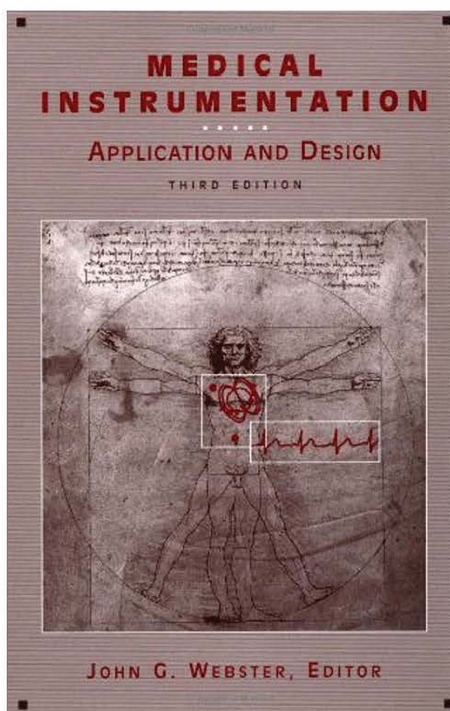




BIOPOTENTIAL ELECTRODES

Michael R. Neuman

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. This interface function is carried out by biopotential 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.

Our first impression is that this is a rather simple function to achieve and that biopotential electrodes should be relatively straightforward. But when we consider the problem in more detail, we see that 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. We shall briefly examine the basic mechanisms involved in the transduction process and shall look at how they affect electrode characteristics. We shall next examine the principal electrical characteristics of biopotential electrodes and discuss electrical equivalent circuits for electrodes based on these characteristics. We shall then cover some of the different forms that biopotential electrodes take in various types of medical instrumentation systems, and finally, we shall look at electrodes used for measuring the ECG, EEG, EMG, and intracellular potentials.

5.1 THE ELECTRODE-ELECTROLYTE INTERFACE

The electrode-electrolyte interface is schematically illustrated in Figure 5.1. A net current that crosses the interface, passing from the electrode to the electrolyte, consists of (1) electrons moving in a direction opposite to that of the current in the electrode, (2) cations (denoted by C^+) moving in the same direction as the current, and (3) anions (denoted by A^-) moving in a direction

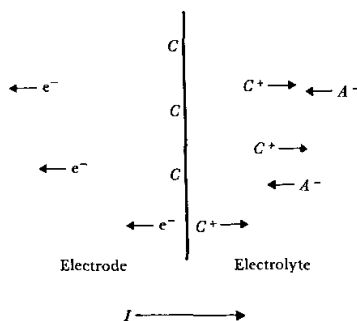


Figure 5.1 Electrode-electrolyte interface The current crosses it from left to right. The electrode consists of metallic atoms C . The electrolyte is an aqueous solution containing cations of the electrode metal C^+ and anions A^- .

opposite to that of the current in the electrolyte. For charge to cross the interface—there are no free electrons in the electrolyte and no free cations or anions in the electrode—something must occur at the interface that transfers the charge between these carriers. What actually occurs are chemical reactions at the interface, which can be represented in general by the following equations:



where n is the valence of C and m is the valence of A . Note that in (5.1) we are assuming that the electrode is made up of some atoms of the same material as the cations and that this material in the electrode at the interface can become oxidized to form a cation and one or more free electrons. The cation is discharged into the electrolyte; the electron remains as a charge carrier in the electrode.

The reaction involving the anions is given in (5.2). In this case an anion coming to the electrode-electrolyte interface can be oxidized to a neutral atom, giving off one or more free electrons to the electrode.

Note that both reactions are often reversible and that reduction reactions (going from right to left in the equations) can occur as well. As a matter of fact, when no current is crossing the electrode-electrolyte interface, these reactions often still occur. But the rate of oxidation reactions equals the rate of reduction reactions, so the net transfer of charge across the interface is zero. When the current flow is from electrode to electrolyte, as indicated in Figure 5.1, the oxidation reactions dominate. When the current is in the opposite direction, the reduction reactions dominate.

To further explore the characteristics of the electrode-electrolyte interface, let us consider what happens when we place a piece of metal into a solution containing ions of that metal. These ions are cations, and the solution, if it is to maintain neutrality of charge, must have an equal number of anions. When the metal comes in contact with the solution, the reaction represented by (5.1) begins immediately. Initially, the reaction goes predominantly either to the left or to the right, depending on the concentration of cations in solution and the equilibrium conditions for that particular reaction. The local concentration of cations in the solution at the interface changes, which affects the anion concentration at this point as well. The net result is that neutrality of charge is not maintained in this region. Thus the electrolyte surrounding the metal is at a different electric potential from the rest of the solution. A potential difference known as the *half-cell potential* is determined by the metal involved, the concentration of its ions in solution, and the temperature, as well as other second-order factors. Knowledge of the half-cell potential is important for understanding the behavior of biopotential electrodes.

The distribution of ions in the electrolyte in the immediate vicinity of the metal-electrolyte interface has been of great interest to electrochemists, and several theories have been developed to describe it. Geddes (1972) compares the charges and potential distributions for four of these theories, whereas Cobbold (1974), in a discussion of the half-cell potential, considers the Stern model. Rather than analyze these theories here, we shall accept their general conclusion. Some sort of separation of charges at the metal-electrolyte interface results in an electric double layer, wherein one type of charge is dominant on the surface of the metal and the opposite charge is distributed in excess in the immediately adjacent electrolyte. This charge distribution at the electrode-electrolyte interface can affect electrode performance as we will see in Section 5.3.

It is not possible to measure the half-cell potential of an electrode because—unless we use a second electrode—we cannot provide a connection between the electrolyte and one terminal of the potential-measuring apparatus. Because this second electrode also has a half-cell potential, we merely end up measuring the difference between the half-cell potential of the metal and that of the second electrode. There would of course be a very large number of combinations of pairs of electrodes, so tabulations of such differential half-cell potentials would be very extensive. To avoid this problem, electrochemists have adopted the standard convention that a particular electrode—the hydrogen electrode—is defined as having a half-cell potential of zero under conditions that are readily achievable in the laboratory. We can then measure the half-cell potentials of all other electrode materials with respect to this electrode.

Table 5.1 lists several common materials that are used for electrodes and gives their half-cell potentials. The table also gives the oxidation-reduction reactions that occur at the surfaces of these electrodes and enable us to arrive

Table 5.1 Half-cell Potentials for Common Electrode Materials at 25 °C

The metal undergoing the reaction shown has the sign and potential E^0 when referenced to the hydrogen electrode.

Metal and Reaction	Potential E^0 , V
$\text{Al} \rightarrow \text{Al}^{3+} + 3\text{e}^-$	-1.706
$\text{Zn} \rightarrow \text{Zn}^{2+} + 2\text{e}^-$	-0.763
$\text{Cr} \rightarrow \text{Cr}^{3+} + 3\text{e}^-$	-0.744
$\text{Fe} \rightarrow \text{Fe}^{2+} + 2\text{e}^-$	-0.409
$\text{Cd} \rightarrow \text{Cd}^{2+} + 2\text{e}^-$	-0.401
$\text{Ni} \rightarrow \text{Ni}^{2+} + 2\text{e}^-$	-0.230
$\text{Pb} \rightarrow \text{Pb}^{2+} + 2\text{e}^-$	-0.126
$\text{H}_2 \rightarrow 2\text{H}^+ + 2\text{e}^-$	0.000 by definition
$\text{Ag} + \text{Cl}^- \rightarrow \text{AgCl} + \text{e}^-$	+0.223
$2\text{Hg} + 2\text{Cl}^- \rightarrow \text{Hg}_2\text{Cl}_2 + 2\text{e}^-$	+0.268
$\text{Cu} \rightarrow \text{Cu}^{2+} + 2\text{e}^-$	+0.340
$\text{Cu} \rightarrow \text{Cu}^+ + \text{e}^-$	+0.522
$\text{Ag} \rightarrow \text{Ag}^+ + \text{e}^-$	+0.799
$\text{Au} \rightarrow \text{Au}^{3+} + 3\text{e}^-$	+1.420
$\text{Au} \rightarrow \text{Au}^+ + \text{e}^-$	+1.680

SOURCE: Data from *Handbook of Chemistry and Physics*, 55th edition, CRC Press, Cleveland, Ohio, 1974-1975, with permission.

at the potentials. The hydrogen electrode is based on the reaction



where H_2 gas bubbled over a platinum electrode is the source of hydrogen molecules. The platinum also serves as a catalyst for the reaction on the left-hand side of the equation and as an acceptor of the generated electrons.

5.2 POLARIZATION

The half-cell potential of an electrode is described in Section 5.1 for conditions in which no electric current exists between the electrode and the electrolyte. If, on the other hand, there is a current, the observed half-cell potential is often altered. The difference is due to polarization of the electrode. The difference between the observed half-cell potential and the equilibrium zero-current half-cell potential is known as the *overpotential*. Three basic mechanisms contribute to this phenomenon, and the overpotential can be separated into three components: the ohmic, the concentration, and the activation overpotentials.

The *ohmic overpotential* is a direct result of the resistance of the electrolyte. When a current passes between two electrodes immersed in an electrolyte, there is a voltage drop along the path of the current in the electrolyte as a result of its resistance. This drop in voltage is proportional to the current and the resistivity of the electrolyte. The resistance between the electrodes can itself vary as a function of the current. Thus the ohmic overpotential does not necessarily have to be linearly related to the current. This is especially true in electrolytes having low concentrations of ions. This situation, then, does not necessarily follow Ohm's law.

The *concentration overpotential* results from changes in the distribution of ions in the electrolyte in the vicinity of the electrode-electrolyte interface. Recall that the equilibrium half-cell potential results from the distribution of ionic concentration in the vicinity of the electrode-electrolyte interface when no current flows between the electrode and the electrolyte. Under these conditions, reactions (5.1) and (5.2) reach equilibrium, so the rates of oxidation and reduction at the interface are equal. When a current is established, this equality no longer exists. Thus it is reasonable to expect the concentration of ions to change. This change results in a different half-cell potential at the electrode. The difference between this and the equilibrium half-cell potential is the concentration overpotential.

The third mechanism of polarization results in the *activation overpotential*. The charge-transfer processes involved in the oxidation-reduction reaction (5.1) are not entirely reversible. In order for metal atoms to be oxidized to metal ions that are capable of going into solution, the atoms must overcome an energy barrier. This barrier, or *activation energy*, governs the kinetics of the reaction. The reverse reaction—in which a cation is reduced, thereby plating out an atom of the metal on the electrode—also involves an activation energy, but it does not necessarily have to be the same as that required for the oxidation reaction. When there is a current between the electrode and the electrolyte, either oxidation or reduction predominates, and hence the height of the energy barrier depends on the direction of the current. This difference in energy appears as a difference in voltage between the electrode and the electrolyte, which is known as the *activation overpotential*.

These three mechanisms of polarization are additive. Thus the net overpotential of an electrode is given by

$$V_p = V_r + V_c + V_a \quad (5.4)$$

where

V_p = total overpotential, or polarization potential, of the electrode

V_r = ohmic overpotential

V_c = concentration overpotential

V_a = activation overpotential

When two aqueous ionic solutions of different concentration are separated by an ion-selective semipermeable membrane, an electric potential exists across this membrane. It can be shown (Plonsey, 1969) that this potential is given by the Nernst equation

$$E = -\frac{RT}{nF} \ln \left(\frac{a_1}{a_2} \right) \quad (5.5)$$

where a_1 and a_2 are the activities of the ions on each side of the membrane. [Other terms are defined in (4.1) and the Appendix.] In dilute solutions, ionic activity is approximately equal to ionic concentration. When intermolecular effects become significant, which happens at higher concentrations, the activity of the ions is less than their concentration.

The half-cell potentials listed in Table 5.1 are known as the standard half-cell potentials because they apply to standard conditions. When the electrode-electrolyte system no longer maintains this standard condition, half-cell potentials different from the standard half-cell potential are observed. The differences in potential are determined primarily by temperature and ionic activity in the electrolyte. *Ionic activity* can be defined as the availability of an ionic species in solution to enter into a reaction.

The standard half-cell potential is determined at a standard temperature; the electrode is placed in an electrolyte containing cations of the electrode material having unity activity. As the activity changes from unity (as a result of changing concentration), the half-cell potential varies according to the Nernst equation

$$E = E^0 + \frac{RT}{nF} \ln (a_{c^{n+}}) \quad (5.6)$$

where

E = half-cell potential

E^0 = standard half-cell potential

n = valence of electrode material

$a_{c^{n+}}$ = activity of cation C^{n+}

Equation (5.6) represents a specific application of the Nernst equation to the reaction of (5.1). The more general form of this equation can be written for a general oxidation-reduction reaction as



where n electrons are transferred. The general Nernst equation for this situation is

$$E = E^0 + \frac{RT}{nF} \ln \left(\frac{a_{\alpha}^{\alpha} a_{\beta}^{\beta}}{a_{\alpha}^{\beta} a_{\beta}^{\alpha}} \right) \quad (5.8)$$

where the a 's represent the activities of the various constituents of the reaction.

