Bio-Signals and Transducers

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Abstract: The signals that are produced in a living body are known as bio-signals or bioelectric potentials. The measurement of bio-signals from certain organs of human body is important for the diagnosis of physical fitness of a person. In order to measure and record potentials and, hence currents in the body, appropriate transducers are used. This paper first mentions the origin of biopotentials and then it goes on to describe the basic mechanisms and operating principles of some of the transducers used to pick up biopotentials in biomedical applications. The purpose and the electrical behavior of the biopotential electrodes are also discussed in brief. Photographs of some practical electrodes and sensors are also included at the end of the paper.

1. Introduction

Since the beginning of medicine, physicians have been using their senses to determine various physical parameters of the patient, such as position of body organs, temperature of the body, color of the skin, and so on. In an attempt to quantify the measurement of these and additional parameters from the living system, we have seen an increased application of technology to clinical and biomedical research. In many cases, instruments that were developed originally for the physical sciences were adapted for specific medical application.

A signal can be defined as a function that conveys information, generally about the state or behavior of a physical system. Although signals can be represented in many ways, in all cases the information is contained in some pattern of variations. Signals are used

- to communicate between humans and between humans and machines;
- to probe our environment to uncover details of structure and state not easily observable; and
- to control and utilize energy and information.

Examples of various types of signals include:

- Speech - in telephony, radio
- Biomedical signals - Electroencephalogram (EEG-brain signals), ECG
- Sound and music (produced by CD player)
- Video and Image (that we watch on TV).
- Radar signals (used to determine the range and bearing of distant targets)

Signals are represented mathematically as functions of one or more independent variables. A signal is represented in the time domain, which shows instantaneous amplitude as a function of time.

Analog signals are those for which both time and amplitude are continuous. (Fig. 1). Digital signals are those for which both time and amplitude are discrete (Fig. 2).

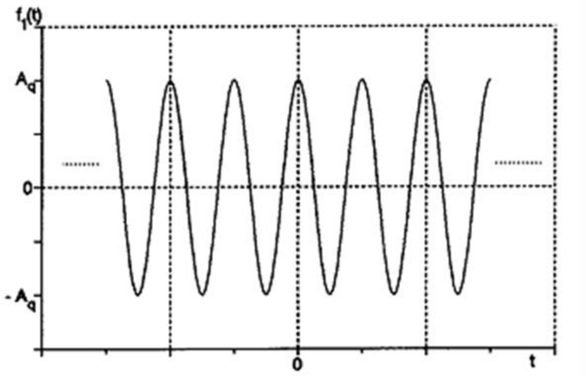

Fig. 1: Analog signal

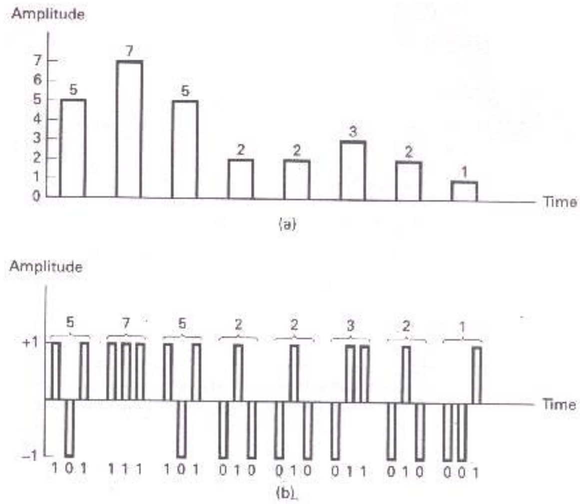

Fig. 2: Digital Signals: (a) digital signal (b) encoded digital signal.

The signals that are produced in a living body are known as bio-signals or bioelectric signals. The physicians and biomedical researchers are interested in measuring the size, shape, and position of the organs and tissues of the body. Variations in these parameters are important in discriminating normal from abnormal function. Some examples of Bio-signals are:

1. ENG: Electroneurogram (signals from nerves)
2. EMG: Electromyogram (signals from muscles)
3. ECG: Electrocardiogram (signals from heart)
4. ERG: Electroretinogram (signals from retina of an eye)
5. EOG: Electro-Oculogram (signals from cornea and retina of an eye)
6. EEG: Electroencephalogram (signals from brain using electrical means or electrodes)
7. MEG: Magnetoencephalogram (signals from brain using magnetic field mapping)
8. Ultrasonogram (imaging from ultra-sound reflection from the internal organs of the body).

A *transducer* is a device that converts one form of energy to another. A sensor, on the other hand, converts a physical parameter to an electric output signal [1, 2]. For example, an electrical speech signal is produced using a microphone. Therefore, the microphone is a transducer or sensor. This signal is then processed and displayed so that humans can perceive the information. An electrical output from the sensor is normally desirable because of the advantages it offers in further signal processing.

Bioelectric potentials are produced as a result of electrochemical activity of a certain class of cells, known as excitable cells, that are components of nervous, muscular, or glandular tissue. Electrically they exhibit a ***resting potential*** and, when appropriately stimulated, an ***action potential*** [1]. Appropriate electrodes and/ or transducers are needed in most of the applications to pick-up bio-signals from certain organs of the human body. In biomedical applications, an *electrode* is used to pick up signal from the body. It is a special *transducer* that transforms an ionic current in into an electronic current.

Section 2 examines the principal functions and requirements of biopotential electrodes. It also covers in brief the different forms of biopotential electrodes used in various types of medical instrumentation systems. Section 3 deals with the basic mechanisms and principles of other transducers and sensors used in biomedical applications. Section 4 describes the process of measuring body temperature using both direct and indirect methods. Section 5 describes, in brief, the ultrasonogram transducers, conclusions are included in section 6 and references in section 7. Finally, some photographs of biopotential electrodes and sensors are included in Appendix-A.

2. Biopotential Electrodes

Interface is the surface common to two areas or the meeting point between two electrical or electronic circuits. Proper matching or mediating is required for successful interfacing between the two entities. In biomedical engineering, two types of interfacing may be considered, which are

- Interfacing with skin and electrode, and
- Interfacing with electrical or electronic circuits linking one device with another.

In order to measure and record potentials and, hence, currents in the body, it is necessary to provide some interface between the body and the electronic measuring apparatus/circuit. This interface function is carried out by biopotential electrodes.

2.1 Electrical Characteristics of Electrodes

In any practical measurement of potentials, current flows in the measuring circuit for at least a fraction of the period of time over which the measurement is made. Ideally this current should be very small. However, in practical situations, it is never zero. Biopotential electrodes must therefore have the capability of conducting a current across the interface between the body and the electronic measuring circuit. The electrode actually carries out a transducing function, because current is carried in the body by ions, whereas it is carried in the electrode and its lead wire by electrons. Thus the electrode must serve as a transducer to change an ionic current into an electronic current. This greatly complicates electrodes and places constraints on their operation [1, 3].

Theoretically, two types of electrodes are possible: those that are perfectly polarizable and those that are perfectly nonpolarizable. This classification refers to what happens to an electrode when a current passes between it and the electrode.

Perfectly polarizable electrodes are those in which no actual charge crosses the electrode-electrolyte interface when a current is applied. Of course, there has to be current across the interface, but this current is a displacement current, and the electrode behaves as though it were a capacitor.

Perfectly nonpolarizable electrodes are those in which current passes freely across the electrode-electrolyte interface, requiring no energy to make the transition. Thus, for perfectly nonpolarizable electrodes there are no overpotentials.

Neither of these two electrodes can be fabricated; however, some practical electrodes can come close to acquiring their characteristics. Electrodes made of noble metal come closes to behaving as perfectly *polarizable* electrodes. The electrical characteristics of such an electrode produce a strong capacitive effect, and is not suitable for practical applications.

The Silver–Silver Chloride electrode is a practical electrode that approaches the characteristics of a perfectly *nonpolarizable* electrode and can be easily fabricated in the laboratory [4]. It is a member of a class of electrodes each of which consists of a metal coated with a layer of slightly soluble ionic compound of that metal with a suitable anion. The whole structure is immersed in an electrolyte containing the anion in relatively high concentrations. The structure is shown in Fig. 3.

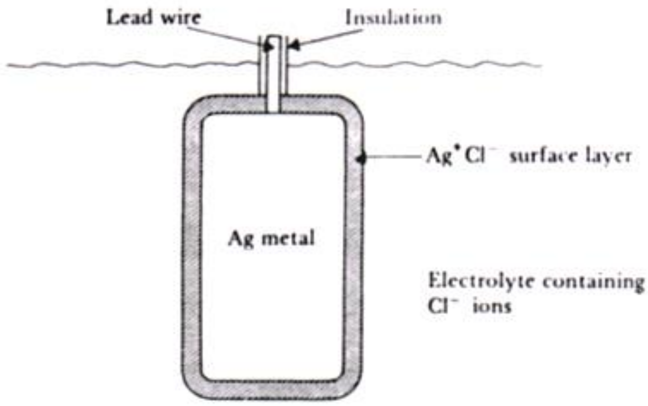

Fig. 3: A silver metal base with attached insulated lead wire is coated with a layer of the ionic compound AgCl. (This material – AgCl – is only very slightly soluble in water, so it remains stable). The electrode is then immersed in an electrolyte bath in which the principal anion of the electrolyte is Cl^-. For best results, the electrolyte solution should also be saturated with AgCl so that there is no chance for any of the surface film on the electrode to dissolve.

