CHAPTER 5

Output Forms: Data Analysis and Applications

Barney LeVeau, Ph.D., PT
Gunnar B.J. Andersson, M.D., Ph.D.

Barney LeVeau, Ph.D., PT, received a certificate in physical therapy from the Mayo Clinic School of Physical Therapy and a Ph.D. degree in Biomechanics from The Pennsylvania State University. He has taught courses at West Chester State College, University of North Carolina, and The University of Texas Southwestern Medical Center at Dallas. He has developed graduate courses in Analysis of Human Motion and EMG Kinesiology, and has performed several EMG research studies. He is the author of the text, *Biomechanics of Human Motion.* He serves as a manuscript reviewer of *Physical Therapy, Journal of Orthopedic and Sports Physical Therapy,* and *Biofeedback and Self-Regulation.* He is presently Professor and Chairman at The University of Texas Southwestern Medical Center at Dallas, Department of Physical Therapy.

Gunnar B.J. Andersson, M.D., Ph.D., is Professor and Chairman at the Department of Orthopedic Surgery, Rush-Presbyterian-St. Luke's Medical Center, Chicago, Illinois. He received his M.D. from the University of Gothenburg in 1967 and a Ph.D. in Medical Sciences from the same University in 1974. He moved to Chicago in 1985. His primary interest is in spinal conditions and his research focuses on epidemiology and occupational biomechanics. He is the author of more than 150 original papers, several textbooks, more than 70 book chapters, and is a member of several national and international societies. He is a past president of the International Society for the Study of the Lumbar Spine, and the International Society of Electrophysiologic Kinesiology. He is on the editorial board of *Spine, Journal of Biomechanics, The Back Letter, Journal of Back and Musculoskeletal Rehabilitation,* and *American Journal of Neuroradiology.*

OUTPUT FORMS: DATA ANALYSIS AND APPLICATIONS
Interpretation of the Electromyographic Signal

Barney LeVeau, PhD, PT
Gunnar Andersson, MD, PhD

INTRODUCTION

The electromyogram (EMG), based on changes in amplitude and frequency, can be quantified and used to classify the electrical activity level that produces a certain muscular tension. The change in the myoelectric signal is based on the recruitment and firing rate of motor units within the muscle. In general, as more force is needed, more motor units are recruited, and the motor units already firing increase their frequency of firing. This general reaction, however, is not exactly the same for every muscle.[1] The interpretation of the changes in recruitment and changes in firing rate can provide information concerning the muscle's level of force or its level of fatigue. The information in this chapter presents a variety of ways by which the ergonomist may analyze or subsequently interpret myoelectric activity.

NORMALIZATION
Definition

Quantification of the myoelectric signal, although not the goal, is done so that comparisons may be made among muscles, individuals, and activities. The myoelectric signal amplitude is used as an indirect measure of contraction-force. Because there is not a one-to-one relationship between the two, a standard of reference must be established for any comparison among subjects, muscles, or activities. Such a process if referred to as normalization. This process also is a form of force calibration.

The myoelectric signal may change from one time to the next for several reasons such as slight change in electrode location, change in tissue properties, or change in tissue temperature. The absolute values of microvolts could give an inaccurate comparison of muscle function during different activities. Therefore, a normalization procedure must be made at each specific testing time for each subject tested.

After applying the electrodes at an appropriate site one the muscle, one or several contractions are performed for each muscle to be studied. The reference contractions must be well defined in terms of electrode placement, type of contraction (eg, extension or flexion), and joint position. In ergonomic studies, maximum efforts during functional activities may also be used. An example of this reporting is shown in Table 5-1, based on work done by Ericson and associates.[2]

Isometric Maximal Voluntary Contraction

The most common method of normalization is to perform one reference contraction, usually an isometric maximal voluntary contraction (MVC or MVIC). The myoelectric values subsequently obtained are expressed as a percentage of the MVC. Examples of this method are shown in Tables 5-1 and 5-2 and Figures 5-1, 5-2, 5-3, and 5-4.[3-6]

