10

KINESIOLOGICAL ELECTROMYOGRAPHY

10.0 INTRODUCTION

The electrical signal associated with the contraction of a muscle is called an *electromyogram*, or EMG. The study of EMGs, called *electromyography*, has revealed some basic information; however, much remains to be learned. Voluntary muscular activity results in an EMG that increases in magnitude with the tension. However, there are many variables that can influence the signal at any given time: velocity of shortening or lengthening of the muscle, rate of tension build-up, fatigue, and reflex activity. An understanding of the electrophysiology and the technology of recording is essential to the appreciation of the biomechanical relationships that follow.

10.1 ELECTROPHYSIOLOGY OF MUSCLE CONTRACTION

It is important to realize that muscle tissue conducts electrical potentials somewhat similarly to the way axons transmit action potentials. The name given to this special electrical signal generated in the muscle fibers as a result of the recruitment of a motor unit is a *motor unit action potential* (m.u.a.p.). Electrodes placed on the surface of a muscle or inside the muscle tissue (indwelling electrodes) will record the algebraic sum of all m.u.a.p.'s being transmitted along the muscle fibers at that point in time. Those motor units far away from the electrode site will result in a smaller m.u.a.p. than those of similar size near the electrode.

 ISBN: 978-0-470-39818-0

10.1.1 Motor End Plate

For a given muscle, there can be a variable number of motor units, each controlled by a motor neuron through special synaptic junctions called *motor end plates*. An action potential transmitted down the motoneuron (sometimes called the *final common pathway*) arrives at the motor end plate and triggers a sequence of electrochemical events. A quantum of ACh is released; it crosses the synaptic gap (200–500 Å wide) and causes a depolarization of the postsynaptic membrane. Such a depolarization can be recorded by a suitable microelectrode and is called an *end plate potential* (EPP). In normal circumstances, the EPP is large enough to reach a threshold level, and an action potential is initiated in the adjacent muscle fiber membrane. In disorders of neuromuscular transmission (e.g., depletion of ACh), there may not be a one-to-one relationship between each motor nerve action potential and a m.u.a.p. The end plate block may be complete, or it may occur only at high stimulation rates or intermittently.

10.1.2 Sequence of Chemical Events Leading to a Twitch

The beginning of the m.u.a.p. starts at the Z line of the contractile element (see Section 9.0.1) by means of an inward spread of the stimulus along the transverse tubular system. This results in a release of Ca^{2+} in the sarcoplasmic reticulum. Ca^{2+} rapidly diffuses to the contractile filaments of actin and myosin, where ATP is hydrolyzed to produce ADP plus heat plus mechanical energy (tension). The mechanical energy manifests itself as an impulsive force at the cross-bridges of the contractile element. The time course of the contractile element's force has been the subject of considerable speculation and has been modeled mathematically, as described in Section 9.0.5.

10.1.3 Generation of a Muscle Action Potential

The depolarization of the transverse tubular system and the sarcoplasmic reticulum results in a depolarization "wave" along the direction of the muscle fibers. It is this depolarization wave front and the subsequent repolarization wave that are "seen" by the recording electrodes.

Many types of EMG electrodes have developed over the years, but generally they can be divided into two groups: surface and indwelling (intramuscular). Basmajian (1973) gives a detailed review of the use of different types along with their connectors. Surface electrodes consist of disks of metal, usually silver/silver chloride, of about 1 cm in diameter. These electrodes detect the average activity of superficial muscles and give more reproducible results than do indwelling types (Komi and Buskirk, 1970; Kadaba et al., 1985). Smaller disks can be used for smaller muscles. Indwelling electrodes are required, however, for the assessment of fine movements or to record from deep muscles. A needle electrode is nothing more than a fine hypodermic

needle with an insulated conductor located inside and bared to the muscle tissue at the open end of the needle; the needle itself forms the other conductor. For research purposes, multielectrode types have been developed to investigate the "territory" of a motor unit (Buchthal et al., 1959), which has been found to vary from 2 to 15 mm in diameter. Fine-wire electrodes, which have a diameter about that of human hairs, are now widely used. They require a hypodermic needle to insert. After removal of the needle, the fine wires with their uninsulated tips remain inside in contact with the muscle tissue. A comparison of this experimental investigation of motor unit territory was seen to agree with theoretical predictions (Boyd et al., 1978).

Indwelling electrodes are influenced not only by waves that actually pass by their conducting surfaces but also by waves that pass within a few millimeters of the bare conductor. The same is true for surface electrodes. The field equations that describe the electrode potential were originally derived by Lorente de No (1947) and were extended in rigorous formulations by Plonsey (1964, 1974) and Rosenfalck (1969). These equationswere complicated by the formulation of current density functions to describe the temporal and spatial depolarization and repolarization of the muscle membrane. Thus, simplification of the current density to a dipole or tripole (Rosenfalck, 1969) yielded a reasonable approximation when the active fiber was more than 1 mm from the electrode surface (Andreassen and Rosenfalck, 1981).

In the dipole model (see Figure 10.1), the current is assumed to be concentrated at two points along the fiber: a source of current, I, representing

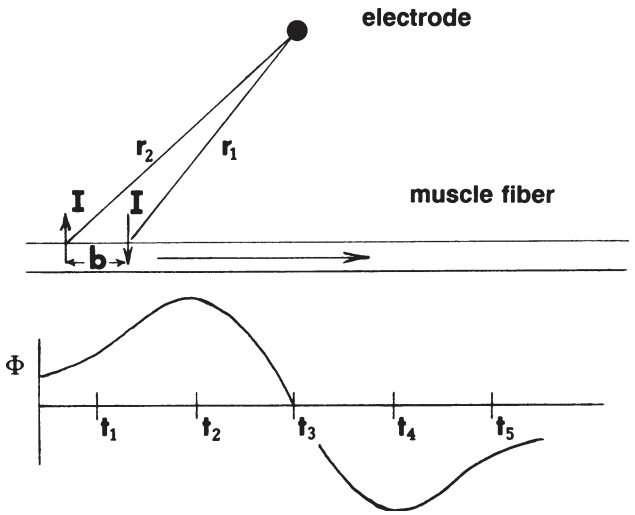


Figure 10.1 Propagation of motor unit action potential wave front as it passes beneath a recording electrode on the skin surface. The electrode voltage is a function of the magnitude of the dipole and the distances r_1 and r_2 from the electrode to the depolarizing and repolarizing currents.

the depolarization, and a sink of current, $-I$, representing the repolarization, both separated by a distance b. The potential $\mathbf{\Phi}$, recorded at a point electrode at a distance r from the current source, is given by:

$$\mathbf{\Phi} = \frac{I}{4\pi\sigma} \cdot \frac{1}{r} \tag{10.1}$$

where σ is the conductivity of the medium, which is assumed to be isotropic (uniform in all spatial directions).

The net potential recorded at the point electrode from both source and sink currents is:

$$\begin{aligned} \mathbf{\Phi} &= \frac{I}{4\pi\sigma} \cdot \frac{1}{r_1} - \frac{I}{4\pi\sigma} \cdot \frac{1}{r_2} \\ &= \frac{I}{4\pi\sigma} \left(\frac{1}{r_1} - \frac{1}{r_2} \right) \end{aligned} \tag{10.2}$$

where r_1 and r_2 are the distances to the source and sink currents.

The time history of the action potential then depends on r_1 and r_2, which vary with time as the wave propagates along the muscle fiber. At t_1, as the wave approaches the electrode ($r_1 < r_2$), the potential will thus be positive and will increase. It will reach a maximum at t_2; then as r_1 becomes nearly equal to r_2, the amplitude suddenly decreases and passes through zero at t_3 when the dipole is directly beneath the electrode ($r_1 = r_2$). Then it becomes negative as the dipole propagates away from the electrode ($r_1 > r_2$). Thus, a biphasic wave is recorded by a single electrode.

A number of recording and biological factors affect the magnitude and shape of the biphasic signal. The duration of each phase is a function of the propagation velocity, the distance b (which varies from 0.5 to 2.0 mm) between source and sink, the depth of the fiber, and the electrode surface area. Equation (10.2) assumes a point electrode. Typical surface electrodes are not point electrodes but have a finite surface area, and each point on the surface can be considered an area of point sources; the potential on the surface is the average of all point-source potentials. Figure 10.2 is presented from a study by Fuglevand et al. (1992) to demonstrate the influence of the electrode size and shape on the electrode potential. Two different orientations of strip electrodes are shown: A—along an axis parallel to the muscle fiber—and D—at right angles to the muscle fiber. B and E show the location of a strip of point-source electrodes to represent these two strip electrodes, each 10 mm long with 10 point sources; C and F show the action potentials at each point source. For the strip electrode in A, the distance from the points on the electrode to the fiber, r_f, is constant; thus, each action potential has the same amplitude and shape. However, there is a phase difference between each of the point-source action potentials such that the average action potential is lower and longer than the individual action potentials. For the strip electrode in D, the distance r_f is

