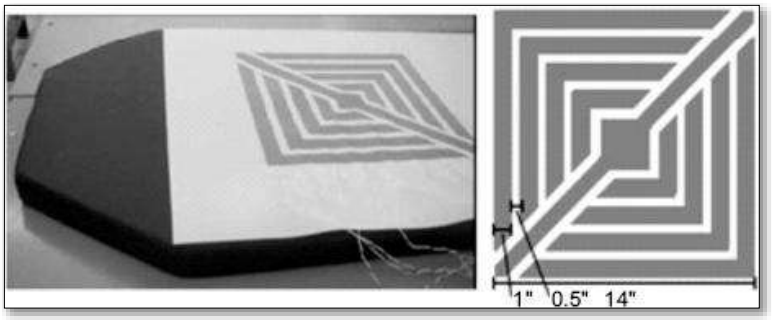

Figure 18 SmartPad electrode array pattern (After Chen et al., 2008)

The electrode pattern consists of a concentric series of 'L' shaped strips with a diagonal centre strip. The electrode strips are one inch wide and are spaced approximately 0.5 inches apart. This geometry is designed so that the optimum ECG signal can be extracted by insuring that at least one combination of two electrodes form a vector across the patient's heart, regardless of the size and orientation of the patient. The multiplexing front-end chooses from one of four possible 'L' electrodes for both the *In+* and *In-* inputs. The three-lead ECG circuit used also requires an additional electrode contact to perform the common-mode cancellation. The pool of possible common-mode electrodes includes the central diagonal electrode along with seven of the 'L' electrodes. The electrode configuration and detection algorithm enables hands free operation on the part of the clinician and shows the noise benefits of using a wireless acquisition system.

ELECTRODES FOR EEG

Small metal discs usually made of stainless steel, tin, gold or silver covered with a silver chloride coating are generally used for recording EEG. They are placed on the scalp in special positions. These positions are specified using the International 10/20 system. Each electrode site is labeled with a letter and a number. The letter refers to the area of brain underlying the electrode e.g. F-Frontal lobe and T-Temporal lobe. Even numbers denote the right side of the head and odd numbers the left side of the head. Figure 19 shows EEG cables with disc electrodes to which electrode gel is applied to maximize skin contact and allow for a low resistance recording of EEG from the subject's scalp.

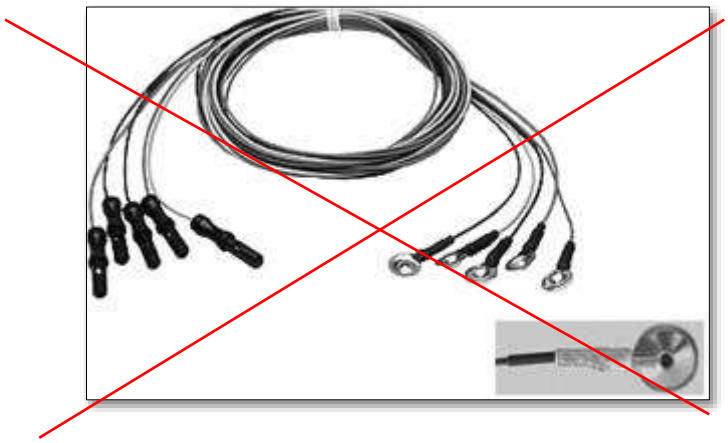

Figure 19 EEG cables with disc electrodes.

This type of electrode provides enough space to contain electrode gel. In these electrodes, the skin never touches the electrode material directly. The electrode-tissue interface has impedance depending on several factors. Some of these factors are the interface layer (such as skin preparation, fat mass, hair, etc.), area of electrode's surface, and temperature of the electrolyte. Another types of electrode for EEG (electroencephalogram) recording is the chlorided silver disc (Figure.19) having approximately 6–8 mm diameter. Contact with the scalp is made via an electrolytic paste through a washer of soft felt. They have ac resistance varying from 3–20 kΩ. 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.

