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4 Principles of bioelectrical measurement

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4.1 INTRODUCTION

The most significant functions of laboratory instruments are measurement and control. Since the scientist's principal activities are observations under controlled conditions, one index of the usefulness of instruments to experimenting psychologists is the extent to which they facilitate the observation, quantification and control of variables relevant to the psychological situation. (Grings, 1954, p. 2)

Parallels in the development of science and technology have been observed in both the physical and the biobehavioral sciences. Technological advances supply the scientist with progressively better tools for measurement and control. Within the physical sciences, for example, the development of the radio telescope allowed astronomers to detect and examine starlike quasars at a level beyond the range of optical imaging (Schmidt, 1963). Brown and Saucer (1958) noted several examples within the behavioral sciences in which technological advances facilitated scientific progress. For example, development of the vacuum tube oscillator greatly improved stimulus control within the study of audition, and the development of suitable amplifiers and display devices (described later in this chapter) enabled the reliable detection and quantitative description of subtle bioelectrical signals as brief as a single action potential (Erlanger & Glasser, 1937). More recently, technological advances permitting detection of weak magnetic fields led to the discovery and quantification of event-related magnetic activity in the brain (Beatty, Barth, Richer, & Johnson, 1986).

Although technological advances may permit scientists to venture into frontier areas of research, inadequate or incomplete technical knowledge may lead to serious errors of inference. For example, soon after the discovery of X-rays, numerous scientists employing the technology for observing the effects of X-rays reported a related phenomenon called *N*-rays (Rostron, 1960, cited in Barber, 1976). The effects misattributed to *N*-rays were later shown to be the result of difficulties involved in estimating by eye the brightness of faint objects (Wood, 1904). In addition, soon after the birth of experimental psychology, the reaction times being measured in Wilhelm Wundt's laboratory at the University of Leipzig were discovered by James Cattell to be in error because of asymmetries in the time required for the magnet in the chronoscope to attract and release the armature (Schal, Davis, & Merzbach, 1976). More recently, Ionides (1982) reported a surprising effect involving temporal integration of a sequence of two very briefly presented visual patterns. It was later determined that the method for presentation of the visual patterns failed to properly account for the decay time of the phosphor on the

display screen, so that the patterns actually overlapped in time (Jonides, Irwin, & Yantis, 1983).

The field of psychophysiology is, at present, highly dependent on electronic instrumentation for detection, amplification, recording, and quantification of physiological processes. Numerous texts on techniques in psychophysiology attest to the fact that the establishment and maintenance of any psychophysiology laboratory requires certain technical expertise (e.g., Brown, 1967; Coles, Donchin, & Porges, 1986; Greenfield & Sternbach, 1972; Martin & Venables, 1980; Stern, Ray, & Davis, 1980; Venables & Martin, 1967). Although a psychophysiologicalist need not become a bioelectrical engineer, an understanding of some fundamentals of electricity and electrical circuits, the unique properties of bioelectrical signals, and the principles of physiological measurement can guide the researcher in the proper application of existing and yet to be developed instruments.

This chapter begins with a description of the physical properties of electricity and electrical transmission. Next, the unique properties of bioelectrical signals are described and related to techniques for detection of bioelectrical signals. Afterward, both the essential and optional components of sensor systems for detecting, amplifying, and storing physiological responses are described.

4.2 BASICS OF ELECTRICITY

All electrical phenomena¹ are traceable to an intrinsic property of matter called charge. The exact nature of electrical charge is understood only in terms of the behavior of particles. The fundamental particles of electrical charge are protons and electrons. Protons, located in the nucleus of all atoms, possess a unit positive charge and electrons, located in planetarylike orbits about the nucleus of an atom, possess a unit negative charge. The simplest molecular structure is that of a hydrogen atom, with one proton in the nucleus and one orbiting electron. The electrical charge of a single electron is quite small, and the coulomb, the physical unit for electrical charge, was developed before discovery of the electron. One coulomb is the charge of 6.2×10^{18} electrons.

Electrons are maintained in orbit about the nucleus of an atom by a mechanical force of attraction that exists between particles of opposing charges. This force, called *electrical potential*, measured in volts (V), is the force underlying all electrical signals. Although electrical potential can be a measure of force of attraction between any two locations, measures of voltage at a particular location are most often reported relative to an electrically neutral location. Such an electrically neutral location is called electrical or earth *ground*. The electrical potential within and surrounding atoms is best described as a spherical *electrical field* centered about the nucleus. The strength of the electrical potential within this field is inversely proportional to the square of the distance from the center of mass. Consequently, electrons, particularly those most distant from the nucleus of an atom, can move more freely than protons and neutrons. The movement of charged particles through an electrical field is called an *electrical current* and is measured in amperes (A). Within a vacuum electrical current can travel almost as fast as light, but within all other media speeds of conduction are somewhat slower.

Electrons that orbit atomic nuclei are located within nested energy levels, called shells, each with a limited capacity for electrons. The innermost shells are filled in succession as the number of electrons increases across elements, and the number of electrons in the outermost shell of an atom determines the chemical and electrical

1a	2a											0	
1	H	3a	4a	5a	6a	7a	8	9	10			2	
3	Li	5	6	7	8	9	10					He	
4	Be	B	C	N	O	F	Ne						
11	Na	13	14	15	16	17	18						
12	Mg	Al	Si	P	S	Cl	Ar						
19	K	21	22	23	24	25	26	27	28	29	30	31	
20	Ca	Sc	Ti	V	Cr	Mn	Fe	Co	Ni	Cu	Zn	Ga	
37	Rb	39	40	41	42	43	44	45	46	47	48	49	
38	Sr	Y	Zr	Nb	Mo	Tc	Ru	Rh	Pd	Ag	Cd	In	
55	Cs	57	72	73	74	75	76	77	78	79	80	81	
88	Ra	La	Hf	Ta	W	Re	Os	Ir	Pt	Au	Hg	Tl	
89	Ac	LANTHANIDES										85	
		58	59	60	61	62	63	64	65	66	67	68	
		Ce	Pr	Nd	Pm	Sm	Eu	Gd	Tb	Dy	Ho	Er	
		70	71	ACTINIDES									
		90	91	92	93	94	95	96	97	98	99	100	
		Th	Pa	U	Np	Pu	Am	Cm	Bk	Cf	Es	Fm	
		102	103	101	102	103	101	102	103	101	102	103	
		Lw	No	Mv	No	Lw	Mv	No	Lw	Mv	No	Lw	

Figure 4.1. Periodic table of elements.

reactivity of that element. Atomic elements that appear in the same row of the periodic table (Figure 4.1) have differing numbers of electrons in the same outermost shell, whereas elements in the same column have the same number of electrons in different outermost shells. Elements in the last column of the periodic table, such as neon, are chemically stable and electrically unchanged because the outermost electron shell is full and the number of protons and electrons remains equal.

Electrically charged atoms are called *ions*. Elements in the first column, such as sodium, or the seventh column, such as chlorine, are chemically reactive and are rarely found in the electrically unchanged state. Sodium, for example, which has only one electron in its outermost shell, tends to lose one electron, eliminating the outermost shell and forming a *cation*, an atom with a net positive charge. Chlorine, on the other hand, which has seven electrons in its outermost shell, tends to gain one electron, completing the outermost shell and forming an *anion*, an atom with a net negative charge. Chemically reactive elements tend to form *ionic* chemical bonds by actual transfer of electrons from one atom to another. Atoms of less reactive elements, such as carbon, with four electrons in the outermost shell, form *covalent* chemical bonds by sharing electrons among atoms rather than actually transferring electrons from one atom to another.

In sum, the electrical and chemical properties of matter are integrated and inseparable. One of the most familiar sources of electrical voltage, a carbon-zinc (dry-cell) battery, relies on a chemically generated electrical potential between a carbon rod and a zinc plate when separated by an *electrolyte* medium, a paste containing free ions. The electrolyte in a dry-cell battery is composed of zinc chloride, ammonium chloride, and manganese dioxide. Chemical generation of an electrical

potential is based on the production of separate concentrations of cations and anions. Inside the carbon-zinc battery, zinc atoms are dissolved into the electrolyte solution, forming doubly charged zinc ions and leaving behind two free electrons. At the same time, the ammonium ions and the manganese dioxide work together to withdraw electrons from the carbon rod, leaving it positively charged. The negatively charged zinc plate is called the *negative terminal*, or *anode*, and the positively charged carbon rod is called the *positive terminal*, or *cathode*. When the positive and negative terminals of the battery are connected externally as through an electrical circuit, free electrons flow from the zinc plate through the circuit to the carbon rod, generating electrical current in the circuit. As long as the circuit is continuous, current continues to flow until the ionic supply within the battery is exhausted.

The voltage provided by a battery is unidirectional and relatively stable in amplitude throughout the effective life of the battery and thus provides a unidirectional current referred to as *direct current* (DC). The amplitude of DC potential at either terminal of a battery is generally referenced to 0 V. A typical dry-cell battery provides an electrical potential of about 1.5 V DC. In *storage batteries* such as those used in automobiles, the chemical reactions that produce the separate ionic concentrations can be reversed by an external energy source, and potentials of 6, 12, or 25 V DC are common.

The nature of the atomic structure within a material determines the extent to which that material will support the movement of charged particles. Solid materials high in unbound electrons, such as metals, and ionic solutions such as saline are classified as *conductors*, or carriers, of electrical current. Materials low in unbound electrons, such as rubber, paper, and glass, do not carry electrical current, and are used as electrical *insulators*. Materials with an intermediate level of unbound electrons such as carbon, silicon, and germanium, called *semiconductors*, are used in production of solid-state electronic components such as transistors and integrated circuits.

The extent to which a material impedes the flow of a unidirectional electrical current between two points of unequal electrical potential is quantified as *resistance*, measured in ohms. The commonly accepted symbol for ohms is the Greek letter omega, Ω . An ohm is a very small unit, and practical resistances within physiological systems range from kilohms to megohms. Conductors are low in resistance; insulators are high in resistance. Even the best conductors provide some resistance to flow of current. However, conductors with a large cross-sectional area provide many more current paths than those with a small cross-sectional area and thus provide a lower resistance to current flow. It is often useful to describe the ability to conduct current in terms of the reciprocal of resistance, called *conductance*, reported in siemens, S (1 siemen = 1/ohm).

The relationship between electrical potential V , current I , and resistance R is described mathematically as Ohm's Law, $V = I \times R$. An analogy to fluid systems helps to clarify this mathematical relationship. Electrical potential is a force similar to the pressure at a water faucet. The forces of both hydraulic pressure and electrical potential are present even in the absence of water and of current flow, respectively. When a valve in a hydraulic system is closed, the pressure is present, but the resistance offered by the valve prevents the flow of water. Opening the valve reduces this resistance, permitting water to flow. The flow of water is analogous to electrical current. A large-diameter hose attached to the valve provides less resistance to the flow of water than a small-diameter hose. Thus, one can increase the water flow by increasing the water pressure or by decreasing the resistance with

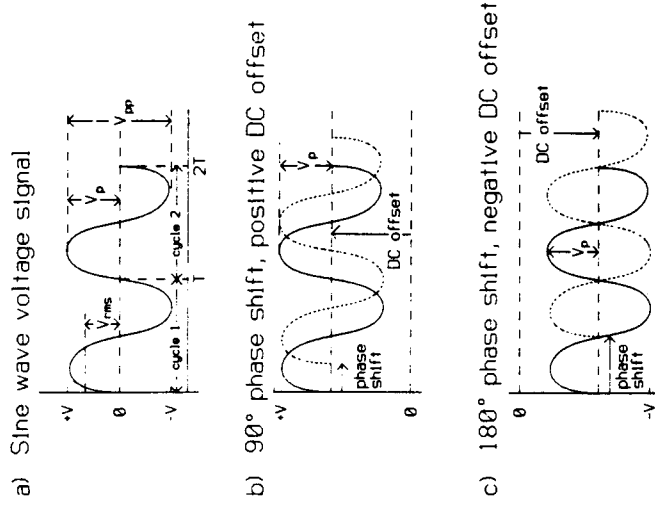


Figure 4.2. Quantification of component parts of simple sine wave voltage signals as function of time. Peak voltage V_p , peak-to-peak voltage V_{pp} , rms voltage V_{rms} , and period T for two-cycle sine wave (a). Sine wave voltage signals with positive and negative DC offsets in (b) and (c), respectively. A second sine wave voltage phase shifted from the original in (a) by 90° and 180° is shown in (b) and (c), respectively.

larger valves and hoses. Likewise, electrical currents are greatest when electrical potentials are high and electrical resistances are low.

The fluid analogy given here may give the false impression that electrical currents and voltages are always unidirectional and remain fixed at a constant level determined by the electrical resistance present.² Although DC sources such as batteries do provide unidirectional, steady-state potentials, the voltages provided by the electrical wall outlets available in a laboratory are not steady but vary regularly with time. Assuming for the moment that electrical resistances remain fixed, Ohm's Law indicates that periodically changing voltages must produce periodically changing currents. *Alternating currents* (AC) periodically vary in both amplitude and direction of flow.

