the contractile fatigue measures which have been described apply to artificial stimulus train, which provides a similar excitation to all the motor units. Natural, voluntary stimulus trains are not similar for the various motor units involved in generating a muscle contraction. For example, the firing rate of latter recruited motor units is lower than that of earlier recruited motor units. Thus, the proposed classification provides a description of the mosaic of the motor unit fatigability but may not provide a description of the motor unit fatigability during a voluntary contraction.

CHAPTER 2

Apparatus, Detection, and Recording Techniques

In this chapter we will discuss details concerning electrode types and configurations, as well as associated instrumentation that has a bearing on the quality of the EMG signal that is detected and subsequently displayed, recorded, or processed. The process of sensing the signal by the electrode is referred to as *detection*. The word *recording* is reserved for describing the process that creates a record on any media (CRT, paper, magnetic tape, etc.).

Before beginning a substantive discussion on electrodes, it is necessary to ensure some minimal knowledge concerning the concepts of "impedances" and "filter functions."

CONCEPT OF IMPEDANCE AND FILTER FUNCTIONS

All forms of matter present an impedance to the transmission of an electric current. The impedance function is a vector quantity, hence it is expressed in terms of complex numbers, the real part of which denotes the resistance and the imaginary part of which denotes the susceptance. This latter part exists due to the presence of capacitance and/or inductance, two basic electrical properties of matter. In media such as muscle tissues, fatty tissue, and skin, the inductance is essentially unmeasurable. However, the capacitance is present in a significant amount and cannot be overlooked.

One of the simplest expressions of an impedance function, which is useful for conceptualizing the electrical characteristics of electrodes and tissue, is the impedance of a resistance in series with a capacitor presented in Figure 2.1. In this configuration the impedance function is expressed as a vector

$$Z(\omega) = R + \frac{1}{j\omega C}, \quad \text{where } j \text{ is } \sqrt{-1}, \text{ an imaginary quantity}$$

R = the resistance (ohms), C = the capacitance (farads), $\omega = 2\pi f$ and f = the frequency (Hz).

A vector may also be expressed in terms of its magnitude and phase (direction). The magnitude is the square root of the sum of the squared real part and the squared imaginary part.

$$|Z(\omega)| = \frac{(1 + \omega^2 C^2 R^2)^{1/2}}{\omega C}$$

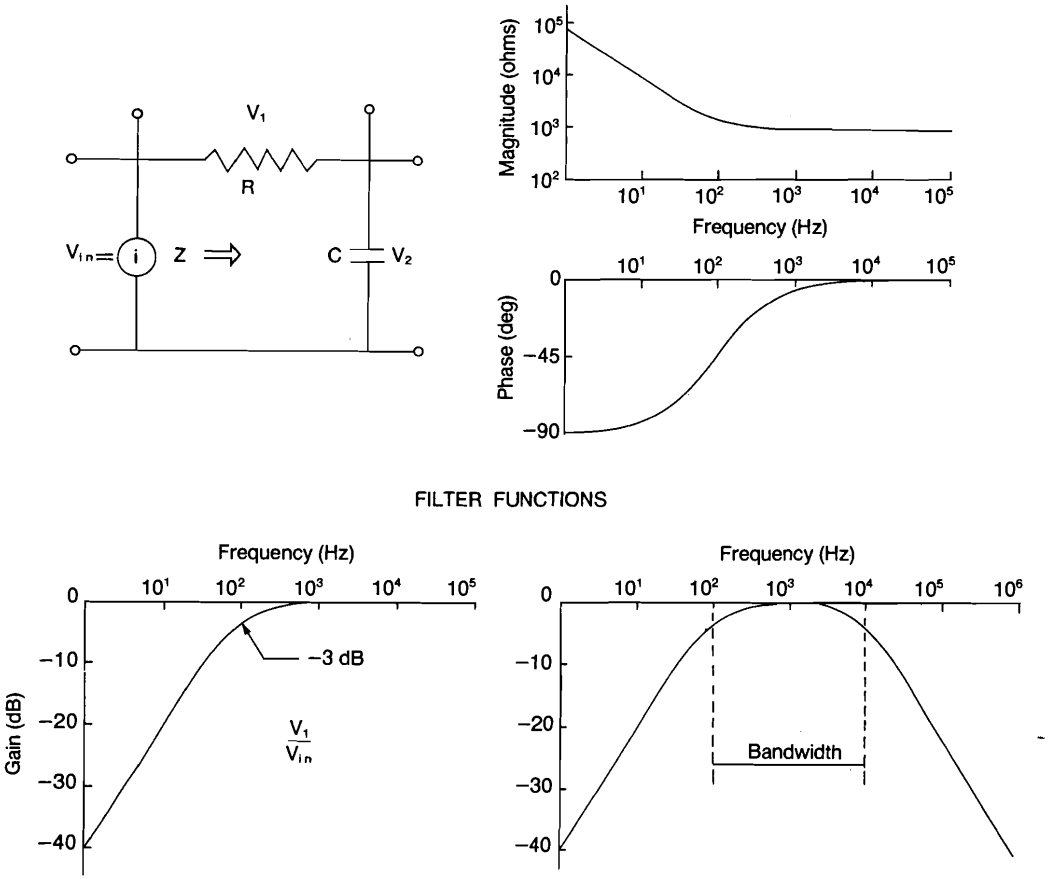

Figure 2.1. (*Top left*) A simple electrical circuit consisting of a resistor and capacitor. (*Top right*) The magnitude and phase of the impedance function of the electrical circuit, with $R = 1k\Omega$ and $C = 1.59\mu f$. Note that when the magnitude decreases by a factor of 0.707, the phase angle is 45°. (*Bottom left*) The filter function of the ratio of V_1/V_{in}. This is a high-pass filter. (*Bottom right*) A band-pass filter. The bandwidth is defined as the range of frequency between the high and low 3 dB points.

The phase is the inverse tangent of the ratio of the imaginary part to the real part.

$$\Phi Z(\omega) = -tan^{-1}\left(\frac{1}{\omega CR}\right)$$

The magnitude is measured in units of ohms and the phase in units of degrees. The above two functions are plotted on the top right-hand quadrant of Figure 2.1, with $R = 1k\Omega$ and $C = 1.59\mu f$. Note that as $f \rightarrow 0$, $|Z(\omega)| \rightarrow \infty$, and as $f \rightarrow \infty$, $|Z(\omega)| \rightarrow R$.

The impedance of a circuit describes the relationship between the voltage and current, $V = ZI$. Thus an alternative way of looking at the frequency-dependent characteristics of the impedance function is as a filter between the current and the voltage. In the specific example provided here, the impedance behaves as a high-pass filter because at higher frequencies the impedance is less. This filtering concept will be useful in describing the filtering properties of the electrode-electrolyte junction in latter portions of this chapter.

A more straightforward description of a filter function is expressed by the ratio of an output voltage to the input voltage. In our example

$$\left|\frac{V_1}{V_{in}}\right| = (1 + 1/\omega^2R^2C^2)^{-1/2}$$

$$\left|\frac{V_2}{V_{in}}\right| = (1 + \omega^2R^2C^2)^{-1/2}$$

Note that the ratio will be unitless and is therefore measured as a gain. The gain of $|V_1/V_{in}|$ will be greater at higher frequencies; thus it is referred to as a high-pass filter, whereas $|V_2/V_{in}|$ is a low-pass filter. The filter function for the high-pass filter is presented in the bottom left-hand quadrant. Note that in both examples of the two types of filters, when $f = \frac{1}{2}\pi RC$, the magnitude and gain decrease by $\sqrt{1/2} = 0.707$. It is common to measure the magnitude and gain in the decibel (dB) scale:

$$\text{gain} = 20 \log \left|\frac{V_{out}}{V_{in}}\right|$$

and a gain of 0.707 = 3 dB. The value of the frequency where the gain decreases by 3 dB is referred to by various terms such as the corner frequency, the break frequency, the cutoff frequency, and the 3 *dB* point. When the magnitude function is plotted on a log-log scale, as is the case in Figure 2.1, the slope of the function is 6 dB/octave or 20 dB/decade. An octave represents a doubling of the frequency and a decade indicates an order of magnitude. Note that 20 dB is equivalent to a factor of 10.

The combination of a low-pass and a high-pass filter yields a band-pass filter, as demonstrated in Figure 2.1 (*bottom*). The width of the "frequency-window" of a band-pass filter is measured by its bandwidth. This is the frequency range between the lower and upper 3 dB points. It should now be apparent that the bandwidth may be increased by either increasing the slope of the rolloff or by shifting the 3 dB points. As was the case in the impedance function, at the cutoff frequency, the phase angle distortion is −45°, a relatively large amount. Hence, if the signal processed by such a filter were to contain considerable energy at this frequency, considerable distortion of the signal would occur. This diffi-

culty may be overcome by choosing a cutoff frequency that is 10 times the greatest signal frequency of interest, in which case the filter would pass the signal with negligible amplitude reduction and a maximal phase shift of about 5°.

ELECTRODES

The electrodes used in electromyography could well be, and actually are, of a wide variety of types and construction. Their use depends on the first principle that they must be relatively harmless and must be brought close enough to the muscle under study to pick up the current generated by the ionic movement. The segment of the electrode which makes direct electrical contact with the tissue will be referred to as the *detection surface.* In electromyography these are used either singularly or in pairs. These configurations will be referred to as *monopolar* and *bipolar.*

The two main types of electrodes used for the study of muscle behavior are surface (or skin) electrodes and inserted (wire and needle) electrodes. Each has its advantages and its limitations, and they will now be described.

Surface Electrodes

Surface electrodes can be constructed as either passive or active. In the passive configuration, the electrode consists of a detection surface that senses the current on the skin through its skin-electrode interface. In the active configuration, the input impedance of the electrodes is greatly increased, rendering it less sensitive to the impedance (and therefore quality) of the electrode-skin interface.

One of the earliest, if not the earliest, reported usages of surface electrodes specifically for the purpose of detecting EMG signals from a human muscle was by Piper (1912). He used a metal plate. The design of passive surface electrodes has not changed much since Piper's days; conceptually, the metal electrodes used for this purpose today function similarly.

Often one finds that the simple silver discs used widely in electroencephalography are also used as passive surface electrodes in electromyography (Fig. 2.2). Their advantages revolve around one point: convenience. For example, they are readily obtained from supply houses; they can be applied to the skin after very little training and with reasonable success (within the limitations to be discussed); and they give little discomfort to the subject.

Since a poor contact must be avoided, continued pressure is important. The pressure provided by the adhesive strips or collars used to secure the electrodes is usually adequate. Electrical contact is greatly improved by the use of a saline gel or paste; this is retained between electrode and skin by making the silver disc slightly concave on the aspect to be applied to the skin. The dead surface layer of the skin, along with its protective oils, must be removed to lower the electrical impedance. This is best

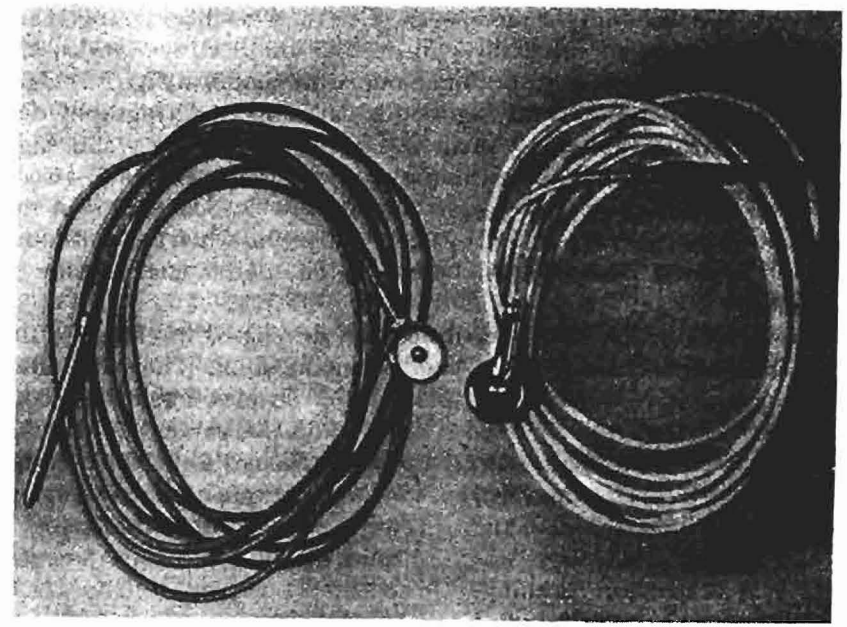

Figure 2.2. Silver disc surface electrodes (electroencephalographic type).

done by light abrasion of the skin at the site chosen for electrode application. In recent years, we have found that it is best produced by "rubbing in" those types of electrode gels that have powdered abrasives included in their formula. For additional details concerning the advantages of skin abrasion, the reader is referred to Tam and Webster (1977) and Burbank and Webster (1978).

In attempting to reduce the mass of the electrode, silver-metal films have been painted on the skin. These electrodes have been described by Rositano (1970). Although they may be convenient for special applications, such as detecting perioral muscle activity (Allen et al, 1972) or for long-lasting recording sessions such as space flights, they generally provide an inferior performance, as compared to that of conventional passive surface electrodes (Konopacki and Cole, 1982).

The lack of chemical equilibrium at the metal-electrolyte junction sets up a polarization potential that may vary with temperature fluctuations, sweat accumulation, changes in electrolyte concentration of the paste or gel, relative movement of the metal and skin, and the amount of current flowing into the electrode. Various construction designs have been implemented attempting to stabilize the polarization potential (Boter et al, 1966; Girton and Kamiya, 1974, among others). It is important to note that the polarization potential has both a dc and an ac component. The ac component is greatly reduced by providing a reversible chloride

exchange interface with the metal of the electrode. Such an arrangement is found in the widely used silver-silver chloride electrodes which are commercially available (e.g., Beckman miniature model in Fig. 2.3). This type of electrode has become highly popular in electromyography due to its light mass (250 mg), small size (11-mm diameter), and high reliability and durability. The diminished polarization potential associated with this electrode is a major benefit. The dc component of the polarization potential is nullified by electronic means when the electrodes are used in pairs. This point will be elaborated upon in later sections of this chapter.