The use of the MVC as a reference contraction is based on the idea that the amount of force produced varies directly with the myoelectric output. This is not quite true, although many researchers have found a linear or near linear relationship between the myoelectric signal and the force produced.[4,7-13] Although the MVC may vary from time to time in quantity and quality, Viitasalo and Komi state that using the MVC "may be an acceptable way to standardize" the testing situations.[14]

Caution should be taken, however, in using the isometric MVC for all investigations. Several factors should be considered when selecting a reference contraction. These include the fact that the EMG-force relationship does not appear to be linear over the entire force range and that the relationship varies among subjects and muscles (see Chapter 6).

The motor unit recruitment pattern for each muscle is also known to be different.[1,15] Woods and Bigland-Ritchies found that muscles with near uniform fiber type composition had a linear relationship between EMG and force, but muscles with mixed fiber type composition had nonlinear relationships.[4] Lawrence and DeLuca, for example, found that the first dorsal interosseous muscle had a linear EMG-force relationship, but the biceps brachii and deltoid had a nonlinear relationship.[1]

In general, the number of active motor units increases with increasing force at low force levels, but the firing rate increases at higher force levels.[10] Slow motor units tend to become active later and continue to fire at higher force levels.

Researchers have found that comparing a subject to themselves is more precise than comparison across individuals. For the same muscle, the EMG-force relationship demonstrated small intrasubject variation but large intersubject variation.[1,16] Comparisons made on the same subject therefore, are more valid.

TABLE 5-1

The Muscles Investigated, The Approximate Position of the Bipolar Electrodes, The Attempted Movement (Isometric) and Joint Position at which the EMG Normalization Was Performed[a,b]

Muscle	Electrode position	EMG-Normalization type of isometric contraction	Joint position
Gluteus maximus	20% of d between spinous process S2 and a point 10 cm distal to greater trochanter	Hip extension in exercise table	45 deg. hip flexion
Gluteus medius	10 cm distally on a line from gluteus medius insertion towards greater trochanter	Leg abduction against manual resistance lying on the floor	Mid hip joint pos.
Rectus femoris	50% of d between SIAS and apex of patella	Knee extension in exercise table	45 deg. knee flexion
Vastus medialis	20% of d between SIAS and medial knee joint space	Knee extension in exercise table	45 deg. knee flexion
Vastus lateralis	25% of d between SIAS and lateral knee joint space	Knee extension in exercise table	45 deg. knee flexion
Biceps femoris	50% of d between ischial tuberosity and caput fibulae	Knee flexion in exercise table	45 deg. knee flexion
Medial hamstring	50% of d between ischial tuberosity and the medial knee joint space	Knee flexion in exercise table	45 deg. knee flexion
Gastrocnemius medialis	35% of d between medial knee joint space and tuberosity of calcaneous	Ankle plantar flexion standing on floor rising against manual resistance	Mid ankle joint pos.
Gastrocnemius lateralis	30% of d between lateral knee joint space and tuberosity of cancaneous	Ankle plantar flexion standing on floor rising against manual resistance	Mid ankle joint pos.
Soleus	50% of d between head of fibula and tuberosity of calcaneous	Ankle plantar flexion standing on floor rising against manual resistance	Mid ankle joint pos.
Tibialis anterior	75% of d between lateral knee joint space and lateral malleolus	Dorsiflexion against manual resistance lying supine with 30 deg. knee flexion	Mid ankle joint pos.

[a]Reproduced with permission from Ericson, et al: Muscular activity during ergometer cycling. Scand J Rehabil Med 17:53-61, 1985, Tab 1, p 55.

[b]d = distance. SIAS = Spina iliaca anterior superior. deg = degrees, pos. = position.