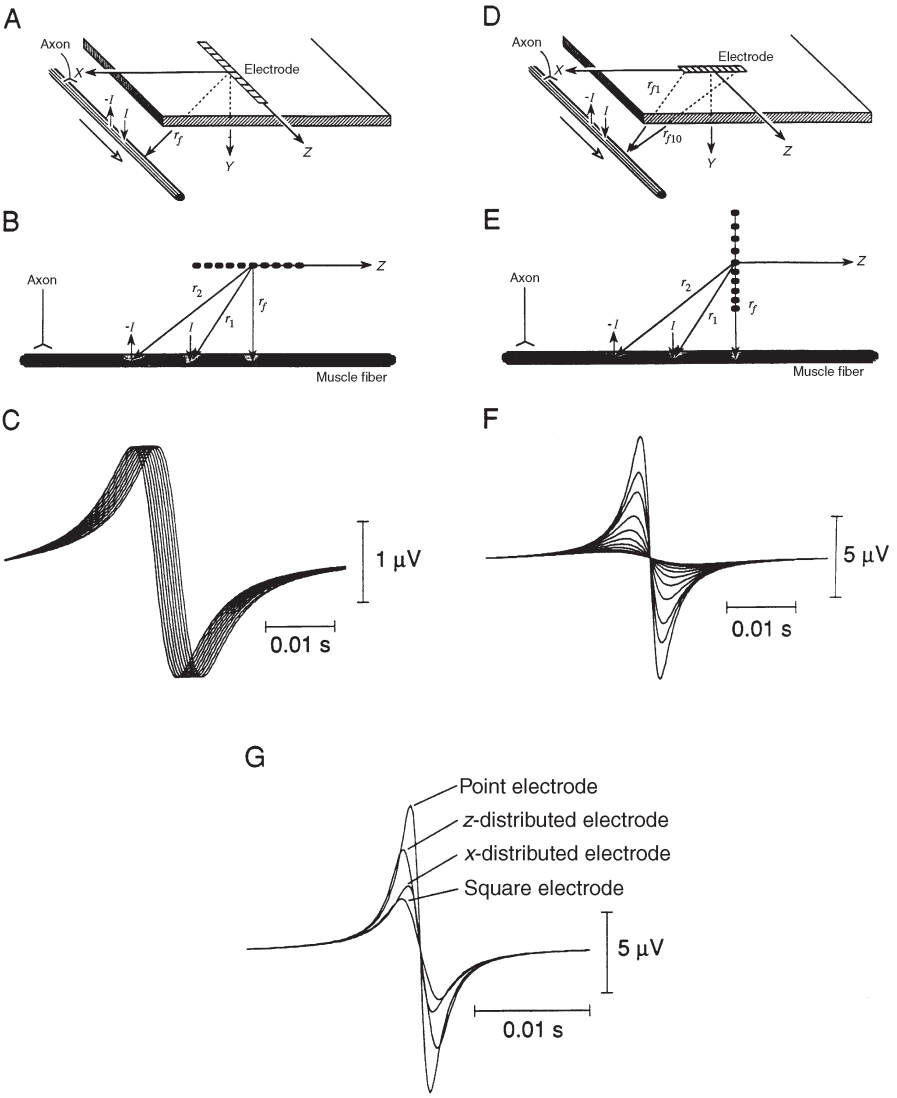


Figure 10.2 Influence of electrode geometry on the predicted action potentials from a single electrode with a variety of shapes and sizes. The potential was computed as the algebraic mean of the potentials detected by an array of point-source electrodes. See the text for detailed discussion. (Reproduced by permission from *Biological Cybernetics*, "Detection of Motor Unit Action Potentials with Surface Electrodes: Influence of Electrode Size and Shaping," A. F. Fuglevand, D. A. Winter, A. E. Patla, and D. Stashuk, 67: 143–153, 1992; Fig. 3A–G. With kind permission of Springer Science + Business Media.)

different for every point source; thus, each action potential decreases as r_f increases, as is seen in F. The phase of each point-source action potential has the same zero-crossing time, but the smaller action potentials from the further point sources are slightly longer in duration. Finally, G compares the net action potentials from a point source, strip A, strip D, and a square electrode $10 \times 10\,\text{mm}^2$. As is evident from these action potentials, it is desirable to use electrodes with as small a surface area as possible. Also, as will be seen in Section 10.2.5, the smaller surface area electrodes are less influenced by cross-talk.

The zone of pick-up for individual fibers by an electrode depends on two physiological variables, I and r. Larger muscle fibers have larger dipole currents, which increase with the diameter of the fiber. For an m.u.a.p., the number of fibers innervated must also be considered. The potential detected from the activation of a motor unit will be equivalent to the sum of the constituent fiber potentials. Therefore, the pick-up zone for a motor unit that innervates a large number of fibers will be greater than that for a unit with fewer fibers. An estimate of the detectable pick-up distance for small motor units (50 fibers) is about 0.5 cm and, for the largest motor units (2500 fibers), about 1.5 cm (Fuglevand et al., 1992). Thus, surface electrodes are limited to detecting m.u.a.p.'s from those fibers quite close to the electrode site and are not prone to pick up from adjacent muscles (called *cross-talk*) unless the muscle being recorded from is very small.

Most EMGs require two electrodes over the muscle site, so the voltage waveform that is recorded is the difference in potential between the two electrodes. Figure 10.3 shows that the voltage waveform at each electrode is almost the same but is shifted slightly in time. Thus, a biphasic potential waveform at one electrode usually results in a difference signal that has three phases. The closer the spacing between the two electrodes, the more the difference signal looks like a time differentiation of that signal recorded at a single electrode (Kadefors, 1973). Thus, closely spaced electrodes result in the spectrum of the EMG having higher-frequency components than those recorded from widely spaced electrodes.

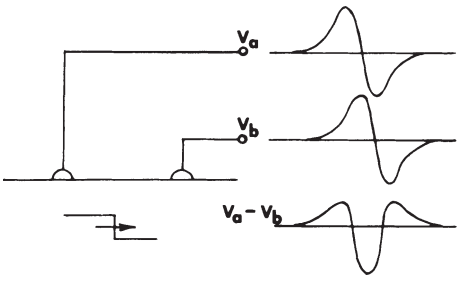


Figure 10.3 Voltage waveform present at two electrodes because of a single propagating wave. Voltage recorded is the difference voltage $V_a - V_b$, which is triphasic in comparison with the biphasic waveform seen at a single electrode.

10.1.4 Duration of the Motor Unit Action Potential

As indicated in the previous section, the larger the surface area, the longer the duration of the m.u.a.p. Thus, surface electrodes automatically record longer duration m.u.a.p.'s than indwelling electrodes (Kadefors, 1973; Basmajian, 1973). Needle electrodes record durations of 3–20 ms, while surface electrodes record durations about twice that long. However, for a given set of electrodes, the duration of the m.u.a.p.'s is a function of the velocity of the propagating wave front and the depth of the motor unit below the electrode's surface. The velocity of propagation of the m.u.a.p. in normals has been found to be about 4 m/s (Buchthal et al., 1955). The faster the velocity, the shorter the duration of the m.u.a.p. Such a relationship has been used to a limited extent in the detection of velocity changes. In fatigue and in certain myopathies (muscle pathologies), the average velocity of the m.u.a.p.'s that are recruited is reduced; thus, the duration of the m.u.a.p.'s increases (Johansson et al., 1970; Gersten et al., 1965; Kadefors, 1973). Because the peak amplitude of each phase of the m.u.a.p. remains the same, the area under each phase will increase. Thus, when we measure the average amplitude of the EMG (from the full-wave rectified waveform), it will appear to increase (Fuglevand, et al., 1992). During voluntary contractions, it has been possible, under special laboratory conditions (Milner-Brown and Stein, 1975), to detect the duration and amplitude of muscle action potentials directly from the EMG. However, during unconstrained movements, a computer analysis of the total EMG would be necessary to detect a shift in the frequency spectrum (Kwatny et al., 1970), or an autocorrelation analysis can yield the average m.u.a.p. duration (Person and Mishin, 1964).

The distance between the motor unit and the surface of the electrode severely influences the amplitude of the action potential, as is predicted in Equation (10.2), and also influences the duration of the action potential. Figure 10.4, taken from Fuglevand et al. (1992), predicts the duration and amplitude of the action potential from a 50-fiber motor unit with electrode-unit distances out to 20 mm. It is evident that the duration of the action potential increases as the amplitude decreases with distance. Thus, the frequency content of the m.u.a.p.'s from the more distant sites decreases. Fuglevand et al. (1992) showed that the mean power frequency of the m.u.a.p. would decrease from 160 Hz at an electrode-motor unit distance of 1 mm to 25 Hz at an electrode-motor unit distance of 20 mm.