The electrical characteristics of electrodes have been the subject of much study. Often the current-voltage characteristics of the electrode-electrolyte interface are found to be nonlinear, and, in turn, nonlinear elements are required for modeling electrode behavior. Specifically, the characteristics of an electrode are sensitive to the current passing through the electrode, and the electrode characteristics at relatively high current densities can be considered different from those at low current densities. The characteristics of electrodes are also waveform-dependent. When sinusoidal currents are used to measure the electrode's circuit behavior, the characteristics are also frequency-dependent.

For sinusoidal inputs, the terminal characteristics of an electrode have both a resistive and a reactive component. The simple series equivalent circuit, however, does not present the entire picture. If we combine the series resistance-capacitance equivalent circuit with a voltage source representing the half-cell potential and a series resistance representing the interface effects and resistance of the electrolyte, we can arrive at the biopotential electrode equivalent circuit model shown in Fig. 4 [1].

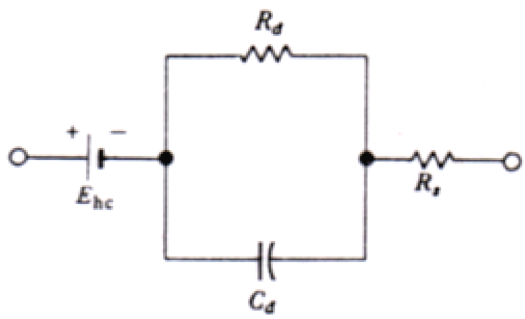

Fig. 4: Equivalent circuit for a biopotential electrode in contact with an electrolyte. E_{hc} is the half-cell potential. R_d and C_d make up the impedance associated with the electrode-electrolyte interface and polarization effects, and R_s is the series resistance associated with interface effects and due to resistance in the electrolyte.

When biopotentials are recorded from the surface of the skin, we must consider an additional interface – the interface between the electrode-electrolyte and the skin – in order to understand the behavior of the electrodes.

In coupling an electrode to the skin, we generally use a transparent electrolyte gel containing Cl^- as the principal anion to maintain good contact. Alternatively, we may use an electrode cream, which contains Cl^- and has the consistency of hand lotion. The interface between this gel and the electrode is an electrode-electrolyte interface, as described above. However, the interface between the electrolyte and the skin is different and requires some explanation. To review the structure of the skin, let us look at Fig. 5, which shows a cross-sectional diagram of the skin [1].

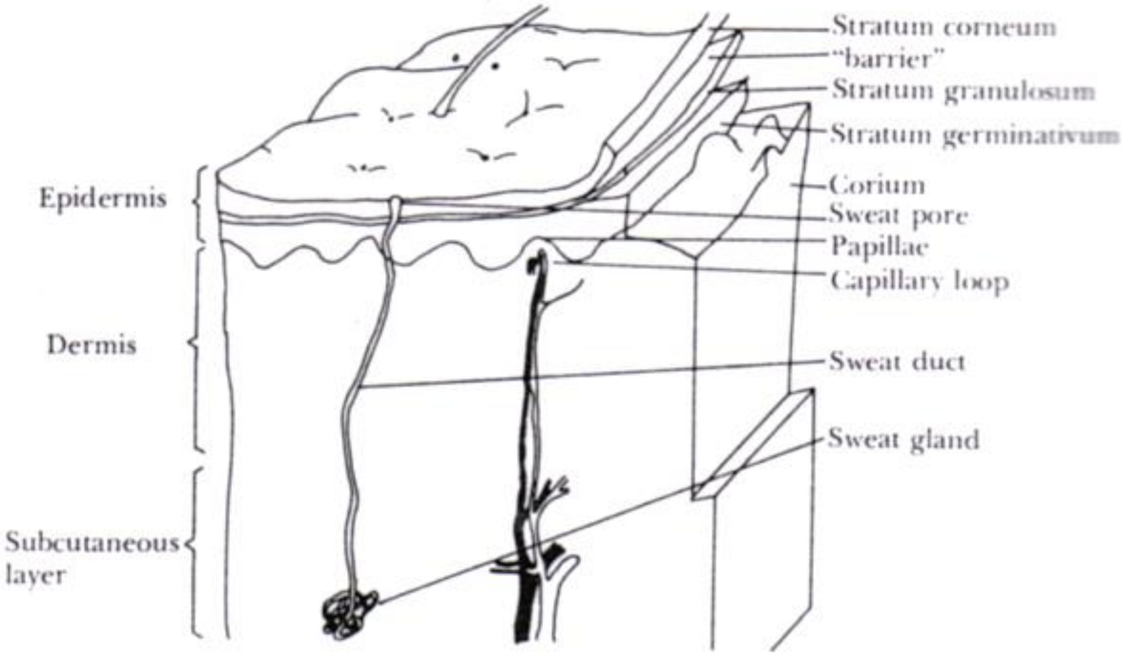

Fig. 5: Magnified section of skin, showing the various layers.

The skin consists of three principal layers that surround the body to protect it from its environment and that also serve as appropriate interfaces. The outermost layer, or *epidermis*, plays the most important role in the electrode-skin interface. This layer, which consists of three sublayers, is constantly renewing itself. Cells divide and grow in the deepest layer, the *stratum germinativum*, and are displaced outward as they grow by the newly forming cells underneath them. As they pass through the *stratum granulosum*, they begin to die and lose their nuclear material. As they continue their outward journey they degenerate further into layers of flat keratinous material that forms the *stratum corneum*, or horny layer of dead material on the skin's surface. These layers are constantly being worn off and replaced at the *stratum granulosum* by new cells. The *epidermis* is thus a constantly changing layer of the skin, the outer surface of which consists of dead material that has different electrical characteristics from live tissue.

The deeper layers of the skin contain the vascular and nervous components of the skin as well as the sweat glands, sweat ducts, and hair follicles. These layers are similar to other tissues in the body and, with the exception of the sweat glands, do not bestow any unique electrical characteristic on the skin. To represent the electric connection between an electrode and the skin through the agency of electrolyte gel, our equivalent circuit of Fig. 4 must be expanded, as shown in Fig. 6 [1].

2.2 Types of Electrodes

Over the years many different types of electrodes for recording various potentials on the body surface have been developed. The electrodes may be broadly classified into the two, namely, the **Body Surface Electrodes** ,which are used on the surface of the body, and **Internal Electrodes**., which are inserted into the body in the form of needles, wires, or implanted electronic circuits such as radiotelemetry transmitter.

2.2.1 Body Surface Electrodes

Among the various types of body surface electrodes, the following are worth mentioning [1]:

(a) Metal-plate electrodes
(b) Suction electrodes
(c) Floating electrodes.

(d) Flexible electrodes
(e) Dry electrodes.

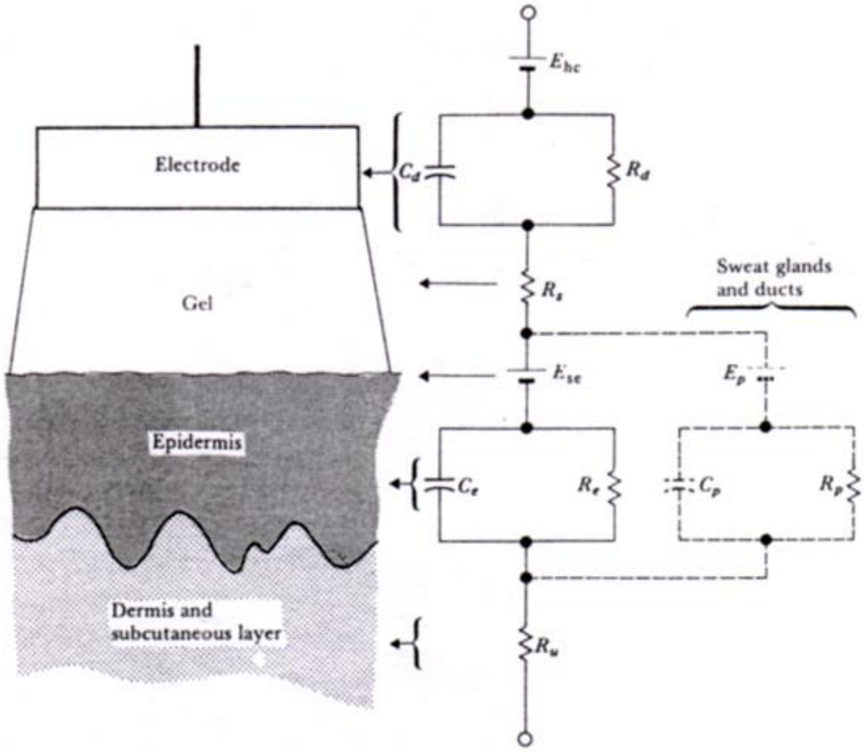

Fig. 6: A body-surface electrode is placed against skin, showing the total electrical equivalent circuit obtained in this situation. Each circuit element on the right is approximately the same level at which the physical process that it represents would be in the left-hand diagram.