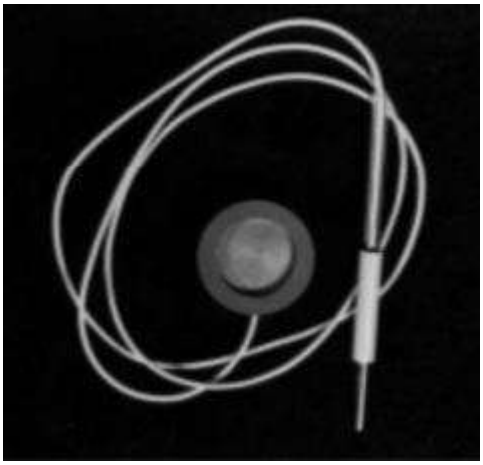

Figure 20 EEG electrode which can be applied to the surface of the skin by an adhesive tape (Courtesy: In Vivo Metrics, U.S.A.)

Another type of EEG electrode is dry electrode (Figure.20). This type of electrode does not need a long set-up time and it is convenient for long-term recordings. These properties are specially advantageous for neurofeedback applications. The electrode consists of 1.5 mm thick silicone conductive rubber-shaped discs of 8 mm diameter. The active side of the electrode is capacitive and coupled through a layer of insulating silicon rubber with a metal shield wired to the active guard shield. The impedance of the realized electrodes at 100 Hz is greater than 20 M with a parasitic capacitance smaller than 2 pF (Gargiulo, 2008).

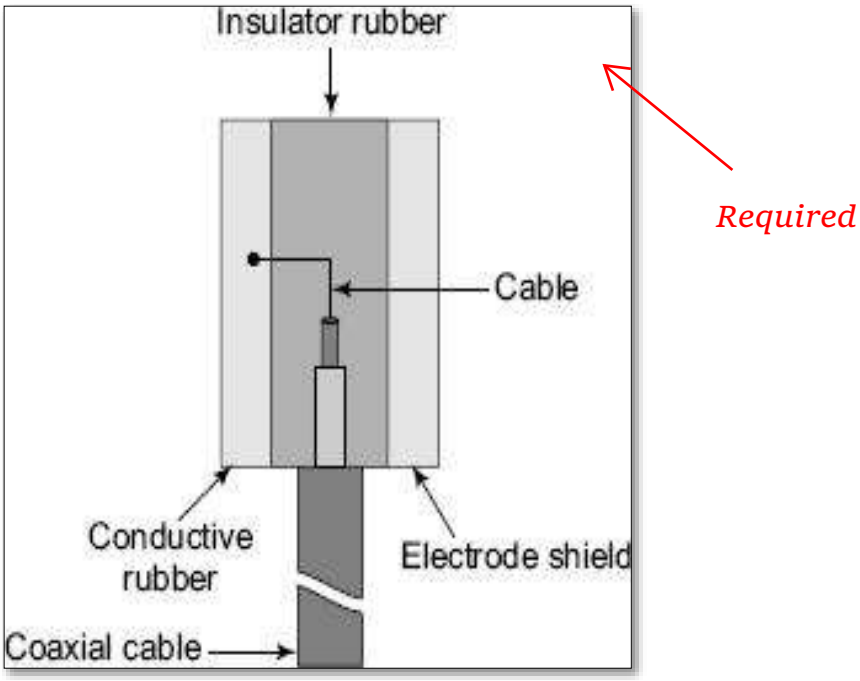

Figure 21 Construction of EEG dry electrode

The EEG electrodes can be classified as disposable, reusable disc and cup shaped, subdermal needles (single-use needles that are placed under the skin), and implanted electrodes. Needles are available with permanently attached wire leads, where the whole assembly is discarded. They are made of stainless steel or platinum. Some of EEG electrodes can be used for special applications. For example, implanted EEG electrodes also can be used to stimulate the brain and map cortical and subcortical neurological functions (Usakli, 2010).

ELECTRODES FOR EMG

Electrodes for electromyographic work are usually of the needle type. 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 unfavourable electrode material from the point of view of noise, it is preferred in EMG work due to its mechanical solidity and low price. 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. 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.

Multi-element needle electrodes are used to pick up the signals from individual fibres of muscle tissue. Special needles are available using 25-micron diameter electrode surfaces and having up to 14 pick up surfaces down the side of one needle. From the point of view of construction, needle electrodes are the simplest. However, edging of the needle point to the suitable angle, providing a proper plastic coating, making them resistant against thermal and chemical stresses and ensuring histological suitability is a difficult manufacturing process.