10.1.5 Detection of Motor Unit Action Potentials from Electromyogram during Graded Contractions

Using recordings from multiple indwelling electrodes, DeLuca et al. (1982) pioneered techniques to identify individual m.u.a.p.'s during a low-level graded contraction. More recent techniques use a quadrifilar needle electrode recording of three channels followed by decomposition algorithms involving

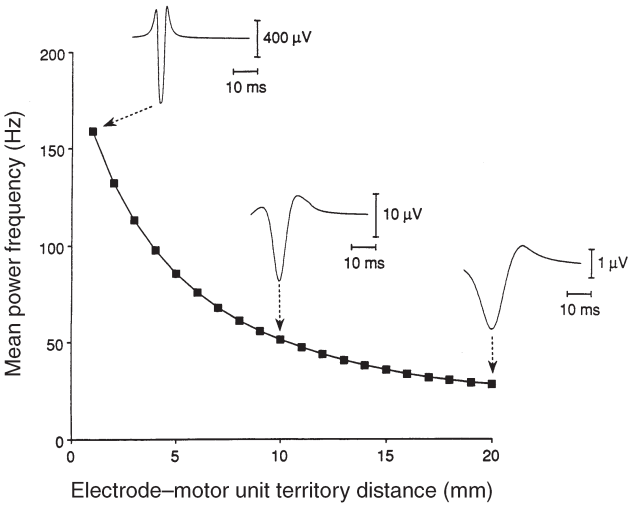


Figure 10.4 Variation in the amplitude and frequency content of the motor unit action potential with increased electrode-motor unit distance. Predictions were for a small motor unit (50 fibers) with $4 - \text{mm}^2$ bipolar electrodes with an interelectrode spacing of 11 mm. Shown are the m.u.a.p.'s at 1-, 10-, and 20-mm distances. (Reproduced by permission from *Biological Cybernetics*, "Detection of Motor Unit Action Potentials with Surface Electrodes: Influence of Electrode Size and Shaping," A. F. Fuglevand, D. A. Winter, A. E. Patla, and D. Stashuk, 67:143–153, 1992; Fig. 8. With kind permission of Springer Science + Business Media.)

template matching, template updating, and motor unit firing statistics (De Luca, 1993). As many as four motor units were detected and their firing rates tracked from 0 to 100% of maximum voluntary contraction (MVC) of the tibialis anterior muscle (Erim et al., 1996). It should be noted that only a fraction of all active motor units can be tracked—only those within the pick-up region of the electrode array. EMG recordings from surface electrodes have been decomposed in order to determine the firing profile of those motor units detected (McGill et al., 1987).

10.2 RECORDING OF THE ELECTROMYOGRAM

A biological amplifier of certain specifications is required for the recording of the EMG, whether from surface or from indwelling electrodes. It is valuable to discuss the reasons behind these specifications with respect to considerable problems in getting a "clean" EMG signal. Such a signal is the summation of m.u.a.ps and should be undistorted and free of noise or artifacts. Undistorted means that the signal has been amplified linearly over the range of the amplifier and recording system. The larger signals (up to 5 mV) have been amplified as much as the smaller signals (100 μV and

below). The most common distortion is overdriving of the amplifier system such that the larger signals appear to be clipped off. Every amplifier has a dynamic range and should be such that the largest EMG signal expected will not exceed that range. Noise can be introduced from sources other than the muscle, and can be biological in origin or man-made. An electrocardiogram (ECG) signal picked up by EMG electrodes on the thoracic muscles can be considered unwanted biological noise. Man-made noise usually comes from power lines (hum) or from machinery, or is generated within the components of the amplifier. Artifacts generally refer to false signals generated by the electrodes themselves or the cabling system. Anyone familiar with EMG recording will recall the lower-frequency baseline jumps, called *movement artifacts*, which result from touching of the electrodes and movement of the cables.

The major considerations to be made when specifying the EMG amplifier are:

1. Gain and dynamic range
2. Input impedance
3. Frequency response
4. Common-mode rejection

10.2.1 Amplifier Gain

Surface EMGs have a maximum amplitude of 5 mV peak to peak, as recorded during a MVC. Indwelling electrodes can have a larger amplitude, up to 10 mV. A single m.u.a.p. has an amplitude of about 100 μV. The noise level of the amplifier is the amplitude of the higher-frequency random signal seen when the electrodes are shorted together, and should not exceed 50 μV, preferably 20 μV. The gain of the amplifier is defined as the ratio of the output voltage to the input voltage. For a 2-mV input and a gain of 1000, the output will be 2 V. The exact gain chosen for any given situation will depend on what is to be done with the output signal. The EMG can be recorded on a pen recorder or magnetic tape, viewed on an oscilloscope, or fed straight into a computer. In each case, the amplified EMG should not exceed the range of input signals expected by this recording equipment. Fortunately, most of these recording systems have internal amplifiers that can be adjusted to accommodate a wide range of input signals. In general, a good bioamplifier should have a range of gains selectable from 100 to 10,000. Independent of the amplifier gain, the amplitude of the signal should be reported as it appears at the electrodes, in millivolts.

10.2.2 Input Impedance

The input impedance (resistance) of a biological amplifier must be sufficiently high so as not to attenuate the EMG signal as it is connected to the input

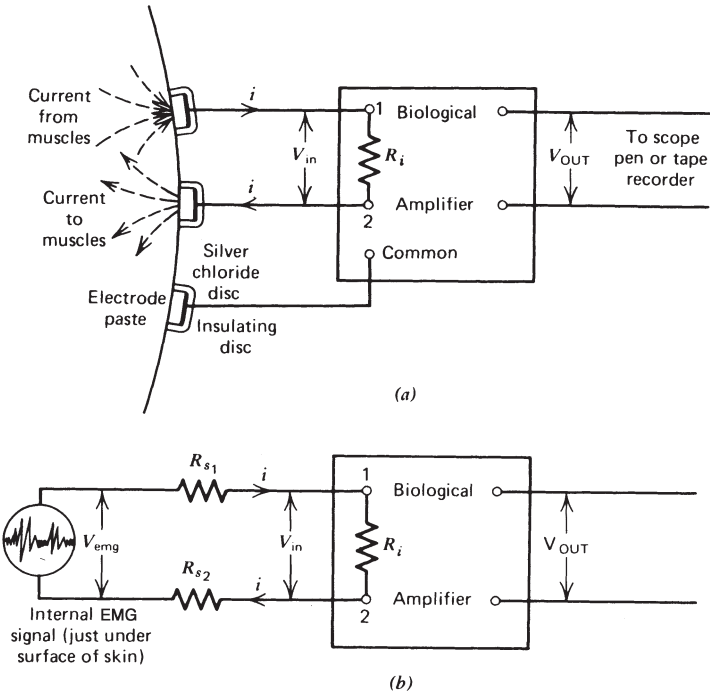


Figure 10.5 Biological amplifier for recording electrode potentials. *(a)* Current resulting from muscle action potentials flows across skin—electrode interface to develop a voltage V_{in} at the input terminals of the amplifier. A third, common electrode is normally required because the amplifier is a differential amplifier. *(b)* Equivalent circuit showing electrodes replaced by series resistors R_{s1} and R_{s2}. V_{in} will be nearly equal to V_{EMG} if $R_i >> R_s$.

terminals of the amplifier. Consider the amplifier represented in Figure 10.5*a*. The active input terminals are 1 and 2, with a common terminal *c*. The need for a three-input terminal amplifier (differential amplifier) is explained in Section 10.2.4.

Each electrode-skin interface has a finite impedance that depends on many factors: thickness of the skin layer, cleaning of the skin prior to the attachment of the electrodes, area of the electrode surface, and temperature of the electrode paste (it warms up from room temperature after attachment). Indwelling electrodes have a higher impedance because of the small surface area of bare wire that is in contact with the muscle tissue.

In Figure 10.5*b*, the electrode-skin interface has been replaced with an equivalent resistance. This is a simplification of the actual situation. A correct representation is a more complex impedance to include the capacitance effect between the electrode and the skin. As soon as the amplifier is connected to the electrodes, the minute EMG signal will cause current to flow through the electrode resistances R_{s1} and R_{s2} to the input impedance of the amplifier R_i.

The current flow through the electrode resistances will cause a voltage drop so that the voltage at the input terminals V_{in} will be less than the desired signal V_{EMG}. For example, if $R_{s1} = R_{s2} = 10{,}000\,\Omega$ and $R_i = 80{,}000\,\Omega$, a 2-mV EMG signal will be reduced to 1.6 mV at V in. A voltage loss of 0.2 mV occurs across each of the electrodes. If R_{s1} and R_{s2} were decreased by better skin preparation to $1000\,\Omega$, and R_i were increased to $1\,\text{M}\,\Omega$, the 2-mV EMG signal would be reduced only slightly, to 1.998 mV. Thus, it is desirable to have input impedances of $1\,\text{M}\,\Omega$ or higher, and to prepare the skin to reduce the impedance to $1000\,\Omega$ or less. For indwelling electrodes, the electrode impedance can be as high as $50{,}000\,\Omega$, so an amplifier with at least $5\,\text{M}\,\Omega$ input impedance should be used.

10.2.3 Frequency Response

The frequency bandwidth of an EMG amplifier should be such as to amplify, without attenuation, all frequencies present in the EMG. The bandwidth of any amplifier, as shown in Figure 10.6, is the difference between the upper cutoff frequency f_2 and the lower cutoff frequency f_1. The gain of the amplifier at these cutoff frequencies is 0.707 of the gain in the midfrequency region. If we express the midfrequency gain as 100%, the gain at the cutoff frequencies has dropped to 70.7%, or the power has dropped to $(0.707)^2 = 0.5$. These are also referred to as the half-power points. Often amplifier gain is expressed in logarithmic form and expressed in decibels,

$$\text{gain (dB)} = 20 \log_{10}(\text{linear gain}) \tag{10.3}$$

If the linear gain were 1000, the gain in decibels would be 60, and the gain at the cutoff frequency would be 57 dB (3 dB less than that at midfrequency).

