CHAPTER 7

EMG Signal Amplitude and Force

The surface EMG signal may be conveniently detected with minimal insult to the subject. For this reason it has become very useful in many applications which require an assessment of the muscular effort. The reader is referred to the material in Chapter 2, which addresses the details of this issue.

A considerable controversy exists concerning the description of this relationship. Early theoretical studies (Person and Libkind, 1967; Bernshtein, 1967; Moore, 1967; Libkind, 1968 and 1969) all suggested that, for isometric contractions, the amplitude of the EMG signal should increase as the square root of force generated by the muscle when the motor units are activated independently. These studies were instrumental in generating interest in providing a more structured approach to the interpretation of the EMG signal-force relationship. It is now clear that the assumptions and approximations which were made were simply too generous. The reader is referred to the relationships and associations between the EMG signal, as a function of force and time, with known physiological correlates displayed in Figure 3.9. In those expressions it is apparent that the relationship is complex. In fact, surprisingly few experimental results support the square root relationship. Almost without exception, investigators report either linear relationships or a more than linear increase of the EMG signal with increasing force.

RELATIONSHIP DURING ISOMETRIC CONTRACTIONS

Monotonically Increasing Contractions

Table 7.1 provides a sample of the studies relating to this issue which have been reported in the literature between 1952 and 1979. No attempt has been made to include all the published reports. The contents of the table were designed to represent the wide variety in and disparity among the wealth of studies which have been performed. These investigations are characterized by considerable variability in the muscles examined, the types of contractions performed, and the quantities derived from the raw data to represent the amplitude of the EMG signal.

Beyond the obvious disparities among the reported studies, it is necessary to consider particulars which are specific to the muscle or muscle group which is involved in the force generation process. For example, (*a*) Relatively small muscles, such as those in the hand, and relatively large muscles in the limbs are controlled by different firing rate-recruitment schemes. For additional details on this point, the reader is referred

Table 7.1.
Selection of Previous Investigations of the EMG Signal-Force Relationship