An electrode–electrolyte interface is not required for a potential difference to exist. If two electrolytic solutions are in contact and have different concentrations of ions with different ionic mobilities, a potential difference, known as a *liquid-junction potential*, exists between them. For solutions of the same composition but different activities, its magnitude is given by

$$E_j = \frac{\mu_+ - \mu_-}{\mu_+ + \mu_-} \frac{RT}{nF} \ln \left(\frac{a'}{a''} \right) \quad (5.9)$$

where μ_+ and μ_- are the mobilities of the positive and negative ions, and a' and a'' are the activities of the two solutions. Though liquid-junction potentials are generally not so high as electrode–electrolyte potentials, they can easily be of the order of tens of millivolts. For example, two solutions of sodium chloride, at 25 °C, with activities that vary by a factor of 10, have a potential difference of approximately 12 mV. Note that you can generate potentials of the order of some biological potentials by merely creating differences in concentration in an electrolyte. This is a factor to consider when you are examining actual electrode systems used for biopotential measurements.

5.3 POLARIZABLE AND NONPOLARIZABLE ELECTRODES

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 electrolyte. *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 metals such as platinum come closest to behaving as perfectly polarizable electrodes. Because the materials of these electrodes are relatively inert, it is difficult for them to oxidize and dissolve. Thus current passing between the electrode and the electrolyte changes the concentration primarily of ions at the interface, so a majority of the overpotential seen from this type of electrode

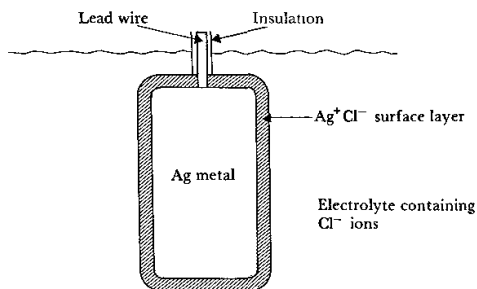


Figure 5.2 A silver/silver chloride electrode, shown in cross section.

is a result of V_c , the concentration overpotential. The electrical characteristics of such an electrode show a strong capacitive effect.

THE SILVER/SILVER CHLORIDE ELECTRODE

The silver/silver chloride (Ag/AgCl) electrode is a practical electrode that approaches the characteristics of a perfectly nonpolarizable electrode and can be easily fabricated in the laboratory. It is a member of a class of electrodes each of which consists of a metal coated with a layer of a 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 Figure 5.2. 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 little chance for any of the surface film on the electrode to dissolve.

The behavior of the Ag/AgCl electrode is governed by two chemical reactions. The first involves the oxidation of silver atoms on the electrode surface to silver ions in solution at the interface.



The second reaction occurs immediately after the formation of Ag^+ ions. These ions combine with Cl^- ions already in solution to form the ionic compound AgCl . As mentioned before, AgCl is only very slightly soluble in water, so most of it precipitates out of solution onto the silver electrode and contributes to the silver chloride deposit. Silver chloride's rate of precipitation and of returning to solution is a constant K , known as the *solubility product*.

Under equilibrium conditions the ionic activities of the Ag^+ and Cl^- ions must be such that their product is the solubility product.

$$a_{\text{Ag}^+} \times a_{\text{Cl}^-} = K_s \quad (5.12)$$

In biological fluids the concentration of Cl^- ions is relatively high, which gives it an activity somewhat less than unity. The solubility product for AgCl , on the other hand, is of the order of 10^{-10} . This means that, when an Ag/AgCl electrode is in contact with biological fluids, the activity of the Ag^+ ion must be very low and of the same order of magnitude as the solubility product.

We can determine the half-cell potential for the Ag/AgCl electrode by writing (5.6) for the reaction of (5.10).

$$E = E_{\text{Ag}}^0 + \frac{RT}{nF} \ln (a_{\text{Ag}^+}) \quad (5.13)$$

By using (5.12), we can rewrite this as

$$E = E_{\text{Ag}}^0 + \frac{RT}{nF} \ln \left(\frac{K_s}{a_{\text{Cl}^-}} \right) \quad (5.14)$$

or

$$E = E_{\text{Ag}}^0 + \frac{RT}{nF} \ln (K_s) - \frac{RT}{nF} \ln (a_{\text{Cl}^-}) \quad (5.15)$$

The first and second terms on the right-hand side of (5.15) are constants; only the third is determined by ionic activity. In this case, it is the activity of the Cl^- ion, which is relatively large and not related to the oxidation of Ag , which is caused by the current through the electrode. The half-cell potential of this electrode is consequently quite stable when it is placed in an electrolyte containing Cl^- as the principal anion, provided the activity of the Cl^- remains stable. Because this is the case in the body, we shall see in later sections of this chapter that the Ag/AgCl electrode is relatively stable in biological applications.

There are several procedures that can be used to fabricate Ag/AgCl electrodes (Janz and Ives, 1968). Two of them are of particular importance in biomedical electrodes. One is the electrolytic process for forming Ag/AgCl electrodes. An electrochemical cell is made up in which the Ag electrode on which the AgCl layer is to be deposited serves as anode and another piece of Ag —having a surface area much greater than that of the anode—serves as cathode. A 1.5-V battery serves as the energy source, and a series resistance limits the peak current, thereby controlling the maximal rate of reaction. A milliammeter can be placed in the circuit to observe the current, which is proportional to the rate of reaction.

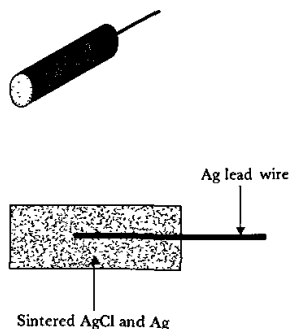


Figure 5.3 Sintered Ag/AgCl electrode

The reactions of (5.10) and (5.11) begin to occur as soon as the battery is connected, and the current jumps to its maximal value. As the thickness of the deposited AgCl layer increases, the rate of reaction decreases and the current drops. This situation continues, and the current approaches zero asymptotically. Theoretically, the reaction is not complete until the current drops to zero. In practice this never occurs because of other processes going on that conduct a current. Therefore the reaction can be stopped after a few minutes, once the current has reached a relatively stable low value—of the order of $10\ \mu\text{A}$ for most biological electrodes.

The second process for producing Ag/AgCl electrodes useful in medical instrumentation is a sintering process that forms pellet electrodes, as shown in Figure 5.3. The electrode consists of an Ag lead wire surrounded by a sintered Ag/AgCl cylinder. It is formed by placing the cleaned lead wire in a die that is then filled with a mixture of powdered Ag and AgCl. The die is compressed in an arbor press to form the powdered components into a pellet, which is then removed from the die and baked at $400\ ^\circ\text{C}$ for several hours. These electrodes tend to have a greater endurance than the electrolytically deposited AgCl electrodes, and they are best applied when repeated usage is necessary. The electrolytically deposited AgCl has a tendency to flake off under mechanical stress, leaving portions of metallic Ag in contact with the electrolyte.

Silver chloride is not a very good conductor of an electric current. If the powder that was compressed to make the sintered electrode consisted only of finely ground silver chloride, this would result in a high-resistance electric connection between the electrolytic solution and the silver wire in the center of the sintered electrode. Electrochemists found that they could increase the conductivity of the silver chloride pellet by including metallic silver powder along with the silver chloride powder. The amount of metallic silver is small enough to make highly unlikely any direct connection from the silver wire to the electrode through silver particles. Instead, there is always some silver

chloride between the silver particles, but the presence of the silver particles makes the current-conduction pathway through the silver chloride much shorter.

A similar situation occurs in the electrolytically prepared silver/silver chloride electrode. Although the silver chloride layer is much thinner in this case than it is for the sintered electrode, it remains a pure silver chloride layer for only a short time after it is deposited. Silver chloride is a silver-halide salt, and these materials are photosensitive. Light striking these salts can cause the silver ions to be reduced to metallic silver atoms. Thus, for all practical purposes, the electrolytically deposited silver chloride layer contains fine silver particles as well. Evidence of their presence appears when the layer is grown and immediately becomes dark gray because of the fine silver particles (pure silver chloride is amber-colored).

In addition to its nonpolarizable behavior, the Ag/AgCl electrode exhibits less electric noise than the equivalent metallic Ag electrodes. Geddes and Baker (1967) showed that electrodes with the AgCl layer exhibited far less noise than was observed when the AgCl layer was removed. Also, a majority of the noise for the purely metallic electrodes was at low frequencies. This would provide the most serious interference for low-frequency recordings, such as the EEG.

A second kind of electrode that has characteristics approaching those of the perfectly nonpolarizable electrode is the *calomel electrode*. It is used primarily as a reference electrode for electrochemical determinations and is frequently applied as the reference electrode when pH is measured (see Section 10.2). The calomel electrode is often constructed as a glass tube with a porous glass plug at its base filled with a paste of mercurous chloride or calomel (Hg_2Cl_2) mixed with a saturated potassium chloride (KCl) solution. Like AgCl, the Hg_2Cl_2 is only slightly soluble in water, so most of it retains its solid form. A layer of elemental mercury is placed on top of the paste layer with an electric lead wire within it. This entire assembly is then positioned in the center of a larger glass tube with a porous glass plug at its base. The tube is filled with a saturated KCl solution so that the Hg_2Cl_2 layer of the inner tube is in contact with this electrolyte through the porous plug of the inner tube. We have a half-cell made up of Hg in intimate contact with an Hg_2Cl_2 layer, which is in contact with the saturated KCl electrolyte. The porous plug at the bottom of the electrode assembly is used to make contact between the internal KCl solution and the solution in which the electrode is immersed. This is actually a liquid-liquid junction that can result in a liquid-liquid junction potential, which will add to the electrode half-cell potential.

Silver/silver chloride electrodes can be fabricated in the same form as the calomel electrode and used for electroanalytical chemical measurements. In this case, the mercury is replaced by silver and AgCl replaces the Hg_2Cl_2 in the electrode structure.

Using the same argument as that used for the Ag/AgCl electrode, we can show that the half-cell potential of this electrode is dependent on the Cl^- activity in the saturated KCl solution. This is stable at a given temperature,

because the solution is saturated. In application, the tip of this electrode assembly that contains the porous plug is dipped into the electrolytic solution that it is to contact. In pH measurements, a pH electrode is also dipped into the solution, and the potential difference between the two electrodes is measured.

5.4 ELECTRODE BEHAVIOR AND CIRCUIT MODELS

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 considerably 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.

The characterization of electrode–electrolyte interfacial impedances has been well reviewed by Geddes (1972), Cobbold (1974), Ferris (1974), and Schwan (1963). It is only summarized here. For sinusoidal inputs, the terminal characteristics of an electrode have both a resistive and a reactive component. Over all but the lowest frequencies, this situation can be modeled as a series resistance and capacitance. We should not be surprised to see a capacitance entering into this model, because the half-cell potential described earlier was the result of the distribution of ionic charge at the electrode–electrolyte interface that had been considered a double layer of charge. This, of course, should behave as a capacitor—hence the capacitive reactance seen for real electrodes.

The series resistance–capacitance equivalent circuit breaks down at the lower frequencies, where this model would suggest an impedance going to infinity as the frequency approaches dc. To avoid this problem, we can convert this series RC circuit to a parallel RC circuit that has a purely resistive impedance at very low frequencies. If we combine this 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 Figure 5.4.

In this circuit, R_d and C_d represent the resistive and reactive components just discussed. These components are still frequency- and current-density-dependent. In this configuration it is also possible to assign physical meaning to the components. C_d represents the capacitance across the double layer of charge at the electrode–electrolyte interface. The parallel resistance R_d represents the leakage resistance across this double layer. All the components of this equivalent circuit have values determined by the electrode material,

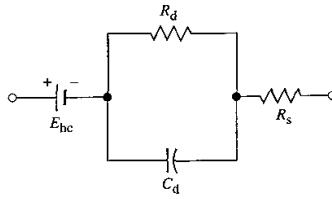


Figure 5.4 Equivalent circuit for a biopotential electrode in contact with an electrolyte. E_{hc} is the half-cell potential, R_d and C_d make up the impedance associated with the electrode-electrolyte interface and polarization effects, and R_s is the series resistance associated with interface effects and due to resistance in the electrolyte.

and—to a lesser extent—by the material of the electrolyte and its concentration.

The equivalent circuit of Figure 5.4 demonstrates that the electrode impedance is frequency-dependent. At high frequencies, where $1/\omega C \ll R_d$, the impedance is constant at R_s . At low frequencies, where $1/\omega C \gg R_d$, the impedance is again constant but its value is larger, being $R_s + R_d$. At frequencies between these extremes, the electrode impedance is frequency-dependent.

The impedance of Ag/AgCl electrodes varies significantly from that of a pure silver electrode at frequencies under 100 Hz. This has been demonstrated by Geddes *et al.* (1969), whose data are reproduced in Figure 5.5.

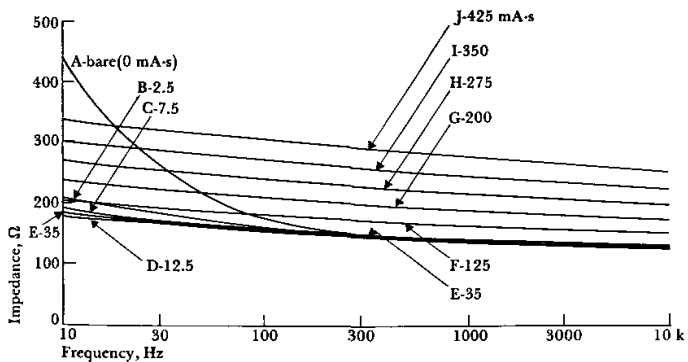


Figure 5.5 Impedance as a function of frequency for Ag electrodes coated with an electrolytically deposited AgCl layer. The electrode area is 0.25 cm^2 . Numbers attached to curves indicate number of $\text{mA} \cdot \text{s}$ for each deposit. (From L. A. Geddes, L. E. Baker, and A. G. Moore, "Optimum Electrolytic Chloriding of Silver Electrodes," *Medical and Biological Engineering*, 1969, 7, pp. 49–56.)