For the measurement of potentials from a specific part of the brain, longer needles are actually inserted into the brain. The needles are precisely located by means of a map or atlas of the brain. Generally, a special instrument called a stereotaxic instrument is used to hold the subject's head and guide the placement of electrodes. Often, these electrodes are implanted to permit repeated measurements over
an extended period of time.

Figure 22 shows different types of intra muscular needle electrodes. The choice of the type of electrode that can best measure the physiological function of interest depends upon the specific features associated with each electrode type, their limitations, cost, availability and potential risk.

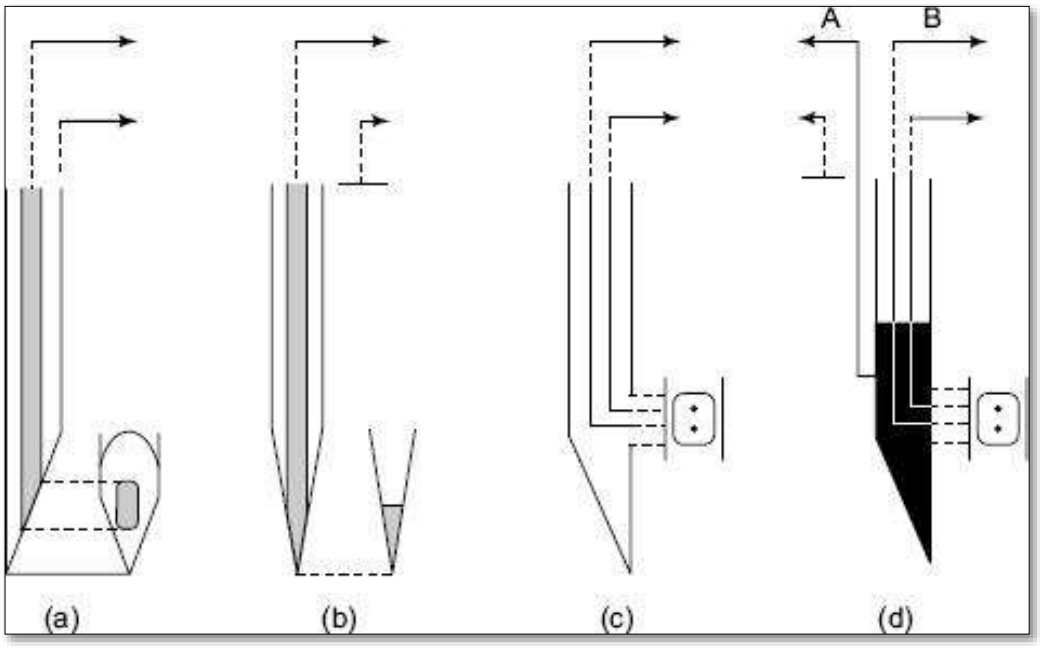

Figure 22 Four common types of needle EMG electrodes. (a) Concentric needle electrode. The recording wire is represented by the stippled area, running the length of a hypodermic needle. (b) A monopolar needle electrode that is simply a wire insulated all around except at the tip. (c) Single-fiber electrode. (d) Macro-EMG electrode. The setup on the right side (4B) is identical to that of the single-fiber electrode. At the tip of the needle is a very large recording surface (the area in black measuring 1.5 cm in length) for detecting the action potentials generated by all the constituent muscle fibers within the motor unit territory (A). (Adapted from Stalberg, 1986)

The commonly used concentric needle electrode (Figure. 22 a) has a single insulated wire inside the cannula of a hypodermic needle, fixed in place by epoxy glue and cut flush with the needle tip. This recording wire, with a recording surface of 150 by 600 μm at the tip, is referenced to the cannula.

Another commonly used electrode is a monopolar needle electrode (Figure. 22 b) that is made up of an insulated solid needle except at the most distal 300 μm at the tip, referenced to a surface electrode; thus, it has a slightly larger pickup area.

To study electrical transmission in single muscle fibres, an electrode with a much smaller recording area is required. This electrode (Figure 22 c) with a recording surface of 25 μm is located in a side port 3 mm back from the needle tip on the opposite side of the bevel. This configuration helps to minimize the risk of studying muscle fibres damaged by the needle tip during insertion. Diameter of the uptake area is about 300 μm.

The configuration of macro-EMG electrode (Figure. 22 d) is similar to the single fibre electrode except that the distal 1.5 cm of the needle electrode is bare. This recording surface, referenced to a surface electrode on the skin, has a very large pickup area and can therefore detect the entire motor unit territory.