In a high-fidelity amplifier used for music reproduction, f_1 and f_2 are designed to accommodate the range of human hearing, from 50 to 20,000 Hz. All frequencies present in the music will be amplified equally, producing a

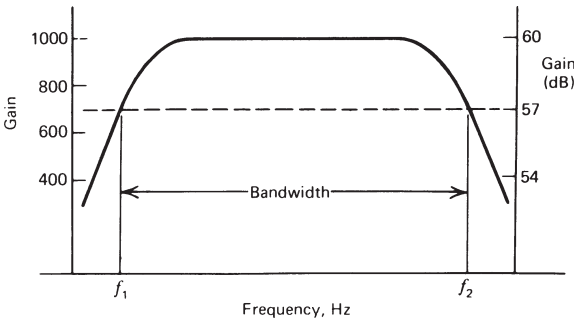


Figure 10.6 Frequency response of the biological amplifier showing a gain of 1000 (60 dB), and lower and upper cutoff frequencies f_1 and f_2.

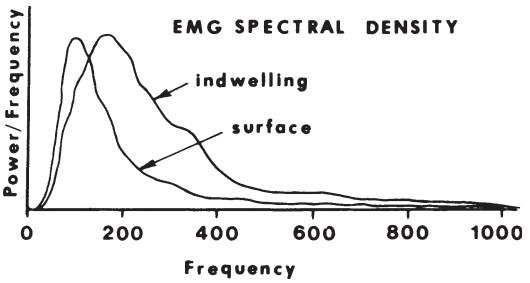


Figure 10.7 Frequency spectrum of EMG as recorded via surface and indwelling electrodes. The higher-frequency content of indwelling electodes is the result of the closer spacing between electrodes and their closer proximity to the active muscle fibers.

true undistorted sound at the loudspeakers. Similarly, the EMG should have all its frequencies amplified equally. The spectrum of the EMG has been widely reported in the literature, with a range from 5 Hz at the low end to 2000 Hz at the upper end. For surface electrodes, the m.u.a.p.'s are longer in duration and, thus, have negligible power beyond 1000 Hz. A recommended range for surface EMG is 10–1000 Hz, and 20–2000 Hz for indwelling electrodes. If computer pattern recognition of individual m.u.a.p.'s is being done, the upper cutoff frequencies should be increased to 5 and 10 kHz, respectively. Figure 10.7 shows a typical EMG spectrum, and it can be seen that most of the signal is concentrated in the band between 20 and 200 Hz, with a lesser component extending up to 1000 Hz.

The spectra of other physiological and noise signals must also be considered. ECG signals contain power out to 100 Hz, so it may not be possible to eliminate such interference, especially when monitoring muscle activity around the thoracic region. The major interference comes from power line hum: in North America it is 60 Hz, in Europe 50 Hz. Unfortunately, hum lies right in the middle of the EMG spectrum, so nothing can be done to filter it out. Movement artifacts, fortunately, lie in the 0–10 Hz range and normally should not cause problems. Unfortunately some of the lower-quality cabling systems can generate large low-frequency artifacts that can seriously interfere with the baseline of the EMG recording. Usually such artifacts can be eliminated by good low-frequency filtering, by setting f_1 to about 20 Hz. If this fails, the only solution is to replace the cabling or go to the expense of using microamplifiers right at the skin surface.

It is valuable to see the EMG signal as it is filtered using a wide range of bandwidths. Figure 10.8 shows the results of such filtering and the obvious distortion of the signal when f_1 and f_2 are not set properly.

10.2.4 Common-Mode Rejection

The human body is a good conductor and, therefore, will act as an antenna to pick up any electromagnetic radiation that is present. The most common

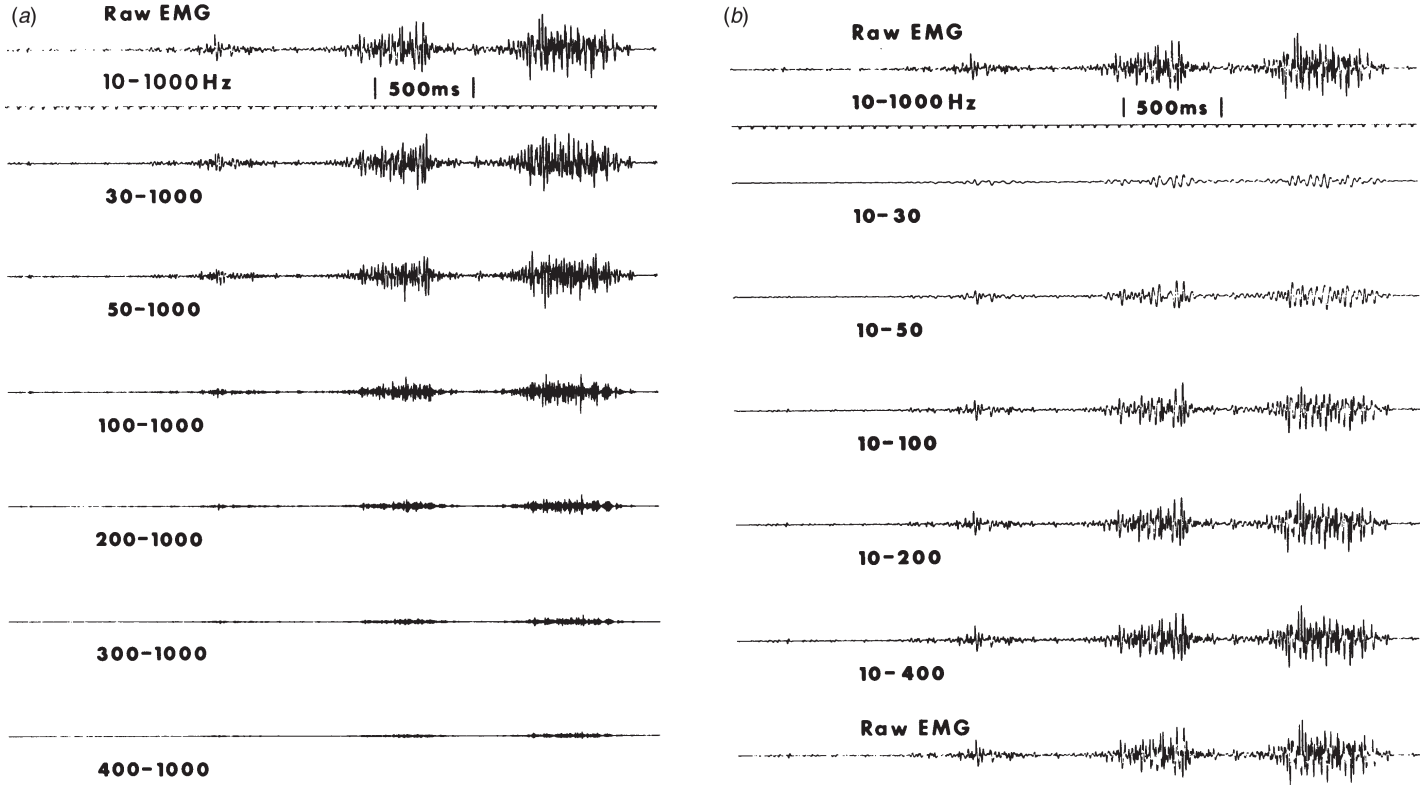


Figure 10.8 Surface EMG filtered with varying cutoff frequencies. (*a*) With lower cutoff frequency varied from 30 to 400 Hz, showing the effect of rejecting lower frequencies. (*b*) With upper cutoff frequency varied from 30 to 400 Hz, showing the loss of higher frequencies.

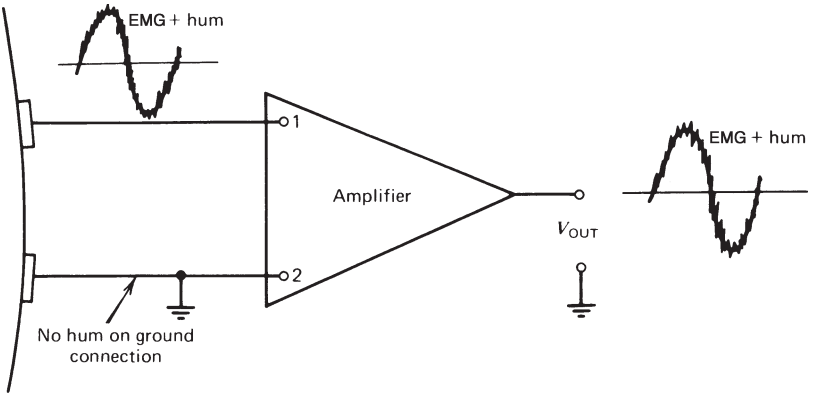


Figure 10.9 Single-ended amplifier showing lack of rejection of hum present on ungrounded active terminal.

radiation comes from domestic power: power cords, fluorescent lighting, and electric machinery. The resulting interference may be so large as to prevent recording of the EMG. If we were to use an amplifier with a single-ended input, we would see the magnitude of this interference. Figure 10.9 depicts hum interference on the active electrode. It appears as a sinusoidal signal, and if the muscle is contracting, its EMG is added. However, hum could be 100 mV, and would drown out even the largest EMG signal.