TABLE 5-2
Mean (SD) of Processed Electrical Activity[a,b]

	MVC							
	10%	20%	30%	40%	50%	60%	70%	8
M biceps brachii, 10 subjects								
Integrated activity per 100 msec	10.6 (3.6)	16.2 (4.5)	22.5 (5.3)	29.4 (5.2)	35.7 (5.6)	43.9 (8.4)	54.0 (12.3)	68.2 (22.3)
Zero crossings per 100 msec	7.2 (1.6)	11.2 (1.9)	13.6 (1.8)	15.5 (2.4)	16.0 (3.2)	16.8 (3.7)	16.7 (3.4)	14.9 (2.9)
Integrated activity/zero crossings	1.32 (0.38)	1.32 (0.41)	1.65 (0.54)	2.03 (0.61)	2.35 (0.67)	2.89 (0.71)	3.47 (0.97)	4.93 (2.01)
Turns per 100 msec	30.2 (6.0)	42.3 (5.8)	51.7 (6.9)	57.3 (8.2)	62.6 (12.0)	67.0 (14.8)	72.6 (14.8)	66.2 (9.7)
Mean amplitude (μV)	300 (74)	379 (74)	471 (78)	532 (67)	634 (96)	695 (84)	775 (115)	888 (161)
Turns/mean amplitude	0.100 (0.022)	0.115 (0.017)	0.114 (0.028)	0.111 (0.018)	0.098 (0.019)	0.093 (0.018)	0.075 (0.019)	(0.019)
M tibialis anterior, 10 subjects								
Integrated activity per 100 msec	8.6 (3.5)	12.0 (5.2)	18.5 (5.1)	25.2 (5.4)	33.0 (10.2)	42.3 (9.5)	52.0 (11.4)	64.5 (14.4)
Zero crossings per 100 msec	9.2 (2.7)	12.0 (2.3)	13.9 (1.6)	15.4 (1.4)	15.9 (1.4)	16.1 (1.6)	16.6 (2.3)	16.2 (2.7)
Integrated activity/zero crossings	0.79 (0.42)	1.00 (0.43)	1.26 (0.36)	1.63 (0.50)	2.12 (0.95)	2.70 (0.96)	3.44 (1.20)	4.74 (1.59)
Turns per 100 msec	26.6 (7.4)	37.3 (5.2)	44.1 (4.3)	51.3 (4.2)	55.1 (4.7)	57.8 (5.0)	61.7 (7.9)	61.3 (8.1)
Mean amplitude (μV)	314 (85)	393 (101)	487 (96)	608 (109)	684 (117)	805 (101)	906 (111)	987 (81)
Turns/mean amplitude	0.062 (0.012)	0.081 (0.081)	0.777 (0.077)	0.069 (0.069)	0.064 (0.064)	0.055 (0.055)	0.049 (0.049)	0.039 (0.039)

[a]Reproduced with permission from Christensen H, LoMonraco M, Dahl K, et al: Processing of electrical activity in human muscle during a gradual increase in force. Electromyogr Clin Neurophysiol 58:230-239, 1983, p 33.

When dynamic activities are studied, a further complication arises, making normalization difficult. The EMG-force relationship found in isometric contractions does not remain when muscles are allowed to change length as they contract.[17] This situation is because of the relationship of the length-tension and force-velocity (discussed in Chapter 6) and because of the changes in location of firing motor units relative to the surface electrodes.

