EGG which is disorganized for several weeks following operation. Differences have been shown following different types of vagotomy.

The EGG has a relatively short history compared with other electrophysiological measures and therefore the research concerned with its validation and application is still far from complete. The technique holds a lot of promise for clinical investigations of gastric disorders but must currently be used with very great care.

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ELECTROMYOGRAPHY

The now classical definition of electromyography describes it as the study of muscle function through the inquiry of the electrical signal the muscle emanates. This definition had its origin among the physiological and clinical practitioners. Today the collection of participants has expanded to include the natural scientist and engineer. Thus, a more comprehensive and appropriate definition would be the following: Electromyography is the discipline that deals with the detection, analysis, and use of the electrical signal that emanates from muscles.

The electrical signal that emanates from contracting muscles has been referred to as the electromyographic (EMG) signal, a term which was more appropriate in the past than in the present. In days past, the only way to capture the signal for subsequent study was to obtain a "graphic" representation. Today, of course, it is possible to store the signal on magnetic tape, disks, and electronics components. Even more means will become available in the near future. This evolution has made the graphics aspect of the nomenclature a limited descriptor. Although a growing number of practitioners choose to use the term "myoelectric (ME) signal," the term "EMG" still commands dominant usage, especially in clinical environments.

The novice in this field may well ask, why study electromyography? Why bother understanding the EMG signal? There are many and varied reasons for doing so. Even a superficial acquaintance with the scientific literature will uncover various current applications in fields such as neurophysiology, kinesiology, motor control, psychology, rehabilitation medicine, and biomedical engineering. Although the state of the art provides a sound and rich complement of applications, it is the potential of future applications that generates genuine enthusiasm.

HISTORICAL PERSPECTIVES

According to the modified definition, electromyography had its earliest roots in the custom practiced by the Greeks of using electric eels to "shock" ailments out of the body. The origin of the shock that accompanied this earliest detection and application of the EMG signal was not understood until 1666 when an Italian, Francesco Redi, associated it with muscle tissue (1). This concept was proved by Luigi Galvani (2) in 1791 who staunchly defended the notion. During the ensuing six decades, a few investigators dabbled with this newly discovered phenomenon, but it remained for DuBois-Raymond (3) in 1849 to prove that the EMG signal could be detected from human muscle during a voluntary contraction. This pivotal discovery remained untapped for eight decades awaiting the development of technological implements to exploit its prospects. This interval brought forth new instruments such as the cathode ray tube, vacuum-tube amplifiers, metal electrodes, and the revolutionary needle electrode which provided means for conveniently detecting the EMG signal. This simple implement introduced by Adrian and Bronk (4) in 1929 fired the imagination of many clinical researchers who embraced electromyography as an essential resource for diagnostic procedures. Noteworthy among these was the contribution of Buchthal and his associates.

Guided by the work of Inman et al. (5), in the mid-1940s to the mid-1950s several investigations revealed a monotonic relationship between the amplitude of the EMG signal and the force and velocity of a muscle contraction. This significant finding had a considerable impact: It dramatically popularized the use of electromyographic studies concerned with muscle function, motor control, and kinesiology. Kinesiological investigations received yet another impetus in the early 1960s with the introduction of wire electrodes. The properties of the wire electrode were diligently exploited by Basmajian and his associates during the next two decades.

In the early 1960s, another dramatic evolution occurred in the field: myoelectric control of externally powered prostheses. During this period, engineers from several countries developed externally powered upper-limb prostheses which were made possible by the miniaturization of electronics components and the development of lighter, more compact batteries that could be carried by the amputee. Noteworthy among the developments of externally powered prostheses was the work of the Yugoslavian engineer Tomovic and the Russian engineer Kobrinski, who in the late 1950s and early 1960s provided the first examples of such devices.

In the following decade, a formal theoretical basis for electromyography began to evolve. Up to this time, all knowledge in the field had evolved from empirical and often anecdotal observations. De Luca (6,7) described a mathematical model that explained many properties of the time-domain parameters of the EMG signal, and Lindström (8) described a mathematical model that explained many properties of the frequency-domain parameters of the EMG signal. With the introduction of analytical and simulation techniques, new approaches to the processing of the EMG signal surfaced. Of particular importance was the work of Graupe and Cline (9) who employed the autoregressive moving average technique for extracting information from the signal.

The late 1970s and early 1980s saw the use of sophisticated computer algorithms and communication theory to decompose the EMG signal into the individual electrical activities of the muscle fibers (10,11). Today, the decomposition approach promises to revolutionize clinical electromyography and to provide a powerful tool for investigating the detailed control schemes used by the nervous system to produce muscle contractions. Other techniques such as the use of median and mean frequencies of the EMG signal to describe the functional state of a muscle and the use of the conduction velocity of the EMG signal to provide information on the morphology of the muscle fibers are currently being developed.

The reader who is interested in more historical and factual details is referred to the book *Muscles Alive* (5th ed.) by Basmajian and De Luca (12).