A sine wave, one of the simplest of periodic waveforms, along with several standard measures of change in the time and amplitude dimensions, is illustrated in figure 4.2 (top panel). The time interval required for one cycle is called the *period* (T). The reciprocal of the period is the *frequency* (f), typically reported as the number of cycles per second, or *hertz* (Hz). The relationship between period and frequency for selected values is shown in Table 4.1. For a sinusoidal potential signal that is symmetrical about 0 V, the peak voltage V_p is the maximum voltage observed, and the peak-to-peak voltage V_{pp} is twice V_p . Likewise, for a sinusoidal current, symmetrical about 0 A, the maximum current observed is called the peak current I_p , and the peak-to-peak current I_{pp} is twice I_p . Because the amplitude of an AC signal

Table 4.1. Relationship between period and frequency

Period(s)	Frequency (Hz)
1.0	1
0.5	2
0.1	10
0.05	20
0.001	100
0.0001	1,000
0.00001	10,000

changes across time, the peak value is not the best indicant of the average effective amplitude. The *root-mean-square (rms) value* for an alternating current or voltage provides a time-averaged amplitude that quantifies variations across time. The general formula for calculation of the rms value for a set of sampled voltages containing n different values V_1, \dots, V_n is given by

$$V_{\text{rms}} = \sqrt{\frac{1}{n} \sum_{i=1}^n V_i^2}$$

This formula can be used with any voltage or current signal that changes with time, including complex physiological signals described later. However, the rms value of a pure sine wave voltage or current V_{rms} or I_{rms} can be reduced to the peak value V_p or I_p divided by the square root of 2. Instruments for measuring AC potentials and currents are universally calibrated in terms of rms values. Unless stated otherwise, it is generally understood that AC potentials and currents are characterized by their rms values. Laboratory AC power lines operate at frequency of 60 Hz in the United States and 50 Hz in Europe with rms amplitudes of 117 and 220 V, respectively. As we shall see, these frequencies overlap with those present in some biopotentials, and the voltages are considerably higher than physiological levels.

In addition to frequency and amplitude, AC signals can be compared in terms of *phase shift* and *DC offset*. Phase shift refers to displacement along the time dimension, and DC offset refers to displacement along the amplitude dimension. Although phase shifts represent temporal displacements, they are generally reported as a *phase angle*, where the period T corresponds to an angle of 360° . When two identical sine wave signals are phase shifted by 0° , 360° , or multiples of 360° (or 2π radians), they appear synchronized in time and are called "in phase." When one of two identical sine wave signals is phase shifted by 90° , the peaks and valleys of one signal correspond in time to the median amplitude level for the other signal. When the peaks of one sine wave signal correspond in time to the valleys of the other signal, the signals are completely out of phase, and the phase angle is 180° . Two identical sine waves that are 90° and 180° out of phase are illustrated in the middle and bottom panels of Figure 4.2.

When the median level of an AC signal is displaced above or below the 0 V level, the waveform can be described as the sum of two waveforms, one AC signal symmetrical about zero plus one constant DC offset signal, either positive or negative, which accounts for the constant displacement from zero. Positive and

negative DC offsets are shown in the middle and bottom panels of Figure 4.2, respectively. It is often useful to be able to separate the slow DC-like and fast AC-like components of a single bioelectrical signal. For example, sensor systems for detection of cyclic respiration rate often include an uninteresting DC offset in the signal. Compensatory circuitry that effectively removes the DC offset in the signal can allow selective amplification of the AC signal of interest. However, it is not always appropriate to disregard the DC offset. In particular, some electrical properties of the skin discussed later in this chapter include discrete responses superimposed on a DC offset, the latter of which reflects the basal level of electrodermal activity (see Dawson, Schell, & Filion, chapter 10).

It is important to note that current, the amount of flow of electrical charges, not voltage, is what poses a risk for electrocution. Thus, although *static electricity*, the separation of charged ions due to friction, is characterized by high voltages, the currents associated with static discharge are minute and pose no risk to subjects. The main sources of tissue-damaging currents are AC power lines and devices attached to them. The typical current flow through a 100-W light bulb attached to an AC power line is about 1 A. However, currents in the milliampere (10^{-3} A) range are hazardous to living organisms. The threshold for detection of current is about 1 mA, and currents of 10 mA are painful. At about 20 mA the muscles exhibit tetanic contraction, making it impossible to let go of a wire that may have been grasped. Muscle damage can occur at 50 mA, and at 100 mA disorganized contractions of the heart, called *fibrillation*, occur. Procedures designed to prevent placing the subject at risk from electrical hazard are discussed in detail in Section 4.6.

4.3 SIMPLE CIRCUITS

An *electrical circuit* consists of a complete conductive path that permits the flow of current. The simplest DC circuit is a single conductive wire connecting the anode and cathode of a battery. The true direction of electron flow is from anode (negative) to cathode (positive). The convention for the direction of current in electronics, however, is based on the direction of movement of a positive charge, which is from positive to negative. Using Ohm's Law, the amount of current flow within the wire can be computed as $I = V/R$, where V is the battery voltage and R is the resistance of the wire. Since R for most conductors is small, such a simple circuit would permit rapid current flow and would quickly deplete the battery potential.

Most practical electrical circuits include components that have more resistance than a single conductive wire. Increasing the electrical resistance within a DC circuit reduces the current flow. For circuits using batteries as the DC potential source, greater resistance means longer life for the battery. It is often useful to add components of known resistance, called *resistors*, to a circuit to control current flow. When voltage is constant, as in DC circuits, increasing circuit resistance also serves to reduce power consumption. Specifically, when voltage is constant, power P , computed as $P = I^2R$, can be reduced to $P = V^2/R$ by substitution of V/R for I (Ohm's Law). The unit of measurement for power is the watt (W).

An electrical circuit that provides a single current path between the positive and negative potential source is called a *series circuit*. A simple series circuit consisting of a potential source such as a battery or an AC wall outlet, a switch, a fuse, an indicator lamp, and a box used to designate a circuit complex within some laboratory instrument is illustrated in Figure 4.3 (top panel). The circuit components

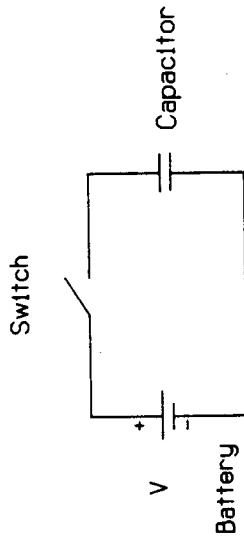


Figure 4.4. DC circuit to charge capacitor.

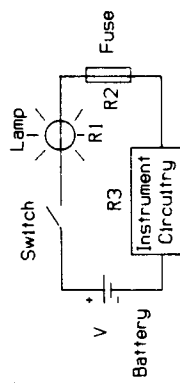
to the original tank. The amount of water flow from the tank is the lowest when only one outlet is opened, and as more water outlets are opened, the addition of more paths for the water to follow permits a greater flow of water. The equivalent resistance of resistors in parallel is always smaller than the smallest of the resistors, and resistors in parallel add as reciprocals. That is, the equivalent resistor for resistors in parallel is given by $1/R_{parallel} = 1/R_1 + 1/R_2 + 1/R_3 + \dots + 1/R_n$.

For the series and parallel circuits illustrated in Figure 4.3, it was assumed that the potential source could be either DC or AC. There are, however, many circuits that react differently to AC compared to DC potentials. These differences are quantified by Ohm's Law for AC circuits in which the resistance term R for DC circuits is replaced by a term for resistance to alternating currents, called *impedance* (Z). Impedance, measured in ohms, includes DC resistance plus two forms of resistance specific to alternating currents, *capacitive reactance* and *inductive reactance*, both described later in this chapter. An understanding of *reactance* (X) is facilitated by an examination of some simple circuits that react differently to AC compared to DC potentials.

One unique feature of AC potentials is the ability to conduct electrical current across thin, nonconductive barriers that cannot be crossed by DC potentials. Consider a simple series circuit consisting of a source of DC potential such as a battery, a switch, and conducting plates separated by a thin insulator, as illustrated in Figure 4.4. Closing the switch connects the positive and negative terminals of the battery to different plates. The potential difference between the two plates causes positive and negative charged particles, attracted to each other across the thin insulating gap, to line up along the opposing plates. When this occurs, the plates are said to be *polarized*, that is, one plate contains a matrix of positively charged particles and the other a matrix of negatively charged particles. The amount of charge that can be stored on either side of such a gap is called *capacitance* (C). The unit for capacitance, the farad, is very large, and typical capacitances are in the range of picofarads (pF) to microfarads (μ F). Capacitance has a special significance to the psychophysicologist because, as will be described later, the detection of many bioelectrical potentials requires a capacitive connection between the laboratory data collection equipment and the subject at the electrode interface.

Capacitance depends upon the size of the conductive plates and the size of the gap. The highest capacitances are found with large plates and small gaps since these conditions maximize the concentrated attractive force between opposing charges on the two plates. With DC potentials, current (i.e., movement of charged particles) flows in the circuit for the brief instant after closure of the switch during which the

a) Series Circuit



b) Parallel Circuit

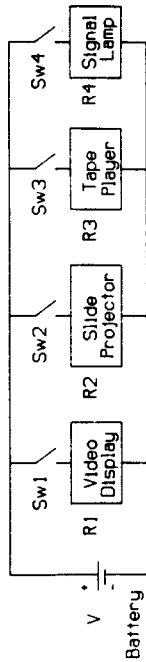


Figure 4.3. Series circuit (a) and parallel circuit (b).

are all connected end-to-end to form a single loop. Each component has an associated internal resistance, designated as R_1 , R_2 , R_3 , and R_4 . Closing the single switch would supply electrical power to all four circuit components. It is common practice to connect fuses and power (on-off) indicator lamps in series with all other internal circuitry within an instrument. A lighted power lamp then verifies the continuity of the internal circuitry, and a single fuse can disconnect all internal components from the power source when rendered discontinuous (blown) by a current surge. The current in this simple circuit is given by the voltage V divided by the total resistance encountered in the current loop. Stated another way, the equivalent resistance of resistors in series is a simple sum of the individual resistances, or $R_{series} = R_1 + R_2 + \dots + R_n$. Thus, each component added to series circuit reduces the current flow through all components. Furthermore, a breakdown of the conducting path through any component in such a series circuit would disrupt the single current path and prevent the equipment from operating.

A simple *parallel circuit* for four independent pieces of laboratory equipment plus four separate switches and a potential source is illustrated in the bottom panel of Figure 4.3. In this parallel circuit each piece of equipment and its associated resistance is connected to both sides of the potential source. One familiar example of a parallel circuit is the separate sockets on a multitap AC wall outlet or a power strip. Most electrical circuits contain some components connected in series with the voltage source and some components connected in parallel with the voltage source, but separate examination of the embedded series and parallel circuits can provide useful information such as the current flow through each component. Connecting resistance in parallel to a voltage source provides multiple current paths for current to flow from the voltage source. Each resistance added to a parallel circuit increases the total current flow through the voltage source without reducing or disrupting current flow through the other components. One hydraulic model for resistors in parallel is a closed water system consisting of a large tank of water and multiple water outlets that convey the water along hoses or pipes before returning the water

plates of the capacitor are polarized. Once the plates are fully polarized, also called charged, the potential across the gap would be equal and opposite to the DC source potential, and current would no longer flow (Figure 4.4). By reversing the connections to the battery, the plates can be discharged and polarized in the opposite direction, and another brief current flow can be detected. If the battery were replaced with a laboratory source of AC potential (e.g., 60 Hz, 117V), the polarity of the plates would be reversed 60 times per second, producing an effective current that alternated in direction. The process of charging and discharging the plates of a capacitor requires time, and the resulting current is phase shifted from the input potential by $+90^\circ$.

The amplitude of current flow across a capacitive gap increases with increases in the frequency of alternation of the AC potential. The extent to which a capacitive gap impedes current flow, termed capacitive reactance (X_c), is inversely related to capacitance C and the frequency of the AC potential (ω) as given by $X_c = 1/\omega C$. Thus, a capacitor appears like an *open switch* to DC signals, and as the frequency of the signal is *increased*, the amount of current flow through a capacitor *increases*. Connecting capacitors in a parallel circuit is analogous to increasing the area of the plates, thus increasing the effective capacitance, and capacitances in a parallel circuit add directly (i.e., $C_{\text{parallel}} = C_1 + C_2 + \dots + C_n$). Connecting capacitors in a series circuit can be compared to increasing the effective size of the gap to be crossed, thus decreasing the effective capacitance, and capacitances in series add as reciprocals (i.e., $1/C_{\text{series}} = 1/C_1 + 1/C_2 + \dots + 1/C_n$).

A second unique property of AC compared to DC potentials results from the magnetic field that exists surrounding the current in any conductor. The lines of force of the magnetic field produced by an electrical current in a straight wire are circular and perpendicular to the current flow. The orientation of the magnetic field surrounding a conductive wire is conventionally described by the "right-hand rule." That is, when grasping the wire with the right hand so that the thumb points toward the anode, the fingers will point in the direction of the magnetic field. The strength of an electrically produced magnetic field can be concentrated in one region by looping conductive wire into a coil and adding conductive material such as iron in the center of the coil, as is done in most electromagnets. Many instruments used in the psychophysiology laboratory operate on electromagnetic principles, including the physical displacement of pens on an ink-writing chart recorder, galvanometers on biofeedback display equipment, and mechanical relay switches used to gate stimulus events. Furthermore, the electrical activity of the brain produces corresponding low-intensity magnetic fields, and localized changes in neuromagnetic activity have been elicited by visual, auditory, and somatosensory stimuli (Beatty et al., 1986).