First author	Yr	Muscle	EMG-F relation	Measure of EMG	Contractions: Type	Contractions: Angle (from full extension)	Position	Electrodes: Type	Electrodes: Diam (cm)	Electrodes: Sep (cm)	Subjects: #	Subjects: Sex	Subjects: Ages (range or avg)
Inman	1952	Biceps, triceps, pectoral, ant. tibialis	Linear	IEMG	Isom F-vary (ramp)			Surf bip, wire & needle	1.5		11 (Amputees)		
Lippold	1952	Soleus, gastrocnemius	Linear	IEMG	Isom	90° ca. knee	Plantar flexion	Surf bip	0.6		30		
Bigland	1954	Calf, finger extensor	Linear	IEMG	Isom isokin, isot isokin	90° ca. knee	Plantar flexion	Surf bip	0.6			M/F	Young adult
Scherrer	1959	Triceps	Linear	IEMG	Isom (ergonometric reps)			Surf bip, needle	~0.7		5		
Close	1960	Soleus	Linear	MUAP counts	Isom isot & isom isokin	90° ca. knee, angle varied at ankle		Wire			6		Young
Liberson	1962	Biceps	Linear	IEMG	Isom isoton			Surf bip		5.0	10		Adult
Mason	1969	Abd. and ext. digiti minimi	Not regular	Mean amplitude of AP spike	Isom	10° between 4th & 5th fingers	Wrist pronated	Needle			12	M	18–40
Komi	1970	Biceps	(a)Linear (b)Non/incr	IEMG IEMG	Isom isoton, isot isokin	90° ca. elbow 10°–115°	Wrist supinated	Surf bip	1.1	5.0	8 29	M	17–29
Bouisset	1971	Biceps, triceps	Linear	IEMG	Isom isoton	90° ca. elbow		Surf bip			4		
Stephens	1972	FDI	Linear	Smoothed rect. EMG (sre)	Isom F-vary (ramp)			Surf bip	1.0		18		Student
Komi	1973	Biceps, brachialis, brachioradialis	Linear	IEMG	Isot isokin	10°–130° meas. at 80°	Wrist supinated	Wire			10	M	
Vredenbregt	1973	Biceps, brachioradialis	Non/incr	Slope of rect. integrated EMG	Isom isoton	0–140°	Wrist supinated	Surf bip	0.4	4.0	1		
Stevens	1973	Biceps, triceps, brachialis, brachioradialis, coracobrachialis, deltoid	Lin and non/decr	Mean ampl. of APs	Isom isoton	90° ca. elbow	Wrist sup, semisup, pron	Surf bip			47	25M 22F	20–40
Bigland-Ritchie	1974	Quadriceps	Linear	IEMG	Isoton	Varied (on ergonometer)		Surf bip	1.0	15.0	3	2M 1F	19–29
Seyfert	1974	Biceps, vastus	Linear	Mean ampl. of EMG	Isom isoton	90° ca. elbow	Wrist semisupinated	Needle			20	M	20–30
Milner-Brown	1975	FDI	Linear	Mean rect. EMG	Isom			Surf bip	0.9	3.0	6		
Soechting	1975	Biceps, triceps	Linear	IEMG	Isom isoton, isom F-vary (sinusoid)	90° ca. elbow	Wrist supinated	Surf bip	0.9		4		
Komi	1976	Rectus femoris	Non/incr	IEMG	Isom isoton	60° ca. knee		Surf bip	0.4	1.0	12	8M 4F	13–15
Thorstensson	1976	Rectus femoris, vastus lateralis	Non/incr	IEMG	Isom isoton	60° ca. knee		Surf bip	Mini		8	M	24.0
van Hoecke	1978	Biceps, pronator teres	Linear & non/incr	IEMG	Isom isoton	0°–135°	Pron, s-p, sup						
Komi	1978	Rectus femoris	Non/incr	IEMG	Isom isoton			Surf bip	0.4		12	8M 4F	13–15
Moritani	1978	Biceps	Linear	IEMG	Isom	90° ca. elbow	Wrist semi-sup	Surf bip, surf unip	1.4	5.1	26	M	21.1

Table 7.1—*Continued*

First author	Yr	Muscle	EMG-F relation	Measure of EMG	Contractions: Type	Contractions: Angle (from full extension)	Position	Electrodes: Type	Electrodes: Diam (cm)	Electrodes: Sep (cm)	Subjects: #	Subjects: Sex	Subjects: Ages (range or avg)
Hagberg	1979	Biceps	Linear	Rect. & filtered EMG	Isom isoton	90° ca. elbow & shoulder	Wrist semi-sup	surf bip			5	3M 2F	20–34
Lam	1979	Soleus (cat)	Linear	IEMG	Isom, isom isoton			Wire			1		

Notes:

1. Complete references are included in the bibliography.
2. The experiments listed involved nonfatiguing contractions, except for Scherrer (1959).
3. In general, where details are not provided, the information was not explicitly available.
4. Additional specifications which could be included are:
   a. Specific electrode placement
   b. Skin preparation, if any
   c. Type of feedback, if any
   d. Range of force investigated as a percent of MVC
   e. Procedures followed to avoid fatigue
   f. Level of exercise training of subjects
   g. Whether contractions were practiced to achieve some degree of accuracy before actual experiments
   h. Data normalization techniques, if any

Abbreviations:

EDI, first dorsal interosseous
Non/incr, a nonlinear MES-F relation in which the MES increases more than linearly with increasing force
MUAP, motor unit action potential
Isom, isometric
Isot or isoton, isotonic
Isokin, isokinetic
Sup, supinated
Pron, pronated
s-p, semisupinated
Meas, measured
Surf bip, surface bipolar
Surf unip, surface unipolar

to Chapter 5. (*b*) The degree of synergistic action of muscle groups is influenced by the relative spatial orientation of individual muscles. (*c*) Varying amounts of cocontraction among antagonist muscle groups may bias the recorded signals attributed to individual muscles. It is conceivable that any or all of these factors, as well as others, could contribute deterministically to the EMG signal-force relationship.