The active surface electrodes have been developed to eliminate the need for skin preparation and conducting medium. They are often referred to as "dry" electrodes or "pasteless" electrodes. These electrodes can be either resistively coupled (Lewes, 1965; Bergey et al, 1971) or capacitively coupled to the skin (Lopez and Richardson, 1969; Potter and Menke, 1970; Betts and Brown, 1976). In the case of the capacitively coupled electrode, the detection surface is coated with a thin layer of dielectric (nonconducting) substance, and the skin electrode junction behaves as a capacitor. Although the capacitively coupled electrodes have

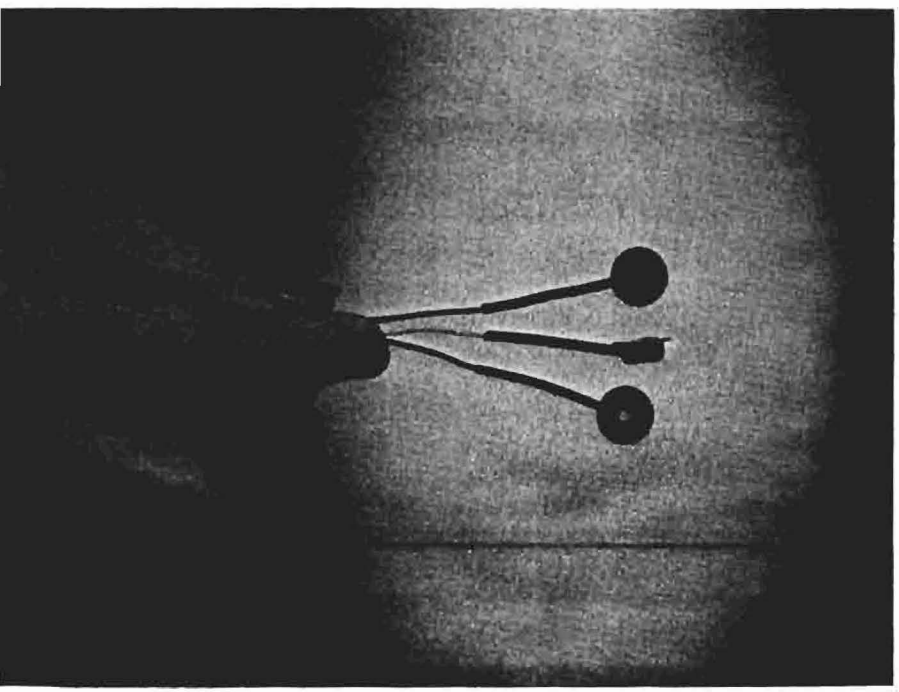

Figure 2.3. Miniature silver-silver chloride electrodes (Beckman type).

the advantage of not requiring a conductive medium, they have a higher inherent noise level (Potter and Menke, 1970). Also, these electrodes do not have long-term reliability because their dielectric properties are susceptible to change with the presence of perspirtion and the erosion of the dielectric substance. For these reasons they have not yet found a place in electromyography.

An adequately large input impedance is achieved when the resistance is in the order of 10^{12} ohm and the capacitance is small (typically, 3 or 4 pF). The advent of JFET microelectronics has made possible the construction of amplifiers housed in integrated circuitry which have the required input impedance and associated necessary characteristics. However, the physical construction of the active electrode remains important because the input capacitance from the metal surfaces to the input of the active circuitry is to be minimized. Two examples of such electrodes are presented in Figure 2.4. These electrodes were conceptualized and designed by De Luca and were constructed in the NeuroMuscular Research Laboratory of the Liberty Mutual Research Center in Hopkinton, MA. They each have two detection surfaces and associated electronics circuitry within their housing; the circular one contains a stainless steel ring around its perimeter which serves as a ground. These electrodes

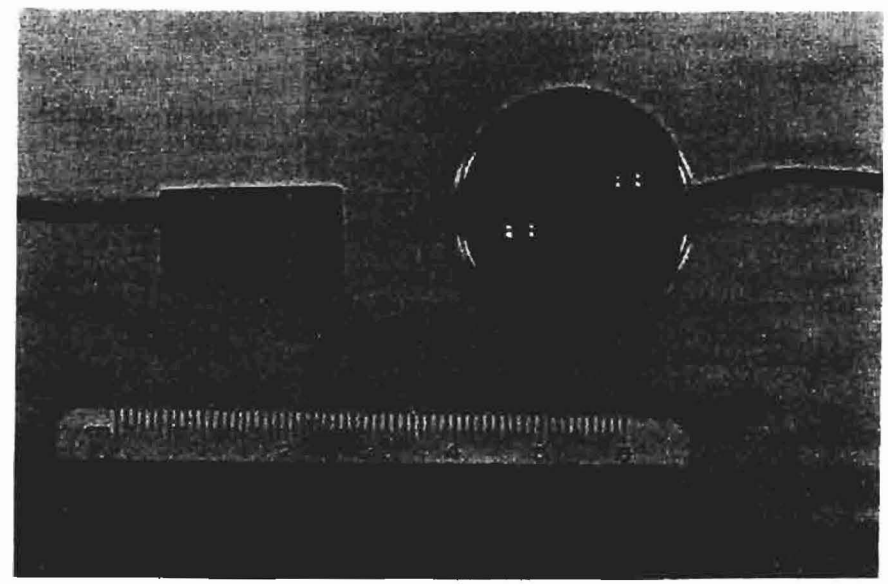

Figure 2.4. Active surface electrodes in bipolar configurations. The circular unit contains a ground ring around the perimeter of the electrode (patent pending). These electrodes do not require any skin preparation or conductive paste or gels. (Conceptualized and designed by C.J. De Luca.)

are an outgrowth of several years of research in the NeuroMuscular Laboratory, and their precursor was reported by De Luca et al (1979). The details of the electronics circuit design considerations will be discussed in a later section in this chapter.

The active surface electrodes are preferable not only because they provide an EMG signal of greater fidelity, but also because they are convenient to use. The simplicity and speed with which they may be applied to the skin is rapidly making them the electrode of choice for pragmatic applications as in busy clinical environments and myoelectrically controlled prosthetics. Within a few years, they will inevitably be the preferred type of surface electrode in the research environment. In our laboratories, active surface electrodes have been in common usage for the past 4 years.

The chief disadvantages of surface electrodes are that they may be used effectively only with superficial muscles and that they cannot be used to detect signals selectively from small muscles. In the latter case, the detection of "cross-talk" signals from other adjacent muscles becomes a concern. These limitations are often outweighed by their advantages in the following circumstances:

1. When representation of the EMG signal corresponding to a substantial part of the muscle is required.
2. In motor behavior studies when the time of activation and the magnitude of the signal contain the required information.
3. In psychophysiological studies of general gross relaxation of tenseness, such as in biofeedback research and therapy.
4. In the detection of EMG signals for the purpose of controlling external devices such as myoelectrically controlled prostheses and other like aids for the handicapped population.
5. In clinical environments where a relatively simple assessment of the muscle involvement is required, e.g., in physical therapy evaluations and sports medicine evaluations.
6. Where the simultaneous activity or interplay of activity is being studied in a fairly large group of muscles under conditions where palpation is impractical, e.g., in the muscles of the lower limb during walking.
7. In studies on children or other individuals who object to needle insertions.

The surface electrodes are commonly used to detect gross EMG signals consisting of the electrical activity from numerous individual motor units within the pickup area of the detection surfaces. However, we have often used such electrodes to detect MUAPTs during low level muscle contractions. Gydikov and Kosarov (1972) have reported similar accomplishments. The reader is cautioned that this detection technique requires practice and may be frustrating.

Needle Electrodes

By far the most common indwelling electrode is the needle electrode. As such, a wide variety of needle electrodes are now commercially

available. The most common needle electrode, in turn, is the "concentric" electrode, first described and used by Adrian and Bronk (1929) and now used widely by clinical and research electromyographers. The monopolar configuration contains one insulated wire in the cannula (Fig. 2.5). The tip of the wire is bared and acts as a detection surface. The bipolar configuration contains a second wire in the cannula and provides a second detection surface. The needle electrode has two main advantages. One is that its relatively small pickup area enables the electrode to detect individual motor unit action potentials conveniently, especially during relatively low-force contractions. The other is that they may be conveniently repositioned within the muscle (after insertion) so that new territories may be explored or the signal quality may be improved. These amenities have naturally led to the development of various specialized versions to study particular properties or aspects of the motor unit.

Buchthal and his colleagues used a multifilament electrode with 12 individually insulated wires located longitudinally along the length of the cannula. With this electrode, they were able to ascertain the size of the territory of the motor unit (Buchthal et al, 1957a) and measure the voltage decrease as a function of distance in muscle tissues (Buchthal et al, 1957b).

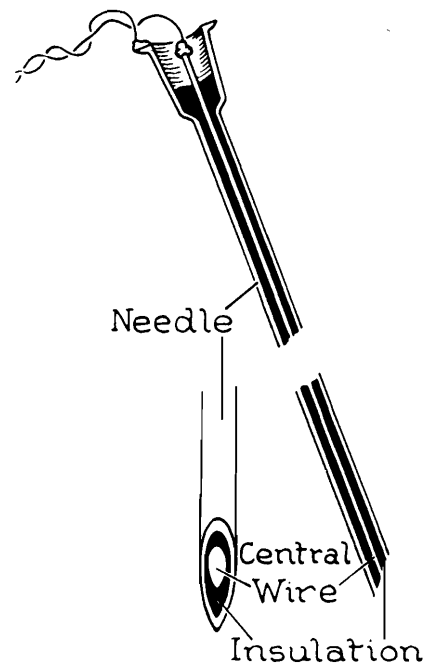

Figure 2.5. Concentric needle electrode in the monopolar configuration. A bipolar configuration may be realized by locating two insulated wires within the cannula of the needle.

In 1972, De Luca and Forrest described the design of a four-filament needle electrode for simultaneously detecting more than one electrical representation of the activity of the motor units within the pickup area of the electrode. This electrode contained four detection surfaces consisting of the cross-sectional area of the 25-μm diameter wires located at the corner of a square in the tip of the cannula. With this electrode, De Luca and Forrest demonstrated that multichannel detection of EMG signals is essential to eliminate ambiguity in the identification of individual MUAPs. Recently, in collaboration with our colleagues Broman and Mambrito, we produced an improved version of this electrode. This latter version, along with the preferred detection configuration, may be seen in Figure 2.6. This version houses somewhat larger (75-μm diameter) detection surfaces on the side of the cannula. This particular architecture of the electrode has proven to provide the necessary pickup selectivity while still detecting several MUAPTs.

Another productive exploit of the architecture of the needle electrode was originally reported by Ekstedt and Stålberg (1973) and has been subsequently popularized by Stålberg and his colleagues in Uppsala. This type of electrode consists of a modified monopolar needle electrode whose central wire is relatively small in diameter (25 μm) and is exposed on the side of the cannula. By detecting from this wire with respect to a

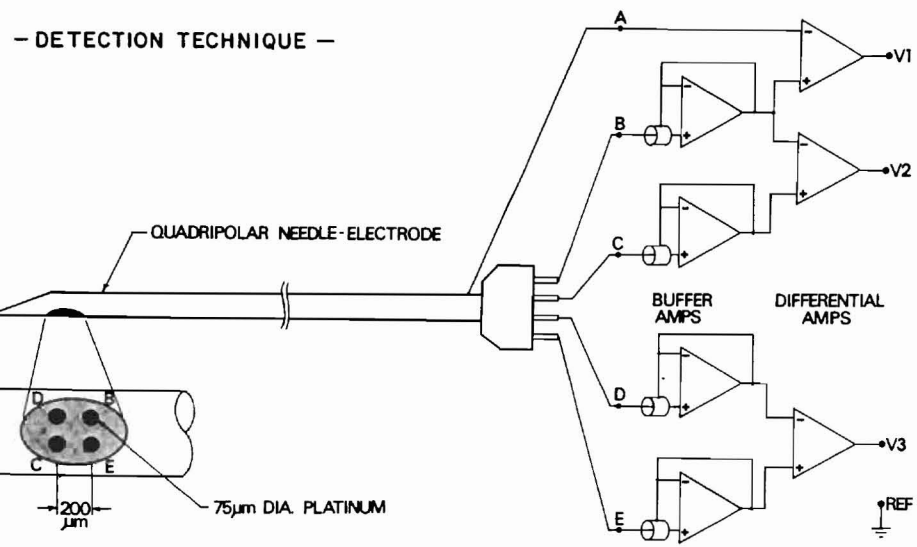

Figure 2.6. A schematic representation of a lightweight quadripolar needle electrode configured to detect three independent channels of EMG signals. Any two of the detection surfaces (cross-sectional areas of the wires) may be used, as a bipolar pair.

distant electrode it is possible to obtain the signal activity of only one or two muscle fibers of a motor unit in most muscles. This type of electrode is presented in Figure 2.7.

In 1980, Stålberg modified the single fiber electrode and constructed a "macro" electrode which detects the MUAP from many fibers of the motor unit. This feat is accomplished by insulating the cannula of the needle to within 15 mm of its tip and using the exposed part of the cannula as the main electrode contact (refer to Fig. 2.7). This uncommonly large detection surface will obviously detect MUAPs from numerous motor units whose fibers are scattered throughout the 15-mm length of the cannula. A chosen MUAP is recovered by locating it with the single fiber electrode (on the same needle) and using this MUAP to trigger average the signal detected by the exposed part of the cannula. The process of trigger averaging consists of using a clearly detectable signal to denote the time occurrence of a related event in a noisier signal. This is accomplished by repetitively averaging a segment of the noisy signal, beginning at a time denoted by the clearly detectable signal or the "trigger" signal. If the noise is random in nature it will cancel out, while the signal of the related event will be enhanced.

Some clinicians prefer the monopolar type of electrode, which is an outgrowth of the electrode introduced by Jasper and Ballem (1949).

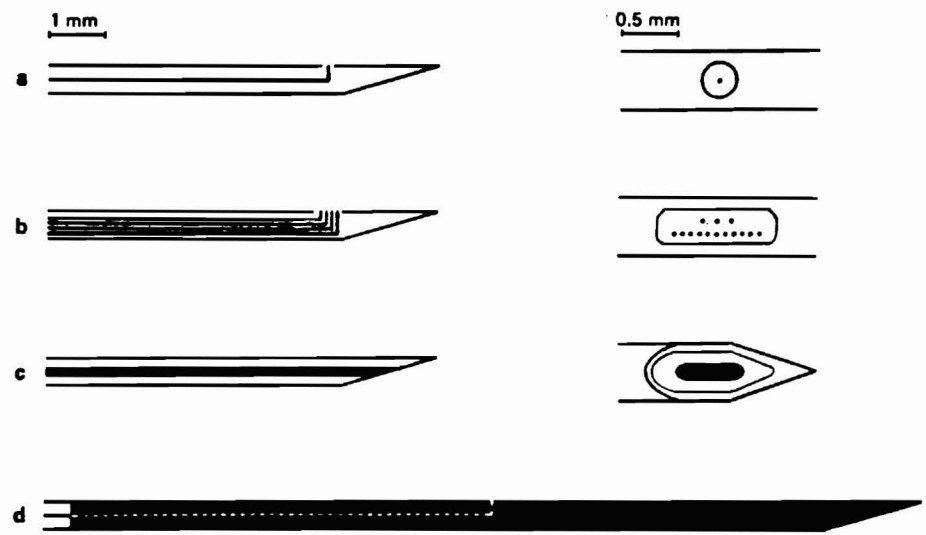

Figure 2.7. Examples of various needle electrodes. (a) Single fiber electrode with one detection surface. (*b*) Multipolar electrode. (c) Concentric needle electrode. (*d*) Macroneedle electrode. (*Black area* is noninsulated.) (From E. Stålberg, © 1980, *Journal of Neurology, Neurosurgery, and Psychiatry*.)

Unlike the concentric monopolar electrode, this electrode consists of a metallic needle that is insulated throughout, except for the tip, which is the detection surface. This concept has now reached a refinement that has proven useful for recording directly from nerve fibers. This microneurography detection was originated by Vallbo et al (1979). The electrode consists of an insulated tungsten filament approximately 150 to 200 μm in diameter and exposed for 5 to 10 μm at the tip. This tungsten filament is inserted as a monopolar electrode in the nerve, and the neuroelectric signal is detected with respect to a reference electrode located on the surface of the skin.

