CHAPTER 6

Functional Muscle: Effects on Electromyographic Output

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Mark S. Redfern, PhD

INTRODUCTION

Electromyography (EMG) has been used in ergonomics to investigate such important topics as musculoskeletal injury, low-back pain, carpal tunnel syndrome, and muscle fatigue from overexertion. Significant progress has been made in understanding these topics by EMG analysis. Although modern instrumentation has facilitated easy acquisition of EMG data, many issues remain unresolved in the interpretation of EMG signals. The information presented here attempts to summarize what is currently known about EMG analysis with respect to muscle function. Material presented is based on a review of recent literature and is presented within the framework of surface EMG applied to the occupational setting. Because there still is controversy about many of the subjects discussed, extensive references are provided with each concept.

This chapter presents three major topics related to EMG: 1) temporal estimations of muscle activity, 2) muscle force estimations, and 3) muscle fatigue. The temporal estimation section discusses the use of surface recorded EMG in the investigations of the timing of whole muscle firing. The time delay found between the recorded surface EMG event and produced muscle force is discussed along with the effects of using an EMG preprocessor. The second section discusses the relationship between EMG and muscle force. The effects of muscle length, velocity of shortening, and cocontraction of synergistic and antagonistic muscles are presented. The relationships are explained in terms of both the empirical results and the physiologic basis. Because this use of EMG is the most common and in many ways the most complicated, a subsection containing specific recommendations for using EMG-muscle force relationships in applied EMG analysis is presented. The third section discusses muscle fatigue and its effect on the recorded EMG. Different types of spectral measurement techniques are discussed along with the limitations of each method.

TEMPORAL ASPECTS

EMG Muscle Force Timing

Electromyography has been used extensively to understand the temporal actions or timing of muscles during various types of exertions. The most basic information obtained is the onset and duration of myoelectric activity. This often is equated with the timing of produced muscle tension. In most cases where general muscle actions are of interest, this assumption is reasonable. In the occupational setting, timing patterns from EMG recordings can be useful. For example, gripping actions of the hand during manual activities are of interest in looking at various types of cumulative trauma disorders such as carpal tunnel syndrome.[1,2] Unfortunately, timing of grip forces during work is difficult to monitor because of varied hand movements and contact with objects. Electromyography has been used to estimate the timing of gripping forces during work.[1-3] The EMGs of the forearm are relatively easy to acquire, and they reflect the actions of the muscles that control grip. Although estimates of exact force magnitudes by EMG analysis during grip are limited, timing information can be used to better understand the gripping requirements of the job.

Differences do exist, however, between the temporal characteristics of the EMG and the produced tension. Although these differences may not be important for most general timing studies, they should be considered. The most apparent is the pure time delay. Ralston et al looked at the delay in the rectus femoris using raw EMG.[4] Time delays of 30 to 40 ms were found between the onset of the EMG and tension. Time delays of 200 to 300 ms occurred between the cessation of the electrical activity and tension. Redfern found similar results when looking at the triceps brachii during elbow extensions.[5] He found delays between EMG and force onsets from rest to be about 50 ms and cessation time delays of 180 to 220 ms. In recent studies, Komi et al have shown electromechanical delays using buckle transducers on the Achilles tendon.[6] Figure 6-1 demonstrates the relationship found between EMG recordings of the triceps surae and the Achilles tendon force. The tendon delay times at the onset of contraction were about 30 ms and the delays registered at the foot plate were about 70 ms. There appears to be, therefore, about a 40 ms difference between the tendon forces and the response of the foot that is measured externally. The exact delay times appear to be muscle dependent. In each case, however, the time delays at the onset are much shorter than the decay times at the cessation of EMG activity.

Initial tension levels in the muscle also have an effect on the delay times seen at onset. If the muscle is held at a baseline tension level before a step increase in force, then the delay time between the rise in EMG and the rise in force will be shorter. This is due primarily to the mechanical slack being taken out of the muscle. The amount of slack in the muscle also has a major effect on the relationship between the peak EMG and the peak force detected in a muscle. The rise time to attain tension levels during a contraction also is affected. It has been proposed that the electrochemical response of the muscle is increased when a baseline of force is used, although, in general, in applied surface EMG the importance of this effect is questionable.

EMG Processing Effects on Timing

In most EMG-muscle force investigations, the EMG signals are processed initially. Methods such as integration, root-mean-square (RMS), and Butterworth filtering have been proposed[7-9] and were discussed in Chapters 4 and 5. The temporal aspect of the EMG-force relationship is affected by the specific processing methodology applied. The primary factor is the low-pass properties associated with the filtering function used. In most cases, such as integration and RMS, the filtering function is an exponential window with some associated time constant. The time constant of the window is chosen generally by adjusting elements of the electrical circuits used (see Chapter 4). A longer time constant produces a smoother estimate of the electrical activity of the muscle. This is beneficial during static exertions where the electrical state of the muscle is stationary. During dynamic exertions, however, the response of the processor may be too slow to capture the changes occurring in the electrical state. Thus, the processed EMGs taken during rapid transitions will not reflect the dynamics of the electrical signals in the muscles but rather the dynamics of the processor itself. This different reflection is particularly true during investigations of ballistic activities. The choice of an appropriate time constant, therefore, is important for the type of activities under investigation. For isometric, static exertions, long time constants (usually $>$ 150 ms) that give smoother signals are desirable. For more dynamic studies, shorter time constants must be chosen ($<$ 60 ms).[10]

Another factor affecting the timing of processed EMG to muscle force is the processing method itself. The two most common processors, integration and RMS, have been shown to affect the temporal aspects of EMGs differently.[10] Additionally, the two common methods of estimating the RMS have different rise and fall times. Thus, not only the time constant but the dynamics of the specific processor are important considerations if temporal information is to be derived from processed EMG.

In general, muscle activation timing will be obtained more easily by the ergonomist by means of visual inspection of the original (nonprocessed) EMG data rather than through some processed quantity. If some estimation of magnitude of EMG activity is also desired, then a processing method should be used; however, the effect of that processing method on temporal aspects of the signal must be considered.

ELECTROMYOGRAM-FORCE

A great deal of confusion exists regarding the relationship between processed EMG and muscle force. Surface EMGs are complicated recordings of the electrochemical activations of muscle. Research has long suggested that the EMG could be used to represent the active control input of the muscle, and that some relationship must exist between the two. Some researchers have presented EMG as a direct indication of muscle force but others have presented very complex models using these signals to predict force. In choosing an appropriate model, it is important to realize that a number of factors influence the relationship between EMG and force. The kinematics of the movement, the processing methods used, and the acquisition procedure, all have an effect on the muscle force-EMG relationship. The following describes some of these factors and their influence on the EMG.

Isometric Response

The first investigations into the relationships between EMG and force were performed under isometric conditions. Lippold looked at the relationship between surface recorded integrated EMGs (IEMGs) and isometric tension, in the triceps surae.[11] This work showed a distinct linearity between IEMG and isometric tensions within subjects with coefficients of determination ($\underline{r}^2$) between .95 and .99. The slopes of these relationships were found to differ from subject to subject. Since that time, many others also have observed this linear relationship. Other investigators, however, have reported curvilinear relationships with the force-EMG slope decreasing at higher force levels. Table 6-1 presents a partial list of researchers who have reported these linear and curvilinear responses.