In addition, variability in detection and data processing techniques may explain some of the inconsistencies which have been reported for specific muscles. The reader is referred to the discussion presented in Chapter 2 which directly addresses the essential technical issues. A review of the reported literature reveals that no consensus has existed on a specific anatomical detection site on the surface muscle from which to detect the signal. All imaginable types of electrodes have been used (including surface, indwelling needle, and wire electrodes) in both the monopolar and bipolar combination. High input impedance amplifiers, employing modern FET-technology eliminate the need to reduce and monitor skin impedance. However, past investigators did not have this benefit and used a variety of methods to prepare the skin. Amplifiers and filter specifications influence the final form of the processed EMG signal, and the specifications which have been used have not been consistent.

There has not been any consistent usage of one parameter measure of the amplitude. This search for a better or different parameter has, to some extent, reflected the lack of universal agreement on which parameter to use, as well as technological developments in electronics which have facilitated the necessary signal processing. Parameters which have been used include the smooth rectified amplitude (Stephens and Taylor, 1972), mean rectified amplitude (Milner-Brown and Stein, 1975), root-mean-squared amplitude (Stulen and De Luca, 1978), and several versions of integrated amplitude (Inman et al, 1952; Komi and Buskirk, 1970; Bouisset and Goubel, 1971; and several others). The distinction and limitations of each of these parameters are discussed and compared in Chapter 3.

The great physiological and technological variability so far described has made comparison and reproducibility of experimental results extremely difficult. Beyond these methodological inconsistencies, a general representation of the system output variables, i.e., the signal amplitude and force level, should contain a formulation which allows for valid comparisons between different muscles and individuals, and for retrials on the same subject. This formulation may be realized by normalizing these variables with respect to some convenient and referable quantities, such as their maximal values in a particular experimental procedure. The absence of normalization often constitutes a deficiency in many reported investigations which have compared or averaged groups of subjects.

In the remainder of this chapter we will consider only the characteristics of EMG signals detected with surface electrodes. It has been shown that the intrasubject variability of the signal is much greater when it is detected with indwelling, rather than surface, electrodes (refer to Fig. 3.12 for an example).

A direct comparison of the relationship between the EMG signal and force ouput has been reported by Lawrence and De Luca (1983). This study investigated the following aspects: (1) Whether the normalized surface-detected EMG signal amplitude vs. normalized force relationship varies in different muscles; (2) whether the relationship is dependent on exercise level; and (3) how much variability exists among the same muscles of different individuals. The data was obtained from the biceps brachii, deltoid, and first dorsal interosseous of accomplished pianists, world-class long distance swimmers, world-class power lifters, and normal subjects during voluntary isometric linearly force-varying contractions.

The signal was detected with two surface electrodes (3 mm in diameter) whose centers were spaced 2 cm apart. The electrodes were oriented parallel to the muscle fibers, and in the case of the deltoid and biceps brachii, the electrode pair was located between the innervation zone and tendinous insertion. The root-mean-squared value of the signal amplitude was used as the variant parameter because, as has been explained in Chapter 3, it is the parameter which more completely reflects the physiological correlates of the motor unit behavior during a muscle contraction.

Figure 7.1 presents the average value of the EMG signal-force data of the three muscles. Each of these three sets of data were obtained from 13 subjects. Each subject performed several contractions. The total number of contractions for each set of data is indicated in the figure. The standard deviation is not indicated because of the visual complexity that would result from the overlap. The standard deviation was generally 25% of the mean value and remained essentially constant over the entire force range. It is apparent that in the case of the first dorsal interosseous, the relationship is quasilinear. A polynomial regression analysis would reveal that a second order (nonlinear) polynomial would only improve the fit by 2 or 3%. Thus, for practical purposes it is safe to consider it as linear. No such claim may be made for the curves of the deltoid and biceps brachii data. They are nonlinear, with signal amplitude increasing more than the force.