The ground electrode for EMG studies usually consists of a conducting strip which is inserted into a saline soaked strap and wrapped around the patient's limb. The ground electrode is usually positioned over bony structures rather than over large muscle masses, in the vicinity of the recording and stimulating electrodes, and where possible, equidistant from them. Surface electrodes are employed for recording gross electrical activity from a particular group of underlying muscles in nerve-conduction velocity measurements. A single surface electrode may also be used as the reference (indifferent) electrode with monopolar needle electrodes. Surface electrodes can be easily and quickly attached and are generally comfortable to wear over long periods. Surface electrodes usually consist of square or circular metal (chlorided silver) plates with leadoff wires attached. They are held in place by straps or adhesive tapes. To reduce electrical resistance between the skin and the electrode, the use of saline soaked felt pads or a small amount of electrode gel between the electrode surface and the skin is recommended. Disposable, adhesive type electrodes are also used for EMG work.

For surface electrodes, simple platinum or silver disc electrodes, pre-gelled Ag-AgCl electrodes, and wet-gel electrodes are commonly used. The disc electrodes (Figure 23 a) are reusable while the gel electrodes (Figure 23b) are designed for single use. Whenever an electrical stimulation is applied, the electrodes used must be properly designed to deliver such electrical stimulations otherwise the power density generated at the skin contact can result in patient injury. Just like ECG electrodes, the EMG electrodes can be pre-gelled or wet gelled.

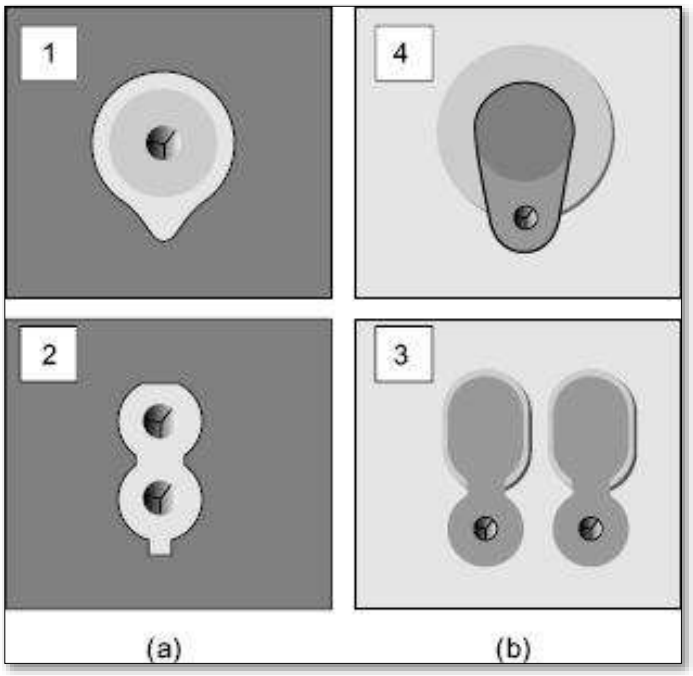

Figure 2.23 EMG surface silver-silver chloride electrodes (a) pre-gelled electrodes (1, 2) (b) wet-gels (3, 4) (After Quach, 2007)

When intramuscular EMG is required to look into the electrical activity of deeper or overlaid muscles, thin and flexible fine wire electrodes are used. The fine wire electrode is inserted into the muscle site of interest. The needle or steel cannula is removed, and the electrode wires are connected to the steel spring adapters to minimize movement artefacts.

For all electrode types, additional measures can be taken to affix the electrode cabling to the patient body to minimize movement artifacts. Regular adhesive tape, hook and loop fasteners, and elastic straps are commonly used to secure cabling onto the body, but never the electrodes as this will affect the readings.

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. The contact impedance of the skin depends upon the type of electrolyte used and the time (Trimby, 1976).

Figure 24 shows the effect of these parameters. A low concentration sodium chloride electrolyte has 0.5% NaCl and a high concentration electrolyte has a concentration in the range of 5 to 10% NaCl. 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.