The discrepancies between these studies is disturbing. Explanations have been proposed and can be categorized as physiologic and experimental. On the experimental side, Moritani and deVries found that electrode configuration had an effect on the shape of the curve.[12,13] Bipolar recordings of the elbow flexors produced a curvilinear relationship but unipolar recordings had a linear result. A more physiologic rationale was presented by Bawa and Stein.[14] They showed in frequency response studies of isometric human soleus under controlled neural stimulation that the gain and phase fall off at rates higher than about 5 Hz. This implies a

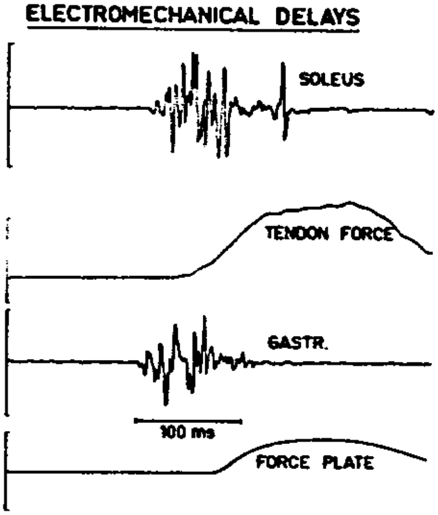

FIGURE 6-1

The temporal relationship between EMG and produced muscle tension in the human triceps surae from in vivo measurements of Achilles tendon forces.

Reprinted with permission from Komi PV, Salonen M, Jarvinen M, et al: In vivo registration of Achilles tendon forces in man: I. Methodological development. Int J Sports Med 8(Suppl 1):3-8, 1987.

nonlinearity between the isometric muscle force output and neural firing rates as they increase. These firing rates have a direct effect on the measured EMG and subsequent IEMG signals.

Woods and Bigland-Ritchie investigated the effects of muscle fiber composition of the surface EMG-force relationship in humans.[15] A variety of muscles with different fiber-type compositions and distributions were used. They found that the shape of the relationship was dependent on fiber composition. Uniform fiber composition led to linear relationships but mixed fiber compositions yielded nonlinear relationships. In other studies, Lawrence and DeLuca explored the RMS EMG-force relationships in three different muscles using different groups of subjects from weight lifters to pianists.[16] Figure 6-2 shows the results from their study. These graphs are of the biceps brachii, deltoid and the first dorsal interosseous muscles. They show the relationship between the RMS EMG normalized by the maximum value attained and force normalized to the maximum voluntary contraction (MVC). Note that the curves for the different types of subjects were very similar in shape, but the different muscles exhibited distinct differences. They concluded from the study that **the myoelectric signal-force relationship** was primarily determined by the muscle under investigation and was **generally independent of the subject group.** This conclusion is consistent with the findings of Woods and Bigland-Ritchie[15] and seems to be the general consensus among researchers in the area today.

If a curvilinear relationship is seen, the mathematical representation of the processed EMG-muscle force relationship should incorporate some second order term. Many methods have been used. The predominant mathematical expressions are either a second order polynomial or an exponential term, as follows:

Polynomial relationship:	$F = aE + bE^2$	(1)
Exponential relationship:	$F = Ae^{(qE)}$	(2)

where F is the muscle force and E is the processed EMG. In each case, two parameters must be estimated. Any higher order terms, such as a third order polynomial, usually provides no significant improvement in the fit of the relationships. The best form (polynomial or exponential) is debatable from both a theoretical or empirical viewpoint and probably is dependent on the exact muscle and instrumentation used. It is suggested that both relationships be tried during calibration of curvilinear data.

Length-Tension Effect

Another factor of importance to EMG studies in work environments is that muscle length has an effect on the force output of a muscle. The mechanics and physiologic basis of this relationship were discussed briefly in Chapter 2. The question to be addressed here is how does muscle length affect the EMG-force relationship? To answer this question, the ergonomist must first realize that the EMG-force relationship pertains to the active force producing capabilities. Hence, it is the effect of length change on the active components on muscle tension and not the passive properties that is of concern in this section.

As shown in Chapter 2, active tension is altered by changes in muscle length. One would expect the muscular response to neural stimulation rate (and, therefore, also to the EMG) to be affected. In their study, Rack and Westbury looked at the effect of length, stimulation level and isometric tension in the cat soleus.[17] The soleus was cut and attached to a force transducer. The nerve controlling the muscle was severed and then stimulated at various levels by an electrical pulse generator. The study found that at constant muscle lengths, the relationships between tension and stimulation level was similar to the EMG-tension curves recorded by other researchers. One major difference was the slight nonlinearity at very low

TABLE 6-1
A Partial List of Researchers Reporting Linear and Curvilinear Relationships Between Processed EMG and Muscle Force

Linear	Curvilinear
Inman et al[65]	Zuniga and Simons[69]
Lippold[11]	Komi and Buskirk[70]
Close et al[66]	Kuroda et al[71]
deJong and Freund[67]	Bouisset[28]
deVries[68]	Lawrence and DeLuca[16]
Woods and Bigland-Ritchie[15]	Woods and Bigland-Ritchie[15]

stimulation rates. When the stimulation rate was held constant, the length was found to have a profound effect on the tension. Figure 6-3 shows a graph of these results. Notice that the plots at different rates of stimulation are similar in shape, but are shifted with respect to muscle length. One conclusion from this data is that stimulation rate and muscle length are interrelated with regards to their effect of muscle tension.

Because the stimulation rate-tension relationship is affected by muscle length, one would expect the EMG, which is in some way a reflection of this stimulation, to also be affected. This, in fact, is the case. Grieves and Pheasant found a family of EMG-muscle length curves at different force levels for the gastrocnemius and soleus.[18] Vredenbregt and Rau examined the relationship between EMG, force, and muscle length in the biceps brachii. They showed that the slope between IEMG and the force varies with the position of the joint. Figure 6-4 shows these results as a series of curves for each joint position, reflecting changes in muscle length. Notice that these results are not linear but curvilinear. This result seems to be prevalent in the biceps brachii. In Figure 6-4 C, the data have been normalized by the maximum force exerted at each angle Fmax. Notice that this data fit one generalized curve. This normalization of muscle forces by the Fmax at each given angle provides a good way to present these relationships not only within subjects but also across subjects.

Velocity-Tension Effect

Dynamic muscular exertions can be divided into concentric (muscle shortening) and eccentric (muscle lengthening) contractions. During concentric contractions, the velocity of shortening affects the muscle tension produced. These results, discussed in Chapter 2, are summarized in a muscle equation relating the two variables, shown as the characteristic equation. The relationship, however, does not hold for eccentric contractions, thus producing a more complicated situation for the study evaluating functional activities commonly seen at the worksite. Although the effect of both concentric and eccentric velocity has been shown, few studies have been conducted on the EMG-force relationship during these movements.