If we replace the single-ended amplifier with a differential amplifier (see Figure 10.10), we can possibly eliminate most of the hum. Such an amplifier takes the difference between the signals on the active terminals. As can be seen, this hum interference appears as an equal amplitude on both active terminals. Because the body acts as an antenna, all locations pick up the same hum signal. Because this unwanted signal is common to both active terminals, it is called a *common-mode signal*. At terminal 1, the net signal is $V_{\text{hum}} + \text{emg}_1$; at terminal 2 it is $V_{\text{hum}} + \text{emg}_2$. The amplifier has a gain of A. Therefore, the ideal output signal is:

$$\begin{aligned} e_o &= A(e_1 - e_2) \\ &= A(V_{\text{hum}} + \text{emg}_1 - V_{\text{hum}} - \text{emg}_2) \\ &= A(\text{emg}_1 - \text{emg}_2) \end{aligned} \tag{10.4}$$

The output *eo* is an amplified version of the difference between the EMGs on electrodes 1 and 2. No matter how much hum is present at the individual electrodes, it has been removed by a perfect subtraction within the differential amplifier. Unfortunately, a perfect subtraction never occurs, and the measure of how successfully this has been done is given by the common-mode rejection ratio (CMRR). If CMRR is 1000:1, then all but 1/1000 of the hum will

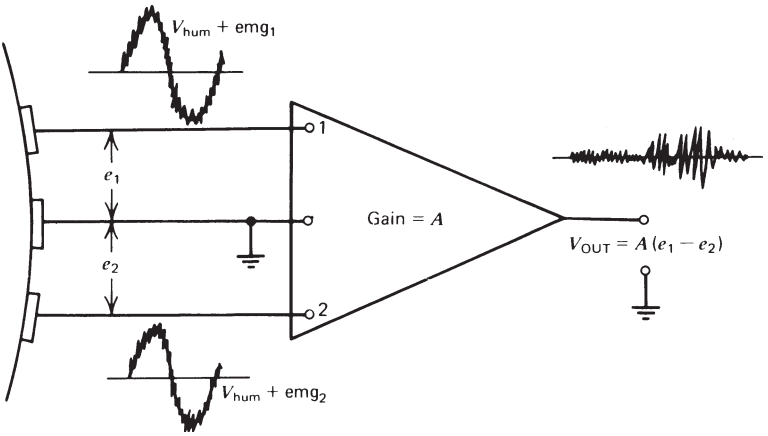


Figure 10.10 Biological amplifier showing how the differential amplifier rejects the common (hum) signal by subtracting the hum signal that is present at an equal amplitude at each active terminal. Different EMG signals are present at each electrode; thus, the subtraction does not result in a cancellation.

be rejected. Thus, the hum at the output is given by:

$$V_o \text{ (hum)} = \frac{A \times V_{\text{hum}}}{\text{CMRR}} \tag{10.5}$$

Example 10.1. Consider the EMG on the skin to be 2 mV in the presence of hum of 500 mV. CMRR is 10,000:1 and the gain is 2000. Calculate the output EMG and hum.

$$\begin{aligned} e_o &= A(\text{emg}_1 - \text{emg}_2) \\ &= 2000 \times 2\,\text{MV} = 4\ \text{V} \\ \text{output hum} &= 2000 \times 500\,\text{mV} \ \div 10{,}000 = 100\,\text{mV} \end{aligned}$$

Hum is always present to some extent unless the EMG is being recorded with battery-powered equipment not in the presence of domestic power. Its magnitude can be seen in the baseline when no EMG is present. Figure 10.11 shows two EMG records, the first with hum quite evident, the latter with negligible hum.

CMRR is often expressed as a logarithmic ratio rather than a linear ratio. The units of this ratio are decibels,

$$\text{CMRR (dB)} = 20 \log_{10} \text{ CMRR(linear)} \tag{10.6}$$

If CMRR $= 10,000 : 1$, then CMRR (dB) $= 20 \log_{10} 10{,}000 = 80$ dB. In good-quality biological amplifiers, CMRR should be 80 dB or higher.

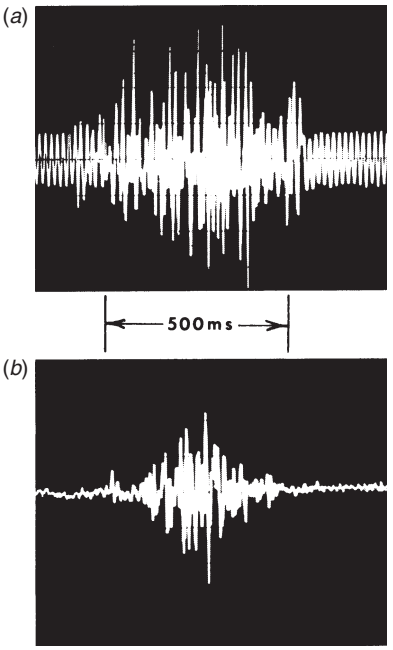


Figure 10.11 Storage oscilloscope recordings of an EMG signal. (*a*) With hum present. (*b*) Without hum.

10.2.5 Cross-Talk in Surface Electromyograms

The detectable pick-up distance was estimated to be about 0.5 cm for small motor units and about 1.5 cm for the largest units (Fuglevand et al., 1992). In Figure 10.12, for example, we see a number of surface electrodes over several thigh muscles in the midthigh region. The range of pick-up is shown as an arc beneath each electrode. Those m.u.a.p.'s whose fibers are close to each electrode are not subject to cross-talk; however, there is an overlapping zone of pick-up where both electrodes detect m.u.a.p.'s from the same active motor units. This common pick-up is called cross-talk.

Experienced researchers who are knowledgeable about their surface anatomy chose sites that minimize cross-talk. However, because of close proximity of the desired muscles, it may be necessary to check that cross-talk is minimal. The most common way to test for cross-talk is to conduct a manual resistance test prior to any given experiment. Consider two adjacent muscles A and B that, during the course of a given movement, have common periods of activity. The purpose of the manual resistance tests is to get a contraction on muscle A without any activity on B and vice versa. Consider electrodes placed on the leg on three muscles that are fairly close together: tibialis anterior (TA), peroneus longus (PER), and lateral gastrocnemius (LG). Functionally, the TA is a dorsiflexor and invertor, PER

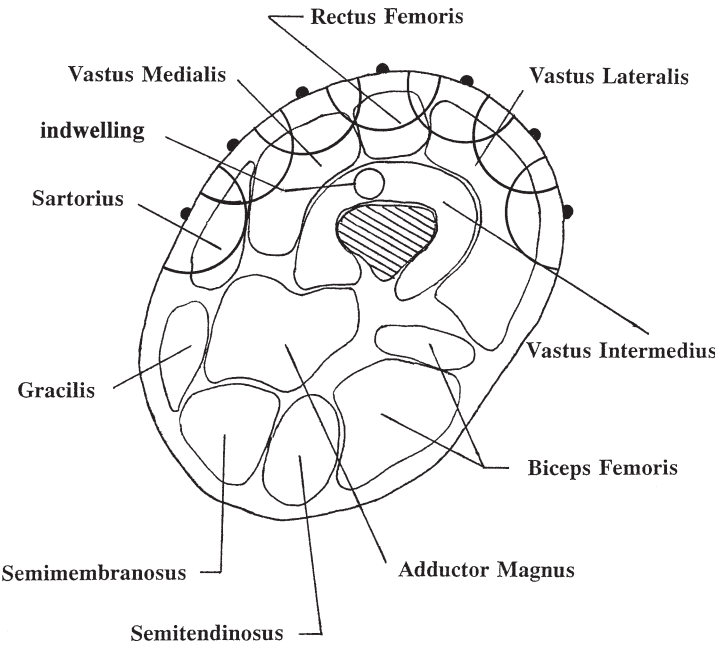


Figure 10.12 Location of seven surface electrodes placed across the midthigh showing the region of pick-up for each site relative to the underlying muscles. The zone of pick-up for each is shown as an arc beneath electrode; there is an overlapping zone between adjacent electrodes that will result in some cross-talk. Deeper muscles, such as the vastus intermedius and adductor magnus, require indwelling electrodes whose zone of pick-up is small because of the small surface area and close spacing of the electrode tips.

is an evertor and plantarflexor, and LG is a plantarflexor. The results of a functional test for cross-talk from the PER to the TA and LG is presented in Figure 10.13. Because it is difficult for many subjects to voluntarily create a unique contraction (such as eversion for the PER), it is very helpful for the subject to view the EMG recording and use visual feedback to assist in the test. Here, we see two very distinct PER contractions. The first was eversion with a small amount of dorsiflexion: note the minor activity on TA and negligible amount on LG. The second contraction was eversion with a minor amount of plantarflexion: note the minor activity on the LG and none on TA. It is not critical to be successful on both tests simultaneously, or even on the very first attempt. The student is referred to Winter et al. (1994) for more details and examples of manual resistance tests.