Not only does any electrical current generate a surrounding magnetic field, but also the force of any *changing* magnetic field will induce an electrical current in any nearby conductors. The strength of the induced current is directly proportional to the rate of change of the magnetic field affecting the conductor. Regardless of the orientation of the conductor, the induced current is always oriented to produce a magnetic field that directly opposes the original changing magnetic field. Thus, at any instant in time, the induced current in a conductive wire placed parallel to a wire conducting an alternating current is opposite in direction to the direction of the original alternating current. The strength of this magnetic opposition to changes in current is quantified as *inductance*, abbreviated L and measured in *henries* (H). The opposition to changes in current offered by inductance in electrical systems has

been likened to the resistance to motion in mechanical systems, called inertia (DeMarre, Kantrowitz, Zucker, & Simmons, 1979).

The principles of inductance are uniquely linked to AC as compared to DC potentials in two important ways. First, the common method for generating alternating currents relies on inductance. Typically, a large coil of wire wound around a soft iron core is placed near a pivoted magnet that is mechanically forced to rotate by means of water, steam, or wind. The fluctuations in the magnetic field surrounding the coil of wire produced by the rotating magnet induce a current that changes in amplitude and direction with each half rotation of the magnet. Second, any conductor that carries an alternating current is surrounded by a changing magnetic field capable of inducing current in other nearby conductors. The concentrated magnetic field produced by an *inductor* such as a coil of wire, resists the passage of alternating current by inducing a current opposite in direction to the applied current. The resulting alternating current through an inductor is phase shifted by -90° relative to the input potential. It is the principle of inductance that makes electromagnetism a mixed blessing for the psychophysicologist. Specifically, the amplitude and frequency of current induced by AC power lines is sufficient to mask many biopotential signals, and psychophysiological laboratories should be designed or selected to isolate sources of magnetic fields from physiological signals.

Inductive coils conduct direct current without distortion, since the magnetic field produced by direct current is constant and no opposing currents are induced. The extent to which an inductor impedes the flow of alternating current, termed *inductive reactance* (X_L) is directly related to inductance (L) and the frequency of the AC potential, ω , as given by $X_L = \omega L$. Thus, an inductor appears as a *closed switch* to a DC signal, and current flow *decreases* as the signal frequency *increases*. Connecting inductors in series has the same effect as increasing the number of turns on the coil of a single inductor, increasing the effective inductance, and inductances in series add directly (i.e., $L_{\text{series}} = L_1 + L_2 + \dots + L_n$). Connecting inductances in parallel permits multiple current paths, reducing the effective inductance, and inductors in parallel add as reciprocals (i.e., $1/L_{\text{parallel}} = 1/L_1 + 1/L_2 + \dots + 1/L_n$). Inductors are used in many laboratory instruments, particularly as *voltage transformers*. A voltage transformer operates on the principle of *mutual inductance*, in which the changing magnetic field around one multiturn coil induces current in an adjacent coil that in turn generates a magnetic field in opposition to the initial field. The strength of magnetic fields, like that of electrical fields, decreases as the square of the distance from the source, and thus the amplitude of induced currents is highly dependent on the distance between the two coils. Mutual inductance is maximized when the two coils are wound around the same iron core, as shown in Figure 4.5. Voltage transformers can be designed to step up (increase) or step down (decrease) the input voltage, depending on the ratio of the number of turns on the primary coil (n_1) generating the magnetic flux and that on the secondary coil (n_2) intercepting the magnetic flux as current. The oldest means for administration of controlled electrical current (shock), the *inductorium*, is based on mutual inductance between coupled coils (Grings, 1954, p. 269).

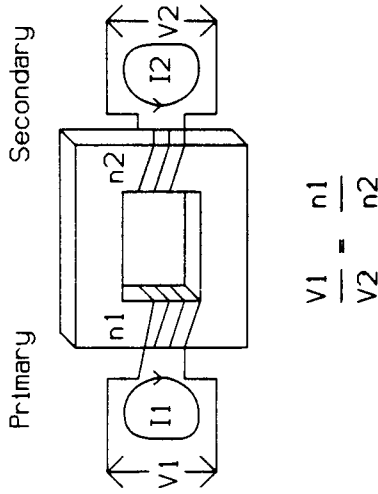


Figure 4.5. Voltage transformer based on mutual inductance.

perturb the activity being measured. Although electrical measurement can be seen as a tool for design and repair of electrical instruments, the principles of electrical measurement provide the psychophysicologist with useful tools for assessing physiological responses. The three most common tools for electrical measurement are the *multimeter*, the *oscilloscope*, and the *bridge circuit*. Each of these tools can be extremely valuable in the psychophysiology laboratory, and the principles for their use generalize to all types of psychophysiological measurement.

Typical multimeters include separate internal circuits for measuring current (*ammeter*), potential (*voltmeter*), and resistance (*ohmmeter*). The appropriate use of each of these three meter circuits is illustrated in Figure 4.6. A very simple electrical circuit that includes a potential source V and an attached load resistance R_{load} is shown in panel *a* of Figure 4.6. To measure the current passing through R_{load} , the ammeter circuit is attached anywhere in the current loop in series with R_{load} . This requires that the existing current loop be opened (physically separated) and the meter circuit inserted in the opening to complete the loop, as shown in panel *b* of Figure 4.6. Because resistances in series add directly, it is important for the internal resistance of an ammeter circuit to be low relative to R_{load} so that addition of the ammeter to the circuit will not substantially reduce the current in the loop being measured.

To measure the electrical potential across R_{load} , the voltmeter circuit is connected in parallel with R_{load} , as shown in panel *c* of Figure 4.6. Note that connecting a meter in parallel need not disrupt the existing circuit loop. Voltmeter circuits are designed with high internal resistance so that the amount of current diverted from the circuit element being measured is minimal. The ability to measure voltage without disruption of the circuitry generating the signal makes measurement of bioelectrical potentials possible at the skin surface, where direct measurement of bioelectrical currents is difficult.³

It is possible to compute the resistance of R_{load} in Figure 4.6 using an ammeter to measure the current through R_{load} and a voltmeter to measure the voltage across R_{load} and then applying Ohm's Law. However, an ohmmeter circuit can be used to measure resistances directly. The ohmmeter circuit must be connected in parallel with the resistance being measured, as shown in panel *d* of Figure 4.6.

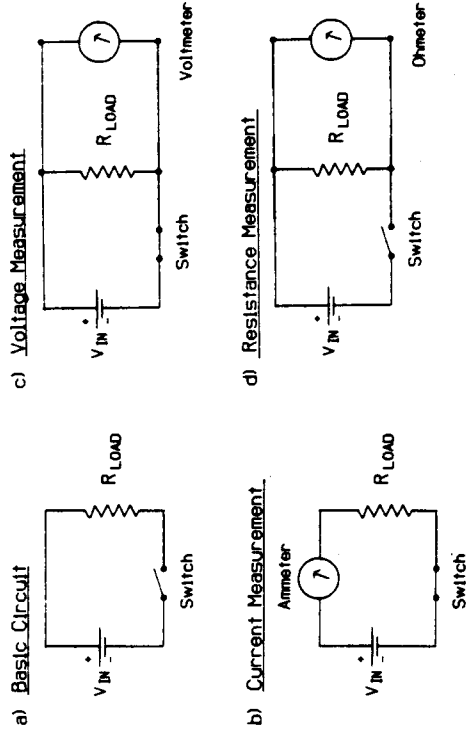


Figure 4.6. Simple circuit consisting of voltage source and load resistance (*a*) and appropriate circuit for measurement of current with ammeter (*b*), voltage with voltmeter (*c*), and resistance with ohmmeter (*d*).

Because most ohmmeter circuits measure resistance by applying a potential and assessing current, it is important that potential sources outside the ohmmeter circuit be switched off when using the ohmmeter. Thus, in Figure 4.6*d* the switch that completes the circuit loop containing the potential source and the load resistance is in the open position. The ohmmeter is also a useful tool for assessing the integrity of electrical connections, since the resistance between two unconnected conductors approaches infinity and the resistance between two connected conductors approaches zero.

Most multimeters can accurately measure both AC and DC currents and potentials but generally measure only resistance. Measurement of impedance (resistance and reactance) requires a specialized form of an ohmmeter that uses AC test currents rather than DC test potentials. The ability to measure resistance and impedance is useful in psychophysiology because the amplitude and integrity of the bioelectrical potentials detected from surface electrodes increases as the resistance/impedance at the electrode interface decreases. More specifically, the skin-electrode barrier acts as if a resistor were placed in series between the biopotential source and the amplification circuitry, thus potentially altering the measurable amplitude and frequency of the voltage signal. In selecting commercial *impedance meters* for psychophysiological applications, it is important that the amplitude of the test currents be well below $100\ \mu\text{A}$, preferably below $10\ \mu\text{A}$, and that the frequency range of the test currents be matched to the frequency range of the physiological signals to be measured. Circuit diagrams for two simple impedance meters can be found in Dunsheath (1982) and in Loeb and Gans (1986, p. 192).

A multimeter measures instantaneous amplitude and polarity of potentials and currents but provides no information about phase or variation with time. Most bioelectrical signals, however, vary almost continuously with time. The *cathode-ray oscilloscope* is an instrument used to simultaneously quantify the amplitude and time components of an electrical signal. An electron gun in the oscilloscope directs a high-velocity focused beam of electrons onto a phosphor-covered glass faceplate.

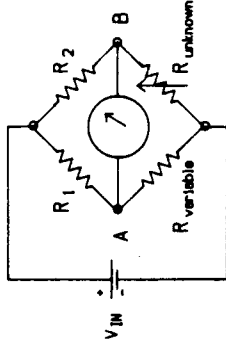
When the phosphor is struck by an electron, it fluoresces, emitting light visible on the X-Y grid coordinates of the faceplate display screen. The beam of electrons can be deflected both vertically and horizontally by pairs of deflecting plates that flank the electron beam. The amount and direction of deflection depend on the voltage applied across the pairs of plates.

Changes in a voltage signal across time can be traced on the oscilloscope display screen by applying the input voltage across the vertical deflection plates and an internal timing voltage across the horizontal deflection plates. Internal timing voltages, also called *horizontal-sweep voltages*, are oscillating ramp signals that show linear increases in amplitude for a selected duration followed by rapid decreases to zero. During each cycle of the horizontal-sweep voltage the electron beam moves across the oscilloscope display screen from left to right, tracing any corresponding changes in the input voltage along the vertical axis. Calibrated vertical and horizontal gain and DC offset controls permit measurements of voltage and time components to be made directly from the display screen. Although the oscilloscope is basically a voltage-sensitive device, current can be indirectly assessed as a voltage change across a fixed resistance or measured directly with a specialized input to the oscilloscope called a *current probe* positioned to encircle the conducting cable and measure the magnetic field produced by the current. Most oscilloscopes can be set to display continuously or to begin displaying when a trigger voltage is detected. Triggering the oscilloscope display from an electrical event can be useful in measurement of event-locked activity. Some oscilloscopes have the capacity to simultaneously display multiple signals as separate channels, and some *storage oscilloscopes* can temporarily fix a tracing on the display screen to facilitate measurements.

When very accurate measurements of resistance/conductance, capacitance, or inductance are required, an instrument with a *bridge circuit* is frequently used. For example, one class of psychophysiological signals, that of electrodermal activity (EDA), results from changes in the resistivity/conductivity of the skin. In addition, bridge circuits are central to many transducer circuits for detection of physiological signals such as respiration and blood volume. Although bridge circuits can be built using resistors, capacitors, or inductors and can be driven by DC potentials or AC currents, all bridge circuits work on the same principle, that of balance. The Wheatstone bridge circuit for measurement of resistance, consisting of a potential source, four resistors, and an ammeter, is illustrated in panel *a* of Figure 4.7. Two of the resistors, R_1 and R_2 , have resistances that are known and fixed, one, R_{variable} , can be adjusted, and the fourth, R_{unknown} , is the resistance to be measured. When all four resistances are equal, the current through each resistor is the same, no potential difference exists between points *A* and *B* in the circuit, and thus no differential current flows through the ammeter. When no differential current is present at the ammeter, the bridge is said to be balanced.

To better understand the conditions under which points *A* and *B* in panel *a* of Figure 4.7 will be equipotential, the Wheatstone bridge circuit is redrawn in the bottom panels of Figure 4.7 as two separate series circuits. Within each of these two series circuits the same current passes through both of the resistors. The current is determined by Ohm's Law to be $V_{\text{in}}/R_{\text{total}}$. Thus, the current through R_1 and R_{var} called I_1 , equals $V_{\text{in}}/(R_1 + R_{\text{var}})$, and the current through R_2 and R_{unknown} , called I_2 , equals $V_{\text{in}}/(R_2 + R_{\text{unknown}})$. Circuits with resistors in series are sometimes called *voltage dividers* because the voltage measured across any one resistor such as R_1

a) Wheatstone Bridge



b) Two Voltage Dividers

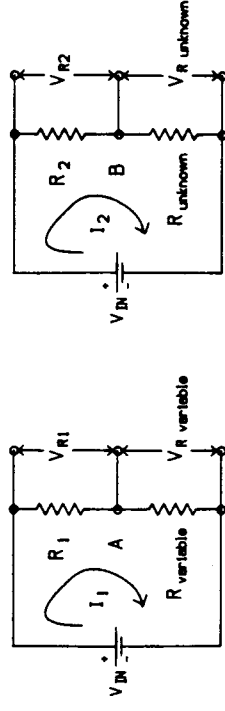


Figure 4.7. Wheatstone bridge circuit (a) and same circuit redrawn as two voltage-divider circuits (b).

is a fraction of the total input voltage V_{in} determined by the ratio of R_1 to the total resistance $R_1 + R_{\text{var}}$, or $V_{R1} = R_1/(R_1 + R_{\text{var}})$. Likewise, the voltage across R_2 is computed as $V_{R2} = R_2/(R_2 + R_{\text{unknown}})$. In order for point *A* and *B* in the circuit to be equipotential, it is not necessary for I_1 to equal I_2 , but it is necessary for the voltage across R_1 to equal the voltage across R_2 . Setting $V_{R1} = V_{R2}$ gives $R_1/(R_1 + R_{\text{var}}) = R_2/(R_2 + R_{\text{var}})$. By cross multiplying and subtracting R_1R_2 from both sides of this equation, the balance conditions for the bridge can be reduced to $R_1R_{\text{unknown}} = R_2R_{\text{var}}$, or the equivalent expression, $R_1/R_{\text{var}} = R_2/R_{\text{unknown}}$.