In their classical study, Bigland and Lippold investigated the relationship between force, velocity, and the integrated EMG in humans.[20] From Lippold's experiences with isometric tension, he anticipated that IEMGs were a good representation of the stimulation rate of **in vivo** muscle.[11] Their subsequent studies showed, in the case of plantar flexion of the foot, that if the load is plotted as a function of velocity at constant EMG levels for shortening muscle, a classic force-velocity curve (Hill's characteristic equation) is seen. Actually, a family of these curves are found, one at each IEMG level. Figure 6-5 is representative of these findings. These results were later substantiated for the biceps brachii by Zahalak et al for forearm rotation.[21] Zahalak et al also showed that the shortening velocity curves at a given IEMG level could be fit by the characteristic equation of Hill.

The results for eccentric contractions are quite different. Asmussen found that the IEMG of muscle under concentric and eccentric contractions with the same tension levels were different: eccentric contractions evoked greater EMG levels than the concentric contractions.[22] The shapes of the eccentric velocity-tension curves at constant IEMG levels are not as well documented as those for the concentric contractions. Some contend that no statistical increase can be found.[21] Others have seen a slight increase in force as eccentric velocity is increased that is similar to the strength curves described in Chapter 2.[23] Komi found, for example, that for elbow flexion the IEMG-force relationship increases with eccentric contractions during maximal exertions (Figure 6-6). Maximal IEMGs were recorded from both the biceps brachii and the brachioradialis muscles at different velocities (both concentric and eccentric). The IEMG levels were fairly constant over all the trails. The force output, however, changed as a function of velocity as shown in Figure 6-6 by the dashed line. This indicates that the IEMG-force relationship is affected by both the direction and level of muscle velocity.

Postural Dynamics

In EMG analysis, as applied in ergonomics, the muscle lengths and velocities are controlled by the postures during the task. Muscles transmit their forces through insertions in the skeletal structure about joints, thus creating torques. It is quite common to see EMG signals related

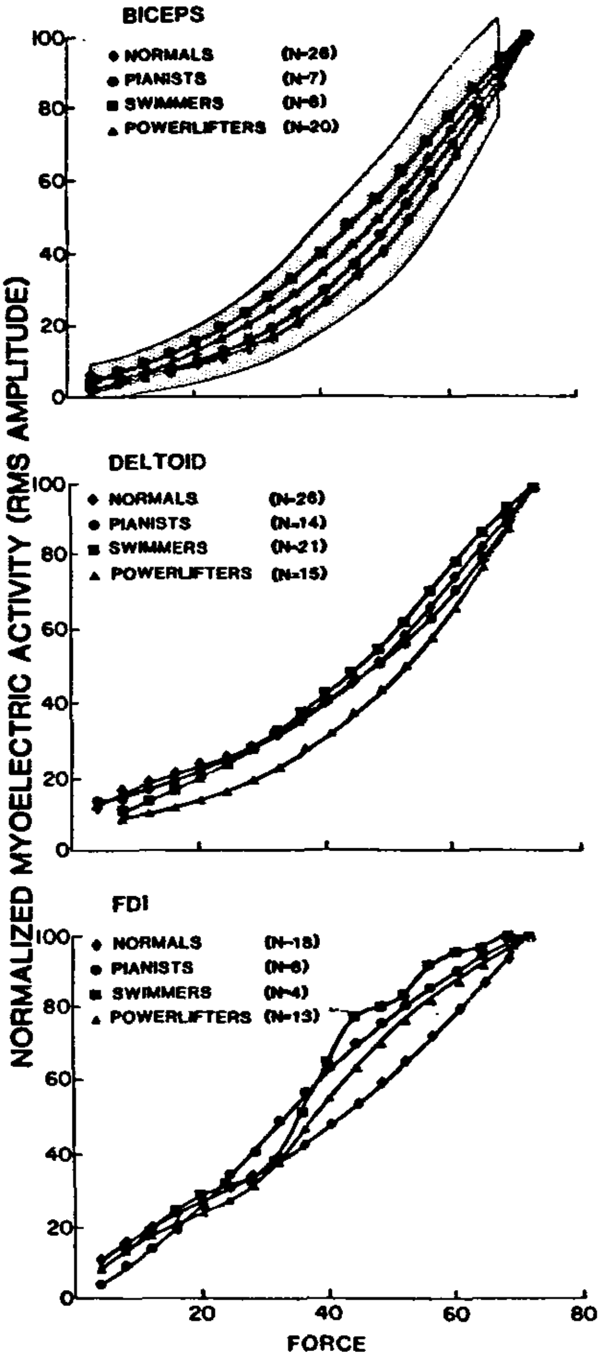

FIGURE 6-2

The RMS EMG-force relationship for three muscle groups of different subject populations. Standard deviations are indicated by shaded area in top graph.

Reprinted with permission from Lawrence JH, DeLuca CJ: Myoelectric signal versus force relationship in different human muscles. J Appl Physiol 54:1653-1659, 1983.

to the torques that are generated, instead of to the internal muscle forces.[24] These torques, then, are a function of the moment arm of the insertion and the angle at which the muscle is applying the force. The angle of pull and even the moment arm to the center of rotation usually change significantly during normal movements. Additionally, the angle of pull may vary within a muscle depending on the pennate structure of the particular muscle. It is very important, therefore, to recognize this dependence of EMG-torque relationships to the biomechanical factors inherent in the system.

Another common practice in EMG modeling is to substitute joint angular data for muscle length information when length and velocity compensations are used.[24] Joint angle is used in place of muscle length, and angular velocity is used for muscle shortening velocities. There are inherent problems with this type of analysis. In a study of the ankle, Redfern compared the use of joint angle data with muscle length and velocity.[25] He found that the relationship between ankle angle and estimated muscle length was nearly linear for the three muscles investigated (tibialis anterior, soleus, and gastrocnemius). A significant nonlinear difference was found, however, between angular velocity and estimated muscle shortening velocity. The geometric relationship between muscle length and ankle angle causes differences in the muscle shortening velocities up to 40%, under constant angular velocity conditions. This was found over a normal range of motion of the ankle. The discrepancy was found to be caused primarily by the insertion points to the axis of rotation. These results indicate that substituting ankle angle for muscle length will result in the force-length relationships differing from the force-angle relationship by a constant multiplicative factor. Thus the shape of the relationship will be preserved. If ankle angular velocity is substituted for muscle shortening velocity, however, major differences in the shape of relationships will occur. The force-angular velocity relationship will not appear as the classic Hill's relationship found on isolated muscle. Again, it becomes apparent that biomechanics of the musculoskeletal system must by considered by the ergonomist when doing dynamic EMG-force estimations.