Comparable measurements on small and large muscles have been reported (Clamann and Broecker, 1979; Woods and Bigland-Ritchie, 1983). The first group observed the relationship between the smoothened rectified amplitude of the signal from the triceps brachii, biceps

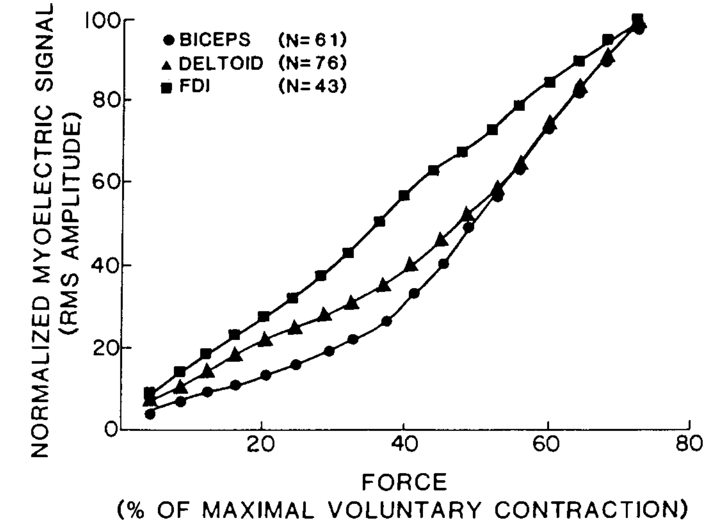

Figure 7.1. Effect of muscle on the EMG signal-force relationship. *N* represents the number of contractions averaged for each muscle. Each set of data was obtained from 13 subjects. (From J.H. Lawrence and C.J. De Luca, © 1983, *Journal of Applied Physiology*.)

brachii, adductor pollicis, and first dorsal interosseous muscles. The second group observed the relationship between the highly smoothened integrated rectified amplitude of the signal from the biceps brachii, triceps brachii, adductor pollicis, and soleus muscles.

Although the methodology was somewhat dissimilar among the three considered studies, one point is strikingly clear. The signal amplitude force output relationship for the small hand muscles, the first dorsal interosseous, and the adductor pollicis was always found to be quasilinear, whereas in the larger muscles, the biceps brachii, triceps brachii, deltoid, and soleus, the relationship was almost unanimously nonlinear. The only exception was the data for the biceps brachii from Clamann and Broecker's study: it presented a quasilinear relationship. This anomaly should serve as a reminder that even among well-executed studies, it is difficult to compare the data because the detected signal is a function of the detection procedure as well as the physiological events.

This distinction in the behavior of the signal-force relationship finds considerable support among the studies listed in the accompanying table. The pattern also appears to be unaltered with rigorous but diverse training regimens, as may be seen in Figure 7.2. These data implicitly indicate that considerable differences in fiber-type constituency do not significantly affect the relationship of the *normalized* signal amplitude and force output. This indirect association is based on the well-docu-