DESCRIPTION OF THE MYOELECTRIC SIGNAL

The EMG signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle. The signal represents the current generated by the ionic flow across the membrane of the muscle fibers which propagates through the intervening tissues to reach the detection surface of an electrode located in the environment. It is an exceedingly complicated signal which is affected by the anatomical and physiological properties of muscles and the control scheme of the nervous system, as well as the characteristics of the instrumentation used to detect and observe it. Most of the relationships between the EMG signal and the properties of a contracting muscle that are currently in use have evolved serendipitously. The lack of a proper description of the EMG signal is probably the greatest single factor that has hampered the development of electromyography into a precise discipline.

Motor Unit Action Potential

The most fundamental functional unit of a muscle is called the motor unit. It consists of an α motoneuron and all the muscle fibers that are innervated by the motoneuron's branches. The electrical signal that emanates from the activation of the muscle fibers of a motor unit that are in the detectable vicinity of an electrode is called the motor unit action potential (MUAP). This constitutes the fundamental unit of the EMG signal. A schematic representation of the genesis of a MUAP is presented in Fig. 1. Note the many factors that influence the shape of the MUAP. Some of these are (1) the relative geometrical relationship of the detection surfaces of the electrode and the muscles fibers of the motor unit in its vicinity; (2) the relative position of the detection surfaces to the innervation zone, that is, the region where the nerve branches contact the muscle fibers; (3) the size of the muscle fibers (because the amplitude of the individual action potential is proportional to the diameter of the fiber); and (4) the number of muscle fibers of an individual motor unit in the detectable vicinity of the electrode.

The last two factors have particular importance in clinical applications. Considerable work has been performed to identify morphological modifications of the MUAP shape resulting

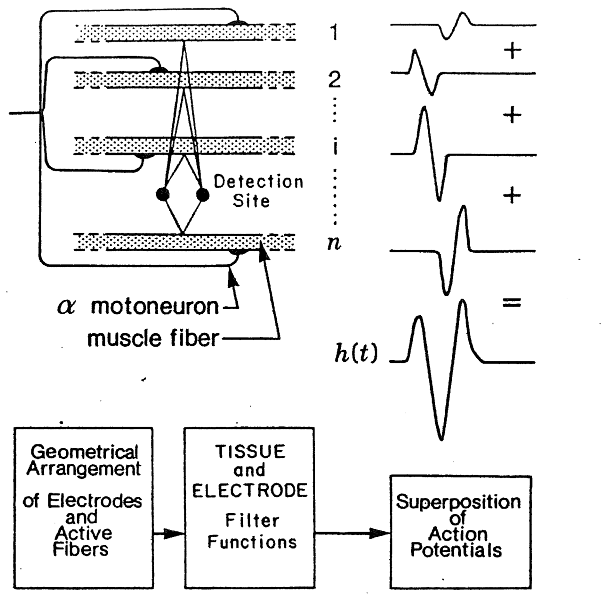

Figure 1. Schematic representation of the generation of the motor unit action potential.

from modifications in the morphology of the muscle fibers (such as hypertrophy and atrophy) or the motor unit (such as loss of muscle fibers and regeneration of motoneurons). Although usage of MUAP shape analysis is common practice among neurologists, interpretation of the results is not always straightforward and relies heavily on the experience and disposition of the observer.

Motor Unit Action Potential Train

The electrical manifestation of a MUAP is accompanied by a contractile twitch of the muscle fibers. To sustain a muscle contraction, the motor units must be activated repeatedly. The resulting sequence of MUAPs is called a motor unit action potential train (MUAPT). The waveform of the MUAPs within a MUAPT will remain constant if the geometric relationship between the electrode and the active muscle fibers remains constant, if the properties of the recording electrode do not change, and if there are no significant biochemical changes in the muscle tissue. Biochemical changes within the muscle can affect the conduction velocity of the muscle fiber and the filtering properties of the muscle tissue.

The MUAPT may be completely described by its interpulse intervals (the time between adjacent MUAPs) and the waveform of the MUAP. Mathematically the interpulse intervals may be expressed as a sequence of Dirac delta impulses $\delta_i(t)$ convoluted with a filter $h(t)$ that represents the shape of the MUAP. Figure 2 presents a graphic representation of a model for the MUAPT. It follows that the MUAPT, $u_i(t)$, can be expressed as

$$u_i(t) = \sum_{k=1}^{n} h_i(t - t_k)$$

where $t_k = \sum_{l=1}^{k} x_l$ for $k, l = 1, 2, 3, \ldots, n$.

In the above expressions, t is a real continuous random variable, t_k represents the time locations of the MUAPs, x represents the interpulse intervals, n is the total number of interpulse intervals in a MUAPT, and i, k, and l are integers that denote specific events.

By representing the interpulse intervals as a renewal process and restricting the MUAP shape so that it is invariant throughout the train, it is possible to derive the approximations

$$\text{Mean rectified value} = E\{|u_i(t, F)|\} \cong \lambda_i(t, F) \int_0^\infty |h_i(t)|dt$$

$$\text{Mean-squared value} = MS\{u_i(t, F)\} \cong \lambda_i(t, F) \int_0^\infty h_i^2(t)dt$$

where F is the force generated by the muscle and λ is the firing rate of the motor unit.