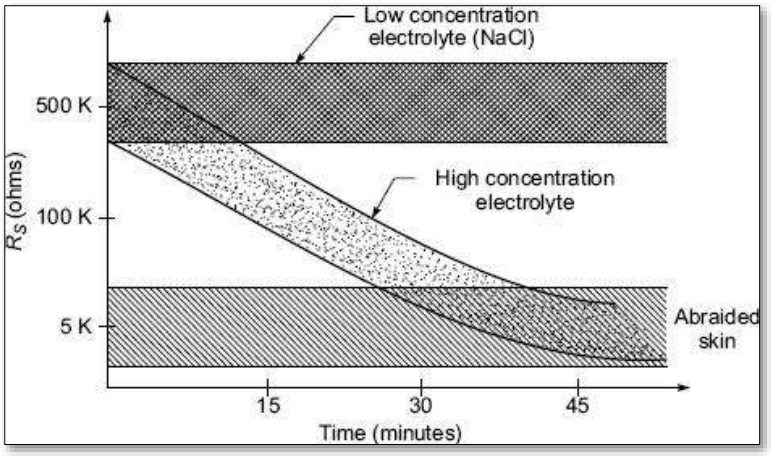

Figure 24 Variation of contact-impedance with electrolyte concentration and time (Redrawn After Trimby, 1976; Courtesy: Hewlett Packard, U.S.A.)

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. Jenkner (1967) used silastic S-2086 by Dow Corning with EEG electrodes and showed that contact resistance was almost the same as with a conventional electrode which make use of electrode jelly.

The design of a wet electrode relies on the fact that the skin has a dry dielectric outer layer requiring gel compositions and electrode pads in order to establish a strong conductive path from the patient to the ECG system. Over time, these gels begin to dry out (as well as possibly cause skin irritation), changing the impedance characteristics between the pad and patient. Air gaps may develop between the electrode and skin, especially if there is hair on the skin surface, increasing the contact impedance. Figure.25 shows the typical impedance variation of both a wet and dry electrode over a five-hour period when connected to a subject. The impedance of the wet electrode varied by more than 10 kΩ as the conductive gels dried out over time.

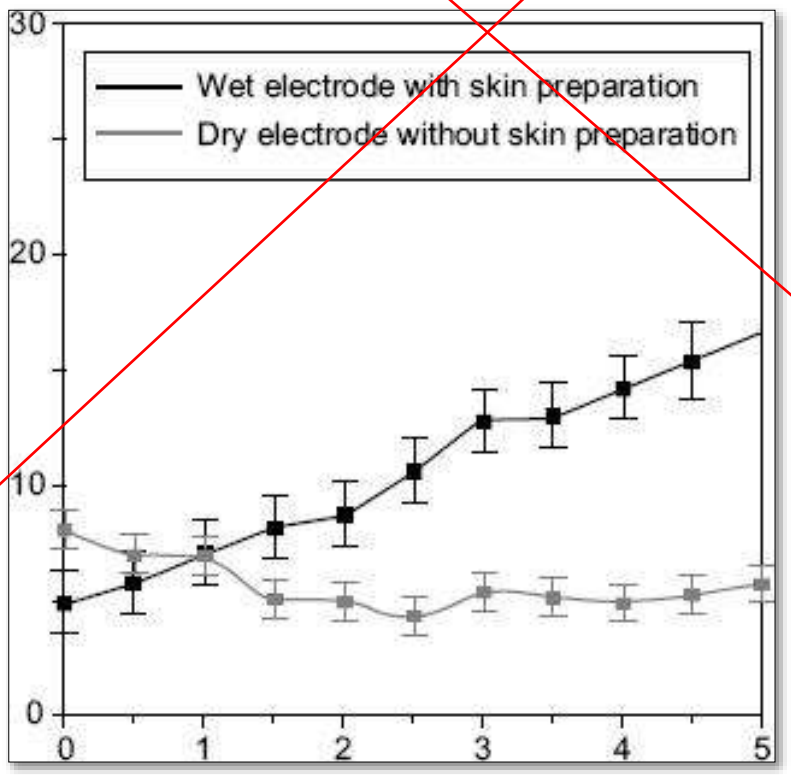

Figure 25 Wet and dry electrode impedance variations over time (Courtesy: M/s Texas Instruments, 2012)

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 [Figure.26(a)] and glass microcapillaries (or micropipette) [Figure. 26(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.