Postural changes, therefore, affect muscle length, velocity of contraction, and biomechanical lines of action that all have been shown to affect EMG-muscle force relationships. Minimizing postural changes, therefore, is recommended whenever possible. For example, consider a study of low back EMGs during lifting. The preferred analysis (from an EMG standpoint) would be a static one, with consistent postures. Muscle lengths would be constant from trial to trial with no velocity of contraction. Obviously, static postures cannot always be used to answer pertinent questions. For dynamic studies, controlled postural changes would then be desired, with

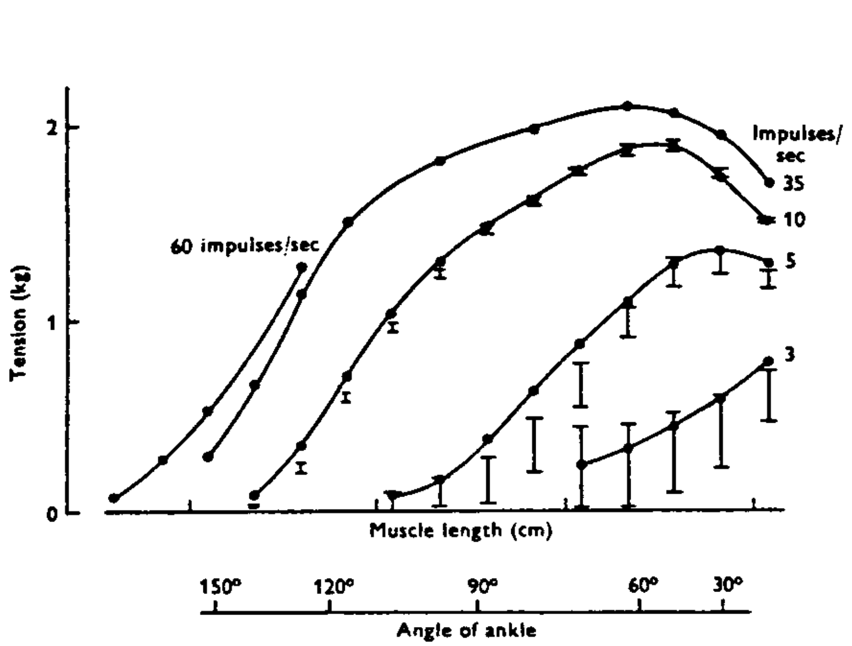

FIGURE 6-3
The effects of stimulation rate and muscle length on produced isometric tension.

Reprinted with permission from Rack PMH, Westbury DR: The effects of length and stimulus rate on tension in the isometric cat soleus muscle. J Physiol (London) 204:443-406, 1969.

the same lifting style and speed used throughout the study. Analysis of these dynamic EMGs could then be made at specific postures throughout the lift, keeping consistent the muscle length, velocity, and lines of action. In summary, the basic principle of EMG recording is that more control of postures means less variability. Some studies done in the workplace, however, produce uncontrollable postures. In this case, EMG analysis should be done with the effect of postural changes and muscle mechanics taken into consideration.

Cross Talk

Selectivity of the electrodes over a muscle group always is a consideration in the analysis of the recorded EMG. Cross talk from the electrical activity of other muscles picked up by the electrodes will cause some error in the analysis. This is true particularly if the signal is to be used to just measure temporal activation (whether the muscle is on or off). Cross talk is less of a problem, however, when the EMG is quantified. As pointed out by Hof, the intensity of the recorded EMG is not simply the sum of the primary muscle of interest and the secondary muscle causing cross talk noise.[26] Because the EMG emitted from each muscle is stochastic, the recorded signal is as follows:

$$E = (E_a^2 + E_b^2)^{1/2} \tag{3}$$

where E is the signal recorded from the electrodes, E_a is the EMG from the primary muscle and E_b is the EMG from the cross talk muscle, so, for example, if E_b is 50% of E_a then $E = 1.12\ E_a$. In reality, E_b is likely to be much less than 50% of E_a, thereby further reducing the cross talk effects. Concerning the importance of cross talk, Hof states the following:

> The results of crosstalk is thus a **noisy baseline,** which at times may suggest a slight activity while there is none, but which does not seriously affect the quantitative interpretation at higher force levels. The possibility of cross talk should be considered seriously, nevertheless, when choosing the electrode location: the ratio wanted/unwanted signal is what matters, not so much the signal strength itself.[26]

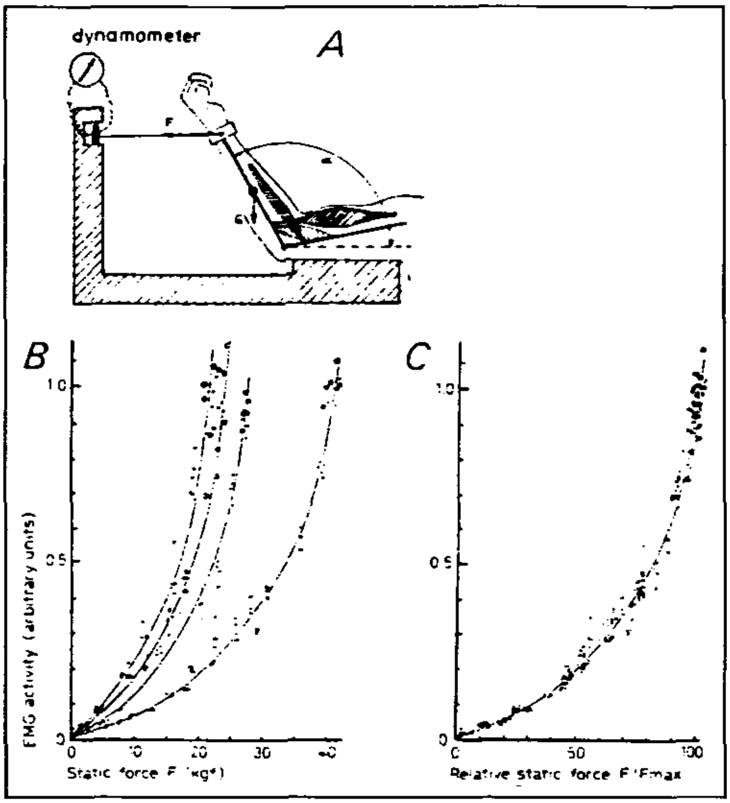

FIGURE 6-4

The relationship between IEMG, muscle length, and tension in the biceps brachii. A. Measurement diagram. B. The force (F) exerted under static conditions in relation to the measured EMG level. The different curves were obtained at different muscle lengths. C. The relative force F/Fmax in relation to the EMG.

Reprinted with permission from Vredenbregt J, Rau G: Surface electromyography in relation to force, muscle length and endurance. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1.