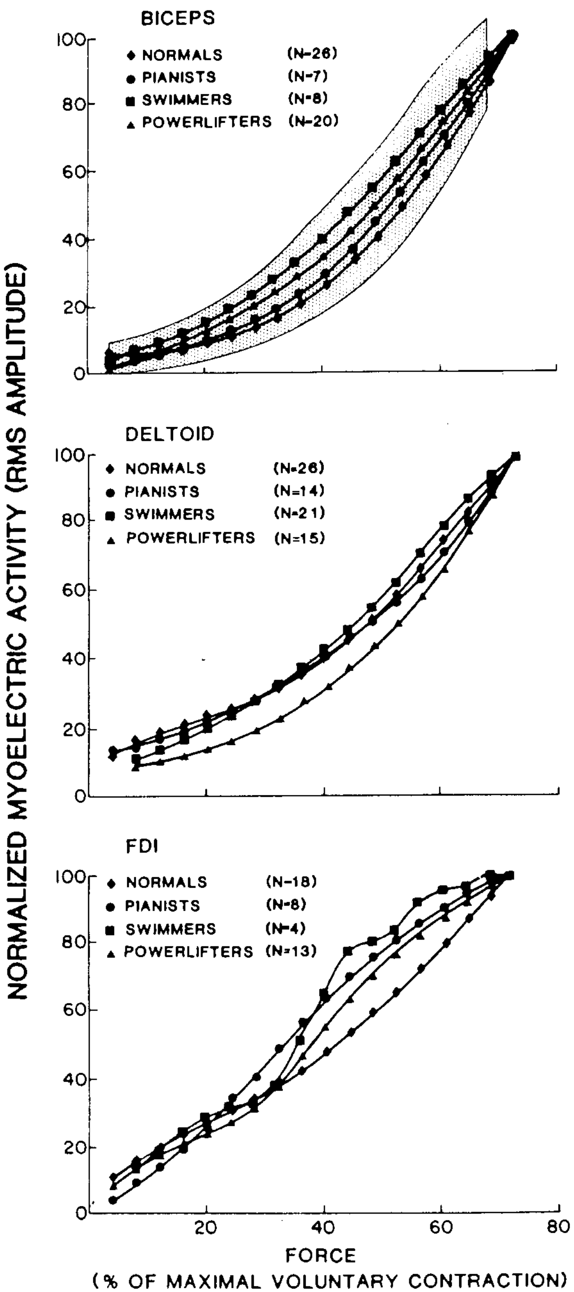

Figure 7.2. Effect of rigorous training regimens on the EMG signal-force relationship in three different muscles. Standard deviations of the raw data are indicated by the *shaded area* in the *top graph*. *N* represents the number of contractions that were averaged to obtain the plotted curves. The data was obtained from 13 subjects. (From J.H. Lawrence and C.J. De Luca, 1983, © *Journal of Applied Physiology*.)

mented fact that elite long distance swimmers have considerably different fiber-type composition in the upper limb musculature than do elite powerlifters (Gollnick et al, 1973).

The distinction in the signal amplitude-force curves remains when the force is generated at different rates ranging from 10% MCV/s to 40% MCV/s (see Fig. 7.3 for details).

A variety of phenomena that may contribute to the muscle-dependent difference in the EMG signal-force relationship can be identified. Some of them are: (*a*) motor unit recruitment and firing rate properties; (*b*) relative location of fast twitch fibers within a muscle and with respect to the detection electrodes; (*c*) cross-talk from signals originating in adjacent muscles; and (*d*) agonist-antagonist muscle interaction.

The agonist-antagonist interaction of simultaneously contracting muscles is an important consideration, especially in isometric contractions, where the joints must be stabilized. In all the studies reported in the literature only the net force or torque resulting from the agonist-antagonist interaction is measured. In many cases this approach provides the correct information with respect to the involvement of the agonist as the muscle of interest. However, in various circumstances involving the need to stiffen the joint, the antagonist(s) may be active. This situation is more likely to occur as the force output increases. In such cases, the net force is customarily assumed to be linear with respect to that of the agonist muscle. However, this relationship may be altered by numerous factors such as joint angle, limb position, and pain sensation. Thus, the signal-force relationship (from the muscle of interest) may be altered.

The electrical cross-talk from adjacent muscles is unquestionably a possible contribution to the behavior of the relationship. Again, its contribution would manifest itself more prominently as the force output of the muscle increases. The presence of cross-talk would be more dominant in smaller muscles, where the electrodes (especially the surface types) are constrained to be located near the adjacent musculature. Cross-talk may also account for the difference in the behavior of the signal-force relationship when the signal is detected with monopolar and bipolar electrodes. Conflicting reports have appeared on occasion (for example, in the work of Moritani and deVries, 1978). As discussed in Chapter 2, these two types of electrodes have considerably different frequency characteristics and detect different amounts of electrical signals from adjacent muscle tissue. The complexity of cross-talk is further compounded by the anisotropy of the muscle tissue itself and the inhomogeneity of the tissues adjacent to the muscle. For this reason, it is often not possible to accurately identify the source of the contaminating physiological signal.