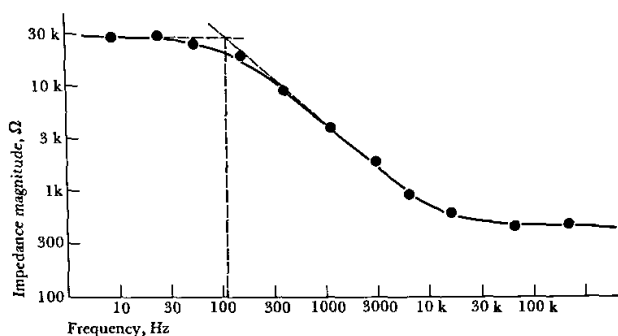


Figure 5.6 Experimentally determined magnitude of impedance as a function of frequency for electrodes.

A metallic silver electrode having a surface area of 0.25 cm^2 had the impedance characteristic shown by curve A. At a frequency of 10 Hz, the magnitude of its impedance was almost three times the value at 300 Hz. This indicates a strong capacitive component to the equivalent circuit. Electrolytically depositing $2.5 \text{ mA} \cdot \text{s}$ of AgCl greatly reduced the low-frequency impedance, as reflected in curve B. Depositing thicker AgCl layers had minimal effects until the charge deposited exceeded approximately $100 \text{ mA} \cdot \text{s}$. The curves were then seen to shift to higher impedances in a parallel fashion as the amount of AgCl deposited increased. Geddes (1972) points out that depositing an AgCl layer using a charge of between 100 and $500 \text{ mA} \cdot \text{s}/\text{cm}^2$ provides the lowest value of electrode impedance. If the current density is maintained at greater than $5 \text{ mA}/\text{cm}^2$, we can adjust current and time to provide the most convenient values for depositing the desired layer.

The impedance decreases with frequency for different electrode materials as shown in Figure 5.6. For 1 cm^2 at 10 Hz, nickel- and carbon-loaded silicone rubber have an impedance of approximately $30 \text{ k}\Omega$, whereas Ag/AgCl has an impedance of less than 10Ω (Das and Webster, 1980).

5.5 THE ELECTRODE–SKIN INTERFACE AND MOTION ARTIFACT

In Section 5.1 we examined the electrode–electrolyte interface and saw how it influenced the electrical properties that are seen in practical electrodes. 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

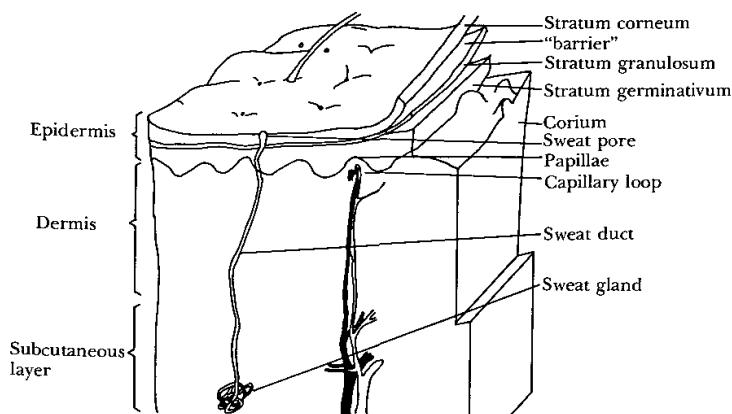


Figure 5.7 Magnified section of skin, showing the various layers (Copyright © 1977 by The Institute of Electrical and Electronics Engineers. Reprinted, with permission, from *IEEE Trans. Biomed. Eng.*, March 1977, vol. BME-24, no. 2, pp. 134–139.)

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. Before we give this explanation, let us briefly review the structure of the skin.

Figure 5.7 shows a cross-sectional diagram of the skin. 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 characteristics on the skin.

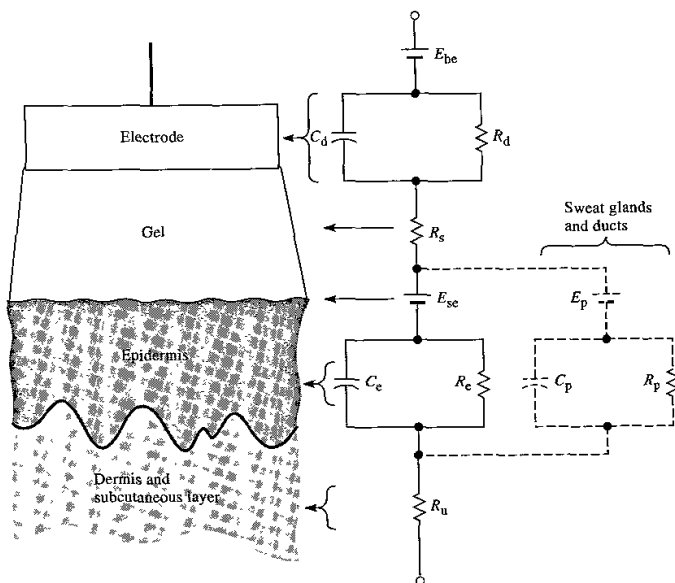


Figure 5.8 A body-surface electrode is placed against skin, showing the total electrical equivalent circuit obtained in this situation. Each circuit element on the right is at approximately the same level at which the physical process that it represents would be in the left-hand diagram.

To represent the electric connection between an electrode and the skin through the agency of electrolyte gel, our equivalent circuit of Figure 5.4 must be expanded, as shown in Figure 5.8. The electrode-electrolyte interface equivalent circuit is shown adjacent to the electrode-gel interface. The series resistance R_g is now the effective resistance associated with interface effects of the gel between the electrode and the skin. We can consider the epidermis, or at least the stratum corneum, as a membrane that is semipermeable to ions, so if there is a difference in ionic concentration across this membrane, there is a potential difference E_{sc} , which is given by the Nernst equation. The epidermal layer is also found to have an electric impedance that behaves as a parallel RC circuit, as shown. For 1 cm^2 , skin impedance reduces from approximately $200 \text{ k}\Omega$ at 1 Hz to 200Ω at 1 MHz (Rosell *et al.*, 1988). The dermis and the subcutaneous layer under it behave in general as pure resistances. They generate negligible dc potentials.

Thus we see that—if the effect of the stratum corneum can be reduced—a more stable electrode will result. We can minimize the effect of the stratum corneum by removing it, or at least a part of it, from under the electrode. There are many ways to do this, ranging from vigorous rubbing with a pad

soaked in acetone to abrading the stratum corneum with sandpaper to puncture it. In all cases, this process tends to short out E_{se} , C_e , and R_e , as shown in Figure 5.8, thereby improving the stability of the signal.

A factor that is sometimes important in examination of, for example, psychogenic electrodermal responses or the galvanic skin reflex (GSR), is the contribution of the sweat glands and sweat ducts. The fluid secreted by sweat glands contains Na^+ , K^+ , and Cl^- ions, the concentrations of which differ from those in the extracellular fluid. Thus there is a potential difference between the lumen of the sweat duct and the dermis and subcutaneous layers. There also is a parallel $R_p C_p$ combination in series with this potential that represents the wall of the sweat gland and duct, as shown by the broken lines in Figure 5.8. These components are often neglected when we consider biopotential electrodes that are not used to measure the electrodermal response or GSR (Boucsein, 1992).

When a polarizable electrode is in contact with an electrolyte, a double layer of charge forms at the interface. If the electrode is moved with respect to the electrolyte, this movement mechanically disturbs the distribution of charge at the interface and results in a momentary change of the half-cell potential until equilibrium can be reestablished. If a pair of electrodes is in an electrolyte and one moves while the other remains stationary, a potential difference appears between the two electrodes during this movement. This potential is known as *motion artifact* and can be a serious cause of interference in the measurement of biopotentials.

Because motion artifact results primarily from mechanical disturbances of the distribution of charge at the electrode–electrolyte interface, it is reasonable to expect that motion artifact is minimal for nonpolarizable electrodes.

Observation of the motion-artifact signals reveals that a major component of this noise is at low frequencies. Section 6.6 and Figure 6.16 will show that different biopotential signals occupy different portions of the frequency spectrum. Figure 6.16 shows that low-frequency artifact does not affect signals such as the EMG or axon action potential (AAP) nearly so much as it does the ECG, EEG, and EOG. In the former case, filtering can be effectively used to minimize the contribution of motion artifact on the overall signal. But in the latter case, such filtering also distorts the signal. Consequently, it is important in these applications to use a nonpolarizable electrode to minimize motion artifact stemming from the electrode–electrolyte interface.

This interface, however, is not the only source of motion artifact encountered when biopotential electrodes are applied to the skin. The equivalent circuit in Figure 5.8 shows that, in addition to the half-cell potential E_{hc} , the electrolyte gel–skin potential E_{se} can also cause motion artifact if it varies with movement of the electrode. Variations of this potential indeed do represent a major source of motion artifact in Ag/AgCl skin electrodes (Tam and Webster, 1977). They have shown that this artifact can be significantly reduced when the stratum corneum is removed by mechanical abrasion with a fine abrasive paper. This method also helps to reduce the epidermal component of the skin impedance. Tam and Webster (1977) also point out, however, that

removal of the body's outer protective barrier makes that region of skin more susceptible to irritation from the electrolyte gel. Therefore, the choice of a gel material is important. Remembering the dynamic nature of the epidermis, note also that the stratum corneum can regenerate itself in as short a time as 24 hours, thereby renewing the source of motion artifact. This is a factor to be taken into account if the electrodes are to be used for chronic recording. A potential between the inside and outside of the skin can be measured (Burbank and Webster, 1978). Stretching the skin changes this skin potential by 5–10 mV, and this change appears as motion artifact. Ten 0.5-mm skin punctures through the barrier layer shortcircuits the skin potential and reduces the stretch artifact to less than 0.2 mV. de Talhouet and Webster (1996) provide a model for the origin of this skin potential and show how it can be reduced by stripping layers of the skin using Scotch tape.

5.6 BODY-SURFACE RECORDING ELECTRODES

Over the years many different types of electrodes for recording various potentials on the body surface have been developed. This section describes the various types of these electrodes and gives an example of each. The reader interested in more extensive examples should consult Geddes (1972).

METAL-PLATE ELECTRODES

One of the most frequently used forms of biopotential sensing electrodes is the metal-plate electrode. In its simplest form, it consists of a metallic conductor in contact with the skin. An electrolyte soaked pad or gel is used to establish and maintain the contact.

Figure 5.9 shows several forms of this electrode. The one most commonly used for limb electrodes with the electrocardiograph is shown in Figure 5.9(a). It consists of a flat metal plate that has been bent into a cylindrical segment. A terminal is placed on its outside surface near one end; this terminal is used to attach the lead wire to the electrocardiograph. A post, placed on this same side near the center, is used to connect a rubber strap to the electrode and hold it in place on an arm or leg. The electrode is traditionally made of German silver (a nickel–silver alloy). Before it is attached to the body, its concave surface is covered with electrolyte gel. Similarly arranged flat metal disks are also used for this type of electrode. Although based upon preceding sections of this chapter one would expect that better electrode designs could be used with electrocardiographs today, these traditional electrodes remain popular and are frequently used.

A second common variety of metal-plate electrode is the metal disk illustrated in Figure 5.9(b). This electrode, which has a lead wire soldered or welded to the back surface, can be made of several different materials. Sometimes the connection between lead wire and electrode is protected by a layer of insulating

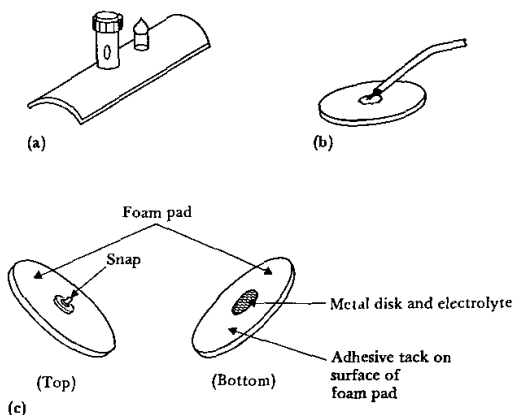


Figure 5.9 Body-surface biopotential electrodes (a) Metal-plate electrode used for application to limbs. (b) Metal-disk electrode applied with surgical tape. (c) Disposable foam-pad electrodes, often used with electrocardiographic monitoring apparatus.

material, such as epoxy or polyvinyl chloride. This structure can be used as a chest electrode for recording the ECG or in cardiac monitoring for long-term recordings. In these applications the electrode is often fabricated from a disk of Ag that may or may not have an electrolytically deposited layer of AgCl on its contacting surface. It is coated with electrolyte gel and then pressed against the patient's chest wall. It is maintained in place by a strip of surgical tape or a plastic foam disk with a layer of tack on one surface.