In most applications of the Wheatstone bridge R_1 and R_2 are fixed resistances either of equal value or switch selectable in ratios differing by powers of 10. In the simplest case where R_1 and R_2 are equal, the bridge will only be balanced when R_{variable} is adjusted to equal R_{unknown} . If a calibrated scale for R_{variable} is provided that accounts for changes in R_1/R_2 , the value of R_{unknown} can be read from the calibrated scale at the balance point. Similar bridge circuits using two fixed and one variable capacitor or inductor can be used to measure unknown capacitances and inductances, respectively.

The Wheatstone bridge circuit is commonly used in the measurement of skin resistance. With this procedure, the subject contributes the unknown resistance. Baseline (tonic) skin resistance levels are measured by adjusting the variable resistance, either manually or automatically, until the voltage (and thus the current) between *A* and *B*, often monitored with a voltage-activated pen recorder, is zero. Once adjustments to offset the baseline skin resistance level have been made, phasic skin resistance responses can be measured as calibrated changes in the voltage between *A* and *B*. Skin conductance, the reciprocal of skin resistance, can be measured directly with a variation of the basic bridge circuit (see Dawson et al., chapter 10).

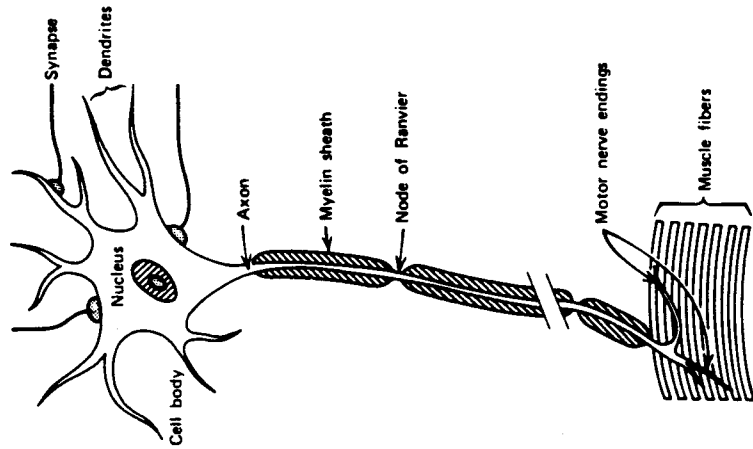


Figure 4.8. Schematic of motor neuron. (Reprinted with permission from J. R. Cameron & J. G. Skofronick, *Medical Physics*, © John Wiley & Sons, Inc., New York, 1978.)

4.5 BIOPOTENTIALS

Bioelectrical signals⁴ generated inside the body control, either directly or indirectly, all functions and activities of the brain, muscles, and organs (see Matsumoto, Walker, Walker, & Hughes, chapter 3). These bioelectrical signals are the result of sudden, reversible changes in the distribution of ions within and outside of specialized cells. Nerve cells, or *neurons* (Figure 4.8), are the major source of biopotentials, but other sources include muscle and cardiac cells. When a neuron is at rest, the cell membrane is differentially permeable to positively charged potassium, K^+ , and negatively charged chlorine, Cl^- , ions compared to positively charged sodium, Na^+ . The Na^+ and Cl^- ions are more concentrated outside the cell whereas K^+ ions are more concentrated inside the cell. In addition, neurons contain concentrations of negatively charged protein molecules too large to pass through the cell membrane. This concentration of protein molecules inside the cell is normally too high to be balanced by the resting distribution of the K^+ , Na^+ , and Cl^- ions. As a result, when a neuron is at rest, the potential inside the cell is more negative, about -70 mV, than the potential outside the cell.

Neurons receive electrical signals from sensory receptors or other neurons

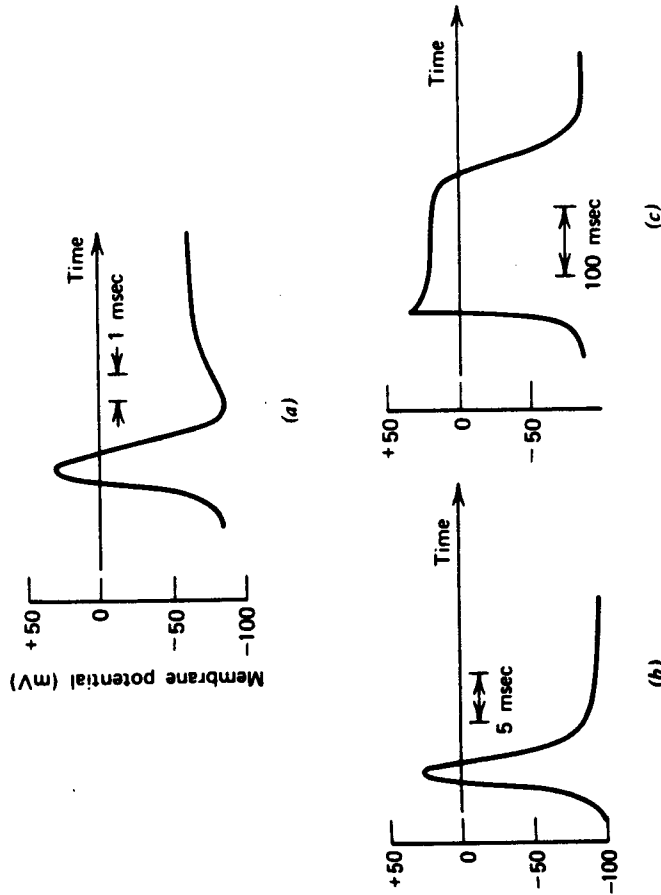


Figure 4.9. Waveform of action potentials from (a) nerve axon, (b) skeletal muscle cell, and (c) cardiac muscle cell. Note different time scales. (Reprinted with permission from J. R. Cameron & J. G. Skofronick, *Medical Physics*, © John Wiley & Sons, Inc., New York, 1978.)

through contacts called *synapses* located on the *dendrites* or the *cell body* (see Figure 4.8). When a neuron is sufficiently stimulated, the cell membrane rapidly increases in permeability to Na^+ ions, permitting the temporary influx of Na^+ ions followed by an efflux of K^+ . This migration of charges produces a temporary increase in the potential inside versus outside the neuron, termed *depolarization*, and a subsequent return to the resting potential due to a compensatory *repolarization*, which are together measured as an *action potential*. Action potentials are then transmitted along a fiber called an *axon* to muscles, glands, or other neurons. The amplitude of a single action potential is less than 100 mV, and the duration ranges from a few milliseconds in the case of nerve axon to 150–300 ms for cardiac muscle cells (Figure 4.9). Immediately after an action potential, a neuron is initially incapable of carrying another action potential and hence is temporarily less excitable, for a period 1 to 3 times the duration of the action potential. The duration of this *refractory time*⁵ appears to vary randomly and permits some, but not all, Na^+ ions to be pumped out of the cell and be replaced by K^+ ions.

Figure 4.10 is a schematic representation of the transmission of a wave of depolarization along an axon and the corresponding potential observed at point *P* on the axon as a function of time. The velocity of conduction of action potentials varies as a function of the axon diameter, the presence or absence of myelin, and the type of cell. First, as with any conductor, the internal resistance of an axon decreases with increases in diameter, and large-diameter axons tend to conduct action

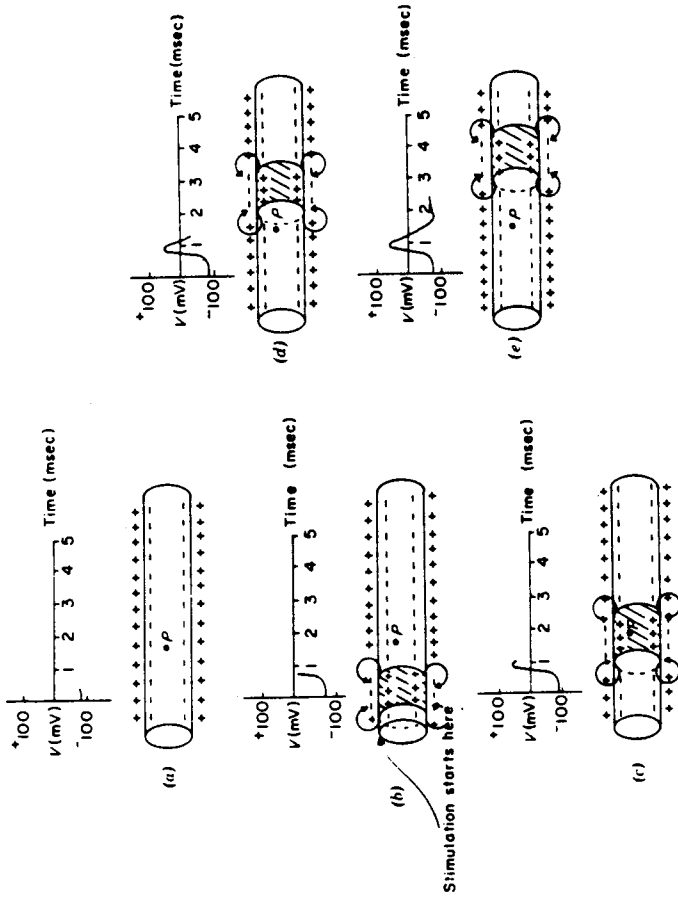


Figure 4.10. Transmission of nerve impulse along axon. Graphs show potential at point P. (a) Axon has resting potential of about -80 mV. (b) Stimulation on left cause Na^+ ions to move into cell and depolarize membrane. (c) Positive current flow on leading edge, indicated by arrows, stimulates regions to right so that depolarization takes place and potential change propagates (d, e). Meanwhile K^+ ions move out of core of axon and restore resting potential (repolarize membrane). Voltage pulse moving along nerve is action potential. (Reprinted with permission from J. R. Cameron & J. G. Skofronik, *Medical Physics*, © John Wiley & Sons, Inc., New York, 1978.)

potentials faster than small-diameter axons. In addition, the membranes of some nerve axons are covered with a fatty insulating layer (*myelin*) that has small uninsulated gaps (*nodes of Ranvier*) every few millimeters (see Figure 4.8). The insulating properties of myelin reduce the capacitive charge stored on the axon membrane. Lower stored charge means faster depolarization, since fewer ions must be moved, and results in correspondingly faster conduction velocities. Action potentials traveling along myelinated fibers appear to jump from one node to the next due to faster conduction velocities between nodes, where myelin is present, compared to within nodes, where no myelin is present. Myelinated human nerve fibers exhibit propagation rates of around 100 m/s, compared to velocities of 0.3 m/s in unmyelinated axons (Curtis, Jacobson, & Marcus, 1972). Much slower transmission rates, in the range of 30 – 45 cm/s, have been observed in human cardiac tissue where transmission delays on the order of 100 ms may be implicated in the synchronization of the action of the upper and lower chambers.

The occurrence of an action potential is an all-or-none event, but subthreshold levels of stimulation can affect the threshold for elicitation of an action potential in a given cell by altering the cell's resting potential. The changes in the resting potential

of cell bodies and dendrites, called *synaptic potentials*, can be graded rather than all or none. An *excitatory postsynaptic potential* (EPSP) lowers the threshold for an action potential by raising the resting potential, and an *inhibitory postsynaptic potential* (IPSP) raises the threshold by lowering the resting potential. Almost all interactions among neurons in the mammalian nervous system occur at synapses, and the EPSP and IPSP are the major kinds of synaptic processes demonstrated in mammals (Thompson, 1975). Another type of synaptic interaction known to occur in the mammalian brain, *presynaptic excitation* and *inhibition*, alters synaptic transmission by modulating the axonal output before it reaches its target cell body or dendrite. Specifically, a modulating axon terminal ends on another axon terminal instead of on a cell body or dendrite and either excites or inhibits the effect of the axonal output of the target cell.

The bioelectrical potentials studied by psychophysicists are typically not single action potentials like those shown in Figure 4.9. One reason for this is that isolation of single nerve and muscle cells is difficult and invasive. A second and more important reason is that the electrical activity of a single cell may not be representative of the activity of a system of interacting cells. Psychophysicists interested in monitoring the activity of a physiological system (e.g., central nervous system, skeletomuscular system) often measure the combined electrical activity of groups of cells located at varying distances from the site of interest. As a result, biopotential signals have waveforms much more complex than the sine waves shown in Figure 4.2. These complex waveforms can be decomposed into a set of sine waves of various amplitudes and frequencies. A Fourier series is a weighted sum of sine waves varying in frequency, amplitude, and phase that characterizes a complex waveform (see Dorfman & Cacioppo, chapter 20; Gottman, chapter 22; Porges & Bohrer, chapter 21).