However, there are many situations when a manual resistance test cannot be done, and there is a good chance that adjacent channels may have some common EMG signal. The question is how much do they have in common? This question was initially addressed by Winter et al. (1994) using the well-recognized signal-processing technique called cross-correlation (see

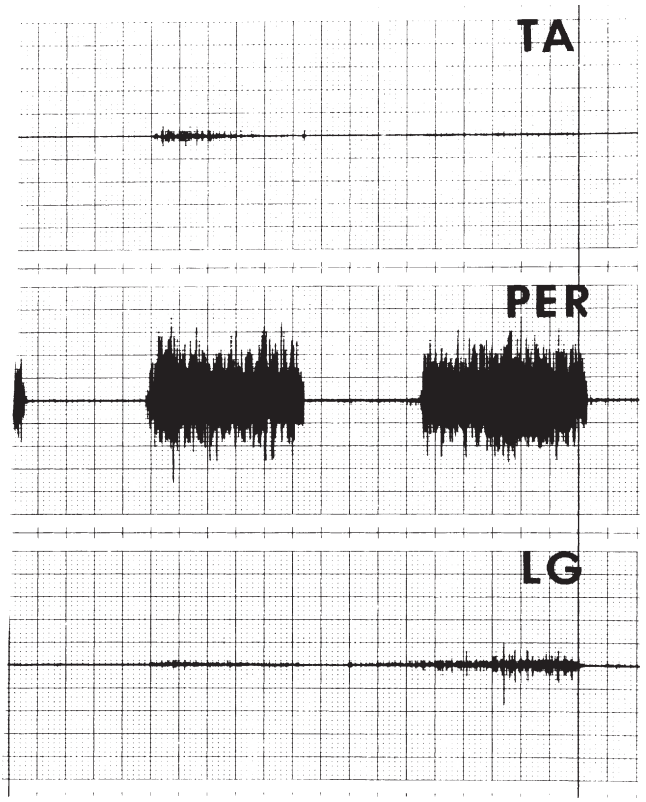


Figure 10.13 Results of a manual resistance test to ensure no cross-talk between the peroneii (PER) muscle and the tibialis anterior (TA) and lateral gastrocnemius (LG) muscles. The first trial showed negligible cross-talk between PER and TA; the second contraction showed negligible cross-talk between PER and LG. See text for further details. (Reprinted from the *Journal of Electromyography and Kinesiology*, Vol. 4, Winter, D. A., A. J. Fuglevand, and S. Archer. "Cross-talk in Surface Electromyography: Theoretical and Practical Estimates," pp. 15–26, 1994, with permission of Elsevier.)

Section 2.1). The adjacent channels, $x(t)$ and $y(t)$, are isometrically contracted for about 10 s at a moderate contraction level. The general equation for cross-correlation, $R_{xy}(\tau)$, between signals $x(t)$ and $y(t)$ is:

$$R_{xy}(\tau) = \frac{\frac{1}{T}\int_0^T x(t) \cdot y(t-\tau)\, dt}{\sqrt{R_{xx}(0) \cdot R_{yy}(0)}} \tag{10.7}$$

where $x(t)$ and $y(t)$, respectively, are at a phase shift of 0. $R_{xx}(0)$ and $R_{yy}(0)$ are the mean squares of $x(t)$ and $y(t)$, respectively, over the time T. τ is normally varied ± 30 ms, and a typical cross-correlation for two electrode pairs located 2 cm apart is presented in Figure 10.14. Note that the peak

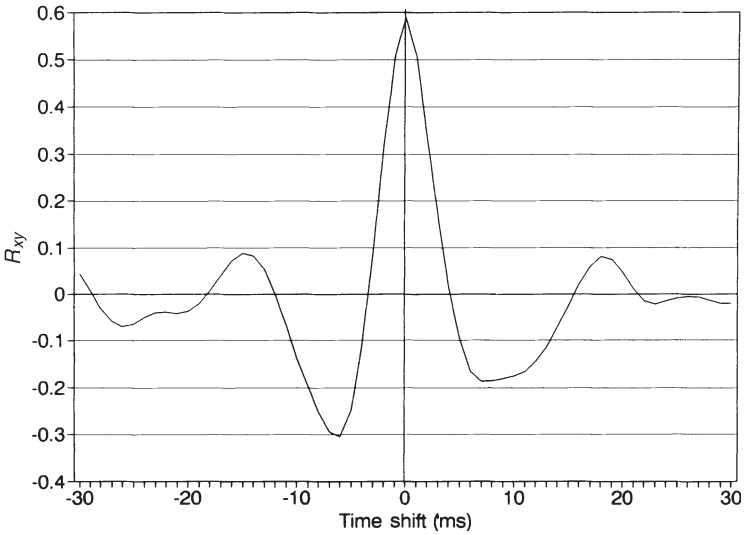


Figure 10.14 Results of a cross-correlation, *Rxy*, between two adjacent electrodes in Figure 10.13. Here, $R_{xy} = 0.6$, indicating a common pick-up of $R_{xy}^2 = 0.36$, or 36% cross-talk. (Reprinted from the *Journal of Electromyography and Kinesiology*, Vol. 4, Winter, D. A., A. J. Fuglevand, and S. Archer. "Crosstalk in Surface Electromyography: Theoretical and Practical Estimates," pp. 15–26, 1994, with permission of Elsevier.)

correlation is 0.6 at zero phase shift. Because the electrode pairs were placed parallel with the muscle fibers, each pair was the same distance from the motor point; thus, any common m.u.a.p.'s will be virtually in phase. With the peak $R_{xy} = 0.6$, the net correlation, $R_{xy}^2 = 0.36$, tells us the two EMG had 36% of their signals in common. R_{xy} drops off sharply as the distance between electrode pairs is increased (Winter et al., 1994). At 2.5 cm, $R_{xy} = 0.48$ (23% cross-talk); at 5.0 cm, $R_{xy} = 0.24$ (6% cross-talk); and at 7.5 cm, $R_{xy} = 0.14$ (2% cross-talk). Reducing the size of the electrodes reduces the pick-up zone of the electrode, and thus the cross-talk. The motor units at a greater distance from the electrodes contribute most to the cross-talk, and the m.u.a.p.'s from those units are longer duration (lower frequency). Thus, by processing the EMG through a differentiator, the higher-frequency (closer) m.u.a.p.'s are emphasized, while the lower-frequency (further) m.u.a.p.'s are attenuated.

10.2.6 Recommendations for Surface Electromyogram Reporting and Electrode Placement Procedures

Over the past 30 years, the availability of EMG recording equipment and the number of laboratories using such equipment has exploded dramatically. Most labs have developed their own protocols regarding the details needed for reporting results and for selection of the electrode sites over the muscles. The

first attempt at developing recommended standards was done by the International Society of Electrophysiological Kinesiology, which resulted in an Ad Hoc Committee Report (Winter et al., 1980). More recently, a considerably more in-depth report was published with the support of the European Union and was called SENIAM (Surface EMG for a Non-Invasive Assessment of Muscles). During 1996–1999, the report was developed and reviewed by over 100 laboratories (Hermens et al., 2000), and the final report was published as a booklet and as a CD-ROM: SENIAM 8: European Recommendations for Surface Electromyography, 1999.

The SENIAM recommendations as to information to be reported are as follows:

1. Electrode size, shape (square, circular, etc.), and material (Ag, AgCl, Ag/AgCl, etc.)
2. Electrode type (monopolar, bipolar, one- or two-dimensional array)
3. Skin preparation, interelectrode distance
4. Position and orientation on the muscle (recommendations for 27 muscles are included)

10.3 PROCESSING OF THE ELECTROMYOGRAM

Once the EMG signal has been amplified, it can be processed for comparison or correlation with other physiological or biomechanical signals. The need for changing the EMG into another processed form is caused by the fact that the raw EMG may not be suitable for recording or correlation. For example, because of the higher frequencies present in the EMG, it is impossible to record it directly on a pen recorder. The frequency response of most recorders (0–60 Hz) means that most of the higher frequency components of the EMG are not seen.

The more common types of online processing are:

1. Half- or full-wave rectification (the latter also called absolute value)
2. Linear envelope detector (half- or full-wave rectifier followed by a low-pass filter)
3. Integration of the full-wave rectified signal over the entire period of muscle contraction
4. Integration of the full-wave rectified signal for a fixed time, reset to zero, then integration cycle repeated
5. Integration of the full-wave rectified signal to a preset level, reset to zero, then integration repeated

These various processing methods are shown schematically in Figure 10.15, together with a sample record of a typical EMG processed all five ways. This is now discussed in detail.