(a) Metal-plate electrodes: One of the most frequently used forms of biopotential sensing electrodes is the metal-plate electrode. In its basic form, it consists of a metallic conductor in contact with the skin. An electrolyte gel is used to establish and maintain the contact. Fig. 7 shows several forms of this electrode.

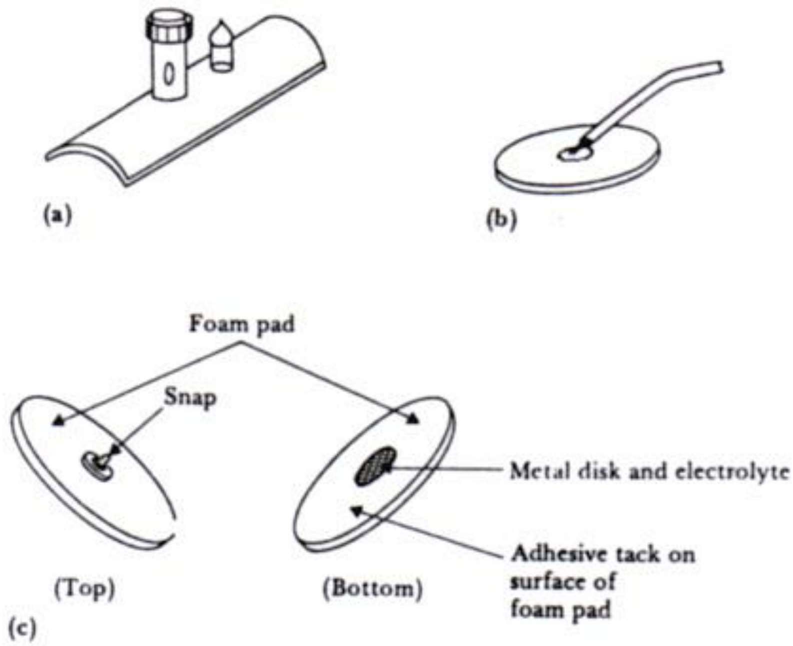

Fig. 7: Body-surface biopotential electrodes: (a) Metal-plate electrode used for application to limbs. (b) Metal-disk electrode applied with surgical tape. (c) Disposable foam-pad electrodes, often used with electrocardiographic monitoring apparatus.

The one most commonly used for limb electrodes with the electrocardiograph is shown in part (a). Before it is attached to the body, its concave surface is covered with electrolyte gel. Part (b) shows the metal disk electrode, which can be used as a chest electrode for recording the ECG. It is also frequently used in cardiac monitoring for long-term recordings. This style of electrode is also popular for surface recordings of EMG or EEG. Part (c) shows a disposable electrode.

(b) Suction electrodes: A modification of the metal-plate electrode that requires no straps or adhesives for holding it in place is the suction electrode illustrated in Fig. 8 (a). Such electrodes are frequently used in electrocardiography as the precordial (chest) leads (Fig. 8 (b)), because they can be placed at particular locations and used to take a recording. They consist of a hollow metallic cylindrical electrode that makes contact with the skin at the base. An appropriate terminal for the lead wire is attached to the metal cylinder, and a rubber suction bulb fits over its other base.

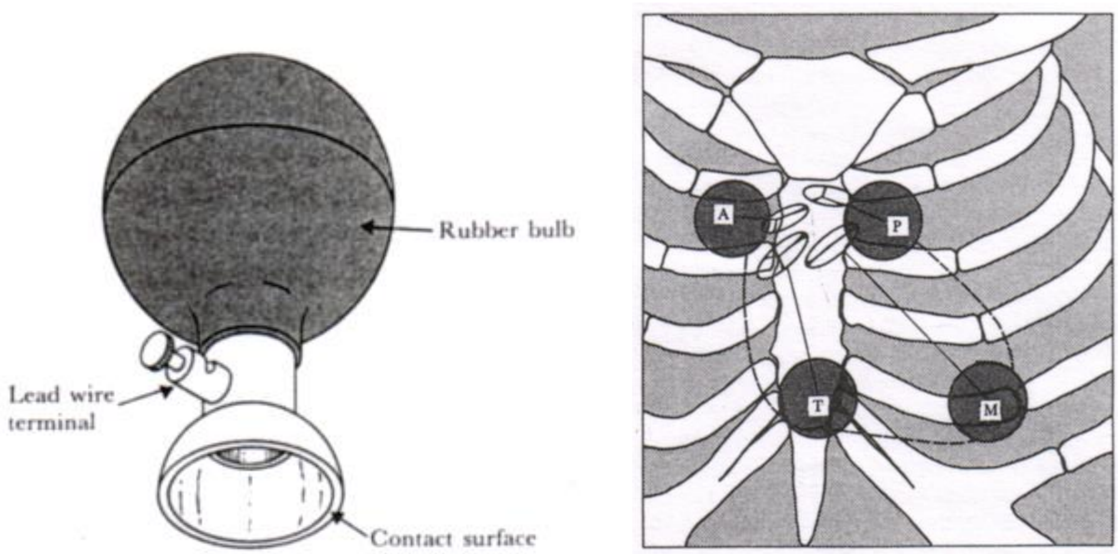

Fig. 8: (a) A metallic suction electrode is often used (b) as a precordial electrode on clinical electrocardiographs.

Electrolyte gel is placed over the contacting surface of the electrode, the bulb is squeezed and the electrode is then placed on the chest wall. The bulb is released and applies suction against the skin, holding the electrode assembly in place. This electrode can be used only for short periods of time; the suction and the pressure of the contact surface against the skin can cause irritation.

(c) Floating electrodes: We know that one source of motion artifact in biopotential electrodes is the double layer of charge at the electrode-electrolyte interface. To reduce this artifact, floating electrodes are used, which offer a suitable technique and stabilize the interface mechanically. Fig. 9(a) depicts a floating electrode known as a top-hat electrode; its internal structure is shown in Fig. 9(b).

The principal feature of the electrode is that the actual electrode element or metal disk is recessed in a cavity so that it does not come in contact with the skin itself. Instead, the element is surrounded by electrolyte gel in the cavity. The cavity does not move with respect to the metal disk, so it does not produce any mechanical movement of the double layer of charge.

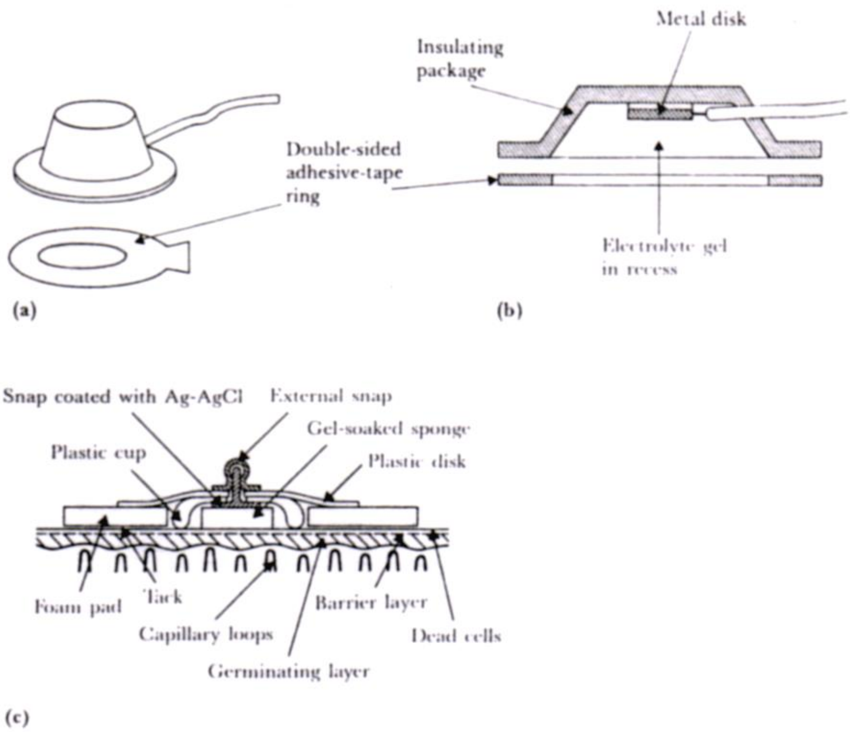

Fig. 9: Examples of floating metal body-surface electrodes: (a) Recessed electrode with top-hat structure. (b) Cross-sectional view of the electrode in (a). (c) Cross-sectional view of a disposable recessed electrode of the same general structure shown in Fig. 7(c). The recess in this electrode is formed from an open foam disk, saturated with electrolyte gel and placed over the metal electrode.