The power density spectrum of a MUAPT was derived from the above formulation by LeFever and De Luca (13) and independently by Lago and Jones (14). It can be expressed as

$$S_{u_i}(\omega, t, F) = S_{\delta_i}(\omega, t, F)|H_i(j\omega)|^2$$
$$= \frac{\lambda_i(t, F) \cdot \{1 - |M(j\omega, t, F)|^2\}}{1 - 2 \cdot \text{Real}\{M(j\omega, t, F)\} + |M(j\omega, t, F)|^2} \{|H_i(j\omega)|^2\}$$

for $\omega \neq 0$

where ω is the frequency in radians per second, $H_i(j\omega)$ is the Fourier transform of $h_i(t)$, and $M(j\omega, t, F)$ is the Fourier transform of the probability distribution function, $p_x(x, t, F)$ of the interpulse intervals.

Myoelectric Signal

The EMG signal may be synthesized by linearly summing the MUAPTs as they exist when they are detected by the electrode. This approach is expressed in the equation

$$m(t, F) = \sum_{i=1}^{p} u_i(t, F)$$

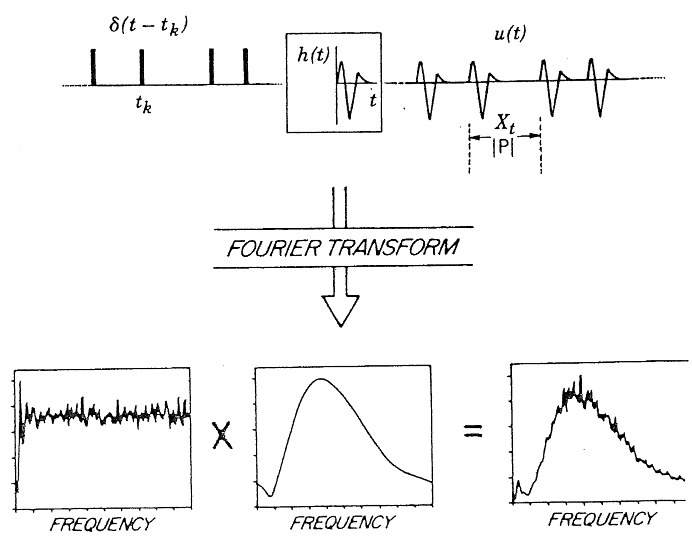

Figure 2. Model for a motor unit action potential train (MUAPT) an the corresponding Fourier transform of the interpulse intervals (IPIs the motor unit action potentials (MUAP), and the MUAPT.

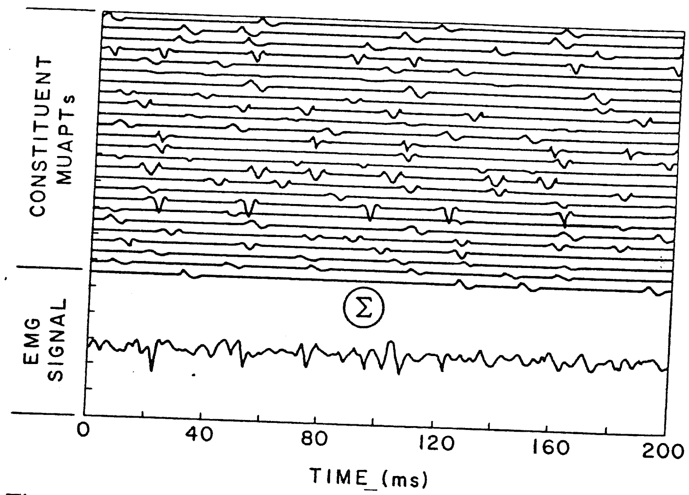

Figure 3. An EMG signal formed by adding (superimposing) 25 mathematically generated MUAPTs.

and is displayed in Fig. 3 where 25 mathematically generated MUAPTs were added to yield the signal at the bottom. This composite signal bears striking similarity to the real EMG signal.

From this concept it is possible to derive expressions for commonly used parameters: mean rectified value, root-mean-squared (rms) value, and variance of the rectified EMG signal. The interested reader is referred to *Muscles Alive* (5th ed.) by Basmajian and De Luca (12).

Continuing with the evolution of the model, it is possible to derive an expression for the power density spectrum of the EMG signal:

$$S_{\mathrm{m}}(\omega, t, F) = R(\omega, d)\left[\sum_{i=1}^{p(F)} S_{u_i}(\omega, t) + \sum_{\substack{i,j=1 \\ i \neq j}}^{q(F)} S_{u_i u_j}(\omega, t)\right]$$

where $R(\omega, d) = K \sin^2(\omega d/2v)$ is the bipolar electrode filter function; d is the distance between detection surfaces of the electrode; ω is the angular frequency; v is the conduction velocity along the muscle fibers; $S_{u_i}(\omega)$ is the power density of the MUAPT, $u_i(t)$; $S_{u_iu_j}(\omega)$ is the cross-power density spectrum MUAPTs $u_i(t)$ and $u_j(t)$; p is the total number of MUAPTs that constitute the signal; and q is the number of MUAPTs with correlated discharges.