Cocontraction Effects

A concern with any study of EMG-force relationships in an ergonomic center is the cocontraction of muscles around the joint under investigation. Both synergistic and antagonistic muscles can have dramatic effects on the results. Cnockaert et al recognized the importance of including synergistic muscles in determining muscle forces from IEMGs during elbow flexion.[27] They proposed a model that related IEMG to muscle torque for the biceps brachii and brachioradialis muscles. These synergistic muscles have been assumed by many to be one "equivalent flexor muscle."[28] The EMGs of one muscle could then be used to describe the actions of both. The results of Cnockaert et al, however, indicated that the two muscles can behave quite differently under different conditions.[27] Thus, the equivalent muscle concept was extremely limited. Falconer and Winter used EMGs to develop an isometric model that estimated the relative cocontraction between the antagonistic muscles, soleus, and tibialis anterior, acting about the ankle during gait.[29] The model produced a measure of the relative cocontraction (termed the cocontraction index) between the two muscles. It could not estimate, however, the torque contributions by the muscle groups. Although this model was limited, it did show that significant levels of cocontraction occur about the ankle during gait and must be taken into consideration. Most other investigations into

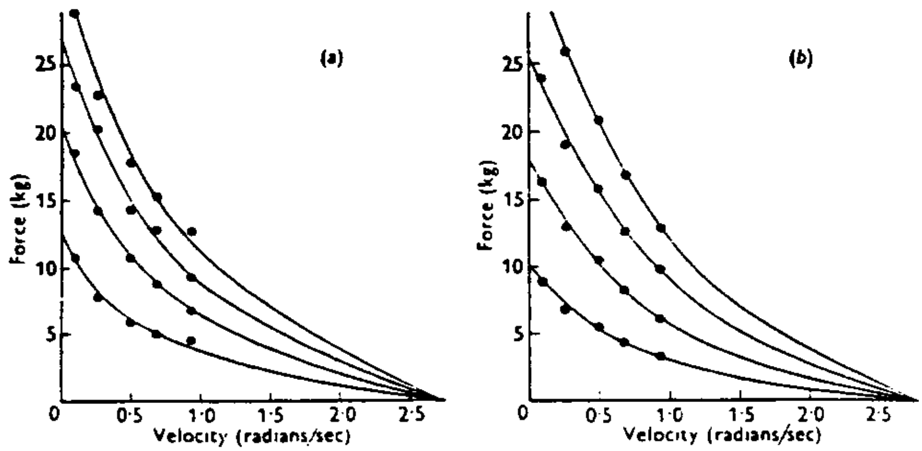

FIGURE 6-5

Force-velocity curves plotted at four different levels of excitation as measured with IEMGs.

Reprinted with permission from Bigland B, Lippold OCJ: The relationship between force, velocity and integrated electrical activity in human muscles. J Physiol (London) 123:214-224, 1954.

processed EMG-muscle force relationships have assumed that no antagonistic or synergistic activity occurs during isometric contractions. If this assumption is invalid, the resulting estimations of the EMG-muscle force relationships would be affected erroneously. Furthermore, any changes in the amount of antagonistic and synergistic muscle activity between trials of an ergonomic study would have an even greater effect, causing increased variance in any calculated model parameters. This error would occur even if the processed EMG-muscle force relationships for the individual muscles were totally stationary.

Redfern and Chaffin examined the isometric torque-processed EMG relationships of the soleus, gastrocnemius, and tibialis anterior during both plantar flexion and dorsiflexion.[25,30] They proposed a method for calibrating the EMG-torque relationships of the three muscles while taking their cocontractive nature into consideration. From their results, they showed significant synergistic and antagonistic activity during torque production in both the plantar and the dorsal directions. Figure 6-7 demonstrates these findings. Figure 6-7 b shows the individual components of torque about the ankle created by each muscle. They were predicted from the EMG data recorded. Figure 6-7 a is a comparison of the measured resultant torque (solid line) and the predicted resultant torque (dashed line). The predicted resultant torque was estimated by summing the three components shown in Figure 6-7 b. These results indicate that significant antagonistic and synergistic activities occur, even during simple isometric torque production. Redfern and Chaffin estimated that a 15% increase in the slopes of the EMG-torque calibrations occurred for the muscles acting on the ankle when cocontraction was taken into consideration instead of assuming no antagonistic activities.[25] Similar adjustments may need to be made when studies incorporating ergonomics are performed.

Ballistic Muscle Actions

Even the most sophisticated dynamic EMG-force models currently are limited to controlled movements. Ballistic activities in particular are not well described by today's modeling techniques. In these types of movements, there are high levels of interaction between the geometry of the muscle, the material properties of the musculotendinous structures, the actin-myosin cross bridge activity, and the electrode-muscle membrane tissue interface. What generally is seen is a large initial burst in the EMG signal during the onset of a ballistic movement that is not reflected in the force output. This large burst often gives large over-estimates of the muscle force from EMG-force models. To overcome this problem, the previously mentioned factors such as length-

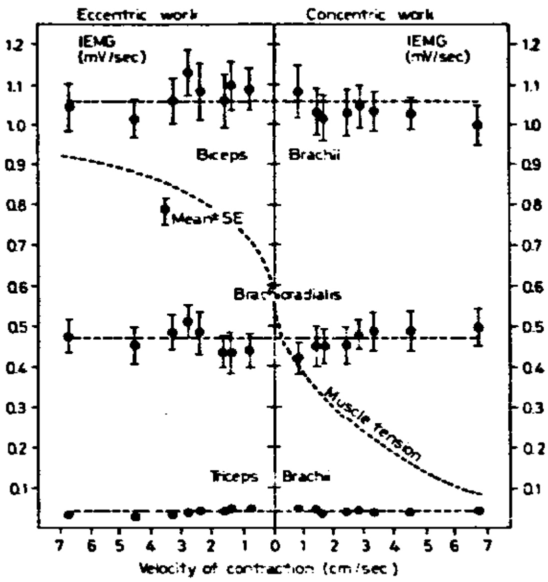

FIGURE 6-6

The relationship between integrated EMG (IEMG) and the velocity of contraction for maximal exertions.

Reprinted with permission from Komi PV: Relationship between muscle tension, EMG and velocity of contraction under concentric and eccentric work. In In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1, pp 596-606.

tension and velocity effects will have to be incorporated with sufficient models of the passive properties of the musculotendinous system. This is pointed out briefly in this chapter, to caution any investigator interested in using EMGs in the quantitative analysis of ballistic activities.

Recommendations

The following are specific recommendations for EMG analysis where force levels are to be estimated by ergonomists. They are based on the factors discussed in this and previous chapters.

1. **Let electrodes stabilize:** The impedance of the electrode-skin interface has a direct effect on the EMG-force relationship. As the electrode paste dissolves into the layers of the skin, the impedance will be reduced. It is important to allow this effect to stabilize. Allow about 20 minutes for this stabilization to occur.
2. **Calibrate every experiment:** Calibrations must be performed after each application of new electrodes. This is because the impedance of the electrode-tissue system will change.
3. **Minimize postural changes:** Because length and velocity have such large effects, try to keep EMG-force relationships to static postures if at all possible. Calibrate the models at the exact postures under investigation.
4. **Consider cocontraction of muscles:** Cocontraction of antagonistic and synergistic muscles always occurs to some degree. This fact should be taken into consideration during calibration of any EMG-force model. Any test must address this problem by either incorporating the other muscles or by making some assumptions and understanding the possible error they incur.
5. **Electromechanical modeling:** There is an electromechanical response time between the EMG and the force output. If dynamically changing forces are to be observed, this factor must be considered.
6. **Minimize fatigue:** Muscle fatigue will change the EMG-force relationship. It is necessary, therefore, to design experiments that will minimize any localized muscle fatigue. Rest periods between exertions should also be used to allow the muscle to recover.