Wire Electrodes

Since 1961 this type of electrode has been popularized by Basmajian and his associates (see Basmajian and Stecko, 1962). Similar electrodes that differ only in the details of their construction have been independently developed by a number of other researchers, notably Close et al. (1960) and Long and Brown (1962). Wire electrodes have proven a boon to kinesiological studies because they are extremely fine, and therefore painless, and are easily implanted and withdrawn.

Wire electrodes may be made from any small diameter, highly nonoxidizing, stiff wire with insulation. Metals such as platinum alloys, silver, and nickel-chromium alloys are preferable. Insulations such as nylon, polyurethane, and Teflon are conveniently available. Such wires are available under the trade names of Karma (Wilber B. Drive Co., Harrison, NJ), Stablohm (Johnson Matthey Metals Ltd., London, England, and California Fine Wire Co., Grover City, CA), and various other manufacturers. Wire diameters in various sizes are available, the smallest being 25 μm. At least one distributor, A-M Systems, Inc. (Toledo, Ohio), has multistrand fine wire available. Of the available metals and alloys which are used to draw the fine wires, we prefer the 90% platinum-10% iridium alloy. It offers the appropriate combination of chemical inertness, mechanical strength, and stiffness. The Teflon or nylon insulators are preferred because they add some mechanical rigidity to the wires, making them easier to handle. Also, these insulators are less apt to crack, thus exposing the wires in unplanned locations and providing additional detection surfaces that may compromise the detection strategy.

The steps for making an electrode are displayed in Figure 2.8 and may be described as follows: (1) A double strand of insulated fine wire is passed through the cannula of a hypodermic needle. A small loop is left distally, and 5 to 7 cm of wire are left proximally. (2) A small amount of the insulation at the distal tip and 3 to 4 cm of each strand proximally is removed, either by burning it off with an alcohol flame or (depending on the type of insulation) etching it off with chemical solvents. (3) The loop is cut, leaving 1 to 2 mm of bared wire distally on each strand. These bare ends are staggered so that they will not come in contact.

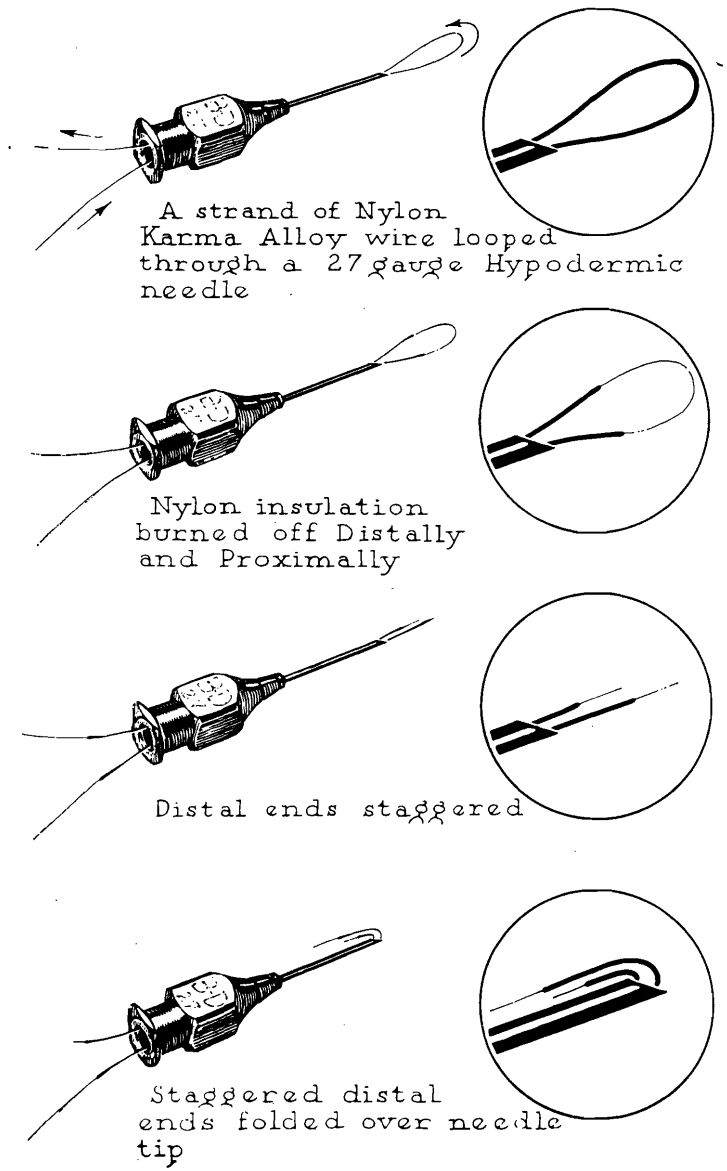

Figure 2.8. Steps in making a bipolar wire electrode with its carrier needle used for insertion. (From Basmajian and Stecko, © 1962, © *Journal of Applied Physiology.*)

They are then bent sharply back to lie against the needle shaft for a short distance. If preferred, the wires may be twisted together.

Such electrode assemblies may be driven easily into a muscle without anesthesia, and the attendant pain is the usual pain resulting from the

needle puncture. If fresh, sharp, 27-gauge needles are used, the pain is minimal and transitory. The needle withdraws easily, and its removal only rarely dislodges the electrodes, for they are retained by the hooks at their ends. The electrodes are taped to the skin at the site of emergence to ensure that an accidental tug does not remove them. At the end of an experiment a gentle pull brings the electrodes out painlessly, for each wire is so pliable that the barb straightens out on traction and offers little if any palpable resistance. We have had no accidental breakage in many thousands of uses; nor would we be disturbed if we had, because the fine wire is innocuous. Jonsson and Bagge (1968) have described the various deformations and dislocations of fine-wire electrodes and report that very vigorous exercise may break 25-μm wires. Fortunately, the wire is available in 50- and 75-μm sizes also, and these have been shown to be much tougher.

If one wishes to insert individual monopolar wire electrodes, the modification of our technique by Scott (1965) is satisfactory. The hypodermic needle is used to insert the electrode and also acts as a cutting instrument once the electrode is deep in place. A single strand of fine wire is passed through the hypodermic needle, and a long loop is turned back from the needle tip. The wire is cut by pulling on its free ends, one of which emerges from the skin alongside the needle and one out of the needle. Both the needle and its contained wire are then withdrawn, leaving one unbarbed wire deep in place.

After a period of use, we found only one tedious step or complication with wire electrodes, i.e., the connection of the almost invisible filament to the input of the amplifiers. Others have met and overcome this problem in different ways. For example, Long and his colleagues at Highland View Hospital in Cleveland relied on preconnection of the fine-wires to standard wires to produce a unit. However, when their needles are withdrawn, they cannot be discarded because they cannot be drawn over the connections. Because we prefer disposable needles and electrodes that are used only once and discarded, we make connections after the needle is entirely removed. Making the connections can be a tedious procedure when it must be done a dozen times for one experimental setup. Therefore, being lazy, we developed a simple method.

After finding that soldering, microwelding, and miniature alligator clamps all have drawbacks, we devised a spring-wire coil connector (Basmajian et al, 1966) which has proved itself in 20 years of extended usage. It is a brass spring (about 4 mm in diameter by 12 mm in length) soldered permanently to the free ends of each amplifier lead-in wire. The spring is tightly wound from a resilient 22 "spring-brass" wire which gives considerable pinch between adjacent coils. This type of hard-brass wire is available through ordinary commercial channels. To make connection to the wire electrodes after they have been inserted and the needle has been discarded, the spring is bent slightly between the thumb and index finger. This spreads the coils and allows the bared end of the electrode wire to be slipped between one or two pairs of coils. Released, the spring clamps the fine wire and gives good electrical connection instantly. Wrapping a bit of adhesive tape around the connection (for protection and insulation) completes a procedure that takes only moments and saves many tedious minutes required by other methods.

The wide acceptance of wire electrodes has led to a variety of innovations exploiting the versatility of this technique. Leifer (1969) used a four-filament electrode ensemble consisting of 25-μm stainless steel wires cemented together with epoxy. The four wires were exposed at their tips and were arranged in a straight line. Hennerz (1974) used a three-filament (100-μm silver wire) arrangement. In the same year, Shiavi (1974) decribed the construction of a wire electrode consisting of three staggered pairs of wires which provided three independent channels of EMG signals. This wire cluster was approximately 250 μm in diameter and was inserted into the muscle by the standard method. All these multifilament fine wire arrangements have, as their main goal, the detection of EMG signals which present a clearer view of the individual MUAPTs. It has been our experience that with the proper combination of wire diameter, staggering arrangement of the wire, and interwire spacing, progress towards the desired intention may be made. However, some trial and error is involved, and the beginner should not easily abandon the task.

Two other innovative users of wire electrodes have been proposed. Caldwell and Reswick (1975) described an electrode that could be inserted in a muscle and would remain within the muscle for months. The electrodes consisted of fine wires wound into a coil filled with silicone rubber. These coils, when inserted via a hypodermic needle, anchored themselves to the muscle fibers and thus resisted dislodgement.

A truly innovative concept has been proposed by Andreassen and A. Rosenfalck (1978). They described a "side-hole" electrode which consists of two 75-μm insulated fine wires which are twisted. Unlike all previous cases, the detection surfaces were two holes (10 to 25 μm) burned into the side of the wires. The electrode was inserted by a curved cannula which entered the skin, passed through the muscle, and exited through the skin. In this fashion, the location of the detection surfaces could be positioned throughout the muscle by pulling on either end of the wires. This approach overcomes the main disadvantages of wire electrodes; the inability to be repositioned after insertion. This interesting technique should be used with caution and, advisably, in an aseptic environment. Continual and excessive movement of the protruding wires may provide a bacterial conduit.

In kinesiological studies in which the main purpose of using wire electrodes is to record a signal that is proportional to the contraction level of muscle, repositioning of the electrode is not important. But, for other applications, such as recording distinguishable MUAPTs, this limitation is damaging. Some of us have used the term "poke-and-hope" to describe the standard wire electrode technique for this particular application. Another limitation of the wire electrode is its tendency to migrate after it has been inserted, especially during the first few contractions of the muscle. In fact, we suggest that the muscle with the electrode be contracted and relaxed at least one-half dozen times before any measurements are taken. The pumping action of the muscle will tend to lodge the barbs at the end of the electrode into the tissue. It will also draw some wire subcutaneously, providing excess wire to be released if the external tension on the wire increases. The reproducibility of the EMG signal detected with wire electrodes discussed by Jonsson and Reichmann (1968), Komi and Burskirk (1970) and, more recently, by Gans and Gorniak (1980) applies to parameters of the total EMG signal (such as amplitude), not to the more sensitive parameters of the individual MUAPs.

ELECTRODE TREATMENT

Cleaning

It is good practice to clean needle electrodes with a swab soaked with 70% ethanol and 30% distilled water solution after every use of the electrode. This procedure will remove the debris (such as skin particles, coagulated blood, muscle tissue, etc.) which tends to accumulate near the detection surfaces. Alternatively, the needle electrodes may be cleaned by placing them in an ultrasonic vibrator, of the type that is used to clean small pieces of jewelry. These vibrators may be purchased as small receptacles which contain a cleansing fluid and conveniently hold several needles simultaneously. We have been very pleased with our experience with such devices. Regardless of the technique used, it is imperative to remove the attached debris from the tip of the needles. Their accumulation will change the impedance properties of the electrode, which will result in a deterioration of the quality of the detected signal.

Surface electrodes should also be cleaned after each application. If the electrodes are of the type which require a conductive paste or gel, then any residual paste or gel should be removed by wiping them with gauze dampened in distilled water before the conductive material hardens. Once the conductive material hardens (usually within 30 minutes of exposure to the air), it may still be removed by more vigorous wiping action with gauze dampened with distilled water. The use of solvents or cleansing agents is greatly discouraged. If the surface electrode is of the

type which does not require conductive paste or gel, it is recommended that the metallic contacts of the electrode be regularly cleaned in order to remove any oxide layer which may accumulate on the detection surface. This should be done by swabbing them with 70% ethanol-30% distilled water mixture. This mixture will evaporate without leaving a film.

Impedance Reduction

Needle electrodes, especially those that have detection surfaces with small surface areas, tend to have a high impedance due to the high resistance of the exposed area and low capacitance of the metal-electrolyte interface. The impedance of the electrode may be effectively reduced by an electrolytic treatment originally described by Buchthal et al (1957a). A modified procedure was described by De Luca and Forrest (1972). This procedure consists of placing the needle electrode in a receptacle containing a 1 N saline solution (0.9% NaCl) and a platinum metal strip. The terminals of the electrode are connected to the negative terminals of a variable power supply, and the platinum strip is connected to the positive terminal. A dc current of 1 mA is passed through the needle electrode until small bubbles are seen forming and effervescing from the detection surfaces of the electrodes. Continue to supply the current until the bubbles rise to the surface in a continuous stream. If the bubbles do not form, increase the current gradually until the bubbles are in evidence. The reader is cautioned that an excessive amount of current will damage the detection surfaces of the electrode.

This electroplating procedure deposits a rough layer of salts on the metal of the contact area and thus increases the surface area. This results in a decreased impedance. The procedure also neutralizes some of the corrosive processes which degrade the metallic surfaces as a function of time, and, in the case of bipolar electrodes, it will tend to balance the impedances of the two detection surfaces. The usefulness of this latter development will become apparent in subsequent sections of this chapter. This approach of balancing input impedances of the detection surfaces has also been applied after the electrode has been inserted. Basmajian (1973) described such usage for the purpose of eliminating occasional high frequency artifacts from indwelling wire electrodes.

Sterilization

Prior to piercing any electrode through the skin and into a muscle, it is manditory to sterilize it. This may be accomplished with dry heat, boiling water, or steam. Of these three approaches, autoclaving at 15 lb/inch or 10 newtons/m^2 pressure for 30 minutes is preferred. If this method is not available in your environment, or if it proves to be inconvenient, then we suggest dry heat. In this approach, simply securely

wrap the electrode(s) in a paper folder and place the package in an oven for 60 minutes at a temperature of 130°C. Both these approaches are appropriate for needle and wire electrodes. For needle electrodes, it is sometimes convenient to sterilize them in boiling water that is under more than 1 atmosphere of pressure.

Antibacterial chemical baths are not recommended for sterilizing electrodes.

During sterilization, caution should be exerted to ensure that the temperature does not damage the insulation of the wires and the adhesive used to bond the wires to the needles. The commonly used materials for this purpose may be continuously exposed to the following temperatures without mechanical damage: Teflon, 150 to 200°C; nylon, 80 to 120°C; and expoxy, ~ 175°C. Continual exposure to higher temperature may structurally damage needle and wire electrodes.

HOW TO CHOOSE THE PROPER ELECTRODE

The specifics of the type of electrode that is chosen to detect the EMG signal depend on the particular application and the convenience of use. The application refers to the information that is expected to be obtained from the signal, for example, obtaining individual MUAPs or the gross EMG signal reflecting the activity of many muscle fibers. The convenience aspect refers to the time and effort that the investigator wishes to devote to the disposition of the subject or the patient. Children, for example, are commonly resistant to having needles inserted in their muscles.