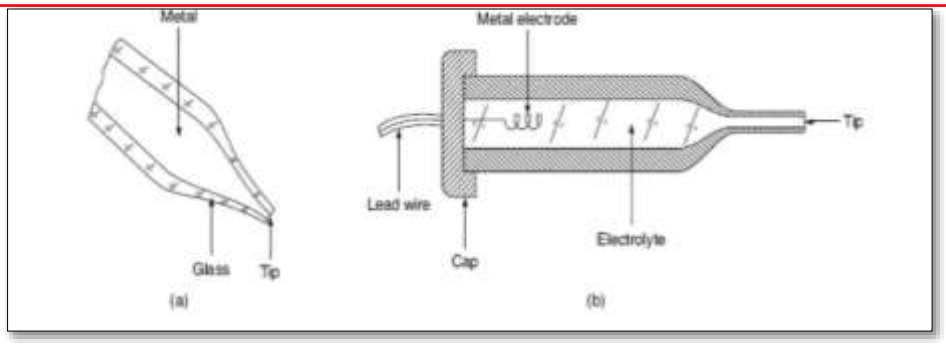

Figure 26 Micro-electrodes(a) Metal microelectrode (b) Micropipete or microcapillaries electrode

The microelectrodes have 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. Similarly, a micropipette tip is filled with an electrolyte which substitutes an electrolytic conductor of small cross-sectional area, which gives a micropipette 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 Microcapillary Electrodes

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 micropipette 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 Microelectrodes

Metal electrodes with very fine tips used for recording from single cells have the advantage over glass micropipettes 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 1mm.

These very 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 mm. 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. Skrzypek and Keller (1975) illustrated a new method of manufacturing tungsten microelectrode permitting close control of microelectrode parameters. In this technique, the tips are dc electroetched to diameters below 500° A and completely covered with polymethyl methacrylate. An electron beam from a scanning electron microscope is then used to expose a precise area on the tip for later removal by chemical methods. Recording results with these electrodes suggested good desirable recording characteristics, i.e. ability to isolate and hold single cells.

Figure 27 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 reproducible performance, with ease of cleaning and maintenance. The metal-electrolyte interface is between the metal film and the electrolyte of the cell.

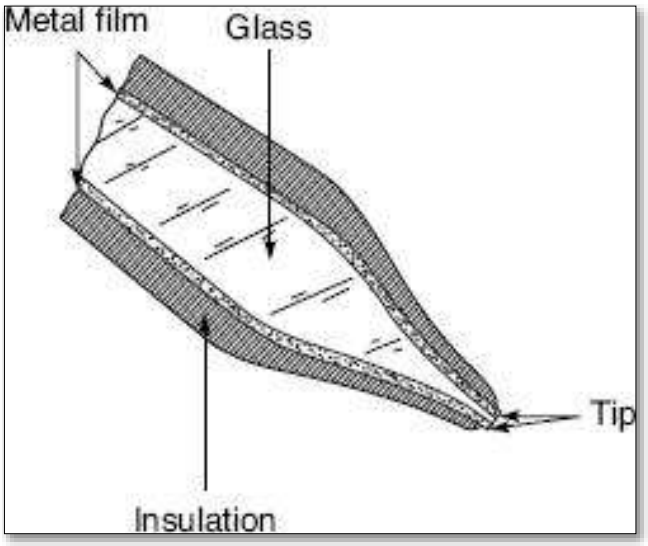

Figure 27 Micro-electrode—Thin metal film coated micropipette

Tungsten is preferred for constructing microelectrodes due to its mechanical strength and its apparent inertness. Although tungsten itself is reactive, a surface layer of tungsten oxide will, in most situations, protect the metal against corrosion. The electrical properties of tungsten microelectrodes made with a taper of the tip of about 1:10 and insulated with lacquer leaving a tip length of about 10–100 μm were studied by Zeuthen (1978). The resting potential in saline was found to be –0.3 V relative to a silver-silver chloride reference electrode for input currents less than 10^{-12} A. The small signal impedance was, ideally that of a capacitor 0.4 pF/mm^2, between 10 and 1000 Hz. Imperfect insulation at the tip caused this impedance to be increasingly resistive. The electrochemical properties of tungsten showed that it behaves as an inert metal within the potentials where it is usually used in biological experiments. Jobling *et al.* (1981) constructed an active microelectrode array using IC fabrication technology to record extracellular potentials in nervous tissues. The array substantially reduces the noise caused by electrostatic pick-up with a good long-term stability.