The amplitude and frequency ranges for several biopotentials of interest to the psychophysicist are shown in Figure 4.11. Note that the amplitude and frequency axes are scaled logarithmically, but the corresponding physical units of millivolts and hertz are also provided. Many of the biopotentials included in Figure 4.11 are described at length in other chapters of this volume. However, a brief description of each source of biopotentials is also provided here.

Electroencephalography (EEG) is the measurement of low-level aggregate electrical activity of the brain, most often detected using sensors placed at specific scalp locations. At present it is widely accepted that the EEG biopotential includes changes in presynaptic and postsynaptic potentials rather than primarily action potentials (Elul, 1972; Li & Jasper, 1953). Typical human EEG signals range in amplitude from 1 to 100 μV and range in frequency from 0.005 to 100 Hz. Note that this is at least 10^8 times smaller than the voltage in a 60 -Hz laboratory AC power line. More detailed information on the origin and characteristics of the EEG can be found in Cooper, Osselton, and Shaw (1980) and Ray (chapter 12).

Electromyography (EMG) is the measurement of the electrical activity of a muscle or muscle fiber that is related to muscle contraction. Electromyography can be detected with sensors placed on the skin over a muscle site or from needle or fine-wire electrodes inserted directly into the muscle. The aggregate EMG signal ranges in frequency from 1 to 2000 Hz. Signal amplitudes for the EMG are higher than that for EEG, ranging from fractions of 1 μV to over 2 mV. With surface electrodes the median frequency is approximately 30 – 140 Hz, a frequency range that spans the 60 Hz frequency for laboratory AC power lines, but as is true of the EEG, at a much lower voltage. More detailed information on specific recording

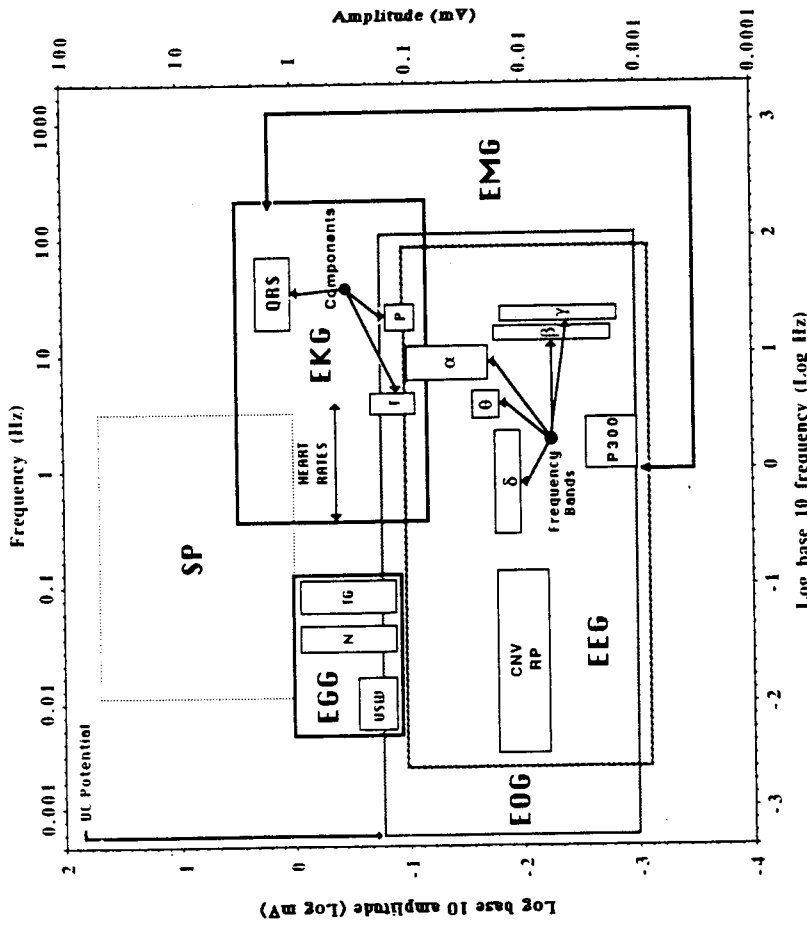


Figure 4.11. Schematized quantitative representation of frequency-amplitude characteristics of major biopotential signals recorded directly from skin surface in humans. Both axes are logarithmically scaled (base 10) and upper and right-hand scales are expressed in corresponding original physical units of hertz and millivolts, respectively. Abbreviations: SP, skin potential; EGG, electrogastragram; USW, ultralow wave; N, normal; TC, tachycardia; EKG, electrocardiogram; EEG, electroencephalogram; CNV, contingent negative variation; RP, readiness potential; ERP, event-related potential; EMG, electromyogram.

techniques and quantification procedures can be found in Basmajian and Deluca (1985), Cacioppo, Tassinari, and Fridlund (chapter 11), and Loeb and Gans (1986). *Electrocardiography* (EKG) is measurement of the electrical activity of the beating heart. The changing electrical potential resulting from propagation of an action potential along the wall of the heart is conducted through the torso and can be detected at the skin surface. The distribution of electrical potential over the entire heart approximates that of an *electric dipole* in which equal positive and negative charges are separated from each other. The instantaneous amplitude and orientation of the heart's dipole potential can be measured as it projects along any axis of the heart. When measured at the skin surface, the heart wave potential is less than 5 mV in amplitude and ranges from 35 to 200 beats per minute. See Papillo and Shapiro (chapter 14) for more information on recording and quantifying cardiac activity.

Electrooculography (EOG) is measurement of the dipole potential of the eye. The EOG can provide an index of eye movements in both the horizontal and vertical planes. The electropotential of the cornea is positive with respect to the retina, and movement of the eye changes the orientation of the eye's dipole field. As is the case with the heart dipole potential, these changes can be detected at the skin surface. The frequency component of EOG signals may be quite slow, at times approximating a DC signal. Amplitudes are generally lower than that of the heart wave potential, in the microvolt to millivolt range. See Oster and Stern (1980), Stern and Dunham (chapter 15), and Young and Sheena (1975) for further information on recording and quantifying the EOG.

Other methods for assessing biopotentials in psychophysiology include *electrogastrography* (EGG), measurement of the electrical activity accompanying gastric contractility (see Stern, Koch, & Vasey, chapter 16), and *skin potential* (SP), measurement of biopotential differences between electrodermally active sites such as the fingers, palm of hand, or sole of foot and an inactive site (see Dawson, Scheil, & Filion, chapter 10).

4.6 INSTRUMENTATION FOR PSYCHOPHYSIOLOGICAL DATA COLLECTION

Before beginning any discussion of instrumentation for the psychophysiology laboratory, it is important to make some general suggestions regarding electrical safety. Two primary concerns in the design of a psychophysiological data collection system are the safety of the subject and the integrity of the data. Fortunately many of the procedures essential for the subject's safety from electrical hazard also help reduce the level of electrical noise (unwanted components in the recorded signal). As mentioned earlier, it is the level of current flow through the body that determines the amount of damage to an organism caused by electricity. Recall that with a regulated voltage source the level of current flow is governed by the resistance of the attached load.

Alternating current sources and AC-powered equipment pose a particular risk because they provide a high-current source. Two procedures are routinely used to protect subjects from dangerous currents, isolation and grounding. Within the psychophysiology laboratory the subject should be physically isolated from all AC sources and, whenever possible, all AC-powered equipment. Many commercial laboratory instruments use optically or magnetically coupled power supplies that transfer power to the instrument without a direct current path to the AC power source to help isolate the operator from hazardous currents. The use of low-current, battery-operated lighting is both safer for the subject and less electrically noisy than using nearby AC-powered lamps. When overhead AC-powered lighting is desirable, grounded wire mesh shields can be placed around the lights to minimize the spread of electrical noise.

Since it is not always possible to completely isolate a subject from high-current sources, it is essential that the subject and all high-current equipment close to the subject or attached to the data collection system be properly grounded. A single ground lead is attached to the subject, and all electrode connections are protected from any current flow above 10 μ A by a current-limiting device that automatically interrupts the connection between the subject and the recording system when higher currents are detected or that simply does not conduct higher currents. Only triple conductor power extension cords should be used, and all AC-powered

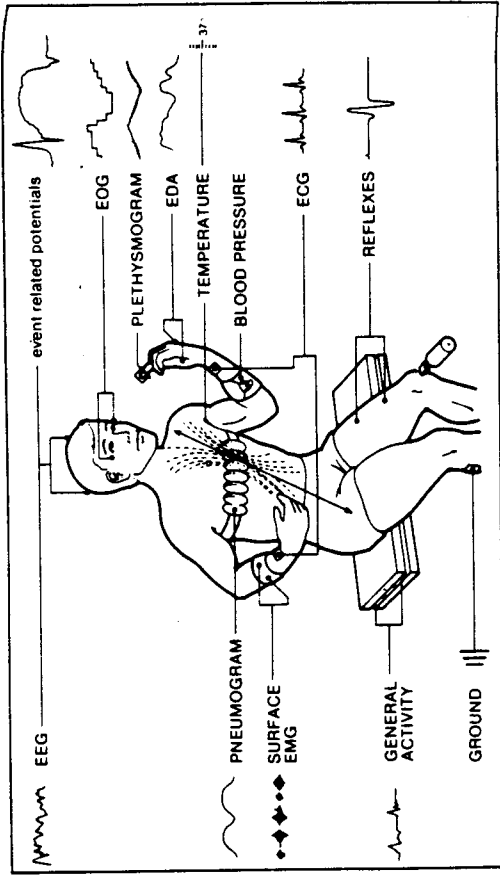


Figure 4.12. Commonly used measures in psychophysiological research. (Adapted with permission from J. Piliard, *Psychophysiology du comportement*, Vol. 3, © Presses Universitaires de France, 1966, and reprinted in C. H. M. Brunia, *Activation*, in J. A. Michon, E. G. J. Eijkman, & Len F. W. Klkerk (Eds.), *Handbook of psychoneuroeconomics*, Vol. 1, © Amsterdam, North-Holland, 1979.)

equipment with a three-prong AC plug must be properly attached to the building power ground rather than attached to ungrounded two-prong adaptors. For additional information on electrical safety precautions, see Spooner (1980).

4.6.1 Detection of physiological signals

The psychophysiological categorizes physiological signals according to the physiological system of origin. A representative sample of response systems monitored by psychophysiologicals along with sample tracings of the amplitude by time recordings is provided in Figure 4.12. The systems illustrated include, for example, the nervous system, the skeletal muscular system, the cardiovascular system, the respiratory system, the gastrointestinal system, and the endocrine system. The nervous system is often further subdivided into central and peripheral, autonomic and somatic, sympathetic and parasympathetic (see Matsumoto, Walker, Walker, & Hughes, chapter 3). However, when discussing instrumentation for the detection of physiological signals, it is more useful to group signals according to electrical and physical properties. By describing physiological signal detection according to the properties of the signal rather than according to the site of origin, it is possible to derive appropriate detection procedures for physiological responses not specifically described here. Three categories of detection procedures are described here: (1) measurement of biopotentials originating in body tissue (e.g., EEG, EMG), (2) measurement of bioelectrical phenomena other than potentials (e.g., skin resistance/conductance), and (3) measurement of physical (nonelectrical) change (e.g., blood pressure or volume, respiration, temperature).

When directly sensing biopotentials such as the EEG and EMG, some form of

electrode is used to detect potential differences between two body (usually skin or scalp) locations. Detection of physiological signals other than potential (e.g., skin conductance, blood pressure) requires a *transducer* to convert the form of energy produced by the body to an electropotential analog of the desired signal. This is because most electronic recording and storage equipment respond only to voltage changes. The transduction of bioelectrical phenomena other than potential, such as measurements of electrical properties of the skin (resistance, conductance, impedance, etc.) and changes in impedance that accompany respiration (*impedance pneumography*), peripheral blood volume (*impedance plethysmography*), cardiac stroke volume (*impedance cardiography*), and brain blood volume (*impedance rheoencephalography*) is based on Ohm's Law relating resistance and impedance to voltage. Specialized transduction systems have been developed for nonelectrical phenomena such as blood pressure, blood flow, respiration rate or flow, cardiac sounds (*phonocardiography*), volume or girth plethysmography, tissue temperature, gastric motility, and body and limb movement.

The selection of specific *biosensors*, whether electrodes or transducers, should be guided by the desired application including not only the frequency and amplitude characteristics of the physiological signals of interest, but also such factors as the placement site on the body, the duration of the recording session, and the amount of subject movement required during recording. Six important design parameters of all dynamic sensors are (1) sample loading, (2) output impedance, (3) damping, (4) frequency response, (5) linearity, and (6) noise (DeMarre & Michaels, 1983). All six parameters are interrelated, and most sensor systems fall short of the ideal on one or more of them. Each is discussed here in turn.

Sample loading, the effect of the sensor on the system it is measuring, should be minimized to avoid distortion of the process of interest. For example, electrodes for detection of facial muscle movements need to be small and lightweight so that they do not restrict or inhibit movement.