We recommend the following electrode usage. However, the reader should keep in mind that crossover applications are always possible for specific circumstances.

Surface electrodes
- Time-force relationship of EMG signals
- Kinesiological studies of surface muscles
- Neurophysiological studies of surface muscles
- Psychophysiological studies
- Interfacing an individual with external electromechanical devices

Needle electrode
- MUAP characteristics
- Control properties of motor units (firing rate, recruitment, etc.)
- Exploratory clinical electromyography

Wire electrodes
- Kinesiological studies of deep muscles
- Neurophysiological studies of deep muscles
- Limited studies on motor unit properties
- Comfortable recording procedure from deep muscles

Each of these suggested categories often has specific applications that require special properties of the electrode. We will now discuss some of these properties. They are: electrode configuration; voltage decrement function of muscle tissue; electrode pickup area and cross-talk electrical properties of electrode-electrolyte interface; and filtering properties of bipolar configuration. The novice in electronics should review the section on impedance and filter functions at the beginning of this chapter prior to reading the following material.

Electrode Configuration

The electrical activity inside a muscle or on the surface of the skin outside a muscle may be easily acquired by placing an electrode with only one detection surface in either environment and detecting the electrical potential at this point with respect to a "reference" electrode located in an environment which is either electrically quiet or contains electrical signals which are unrelated to those being detected. (By unrelated, it is meant that the two signals have minimal physiological and anatomical association.) A surface electrode is commonly used as the reference electrode. Such an arrangement is called monopolar and is at times used in clinical environments because of historical precedents and its relative technical simplicity. In the early days of electromyography (circa 1940) electronics amplifiers were considerably inferior to and much more limited than today's versions. Schematic arrangement of the monopolar detection configuration may be seen in Figure 2.9.

The monopolar configuration has the drawback that it will detect all the electrical signals in the vicinity of the detection surface; that includes unwanted electrical signals from sources other than the muscle being investigated.

The bipolar detection configuration overcomes this limitation. This configuration is also displayed in Figure 2.9. In this case, two detection surfaces are used to detect two potentials in the muscle tissue of interest, each with respect to the reference electrode. The two signals are then fed to a differential amplifier which amplifies the difference of the two signals, thus eliminating any "common mode" components in the two signals. Signals emanating from the muscle tissue of interest near the detection surface will be dissimilar at each detection surface due to the localized electrochemical events occurring in contracting muscle fibers. Whereas "ac noise" signals originating from a more distant source (such as 50- or 60-Hz electromagnetic signals radiating from power cords, outlets, and electrical devices) ~~as well as~~ and "dc noise" signals (such as polarization potentials in the metal-electrolyte junction) will be detected with an essentially similar amplitude at both detection surfaces, ~~and~~ therefore, their will be subtracted prior to being amplified. This idealized behavior of the differential amplifier cannot be achieved with present day electronics. The measure of the ability of the differential amplifier to eliminate the common mode signal is called the common mode rejection ratio.

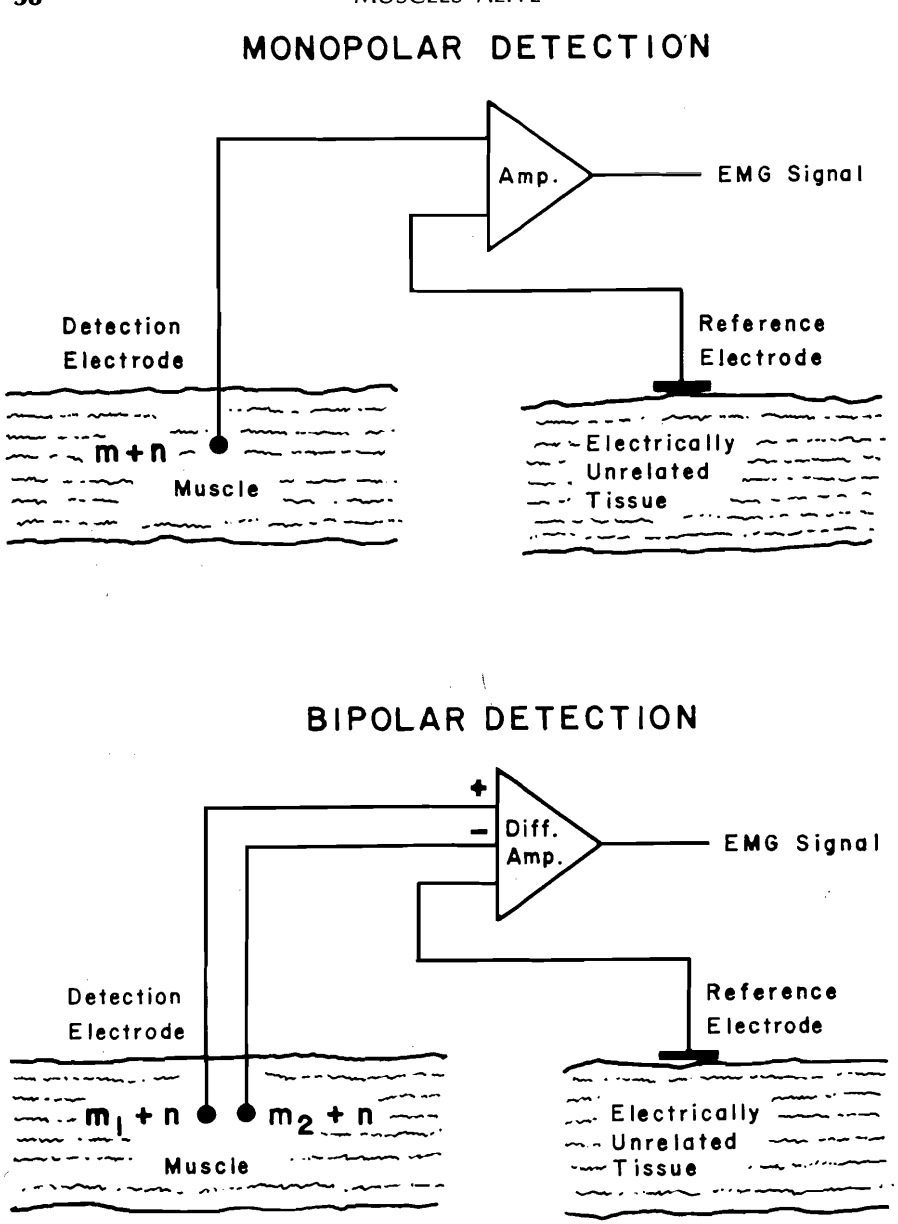

Figure 2.9. (*Top*) Monopolar detection arrangement. (*Bottom*) Bipolar detection arrangement. Note that in the bipolar detection arrangement, the EMG signals are considered to be different, whereas the noise is similar.

Decrement Function of Muscle Tissue

Muscle tissue presents an internal impedance to the propagation of electric currents. The impedance is frequency dependent; it is less for lower frequencies than for higher frequencies. It will also be a function of the distance between the sources of the EMG signal and the detection surfaces of the electrode. Thus, the muscle and adjacent tissues may be considered as a distance-dependent filter. Lindström (1970), through elaborate mathematical modeling, was able to calculate the tissue filter functions. His results, which are a simplified representation of the real environment, are nonetheless very helpful in providing guidance and insight in the behavior of tissue properties. They are presented in Figure 2.10. These curves represent the tissue filter properties as a function of distance perpendicular to the muscle fiber.

In reality, the impedance of muscle tissue is not isotropic, i.e., similar in all directions. In fact, it is highly direction dependent, i.e., anisotropic. The anisotropy is due to the nonhomogeneity of the anatomical construction of a muscle; muscle fibers are normally arranged lengthwise, and the surrounding extracellular fluid forms lengthwise channels parallel to the muscle fibers. These "channels" of lower impedance branching throughout the muscle make it very difficult to define precisely the current distribution within a muscle. In fact, the situation is considerably aggravated when the signal propagates through the fatty tissue and the skin to reach the surface of the skin, where it may be detected by surface electrodes. The considerably different electrical properties of the muscle tissue, fatty tissue, and skin cause inflections in the current field.

This anisotropic property of the muscle tissue impedance has been known since the earliest attempts were made to measure it. Hermann (1871), using the crude instruments available in the last century, found that the magnitude of the impedance in a direction perpendicular to the muscle fibers was considerably greater than that in a direction parallel to the muscle fibers. Recently, Epstein and Foster (1983) have reported detailed measurements which indicate that the magnitude of the impedance in the perpendicular direction is 7 to 10 times greater than that in the longitudinal direction. These results are consistent with other reported measurements of a similar nature.

Referring to Figure 2.10, it can be seen that at higher frequencies, the signal amplitude will decline sharply near the surface of the muscle fiber ($D = 0$) and then gradually diminish. This measure is known as the "decrement function." This function is typically obtained by plotting the peak-to-peak amplitude of a muscle fiber action potential observed as the detecting electrode is moved away from the active muscle fiber along a perpendicular direction. (Note that the peaks of the action potential contain high frequency components. It is for this reason that the high frequency region of Figure 2.10 is used to provide a comparison in the

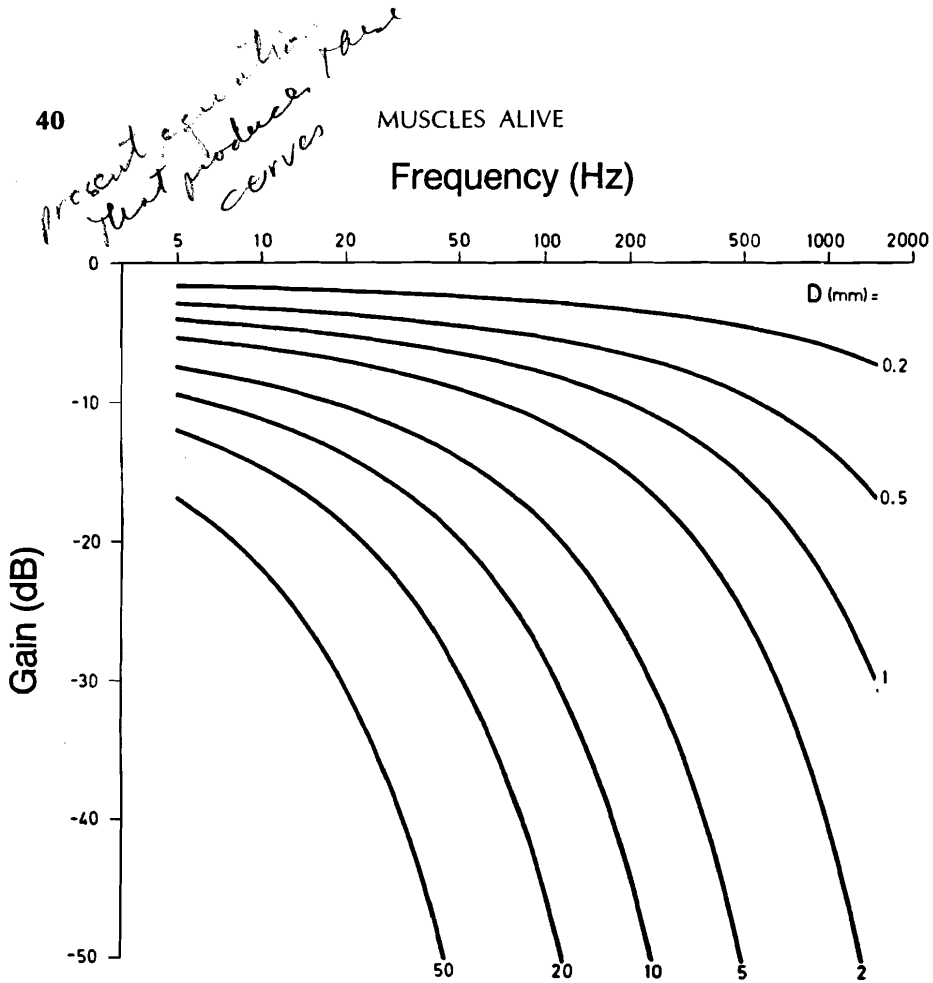

Figure 2.10. Representation of the tissue filter function. The parameter (*D*) indicates the distance from an active fiber to the detection electrode. These curves were obtained by designating the conduction velocity along the muscle fiber to be 4 m/s and the diameter of the muscle fiber to be 100 μm. (From L. R. Lindström, © 1970, *Technical Report, Chalmers University of Technology, Sweden.*)

frequency domain for the peak-to-peak decrease in the time domain.) Figure 2.11 which is taken from the theoretical work of Andreassen and A. Rosenfalck (1978), clearly demonstrates the dramatic decrease in the peak-to-peak amplitude for different electrode configurations and orientations. Gath and Stålberg (1978) reported empirical results supporting the theory.

The information in Figures 2.10 and 2.11 indicates that small displacements of the electrode with respect to the active fibers, when the electrode is near the surface of the active fiber, cause drastic changes in the waveform of the detected signal. If the electrode is moved 100 μm from the surface of a fiber, the peak-to-peak amplitude decreases by approximately 75%. It is this sharp radial decline which accounts for the

sometimes drastic modifications of MUAP waveforms during muscle contractions. Even attempted constant-force isometric contractions may provide sufficient relative movement between the electrode and active muscle fibers to seriously disturb the amplitude and shape of the signal. At times this disturbance may be sufficient to render impossible the identification of MUAPs belonging to the same MUAPT.

Figure 2.11 also demonstrates that *rotation* of the detection surfaces of a needle electrode will greatly modify the amplitude of the signal, especially if they are near the active muscle fiber. Empirical observations of this behavior have been reported by Andreassen and A. Rosenfalck (1978) for indwelling electrodes and by Vigreux et al (1979) for surface electrodes. This aspect of the detecting techniques provides both advantages and disadvantages. The wise investigator will exploit this property

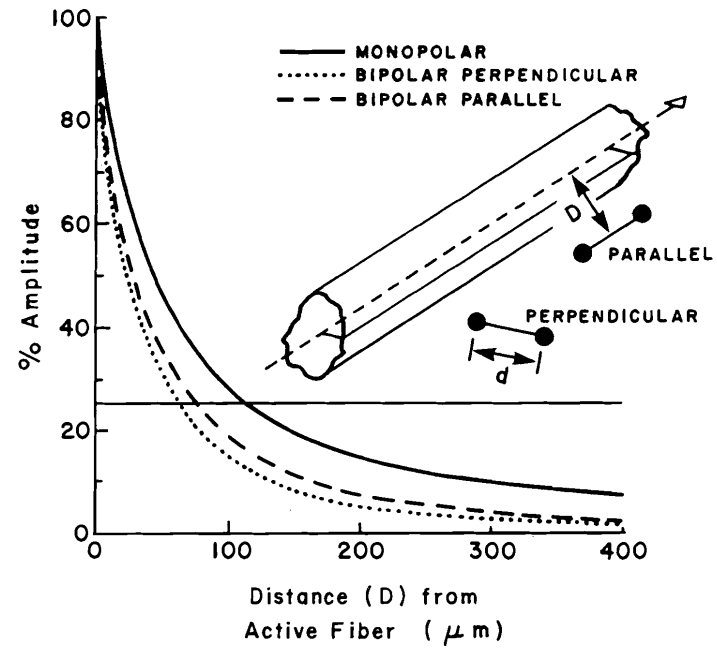

Figure 2.11. Decline of the peak-to-peak amplitude of the muscle fiber action potential for a monopolar and bipolar electrode placed perpendicularly or parallel to the direction of the muscle fibers. The *insert* describes the orientations. The *ordinate* represents the amplitude in terms of percentage of the amplitude at distance 0 from the fiber surface. The *abscissa* presents the distance from the fiber surface to the nearest detection surface. The distance at which the amplitude declines to 25% is expressed for the three electrode arrangements by the *horizontal line*: monopolar, 116 μm; bipolar perpendicular, 63 μm; bipolar parallel, 76 μm. (From S. Andreassen and A. Rosenfalck, © 1978, *IEEE Transactions in Biomedical Engineering.*)

to improve the quality of the detected signal. It also follows that caution must be taken to prevent the needle electrode from unwanted rotations during a contraction, including attempted isometric contractions.