Lindström (8), using a dipole model, arrived at another expression for the power density spectrum:

$$S_{\mathrm{m}}(\omega, t, F) = R(\omega, d)\left[\frac{1}{v^2(t, F)}\, G\left(\frac{\omega d}{2v(t, F)}\right)\right]$$

This representation explicitly denotes the interconnection between the spectrum of the EMG signal and the conduction velocity of the muscle fibers. Such a relationship is implicit in the previously presented modeling approach because any change in the conduction velocity would directly manifest itself in a change in the time duration of $h(t)$ as seen by the two detection surfaces of a stationary bipolar electrode.

ELECTRODES

Two main types of electrodes are used to detect the EMG signal: one is the surface (or skin) electrode and the other is the inserted (wire or needle) electrode. Electrodes are typically used singularly or in pairs. These configurations are referred to as monopolar and bipolar, respectively.

Surface Electrodes

Surface electrodes may be constructed as either passive or active units. In passive units, 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 by electronic means, rendering it less sensitive to the impedance (and therefore quality) of the electrode–skin interface.

The simplest form of passive electrode consists of silver disks that adhere to the skin. Electrical contact is greatly improved by introducing a saline gel or paste between the electrode and skin. The impedance can be further reduced by removing the dead surface layer of the skin along with its protective oils; this is best done by light abrasion of the skin.

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 amount of current flowing into the electrode. 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). This type of electrode has become highly popular in electromyography because of its light mass (250 mg), small size (11 mm diameter), and high reliability and durability. The dc component of the polarization potential is nullified by ac amplification when the electrodes are used in pairs. This point is elaborated upon in later sections of this article.

The active surface electrodes have been developed to eliminate the need for skin preparation and conducting medium. They are often referred to as "dry" or "pasteless" electrodes. These electrodes may be either resistively or capacitively coupled to the skin. Although the capacitively coupled electrodes have the advantage of not requiring a conductive medium, they have a higher inherent noise level. Also, these electrodes do not have long-term reliability because their dielectric properties are susceptible to change with the presence of perspiration 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 resistance is on the order of 10 TΩ and 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. Two examples of such electrodes are presented in Fig. 4. These electrodes were conceptualized and constructed at the NeuroMuscular Research Center under the sponsorship of Liberty Mutual Insurance Company. They each have two detection surfaces and associ-

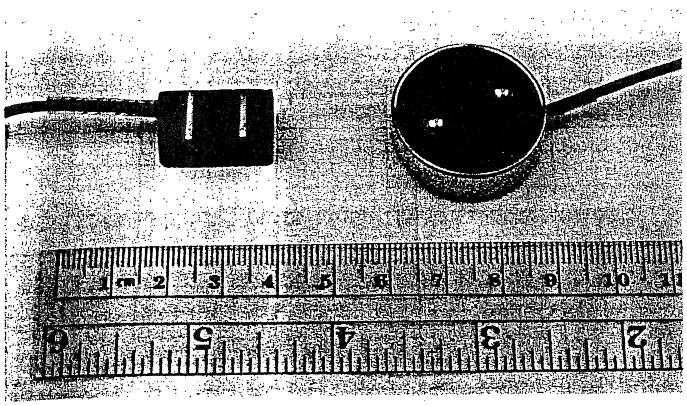

Figure 4. Active surface electrodes in bipolar configurations. The circular unit contains a ground ring around the perimeter of the electrode. These electrodes do not require any skin preparation or conductive paste or gels.

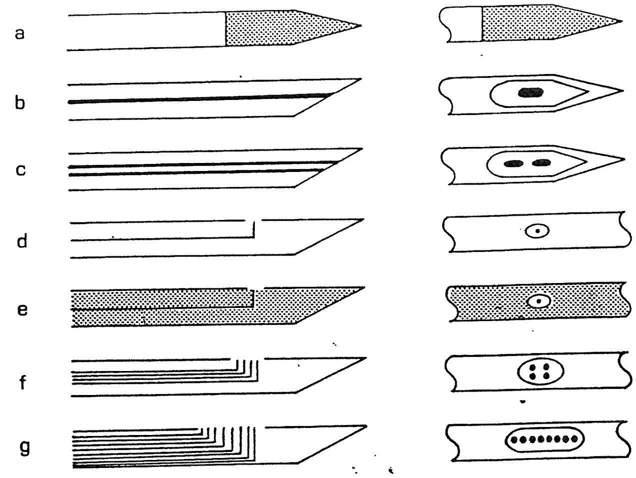

Figure 5. Examples of various needle electrodes: (a) Monopolar with a solid tip having 3 to 4 mm of exposed pickup length. If sufficiently thin, it can be inserted into a nerve bundle and detect neuroelectrical signals. (b) Concentric needle with one monopolar detection surface formed by the beveled cross section of centrally located wire typically 200 μm in diameter. Commonly used in clinical practice. (c) Bipolar needle electrode with two wires exposed in cross section, typically 100 μm in diameter. Used in clinical practice. (d) Single-fiber electrode with 25-μm-diameter wire. Used to detect the activity of individual muscle fibers. (e) Macroelectrode with 25-μm-diameter wire and with the cannula of the needle used as a detection surface. Used to detect the motor unit action potential from a large portion of the motor unit territory. (f) Quadrifilar planar electrode with four 50-μm wires located on the corners of a square 150 μm apart (center-to-center). Used for multiple-channel recordings and in EMG signal decomposition technique. (g) Multifilar electrode consisting of a row of wires. Used to study the motor unit territory.

ated electronic circuitry within their housing. The circular one contains a stainless-steel ring around its perimeter which serves as a ground.