This style of electrode is also popular for surface recordings of EMG or EEG. In recording EMGs, investigators use stainless steel, platinum, or gold-plated disks to minimize the chance that the electrode will enter into chemical reactions with perspiration or the gel. These materials produce polarizable electrodes, and motion artifact can be a problem with active patients. Electrodes used in monitoring EMGs or EEGs are generally smaller in diameter than those used in recording ECGs. Disk-shaped electrodes such as these have also been fabricated from metal foils (primarily silver foil) and are applied as single-use disposable electrodes. The thinness of the foil allows it to conform to the shape of the body surface. Also, because it is so thin, the cost can be kept relatively low.

Economics necessarily plays an important role in determining what materials and apparatus are used in hospital administration and patient care. In choosing suitable cardiac electrodes for patient-monitoring applications, physicians are more and more turning to pregelled, disposable electrodes with the adhesive already in place. These devices are ready to be applied to the patient and are not cleaned after use. This minimizes the amount of time that personnel must devote to the use of these electrodes.

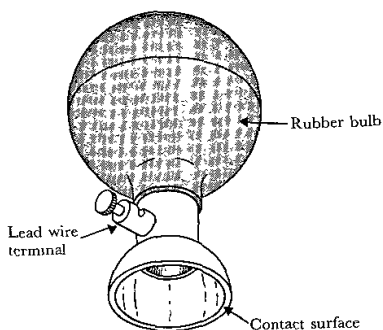


Figure 5.10 A metallic suction electrode is often used as a precordial electrode on clinical electrocardiographs.

A popular type of electrode of this variety is illustrated in Figure 5.9(c). It consists of a relatively large disk of plastic foam material with a silver-plated disk on one side attached to a silver-plated snap similar to that used on clothing in the center of the other side. A lead wire with the female portion of the snap is then snapped onto the electrode and used to connect the assembly to the monitoring apparatus. The silver-plated disk serves as the electrode and may be coated with an AgCl layer. A layer of electrolyte gel covers the disk. The electrode side of the foam is covered with an adhesive material that is compatible with the skin. A protective cover or strip of release paper is placed over this side of the electrode and foam, and the complete electrode is packaged in a foil envelope so that the water component of the gel will not evaporate away. To apply the electrode to the patient, the technician has only to clean the area of skin on which the electrode is to be placed, open the electrode packet, remove the release paper from the tack, and press the electrode against the patient. This procedure is quickly accomplished and no special technique need be learned, such as using the correct amount of gel or cutting strips of adhesive tape to hold the electrode in place.

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 Figure 5.10. Such electrodes are frequently used in electrocardiography as the *precordial* (chest) leads, 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 its base. An appropriate terminal for the lead wire is attached to the metal cylinder, and a rubber suction bulb fits over its other base. 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

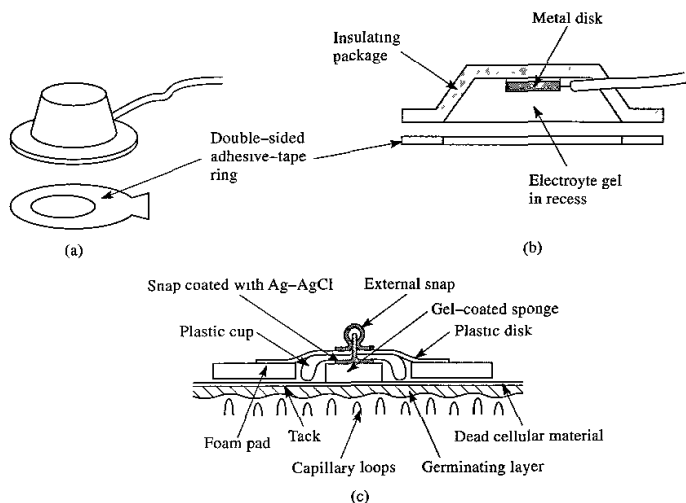


Figure 5.11 Examples of floating metal body-surface electrodes (a) Recessed electrode with top-hat structure. (b) Cross-sectional view of the electrode in (a). (c) Cross-sectional view of a disposable recessed electrode of the same general structure shown in Figure 5.9(c). The recess in this electrode is formed from an open foam disk, saturated with electrolyte gel and placed over the metal 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. Although the electrode itself is quite large, Figure 5.10 shows that the actual contacting area is relatively small. This electrode thus tends to have a higher source impedance than the relatively large-surface-area metal-plate electrodes used for ECG limb electrodes, as shown in Figure 5.9(a).

FLOATING ELECTRODES

In the previous section, we noted that one source of motion artifact in biopotential electrodes is the double layer of charge at the electrode-electrolyte interface. The use of nonpolarizable electrodes, such as the Ag/AgCl electrode, can greatly diminish this artifact. But it still can be present, and efforts to stabilize the interface mechanically can reduce it further. *Floating electrodes* offer a suitable technique to do so.

Figure 5.11 shows examples of these devices. Figure 5.11(a) depicts a floating electrode known as a top-hat electrode; its internal structure is illustrated in cross section in Figure 5.11(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. 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 Figure 5.11. The electrode element can be a disk made of a metal such as silver 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 uses.

A single-use, disposable modification of the floating electrode is shown in cross section in Figure 5.11(c). Its structure is basically the same as that of the disposable metal-plate electrode shown in Figure 5.9(c), but it has one added component—a disk of thin, open-cell foam saturated with electrolyte gel. The foam is firmly affixed to the metal-disk electrode, thereby providing an intermediate electrolyte-gel layer between the electrode and the skin. Because the foam is fixed to the metal disk, the gel contained within it at the disk interface is mechanically stable. The other surface of the foam that is placed against the skin is able to move with the skin, thereby diminishing the motion artifact that sometimes results from differential movement between the skin and the electrolyte gel. Figure 5.11(c) shows that it is possible to lightly abrade through the high-impedance barrier layer of the skin without breaking the deeper capillary loops and drawing blood.

ELECTRODE STANDARDS

During defibrillation, large currents may flow through the electrodes, change the electrode overpotential, and make it difficult to determine whether the defibrillation has been successful. In general Ag/AgCl electrodes are satisfactory, whereas polarizable electrodes are not. Standards for pregelled disposable electrodes (Anonymous, 1984) require face-to-face bench testing to ensure that the offset voltage is less than 100 mV, the noise is less than 150 μ V, the 10-Hz impedance is less than 2 k Ω , the defibrillation overload recovery to four 2-mC charges is less than 100 mV, and the bias current tolerance to 200 nA for 8 h yields less than 100 mV offset (Carim, 1988). The defibrillation recovery voltage versus time for 12 electrode materials shows that the optimal recovery occurs for 500 mC/cm² of AgCl electrodeposited on Ag (Das and Webster, 1980). Additional tests on humans can assess motion artifact, adhesive tack, and skin irritation (Webster, 1984b).

FLEXIBLE ELECTRODES

The electrodes described so far are solid and either are flat or have a fixed curvature. The body surface, on the other hand, is irregularly shaped and can change its local curvature with movement. Solid electrodes cannot conform to this change in body-surface topography, which can result in additional

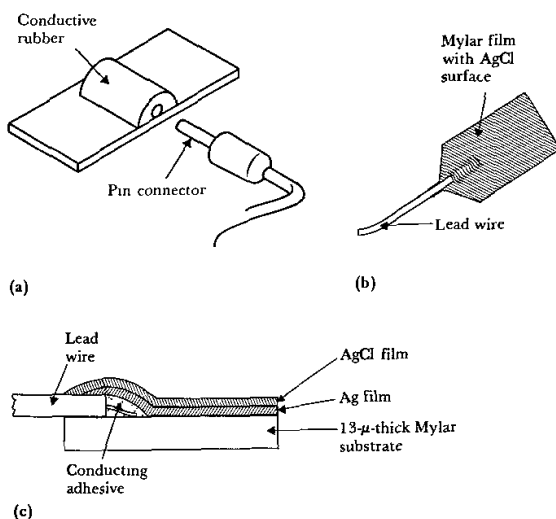


Figure 5.12 Flexible body-surface electrodes (a) Carbon-filled silicone rubber electrode. (b) Flexible thin-film neonatal electrode (after Neuman, 1973). (c) Cross-sectional view of the thin-film electrode in (b). [Parts (b) and (c) are from International Federation for Medical and Biological Engineering. *Digest of the 10th ICMBE*, 1973.]

motion artifact. To avoid such problems, *flexible electrodes* have been developed, examples of which are shown in Figure 5.12.

One type of flexible electrode is a woven, stretchable, nylon fabric impregnated with silver particles. Lead wire bonding is achieved by the use of epoxy. Gel pads are used for short-term monitoring.

Figure 5.12(a) shows another 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; they cannot conform to the shape of the infant's chest and can cause severe skin ulceration at pressure points. They must also be removed when chest x rays are taken of the infant, because they are opaque and can obstruct the view of significant portions of the thoracic cavity. Neuman (1973) developed flexible, thin-film electrodes for use on newborn infants that minimize these problems. The basic electrode consists of a 13- μ m-thick Mylar film on which an Ag and AgCl

film have been deposited, as shown in Figure 5.12(b). The actual structure of the electrode is illustrated in cross section in Figure 5.12(c). The flexible lead wire is attached to the Mylar substrate by means of a conducting adhesive, and a silver film approximately $1\text{ }\mu\text{m}$ thick is deposited over this and the Mylar. An AgCl layer is then grown on the surface of the silver film via the electrolytic process.

In addition to the advantage of being flexible and conforming to the shape of the newborn's chest, these electrodes have a layer of silver thin enough to be essentially x-ray-transparent, so they need not be removed when chest x rays of the infant are taken. All that shows up on the x rays is the lead wire. Consequently, the infant's skin is also protected from the irritation caused by removing and reapplying the adhesive tape that holds the electrode in place. This has been demonstrated to reduce the occurrence of skin irritation significantly in nurseries in which flexible, thin-film electrodes have been used.

The flexible electrodes we have described require some type of adhesive tape to hold them in place against the skin. New electrolytic hydrogel materials have been developed that are in the form of a thin, flexible slab of gelatinous material. This substance has a sticky surface that is similar to the adhesive tack on the tape used to hold electrodes in place. By virtue of the mobile ions that it contains, it is also electrically conductive. A piece of this material the same size as the flexible electrode can be secured on the electrode's surface and used to hold it in place against the skin. Because the electrode and this interface material are both flexible, a good, mechanically secure electric contact can be made between the electrode and the skin. One drawback of this material is its relatively high electric resistance, compared to that of the electrolyte gel routinely used with electrodes. Hydrogels are less effective at hydrating the dry epidermal layer (Jossinet and McAdams, 1990). This is not a severe problem anymore, however, because the amplifiers used with these electrodes now have input impedances of the order of $10\text{ M}\Omega$ or higher, which is much greater than the resistance of the electrolytic material. Often there is less motion artifact when these electrodes are used.

5.7 INTERNAL ELECTRODES

Electrodes can also be used within the body to detect biopotentials. They can take the form of *percutaneous electrodes*, in which the electrode itself or the lead wire crosses the skin, or they may be entirely *internal electrodes*, in which the connection is to an implanted electronic circuit such as a radiotelemetry transmitter. These electrodes differ from body-surface electrodes in that they do not have to contend with the electrolyte-skin interface and its associated limitations, as described in Section 5.5. Instead, the electrode behaves in the way dictated entirely by the electrode-electrolyte interface. No electrolyte gel is required to maintain this interface, because extracellular fluid is present.

There are many different designs for internal electrodes. An investigator

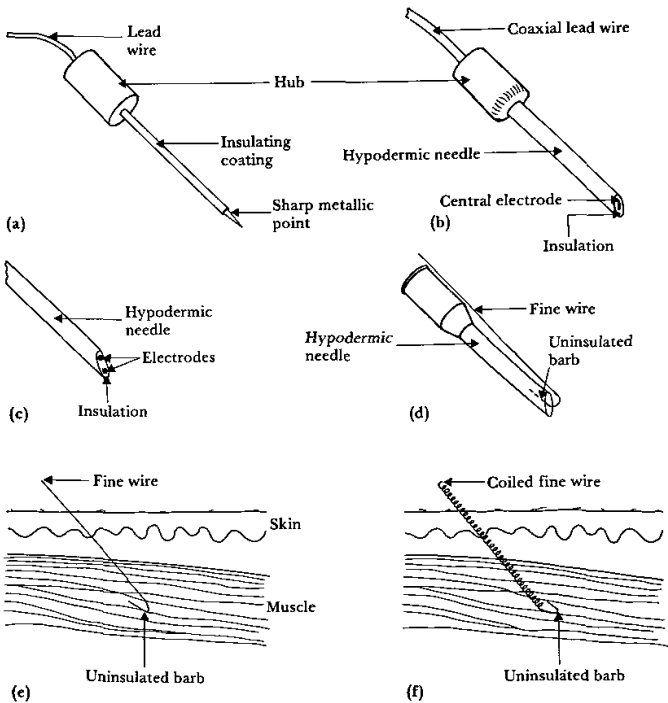


Figure 5.13 Needle and wire electrodes for percutaneous measurement of bio-potentials (a) Insulated needle electrode. (b) Coaxial needle electrode. (c) Bipolar coaxial electrode. (d) Fine-wire electrode connected to hypodermic needle, before being inserted. (e) Cross-sectional view of skin and muscle, showing fine-wire electrode in place. (f) Cross-sectional view of skin and muscle, showing coiled fine-wire electrode in place.

studying a particular bioelectric phenomenon by using internal electrodes frequently designs his or her own electrodes for that specific purpose. The following paragraphs describe some of the more common forms of these electrodes and give examples of their application.