In practice, the electrode is filled with electrolyte gel and then attached to the skin surface by means of a double-sided adhesive-tape ring, as shown in Fig. 9. The electrode element can be a disk made of metal such as silver, and often it is coated with AgCl. Another frequently encountered form of the floating electrode uses a sintered Ag-AgCl pellet instead of a metal disk. These electrodes are found to be quite stable and are suitable for many usage.

(d) Flexible electrodes: Solid electrodes described so far cannot conform to the change in body-surface topography, which can result in additional motion artifact. To avoid such problems, flexible electrodes have been developed [5], examples of which are shown in Fig. 10. Figure 10 (a) shows a technique employed to provide flexible electrodes. A carbon-filled silicone rubber compound in the form of a thin strip or disk is used as the active element of an electrode. A pin connector is pushed into the lead connector hole, and the electrode is used in the same way as a similar type of metal-plate electrode. Flexible electrodes are especially important for monitoring premature infants. Electrodes for detecting the ECG and respiration by the impedance technique are attached to the chest of premature infants, who usually weigh less than 2500 g. Conventional electrodes are not appropriate.

(e) Dry electrodes: All the surface electrodes described so far require an electrolyte gel to establish and maintain contact between the electrode and the skin. Recent advances in solid-state electronic technology have made it possible to record surface biopotentials from electrodes that can be applied directly to the skin without an intermediate layer of electrolyte gel. The significant feature of these electrodes is a self-contained, very-high-input impedance amplifier. An example of a dry-electrode system developed by Kao and Hynecek (1974) is shown in Fig. 11.

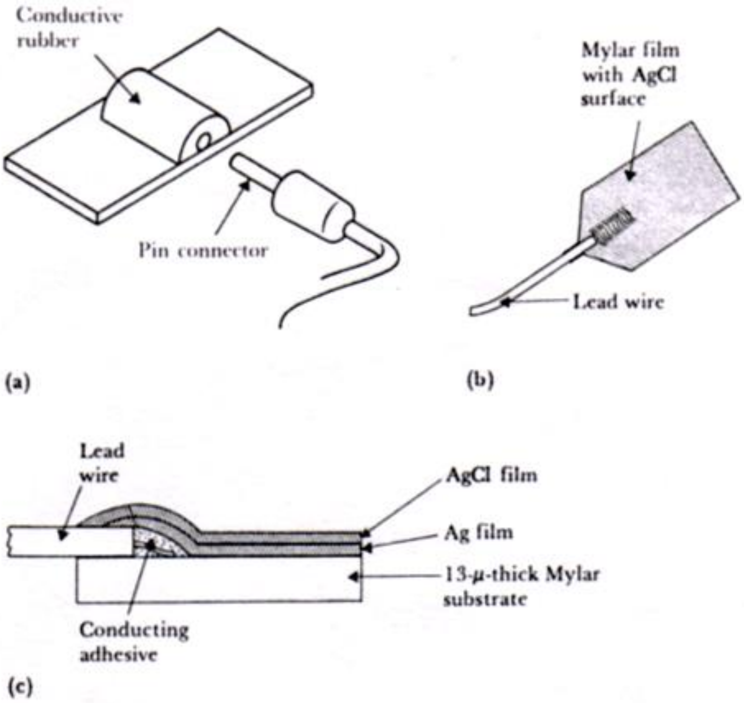

Fig. 10: Flexible Body-Surface Electrodes. (a) Carbon-filled silicone rubber electrode. (b) Flexible thin-film neonatal electrode. (c) Cross-sectional view of the thin-film electrode in (b).

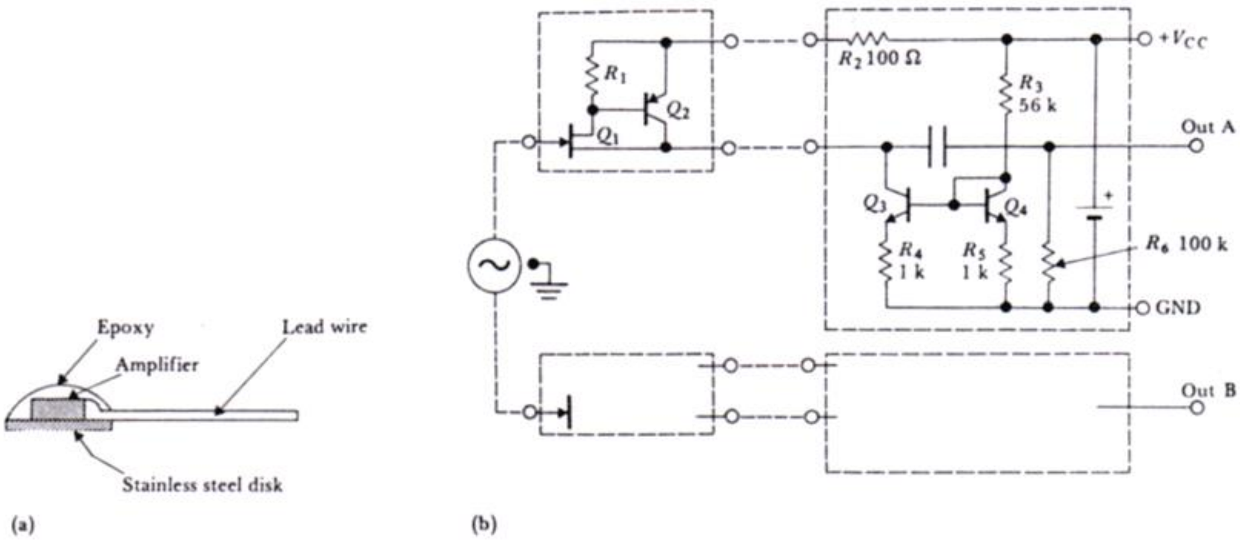

Fig. 11: (a) Dry, active electrode and (b) its amplifier circuit.

2.2.2 Internal Electrodes

Electrodes can also be used within the body to detect biopotentials. They can take the form of (i) *percutaneous electrodes*, in which the electrode itself or the lead wire crosses the skin, or they may be entirely (ii) *internal electrodes*. There are many different designs for internal electrodes. Fig. 12 shows different types of percutaneous needle and wire electrodes.

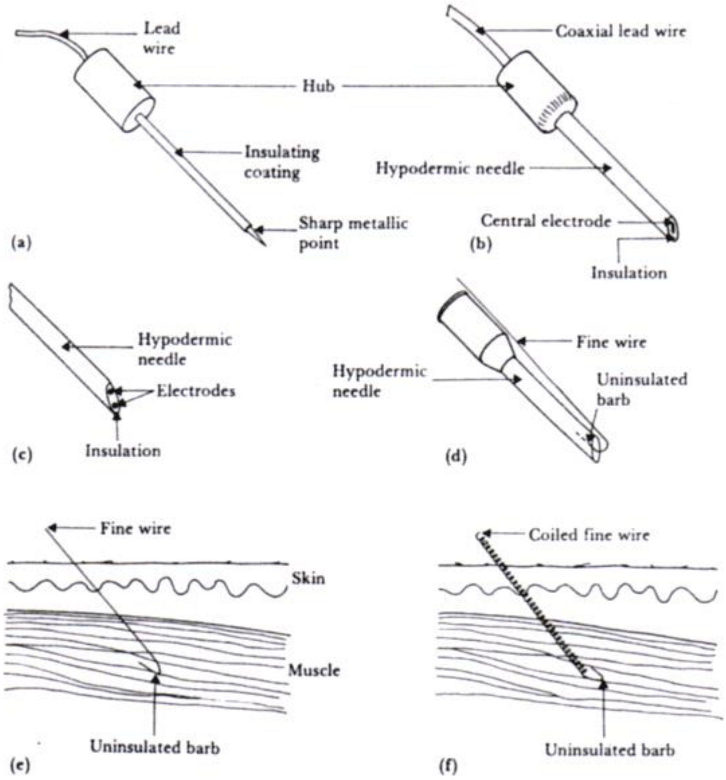

Fig. 12: Needle and wire electrodes for percutaneous measurement of biopotentials: (a) Insulated needle electrode. (b) Coaxial needle electrode. (c) Bipolar coaxial electrode (d) Fine-wire electrode connected to hypodermic needle, before being inserted. (e) Cross-sectional view of skin and muscle, showing coiled fine-wire electrode in place.