Electrode Selectivity or Pick-Up Area and Cross-Talk

When an electrical current propagates in a volume conductor, it is theoretically possible to detect a potential at any location throughout the medium. But, as is evidenced in Figure 2.11, the voltage gradient decreases quickly. Therefore, if an electrode is placed more than 2 or 3 mm from the surface of an active muscle fiber the detected signal will have a very low amplitude, possibly lower than that of the extraneous unwanted signals, and thus will provide no useful information. It is therefore necessary to establish an arbitrary demarcation value that will define the pickup area. Several approaches have been proposed. Pollak (1971) suggested that the demarcation point be where the peak-to-peak amplitude of the action potential diminishes to 10%. Andreassen and A. Rosenfalck (1978) suggested a decrease to 25%, whereas Gath and Stålberg (1978) suggested that it be designated by the distance where the amplitude of the action potential diminishes to 200 μV. The definitions relating to the percentage decrease are more direct than those referring to an absolute value. In the latter case, the diameter of the muscle fibers will effect the measure because it is related to the amplitude of the action potential (P. Rosenfalck, 1969).

The selectivity of an electrode will depend on the area of the detection surface, and in the case of bipolar electrodes, on the distance between the two detection surfaces. By using mathematical derivations based on the earlier work of P. Rosenfalck (1969), Andreasson and A. Rosenfalck (1978) were able to determine the selectivity of different types of electrodes. Pursuing their definitions of the pickup area, they obtained values for monopolar electrodes, bipolar electrodes oriented perpendicularly to the muscle fibers, and bipolar electrodes oriented in parallel to the muscle fibers. Figure 2.12 presents the borders of the pickup area (decrease to 25%) for electrodes whose detection surface has a diameter of 25 μm. It is interesting to note that the monopolar configuration is less sensitive (larger pickup area) than the bipolar configuration; and in the latter case the selectivity increases when the detection surfaces are oriented perpendicularly to the muscle fibers.

The voltage decrement functions for the bipolar configurations presented in Figure 2.11 are obtained by performing the operation depicted in Figure 2.13. In this figure, d represents the distance between the two detection surfaces. Take the amplitude of the monopolar decrement function beginning at a distance, d, and subtract it from the decrement function of the other detection surface. The resultant describes the voltage decrement function of the differential bipolar arrangement.

In their report, Andreassen and A. Rosenfalck (1978) estimated that

Figure 2.12. Territories of pickup areas (decline of 75% of the amplitude) for: monopolar electrode (full semicircle) with detection surface at position 1; a bipolar perpendicular electrode (*dashed line* with two lobes) with detection surfaces of positions 1 and 2; and bipolar parallel electrode (*dashed semicircle*) with detection surfaces at position 1 and 25 μm above or below position 1. At various positions on the micrograph, the number of fibers within the pickup area ranges from 9 to 17 for the monopolar electrode, 2 to 7 for the bipolar perpendicular electrode, 5 to 9 for the bipolar parallel electrode. (From S. Andreassen and A. Rosenfalck, © 1978, *IEEE Transactions in Biomedical Engineering.*)

for the monopolar configuration, 9 to 17 muscle fibers are located in the pickup area. For a bipolar configuration (25-μm diameter wires spaced 50 μm apart), 2 to 9 muscle fibers would be circumscribed by the pickup area if the detection surfaces were oriented perpendicularly to the fibers, 5 to 9 if oriented parallel to the muscle fibers. The reader is cautioned that these numbers would be higher if a more generous measure of the pickup area were used, say, a decrease to 10% of the amplitude. In any case, it is apparent that the most selective electrode is the bipolar electrode, which is constructed with the smallest detection surfaces and with the smallest separation between the detection surfaces. The selectivity is further accentuated by orienting the detection surfaces in a direction perpendicular to that of the muscle fibers.

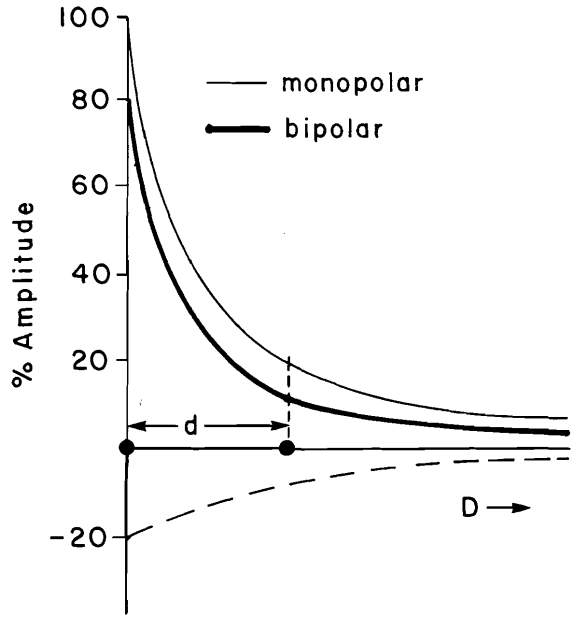

Figure 2.13. Method for calculating the decrement function of bipolar electrodes from the decrement function of a monopolar electrode. The detection surfaces of the bipolar electrode are identical and similar to those of the monopolar electrode. The distance (*D*) describes the center-to-center separation between the detection surfaces of the bipolar electrode. The operation is performed by taking the segment of the monopolar curve, beginning at *d* to infinity; inverting (as required by differential arrangement); shifting it back to $d = 0$; and subtracting its values from the original monopolar curve.

The reader is reminded that during submaximal contractions, not all the muscle fibers will be active and that adjacent fibers commonly belong to different motor units. Therefore, by judiciously placing a highly selective monopolar or bipolar needle electrode in the muscle, it is possible to detect extracellular action potentials from single muscle fibers during submaximal contractions. This is the basis of single fiber electromyography.

The high selectivity, although a blessing for some applications, may be a nightmare for others. Consider the situation where there is some relative movement between the active muscle fibers and electrode. Minor relative movements in the order of 100 μm will locate the pickup area of a selective electrode in an entirely different field of muscle fibers which, given the random intermingling of muscle fibers of different motor units, may belong to different motor units. Such minute relative movements may occur even during attempted constant-force isometric contractions.

Wire electrodes, which generally have larger detection surfaces and

are usually spaced 1 to 2 mm apart, therefore will have a larger pickup area. However, unlike needle electrodes they may move within the muscle without external indications. Such hidden relative movements between the detecting surfaces and the active fibers may cast serious ambiguity on the reliability of data relating to properties of individual motor units.

Surface electrodes, even the ones which have relatively small (less than 2 mm) detection surfaces, are not generally considered to be selective. In fact, an efficient design of surface electrodes is directed at obtaining as much activity as possible from one muscle. However, such attempts must be counterbalanced with the discrimination of EMG signals from adjacent muscles, including muscles deep to the one of interest. This "interference" of EMG signals from muscles other than the one(s) under the electrode is referred to as *cross-talk*. There is no fixed solution for guarding against cross-talk. Each electrode configuration and anatomical architecture of the adjacent musculature requires a specific solution to the design of the surface electrode.

Surface electrodes are generally used in the bipolar configuration. The differential amplification arrangement is essential to remove the unwanted "noise" signals on the surface of the skin which are generally present in most environments. (It should be mentioned that monopolar surface electrodes may be used successfully if the reference electrode is large, makes good electrical contact with the skin, and is judiciously located.) The size of the detecting surfaces is not highly critical. Although, ideally they should be as large as possible, the advantages of increasing their dimension quickly disappears above a diameter of 5 mm. Therefore, the major question with respect to selectivity is how far apart the detection surfaces should be located. This question may easily be answered by performing the calculation represented in Figures 2.13. Lynn et al (1978) proposed a rule-of-thumb to simplify the above calculation: If the electrical characteristics of the tissue(s) beneath the electrode are reasonably homogeneous, the distance between the detecting surfaces corresponds roughly to the distance from which muscle fiber will contribute meaningfully to the EMG signal. The reader is cautioned that the anisotropy of the tissue(s) beneath the electrode may considerably alter the direction of the current path and render this simplification meaningless.

We recommend that a standard interdetection-surface spacing of 1 cm be used in surface electrodes. This spacing is compatible with the anatomical architecture of most muscles in the human body. In the following section, discussing signal bandwidth considerations, it will be seen that this spacing has other advantages.

Filtering Properties of Electrode-Electrolyte Interface

All electrode-electrolyte (including skin) interfaces have an impedance

consisting of resistances and capacitances. This impedance is frequency dependent. Given that the signal generated by the muscle fibers is a current source, the electrode-electrolyte interface can be thought of in terms of a filter for the voltage generated on the electrode. Most electrodes that are used in electromyography may be conveniently considered to be a high-pass filter. The characteristics of the high-pass filter vary with electrode type and with the characteristics of the electrode-electrolyte junction. This behavior is particularly sensitive for needle electrodes having small detection surfaces (see De Luca and Forrest, 1972). The reader who is interested in more details on this topic is referred to the book by Geddes (1972). Figure 2.14 presents the magnitude of the impedance function for monopolar and bipolar needle electrodes. The detection surfaces were formed by the circular cross-section of the 25-μm diameter nickel-chromium alloy wires spaced 50 μm apart (in the bipolar configuration). The figure also shows the effect of the electrolysis procedure mentioned earlier in this chapter. These impedance values are typical of extremely selective (small detection surface) electrodes. The *dashed lines* represent Bode representations of the impedance function. The equations derived from these representations are provided for each case of Figure 2.14. Each additional term in the equations indicates and increases in the complexity of the resistor-capacitor model required to describe the behavior of the magnitude of the impedance.

For wire and surface electrodes, the magnitude of the impedance is considerably less. Antonelli et al (1982) have reported measurements for typical surface and wire electrode bipolar configurations. Their data is presented in Figure 2.15. Note that the magnitude of the impedance is generally higher for wire electrodes and for smaller detection surfaces.

Filtering Property of Bipolar Configurations

This is a property of bipolar electrodes when used in a differential amplification arrangement.

As demonstrated in Figure 2.9, the two input signals are subtracted and then amplified. Now, if the two detection surfaces are placed parallel to the muscle fibers, the action potential waveform will reach one detecting surface before the other. The differences in the times of arrival will be a function of the conduction velocity of the muscle fiber(s) and the interdetection surface separation. It follows that the frequency components of the propagated signal whose wavelength is equal to the interdetection surface separation (d) will cancel out (refer to signal 1 in Fig. 2.16*A* for a graphic explanation). In a similar fashion, it may be argued that the frequencies component whose wavelength is equal to $2d$ will be amplified with no loss. (refer to *signal 2* in Fig. 2.16A). This pattern will be repeated for every multiple integral frequency value. The

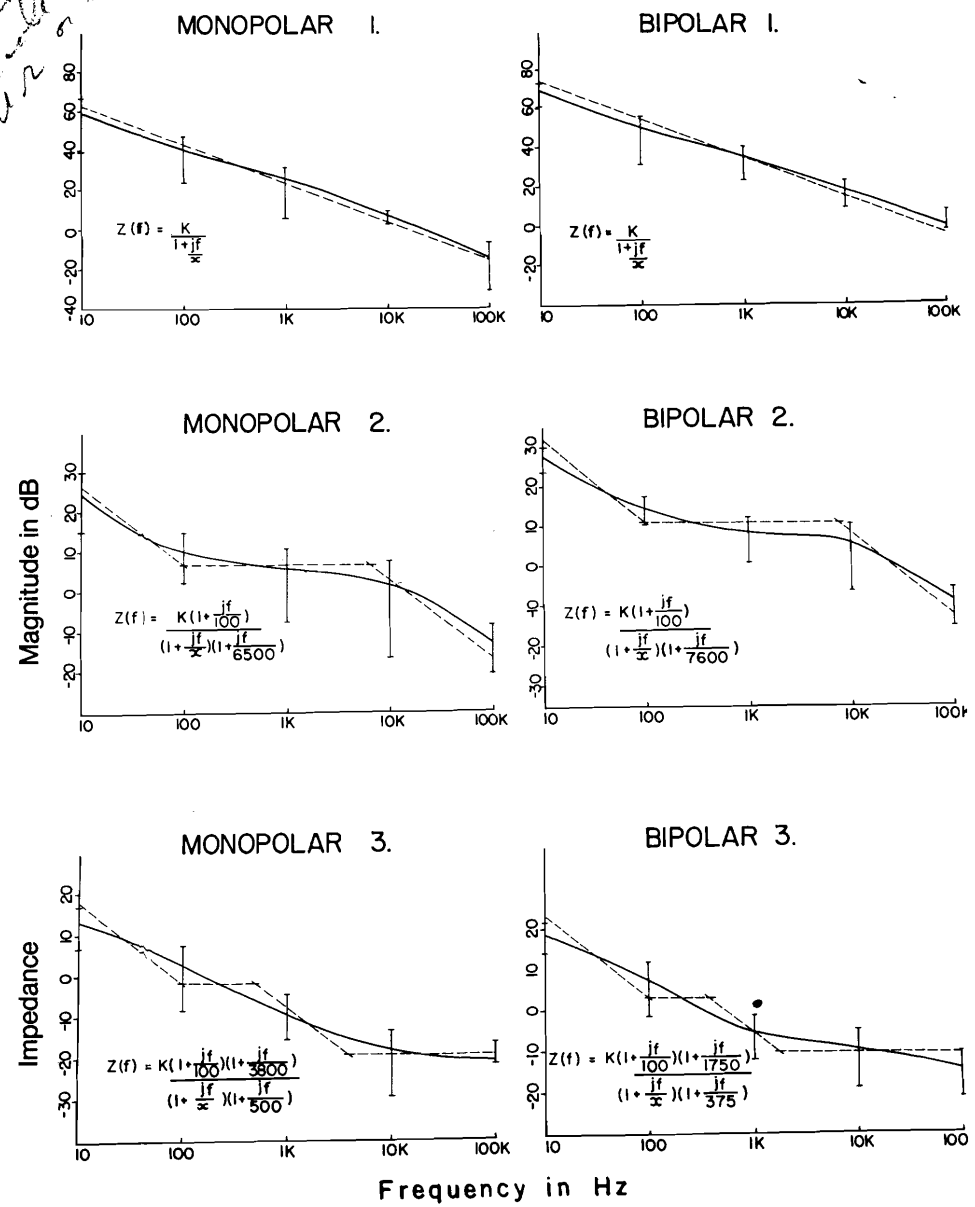

Figure 2.14. Magnitude of the impedance of a monopolar and bipolar needle electrode: (1) before electrolysis; (2) 72 hr after electrolysis and (3) 10 minutes after electrolysis. These values represent the average and standard deviations of 12 electrodes built with 25 μm wires and spaced approximately 50 μm apart. The *dashed lines* represent Bode representations which describe the impedance function. The equations derived from these representations are provided for each case. (From C.J. De Luca and W.J. Forrest, © 1972, *IEEE Transactions in Biomedical Engineering.*)