The *output impedance* of a sensor is the combined DC and AC resistances measured between the two output leads of the sensor. For electrode systems, this includes the impedance of the body tissue between the two electrodes. In most electronic applications it is desirable to match the output impedance of each component in the system to the input impedance of the next component in the circuit for maximum power transfer. However, with sensors attached to living organisms, matching the output impedance of the sensors with the input impedance of the signal amplifier maximizes the level of current allowed to pass between the subject and the amplifier. To prevent damaging current flow between the subject and the amplifier, it is recommended that electrode output impedances be at least 10^6 times less than the input impedance of the amplifier. Because the resistance of the skin imposes a lower limit on output impedance for surface electrodes of the order of $10^{10}\Omega$, bioelectrical amplifiers are generally designed with input impedances on the order of $10^{10}\Omega$ (range $10^7 - 10^{12}\Omega$). Electrode output impedance also acts as a barrier to the transmission of biopotentials, so further reduction in electrode output impedance is desirable to increase sensitivity to the physiological signal. When reduction of electrode output impedance is not possible, as is sometimes the case with subdermal microelectrodes, high-impedance couplers can be placed between the electrode and the amplifier to prevent excess electrode currents and minimize distortion of low-frequency signals.

Damping refers to the extent to which a sensor faithfully reproduces the

frequency components of the input physiological signal. In transducers of physical movement, damping may result from inertia, producing a slowed response to rapid initiation and cessation of movement. A critically damped sensor neither overreacts (the underdamped state) nor underreacts (the overdamped state) to the input signal. The *frequency response* of a sensor is related to the damping and refers to the range of frequencies to which it will respond. The effective bandwidth of frequencies for a sensor should be selected to approximate the frequency range of the event being measured.

The ideal relationship between the amplitude of physiological input to a sensor and the amplitude of the electrical output to the amplifier is a linear one. Although obtaining *linearity* within restricted amplitude and frequency ranges is not typically a problem with bioelectrode systems, it is important that the range of input amplitudes and frequencies over which a transducer provides linear output be matched to the range of physiological input amplitudes and frequencies.

The last design parameter, that of *noise*, refers to the extent to which a sensor either produces or conducts undesired signals. Noise is any unwanted signal and may arise from the sensor itself, from physiological sources such as limb movement, and from ambient electrical noise generated by the proximity of necessary AC-powered equipment such as lighting, video display monitors, and so on. Ambient electrical noise may be impossible to eliminate, but it can be attenuated using grounded shielding on all power sources to reduce magnetic flux, and pickup of ambient noise can be reduced using grounded shields on all signal cables. In addition, the use of twisted pairs of sensor leads that are then exposed to the same ambient electrical noise can be applied to differential amplifiers, described later, which magnify the difference between the two signals, attenuating the noise component common to both leads.

4.6.1.1 Electrode systems

Biopotential electrodes may be separated into three categories based on their physical construction: (1) microelectrodes, (2) needle or fine-wire electrodes, and (3) skin surface electrodes. Microelectrodes for recording from a single cell have a fine point in the micrometer range and can be constructed from a micropipette and filled with a conductive electrolyte. The use of microelectrodes is most common in physiology, and their use in psychophysiology is quite limited. Stainless steel or platinum needle and fine-wire electrodes are sometimes used for human EEG and EMG recording and are used frequently in animal research. Skin surface electrodes, the most prevalent type in human psychophysiology, pose the least risk to the subject but are the most distant from the signal source(s) and are separated from the source(s) by the skin, which acts as an electrical insulator.

The ideal biopotential electrode is a perfect conductor that provides a stable current path between the source of the desired physiological signal and the laboratory data collection system. The conducting material must be inert to the chemicals on or in the body to prevent polarization, an accumulation of ions on the electrode surface that blocks conduction of the signal due to capacitive reactance as well as reducing the possibility of toxic reactions. For this reason, less chemically active metals such as silver and platinum are preferable over active metals such as zinc and nickel. Although more reactive metals such as copper tend to be electrically consistent over time, *generating* less irrelevant electrical potentials than

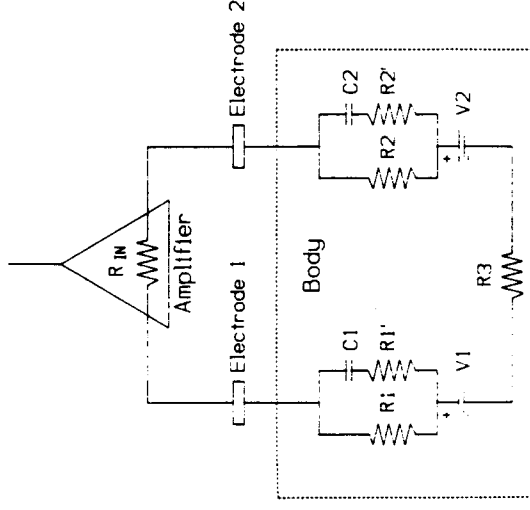


Figure 4.13. Equivalent circuit for pair of surface electrodes. (Adapted with permission from J. J. Carr & J. M. Brown, *Introduction to biomedical equipment technology*. © John Wiley & Sons, Inc., New York, 1981.)

the less reactive metals, their use is limited to very short term exposures (DeMarre & Michaels, 1983). Silver-silver chloride (Ag-AgCl) surface electrodes are highly recommended because they provide good current transfer with minimum polarization. Typical silver-silver chloride electrodes consist of circular disks of silver electroplated with silver chloride or of cylindrical blocks of sintered particles of silver and silver chloride.⁵

When using surface electrodes to measure low-level biopotentials (EEG, EMG), it is important to reduce the natural insulating properties of the skin by thoroughly cleansing it with alcohol to remove any dirt or oils and lightly abrading to remove the surface layer of dead skin cells. In addition, an electrolyte paste or gel is used between the skin and the surface of the electrode to improve the conductive path from underlying tissues to the external electrode and to maintain a stable electrode interface. Some of the electrolyte penetrates the skin at the electrode site, and the remaining electrolyte forms a conductive bridge between the electrode surface and the skin surface that is not easily disrupted by bodily movements. When measuring higher amplitude biopotentials (EKG), abrading the skin is not necessary.

The equivalent circuit for a pair of biopotential electrodes is shown in Figure 4.13. The biopotential electrode surface and the skin surface form a chemical half-cell (battery) between the metal in the electrode and the electrolytic fluid beneath the electrode, capable of developing an electrical potential at each electrode. The electrode offset potential observed at the amplifier is the difference between V_1 and V_2 . The amplitude and stability of this electrode potential depends mainly on the metallic content of the electrode. The electrode potential in series with the biopotential being measured adds DC offset to the biopotential signal. Large and variable DC offset potentials can mask the characteristics of the biopotential and may need to be removed from the signal at the time of signal conditioning. When

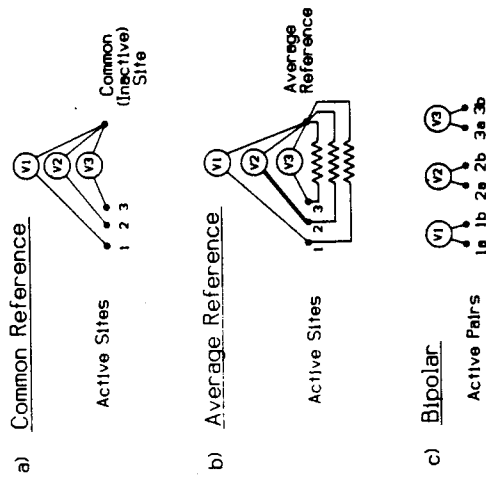


Figure 4.14. Common reference (a), average reference (b), and bipolar (c) placement patterns for electrodes.

the electrodes are identical and differential amplifiers are used, electrode offset potential does not pose a problem to detection of physiological signals. In addition to the signal distortion produced by electrode potentials, the insulating properties of the skin that are not fully overcome by cleaning and abrading at each electrode site add resistances R_1 and R_2 in series with the electrodes that attenuate the level of the output signal. The body fluids and tissue under the skin between the two electrodes develop an additional resistance, R_3 . Finally, a double layer of ionic charges may add at the electrode interface, producing an effective capacitance, C_1 and C_2 , and a corresponding capacitive reactance, R_1' and R_2' . Capacitive reactance is also called the polarization impedance. Recall that capacitors appear as open circuits to DC signals, and current flow increases directly with increases in signal frequency. Thus, fast AC-like biopotentials produce greater current flow across the electrode interface than slower DC-like signals. What is not clear from Figure 4.13 is that the resistances and capacitances are not stable either between subjects or within a single subject across time. However, since the electrical parameter of interest is voltage and not current, it is neither necessary nor desirable to maintain high current flow at the interface between the electrode and the skin. Amplifiers with high input impedance, R_{in} , minimize the current flow between the electrode and thus attenuate the variable effects of electrode polarization.

Three types of placement patterns are used with biopotential electrodes: common reference, average reference, and bipolar (Cooper, Osselton, & Shaw, 1980). All three techniques compare the potential between a set of electrodes. With common reference placements, one electrode is placed over an electrically active site, and the other is placed over an electrically inactive site (see Figure 4.14, panel a). The output of common reference electrodes yields information about the absolute level of electrical activity at the active site. For example, in EEG recording, each individual active scalp placement may be compared to one electrode placed at a relatively inactive site such as the mastoid area.

With average reference placements, the activity at each electrically active site is

compared to the average of the activity at all electrode sites used. The average reference signal is generated by connecting each active signal electrode through equal resistors to a single point (see Figure 4.14, panel b). If the potentials at the active sites are unrelated, the average reference potential tends toward zero as the number of active electrode sites increases. However, widespread synchronous activity that affects many active sites or large-amplitude artifacts at a single site will raise the amplitude of the reference potential. The output of average referenced electrodes yields information about the level of electrical activity unique to each active site relative to the level of overall activity.

With bipolar placements, the two electrodes are placed close together over an active site (see Figure 4.14, panel c). This placement pattern is particularly sensitive to localized changes in electrical activity. Bipolar placements are common in EMG recording, with the two electrodes positioned parallel to the muscle fibers close to the middle of the muscle. When a series of electrodes is placed in a straight line across an electrically active site, montages formed from the potential between adjacent pairs of electrodes yield information about the gradient of the electrical potential across the active site. In EKG recording, a variety of bipolar placements are possible, ranging from a simple two-electrode arrangement using sites on the right arm and the left leg or the right collarbone and left lower rib when only heart rate is required, to more complex three- and five-electrode arrangements when time and amplitude components of the complete cardiac waveform are required. Each pair of EKG electrodes spans a different axis of the heart, and in clinical settings the type and site of cardiac disorders can be diagnosed from the resulting pattern of waveform anomalies.

In addition to the detection of biopotentials, electrode systems are used to detect other electrical properties such as resistance and impedance. The electrodes used for detection of skin conductance/resistance and for low-frequency (less than 5 Hz) skin impedance changes require the same characteristics as biopotential electrodes. That is, they should be electrochemically stable and nonpolarizable. The measurement of DC resistance (and its reciprocal, conductance) and/or impedance (and its reciprocal, admittance) at the skin's surface requires the application of known signals between two surface sites. Most often a constant (low) voltage is applied and changes in current are monitored, but the reverse is also possible. However, for impedance measurements using high-frequency AC signals (greater than 10 kHz), such as impedance pneumography, plethysmography, cardiography, and rheoencephalography, the speed at which the direction of current flow changes eliminates the polarization problem. Both copper and aluminum electrodes can be used for impedance plethysmography. Though nonpolarizable electrodes are not necessary with high-frequency measurements, they can be used as well.

The detection of skin conductance/resistance is described in detail elsewhere (see Dawson, Schell, & Filion, chapter 10), but the general procedure is described briefly here to illustrate the application of electrical principles to detection of bioelectrical responses other than potentials. It is generally agreed that electrodermal activity is related to eccrine sweat gland activity, and it is affected by the level of hydration of the skin (Fowles, 1986; Venables & Christie, 1973). Since hydration of the skin is the peripheral physical parameter of interest, a special-purpose electrolyte is used that closely approximates the ionic concentrations in the body fluids and thus does not hydrate the skin (Fowles et al., 1981; Schneider & Fowles, 1978). The electrode site is cleansed with soap and water to remove surface dirt and oil, but procedures that

might alter the inherent conductive/resistive properties of the skin are avoided. Specifically, cleansing with alcohol, which causes drying, and abrading the skin, which removes an insulating layer of skin, are not recommended. Two identical electrodes on electrodermally active sites (bipolar) can be used or one electrode at an active site such as the palm can be compared to another electrode at an inactive site such as the earlobe (common reference). Either a constant current (e.g., 8 $\mu\text{A}/\text{cm}^2$) or a constant voltage (e.g., 0.5 V) is applied across the two electrodes, and a type of bridge circuit is used to quantify changes in resistance or conductance.

Two additional electrode systems for bioelectrical responses also merit discussion here: impedance pneumography and impedance plethysmography. In impedance pneumography, a low-voltage 50–500 kHz AC signal is applied to the subject via a pair of surface electrodes on the chest (Carr & Brown, 1981). Large fixed resistors are connected in series with each electrode, creating a constant-current source. The voltage drop across these fixed resistors represents the thoracic impedance. Without respiration the current through the subject's chest is nearly constant, but it changes with respiration. The impedance pneumograph is a reliable and valid indicator of respiration rate and relative amplitude; however, absolute measures of respiration amplitude are possible over only a limited portion of the range of observed tidal volumes (see Lorig & Schwartz, chapter 17).

In impedance plethysmography, changes in the instantaneous blood volume of the whole body or selected limbs are inferred from changes in impedance between pairs of surface electrodes. Impedance cardiography is a special case of plethysmography for cardiac stroke volume. Impedance rheoencephalography is a special case of plethysmography for cranial (forehead or mastoid) sites. Detection of whole-body and single-limb blood volume changes typically use pairs of band electrodes that encircle the site of interest such as the neck or the abdomen and thorax. With whole-body and limb measurements, a typical applied current is a 100-kHz, 6-mA signal (DeMarre & Michaels, 1983). Circular disk electrodes are common for cranial sites, with applied currents of 1–10 mA at 30–70 kHz (Geddes & Baker, 1975). Although absolute blood volume measurements are possible, percentage or ratio change measurements are more common in psychophysiological research.