FATIGUE

Fatigue caused from environmental and job related stresses continues to be a major concern in the workplace. The term fatigue, however, is not easily defined and therefore is difficult to measure. One type of fatigue is systemic, affecting the person as a whole. Examples of contributing factors involved in systemic fatigue would be high levels of heat or cold, aerobic requirements, lactic acid, or even psychological stress (see Edwards and Lippold,[31] Hermans et al,[32] and Bouisset,[28] for reviews of these factors). Fatigue also occurs on a local level within the body. This is true particularly in musculoskeletal exertions. Typical externally visible symptoms are loss of force production capabilities, localized discomfort and pain. This type of fatigue has become known as "localized muscle fatigue" (LMF).[33] Localized muscle fatigue continues to be of concern in ergonomic assessment of jobs. Muscle exertion levels do not necessarily need to be high to cause LMF. Isometric contractions of as low as 10% of MVC have shown signs of LMF. Much higher contraction levels are common in the workplace,

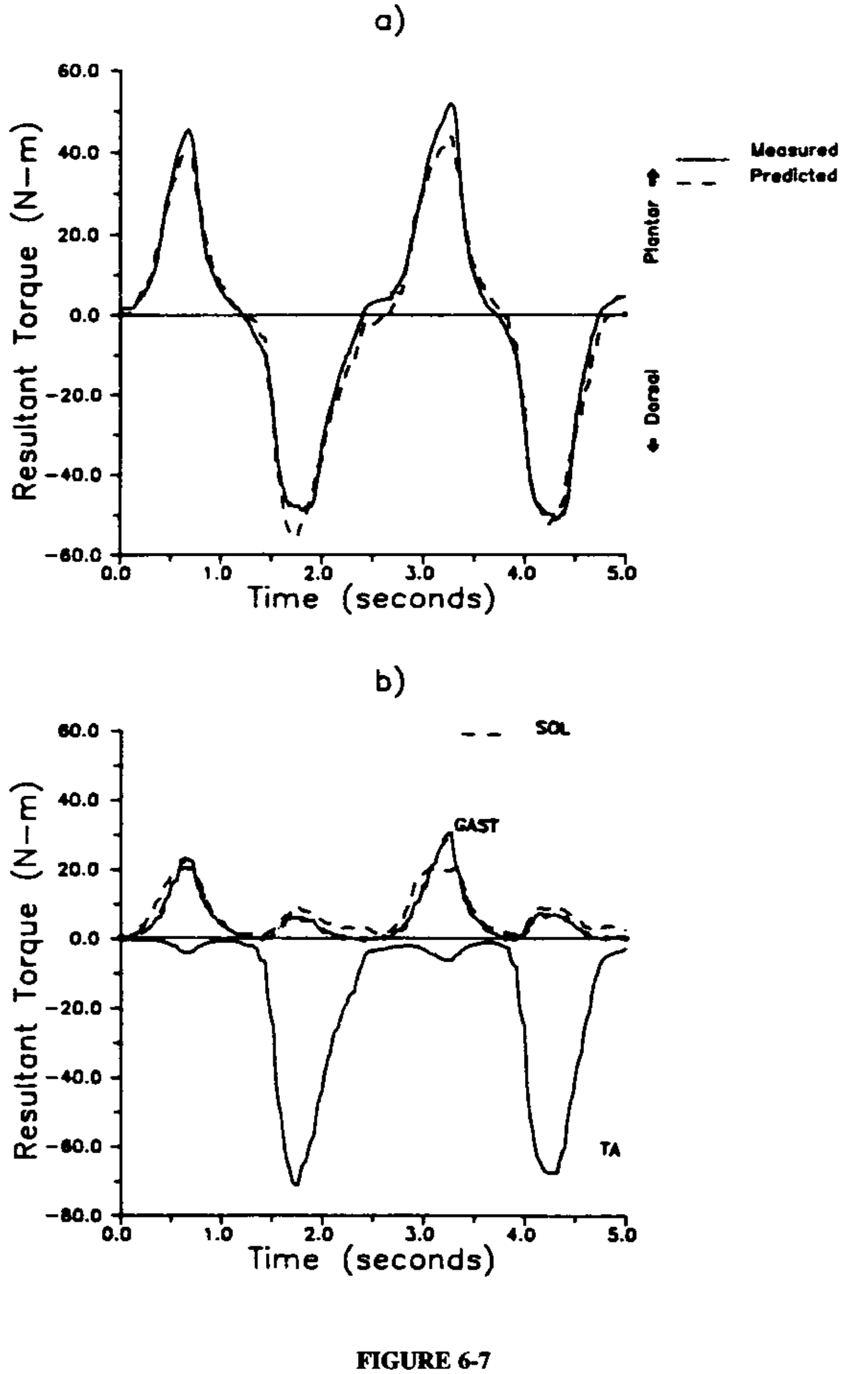

FIGURE 6-7

Expected value prediction results from a slow sinusoidal trial with large knee angle: a) comparison of predicted versus measured resultant ankle torques; b) predicted torque contributions from the tibialis (TA), soleus (SOL), and gastrocnemius (GAST).

Reprinted with permission from Redfern MS, Chaffin DB: Modeling EMG-torque relationships of muscle groups around the ankle joint considering co-contraction. Submitted to J Biomech 1988.

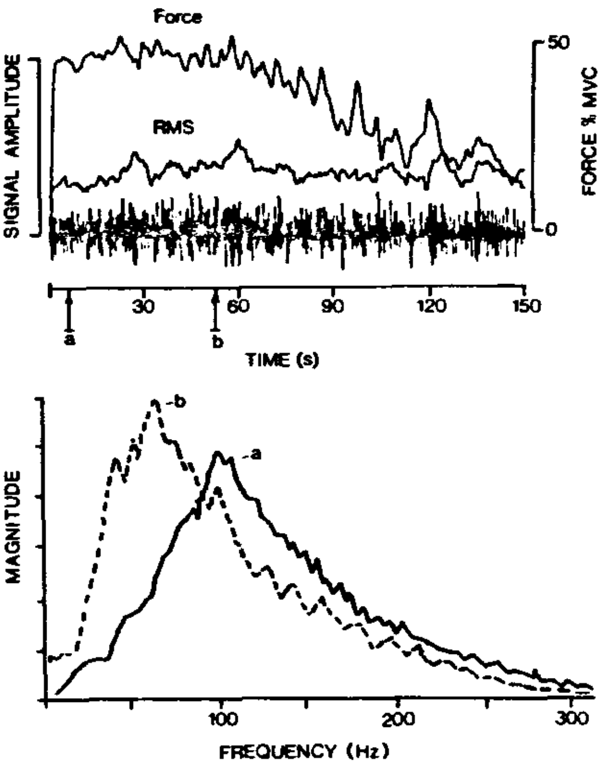

FIGURE 6-8

(Top) Amplitude changes and (bottom) frequency spectral shifts resulting from local muscle fatigue during a sustained, isometric contraction of the first dorsal interosseous muscle.