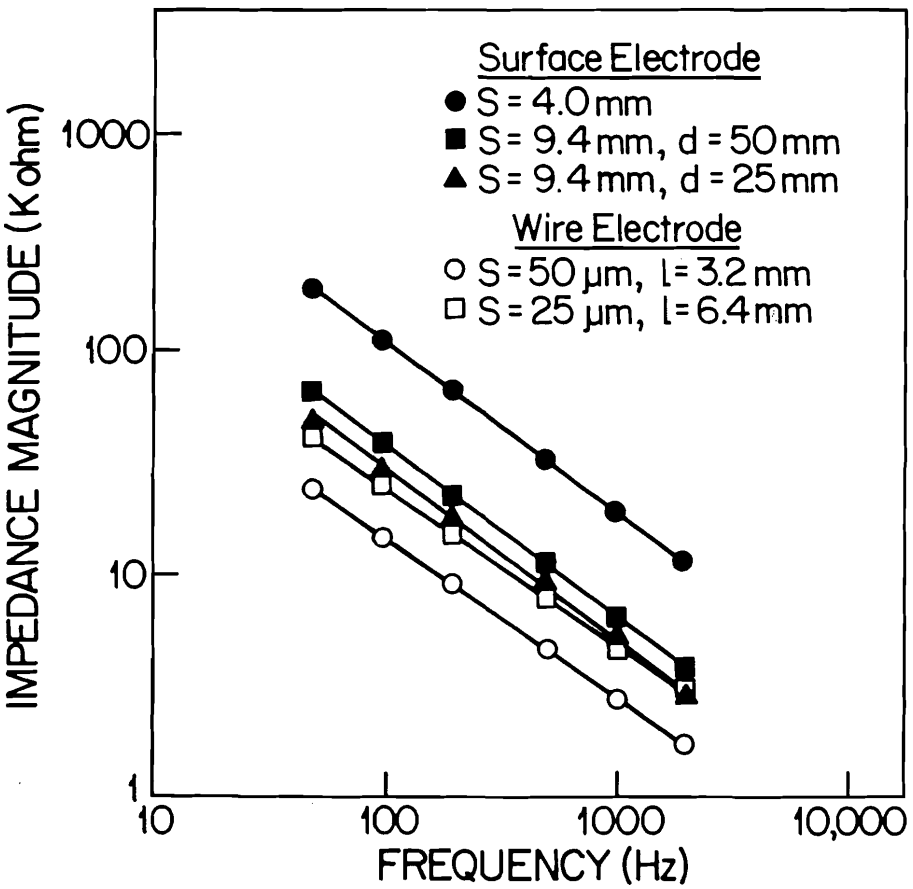

Figure 2.15. Typical values of the magnitude of the impedance of surface and wire electrodes. The *filled circle* represents a monopolar arrangement; the rest are all bipolar. *S*, diameter of detection surface; *D*, interdetection surface spacing; *L*, exposed tip length. (From D. Antonelli et al, © 1982, Pathokinesiology Lab, Rancho Los Amigos Hospital, California.)

corresponding cancellation frequency and pass frequency values may be expressed as:

$$f_{cancellation} = \frac{nv}{d} \qquad n = 1, 2, 3, \cdots$$

$$f_{pass} = \frac{nv}{2d} \qquad n = 1, 3, 5, \cdots$$

where v is conduction velocity along muscle fibers and d is interdetection surface separation. It further stands to reason that frequencies having other values will be subject to some attentuation.

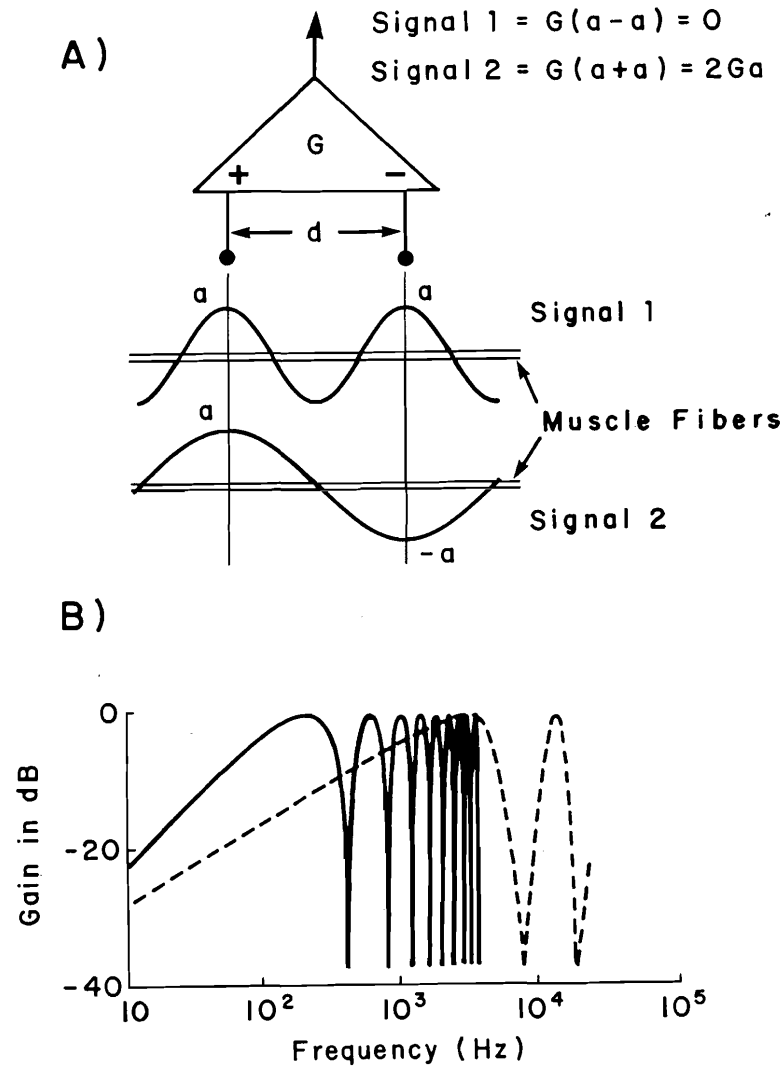

Figure 2.16. (*A*) Schematic representation of the filtering aspects of the differentially amplified bipolar detection. As the signal travels along a muscle fiber at its conduction velocity, it will pass by both detection surfaces sequentially, with a delay proportional to the interdetection surface spacing (*d*). Some of the frequency components of the signal will have wavelengths which are multiples of the distance *d* (cancellation frequency); these will cancel out when amplified differentially. When the wavelength is equal to *2d* (as in signal 2), the signal will add (pass frequency). (*B*) Alternating behavior of the filter function and cancellation frequencies. The *solid line* represents the filter function of a surface electrode and is calculated for an interdetection surface spacing of 1 cm and a conduction velocity of 4 m/s. The *dashed line* represents the filter function of a typical needle electrode and is calculated for an interdetection surface spacing of 0.5 mm and a conduction velocity of 4 m/s. It is apparent that the bipolar filter function is of concern for surface electrodes and has minor relevance in needle electrodes.

By applying elegant mathematical modeling, Lindström (1970) was able to derive the complete differential filter function, whose magnitude may be expressed as:

$$R(\omega,d) = K \sin^2\left(\frac{\omega d}{2v}\right)$$

where K is a scaling factor representing the various gain factors of the electrode-electrolyte interface. This function is displayed in Figure 2.16*B*. This particular example was calculated by letting $d = 1$ cm and $v = 4$ m/s.

In the case of needle or wire electrodes, where the interdetection surface spacing is considerably smaller, the first cancellation frequency would occur at considerably higher values. Let us take, for example, a needle electrode with an interdetection surface separation of 0.5 mm. For a conduction velocity of 4 m/s, the first cancellation point would occur at 8 kHz, and the first pass frequency would be at 4 kHz. These values are located well into the higher end of the EMG signal spectrum. Hence, only the first and possibly the second cycles will influence the EMG signal. The shape of the latter filter function is also shown in Figure 2.16*B*.

Now it must be pointed out that these functions have been calculated for one muscle fiber which is modeled as an infinitely long cylinder in an unbounded medium, a clearly unrealistic representation which, nonetheless, does provide a useful expression of the filtering characteristics of the filter. Of greater concern in the environment of real muscle fibers is that they are not all necessarily oriented parallel to each other and, therefore, parallel to the electrode, and that they do not all have identical conduction velocities. Thus, the resulting filter function of the bipolar electrodes to the EMG signal, which may be thought of as a summation of the individual filter functions of the action potential, in all likelihood will not be so well defined, nor will it have such well-defined "dips."

The above equation indicates that the bandwidth of the electrode filter function increases as the interdetection surface distance d decreases. Empirical verification of this fact has been reported by Parker and Scott (1973), Zipp (1978), and Lynn et al (1978), all of whom presented evidence that the bandwidth of the detected EMG signal increases as the interdetection surface distance decreases.

ELECTRONICS CONSIDERATIONS

It is now apparent that the EMG signal is filtered by the tissue and the electrode in the process of being detected. Before the signal may be observed, it is necessary to amplify it. This latter procedure may also modify the frequency characteristics of the signal. In order to describe

this process, it is necessary to describe some properties and parameters of electronic amplifiers. They are:

(a) Noise characteristics
(b) Signal-to-noise ratio
(c) Gain
(d) Common mode-rejection ratio
(e) Input impedance and input bias current
(f) Bandwidth

For this purpose it is useful to refer to Figure 2.9 and define the following terms:

G = gain of the amplifier
m = the detected wanted signal (the EMG signal)
n = the detected unwanted signal (the noise)

Noise

This term can be defined as any unwanted signal which is detected together with the wanted signal. Our environment is inundated with myriad electrostatic and electromagnetic fields. The presence of electrostatic fields has been completely overlooked in electromyogaphy. This oversight has been mainly due to the fact that the equipment used generally filters out DC signals. However, the presence of "static electricity" on the surface of a subject may reach proportions which may damage the electrode characteristics and, possibly, the amplifiers. High levels of static electricity are often present when a subject wears polyester clothing and the humidity level of the air is low. Electromagnetic fields are ever present in a variety of forms such as 50 or 60 Hz from power lines and electrical devices which operate on line current, radio signals, television signals, and communications signals, to name a few.

To these, also add electrical noise generated by the very equipment which we employ to detect and record the EMG signal. These are: (1) the "thermal noise" generated by the electrodes; this physical property of metals is proportional to the square root of the resistance of the detection surface and cannot be eliminated but may be reduced to the point that it is not a factor of concern by cleaning the electrode contacts, as described in a previous section in this chapter; (2) the noise generated by the first stage of the amplifiers; this is a physical property of semiconductors and cannot be completely eliminated, but may be reduced by the continual advances that are being made in semiconductor physics. The user has no recourse but to choose (or construct) an amplifier that has low noise.

To the above sources we should add another which assumes particular importance in electromyography, that is, *motion artifact*. This disturbance may occur in two locations; at the electrode-tissue interface or at the

wire leads connecting the electrodes to the amplifier. The prior source has two origins. One is any relative movement of the electrode and tissue. As described previously in this chapter, when any two materials having dissimilar electrical properties come in contact with each other, there is a lack of chemical equilibrium at the junction, which in turn generates a polarization potential. Any relative movement at the junction modulates the polarization potential and generates an AC current which generates the noise signal. The other is the "skin potential." Under normal conditions, a voltage of approximately 20 mV exists across the skin layers. It is generally believed that this potential is originated by "injury currents" of the dead cells as they migrate to the surface of the skin. In any case, the voltage varies as the skin is stretched, as is the case when the muscles underneath it contract or as a limb is displaced. It is of interest to point out that abrasion of the skin reduces this component of motion artifact because as the skin is pierced, the voltage across the skin is shorted out (refer to Burbank and Webster, 1978, for details). The noise resulting from the leads movement is caused by the natural phenomenon that is used to create current in a generator. That is, a metallic wire (the lead) is moved through electromagnetic fields (which as described before are pervasive in our environment). The voltage resulting from these mechanical artifacts may be large (several millivolts) so that they seriously contaminate the EMG signals. This problem is accentuated when the input impedance of the amplifier is high because a small current passing through a high impedance may generate a high voltage. The reader is reminded of the basic relation ($V = IZ$) between voltage (V), impedance (Z), and current (I). It should also be pointed out that because these forms of electrical noise are induced by movements of the body tissues, they will be limited to frequency components that are less than 30 Hz. The human body, or parts of it, will not oscillate with any easily measurable energy or higher frequencies.

All these noise sources are the critical ones because they are added to the wanted signal prior to amplification; therefore any amplification will increase both the wanted and unwanted signals.

Signal-to-Noise Ratio

In any scheme for detecting, amplifying, or recording signals, the ratio of the wanted signal to the unwanted signal is the single most important factor to be considered. It is the factor which measures the quality of the signal.

Gain

Referring to Figure 2.9, we can describe the amplification of the detected signal in the following fashion:

Monopolar amplified signal = $G(m + n)$

Bipolar amplified signal = $G[(m_1 + n) - (m_2 + n)]$
$= G(m_1 - m_2)$

The advantage of the bipolar configuration is now apparent. Ideally, the noise component is removed. This idealized representation for the differential amplification associated with the bipolar detection configuration indicates that if the noise signal fed to the amplifier is similar in all respects (amplitude, phase and frequency components), then it will be totally eliminated. This perfect cancellation does not occur in real differential amplifiers for two reasons. First, the amplifiers cannot subtract perfectly. The measure of how well the differential amplifier subtracts (reject) the common mode signal is called the "common mode rejection ratio," and will be addressed in the following section. The second reason is that the noise signal reaching the two input stages of the differential amplifier is not necessarily common mode. This is particularly true if the tissue media is anisotropic.

Another point that should be mentioned concerns the amount of amplification required to observe or record the EMG signal. It is apparent in the above formulation that the bipolar configuration will require greater amplification. However, this is of no concern because the values of the gains required in both cases are well within the capabilities of ordinary electronics amplifiers.

Common Mode-Rejection Ratio

In practice, the performance of differential amplifier circuits departs from the ideal characteristics mentioned above. Gain imbalance and nonlinearities in the amplifier's differential input stages cause errors in the subtraction process. As a result, signals common to each input (E_{cm}) do not cancel completely and produce an undesirable common-mode error voltage at the amplifier output (E_e). The ratio between the common-mode voltage (E_{cm}) of the amplifier and its common-mode error voltage (E_e) is defined as the common-mode rejection ratio (CMRR)

$$CMRR = \frac{E_{cm}}{E_e}$$

The importance of the CMRR becomes apparent when dealing with the effects of external fields such as power line-induced interference radiating from the environment. Referring to Figure 2.17, we can model the effect of an external signal field acting on the tissue media as two current sources (i_n) in parallel with their respective tissue impedances (Z_{tn}). If the tissue media impedance (Z_n) is isotropic, and the external field gradient across the tissue media is constant, then the fields induced currents (i_n) at each input are equal and will cancel. Obviously, the higher the CMRR of the amplifier, the better the cancellation of these undesirable currents.