Another group of percutaneous electrodes are those used for monitoring fetal heartbeat. In this case it is desirable to get the electrocardiogram from the fetus during labor by direct connection to the presenting part (usually the head) through the uterine cervix (the mouth of the uterus). The fetus lies in a bath of amniotic fluid that contains ions and is conductive, so surface electrodes generally do not provide an adequate ECG as a result of the shorting effect of the amniotic fluid. Thus electrodes used to obtain the fetal ECG must penetrate the skin of the fetus. An example of a suction electrode that does this is shown in Fig. 13 (a). A sharp-pointed probe in the center of a suction cup can be applied to the fetal presenting part, as shown in Fig. 13 (b).

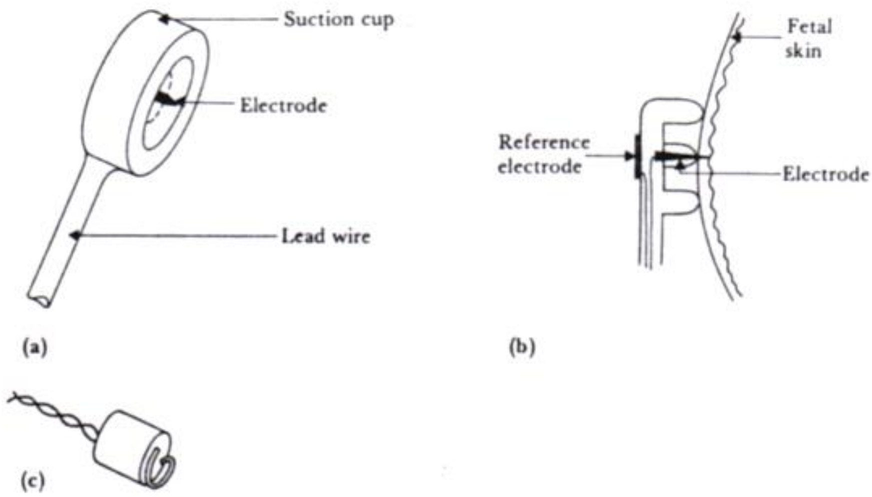

Fig. 13: Electrodes for detecting fetal electrocardiogram during labor, by means of intracutaneous needles. (a) Suction electrode. (b) Cross-sectional view of suction electrode in place, showing penetration of probe through epidermis. (c) Helical electrode, which is attached to fetal skin by corkscrew-type action.

In studying the electrophysiology of excitable cells, it is often important to measure potential differences across the cell membrane. To be able to do this, we must have an electrode within the cell. Such electrodes must be small with respect to the cell dimensions to avoid causing serious cellular injury and thereby changing the cell's behavior. These electrodes are known as microelectrodes and they may be of three types, namely, metal microelectrode , supported metal microelectrodes and micropipet electrodes.

The technology used to produce transistors and integrated circuits can also be used to micromachine small mechanical structures. This technique has been used by several investigators to produce metal microelectrodes. This is essentially a fine needle of a strong metal that is insulated with appropriate insulator up to its tip, as shown in Fig. 14.

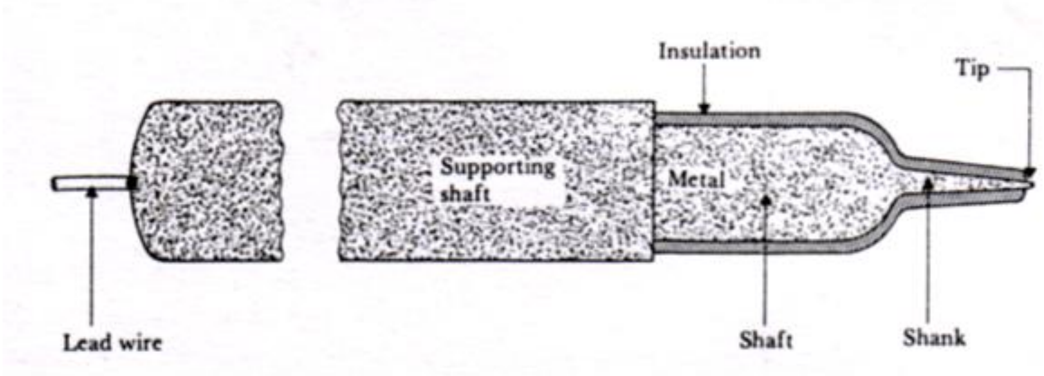

Fig. 14: The structure of a metal microelectrode for intracellular recordings.

On the other hand, in a micropipet electrode, its tip is fabricated in the shape of a pipet. The tip diameter is in the order of 1 μm as shown in Fig. 15.

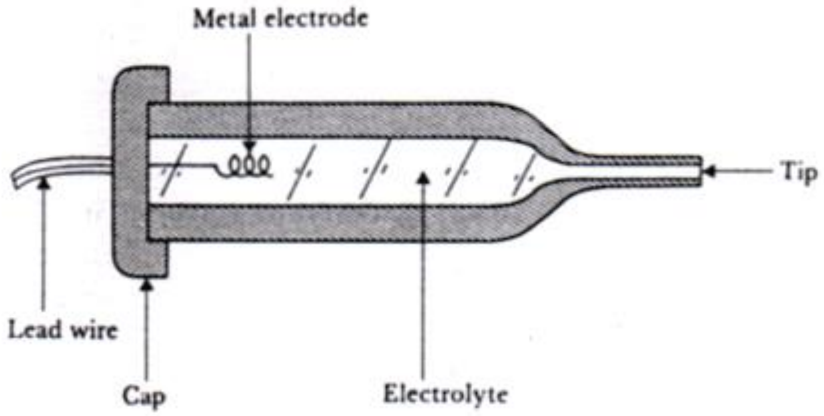

Fig. 15: A glass micropipet electrode filled with an electrolytic solution.

3. Sensors

As mentioned in section 1, the physicians and biomedical researchers are interested in measuring the size, shape, and position of the organs and tissues of the body. Variations in these parameters are important in discriminating normal from abnormal function. Displacement sensors can be used in both direct and indirect systems of measurement. Direct measurements of displacement are used to determine the change in diameter of blood vessels and the changes in volume and shape of cardiac chambers. There are many methods used to convert physiological events to electric signals. Dimensional changes may be measured by variations in resistance, inductance, capacitance, and piezoelectric effect. Thermistors and thermocouples are employed to measure body temperatures. Electromagnetic-radiation sensors include thermal and photon detectors [6].

The displacement-sensitive measurement methods are: resistive, inductive, capacitive, and piezoelectric [1, 2].

3.1 Resistive Sensors

Potentiometers and strain gages are resistive sensors. The potentiometers can measure translational displacements from 2 to 500 mm and some can measure rotational displacements ranging from 10° to more than 50°. The resistance elements (composed of wire-wound, carbon-film, metal-film, conducting-plastic, or ceramic material) may be excited by either dc or ac voltages. These potentiometers produce a linear output (within 0.01% of full scale) as a function of displacement, provided that the potentiometer is not electrically loaded.

In a strain gage, a fine wire (25 μm) is strained within its elastic limit. As strained, the wire's resistance changes because of changes in the diameter, length, and resistivity. The resulting strain gages may be used to measure extremely small displacements, on the order of nanometers. Strain gages can be classified as either unbonded or bonded. An unbonded strain-gage unit is shown in Fig. 16(a). The four sets of strain-sensitive wires are connected to form a Wheatstone bridge, as shown in Fig. 16(b). This type of sensor may be used for converting blood pressure to diaphragm movement, to resistance change, then to an electric signal.