Figure 5.13 shows different types of percutaneous needle and wire electrodes. The basic needle electrode consists of a solid needle, usually made of stainless steel, with a sharp point. The shank of the needle is insulated with a coating such as an insulating varnish; only the tip is left exposed. A lead wire is attached to the other end of the needle, and the joint is encapsulated in a plastic hub to protect it. This electrode, frequently used in electromyography, is shown in Figure 5.13(a). When it is placed in a particular muscle, it obtains an EMG from that muscle acutely and can then be removed.

A variation of this type of electrode is used on patients undergoing surgery to monitor the ECG continuously. The electrode consists of stainless steel hypodermic needles placed subcutaneously on each limb. Lead wires with special connectors attached to the needle at the hub connect the electrodes to the cardioscope (Section 6.9). These electrodes remain in place as the patient is manipulated during the surgical procedure. They are away from the surgical field. Electrolyte gel is not necessary.

A shielded percutaneous electrode can be fabricated in the form shown in Figure 5.13(b). It consists of a small-gage hypodermic needle that has been modified by running an insulated fine wire down the center of its lumen and filling the remainder of the lumen with an insulating material such as an epoxy resin. When the resin has set, the tip of the needle is filed to its original bevel, exposing an oblique cross section of the central wire, which serves as the active electrode. The needle itself is connected to ground through the shield of a coaxial cable, thereby extending the coaxial structure to its very tip.

Multiple electrodes in a single needle can be formed as shown in Figure 5.13(c). Here two wires are placed within the lumen of the needle and can be connected differentially so as to be sensitive to electrical activity only in the immediate vicinity of the electrode tip.

The needle electrodes just described are principally for acute measurements, because their stiffness and size make them uncomfortable for long-term implantation. When chronic recordings are required, percutaneous wire electrodes are more suitable. There are many different types of wire electrodes and schemes for introducing them through the skin. [The interested reader should refer to Geddes (1972) for a more detailed review.] The principle can be illustrated, however, with the help of Figure 5.13(d). A fine wire—often made of stainless steel ranging in diameter from 25 to 125 μm —is insulated with an insulating varnish to within a few millimeters of the tip. This noninsulated tip is bent back on itself to form a J-shaped structure. The tip is introduced into the lumen of the needle, as shown in Figure 5.13(d). The needle is inserted through the skin into the muscle at the desired location, to the desired depth. It is then slowly withdrawn, leaving the electrode in place, as shown in Figure 5.13(e). Note that the bent-over portion of wire serves as a barb holding the wire in place in the muscle. To remove the wire, the technician applies a mild uniform force to straighten out the barb and pulls it out through the wire's tract.

A variation on this basic approach has been described by Caldwell and Reswick (1975). Realizing that wire electrodes chronically implanted in active muscles undergo a great amount of flexing as the muscle moves (which can cause the wire to slip as it passes through the skin and increase the irritation and risk of infection at this point, or even cause the wire to break), they developed the helical spiral electrode shown in Figure 5.13(f). It, too, is made from a very fine insulated wire coiled into a tight helix of approximately 150 μm diameter that is placed in the lumen of the inserting needle. The uninsulated barb protrudes from the tip of the needle and is bent back along the needle before insertion. It holds the wire in place in the tissue when the needle is removed from the muscle. Of course, the external end of the electrode now

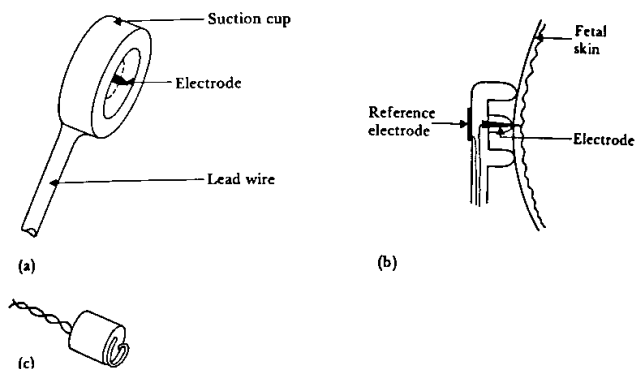


Figure 5.14 Electrodes for detecting fetal electrocardiogram during labor, by means of intracutaneous needles (a) Suction electrode. (b) Cross-sectional view of suction electrode in place, showing penetration of probe through epidermis. (c) Helical electrode, which is attached to fetal skin by corkscrew-type action.

passes through the needle and the needle must be removed—or at least protected—before the electrode is connected to the recording apparatus.

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 Figure 5.14(a). A sharp-pointed probe in the center of a suction cup can be applied to the fetal presenting part, as shown in Figure 5.14(b). When suction is applied to the cup after it has been placed against the fetal skin, the surface of the skin is drawn into the cup and the central electrode pierces the stratum corneum, contacting the deeper layers of the epidermis. On the back of the suction electrode is a reference electrode that contacts the fluid, and the signal seen between these two electrodes is the voltage drop across the resistance of the stratum corneum. Thus, although the amniotic fluid essentially places all the body of the fetus at a common potential, the potentials beneath the stratum corneum can be different, and fetal ECGs that have peak amplitudes of the order of 50–700 μV can be reliably recorded.

Another intradermal electrode that is widely applied for detecting fetal ECG during labor is the helical electrode developed by Hon (1972). It consists of a stainless steel needle, shaped approximately like one turn of a helix,

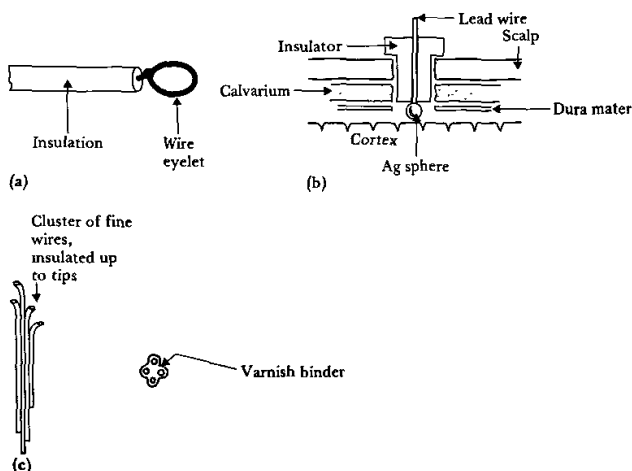


Figure 5.15 Implantable electrodes for detecting biopotentials (a) Wire-loop electrode. (b) Silver-sphere cortical-surface potential electrode. (c) Multielement depth electrode.

mounted on a plastic hub. [See Figure 5.14(c).] The back surface of the hub contains an additional stainless steel reference electrode. When labor has proceeded far enough, this electrode can be attached to the fetal presenting part by rotating it so that the needle twists just beneath the surface of the skin as would a corkscrew shallowly penetrating a cork. This electrode remains firmly attached, and because of the shortness of the helical needle, it does not penetrate deep enough into the skin to cause serious damage. It operates on the same basic principle as the suction electrode.

Often when radio telemetry is used, we want to implant electrodes within the body and not penetrate the skin with the wires. In this case the radio transmitter must be implanted in the body. A wide variety of electrodes are used in this application. Only a few examples are given here.

The simplest electrode for this application is shown in Figure 5.15(a). Insulated multistranded stainless steel wire suitable for implantation has one end stripped so that an eyelet can be formed from the strands of stainless steel. This is best done by individually taking each strand and forming the eyelet either by twisting the wires together one by one at the point at which the insulation stops or by spot-welding each strand to the wire mass at this point. The eyelet can then be sutured to the point in the body at which electric contact is to be established.

Figure 5.15(b) shows another example of an implantable electrode for obtaining cortical-surface potentials from the brain. Critchfield *et al.* (1971) applied this electrode for the radio telemetry of subdural EEGs. The electrode

consists of a 2-mm-diameter silver sphere located at the tip of the cylindrical Teflon insulator through which the electrode lead wire passes. The calvarium is exposed through an incision in the scalp, and a burr hole is drilled. A small slit is made in the exposed dura, and the silver sphere is introduced through this opening so that it rests on the surface of the cerebral cortex. The assembly is then cemented in place onto the calvarium by means of a dental acrylic material.

Deep cortical potentials can be recorded from multiple points using the technique described by Delgado (1964), as shown in Figure 5.15(c). This kind of electrode consists of a cluster of fine insulated wires held together by a varnish binder. Each wire has been cut transversely to expose an uninsulated cross section that serves as the active electrode surface. By staggering the ends of the wires as shown, we can produce electrodes located at known differences in depth in an array. The other ends of the electrodes can be attached to appropriate implantable electronic devices or to a connector cemented on the skull to allow connection to an external recording apparatus.

5.8 ELECTRODE ARRAYS

Although implantable electrode arrays can be fabricated one at a time using clusters of fine insulated wires, this technique is both time-consuming and expensive. Furthermore, when such clusters are made individually, each one will be somewhat different from the other. A way to minimize these problems is to utilize microfabrication technology to fabricate identical two- and three-dimensional electrode arrays. Examples of some of the types of structures that are possible are shown in Figure 5.16. One-dimensional linear arrays of six pairs of biopotential recording electrodes have been described by Mastrototaro *et al.* (1992). These probes are illustrated in Figure 5.16(a) consist of square Ag/AgCl electrodes 40 μm on a side on thin-film gold conductors that have been deposited on either flexible polyimide substrates or more robust molybdenum substrate coated with an anodically grown oxide layer to provide the necessary insulation. The probes were typically 10 mm long, 0.5 mm wide, and 125 μm thick. Lead wires were attached to the bonding pads at the proximal end of the probe. These electrode arrays were designed to be used for measuring transmural potential distributions in the beating myocardium. Their flexibility was important to minimize tissue damage as the muscle contracts and relaxes.

Two-dimensional electrode arrays for mapping the electrical potentials across a region of the surface of an organ such as the heart are shown in Figure 5.16(b). These electrodes essentially represent an extension of the approach used for the one-dimensional arrays described above. A pattern of miniature electrodes is formed on a rigid or flexible surface and connected by conductors to the associated instrumentation. This interconnection can be quite a problem because large arrays require many connections. Sock elec-

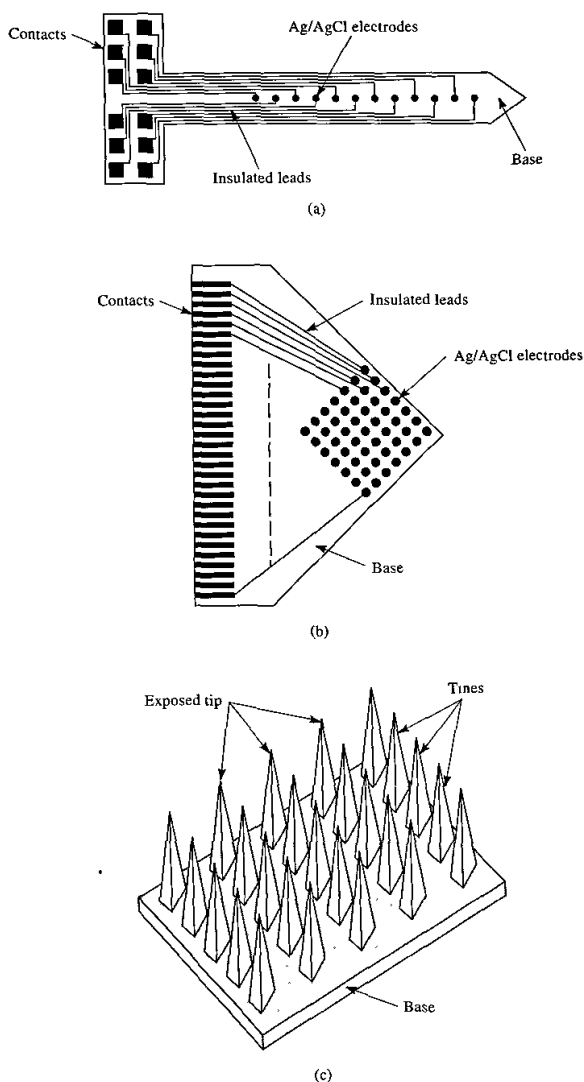


Figure 5.16 Examples of microfabricated electrode arrays. (a) One-dimensional plunge electrode array (after Mastrototaro *et al.*, 1992), (b) Two-dimensional array, and (c) Three-dimensional array (after Campbell *et al.*, 1991).

trodes consisting of individual silver spheres roughly 1 mm in diameter incorporated into a fabric sock that fits snugly over the heart have been used to map epicardial potentials. Each sphere is at the tip of an insulated wire that connects it to the recording apparatus. Needless to say, an array of a large number of electrodes of this type is difficult to build and awkward to use due to the large number of wires coming from the sock.

Ash *et al.* (1992) have shown that this process can be simplified by using a multilayer ceramic integrated circuit package as the electrode array. They have used this structure to map epicardial potentials. These investigators have also used thin-film microfabrication technology to form arrays of 144 miniature Ag/AgCl electrodes on polyimide substrates [Figure 5.16(b)]. The thin gold films serve as conductors as well as the bases for the Ag/AgCl electrodes. The interconnections were completed by using a miniature ribbon cable designed for surface mount microelectronic applications.