The chief disadvantages of surface electrodes are that they can 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, for example, 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, for example, in the muscles of the lower limb during walking
7. In studies of children or other individuals who object to needle insertions.

Needle Electrodes

By far the most common indwelling electrode is the needle electrode. A wide variety are commercially available. The most common needle electrode is the "concentric" electrode now used widely by clinicians. This monopolar configuration contains one insulated wire in the cannula (Fig. 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 MUAPs during relatively low-force contractions. The other is that the electrodes may be conveniently repositioned within the muscle (after insertion) so that new tissue territories may be explored or the signal quality improved. These amenities have naturally led to the development of various specialized versions such as the multifilar electrode developed by Buchthal and Guld (15), the planar quadrifilar electrode of De Luca and Forrest (16), the single-fiber electrode of Ekstedt and Stålberg (17), and the macroelectrode of Stålberg (18). Examples of these electrodes may be seen in Fig. 5.

Wire Electrodes

Since the early 1960s, this type of electrode has been popularized by Basmajian and Stecko (19). Similar electrodes that differ only in minor details of construction were developed independently at about the same time by other researchers. Wire electrodes have proved a boon to kinesiological studies because they are extremely fine, and therefore painless, and easily implanted and withdrawn.

Wire electrodes may be made from any small-diameter, highly nonoxidizing, stiff wire with insulation. Alloys of platinum, silver, nickel, and chromium are preferable. Insulations

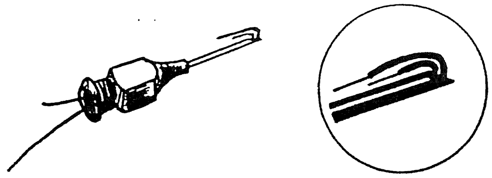

Figure 6. A bipolar wire electrode with its carrier needle used for insertion.

such as nylon, polyurethane, and Teflon® are conveniently available. The preferable alloy is 90% platinum–10% iridium; it offers the appropriate combination of chemical inertness, mechanical strength, and stiffness. The Teflon and nylon insulations are preferred because they add some mechanical rigidity to the wires, making them easier to handle. The electrode is constructed by inserting two insulated fine (25–100 μm in diameter) wires through the cannula of a hypodermic needle. Approximately 1–2 mm of the distal tips of the wire is deinsulated and bent to form two staggered hooks. (See Fig. 6 for completed version.) The electrode is introduced into the muscle by inserting the hypodermic needle and then withdrawing it. The wires remain lodged in the muscle tissues. They may be removed by gently pulling them out. They are so pliable that the hooks straighten out on traction.

In kinesiological studies where 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 have used the phrase "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.

lectrode Maintenance

'roper usage of wire and needle electrodes requires constant urveillance of the physical and electrical characteristics of ne electrode detection surfaces. Particular attention should e given to keeping the tips free of debris and oxidation. The eader is referred to the book *Muscles Alive* (5th ed.) by Basnajian and De Luca (12) for details on these procedures as well s suggestions for sterilization.

ow to Choose the Proper Electrode

ne specific type of electrode chosen to detect the EMG signal epends on the particular application and the convenience of e. The application refers to the information that is expected be obtained from the signal; for example, obtaining individl MUAPs or the gross EMG signal reflecting the activity of any muscle fibers. The convenience aspect refers to the time d effort the investigator wishes to devote to the disposition the subject or patient. Children, for example, are generally sistant to having needles inserted in their muscles.

The following electrode usage is recommended. The reader, wever, should keep in the mind that crossover applications e 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 of motor unit properties
  - Comfortable recording procedure from deep muscles

Where to Locate the Electrode

The location of the electrode should be determined by three important considerations: (1) signal-to-noise ratio, (2) signal stability (reliability), and (3) cross talk from adjacent muscles. 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 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 detects only signals from nearby muscle fibers. Because 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. Nonetheless, it is wise to steer clear of the innervation zone so as to reduce the probability of irritating a nerve ending.

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

For surface electrodes, the issue of cross talk must be considered. Obviously, it is foolish to optimize the signal detected, only to have the detected signal unacceptably contaminated by an unwanted source. A second consideration concerns the susceptibility of the signal to the architecture of the muscle. Both the innervation zone and the tendon–muscle tissue interface have been found to alter the characteristics of the signal. *It is suggested that the preferred location of an electrode is in the region halfway between the center of the innervation zone and the further tendon.*

SIGNAL DETECTION: PRACTICAL CONSIDERATIONS

When attempting to collect an EMG signal, both the novice and the expert should 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 that is generated by the membrane of the muscle fibers. The "distortion" of the signal as it progresses from the source to the elec-