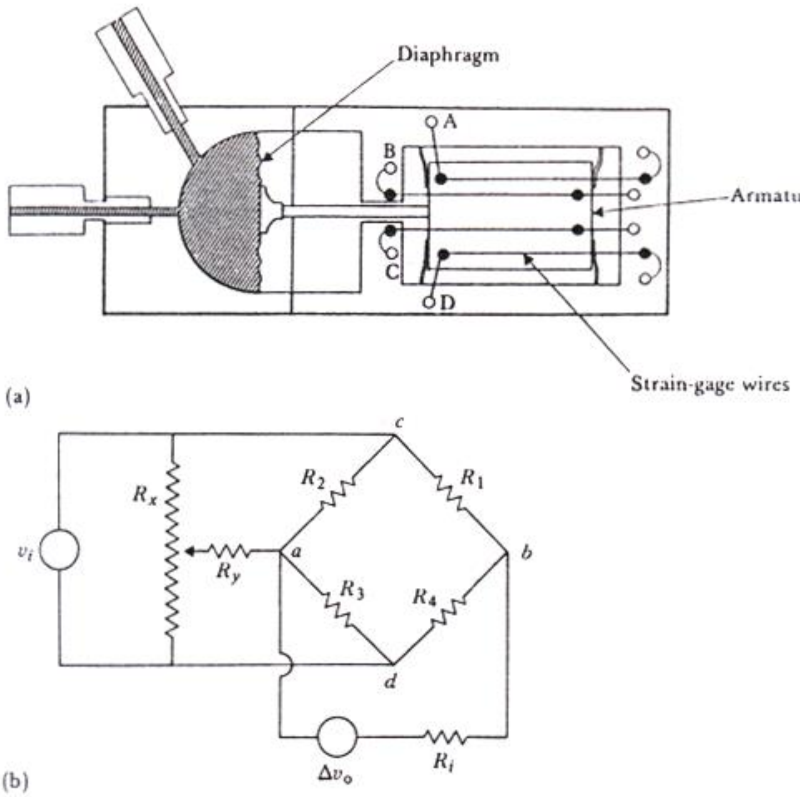

Fig. 16: (a) Unbonded strain-gage pressure sensor. The diaphragm is directly coupled by an armature to an unbonded strain-gage system With increasing pressure, the strain on gage pair B and C is increased, while that on the gage pair A and D is decreased. (b) Wheatstone bridge with four active elements. R_1=B, R_2=A, R_3=D, and R_4=C when the unbonded strain gage is connected for translational motion. Resistor R_y and potentiometer R_x are used to initially balance the bridge. v_i is the applied voltage and Δv_o is the output voltage on a voltmeter or similar device with an internal resistance of R_i.

3.2 Inductive Sensors

An inductance L can be used to measure displacement by varying any three of the coil parameters [Fig. 17]:

$$L = n^2 G \mu \quad \dots\dots\dots (1)$$

where,

n= number of turns of coil
G= geometric form factor
μ =effective permeability of the medium.

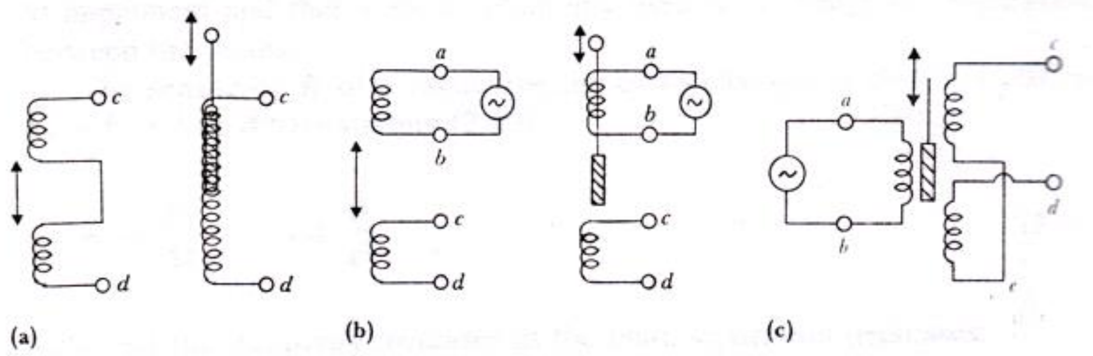

Fig. 17: Inductive displacement sensors (a) self-inductance (b) mutual inductance (c) differential transformer.

An inductive sensor has an advantage in not being affected by the dielectric properties of its environment. However, it may be affected by external magnetic fields due to the proximity of magnetic materials.

3.3 Capacitive Sensors

The capacitance between two parallel plates of area A separated by distance x is

$$C = \varepsilon_0 \varepsilon_r A/x \; \ldots \; \ldots \; \ldots \; (2)$$

where ε_0 is the dielectric constant of free space and ε_r is the relative dielectric constant of the insulator (1.0 for air). In principle it is possible to monitor displacement by changing any of the three parameters ε_r, A, or x. However, the method that is easiest to implement and that is most commonly used is to change the separation between the plates [6]. Fig. 18 shows a capacitance sensor for measuring dynamic displacement changes.

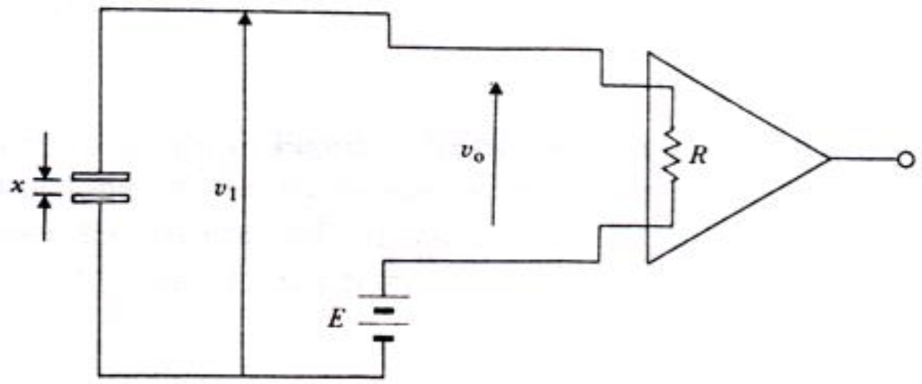

Fig. 18: Capacitance sensor for measuring dynamic displacement changes.

3.4 Piezoelectric Sensors

Piezoelectric sensors are used to measure physiological displacements and record heart sounds. Piezoelectric materials generate an electric potential when mechanically strained, and conversely an electric potential can cause physical deformation of the material. The principle of operation is that, when an asymmetrical crystal lattice is distorted, a charge reorientation takes place, causing a relative displacement of negative and positive charges. The displaced internal charges induce surfaces of opposite polarity on opposite sides of the crystal. Surface charge can be determined by measuring the difference in voltage between electrodes attached to the surfaces [Fig. 19].

The total induced charge q is directly proportional to the applied force F.

$$Q = k\,F \; \ldots \; \ldots \; \ldots \; \ldots \; (3)$$

where k is the piezoelectric constant, C/N.

The change in voltage can be found by assuming that the system acts like a parallel-plate capacitor where the voltage v across the capacitor is charge q divided by capacitance C.

$$V = k\,F/C \quad \dots\dots\dots\dots (4)$$

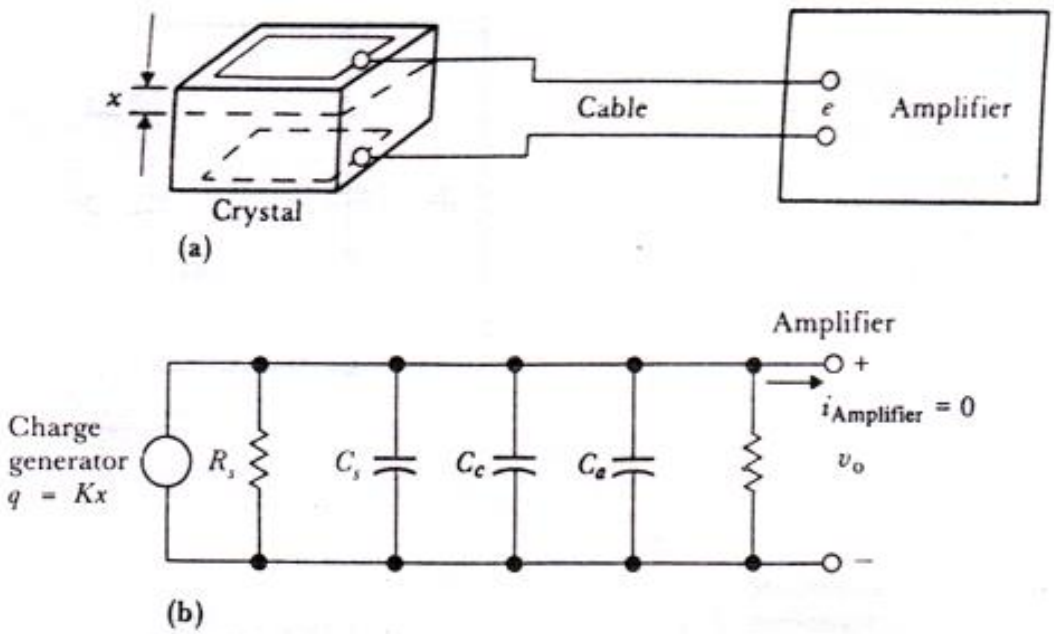

Fig. 19: (a) Piezoelectric sensor (b) Equivalent circuit of piezoelectric sensor, where R_s = sensor leakage resistance, C_s = sensor capacitance, C_c = cable capacitance, C_a = amplifier input capacitance, R_a = amplifier input resistance, and q = charge generator.