Three-dimensional electrode arrays fabricated using silicon microfabrication technology have been described by Campbell *et al.* (1991). Their device has the appearance of a two-dimensional comb [Figure 5.16(c)] with each tine being roughly 1.5 mm long and surrounded with insulating material up to the tip. The exposed tip serves as the electrode, and a wire connection on the base of the structure was needed to make contact with each tine electrode. Although this array is a three-dimensional structure, it really only measures from a two-dimensional array of electrodes because all of the tines are the same length. A truly three-dimensional electrode array can be fabricated by taking a set of one-dimensional electrode array probes such as seen in Figure 5.16(a) or a microelectrode probe as seen in Figure 5.20(b) and assembling them in an array of tines similar in appearance to that of Figure 5.16(c).

5.9 MICROELECTRODES

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. In addition to being small, the electrode used for measuring intracellular potential must also be strong so that it can penetrate the cell membrane and remain mechanically stable.

Electrodes that meet these requirements are known as *microelectrodes*. They have tip diameters ranging from approximately 0.05 to 10 μm . Microelectrodes can be formed from solid-metal needles, from metal contained within or on the surface of a glass needle, or from a glass micropipet having a lumen filled with an electrolytic solution. Examples of each type are given in the following paragraphs. More detailed descriptions can be found in Geddes (1972), Ferris (1974), and Cobbold (1974).

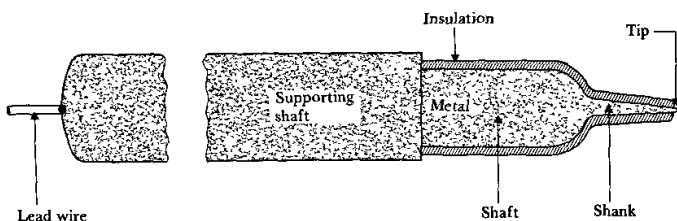


Figure 5.17 The structure of a metal microelectrode for intracellular recordings.

METAL MICROELECTRODES

The metal microelectrode is essentially a fine needle of a strong metal that is insulated with an appropriate insulator up to its tip, as shown in Figure 5.17. A metal needle is prepared in such a way as to produce a very fine tip. This is usually done by electrolytic etching, using a cell in which the metal needle is the anode. The needle is etched as it is slowly withdrawn from the electrolyte solution. Very fine tips can be formed in this way, but a great deal of patience and practice are required to gain the skill to make them. Suitable strong metals for these microelectrodes are stainless steel, platinum-iridium alloy, and tungsten. The compound tungsten carbide is also used because of its great strength.

The etched metal needle is then supported in a larger metallic shaft that can be insulated. This shaft serves as a sturdy mechanical support for the microelectrode and as a means of connecting it to its lead wire. The microelectrode and supporting shaft are usually insulated by a film of some polymeric material or varnish. Only the extreme tip of the electrode remains uninsulated.

SUPPORTED-METAL MICROELECTRODES

The properties of two different materials are used to advantage in supported-metal microelectrodes. A strong insulating material that can be drawn to a fine point makes up the basic support, and a metal with good electrical conductivity constitutes the contacting portion of the electrode.

Figure 5.18 shows examples of supported metal microelectrodes. The classic example of this form is a glass tube drawn to a micropipet structure with its lumen filled with an appropriate metal. Often this type of microelectrode, as shown in Figure 5.18(a), is prepared by first filling a glass tube with a metal that has a melting point near the softening point of the glass. The tube can then be heated to the softening point and pulled to form a narrow constriction. When it is broken at the constriction, two micropipets filled with metal are formed. In this type of structure, the glass not only provides the mechanical support but also serves as the insulation. The active tip is the only metallic area exposed in cross section where the pipet was broken away. Metals such

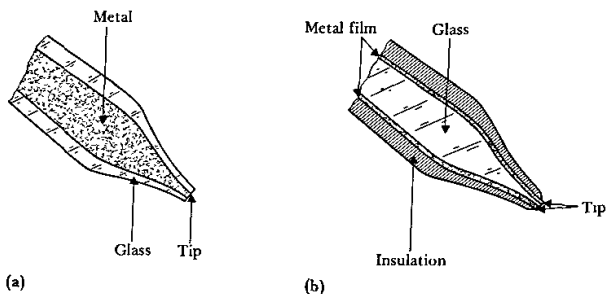


Figure 5.18 Structures of two supported metal microelectrodes (a) Metal-filled glass micropipet. (b) Glass micropipet or probe, coated with metal film.

as silver-solder alloy and platinum and silver alloys are used. In some cases metals with low melting points, such as indium or Wood's metal, are used.

New supported-metal electrode structures have been developed using techniques employed in the semiconductor microelectronics industry. Figure 5.18(b) shows the cross section of the tip of a deposited-metal-film microelectrode. A solid glass rod or glass tube is drawn to form the micropipet. A metal film is deposited uniformly on this surface to a thickness of the order of tenths of a micrometer. A polymeric insulation is then coated over this, leaving just the tip, with the metal film exposed.

MICROPIPET ELECTRODES

Glass micropipet microelectrodes are fabricated from glass capillaries. The central region of a piece of capillary tubing, as shown in Figure 5.19(a), is heated with a burner to the softening point. It is then rapidly stretched to produce the constriction shown in Figure 5.19(b). Special devices, known as microelectrode pullers, that heat and stretch the glass capillary in a uniform reproducible way to fabricate micropipets are commercially available. The two halves of the stretched capillary structure are broken apart at the constriction to produce a pipet structure that has a tip diameter of the order of $1\text{ }\mu\text{m}$. This pipet is fabricated into the electrode form shown in Figure 5.19(c). It is filled with an electrolyte solution that is frequently $3M\text{ KCl}$. A cap containing a metal electrode is then sealed to the pipet, as shown. The metal electrode contacts the electrolyte within the pipet. The electrode is frequently a silver wire prepared with an electrolytic AgCl surface. Platinum or stainless steel wires are also occasionally used.

MICROELECTRODES BASED ON MICROELECTRONIC TECHNOLOGY

The technology used to produce transistors and integrated circuits can also be used to micromachine small mechanical structures. This technique has been

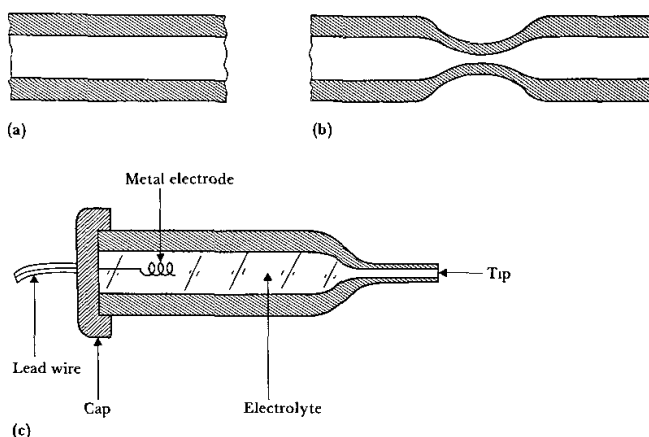


Figure 5.19 A glass micropipet electrode filled with an electrolytic solution (a) Section of fine-bore glass capillary. (b) Capillary narrowed through heating and stretching. (c) Final structure of glass-pipet microelectrode.

used by several investigators to produce metal microelectrodes. The structure shown in Figure 5.20(a) uses the technology for fabricating beam-lead transistors (Wise *et al.*, 1970). The basic structure consists of narrow gold strips deposited on a silicon substrate the surface of which has been first insulated by growing an SiO_2 film. The gold strips are then further insulated by depositing SiO_2 over their surface. The silicon substrate is next etched to a thin, narrow structure that is just wide enough to accommodate the gold strips in the region of the tip. The silicon substrate is etched a millimeter or two back from the tip so that only the gold strips and their SiO_2 insulation remain. The insulation is etched away from the very tip of the gold strips to expose the contacting surface of the electrodes. Although this technology cannot produce tips as small as can be produced with the glass micropipet technique previously described, it is possible to make multielectrode arrays and to maintain very precisely the geometry between individual electrodes in the array. The high reproducibility of microelectronic processing allows many electrodes to be made that all have very similar geometric properties. Thus the characteristics vary little from one electrode array to the next.

Several other designs for sensing and stimulating electrodes have been developed through the use of microelectronic technology. An array of multisite microelectrodes can be grown on a thin silicon probe [Figure 5.20(b)] that can be placed in the cortex of the brain to detect local potentials (Drake *et al.*, 1988). A similar design was utilized by Prohaska *et al.* (1986), but here the actual gold or silver/silver chloride electrode was located in a very small chamber filled with an electrolytic solution such as sodium chloride and was

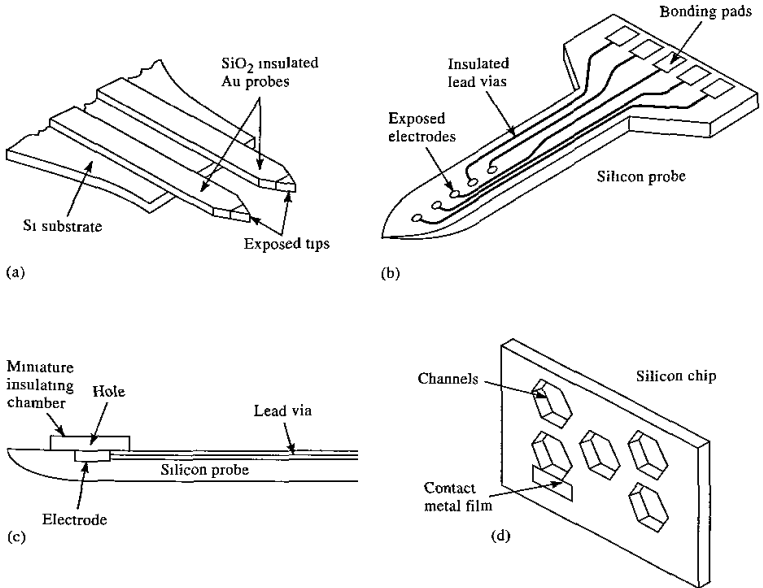


Figure 5.20 Different types of microelectrodes fabricated using microelectronic technology (a) Beam-lead multiple electrode. (Based on Figure 7 in K. D. Wise, J. B. Angell, and A. Starr, "An Integrated Circuit Approach to Extracellular Microelectrodes." Reprinted with permission from *IEEE Trans. Biomed. Eng.*, 1970, BME-17, pp. 238–246. Copyright © 1970 by the Institute of Electrical and Electronics Engineers.) (b) Multielectrode silicon probe after Drake *et al.* (c) Multiple-chamber electrode after Prohaska *et al.* (d) Peripheral-nerve electrode based on the design of Edell.

made from an insulating film with a small hole to allow communication with the nervous tissue in which it was placed [Figure 5.20(c)]. A novel new electrode for sensing signals emitted from peripheral nerves has been developed (Edell, 1986). His electrode consisted of an array of channels etched through a silicon chip [Figure 5.20(d)]. He used these electrodes in animal studies wherein peripheral nerves were transected and each side of the cut nerve was aligned on opposite sides of the silicon chip so that the nerve could regenerate and grow through the channels on the chip to reestablish the connections. Gold metalization on the silicon surface surrounding each channel was used to make electric contact with the nerve fibers that passed through the channels on the silicon chip.

ELECTRICAL PROPERTIES OF MICROELECTRODES

To understand the electrical behavior of microelectrodes, we must derive an electrical equivalent circuit from physical considerations. This circuit differs for metal and micropipet electrodes.