4.6.1.2 Transducer systems

Geddes and Baker (1975, p. 3) describe a transducer as a "sense organ for the electronic processing equipment." The meaning conveyed by this statement is that organisms as well as laboratory equipment only process information of a specific type, and both require peripheral preprocessors to convert incoming signals from various physical forms to an interpretable signal. The range of nonelectrical physiological changes of interest to the psychophysiologicalist include but are not restricted to blood pressure and flow, respiration rate and flow, skin or core temperature, and body or limb movements.

Transducers are by necessity specialized for detection of distinct types of physical change, so that some are useful for measurement of variation in pressure whereas others are sensitive to changes in temperature or mechanical displacement. In addition, any electrical parameter (most often resistance) may be made to vary as a function of the signal of interest. As is the case with measurement of any resistive properties (e.g., resistance/conductance, impedance), most transducers require

some type of bridge circuit such as the Wheatstone bridge described earlier for accuracy at low levels and often require a coupler capable of offsetting changes in DC potential. In addition, in order for the absolute amplitude of the transducer output to be meaningful, it must be calibrated to some physical or physiological standard.

Two general guidelines for transducer selection are offered (Seippel, 1983). First, when measuring effort or potential variables like pressure, force, or voltage, select a transducer that greatly impedes motion or flow. For example, a simple transducer system for momentary blood pressure (cuff auscultation) uses an inflatable cuff to temporarily occlude arterial blood pressure. The pressure in the cuff can be adjusted until it matches the pulsatile systolic blood pressure indicated by disappearance of the pulse sound and then reduced until it matches the arterial pressure indicated by Korotkoff sounds. Likewise, voltmeters and oscilloscopes, both voltage-sensitive devices, are designed with high input impedance to minimize the level of current diverted from the circuit of interest. More generally, when measuring voltage, it is important to minimize the current through the measurement device. Second, when measuring motion or flow, select a transducer that readily allows motion. Thus, for example, proper use of the chest strain gauge for detection of respiration rate requires that the chest band not restrict movement. The nasal thermistor method for respiration detection is another example of a low-resistance transducer.

A wide variety of resistive transducers are available. The simplest ones use linear or rotary *potentiometers*, variable resistors that change resistance with mechanical movement or rotation. Volume controls and dimmer switches are common types of potentiometers. In psychophysiology rotary potentiometers have been used to quantify movements such as eyeblinks (e.g., Braff et al., 1978). For measurement of very small movements, resistive changes due to physical distortion of the metal within a *strain gauge* can be used. For example, one method for transduction of respiration uses a mercury strain gauge attached to a chest band. The band is adjusted so that movement of the chest cavity associated with respiration produces corresponding changes in the length of a flexible tube filled with mercury. The resistance of the mercury increases as the tube lengthens and reduces the cross-sectional area. Because the resistivity of metals exhibits a considerable degree of temperature dependence, metallic *resistance thermometers* and metal-metal oxide *thermistors* are also effective transducers for core and skin temperature. These devices change in resistance as a function of changes in temperature. For example, nasal thermistors can be used to detect respiration rate based on the cyclic cooling and heating that results from the movement of air during nasal breathing (see Lorig & Schwartz, chapter 17).

Three additional types of transducers deserve mention here because of their uses in psychophysiological research: photoelectric, magnetic, and ultrasonic transducers. Each of the three transducer types employ an energy source or transmitter, an energy receiver, and a medium through which the energy must travel. In psychophysiological applications the medium is the body. For example, a light source and a photoconductive cell placed on opposite sides of a finger or an earlobe can transduce changes in blood volume as the reciprocal of light energy reaching the receiver. Likewise, magnetic and ultrasonic sources and receivers can be used to transduce changes such as blood volume in larger body regions not penetrable by light, such as the brain.

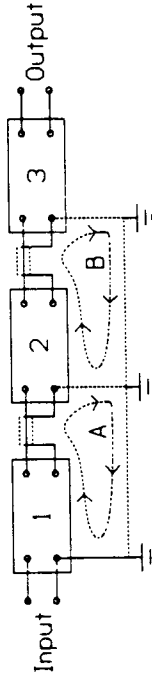


Figure 4.15. Ground-loop circuits resulting from use of multiple grounds in multiple-stage signal-processing system.

4.6.2 Bioelectrical amplifiers and signal conditioners

Following the detection of the physiological signal of interest, it is necessary to selectively magnify and shape the electrical signal to permit display, quantification, and storage. A variety of *signal processors* are used in psychophysiological research that perform mathematical transformations on the detected signal. Signal processors are attached in parallel with the two signal leads of the electrode or transducer and typically provide a single-ended (zero-referenced) output of the transformed signal. In the event that multiple stages of signal processing are required, as is often the case in psychophysiological research, all grounded components of a given stage should be returned to a single ground point, preferably at the initial input stage. If multiple grounds are used, so-called *ground loops*, or current paths through the metal chassis, are formed (Figure 4.15). Large ground loop currents introduce *cross-talk* (undesired signal coupling between stages) that can mask the desired signal.

One essential signal processor for physiological signals of psychological interest is an *amplifier* that increases the amplitude (voltage) of the signal. Other special-purpose signal processors include *filters* that selectively pass or prevent passage of certain frequencies of the input signal, *integrators* that average input amplitudes over a selected time period, and *amplitude-sensitive rate devices* that are triggered when the input signal exceeds a preset threshold. The characteristics and applications for these four signal processors are described in turn.

The low voltage levels for all biopotentials makes amplification the first and most crucial of all signal processing. In addition to the initial or *preamplification* stage, other amplifiers and signal conditioners may be chained in parallel with the input to further magnify and shape the signal. The critical design parameters for preamplifiers (and to some extent, all signal processors) are (1) input impedance, (2) output impedance, (3) gain, (4) common-mode rejection ratio, (5) bandwidth, (6) power distribution, (7) distortion, and (8) noise (DeMarre & Michaels, 1983).

The need for amplifiers with extremely high (10–50 M Ω) *input impedance* was pointed out in the earlier section on detection methods. This is because the electrical parameter of interest is voltage, not current. To maximize sensitivity to changes in voltage, current flow through the measurement instrumentation must be minimized. Minimizing current levels also poses less risk to the subject, since it is current, not voltage, that can damage tissue. Amplifier *output impedance* should be selected to produce an output signal that is an exact amplified replica of the input signal of interest while remaining 10 times less than the input impedance of the next stage of signal processing or recording. Note that successive stages of signal amplification and other processing are chained together in parallel, and each stage is designed to be sensitive to changes in voltage. Thus, the output impedance of each signal processor in the chain is maintained at a level lower than that of the

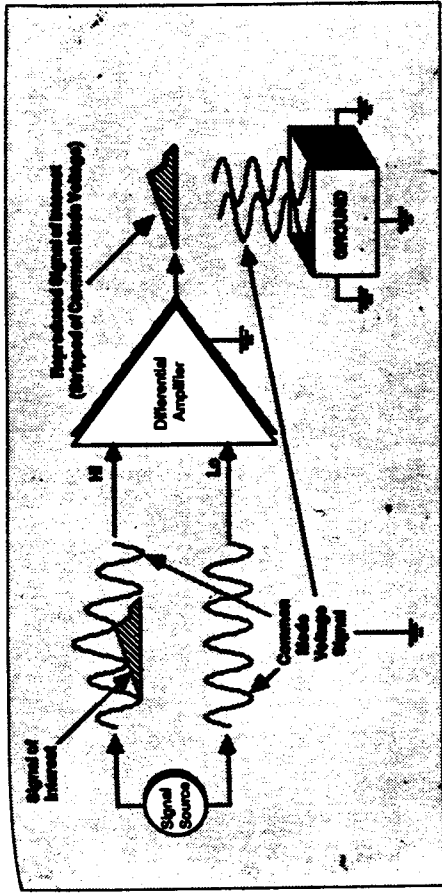


Figure 4.16. Differential amplifier used to reject voltage signal common to two electrode sites. (Reprinted with permission from *Codas analog to digital converter selector guide: application note 4*, © Dataq Instruments, Inc.)

following stage for the same reasons input impedance is maintained at a higher level than the preceding stage, to produce a maximum sensitivity to changes in voltage rather than changes in current. If one stage in the signal-processing sequence is a device with a naturally low input impedance, an additional stage of amplification with high input impedance and low output impedance, called *buffering*, is used to adjust impedances.

The *gain* of a signal processor is the ratio of signal input to signal output for voltage, current, or power and can be (1) fractional, indicating a reduction in the signal level; (2) 1, indicating no change in the signal level; or (3) greater than 1, indicating an increase in signal level. Fractional gains are also called *negative gains*. Bioelectrical amplifiers typically have positive voltage gain, negative current gain, and negative power gain. In practice, several positive voltage gains may be switch selectable. Gain factors are typically categorized as low (10^0 – 10^1), medium (10^1 – 10^3), and high (10^3 – 10^6).

Differential amplifiers are signal-processing devices that selectively magnify the potential difference between two inputs and are extremely useful when the signal amplitude is small relative to the ambient electrical noise. The quality of differential amplifiers is rated by the extent to which they reject or attenuate signals common to both inputs. The effect of differential amplification is illustrated in Figure 4.16. It is not possible for a differential amplifier to completely eliminate common inputs, and the *common-mode rejection ratio* (CMRR), computed as the output amplitude divided by the common input amplitude, is a dimensionless index of the level of attenuation achieved. For measurement of EEG, EMG, and EKG, common-mode rejection ratios of 90–130 dB are recommended.

Amplifiers can effectively process only a limited range of input frequencies. *Bandwidth* and *power distribution* together quantify the frequency response of an amplifier. The bandwidth is the effective range of input frequencies the amplifier is capable of boosting, and the power distribution is a function of the gain at each frequency in the bandwidth. The gain rating provided for an amplifier is the amount

of amplification for most input frequencies in the middle of the bandwidth. Because amplifier gain drops off gradually rather than completely at frequencies above and below those frequencies for which the amplifier is maximally responsive, specification of an amplifier bandwidth requires selection of a criterion for the minimum-gain acceptable percentage of the rated (peak) gain. Two common criteria for amplifier bandwidth specification are ± 3 and ± 6 dB, which include those frequencies where the gain is no less than 70.7 and 50 percent of the rated (peak) gain respectively. Bandwidths of 0.01 Hz–20 kHz are common for bioelectrical amplifiers.

Bioelectrical amplifiers are either DC and AC coupled. Thus, for purposes of signal amplification and processing it is useful to divide bioelectrical signals into two general categories, fast-wave signals and slow-wave signals. Some of the biopotentials shown in Figures 4.11 and 4.12 such as EOG and EEG can include both fast-wave and slow-wave components, and separate but parallel DC-coupled and AC-coupled amplification and processing may be required to detect both signal components from a single set of electrodes. When the input signal to the amplifier remains constant for long periods of time or changes very slowly, as with EGG and electrodermal activity, some form of DC coupling is required. When the input signal changes at frequencies of 0.05 Hz or higher, AC coupling is recommended to eliminate DC offset potentials at the sensors. Fast-wave signals include EMG and EKG.

Regardless of the techniques used, all signal processing introduces some *distortion*. However, the most noticeable distortions of a processed signal occur at the sensor and at the display rather than at the amplifier. If the amplifier is distorting the signal more than 1 percent, it is often traceable to improper impedance matching between two of the stages of amplification. With commercial modular signal-processing equipment using integrated circuitry, amplifier-induced distortion is not a problem. What is a problem is amplification of ambient electrical noise detected at the sensor or induced through unshielded leads. Precautions such as removal of unnecessary AC equipment near the subject, shielding of necessary equipment, use of grounded shields on signal lines from the sensor to the preamplifier, and differential amplification will help, but some signal conditioning to remove noise may also be necessary.

Electronic *filters* are used to selectively attenuate unwanted frequency components of signals. Filters are characterized by the frequency range they pass or reject. Thus, *high-pass filters* selectively attenuate frequencies below a selected cutoff frequency, and *low-pass filters* selectively attenuate frequencies above a selected cutoff frequency. With EEG recording it is often useful to use a low-pass filter cutoff of 30 Hz to reject high-frequency signals above the physiological range for beta waves. Restricted-range *band pass* and *band reject filters* are also available.

A particularly useful band reject filter known as a *notch filter* selectively attenuates signals in the frequency range of 58–62 Hz, the band surrounding 60-Hz AC noise. However, as was noted earlier in the section on biopotentials, 60-Hz filtering is not always appropriate. A major part of EMG signals is in this frequency range, and 60-Hz filtering of AC ambient noise will also remove 60-Hz frequency components of the EMG signal.

When recording psychophysiological signals that include very high frequency components, such as the EMG, it can be useful to electronically time average the input signal prior to display or storage of the signal. *Integrators* are a special type of

low-pass filter that rectify the input (invert the negative components of the signal) and store a cumulative sum of the resulting positive amplitudes for selected periods of time. The storage of these signals decays exponentially with time, and the time taken for the output to decay to approximately 67 percent of the input amplitude is known as the *time constant*. *Amplitude-sensitive rate devices* are also useful for high-frequency signals when the amplitude components of the signal are not required. The output for rate devices is typically a state change (stepwise low to high or high to low voltage) or a brief pulse of fixed duration. The Schmitt trigger is an example of a rate device used for determination of heart rate. The trigger threshold is adjusted so that only the occurrence of the R-spike component of the heart wave is detected, and heart rate is either computed from the time between state changes or read directly as the calibrated amplitude of the output signal (see Papillo & Shapiro, chapter 14).