The relative location of slow-twitch and fast-twitch muscle fibers within the muscle is also an important consideration because of the following

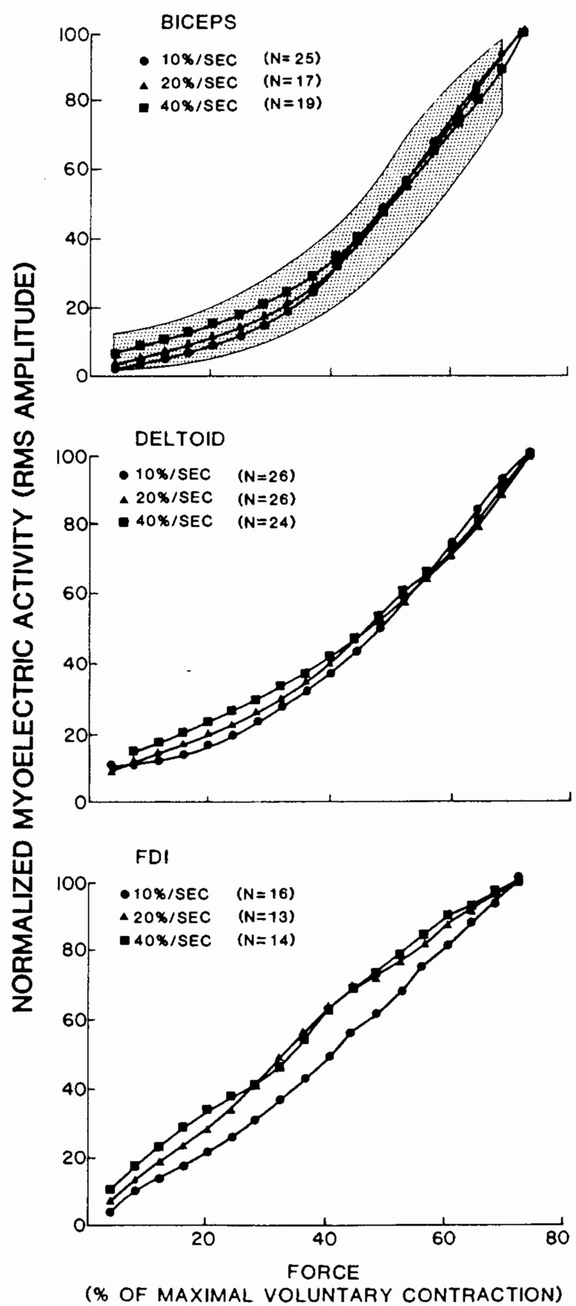

Figure 7.3. Effect of force rate (nonballistic) on the EMG signal-force relationship in three different muscles. The standard deviations of the raw data are indicated by the *shaded area* in the *top graph*. *N* represents the number of contractions that were averaged to obtain the plotted curves. The data was obtained from 13 subjects. (From J.H. Lawrence and C.J. De Luca, 1983, © *Journal of Applied Physiology*.)

reasons. The amplitude of the action potential generated by a single muscle fiber is proportional to the fiber diameter. Fast-twitch fibers, which in the human first dorsal interosseous and biceps brachii muscles are generally larger in diameter (Polgar et al, 1973), have higher amplitude action potentials than slow-twitch fibers. Higher amplitude motor unit action potentials result in a higher amplitude signal. (Refer to discussion in Chapter 3, specifically to Figure 3.11.) However, the amplitude of the motor unit action potential that contributes to the surface signal is a function of the distance between the active fibers and the detection electrodes: the greater this distance, the smaller the amplitude contribution. The larger motor units (containing the larger diameter fast-twitch fibers) are preferentially recruited at high force levels according to the "size principle" (Henneman and Olson, 1965). Therefore, the relative location of the fast-twitch fibers within the muscle and with respect to the recording electrodes determines how the electrical signal from these motor units affects the surface EMG signal.