Figure 5.21 shows metal microelectrodes. The microelectrode contributes

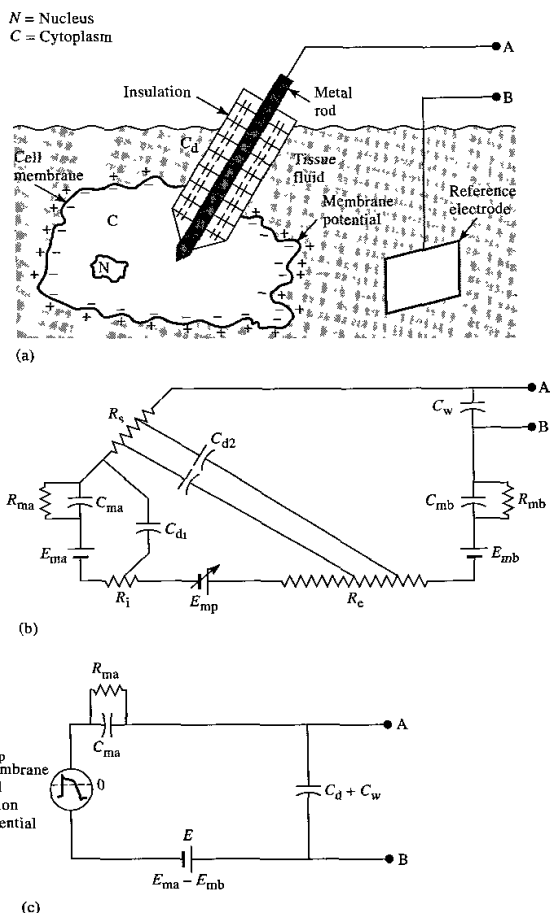


Figure 5.21 Equivalent circuit of metal microelectrode (a) Electrode with tip placed within a cell, showing origin of distributed capacitance. (b) Equivalent circuit for the situation in (a). (c) Simplified equivalent circuit. (From L. A. Geddes, *Electrodes and the Measurement of Bioelectric Events*, Wiley-Interscience, 1972. Used with permission of John Wiley and Sons, New York.)

a series resistance R_s that is due to the resistance of the metal itself. A major contributor to this resistance is the metal in the shank and tip portion of the microelectrode, because the ratio of length to cross-sectional area is much higher in this portion than it is for the shaft. The metal is coated with an insulating material over all but its most distal tip, so a capacitance is set up between the metal and the extracellular fluid. This is a distributed capacitance C_d that we can represent in lumped form by separating the shank and tip from the shaft. In the shank region, we can consider the microelectrode to be a coaxial cylinder capacitor; the capacitance per unit length (F/m) is given by

$$\frac{C_{d1}}{L} = \frac{2\pi\epsilon_r\epsilon_0}{\ln(D/d)} \quad (5.16)$$

where

- ϵ_0 = dielectric constant of free space (Appendix A.1)
- ϵ_r = relative dielectric constant of insulation material
- D = diameter of cylinder consisting of electrode plus insulation
- d = diameter of electrode
- L = length of shank

Of course, this coaxial-cable approximation is not a very good approximation for the shank region, which is tapered, but it is reasonable for a rough calculation. Because insulation thicknesses are usually on the order of $1\text{ }\mu\text{m}$ in the shank and tip, it is important to consider the structure using the coaxial cylinder analog. However, when we consider the shaft portion of the electrode, if the thickness of the insulation is still approximately $1\text{ }\mu\text{m}$, the diameter of the metal shaft can be on the order of several millimeters. Here the ratio of diameters would be practically unity, so we can simplify the calculation by unwrapping the circumferential surface of the shaft and considering the system to be a parallel-plate capacitor of area equal to the circumferential surface area and of thickness equal to t , the thickness of the insulation layer. The capacitance per unit length (F/m) is given by

$$\frac{C_{d2}}{L} = \frac{\epsilon_r\epsilon_0\pi d}{t} \quad (5.17)$$

Note that this capacitance comes from only that portion of the electrode shaft that is submerged in the extracellular fluid. Often only the shank is submerged, so C_{d2} is zero.

The other significant contributions to the equivalent circuit from the metal microelectrode are the components contributed by the metal-electrolyte interface, R_{ma} , C_{ma} , and E_{ma} . A similar set of components, C_{mb} , R_{mb} , and E_{mb} , are associated with the reference electrode. Because of the much larger surface

area of the reference electrode compared with the tip of the microelectrode, the impedance due to these components is much lower. Of course, the half-cell potential due to the reference electrode is unaffected by the surface area. The tip of the microelectrode is within a cell, so there is a series resistance R_i associated with the electrolyte within the cell membrane and another series resistance R_e due to the extracellular fluid. The cell membrane itself can be modeled simply as a variable potential E_{mp} , but in more detailed analyses an equivalent circuit of greater complexity is required. Some of the distributed capacitance of the shank, C_{d1} , is between the microelectrode and the extracellular fluid, as shown in the equivalent circuit, whereas the remainder of it is between the microelectrode and the intracellular fluid.

There is also a capacitance associated with the lead wires, C_w . The physical basis for this equivalent circuit is shown in Figure 5.21(a); the actual equivalent circuit is shown in Figure 5.21(b). Often it is acceptable to simplify this equivalent circuit to that shown in Figure 5.21(c), which neglects the impedance of the reference electrode and the series-resistance contribution from the intracellular and extracellular fluid and lumps all the distributed capacitance together. Under circumstances in which the input impedance of the amplifier connected to this electrode is not sufficiently large, we see that this circuit can behave as a high-pass filter and significant waveform distortion can result.

The effective impedance of metal microelectrodes is frequency-dependent and can be of the order of 10 to 100 M Ω . We can, however, lower this impedance by increasing the effective surface area of the tip of the microelectrode through the application of platinum black, as we did in the case of the hydrogen electrode. Impedance reduction of one or two orders of magnitude can be achieved in this way. At lower frequencies, the impedance can be reduced by applying an Ag/AgCl surface to the electrode tip. Care must be taken in doing this, however, because of the mechanically fragile nature of this film and its tendency to flake off.

The equivalent circuit for the micropipet electrode is somewhat more complicated than that of the metal microelectrode. The physical situation is illustrated in Figure 5.22(a), and the resulting equivalent circuit is shown in Figure 5.22(b). The internal electrode in the micropipet gives the metal-electrolyte interface components R_{ma} , C_{ma} , and E_{ma} . In series with this is a resistive element R_i corresponding to the resistance of the electrolyte in the shank and tip region of the microelectrode. Connected to this is the distributed capacitance C_d corresponding to the capacitance across the glass in this region. The distributed capacitance due to the shaft region has been neglected, because the glass wall of the electrode is much thicker in this region and the capacitive contribution is quite small.

There are two potentials associated with the tip of the micropipet microelectrode. The *liquid-junction potential* E_j corresponds to the liquid junction set up between the electrolyte in the micropipet and the intracellular fluid. In addition, a potential known as the *tip potential* E_t arises because the thin glass wall surrounding the tip region of the micropipet behaves like a glass membrane and has an associated membrane potential.

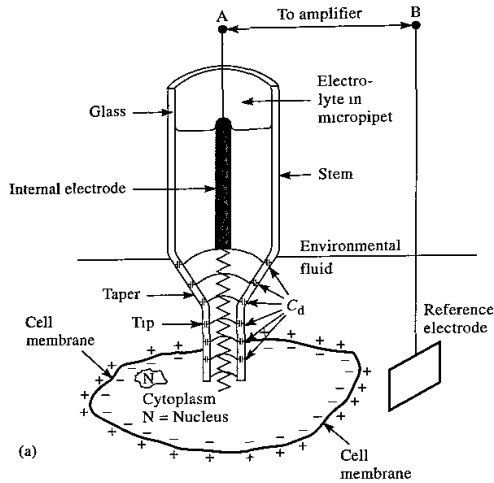
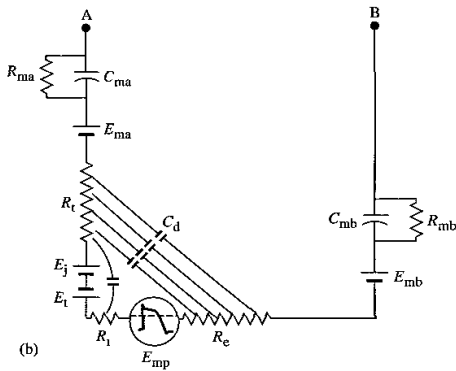
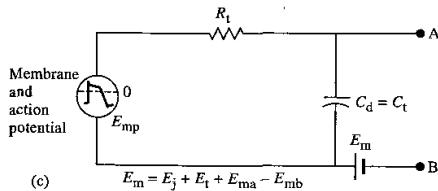


Figure 5.22 Equivalent circuit of glass micropipet microelectrode (a) Electrode with its tip placed within a cell, showing the origin of distributed capacitance. (b) Equivalent circuit for the situation in (a). (c) Simplified equivalent circuit. (From L. A. Geddes, *Electrodes and the Measurement of Bioelectric Events*, Wiley-Interscience, 1972. Used with permission of John Wiley and Sons, New York.)

The equivalent circuit also includes resistances corresponding to the intracellular R_i and extracellular R_e fluids. These are coupled to the microelectrode through the distributive capacitance C_d , as is the case for the metal microelectrode. The equivalent circuit for the reference electrode remains unchanged from that shown in Figure 5.21(b).

Unlike the metal microelectrode, the micropipet's major impedance contribution is resistive. This can be illustrated by approximating the equivalent circuit to give that shown in Figure 5.22(c). Here the overall series resistance of the electrode is lumped together as R_i . This resistance generally ranges in value from 1 to 100 M Ω . The total distributed capacitance is lumped together to form C_i , which can be on the order of tens of picofarads. And all the associated dc potentials are lumped together in the source E_m , which is given by

$$E_m = E_j + E_i + E_{m_2} - E_{m_1} \quad (5.18)$$

Note that the micropipet-type microelectrode behaves as a low-pass filter. The high series resistance and distributed capacitance cause the electrode output to respond slowly to rapid changes in cell-membrane potential. To reduce this problem, positive-feedback, negative-capacitance amplifiers (see Section 6.6) are used to reduce the effective value of C_i .

5.10 ELECTRODES FOR ELECTRIC STIMULATION OF TISSUE

Electrodes used for the electric stimulation of tissue follow the same general design as those used for the recording of bioelectric potentials. They differ in that currents as large as milliamperes cross the electrode-electrolyte interface in stimulating electrodes. Examples of specific electrodes used in cardiac pacemakers, other functional electric stimulators, and cardiac defibrillators (where currents are even larger) are given in Chapter 13. Other types of stimulating electrodes are of the same form as the potential recording electrodes described in this chapter.

In considering stimulating electrodes, we must bear in mind that the net current across the electrode-electrolyte interface is not always zero. When a biphasic stimulating pulse is used, the average current over long periods of time should be zero. However, over the stimulus cycle, there are periods of time during which the net current across the electrode is in one direction at one time and in the other direction at a different time. Also, the magnitudes of the currents in the two directions may be unequal. In studying the electrical characteristics of the electrode-electrolyte interface under such circumstances, we may well imagine that the equivalent circuit changes as the stimulus progresses. Thus the effective equivalent circuit for the electrode is determined by the stimulus parameters, principally the current and the duration of the stimulus.

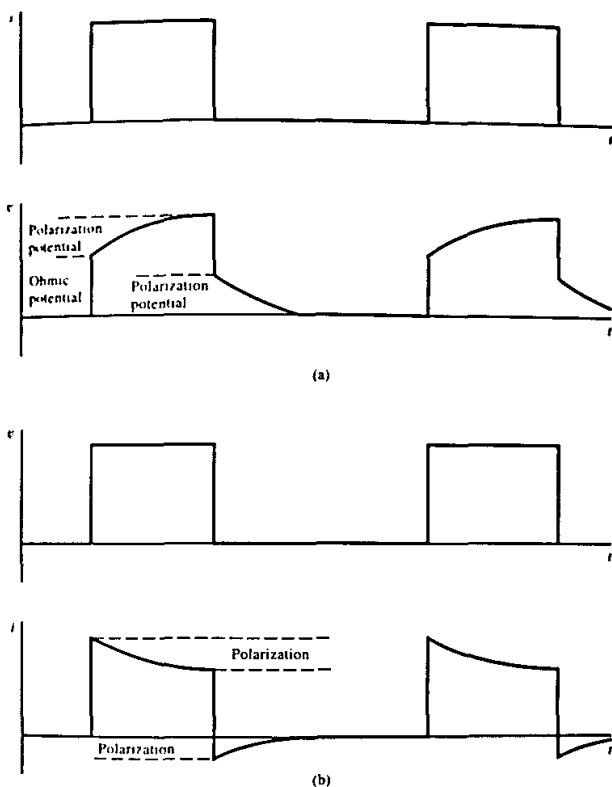


Figure 5.23 Current and voltage waveforms seen with electrodes used for electric stimulation (a) Constant-current stimulation. (b) Constant-voltage stimulation.

Rectangular biphasic or monophasic pulses are frequently used for electric stimulation. However, other waveshapes, such as decaying exponentials in trapezoids, or sine waves, have also been used. Frequently a stimulus is used that is either constant current or constant voltage during the pulse. The response of a typical electrode to this type of stimulus is illustrated in Figure 5.23.

A constant-current stimulus pulse is applied to the stimulating electrodes in Figure 5.23(a), giving the voltage response shown. Note that the resulting voltage pulse is not constant. This is reasonable when we consider that there is a strong reactive component to the electrode-electrolyte interface, or in other words, polarization occurs. The initial rise in voltage corresponding to

the leading edge of the current pulse is due to the voltage drop across the resistive components of the electrode–electrolyte interface, but we see that the voltage continues to rise with the constant current. This is due to the establishment of a change in the distribution of charge concentration at the electrode–electrolyte interface—in other words, to a change in polarization resulting from the unidirectional current. As stated in the description of the simplified electrode–electrolyte equivalent circuit of Figure 5.4, this polarization effect can be represented by a capacitor. Again it is important to remember that the size of this capacitor is determined by several factors, one of which is the current density at the interface, so the equivalent circuit for stimulating electrodes changes with the stimulus.

When the current falls back to its low value, the voltage across the electrodes drops, but not back to its initial value. Instead, after the initial steep fall, there is a slower decay corresponding to the dissipation of the polarization charge at the interface.

Constant-voltage stimulation of the electrodes is shown in Figure 5.23(b). In this case the current corresponding to the rising edge of the voltage pulse is seen to jump in a large step and, as the distribution of the polarization charge becomes established, to fall back to a lower steady-state value. When the voltage pulse falls, the current is seen to change direction and then slowly return to its initial zero value. This is a result of the dissipation of the polarization charge built up at the electrode–electrolyte interface.