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and LMF has certainly been seen under these circumstances. Great interest exists in finding an objective measure of LMF that can be used in job evaluation and design. Electromyographic analysis has been proposed as one method of evaluating LMF during repeated or locally stressful activities.

Localized Muscle Fatigue and the EMG Signal

During LMF, changes occur in the surface recorded EMG signal. Two of the most commonly cited changes are a shift in the frequency content of the signal toward the low end and an increase in the amplitude. Figure 6-8 graphically demonstrates this effect during sustained, isometric contraction. Note that the force level decreases for a given EMG level and the power spectrum shifts from time a to time b. Lindstrom[34] and DeLuca[35] contend that these two phenomena are related. They state that tissue filtering characteristics act as a low pass filter. As the frequency content of the signal shifts to the lower frequencies, more energy is transferred through the tissues to the electrodes. This energy transfer, in turn, increases the amplitude of the recorded signal.

TABLE 6-2

Proposed Physiologic Causes and References for Spectral Shifts and Amplitude Changes of the EMG During Fatigue

Motor unit recruitment	Edwards and Lippold[31] Milner-Brown et al[72] Clamann and Broecker[73] Maton[74]
Motor unit synchronization	Lippold et al[75] Milner-Brown et al[72] Chaffin[76] Bigland-Ritchie et al[37]
Firing rate and interpulse intervals	DeLuca and Forrest[77] Hogan[78]
Motor unit and action potential shape	Lindstrom[34] Broman[79] Kranz et al[38] Mills[45]

Physiologic explanations for the changes in amplitude and spectral characteristics have been proposed by many researchers. Some of the factors believed to be involved are presented in Table 6-2. Motor unit recruitment is believed by some to occur as a muscle becomes fatigued in response to reduced muscle contractility. This increased recruitment would have the desired effects on the amplitude and spectrum of the signal. Synchronization of the recruited motor unit activations has also been observed and proposed as a mechanism for changes in the EMG signal. As this synchronization occurs, an increase in the low frequency content would be expected. There is not total agreement, however, on the influence of motor unit recruitment and synchronization. Basmajian and DeLuca contend that although these phenomena may occur, it is doubtful that either is a major factor in EMG spectral shifts during constant contractions.[36] They propose factors such as firing rate, variations in interpulse intervals, and changes in the shapes of the motor unit actions potentials are the predominant cause of spectral shifts. Conduction velocity along the muscle membranes, which has a direct effect on these factors, also is believed to be important.[34,37] Kranz et al proposed that the change in the conduction velocity accounts for the majority of the shift in the spectrum.[38] In a study of conduction velocity and spectral factors during contractions, Broman et al stated "no single process appears to account for the effects of a high force level contraction on the myoelectric signal. In addition to myoelectric conduction velocity decrease, changes in the firing patterns of the active motor units are suggested."[39]

Shifts in the spectrum are reversible when the muscle has rested. The amount of time required for the spectrum to recover after the cessation of an exertion depends on the type and duration of the exercise performed. For short duration, low level loading, recovery appears to occur within 2 to 5 minutes of rest.[40–42] Mortimer et al showed that conduction velocity also recovered about 2 minutes after exercise.[43] For longer or more strenuous exertions, however, the spectrum will not recover for hours.[44] Spectral recovery does not appear to correspond with mechanical or physiological recovery of the muscle, which may take much longer.[45–47] This brings into question the use of spectral recovery rates to monitor fatigue recovery.

Specific Measures of Spectral Shifts

Despite the continued controversy over the underlying causation of spectral shifts, monitoring these changes in the EMG signal continues to be used to assess the state of the muscle during repeated or constant contraction. Many different metrics have been used to describe the shift in the power spectra. Two of the more popular techniques are presented in Table 6-3.

These are a ratio of high to low frequencies (HLR) and the median power frequency (MPF). Other methods have been used such as peak frequency[48] and zero crossings.[49,50] Although these other measures have shown correlations with changes in the spectrum, their applicability and generality in applied EMG are not as accepted as the MPF and HLR methods. The rest of this section concentrates on the MPF and HLR measurement methods and the results found with each.

Ratio of High to Low Frequencies

Spectral shifts of the EMG during fatigue can be thought of as a decrease in the high frequency band and an increase in the low frequency band. A ratio (HLR) of the power in these two bands then would show the relative shift of the spectrum (assuming the total power remains constant over time). As an example, Bigland-Ritchie et al formed the ratio of the low band power (20-40 Hz) to high band power (130-238 Hz) as HLR.[37] One major advantage of this method is the speed and ease of measurement. This measure shows significant reduction

TABLE 6-3

Selected References on Reported Techniques to Monitor the Shift in the Power Spectrum of the EMG Signal During Muscular Contractions

Ratio of High to Low Frequencies	**Median or Mean Power Frequency**
Ortengren[60]	Herberts et al[80]
Gross et al[55]	Lindstrom[34]
Bigland-Ritchie, et al[37]	Lindstrom et al[81]
Kramer et al[44]	Petrofsky and Lind[82]
	Hagberg[62]
	Hagberg and Ericson[58]
	Baidya and Stevenson[83]

over time during a fatiguing contraction as predicted in Figure 6-9. Others have proposed other HLR methods that define the bands in different ways with varying degrees of success (see Table 6-3 for references).

Although the HLR does correlate well with muscle fatigue, problems exist with this measure of the EMG shift. The ratio is sensitive not only to the power shifts in the spectrum, but also to its shape.[1] These shapes can change as a result of differences in the muscles tested, intersubject variations, muscle length, and other factors.[52,53] Another concern is the dependence of the results on the choice of the frequencies that divide the spectrum in the bands. This method makes any standardization or interpretation between studies difficult.

Median Power Frequency

The MPF (sometimes known as the center frequency) method of analysis is the most widely used measure of spectral shift resulting from fatigue. The MPF is defined as that frequency about which the power is distributed equally above and below. It is calculated as any median of a distribution. Another spectral measure commonly used is the mean power frequency. Both the mean and the median give similar estimations for spectral shifts. In this chapter, discussions of the MPF will refer to either the mean or the median power frequency. Both measures have been found to decrease over time, sometimes as much as 50% during prolonged isometric contractions (see Table 6-3 for references). Some researchers have also seen reliable decreases in the MPF during dynamic contractions,[54,55] although others have reported conflicting results. The rate at which the shift in the MPF occurs over time is dependent on the level of the contraction. Figure 6-10 demonstrates this change in the MPF. The higher the tension level exerted, the faster the MPF shifts to lower frequencies. This is to be expected because muscles under greater tension will