4.6.3 Display and storage systems

Once the psychophysiological signal of interest has been detected and amplified, it is necessary to display and store a representation of the data to permit inspection for artifacts, examination of the topographical features, and extraction of dependent variable measures. For some applications such as stimulus-evoked muscle action potentials, the display screen of a storage oscilloscope may be sufficient to permit detection of the presence or absence of component reflexive responses. When a permanent record of the signal is required, a special oscilloscope camera can be used to photograph the display in single frames or as a continuous-motion picture. For many psychophysiological applications in which multiple physiological response systems are monitored for periods of from several seconds to an hour or more, an analog recorder is particularly useful. The two more common analog recording systems used in psychophysiology are graphic recorders and magnetic tape recorders. Other types of storage systems for physiological data include card and tape punches, videotape recording, and transient recorders with digital memory. For more information on these systems, the reader is referred to Goovaerts, Ross, and Schneider (1979).

All graphic recorders include an electromechanically controlled stylus that converts a voltage input to mechanical movement and a paper drive assembly to move the chart paper past the stylus at a constant speed. Graphic recorders differ in the number of signals that can be simultaneously recorded and in the method of transcription (e.g., ink writing, pressure writing, thermal writing, photographic writing, electrostatic writing). Both ink and pressure writing systems severely attenuate high-frequency inputs (over 100 Hz for ink writing systems, over 250 Hz for pressure writing systems). This limitation does not significantly affect recording of slower frequency signals such as EEG or EDA, but does affect recording of the nonintegrated EMG. Ink-jet recorders can accurately reproduce frequencies up to about 1 kHz, and photographic and electrostatic recorders are accurate to about 5 kHz. To overcome this limitation, some graphic recorders now include memory modules that permit brief data sampling followed by data recording at a slower rate, resulting in undistorted reproduction of higher frequencies.

Magnetic tape recording systems continuously store coded patterns on a moving magnetic tape. These patterns can be reproduced later at a different speed, allowing time expansion or compression. The magnetic code cannot be visually inspected

learns only to apply existing technology may be at a disadvantage when the inevitable technological advances occur. On the other hand, the psychophysiologicalist who understands the principles underlying the procedures for bioelectrical measurement will be in a much better position to incorporate and apply new technologies.

NOTES

- 1 Additional information on the physical and chemical properties of electricity can be obtained from most introductory physics or physical chemistry textbooks. However, for information on the behavior of simple circuits, an electronics textbook such as Brophy (1983) or a self-teaching guide such as Kybett (1986) is recommended.
- 2 Conceptualizing resistors as impediments can be confusing because of the implicit assumption that the more resistors in the circuit, the less the overall flow. As elaborated in section 4.3, adding resistors in parallel actually increases flow rather than decreases flow (see Gentner & Gentner, 1983, for a useful discussion on the limitations of heuristic models of electricity).
- 3 Ionic current flow through a cell membrane can be assessed using voltage clamp techniques that apply controlled currents across the membrane while maintaining a constant membrane potential. Under voltage clamp conditions, the current required to maintain the membrane potential must be equal and opposite any membrane current (see Kandel & Schwartz, 1985, for a more detailed description of voltage clamp techniques).
- 4 See Brazier (1977) for a more detailed description of the sources and properties of bioelectrical activity.
- 5 When electrodes must be implanted or attached for more than a few days, the chemical stability of platinum makes it the best choice.

REFERENCES

- Barber, T. X. (1976). *Pitfalls in human research*. New York: Pergamon.
- Basmajian, J., & DeLuca, C. J. (1985). *Muscles alive: Their functions revealed by electromyography* (5th ed.). Baltimore: Williams & Wilkins.
- Beatty, J., Barth, D. S., Richer, F., & Johnson, K. A. (1986). Neuromagnetometry. In M. G. H. Coles, E. Donchin, & S. W. Porges (Eds.), *Psychophysiology: Systems, and applications* (pp. 26-40). New York: Guilford Press.
- Braff, D. L., Stone, C., Callahan, E., Ceyer, M., Gilck, I., & Bali, L. (1978). Prestimulus effects on human startle reflex in normals and schizophrenics. *Psychophysiology*, 15, 339-343.
- Brazier, M. A. B. (1977). *Electrical activity of the nervous system* (4th ed.). Baltimore: Williams & Wilkins.
- Brophy J. J. (1983). *Basic electronics for scientists* (4th ed.). New York: McGraw-Hill.
- Brown, C. C. (1967). *Methods in psychophysiology*. Baltimore: Williams & Wilkins.
- Brown, C. C., & Saucer, R. T. (1958). *Electronic instrumentation for the behavioral sciences*. Springfield, IL: Charles C. Thomas.
- Carr, J. J., & Brown J. M. (1981). *Introduction to biomedical equipment technology*. New York: Wiley.
- Coles, M. G. H., Donchin, E., & Porges, S. W. (Eds.) (1986). *Psychophysiology: Systems, processes, and applications*. New York: Guilford Press.
- Cooper, R., Osselson, J. W., & Shaw, J. C. (1980). *EEG technology* (3rd ed.). London: Butterworths.
- Curtis, B. A., Jacobson, S., & Marcus, E. M. (1972). *An introduction to the neurosciences*. Philadelphia: Saunders.
- DeMarre, D. A., Kantrowitz, P., Zucker, L., & Simmons, D. (1979). *Applied biomedical electronics for technicians*. New York: Marcel Dekker.
- DeMarre, D. A., & Michaels, D. (1983). *Bioelectronic measurements*. Englewood Cliffs, NJ: Prentice-Hall.
- Dunseath, W. J. R. (1982). A low-cost precision electrode impedance meter. *Psychophysiology*, 19, 117-119.
- Elul, M. R. (1972). The genesis of the EEG. *International Review of Neurobiology*, 15, 227-272.

during recording, and simultaneous monitoring of the input signal with an oscilloscope or graphic recorder is recommended. Three types of recording techniques are available: direct recording, frequency modulation, and pulse-code modulation.

In *direct recording*, the signal voltage is superimposed on a 70-100 kHz bias signal to increase the magnetizing force of the signal. The recording head converts the electrical input to a varying magnetic flux that changes the residual magnetism on the tape as it moves past the recording head. The frequency response of direct recording is poor for frequencies less than 100 Hz, and thus direct recording is not appropriate for most psychophysiological signals. With *frequency modulation (FM)* recording, the input signal is used to modulate the frequency of a carrier signal before storage. The bandwidth for FM recording is proportional to the tape speed and can range from DC to 108,000 Hz at speeds of 60 revolutions per second. In *pulse-code modulation (PCM)* the input signal is converted to a digital signal representing the momentary signal amplitude. The amplitude resolution of PCM recording with a 12-bit analog-to-digital converter (ADC) provides a better signal-to-noise ratio than FM recording.

4.7 SUMMARY AND CONCLUSIONS

This tutorial on bioelectrical measurement began with the basics of electricity and electrical conduction. It was noted that electrical charge is a fundamental property of all matter that is intrinsically associated with chemical reactivity. The relationship between electrical potential, current, and resistance described by Ohm's Law was elaborated as the basis for explanations of the properties of simple circuits and the use of tools for electrical measurement. The distinctive properties of DC and AC signals were delineated and related to capacitance and inductance. The appropriate uses of the three most common tools for electrical measurement - the multimeter, oscilloscope, and bridge circuit - were described and related to measurement of bioelectrical signals. The sources and the frequency and amplitude ranges for illustrative biopotential signals were surveyed.

Some general guidelines for sensor systems were provided, and the detection of physiological signals was subdivided into electrode systems for biopotentials and impedance and transducer systems for nonelectrical physiological changes. Within electrode systems, the electrical properties of the subject-electrode interface were discussed. Within transducer systems, it was noted that transducers are by necessity specialized for distinct types of physical change, and the role of the bridge circuit in transduction was described. The necessity for signal amplification and conditioning was noted, and the application of three types of special-purpose signal processors (filters, integrators, and amplitude-sensitive rate devices) in psychophysiology was presented. Finally, the prototypic systems for display and storage of voltage signals and their application in psychophysiology were described.

Progress in science is limited by the integrity of scientific observations. Whereas technology provides the tools for scientific observation, it is the responsibility of the observer to apply the tools appropriately. Failure to do so can result in a faulty data base and erroneous inferences. With all the sciences, development and distribution of technical tools will expand the ability of the psychophysiologicalist to observe, to quantify, and to control (Tassinary, Marshall-Goodell, & Cacioppo, 1985). Because it is difficult to anticipate future advances in technology, the psychophysiologicalist who

- Erlanger, J., & Glasser, H. S. (1937). *Electrical signs of nervous activity*. Philadelphia: University of Pennsylvania.
- Fowles, D. C. (1986). The eccrine system and electrodermal activity. In M. G. H. Coles, E. Donchin, & S. W. Porges (Eds.), *Psychophysiology: Systems, processes, and applications* (pp. 51-96). New York: Guilford Press.
- Fowles, D. C., Christie, M. J., Edelberg, R., Grings, W. W., Lykken, D. T., & Venables, P. H. (1981). Committee report: Publication recommendation for electrodermal measurement. *Psychophysiology*, 18, 232-239.
- Geddes, L. A., & Baker, L. E. (1975). *Principles of applied biomedical instrumentation* (2nd ed.). New York: Wiley.
- Gentner, D., & Gentner, D. R. (1983). Flowing waters or teaming crowds: Mental models of electricity. In D. Gentner & A. L. Stevens (Eds.), *Mental models*. New York: Erlbaum.
- Goldman, D. (1950). The clinical use of the "average" reference electrode in monopolar recording. *Electroencephalography and Clinical Neurophysiology*, 39, 526.
- Goovaerts, H. G., Ross, H. H., & Schneider, H. (1979). Storage systems. In R. S. Reneman and J. Strackie (Eds.), *Data in medicine: Collection, processing, and presentation* (pp. 181-205). The Hague: Martinus Nijhoff.
- Greenfield, N. S., & Sternbach, R. A. (Eds.). (1972). *Handbook of psychophysiology*. New York: Holt, Reinhart and Winston.
- Grings, W. W. (1954). *Laboratory instrumentation in psychology*. Palo Alto, CA: National.
- Jonides, J. (1982). Integrating visual information from successive fixations. *Science*, 215, 192-194.
- Jonides, J., Irwin, D. E., and Yantis, S. (1983). Failure to integrate information from successive fixations. *Science*, 222, 188.
- Kandel, E. R., & Schwartz, J. H. (1985). *Principles of neural science*. New York: Elsevier.
- Kybett, H. (1986). *Electronics: A self-teaching guide* (2nd ed.). New York: Wiley.
- Lambert, E. H. (1962). Electromyography. In O. Glasser (Ed.), *Medical physics* (Vol. 3). Chicago: Year Book Medical.
- Li, C. L., & Jasper, H. H. (1953). Microelectrode studies of the electrical activity of the cerebral cortex in the cat. *Journal of Physiology*, 121, 117.
- Loeb, G. E., & Gans, C. (1986). *Electromyography for experimentalists*. Chicago: University of Chicago.
- Martin, I., & Venables, P. H. (Eds.). (1980). *Techniques in psychophysiology*. Chichester: Wiley.
- Oster, P., & Stern, J. (1980). Measurement of eye movement: Electrooculography. In I. Martin and P. H. Venables (Eds.), *Techniques in psychophysiology*. Chichester: Wiley.
- Rostron, J. (1960). *Error and deception in science*. New York: Basic Books.
- Schmidt, M. (1963). A star-like object with large red-shift. *Nature*, 197, 1040.
- Schneider, R. E., & Fowles, D. C. (1978). A convenient, non-hydrating electrolyte medium for the measurement of electrodermal activity. *Psychophysiology*, 15, 483-486.
- Seippen, R. G. (1983). *Transducers, sensors, and detectors*. Reston, VA: Reston.
- Sokal, M. M., Davis, A. B., & Merzbach, U. C. (1976). Laboratory instruments in the history of psychology. *Journal of the History of the Behavioral Sciences*, 12, 59-64.
- Spooner, R. B. (1980). *Hospital electrical safety simplified*. Research Triangle Park, NC: Instrument Society of America.
- Stern, R. M., Ray, W. J., & Davis, C. M. (Eds.). (1980). *Psychophysiological recording*. New York: Oxford University Press.
- Tassinari, L. G., Marshall-Goodell, B., & Cacioppo, J. T. (1985). Microcomputers in social psychophysiological research: An overview. *Behavioral Research Methods, Instruments, and Computers*, 17, 532-536.
- Thompson, R. F. (1975). *Introduction to physiological psychology*. New York: Harper & Row.
- Venables, P. H., & Christie, M. J. (1973). Mechanisms, instrumentation, recording techniques and quantification of responses. In W. F. Prokasy & D. C. Raskin (Eds.), *Electrodermal activity in psychological research*. New York: Academic.
- Venables, P. H., & Martin, I. (Eds.). (1967). *A manual of psychophysiological methods*. New York: Wiley.
- Wood, R. W. (1904). The n-rays. *Nature*, 70, 530-531.
- Young, L., & Sheena, D. (1975). Survey of eye movement recording methods. *Behavior Research Methods and Instruments*, 7, 397-429.