In choosing materials for stimulating electrodes, we must take into consideration chemical reactions occurring at the electrode–electrolyte interface. If a stimulating current causes the material of the electrode to be oxidized, the electrode is consumed, which limits its lifetime and also increases the concentration of the ions of electrode material in the vicinity of the electrode. This could be toxic to the tissue. When electrodes such as the Ag/AgCl electrode are used, the stimulating current can result in either the formation of additional Cl^- or the reduction of what is already formed, thereby greatly changing the characteristics of the electrode. Thus the best stimulating electrodes are made from noble metals (or at least stainless steel), which undergo only minimal chemical reactions. It is also possible in this case that the chemical reaction involves water in the vicinity of the electrodes. Thus the local activity of hydrogen or hydroxyl ions in the vicinity of the electrode can greatly change. These changes in acidity or alkalinity can produce tissue damage that limits the overall effectiveness of the noble-metal electrodes. Of course, the polarization of these electrodes is large, and the waveforms, such as those shown in Figure 5.23, are less rectangular than they would be with nonpolarizable electrodes. In extreme cases, electrode voltage and current density can result in the electrolysis of water that leads to the evolution of small bubbles of hydrogen or oxygen gas. Clearly, this is a situation that should be avoided.

Carbon-filled silicone rubber electrodes are used for transcutaneous stimulation used in clinical pain management (Carim, 1988). Arzbaecher (1982) describes a pill electrode that can either record a large P wave or pace the heart from the esophagus. A stimulating electrode based upon the iridium/

iridium oxide system [Robblee *et al.* (1983)] has been shown to inject charge into biologic tissue. This type of electrode has been found to provide maximal current density for biphasic stimuli while minimizing chemical changes that could lead to tissue damage. The passage of charge from the electrode to tissue and in the reverse direction results in a chemical change in the iridium oxide chemical composition as opposed to injecting iridium ions into solution or reducing them.

5.11 PRACTICAL HINTS IN USING ELECTRODES

In using metal electrodes for measurement and stimulation, we should understand a few practical points not mentioned elsewhere in this chapter. 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. It is far better either to weld the lead wire to the electrode or at least to form a mechanical bond through crimping or peening. Dissimilar metals should not be used in contact because their half-cell potentials are different. And, because they are connected and in contact with the same electrolyte, it is more than likely that an electrochemical reaction will be set up between them that can result in additional polarization and often in corrosion of one of the metals. This factor also tends to make half-cell potentials less stable, thereby contributing to increased electric noise from the electrode.

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 means that the net dc potential seen at the input to the amplifier connected to the electrodes is relatively small, possibly even zero. 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. Frequently, this is due to a loss of effectiveness of the tack on the tape holding them in place. However, the problem need not arise if the electrodes are well designed. Lead wires to these surface electrodes should be extremely flexible yet strong. If they are, only tension on the lead wire can apply a force that is likely to remove the electrode. If the lead wire remains loose, it cannot apply any forces to the electrode because of its high flexibility. It is helpful to provide additional relief from strain by taping the lead wire to the skin a few cm from the electrode with some slack in the wire between the tape and the electrode.

The point at which the lead wire enters the electrode is a point of frequent failure. Even though the insulation appears intact, the wire within may be

broken as a result of severe repeated flexing at this point. Well-designed electrodes minimize this problem by providing strain relief at this point, so that there is a gradual transition between the wire and the solid material of the electrode. Using a tapered region of insulation that gradually increases from the diameter of the wire to one closer to that of the electrode often minimizes this problem and distributes the flexing forces over a greater portion of the wire.

Another point to consider is that the insulation of the lead wire and the electrode can also present problems. Electrodes are often in a high-humidity environment or are continually soaked in extracellular fluid or even in cleaning solution (if they are of the reusable type). The insulation of these electrodes is usually made of a polymeric material, so it can absorb water. Some of these materials can become more conductive when they absorb water, and, in the case of implantable electrodes, there may be some high-resistance contact with the lead wire as well as at the electrode itself. If the lead wire is made of a material different from that of the electrode, the problems just described can result, thereby increasing the observed electric noise and possibly leading to a weakening of the lead wires due to corrosion. Thus it is important to understand the insulation material used with the electrode and to make sure that there is a layer of it thick enough to prevent this problem from occurring.

One final point regarding electrodes for measuring biopotentials: In deriving the equivalent circuit for electrodes such as those shown in Figure 5.4, we stressed that for high-fidelity recordings of the measured biopotential, 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. This is demonstrated by Geddes (1966) for electrodes used to record electrocardiograms. Geddes shows how lowering the input impedance of the amplifier causes the recorded signal to take on a more and more biphasic character as well as a reduced amplitude.

PROBLEMS

5.1 A set of biopotential electrodes made of silver are attached to the chest of a patient to detect the electrocardiogram. When current passes through the anode, it causes silver to be oxidized, producing silver ions in solution. There is a $10\text{-}\mu\text{A}$ leakage current between these electrodes. Determine the number of silver ions per second entering the solution at the electrode-electrolyte interface.

5.2 Design a system for electrolytically forming Ag/AgCl electrodes. Give the chemical reactions that occur at *both* electrodes.

5.3 Design an Ag/AgCl electrode that will pass 150 mC (millicoulombs) of

charge without removing all the AgCl. Calculate the mass of AgCl required. Show the electrode in cross section and give the active area.

5.4 When electrodes are used to record the electrocardiogram, an electrolyte gel is usually put between them and the surface of the skin. This makes it possible for the metal of the electrode to form metallic ions that move into the electrolyte gel. Often, after prolonged use, this electrolyte gel begins to dry out and change the characteristic of the electrodes. Draw an equivalent circuit for the electrode while the electrolyte gel is fresh. Then discuss and illustrate the way you expect this equivalent circuit to change as the electrolyte gel dries out. In the extreme case where there is no electrolyte gel left, what does the equivalent circuit of the electrode look like? How can this affect the quality of the recorded electrocardiogram?

5.5 A zinc wire and an aluminum wire accidentally come in contact with a part of the body that has been saturated with a physiological saline solution. Is there a potential difference between these two wires? If your answer is yes, quantitatively state the value of this potential under open-circuit conditions.

5.6 Design the electrode of the smallest area that has an impedance of $10\ \Omega$ at 100 Hz. State your source of information, describe construction of the electrode, and calculate its area.

5.7 A pair of biopotential electrodes is used to detect the electrocardiogram of an adult male. It has become necessary to determine the equivalent-source impedance of this electrode pair so that a particular experiment can be performed. Describe an experimental procedure that can be used to determine this quantity, using a minimum of test equipment.

5.8 Using test equipment found in most labs, design (show a block diagram and wiring connections) a test facility for measuring the impedance versus frequency of 1-cm^2 electrodes. It should use the largest current density that does not cause a change in the impedance.

5.9 A pair of biopotential electrodes is placed in a saline solution and connected to a stimulator that passes a direct current through the electrodes. It is noted that the offset potentials from the two electrodes are different. Explain why this happens during the passage of current. Sketch the distribution of ions about each electrode while the current is on.

5.10 Electrodes having a source resistance of $4\ \text{k}\Omega$ each are used in a bipolar configuration with a differential amplifier having an input impedance of $70\ \text{k}\Omega$. What will be the percentage reduction in the amplitude of the biopotential signal? How can this distortion of the signal be reduced?

5.11 A nurse noticed that one electrode of a pair of Ag/AgCl cardiac electrodes used on a chronic cardiac monitor was dirty and cleaned it by scraping it with steel wool (Brillo) until it was shiny and bright. The nurse then placed the electrode back on the patient. How did this procedure affect the signal observed from the electrode and electrode impedances?

5.12 A metal microelectrode has a tip that can be modeled as being cylindrical. The metal itself is $1\ \mu\text{m}$ in diameter, and the tip region is 3 mm long. The metal has a resistivity of $1.2 \times 10^{-5}\ \Omega \cdot \text{cm}$ and is coated over its circumference with an insulation material $0.2\ \mu\text{m}$ thick. The insulation material has a rela-

tive dielectric constant of 1.67. Only the base of the cylinder is free of insulation.

- a. What is the resistance associated with the tip of this microelectrode?
- b. What is the area of the surface of the electrode that contacts the electrolytic solution within the cell? The resistance associated with the electrode-electrolyte interface of this material is $10^3 \Omega$ for 1 cm^2 . What is the resistance due to this microelectrode's contact with the electrolyte?
- c. What is the capacitance associated with the tip of the microelectrode when the capacitances at the interface of the electrode-electrolytic solution are neglected?
- d. Draw an approximate equivalent circuit for the tip portion of this microelectrode.
- e. At what frequencies do you expect to see distortions when the electrode is connected to an amplifier having a purely resistive input impedance of $10 \text{ M}\Omega$? You may assume that the reference electrode has an impedance low enough so that it will not enter into the answer to this question. If the amplifier's input impedance is raised to $100 \text{ M}\Omega$, how does this affect the frequency response of the system? Is this difference significant for most intracellular biological applications?

5.13 A micropipet electrode has a luminal diameter of $3 \mu\text{m}$ at its tip. At this point, the glass wall is only $0.5 \mu\text{m}$ thick and 2 mm long. The resistance of the electrolyte in the tip is $40 \text{ M}\Omega$. The glass has a relative dielectric constant of 1.63. Estimate the frequency response of this electrode when it is connected to an infinite-input-impedance amplifier. How can this frequency response be improved?

5.14 A pair of biopotential electrodes are used to monitor a bioelectric signal from the body. The monitoring electronic circuit has a low-input impedance that is of the same order of magnitude as the source impedance in the electrodes.

- a. Sketch an equivalent circuit for this situation.
- b. Describe qualitatively what you expect the general characteristics of the frequency response of this system to be. It is not necessary to plot an analytic Bode plot.

5.15 A pair of identical stainless steel electrodes is designed to be used to stimulate skeletal muscles. The stimulus consists of a rectangular constant-voltage pulse applied to the electrodes. The pulse has an amplitude of 5 V with a duration of 10 ms . Draw, on the basis of the equivalent circuits of each of the electrodes, an equivalent circuit for the load seen by the constant-voltage pulse generator. Simplify your circuit as much as possible. What is the waveshape of the current at the generator terminals? Remember that a constant-voltage generator has a source impedance of zero. Explain and sketch the resulting current waveform.

5.16 An exotic new animal, recently discovered, has an unusual electrolyte makeup in that its major anion is Br^- rather than Cl^- . Scientists want to measure the EEG of this animal, which is less than $25 \mu\text{V}$. Electrodes made

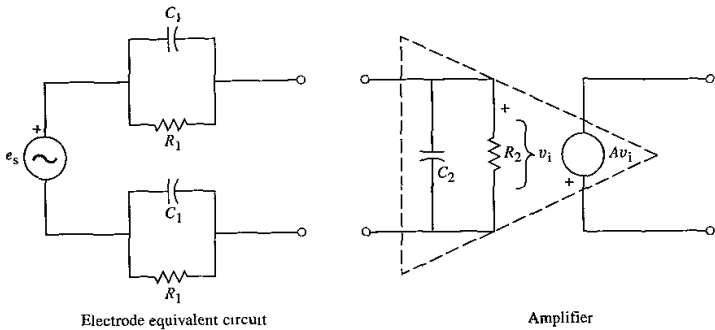


Figure P5.1

of Ag/AgCl seem to be noisy. Can you suggest a better electrode system and explain why it is better?

5.17 Needle-type EMG electrodes are placed directly in a muscle. Figure P5.1 shows their simplified equivalent circuit and also the equivalent circuit of the input stage of an amplifier. The value of the capacitor C_2 in the amplifier may be varied to any desired quantity.

- Assuming $C_2 = 0$ and the amplifier gain is A , write an equation showing the output voltage of the amplifier as a function of e_s (the signal) and frequency.
- Determine a value for C_2 that gives electrode–amplifier characteristics that are independent of frequency.
- What is the amplifier's output voltage in part (b) when the signal is e_s ?

5.18 Figure P5.2 shows the equivalent circuit of a biopotential electrode. A pair of these electrodes are tested in a beaker of physiological saline solution. The test consists of measuring the magnitude of the impedance between the electrodes as a function of frequency via low-level sinusoidal excitation so that the impedances are not affected by the current crossing the electrode–electrolyte interface. The impedance of the saline solution is small enough to be neglected. Sketch a Bode plot (log of impedance magnitude versus log of

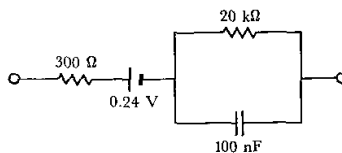


Figure P5.2

frequency) of the impedance between the electrodes over a frequency range of 1–100,000 Hz.

5.19 A pair of biopotential electrodes are implanted in an animal to measure the electrocardiogram for a radio-telemetry system. One must know the equivalent circuit for these electrodes in order to design the optimal input circuit for the telemetry system. Measurements made on the pair of electrodes have shown that the polarization capacitance for the pair is 200 nF and that the half-cell potential for each electrode is 223 mV. The magnitude of the impedance between the two electrodes was measured via sinusoidal excitation at several different frequencies. The results of this measurement are given in the accompanying table. On the basis of all of this information, draw an equivalent circuit for the electrode pair. State what each component in your circuit represents physically, and give its value.

Frequency	Impedance (magnitude), Ω
5 Hz	20,000
10 Hz	19,998
.	.
.	.
.	.
40 kHz	602
50 kHz	600
100 kHz	600

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