Typical values of k are 2.3pC/N for quartz and 140pC/N for barium titanate. For a piezoelectric sensor of 1-cm^2 area and 1-mm thickness with an applied force due to a 10-g weight, the output voltage v is 0.23mV and 14 mV for the quartz and barium titanate crystals, respectively. Piezoelectric materials have a high but finite resistance (on the order of 100GΩ). It is obviously quite important that the input impedance of the external voltage-measuring device be an order of magnitude higher than that of the piezoelectric sensor.

4. Temperature Measurements

A patient's body temperature gives the physician important information about the physiological state of the individual. External body temperature is one of many parameters used to evaluate patients in shock, because the reduced blood pressure of a person in circulatory shock results in low blood flow to the periphery. A drop in the big-toe temperature is a good early clinical warning of shock. Infections on the other hand, are usually reflected by an increase in body temperature, with a hot, flushed skin and loss of fluids. Increased ventilation, perspiration, and blood flow to the skin result when high fevers destroy temperature-sensitive enzymes and proteins. Anesthesia decreases body temperature by depressing the thermal regulatory center. In fact, physicians routinely induce hypothermia in surgical cases in which they wish to decrease a patient's metabolic process and blood circulation [1].

In pediatrics, special heated incubators are used for stabilizing the body temperature of infants. Accurate monitoring of temperature and regulatory control systems are used to maintain a desirable ambient temperature for the infant. In the study of arthritis, physicians have shown that temperatures of joints are closely correlated with the amount of local inflammation. The increased blood flow due to arthritis and chronic inflammation can be detected by thermal measurements. The direct methods of temperature measurement make use of thermocouples, thermistors, and radiation and fiber-optic detectors. Indirect measurement methods, on the other hand, make use of the principle of radiation thermometry.

4.1 Direct Temperature Measurement

The voltage across a p-n junction changes about 2 mV/°C so temperature sensors that use this principle are available in the market [7].

Thermistors are semiconductors made of ceramic materials that are thermal resistors with a high negative temperature coefficient. The resistance of thermistors decreases as temperature increases and increases as temperature decreases. The resistivity of thermistor semiconductors used for biomedical applications is between 0.1 and 100 Ω-m. These devices are small in size (they can be made less than 0.5 mm in diameter), have a relatively large sensitivity to temperature changes (-3 to –5%/°C), and have excellent long-term stability characteristics (± 0.2% of nominal resistance value per year.) The resistance-versus-temperature characteristics of thermistors are not linear.

4.2 Indirect Temperature Measurement (Radiation Thermometry)

The basis of radiation thermometry is that there is a known relationship between the surface temperature of an object and its radiation power. This principle makes it possible to measure the temperature of a body without physical contact with it.

Medical thermography is a technique whereby the temperature distribution of the body is mapped with a sensitivity of a few tenths of a Kelvin. Thermography has been used for the detection of breast cancer, but the method is controversial. It has also been used for determining the location and extent of arthritic disturbances, for gaging the depth of tissue destruction from frostbite and burns, and for detecting various peripheral circulatory disorders (venous thrombosis, carotid-artery occlusions, and so forth).

Every body that is above absolute zero radiates electromagnetic power, the amount being dependent on the body's temperature and physical properties. For objects at room temperature, the spectrum is predominantly in the far- and extreme-far-infrared regions. Infrared detectors and instrument system must be designed with a high sensitivity because of the weak signals. These devices must have a short response time and appropriate wavelength-bandwidth requirements that match the radiation source. Thermal and photon detectors are used as infrared detectors. Suitable instrumentation must be used to amplify, process, and display these weak signals from radiation detectors. Most radiometers make use of a beam-chopper system to interrupt the radiation at a fixed rate (several hundred Hz).

One application of radiation thermometry is an instrument that determines the internal or core body temperature of the human by measuring the magnitude of infrared radiation emitted from the tympanic membrane and surrounding air canal. The tympanic membrane and hypothalamus are perfused by the same vasculature. The hypothalamus is the body's main thermostat, which regulates the core body temperature. Infrared tympanic temperature-monitoring systems require a calibration target in order to maintain their high accuracy.

Thermal sensors and quantum sensors are used as radiation sensors. A thermal sensor absorbs radiation and transforms it into heat, thus causing a rise in temperature in the sensor. Typical thermal sensors are the thermistor and the thermocouple. On the other hand, quantum sensors absorb energy from individual photons and use it to release electrons from the sensor material. Typical quantum sensors are the eye, the phototube, the photodiode, and the photographic emulsion. Such sensors are sensitive over only a restricted band of wavelengths; most respond rapidly. Because none of the common sensors is capable of measuring the radiation emitted by the skin (300K), which has a peak output at 9000 nm, special sensors have been developed, such as InSb sensor.

For cancer therapy or in patient rewarming, a nonmetallic probe is particularly suited for temperature measurement in the strong electromagnetic heating fields used in heating tissue. Fig. 20 shows the details of such a GaAs semiconductor temperature probe coupled with optical fiber.

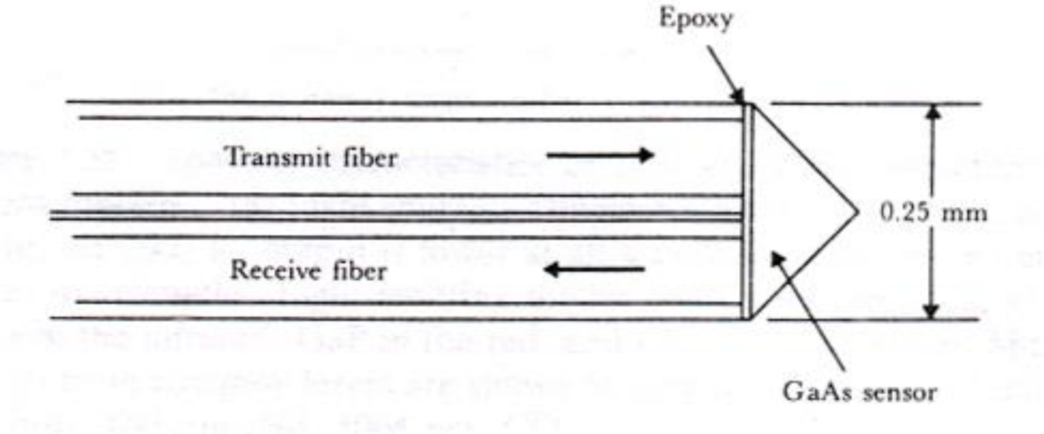

Fig. 20: Details of the fiber/sensor arrangement for the Gas semiconductor temperature probe.

A small prism-shaped sample of single-crystal undoped GaAs is epoxied at the ends of two side-by-side optical fibers. The sensors and fibers can be quite small, compatible with biological implantation after being sheathed. One fiber transmits light from a light-emitting diode source to the sensor, where it is passed through the GaAs and collected by the other fiber for detection in the readout instrument. Some of the optical power traveling through the semiconductor is absorbed, by the process of raising valence-band electrons, across the forbidden energy gap into the conduction band. Because the forbidden energy gap is a sensitive function of the material's temperature, the amount of power absorbed increases with temperature.

5. Ultrasonogram

Ultrasound waves are used to detect the shape and size of the fetus in the mother's womb. We know that pulses of sound waves are used to detect submarines underneath the sea. Similar detection principle is also used here.

Sound and ultrasound follow rules of propagation and reflection similar to those that govern electric signals. Ultrasound transducers use the piezoelectric properties of ceramics such as barium titanate or similar materials. When stressed, these materials produce a voltage across their electrodes. Similarly, when a voltage pulse is applied, the ceramic deforms, If the applied pulse is short, the ceramic element "rings" at its mechanical resonant frequency. With appropriate electronic circuits, the ceramic can be pulsed to transmit a short burst of ultrasonic energy (frequency above 1.0MHz) as a miniature loudspeaker and then switched to act as a microphone receive signals reflected from the interfaces of various tissue types. The gain of the receiver can be varied as a function of time between pulses to compensate for the high attenuation of the tissues.

Muscles and tissues attenuates the ultrasound forward and reflected signals. A 50% decrease occurs through only 2.5cm of muscle. The time delay between the transmitted pulse and its echo is a measure of the depth of the tissue interface. Fine structure of tissues (blood vessels, muscle sheaths, and connective tissue) produce extra echoes within "uniform" tissue structures. At each change of tissue type, a reflection results and that in turn reveal their locations. Fig. 21 shows various ultrasonic transducers and Fig. 22 shows intravascular ultrasonic image of an artery.