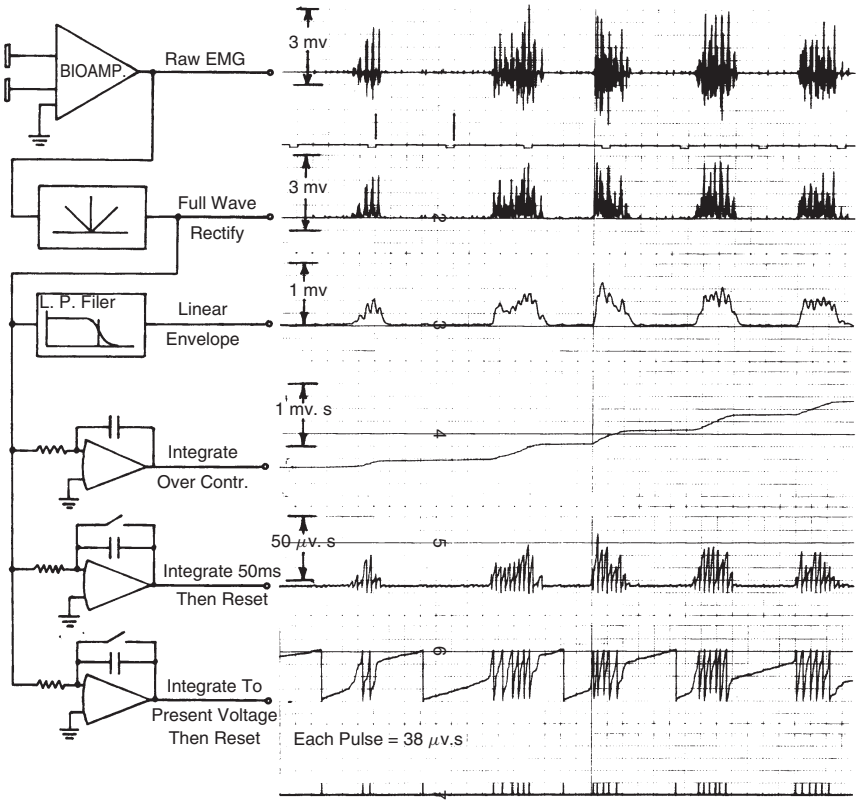


Figure 10.15 Schematic diagram of several common EMG processing systems and the results of simultaneous processing of EMG through these systems. See the text for details.

10.3.1 Full-Wave Rectification

The full-wave rectifier generates the absolute value of the EMG, usually with a positive polarity. The original raw EMG has a mean value of zero because it is recorded with a biological amplifier with a low-frequency cutoff around 10 Hz. However, the full-wave rectified signal does not cross through zero and, therefore, has an average or bias level that fluctuates with the strength of the muscle contraction. The quantitative use of the full-wave rectified signal by itself is somewhat limited; it serves as an input to the other processing schemes. The main application of the full-wave rectified signal is in semi-quantitative assessments of the phasic activity of various muscle groups. A visual examination of the amplitude changes of the full-wave rectified signal gives a good indication of the changing contraction level of the muscle. The proper unit for the amplitude of the rectified signal is the millivolt, the same as for the original EMG.

10.3.2 Linear Envelope

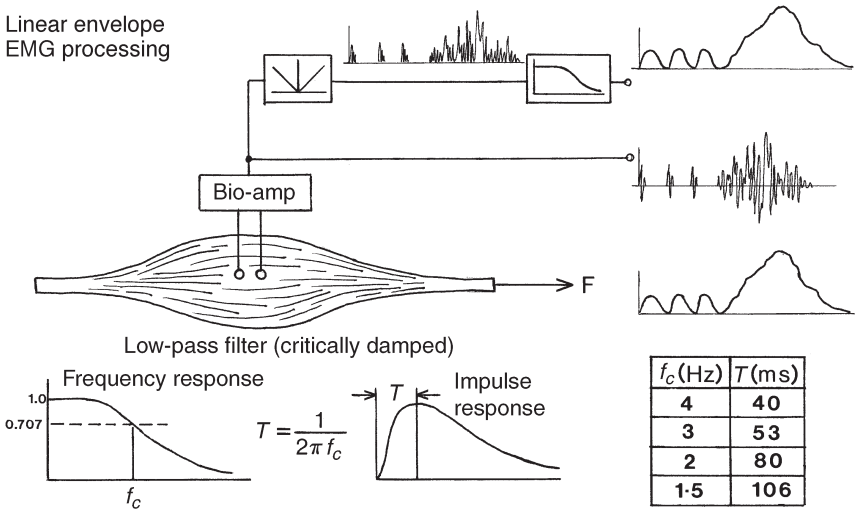
If we filter the full-wave rectified signal with a low-pass filter, we have what is called the *linear envelope*. It can be described best as a moving average because it follows the trend of the EMG and quite closely resembles the shape of the tension curve. It is reported in millivolts. There is considerable confusion concerning the proper name for this signal. Many researchers call it an *integrated EMG* (IEMG). Such a term is quite wrong because it can be confused with the mathematical term *integrated*, which is a different form of processing.

There is a need to process the signal to provide the assessor with a pattern that can be justified on some biophysical basis. Some researchers have full-wave rectified the raw EMG and low-pass filtered at a high frequency but with no physiological basis (Forssberg, 1985; Murray et al., 1985). If it is desired that the linear envelope bear some relationship to the muscle force or the joint moment of force, the processing should model the biomechanics of muscle tension generation. The basic unit of muscle tension is the muscle twitch, and the summation of muscle twitches as a result of recruitment is matched by a superposition of m.u.a.p.'s. There is an inherent delay between the m.u.a.p. and the resultant twitch waveform. If we consider the full-wave rectified signal to be an impulse and the twitch to be the response, we can define the transfer function of the desired processing. The duration of the full-wave rectified m.u.a.p. is about 10 ms, while the twitch waveform peaks at 50–110 ms and lasts up to 300 ms, so this impulse/response relationship is close. Twitch waveforms have been analyzed and have been found to be a second-order system, critically damped or slightly overdamped (Crosby, 1978; Milner-Brown et al., 1973). The cutoff frequency of these second-order responses ranged from 2.3 Hz to 7.8 Hz. Figure 10.16 is presented to show the linear envelope processing of the EMG to model the relationship between the m.u.a.p. and the twitch. A critically damped second-order low-pass filter has a cutoff frequency f_c, which is related to the twitch time T as follows:

$$f_c = 1/2\pi T \tag{10.8}$$

Thus, the table in Figure 10.16 shows the relationship between fc and T for the range of twitch times reported in the literature. The soleus muscle with a twitch time $\approx$ 106ms would require a filter with $f_c = 1.5$ Hz. The reader is referred back to Section 9.0.5 for information on the twitch shape and times.

Good correlations have been reported between the muscle force and linear envelope waveforms during isometric anisotonic contractions (Calvert and Chapman, 1977; Crosby, 1978). Readers are also referred back to Section 9.3.1 and Figure 9.20, which modeled the muscle contraction as a mass-spring-damper system that was critically damped. The critically damped low-pass filters described earlier have exactly the same response as this previously described mechanical system. Thus, the matching of the muscle force waveform with that from the model is exactly the same as



f_c (Hz)	T (ms)
4	40
3	53
2	80
1·5	106

Figure 10.16 Linear envelope processing of the EMG with a critically damped low-pass filter to match the impulse response of the muscle being recorded. The full-wave rectified EMG acts as a series of impulses that, when filtered, mimic the twitch response of the muscle and, in a graded contraction, mimic the superposition of muscle twitches.

would be achieved with a linear envelope detector with the same cutoff frequency as used in the mechanical model.

10.3.3 True Mathematical Integrators

There are several different forms of mathematical integrators, as shown in Figure 10.15. The purpose of an integrator is to measure the "area under the curve." Thus, the integration of the full-wave rectified signal will always increase as long as any EMG activity is present. The simplest form starts its integration at some preset time and continues during the time of the muscle activity. At the desired time, which could be a single contraction or a series of contractions, the integrated value can be recorded. The unit of a properly integrated signal is millivolt-seconds (mV·s). The only true way to find the average EMG during a given contraction is to divide the integrated value by the time of the contraction; this will yield a value in millivolts.

A second form of integrator involves a resetting of the integrated signal to zero at regular intervals of time (40–200 ms). Such a scheme yields a series of peaks that represent the trend of the EMG amplitude with time. Each peak represents the average EMG over the previous time interval, and the series of peaks is a "moving" average. Each peak has units of millivolt-seconds so that the sum of all the peaks during a given contraction yields the total integrated signal, as described in the previous paragraph. There is a close similarity

between this reset, integrated curve and the linear envelope. Both follow the trend of muscle activity. If the reset time is too high, it will not be able to follow rapid fluctuations of EMG activity, and if it is reset too frequently, noise will be present in the trendline. If the integrated peaks are divided by the integration time, the amplitude of the signal can again be reported in millivolts.

A third common form of integration uses a voltage level reset. The integration begins before the contraction. If the muscle activity is high, the integrator will rapidly charge up to the reset level; if the activity is low, it will take longer to reach rest. Thus, the strength of the muscle contraction is measured by the frequency of the resets. A high frequency of reset pulses reflects high muscle activity; a low frequency represents low muscle activity. Intuitively, such a relationship is attractive to neurophysiologists because it has a similarity to the action potential rate present in the neural system. The total number of counts over a given time is proportional to the EMG activity. Thus, if the threshold voltage level and the gain of the integrater are known, the total EMG activity (millivolt-seconds) can be determined.

10.4 RELATIONSHIP BETWEEN ELECTROMYOGRAM AND BIOMECHANICAL VARIABLES

The major reason for processing the basic EMG is to derive a relationship between it and some measure of muscle function. A question that has been posed for years is: "How valuable is the EMG in predicting muscle tension?" Such a relationship is very attractive because it would give an inexpensive and noninvasive way of monitoring muscle tension. Also, there may be information in the EMG concerning muscle metabolism, power, fatigue state, or contractile elements recruited.