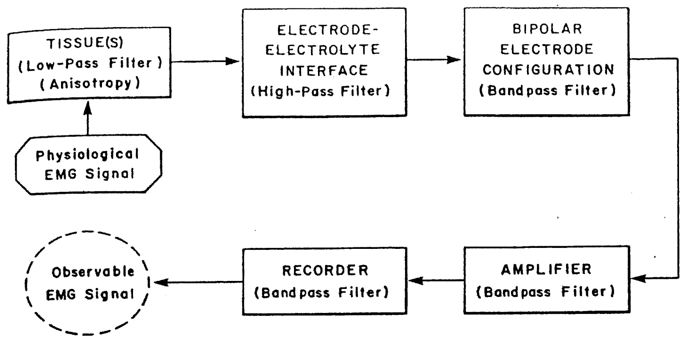

Figure 7. Block diagram of all the major aspects of the signal acquisition procedure. Note the variety of physical properties that act as filters to the EMG signal before it can be observed. The term "physiological EMG signal" refers to the collection of signals that emanate from the surface of the muscle fibers. These are not observable.

trode may be viewed as a filtering sequence. An overview of the major filtering effects is presented in Fig. 7. A brief summary of the pertinent facts follows. The reader interested in additional details is referred to *Muscles Alive* (5th ed.) by Basmajian and De Luca (12).

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 that either is electrically quiet or contains electrical signals unrelated to those being detected. ("Unrelated" means that the two signals have minimal physiological and anatomical associations.) 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 its relative technical simplicity. A schematic arrangement of the monopolar detection configuration may be seen in Fig. 8. The monopolar configuration has the drawback that it will detect all the electrical signals in the vicinity of the detection surface; this includes unwanted signals from sources other than the muscle of interest.

The bipolar detection configuration overcomes this limitation (see Fig. 8). In this case, two 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 because of the localized electrochemical events occurring in the 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) 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. Therefore, they will be subtracted but not necessarily nullified prior to being amplified. The measure of the ability of the differential amplifier to eliminate the common-mode signal is called the common-mode rejection ratio.

Tissue Filtering

1. As the signal propagates through the tissues, the amplitude decreases as a function of distance. The amplitude of the EMG signal decreases to approximately 25% within 100 μm. Thus, an indwelling electrode will detect only signals from nearby muscle fibers.
2. The filtering characteristic of the muscle tissues is a function of the 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 decrease as the distance increases.
3. The muscle tissue is 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. The contact layer between the metallic detection surface of the electrode and the conductive tissue forms an electrochemical junction which behaves as a high-pass filter.
2. The gain and bandwidth will be a function of the area of the detection surfaces and any chemical–electrical alteration of the junction.

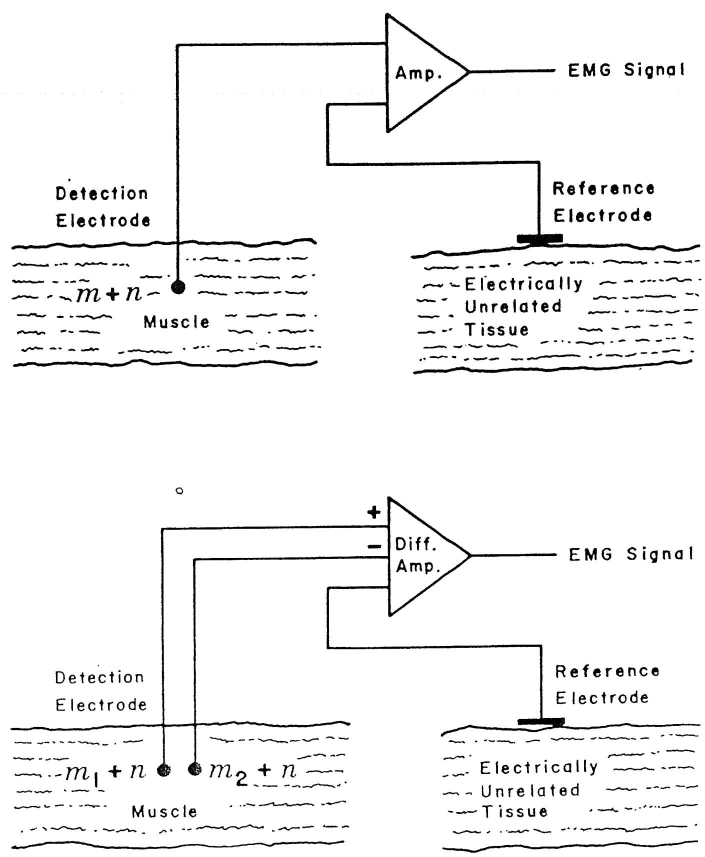

Figure 8. (Top) Monopolar detection arrangement. (Bottom) Bipolar detection arrangement. Note that in the bipolar detection arrangement, the EMG signals are considered to be different, whereas the noise is similar.

Bipolar Electrode Configuration

1. This configuration ideally behaves as a bandpass filter; however, this is true only if the inputs to the amplifier are balanced and 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.
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 that will minimally distort the EMG signal detected by the electrodes.
2. 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 should be located near (within 10 cm) the electrode.
3. Typical settings and characteristics:
   a. Gain: Such that it renders the output with an amplitude of approximately ± 1 V
   b. Input impedance >10 TΩ resistance in parallel with 5 pF capacitance
   c. Common-mode rejection ratio: >80 dB
   d. Input bias current: as low as possible (typically less than 50 pA)
   e. Noise < 5 μV rms
   f. Bandwidth in hertz (3-dB points for 12 dB/octave rolloff):

Surface electrodes	20–500
Wire electrodes	20–2000
Monopolar and bipolar needle electrodes for general use	20–10,000
Needle electrodes for signal decomposition	1000–10,000
Single-fiber electrode	1000–10,000
Macroelectrode	20–10,000

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. 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. In the use of 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.