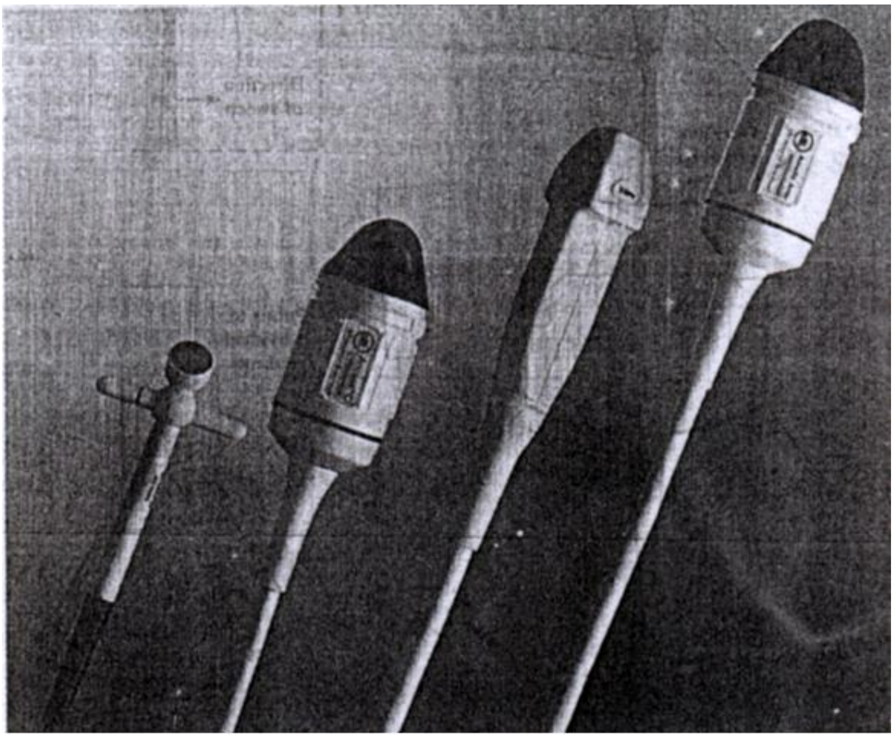

Fig. 21: Different types of ultrasonic transducers range in frequency from 12 MHz for ophthalmic devices to 4 MHz for transducers equipped with a spinning head.

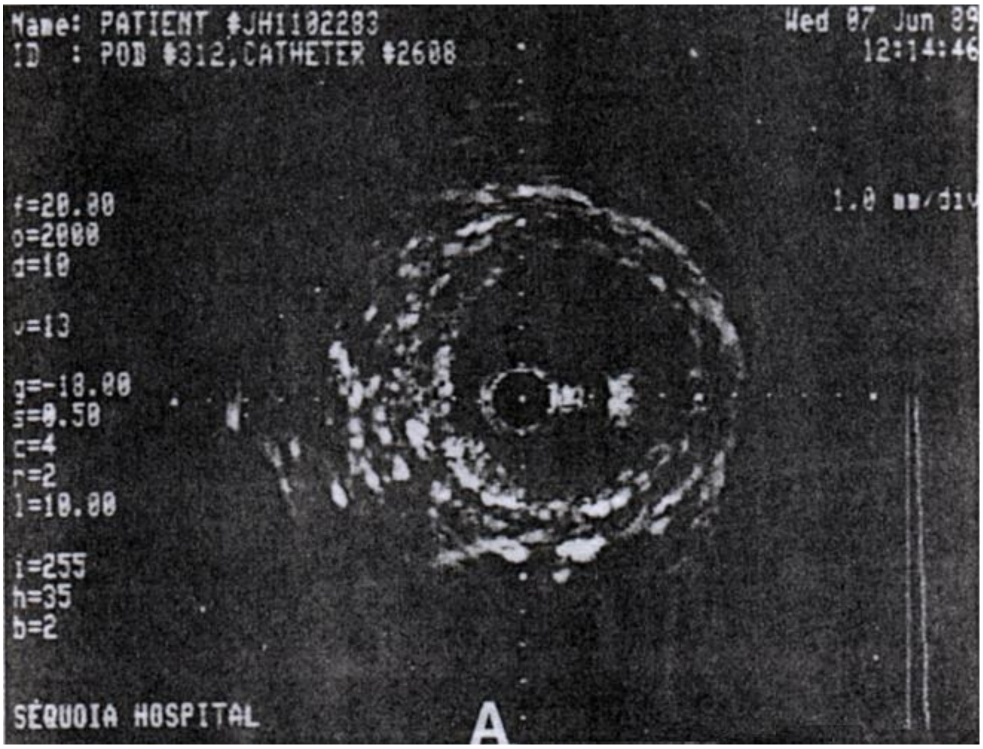

Fig. 22: Intravascular ultrasonic image showing the characteristic three-layer appearance of a normal artery. Mild plaque and calcification can be observed at 7 o'clock.

Apart from biopotential electrodes and sensors, optical systems are also used widely in medical diagnosis [8]. The most common use occurs in the clinical-chemistry lab, in which technicians analyze samples of blood and other tissues removed from the body. Optical instruments are also used during cardiac catheterization to measure the oxygen saturation of hemoglobin and to measure cardiac output.

6. Conclusions

Biomedical sensors couple physiological variables in living and other biological systems to electronic instrumentation for making measurements. This article has described the principles and types of various biopotential electrodes and sensors. Microfabrication technology, such as used in the microelectronics industry, is usually applied in the fabrication of biomedical sensors. Thin- and thick-film processing is especially well-suited to fabricating physical and chemical sensors due to the special properties of these films and the relative low costs for their production compared to other microfabrication technologies. These technologies can yield reproducible, batch-fabricated, and relatively inexpensive sensors that can be applied to biomedical problems in a cost-effective way.

In using metal electrodes for measurement and stimulation, we should understand a few practical points. The first point is the importance of constructing the electrode and any parts of the lead wire that may be exposed to the electrolyte all of the same material. Furthermore, a third material such as solder should not be used to connect the electrode to its lead wire unless it is certain that this material will not be in contact with the electrolyte. When pairs of electrodes are used for measuring differentials, such as in detecting surface potentials on the body or internal potentials within it, it is far better to use the same material for each electrode, because the half-cell potentials are approximately equal. This minimizes possible saturation effects in the case of high gain direct-coupled amplifiers. Electrodes placed on the skin's surface have a tendency to come off. Lead wires to these surface electrodes should be extremely flexible yet strong. It is helpful to provide additional relief from strain by taping the lead wire to the skin approximately 10 cm from the electrode with some slack in the wire between the tape and the electrode. The input impedance of the amplifier to which the electrodes are connected must be much higher than the source impedance represented by the equivalent circuit. If this condition is not met, not only will the amplitude of the recorded signal be less than it should be, but significant distortion also will be introduced into the waveform of the signal.

Finally, it should be emphasized that right form of electrodes and/ or sensors should be used to pick up specific signals from the human body. Operation of instruments in the medical environment imposes important additional constraints. Equipment must be reliable, simple to operate, and capable of withstanding physical abuse and exposure to corrosive chemicals. Electronic equipment must be designed to minimize electric-shock hazards. The safety of patients and medical personnel must be considered in all phases of the design and testing of instruments.

Nearly all biomedical measurements depend either on some form of energy being applied to the living tissue or on some energy being applied as an incidental consequence of sensor operation. X-ray and ultrasonic imaging techniques and electromagnetic or Doppler ultrasonic blood flow meters depend on externally applied energy interfacing with living tissue. Safe levels of these various types of energy are difficult to establish, because many mechanisms of tissue damage are not well understood. A fetus is particularly vulnerable during the early stages of development. The heating of tissue is one effect that must be limited, because even reversible physiological changes can affect measurements. Damage to tissue at the molecular level has been demonstrated in some instances at surprisingly low energy levels.

7. References

[1] Webster J. G. et. al., *"Medical Instrumentation Application and Design"*, 3rd Edition, John Wiley & Sons, Inc. , New York, 2003, pp. 44-87, 183-226.

[2] Cobbold, R. S. C., *Transducers for Biomedical Measurements: Principles and Applications*, New York, Wiley, 1974.

[3] Carin, H. M., "Bioelectrodes" in J. G. Webster (Ed.), *Encyclopedia of Medical Devices and Instrumentation*, New York, Wiley, 1988, pp. 195-226.

[4] Webster J. G., *What is important in Biomedical Electrodes?*, Proc. Annu. Conf. Eng. Med. Biol., 1984.

[5] Neuman, M. R., *Flexible Thin Film Skin Electrodes for Use with Neonates*, Dig. Int. Conf. Med. Biol. Eng., 1973, paper no. 35.11.

[6] Bowman, L., and Meindl, J. D., "Capacitive Sensors,", in J. G. Webstar (Ed.), *Encyclopedia of Medical Devices and Instrumentation*, New York, Wiley, 1988, pp. 551-556.

[7] Re, T. J., and Neuman, M. R., *Thermal Contact-Sensing Electronic Thermometer*, Biomed. Instrum. Technol., 1991, Vol. 25, pp.540-59.

[8] Doebelin, E. O., *Measurement Systems: Application and Design*, 4th Ed., New York: McGraw-Hill, 1990.