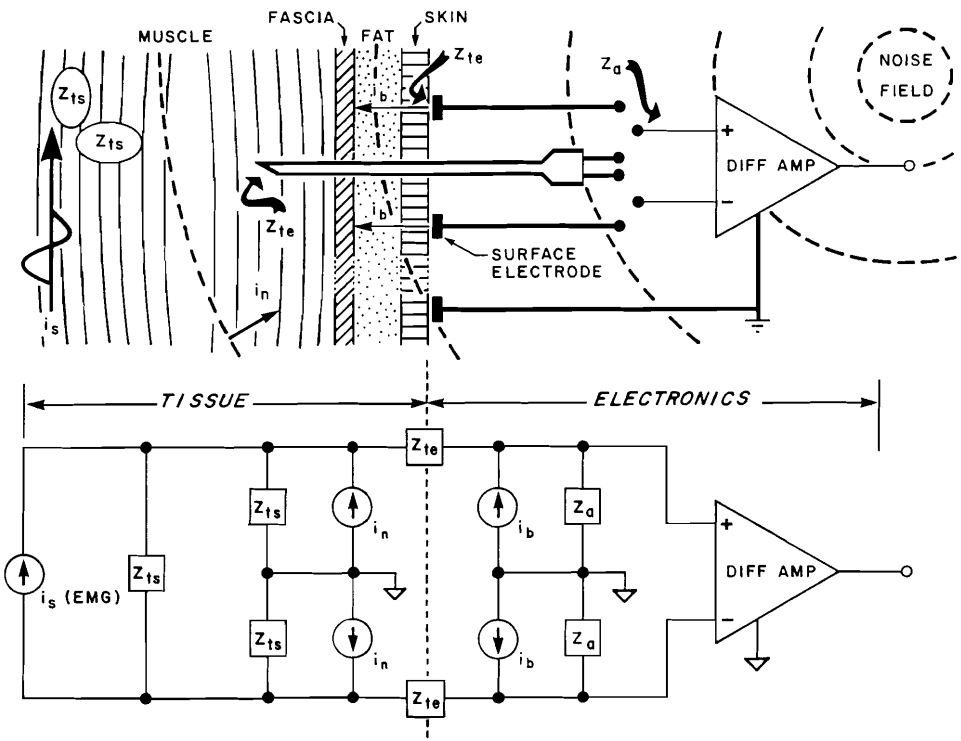

Figure 2.17. (*Top*) Diagramatic representation of impedances and currents in the tissues, electrode, and amplifier:

i_s = source current from EMG signal
i_n = common current from noise
i_b = input bias current from amplifier
Z_{ts} = tissue impedance seen by source current
Z_{tn} = tissue impedance seen by noise current
Z_{te} = tissue electrode impedance of the metal electrolyte interface
Z_a = input impedance of amplifier

(*Bottom*) The schematic diagram representing the electrical interaction of the EMG signal, extraneous noise, electrode, and amplifier.

Input Impedance and Input Bias Current

In order to accurately measure the amplitude and waveshape parameters of the EMG signal, it is necessary to understand how the input impedance and input bias current of the differential amplifier can influence these parameters. The concept of impedance has been discussed at the beginning of this chapter. The input bias current may be thought of as the minimal constant current required to keep the amplifier active. Since the differential amplifier is not ideal, it has a finite impedance at each input and nonzero input bias current. As demonstrated in Figure

2.17, the bias current flows out of the amplifier. Thus, it stands to reason that any signal which has a current less than the bias current will not be amplified. In modern amplifiers this current is considerably small (< 100 pA) so that it does not present any danger to the subject when the electrode is on the skin or in a skeletal muscle. However, in needle electrodes which have small detection surfaces (<100 μm), the current may be sufficient to alter the chemical structure of the surface layer over repeated applications. This, in turn, will alter the metal-electrolyte filtering characteristics of the electrode. (Referring to Fig. 2.17, we can model the input of the amplifier with an impedance (Z_a) and current source (i_b) from each input to ground reference. When an EMG signal source is connected to the inputs, these bias currents flow through the signal source. The output of the signal source is likewise shunted across the input impedance (Z_a). The amount of signal shunting depends on Z_a and the distributed impedance of the EMG signal source.

The distributed impedance of the EMG signal source is determined by the impedance characteristics of the tissue (Z_{ts}) and the tissue-electrode interface, (Z_{te}). As discussed in previous sections, these impedances have both resistive and reactive components due to the capacitive effects of tissue media and electrode interface. The value of the distributed source impedance can vary greatly, depending on the impedance of the needle or surface electrode interface configuration (Z_{te}) and the amount of intervening tissue (Z_{ts}), typically 10^4 to 10^6 ohms at 1 kHz. *To minimize waveshape distortion and attentuation of the signal source due to the shunting by the amplifier, the input impedance (Z_a) should be much larger (10^{12} in parallel with 5 pf) than the distributed source impedance.*

Bandwidth

All amplifiers have limits on the range of frequency over which they operate. In fact, limitations of amplifiers are commonly measured in gain-bandwidth quotient. The value of the quotient is defined by the type of semiconductor components used. This limitation does not present a problem in electromyography because amplifiers providing the required gain over the necessary bandwidth are commonly available and easy to design. The bandwidth of an amplifier may be conceptualized as a window in the frequency domain. The frequencies of a signal that coincide between the borders of the window, i.e., the bandwidth, will pass with minimal, if any, diminution; whereas, frequency components outside the bandwidth will be suppressed or eliminated. In this sense, an amplifier can be considered as a filter with gain (refer to Fig. 2.1 and the discussion at the beginning of the chapter for more details).

Some amplifiers are specially designed to amplify dc signals. Such amplifiers require special circuitry to eliminate "drift" as a function of temperature. They generally have low bandwidths, usually up to 100

Hz. They are typically used to amplify signals such as those generated by force, pressure, and temperature transducers.

In electromyography, it is highly advisable not to DC couple the electrodes to the amplifiers for the following reasons: (1) The dc polarization potential present at the electrode-electrolyte interface may be as large as the EMG signal being detected. (2) Motion artifacts in the lead wire will generally have low frequencies (<20 Hz), and thus they would also be amplified. (3) The frequency components of the EMG signal below 20 Hz are unstable and fluctuate with a considerable amount of unpredictability. This point will be elaborated in Chapter 3. For these reasons, it is recommended that for general applications the low-frequency 3 dB point be set at 20 Hz. The high frequency 3 dB point should be set to a value slightly higher than the highest frequency components of the wanted signal. As will be seen shortly, this value is dependent on the type of electrode used to detect the signal. Any noise signal having frequency components greater than the high 3 dB point will be attentuated, thus increasing the signal-to-noise ratio of the amplified signal.

RECORDING ASPECTS

Having detected and amplified the signal, it is necessary to display it in some fashion so that it may be employed. A variety of devices and media may be used for this purpose. The most common approach is to display the amplified signal on an oscilloscope. If a permanent record is required, it is useful to record the signal on a frequency-modulated (FM) tape recorder. FM tape recorders are used because they have a bandpass which includes dc. The great advantage of storing the signal on FM tape is that it may be replayed and transferred to other media. Permanent visual records are commonly produced on strip-chart recorders, of which there are at least three types. One type prints on paper via an electromechanical pen, another via a galvanographic or oscillographic process on ultraviolet sensitive paper, and the third prints on paper via an ink jet spray.

The primary concern of recording the signal is that the bandwidth of the device be greater than the bandwidth of the amplified signal. If the signal has been amplified with a proper gain (yielding a ±1 V peak-to-peak value), the noise considerations become minimal at this stage. For oscilloscopes, this is a trivial issue, because they all have much wider bandwidth specifications. FM magnetic tape recorders have bandwidths whose upper 3 dB point is linearly related to the speed of the tape. All brand-name commercially available FM tape recorders have bandwidths which are required for EMG signal storage. It is only necessary to ensure that the appropriate speed is used when recording the signal.

Strip chart recorders present a more restrictive barrier. The type which prints with an electrical-mechanical pen has severe bandwidth

limitations, typically 0 to 60 Hz. Clearly any attempt to record the EMG signal directly on such devices sacrifices most of the energy in the signal and, thus, most of the information. However, such devices are obviously convenient. They may be properly employed if the signal is first recorded on FM tape at a relatively high speed, and is subsequently played back to the strip-chart recorder at a lower speed. The ratio of the high to low speeds effectively multiples the bandwidth of the strip-chart recorder. So, if a speed reduction ratio of 32 is used, the strip-chart recorder will have an effective bandwidth of approximately 0 to 2000 Hz. Such a bandwidth is acceptable for surface electrodes and adequate for wire and needle electrodes. This procedure has the disadvantage that it cannot be performed in real time. The ultraviolet sensitive strip-chart recorders, as well as some ink jet recorders which are now appearing on the market, have bandwidths that are considerably higher, typically 0 to 750 Hz. These may be used to record EMG signals detected with surface electrodes directly without sacrificing a significant part of the signal. They may be (and are often) used to record signals directly from indwelling electrodes. Although they provide a visually proper representation of signals obtained by indwelling electrodes, it is wise to remember that the higher frequency components have been removed. Therefore, measurements of the amplitude of peaks and peak-to-peak slopes will misrepresent the actual (postdetected, prerecorded) values. However, these devices may be used in real time, and as such may inherently provide some advantages, especially in clinical environments. Nonetheless, the images traced on the strip-chart must be interpreted with the forewarned caution.

With the exploding popularity of digital computers in laboratories and clinics, it has become possible to record the EMG signals directly on digital storage media (computer memory, disks, or digital magnetic tape). This is accomplished by sampling the signal at regular intervals and expressing the amplitude value at each point as a binary value (power of 2) and storing this value. This operation is known as digitization. In this operation, the sampling rate is an important factor. A minimal requirement to preserve the frequency information of the signal is that the sampling rate be at least twice the value of the highest frequency component of the signal. This is known as the Nyquist frequency. Other considerations such as signal amplitude, prefiltering, aliasing, etc., must also be addressed when digitizing a signal. Although computer programs for performing this operation are readily available, the novice is strongly advised to obtain assistance from competent individuals prior to attempting this operation.

OVERVIEW OF DETECTION AND RECORDING PROCEDURES

It is now apparent that the entire procedure of acquiring an observable EMG signal consists of a catenation of filtering processes, each of which

modifies, in some respects, the amplitude and frequency characteristics of the observed signal.

It is important to remember that the characteristics of the observed EMG signal are a function of the apparatus used to acquire the signal as well as the electrical current which is generated by the membrane of the muscle fibers.

For the sake of convenience, the block diagram in Figure 2.18 presents all the individual filtering steps which have been discussed. Some minor filtering effects due to the size of the detection surfaces, location of the electrode with respect to the neuromuscular junction and tendons, etc. have not been discussed. Their effects are difficult to calculate and do not alter the signal characteristics in any considerable amount.

Practical Considerations

It is always desirable to detect and record the EMG signal with minimal distortion and extraneous contamination. We will now list some practical considerations which will assist in accomplishing the task. These pointers essentially represent a summary of the previous discussions in this chapter, the details of which should now be familiar to the reader.

Tissue Filtering

1. The voltage decrement function decreases rapidly with distance; therefore, inserted electrodes will only detect signals from nearby muscle fibers. The amplitude of action potentials decreases to 25% within 100 μm.

2. The filtering characteristics of the muscle tissues is a function of the

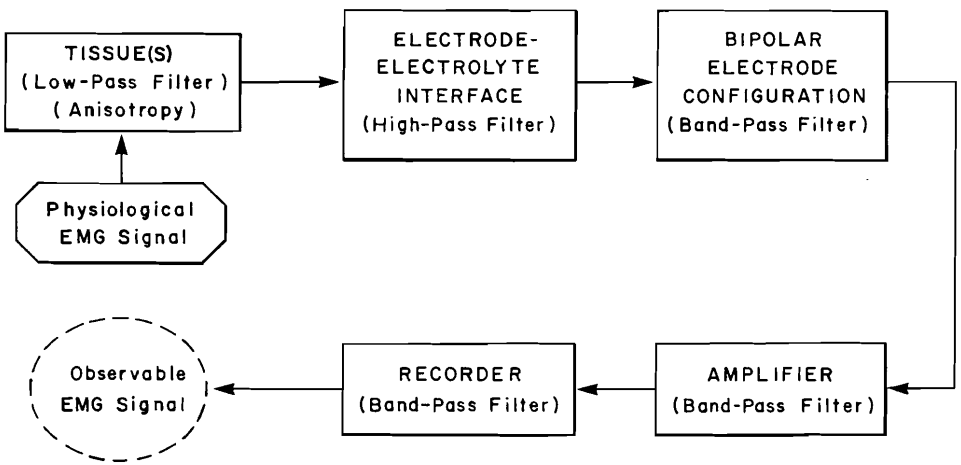

Figure 2.18. Block diagram of all the major aspects of the signal acquisition procedure. Note the variety of physical properties that act as filters to the EMG signal before it can be observed. The term physiological EMG signal refers to the collection of signals which emanate from the surface of the muscle fibers. They are not observable.

distance between the active muscle fibers and the detection surface(s) of the electrode. In the case of surface electrodes, the thickness of the fatty and skin tissues must also be considered. The tissue(s) behaves as a low-pass filter whose bandwidth and gain decreases as the distance increases.

3. The muscle tissue is highly anisotropic. Therefore, the orientation of the detection surfaces of the electrode with respect to the length of the muscle fibers is critical.

Electrode-Electrolyte Interface

1. This electrochemical junction also behaves as a high-pass filter.

2. The gain and bandwidth will be a function of the area of the detection surfaces, electrolytic treatment of the surfaces, and any chemical-electrical alteration of the junction.

3. The detection surfaces should always be kept clean.

Bipolar Electrode Configuration

1. This property ideally behaves as a band-pass filter. However, this is only true if the inputs to the amplifier are balanced and if the filtering aspects of the electrode-electrolyte junctions are equivalent.

2. A larger interdetection surface spacing will render a lower bandwidth. This aspect is particularly significant for surface electrodes.

3. The greater the interdetection surface spacing, the greater the susceptibility of the electrode to detecting measurable amplitudes of EMG signals from adjacent and deep muscles. Again, this aspect is particularly significant for surface electrodes. A rule of thumb is that the electrodes will detect measurable signals from a distance equal to the interdetection surfaces spacing. However, the anisotropy of the tissues beneath the electrode may augment the sensitivity of the electrodes along the surface of the muscle creating cross-talk.

4. An interdetection surface spacing of 1.0 cm is recommended for surface electrodes.

Amplifier Characteristics

1. These should be designed and/or set for values which will minimally distort the EMG signal detected by the electrodes.

2. The length of the leads to the input of the amplifier (actually, the first stage of the amplification) should be as short as possible and should not be susceptible to movement. This may be accomplished by building the first stage of the amplifier (the preamplifier) in a small configuration which may be physically located near (within 10 cm) of the electrode. The necessity of this precaution is accentuated when amplifiers with high input impedance ($>10^7$ ohm) are used.