SIGNAL ANALYSIS TECHNIQUES

The EMG signal is a time- and force (and possibly other parameters) dependent signal whose amplitude varies in a random nature above and below the zero value. Thus, simple averaging of the signal will not provide any useful information.

Rectification

A simple method that is commonly used to overcome the above restriction is to rectify the signal before performing mode pertinent analysis. The process of rectification involves the concept of rendering only positive deflections of the signal. This may be accomplished either by eliminating the negative values (half-wave rectification) or by inverting the negative values (full-wave rectification). The latter is the preferred procedure because it retains all the energy of the signal.

Averages or Means of Rectified Signals

The equivalent operation to smoothing in a digital sense is averaging. By taking the average of randomly varying values of a signal, the larger fluctuations are removed, thus achieving the same results as the analog smoothing operation. The mathematical expression for the average or mean of the rectified EMG signal is

$$\overline{|m(t)|}_{t_j - t_i} = \frac{1}{t_j - t_i} \int_{t_i}^{t_j} |m(t)|\, dt$$

where t_i and t_j are the points in time over which the integration and, hence, the averaging is performed. The shorter the time interval, the less smooth the averaged value will be.

The preceding expression will provide only one value over the time window $T = t_j - t_i$. To obtain the time-varying average of a complete record of a signal, it is necessary to move the time window T duration along the record. This operation is referred to as moving average.

$$\overline{|m(t)|} = \frac{1}{T} \int_{t}^{t+T} |m(t)|\, dt$$

Like the equivalent operation in the analog sense, this operation introduces a lag; that is, T time must pass before the value of the average of the T time interval can be obtained. In most cases, this outcome does not present a serious restriction, especially if the value of T is chosen wisely. For typical applications, values ranging from 100 to 200 ms are suggested. It should be noted that the smaller the time window T, the less smooth will be the time-dependent average (mean) of the rectified signal.

Integration

The most commonly used and abused data reduction procedure in electromyography is integration. The literature of the past three decades is swamped with improper usage of this term, although happily within the past decade it is possible to find increasing numbers of proper usage. When applied to a procedure for processing a signal, the term integration has a well-defined meaning which is expressed in a mathematical sense. It applies to a calculation that obtains the area under a signal or a curve. The units of this parameter are volt seconds (V·s). It is apparent that an observed EMG signal with an average value of zero will also have a total area (integrated value) of zero. Therefore, the concept of integration may be applied only to the rectified value of the EMG signal.

$$I\{|m(t)|\} = \int_t^{t+T} |m(t)|\, dt$$

Note that the operation is a subset of the procedure of obtaining the average rectified value. Since the rectified value is always positive, the integrated rectified value will increase continuously as a function of time. The only difference between the integrated rectified value and the average rectified value is that in the latter case the value is divided by T, the time over which the average is calculated. If a sufficiently long integration time T is chosen, the integrated rectified value will provide a smoothly varying measure of the signal as a function of time. There is no additional information in the integrated rectified value.

Root-Mean-Square (rms) Value

Mathematical derivations of the time- and force-dependent parameters indicate that the rms value provides more information than the previously described parameters (see Ref. 12). Its use in electromyography, however, has been sparse in the past. The recent increase is due possibly to the availability of analog chips that perform the rms operation and to the increased technical competence in electromyography. The rms value is obtained by performing the operations described by the term in reverse order; that is,

$$\mathrm{RMS}\{m(t)\} = \left(\frac{1}{T}\int_t^{t+T} m^2(t)\, dt\right)^{1/2}$$

This parameter is recommended above the others.

Zero Crossings and Turns Counting

This method consists of counting the number of times per unit time that the amplitude of the signal contains either a peak or crosses a zero value of the signal. It was popularized in electromyography by Willison (20). The relative ease with which these measurements could be obtained quickly made this technique popular among clinicians. Extensive clinical applications have been reported, some indicating that a discrimination may be made between myopathic and normal muscle; however, such distinctions are usually drawn on a statistical basis.

This technique is not recommended for measuring the behavior of the signal as a function of force (when recruitment or derecruitment of motor units occurs) or as a function of time during a sustained contraction. Lindström *et al.* (21) have shown that the relationship between the turns or zeros and the number of MUAPTs is linear for low-level contractions. But as the contraction level increases, the additionally recruited motor units contribute MUAPTs to the EMG signal. When the signal amplitude attains the character of Gaussian random noise, the linear proportionality no longer holds.

Frequency-Domain Analysis

Analysis of the EMG signal in the frequency domain involves measurements and parameters that describe specific aspects of the frequency spectrum of the signal. Fast Fourier transform techniques are commonly available and are convenient for obtaining the power density spectrum of the signal.