Although all three causalities described above require consideration when the EMG signal is compared to the force output, it remains difficult to document their intervention quantitatively. The fourth causality mentioned, that is, the motor unit recruitment and firing rate properties, has been documented more concretely. Information described in detail in Chapter 5 provides documentation that the firing rate and recruitment properties of relatively large and small muscles are distinctly different. The equations in Figure 3.11 show that such a distinction has the potential of expressing itself in the value of the parameters that are customarily used to measure the amplitude of the signal.

Oscillating Contractions

An alternative approach to studying the relationship between the EMG signal and the force output of a muscle during isometric contractions is monitoring the signal from the muscle while the force output of the muscle oscillates at a fixed frequency, and repeat the measurement at various frequencies. The modulation of the force output may be induced either by electrical stimulation of the muscle (Buchthal and Schmalbruch, 1970; Crochetière et al, 1967) or by voluntary intention (Gottlieb and Agarwal, 1971; Soechting and Roberts, 1975). The input-ouput relationship (transfer function) between the signal amplitude and the force output as a function of the frequency of the contraction oscillation may be represented as gain and phase functions. (See the discussion at the beginning of Chapter 2 for more details on transfer functions.)

The results of Soechting and Roberts (1975), who detected surface EMG signals, are presented in Figure 7.4. The continuous lines represent a simple transfer function which approximates the data. It is:

$$H(s) = \frac{k}{(s + 5\pi)^2} \qquad \text{where } s = j2\pi f$$

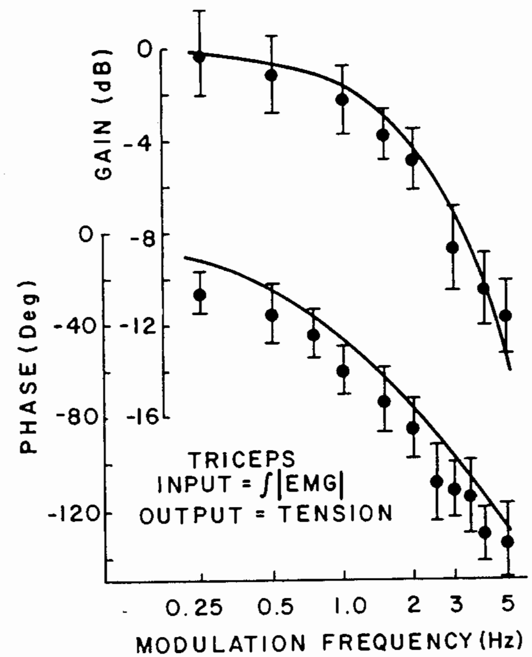

Figure 7.4. Gain and phase relationships between the integrated rectified amplitude of the EMG signal and isometric force in the biceps brachii. (From J.F. Soechting and W.J. Roberts, © 1975, *Journal of Physiology*.)

Although this transfer function does not provide the best fit to the data, it is nonetheless a useful representation of the data because it models the input-output relationship as a simple second order system with a 3 dB point at 2.5 Hz. The data of Figure 7.4 indicate that an increase in the modulation amplitude of the surface EMG signal is required to maintain the same amount of modulation of the output force as the frequency of the modulation increases. In other words, the electrical-mechanical transfer function behaves as a low-pass filter.

RELATIONSHIP DURING ANISOMETRIC CONTRACTIONS

Studies directed at describing the relationship between the signal amplitude and some form of mechanical output of the muscle such as position, velocity, and acceleration are susceptible to additional complications above those associated with isometric contractions. When the contraction is anisometric, that is, one which involves a change in the length of the muscle, the following additional factors must be addressed: (1) the modulation of the EMG signal induced by the relative movement of the electrode with the active fibers; (2) the force-length relationship of muscles; (3) the possible presence of reflex activity; (4) the change in the instantaneous center of rotation of a joint which will effect the moment (force × distance) of the tendon insertion.

A discussion on the complications involving relative movement of the active fibers and electrode is provided near the end of Chapter 2.