3. Typical settings and characteristics are:

a. Gain: such that it renders the output with an amplitude of approximately ± 1 V.

b. Input impedance > 10^{12} ohms resistance in parallel with 5 pf capacitance.
c. Common mode rejection ratio: >100 dB
d. Input bias current: as low as possible (typically less than 50 pA)
e. Noise < 5μV rms
f. Bandwidth (3 dB points for 12 dB/octave rolloff):

Surface electrodes	20–500 Hz
Wire electrodes	20–1000 Hz
Monopolar and bipolar needle electrodes for general use	20–1000 Hz
Needle electrodes for signal decomposition	1000–10,000 Hz
Single fiber electrode	20–10,000 Hz
Macroelectrode	20–10,000 Hz

Recording Characteristics

The effective or actual bandwidth of the device or algorithm that is used to record or store the signal must be greater than that of the amplifiers.

Other Considerations

1. It is preferable to have the subject, the electrode, and the recording equipment in an electromagnetically quiet environment. However, if all the procedures and cautions discussed in this chapter are followed and heeded, high quality recordings will be obtained in the electromagnetic environments found in most institutions, including hospitals.

2. When using indwelling electrodes, great caution should be taken to minimize (eliminate, if possible) any relative movement between the detection surfaces of the electrodes and the muscle fibers. Relative movements of 0.1 mm may dramatically alter the characteristics of the detected EMG signal and may possibly cause the electrode to detect a different motor unit population.

WHERE TO LOCATE THE ELECTRODE

Prior to entertaining a discussion on the preferred location of electrodes in a muscle or on the surface of a muscle, we would like to dispel a common fallacy which, to our knowledge, has been prevalent for many years among electromyographers, that is, the common practice of locating the detection electrode in the vicinity of the motor point of the muscle. (This location is the surface projection of the anatomical center of the innervation zone of the muscle.) The reasoning presumably is that the motor point is to some extent an electrically and anatomically definable point. Another fallacious rationale is at times advanced by the belief that the motor point, being the location where an externally applied electrical current causes the maximal excitation of the muscle, should therefore be the location that provides the greatest signal amplitude. This is not so. Even if it were so, the availability of larger amplitude is not an important consideration. However, there are three important considerations. They are: (1) signal-to-noise ratio; (2) signal stability

(reliability); and (3) cross-talk from adjacent muscles. The importance of the signal-to-noise ratio has been discussed already. The stability consideration addresses the issue of the modulation of the signal amplitude due to relative movement of the active fibers with respect to the detection surfaces of the electrode. The issue of the cross-talk concerns the detection by the electrode of signals emanating from adjacent muscles.

For most configurations of needle electrodes the question of cross-talk is of minor concern, because the electrode is so selective that it only detects from nearby muscle fibers (see Fig. 2.12). Due to the fact that the muscle fibers of different motor units are scattered in a semirandom fashion throughout the muscle, the location of the electrode becomes irrelevant from the point of view of signal quality and information content. The stability of the signal will not necessarily be improved in any one location. It is, nonetheless, wise to stay clear of the innervation zone so as to reduce the probability of irritating a nerve ending. And of course, one should stay clear of suspected locations of blood vessels. These two latter precautions should be taken, not only as a courtesy to the subject, but also for technical considerations. If a nerve ending is irritated, "spontaneous" signals unrelated to the motor aspects of muscles will be detected. If a blood vessel is ruptured, the resulting pool of blood in the interstitial spaces will "short" the input of the electrode by decreasing the impedance between the detection surfaces.

For wire electrodes, all the considerations which have been discussed for needle electrodes also apply; and in this case, any complication will be unforgiving, in that the electrode may not be relocated. Because the wire electrodes have a larger pickup area, a concern arises with respect to how the location of the insertion effects the stability of the signal. This question is even more dramatic in the case of surface electrodes and will be discussed in that context.

The precautions required to reduce cross-talk with surface electrodes has already been discussed. In this section we will discuss the susceptibility of signal to the location on the muscle. In order to measure this susceptibility, it is necessary to devise a procedure whereby the excitation to the muscle remains constant while the location of the electrode is varied. Gerbino, Gilmore, and De Luca satisfied this condition by supramaximally stimulating, with 0.5-ms current pulses, the tibialis nerve of rabbits via implanted stimulation electrodes. In this fashion, a compound action potential of consistent amplitude and shape was generated in the gastrocnemius muscle. The detection electrode was always arranged with its detection surfaces along the length of the fibers of the gastrocnemius muscle and was located in four distinct locations: near the Achilles tendon; halfway between the tendon and the center of the innervation zone; on the innervation zone; and between the innervation zone and the origin. While the electrode was maintained in each location, the

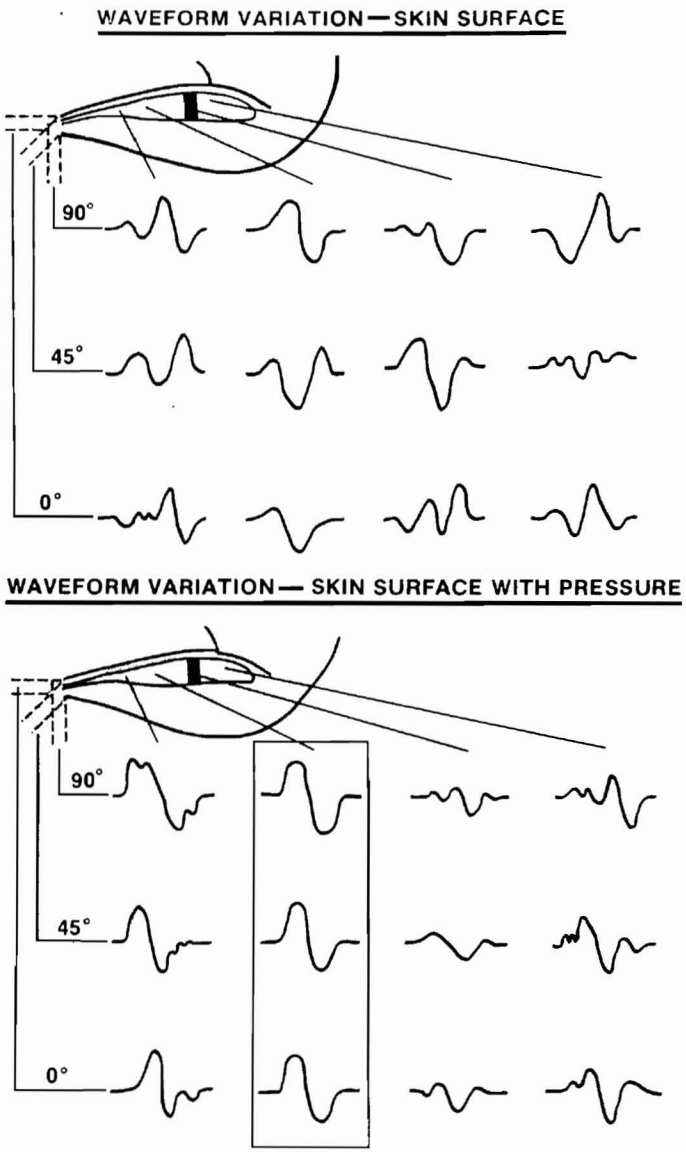

Figure 2.19. Compound action potentials detected with bipolar electrodes from the surface of the skin above the gastrocnemius muscle of a rabbit. A stimulation electrode was implanted around the tibialis nerve, and supramaximal electrical stimulation was applied. The detection electrode was located in four positions: near the Achilles tendon; 1 halfway between the tendon and the center of the innervation zone (*double-shaded* band); the center of the innervation zone; and between the innervation zone and the origin. The angle of the ankle joint was held at 0, 45, and 90. (*Top panel*) The electrode was attached to the skin in the normal fashion. (*Bottom panel*) the electrode was held firmly against the skin and underlying muscle, reducing any relative movement between the muscle and the electrode. Note the more consistent waveforms in the latter case. Also note that the preferred location is the area between the insertion tendon and the innervation zone.

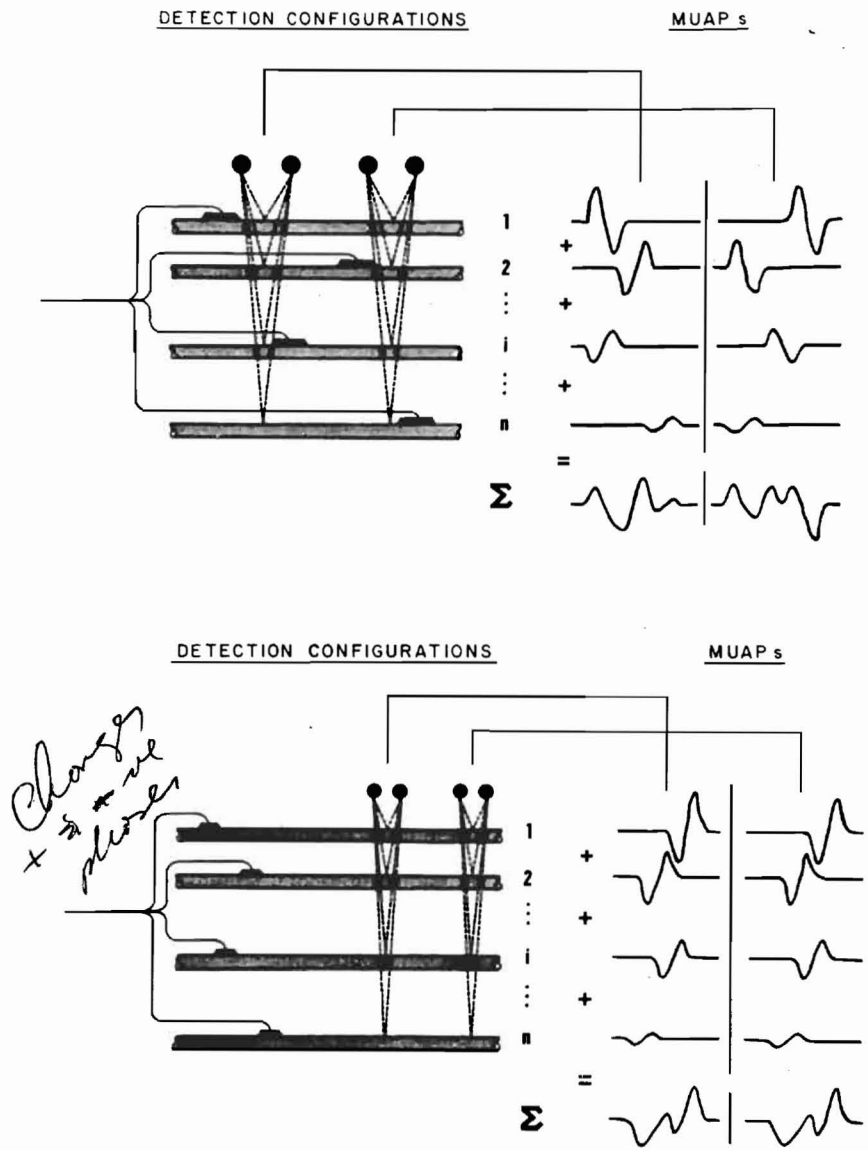

Figure 2.20. Schematic diagram explaining the susceptibility of the motor unit action potential (MUAP) waveform to electrode displacement. *Top panel* represents two detection locations in the innervation zone and the associated modifications in the detected MUAP. *Bottom panel* presents the similar data for two comparable detection locations well away from the innervation zone.

ankle was fixed at 0°, 45°, or 90°, thus simulating the muscle length changes during shortening contractions. The compound action potential was detected for each electrode location and for each ankle position. Two sets of measurements were taken. In one set, the electrode was attached to the skin in a normal fashion; in the other set, the electrode was held firmly against the skin with pressure applied against the muscle, thereby reducing the relative movement of the electrode and skin with respect to the muscle fibers.

A typical example of the results is displayed in Figure 2.19. Note that the shape of the compound action potential is susceptible to the location of the electrode along the muscle, and in the case where no pressure was applied to stabilize the electrode, the shape was altered by the position of the ankle. In both cases, the location which provided the most unreliable results was the one corresponding to the innervation zone.

These observations may be conveniently explained by using a schematic approach for the formulation of the MUAP, which in some sense has a comparable genesis to the compound action potential. (The reader is referred to Chapter 3 for a detailed explanation of this concept.) This schematic representation is presented in Figure 2.20. The MUAP is formed by linearly superimposing the action potentials from all the active muscle fibers in the vicinity of the detection electrode. In the *top panel*, the electrode is located near the innervation zone, in the *bottom panel* away from the zone. It is apparent in the top panel that a slight displacement of the electrode results in a drastically different waveform for the MUAP, whereas in the bottom panel an equivalent displacement does not result in such drastic alterations of the waveform.

It is also apparent from Figure 2.20, and may be noted in the empirical results of Figure 2.19, that the relative movement of the detection electrode with respect to the active fibers (as may easily occur when a muscle shortens and the electrode remains stationary on the skin) may, by itself, alter the characteristics of the detected signal. This modification of the EMG signal would not be related to physiological aspects of the contracting muscle. A situation of this nature is found in investigations which acquire EMG signals during anisometric contraction, such as gait. The modulation of the amplitude of the EMG signals obtained from lower limb muscles during gait should be interpreted with great caution.

We suggest that the preferred location of an electrode is in the region halfway between the center of the innervation zone and the further tendon.

CHAPTER 3

Description and Analysis of the EMG Signal

The EMG signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle. It is an exceedingly complicated signal which is affected by the anatomical and physiological properties of muscles, the control scheme of the peripheral nervous system, as well as the characteristics of the instrumentation that is used to detect and observe it. Most of the relationships between the EMG signal and the properties of a contracting muscle which are presently employed have evolved serendipitously. The lack of a proper description of the EMG signal is probably the greatest single factor which has hampered the development of electromyography into a precise discipline.

This chapter will present two main concepts. The first is a discussion of a structured approach for interpreting the information content of the EMG signal. The mathematical model which is developed is based on current knowledge of the properties of contracting human muscles. These properties are discussed in Chapter 5. The extent to which the model contributes to the understanding of the signal is restricted to the limited amount of physiological knowledge currently available. However, even in its present form, the modeling approach supplies an enlightening insight into the composition of the EMG signal.

The second concept in this chapter concerns a discussion of methodologies that are useful for processing and analyzing the signal.

REVIEW OF NOMENCLATURE

Throughout this chapter, specialized terms will be used to describe distinct aspects of signals. These terms will be defined now so as to eliminate possible confusion.

Waveform—The term which describes all aspects of the excursion of the potential, voltage, or current associated with a signal or a function of time. It incorporates all the notions of shape, amplitude, and time duration.
Amplitude—That quantity which expresses the level of signal activity.
Time duration—The amount of time over which a waveform presents detectable energy.
Phase—In electromyography, this term refers to the net excursion of the amplitude of a signal in either the positive or negative direction.
Shape—The characteristics of a signal which remains unaltered with linear scaling in either the amplitude or time domains. An example of such characteristics is the phases of an action potential.

The distinction between the concept of shape and waveform is depicted