Three parameters of the power density spectrum may be conveniently used to provide useful measures of the spectrum. They are the median frequency, the mean frequency, and the bandwidth of the spectrum. Other parameters such as the mode frequency and ratios of segments of the power density spectrum have been used by some investigators, but are not considered reliable measures given the inevitably noisy nature of the spectrum. The median frequency and the mean frequency are defined by the equations:

$$\int_0^{f_{\mathrm{med}}} S_{\mathrm{m}}(f)\, df = \int_{f_{\mathrm{med}}}^{\infty} S_{\mathrm{m}}(f)\, df$$

$$f_{\mathrm{mean}} = \frac{\int_0^f f\, S_{\mathrm{m}}(f)\, df}{\int_0^f S_{\mathrm{m}}(f)\, df}$$

where $S_{\mathrm{m}}(f)$ is the power density spectrum of the EMG signal. Stulen and De Luca (22) performed a mathematical analysis to investigate the restrictions in estimating various parameters of the power density spectrum. The median and mean frequency parameters were found to be the most reliable, and of these two the median frequency was found to be less sensitive to noise. This quality is particularly useful when a signal is obtained during low-level contractions where the signal-to-noise ratio may be less than 6.

Much of the work presented here is adapted, with permission, from Ref. 12, Chap. 2, pp. 22–34, 36–38, 58–64 (excluding Fig. 2.19), and Chap. 3. The author thanks Williams & Wilkens for permission to extract this material, and Liberty Mutual Insurance Company for financial support.

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ELECTRONEUROGRAPHY. *See* ELECTROMYOGRAPHY.

ELECTRONEUROGRAPHY: NERVE CONDUCTION STUDIES

Electroneurography is the recording and study of action potential propagation along peripheral nerves. This procedure, commonly referred to simply as nerve conduction study, is an easily performed and reliable test of peripheral nerve function and is used routinely as part of the electrodiagnostic examination. The clinical value of these measurements has been demonstrated repeatedly in the examination of diseases and injuries which might otherwise be difficult to diagnose by other means. The localization of the site of a nerve lesion and the diagnosis of diabetic neuropathy, which affects primarily nerve fibers, are but two examples of the diagnostic usefulness of this technique.

The anatomic system of primary interest in nerve conduction studies consists essentially of the peripheral nerves, the myoneural junctions, and the skeletal muscles. The main function of the nervous system is to convey information from one part of the body to another, that is, from the cell body in the spinal cord to the nerve terminal at the muscle. The information propagated along nerve fibers is in the form of short bursts of electrical energy or action potentials. To fully understand and appreciate the procedure for measuring conduction along nerves, it is necessary to review briefly some basic neurophysiology.

A REVIEW OF BASIC NEUROPHYSIOLOGY

Anatomy

Motor nerve fibers innervate striated voluntary muscles, except those in the head, by axons of cells originating in the anterior gray matter of the spinal cord (Fig. 1). Within a muscle, the axon from a single motor nerve cell arborizes into many terminal branches. Each branch is attached to the midpoint of a single muscle fiber, called the myoneural junction. Branching permits a single axon to stimulate a group of muscle fibers which forms a part of the motor unit. The motor unit is the functional unit of the neuromuscular system and consists of the anterior horn cell, its axon, and all of the muscle fibers innervated by that axon. The number of muscle fibers in a single motor unit varies widely for the different skeletal muscles.

Sensory nerve fibers arise from nerve endings situated in a variety of somatic structures, such as cutaneous and subcutaneous tissues, tendons, and skeletal muscle. Signals from these receptors are sent back to the central nervous system (the dorsal horn of the spinal cord) via the sensory nerve fibers.

Nerve Fiber Types

The peripheral nerve contains sensory and motor fibers of various diameters, conduction velocities, action potential configurations, and refractory periods. The fibers can be classified into different types known as A, B, and C fibers. The A fibers are large myelinated fibers that innervate skeletal muscle and also conduct sensory impulses from the proprioceptive receptors in skeletal muscles as well as receptors in the skin. The B group consists of small myelinated, sensory, preganglionic fibers found only in the autonomic nerves. The C fibers are unmyelinated pre- and postganglionic sympathetic fibers, also found as sensories mediating various modalities of sensation (deep pain).

A typical peripheral nerve such as the sciatic nerve contains both A and C fibers. The individual fibers are not of equal size, but cover a wide range of diameters. The A fibers constitute a group of which the most prominent variable is the diameter. In mammalian nerves, the diameter of A fibers can vary from 1 to 22 μm.

The nerve fibers can be subdivided in terms of their conduction velocity as well as their diameters. Because the internal longitudinal resistance is inversely proportional to the diameter, the large-diameter fibers conduct at a greater velocity than the small-diameter fibers. For the A fibers, the conduction velocities have been found to be linearly related to fiber diameter.

Transmembrane Potentials

Surrounding each nerve fiber is a membrane that regulates the interchange of substances between the cell interior and its environment. The nature of the membrane is such that it imposes a restriction on the movement of some ions: The membrane is impermeable to the organic anions within the fiber. Also, certain ions pass freely through the membrane, whereas the diffusion of others is severely limited: The membrane is more permeable to K^+ than it is to Na^+. In addition, the membrane has a mechanism to actively transport Na^+ from inside to outside the fiber and K^+ in the opposite direction (the so-