The effect of length on the force generation capability is well documented. This nonlinear, nonmonotonic relationship is presented in Figure 7.5. The force produced by the muscle consists of two components: the passive elastic force (curve 1) exerted by the elastic components of a muscle, and the excitation-force response (curve 2). The sum of the two components yields curve 3, which represents the force output of the muscle as a function of its length. Note that the maximal force is generated when the muscle is stretched to approximately 1.2–1.3 times its resting length. This increment of stretch often coincides with the length of the muscle in the relaxed position. It appears that the anatomical architecture of the musculoskeletal system is organized so as to benefit from the force-length characteristics of the neuromuscular system. The instantaneous center of rotation of a joint, that is, the point around which the net torque is zero, is not fixed in most joints. De Luca and Forrest (1973c) demonstrated that in the case of the shoulder joint the instantaneous center of rotation undergoes a considerable excursion as a function of limb abduction.

The involvement of some or possibly all of the above considerations is seen in the findings of Miwa and Matoba (1959). They found that during slow flexion, the biceps brachii is much more active at certain angles of the elbow; it reaches a peak of activity when the elbow is at 160° and falls rapidly to almost nil at 90°, and increases again at maximal flexion. In another study Miwa and Matoba (1963) found similar changes to occur in the muscles of the thigh.

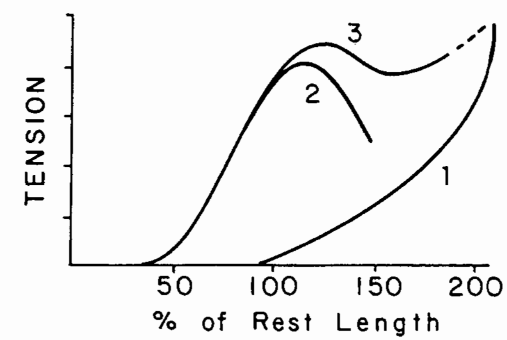

Figure 7.5. Force-length curves for an isolated muscle. Curve 1 is the passive elastic force of a muscle that is stretched. Curve 3 is the total force of a muscle contracting actively at different lengths. Curve 2 is the force developed by the contractile mechanism; it is obtained by subtracting curve 1 from curve 3.

The above discussion should serve as a cautionary warning to resist making facile interpretations concerning the amplitude of the EMG signal when the length of the muscle changes. It is alluring to use the EMG signal amplitude as a parameter to assess the quantitative involvement of the muscle during a functional movement. The literature abounds with reports of such attempts, notably in the area of gait measurements. Admittedly, there are many situations in which the EMG signal does have a useful role in the study of human gait, but in such cases the interpretation of the meaning of the amplitude of the signal should be executed wisely. Appropriate consideration should always be given to the underlying causes and interactions.

SUMMARY

This chapter has described the known behavior of the relationship between the amplitude of the EMG signal and the force output of the muscle. It has been emphasized that the characteristics of the amplitude of the signal can reflect the force output of the muscle, but it may also be affected by the technical details of the detection procedure and by physiological events occurring or originating in muscles not being monitored. Therefore, caution must always be considered when the amplitude of the signal is used as an indication of the muscular mechanical output, especially in contractions that are not isometric.

The relationship between the normalized EMG signal and the normalized force output of the muscle displays the following characteristics.

During a nontonically increasing isometric contraction:

1. There exists a considerable intersubject variation.
2. It is dependent on the muscle; it is quasilinear for the small muscles in the hand and nonlinear (amplitude increasing more than force) for the large muscles of limbs.
3. This distinction in the behavior may possibly reflect the difference in the firing rate and recuitment properties of small and large muscles, as well as other anatomical and electrical considerations.
4. It is independent of training and possibly fiber typing.
5. It is independent of the rate at which the contraction is generated, with the restriction that the contraction be nonballistic and nonfatiguing.

During oscillating isometric contractions the ratio of the force output to the amplitude of the EMG signal decreases as the frequency of the oscillation increases. During anisometric contractions the interpretation of the relationship between EMG signal amplitude and force requires considerable caution because numerous factors other than the force generated by the muscle may affect this relationship.