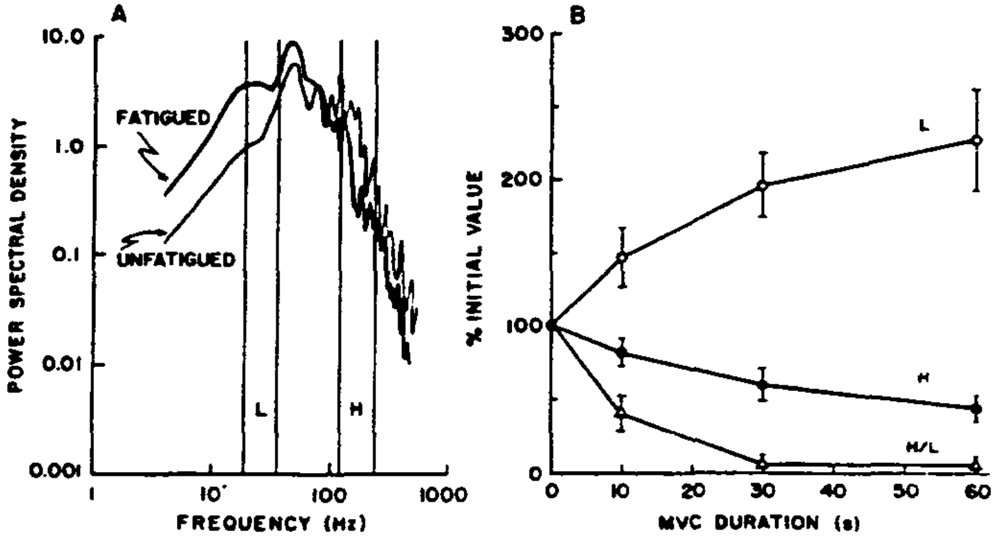

FIGURE 6-9

1A. EMG power spectra obtained before and after 60 seconds of fatigue at room temperature. B. Changes in the mean values H (130–238 Hz), L(20–40 Hz), H/L (± SD) during 60 second sustained maximal contractions.

Reprinted with permission from Bigland-Ritchie B, Donovan EF, Roussos CS: Conduction velocity and EMG power spectrum changes in fatigue of sustained maximal efforts. J Appl Physiol 51:1300-1305, 1981.

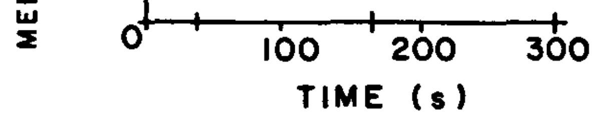

FIGURE 6-10

The affect of tension level on the rate of change of the median power frequency.

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fatigue more rapidly. The total decrease is dependent on the muscle under investigation and on some subject variability. When a muscle is allowed to rest after the end of a contraction, the MPF can return to the higher level. This is not true, however, if the muscle has been fatigued to exhaustion.

Although the rate of change of the MPF is affected by the tension, there is no consensus on the effects of muscle tension levels on the initial MPF. Some researchers have found that in healthy subjects the frequency spectrum is almost independent of the exerted force.[19,32,56,57] There is evidence, however, that tension level can have an effect. Hagberg and Ericson found that the MPF increased with contraction strength for low level contractions.[58] At higher levels in excess of 25% MVC, the MPF became independent of contraction level. They surmised that this dependence at lower contraction levels is due to tissue filtering effects and recruitment of motor units. Moritani and Muro, however, found that the initial MPF increased significantly from 0% through 80% MVC for the biceps brachii.[12] Others have also seen this increase in frequency content with load.[39,59]

Limitations

Although the use of EMGs in measuring or monitoring LMF is well established and frequently used, the technique is not without limitations. It is important to understand some of these limitations before undertaking an EMG analysis in the field of ergonomics. The first problem is in the basic definition of fatigue. Any method used to measure fatigue, including EMG analysis, is based on a basic definition of the problem. Because there is no universal definition of fatigue, agreement on the validity and meaning of EMG measures will be questioned. Other factors in LMF such as pain tolerance, motivation, and synergistic accommodation are not included in EMG analyses and have been argued to be important. Additionally, spectral shifts have been used for short term contraction fatigue, but the use of EMGs in long term fatigue is questioned. For muscle fatigue that occurs over a longer period of time, perhaps hours, the use of EMGs has not been well established.

Besides definition limitations, other problems more methodological in nature occur. Shifts in the various EMG indices have been established for isometric constant contractions, but the timing of these changes may not be synchronized with changes in the state of the muscle. The indices have been shown to decrease rapidly during the initial stages of a contraction, but do not decay as rapidly toward the end of a long session of work.[60] So, although the shift may indicate LMF, it does not follow the degree of fatigue experienced over time.

Another limitation is the difficulty of using EMG analysis around complex joints. In multiple muscle studies, particularly at the shoulder, spectral shifts have been seen during fatigue in some muscles and not in others. This has occurred even though one would expect the unaffected muscles to have equal or greater imposed stresses.[56,61-63] Finally, it appears from the data of Okada that the shape of the frequency spectrum is affected by muscle length.[64] The spectrum shifts to lower frequencies as the muscle length is increased. Postural changes, therefore, must be mobilized if spectral shifts resulting from fatigue and not length changes are to be examined. In general, therefore, the indices of LMF taken from EMG analysis appear to be quite reliable for constant isometric contractions of greater than 10% of MVC. The physiologic cause of EMG changes appear to be related directly to the causes of LMF, which makes it a reasonable measure. There are unresolved questions, however, as to the scope of applicability of EMG analysis in measuring worker fatigue in other dynamic or low level exertion environments. It is within the limitations presented that EMG analysis should be used to investigate fatigue.

SUMMARY

In this chapter, the relationship between EMG and muscle function is discussed. The topics presented are temporal estimation of muscle activity, muscle force estimation, and localized muscle fatigue. The most basic use of EMGs is the temporal estimation of muscle activity. Surface EMGs are a very effective method obtaining a general idea of when a muscle is active. There are differences, however, between the temporal characteristics of the EMG and the produced tension. Time delays between EMG onset and muscle force generation are 30 to 50 ms but delays of 200 to 300 ms occur between the cessation of EMG and muscle force. The effect of EMG processing on these delays also is discussed. For most work in ergonomics, these differences are not important. Researchers, however, should be aware of these differences, particularly in investigations of highly dynamic activities.

Muscle force estimations from EMG recordings often are desired. The relationship between EMG and muscle force is dependent on factors such as muscle length, velocity, cross talk between electrodes, and cocontractions of both synergistic and antagonistic muscles. In controlled, isometric contractions, the relationship between processed EMG (usually integration or RMS) and muscle force has been reported as both linear and curvilinear. The shape of the relationship appears to be dependent on the muscle studied and possibly the type of electrodes used. Models of the curvilinear relationships have included second order polynomials and exponentials. Muscle length and velocity dramatically changs the

EMG-force relationship found in isometric contractions. Methods to compensate for these factors have been attempted with varying successes; they are presented. Cocontraction of synergistic and antagonistic muscles also affect the estimation of the EMG-force relationship. It is often assumed that synergists act as one equivalent muscle and that antagonists are not active. These assumptions are often incorrect and lead to significant errors in force estimations.

Localized muscle fatigue has been associated with change in the EMG. The amplitude of the EMG increases for a given force level and the frequency spectrum shifts to lower frequencies. Two often used measures of the spectral changes are the median power frequency (MPF) and a ratio of high to low frequencies (HLR). The cause of these changes have been attributed to motor unit recruitment, firing rate, synchronization, and action potential shape. Limitations of using spectral shifts to monitor muscle fatigue are discussed.

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