10.4.1 Electromyogram versus Isometric Tension

Bouisset (1973) has presented an excellent review of the state of knowledge regarding the EMG and muscle tension in normal isometric contractions. The EMG processed through a linear envelope detector has been widely used to compare the EMG-tension relationship, especially if the tension is changing with time. If constant tension experiments are done, it is sufficient to calculate the average of the full-wave rectified signal, which is the same as that derived from a long time-constant linear envelope circuit. Both linear and nonlinear relationships between EMG amplitude and tension have been discovered. Typical of the work reporting linear relationships is an early study by Lippold (1952) on the calf muscles of humans. Zuniga and Simons (1969) and Vredenbregt and Rau (1973), on the other hand, found quite nonlinear relationships between tension and EMG in the elbow flexors over a wide range of joint angles. Both these studies were, in effect, static calibrations of

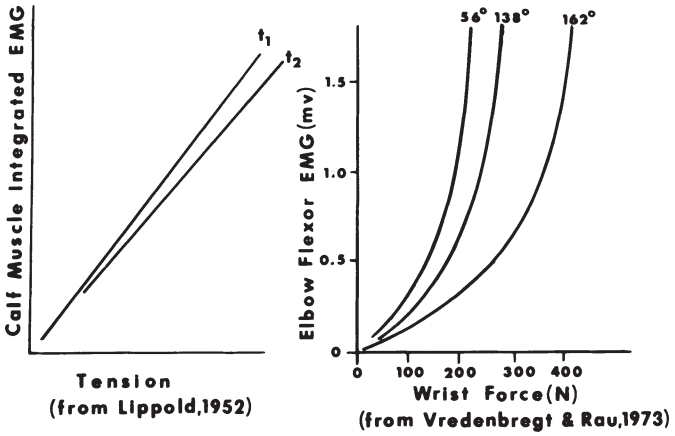


Figure 10.17 Relationship between the average amplitude of EMG and the tension in the muscle in isometric contraction. Linear relationships have been shown by some researchers, while others report the EMG amplitude to increase more rapidly than the tension.

the muscle under certain length conditions; a reproduction of their results is presented in Figure 10.17.

Another way of representing the level of EMG activity is to count the action potentials over a given period of time. Close et al. (1960) showed a linear relationship between the count rate and the integrated EMG, so it was not surprising that the count rate increased with muscle tension in an almost linear fashion.

The relationship between force and linear envelope EMG also holds during dynamic changes of tension. Inman et al. (1952) first demonstrated this by a series of force transducer signals that were closely matched by the envelope EMG from the muscle generating the isometric force. Gottlieb and Agarwal (1971) mathematically modeled this relationship with a second-order low-pass system (e.g., a low-pass filter). Under dynamic contraction conditions, the tension is seen to lag behind the EMG signal, as shown in Figure 10.18. The delay is caused by the fact that the twitch corresponding to each m.u.a.p. reaches its peak 40–100 ms afterward. Thus, as each motor unit is recruited, the resulting summation of twitch forces will also have a similar delay behind the EMG.

In spite of the reasonably reproducible relationships, the question still arises as to how valid these relationships are for dynamic conditions when many muscles act across the same joint:

1. How does the relationship change with length? Does the length change merely alter the mechanical advantage of the muscle, or does the changing overlap of the muscle fibers affect the EMG itself (Vredenbregt and Rau, 1973)?

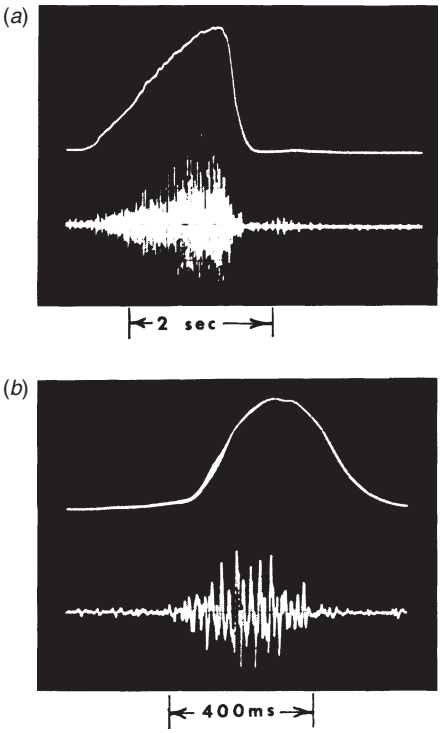


Figure 10.18 EMG and muscle tension recorded on a storage oscilloscope during varying isometric contractions of the biceps muscles. Note the delay between the EMG and the initial build-up of tension, time to reach maximum tension, and drop of tension after the EMG has ceased. (*a*) During a gradual build-up and rapid relaxation. (*b*) During a short 400-ms contraction.

2. How do other agonist muscles share the load at that joint, especially if some of the muscles have more than one function (Vredengregt and Rau, 1973)?
3. In many movements, there is antagonist activity. How much does this alter the force being predicted by creating an extra unknown force?

With the present state of knowledge, it appears that a suitably calibrated linear envelope EMG can be used as a coarse predictor of muscle tension for muscles whose length is not changing rapidly.

10.4.2 Electromyogram during Muscle Shortening and Lengthening

In order for a muscle to do positive or negative work, it must also undergo length changes while it is creating tension. Thus, it is important to see how well the EMG can predict tension under these more realistic conditions. A

major study has been reported by Komi (1973). In it the subject did both positive and negative work on an isokinetic muscle-testing machine. The subject was asked to generate maximum tension while the muscle lengthened or shortened at controlled velocities. The basic finding was that the EMG amplitude remained fairly constant in spite of decreased tension during shortening and increased tension during eccentric contractions (see Section 9.2.1). Such results support the theory that the EMG amplitude indicates the state of activation of the contractile element, which is quite different from the tension recorded at the tendon. Also, these results combined with later results (Komi et al., 1987) indicate that the EMG amplitude associated with negative work is considerably less than that associated with the same amount of positive work. Thus, if the EMG amplitude is a relative measure of muscle metabolism, such a finding supports the experiments that found negative work to have somewhat less metabolic cost than positive work.

10.4.3 Electromyogram Changes during Fatigue

Muscle fatigue occurs when the muscle tissue cannot supply the metabolism at the contractile element, because of either ischemia (insufficient oxygen) or local depletion of any of the metabolic substrates. Mechanically, fatigue manifests itself by decreased tension, assuming that the muscle activation remains constant, as indicated by a constant surface EMG or stimulation rate. Conversely, the maintenance of a constant tension after onset of fatigue requires increased motor unit recruitment of new motor units to compensate for a decreased firing rate of the already recruited units (Vredenbregt and Rau, 1973). Such findings also indicate that all or some of the motor units are decreasing their peak twitch tensions but are also increasing their contraction times. The net result of these changes is a decrease in tension.

Fatigue not only reduces the muscle force but also may alter the shape of the motor action potentials. It is not possible to see the changes of shape of the individual m.u.a.p.'s in a heavy voluntary contraction. However, an autocorrelation shows an increase in the average duration of the recruited m.u.a.p. (see Section 10.1.4). Also, the EMG spectrum shifts to reflect these duration changes; Kadefors et al. (1973) found that the higher-frequency components decreased. The net result is a decrease in the EMG frequency spectrum, which has been attributed to the following:

1. Lower conduction velocity of the action potentials along the muscle fibers below the nonfatigued velocity of 4.5 m/s (Mortimer et al., 1970; Krogh-Lund and Jørgensen, 1991).
2. Some of the larger and faster motor units with shorter duration m.u.a.p.'s dropping out.
3. A tendency for the motor units to fire synchronously, which increases the amplitude of the EMG. Normally, each motor unit fires independently of others in the same muscle so that the EMG can be considered

to be the summation of a number of randomly timed m.u.a.p.'s. However, during fatigue, a tremor (8–10 Hz) is evident in the EMG and force records. These fluctuations are neurological in origin and are caused by motor units firing in synchronized bursts.

The major EMG measures of fatigue are an increase in the amplitude of the EMG and a decrease in its frequency spectrum. One of those measures is median power frequency, f_m, which is the frequency of the power spectral density function below which half the power lies and above which the other half of the power lies:

$$\int_0^{f_m} X^2(f)\,df = \int_{f_m}^{\infty} X^2(f)\,df = \frac{1}{2}\int_0^{\infty} X^2(f)\,df \tag{10.9}$$

where $X(f)$ is the amplitude of the harmonic at frequency f, and $X^2(f)$ is the power at frequency f.

A representative paper that shows the majority of these measures (Krogh-Lund and Jørgensen, 1991) reports a 10% decrease in conduction velocity, a 45% decrease in the median power frequency, and a 250% increase in the root mean square (rms) amplitude. The median power frequency has also been shown to be highly correlated with the amplitude of the low-frequency bandwidth (15–45 Hz) and the ratio of the high-frequency bandwidth (>95 Hz) to the low-frequency bandwidth (Allison and Fujiwara, 2002). An alternate and very common statistical measure (Öberg et al., 1994) is mean power frequency (MPF):

$$\text{MPF} = \frac{\int_0^F f \cdot X^2(f)\,df}{\int_0^F X^2(f)\,df}\text{Hz} \tag{10.10}$$

where F is the maximum frequency analyzed.

There is virtually no difference in these two frequency measures and their correlations with independent measures of fatigue (Kerr and Callaghan, 1999).

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