If an isometric MVC is used as the reference contraction, the investigator must realize that in many cases the myoelectric signal will give an overestimation of the maximum force.[18] This is especially true if the nonlinear relationship exists as described by Lawrence and DeLuca[1] and Heckthorne and Childress.[19]

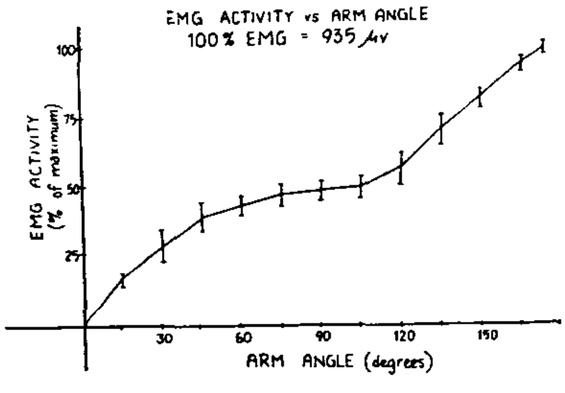

FIGURE 5-1

Activity of upper trapezius. Plateau phase occurs between 45 and 105 degrees of arm elevation (subject #13).

Reprinted with permission from Bagg SD, Forrest WJ: Electromyographic study of the scapular rotators during arm abduction in the scapular plane. Am J Phys Med 65:111-124, 1986, Figure 4, p 118.

Options Other Than Maximal Voluntary Contraction

Because the EMG-force relationship may not be linear for the specific muscle to be studied, the ergonomist may gain increased accuracy by using one or a series of several submaximal isometric contractions to provide a reference for comparison. Such a procedure has been reported by Perry and Bekey[13] and Yang and Winter.[16] This procedure may also be helpful in providing a force level similar to the level of force needed for the activity under investigation. Yang and Winter found that submaximal contractions were more reliable than MVC and therefore should be more desirable to use.[16]

Winkel and Bendix used three different well-defined reference tasks to give myoelectric signal reference values to compare the different seated work tasks studied.[20] Andersson et al also used more than one contraction as a standard.[21] They transformed myoelectric signal microvolts to force data by way of regression analysis using a set of calibration experiments over a force range of interest.

Other alternative procedures for normalization have been reported in the literature. Jonsson and Hagberg[22] and Janda et al[23] used one of the activities being studied as the reference contraction. In comparing muscle activity for different grips, for example, Janda et al used the open grip position as the reference activity.[23]

If the activity under investigation does not include isometric muscle contractions, an isotonic contraction may be used as the standard. Bobet and Norman, for example, used the unloaded walking activity to produce the reference contractions for the loaded walking activities being studied.[24] For a cycling study, Gregor et al used the greatest muscle myoelectric activity values obtained for the cycling activity as the 100% reference.[25] Either of these techniques can be used in ergonomic studies.

In more complex activities, a further complication in the EMG-force relationship arises. Because different synergists and antagonists are active in different proportions, the synergists will share the force production differently and will also have to develop force to overcome antagonistic activity.

The resting, or minimal value, is often subtracted from the myoelectric signal of the reference and of the task. This process serves to eliminate the noise or other instrumentation bias errors.[5,26] This technique, however, is not considered essential by many because this same signal component is included in all of the tasks evaluated.

Normalization of the Task

The task being studied may also need to be normalized in terms of time. A cyclic activity can be set at 100 Ñ for each cycle. This procedure allows for comparisons of data which may vary slightly in duration. This type of normalization has been used in gait analysis[27] (Figure 5-5) and when carrying loads on the back.[24] While studying shoulder joint load and muscle activity during lifting of a box, Arborelius and associates expressed each task as a working cycle ratio.[28,29] The time 0 is when the box left the floor and is when the box was placed on the table (Figure 5-6).

Summary

1. Normalization of the myoelectric activity to a reference contraction is important to compare trials, subjects, and muscles.[5,30,31]
2. The myoelectric signal-to-force relationship is sufficiently linear to use an isometric maximal voluntary contraction as a reference contraction in many situations.[10,12,14,17,18,32] Some limitations to that approach are discussed below.
3. The MVC normalization approach probably results in an overestimate of the force produced.[33]
4. The procedure of normalization is improved when the level of activity is close to the activity under investigation. Submaximal isometric contractions therefore are more accurate as reference contractions. This is particularly true if the EMG-force relationship of the specific muscle is known to be nonlinear.[13,16]

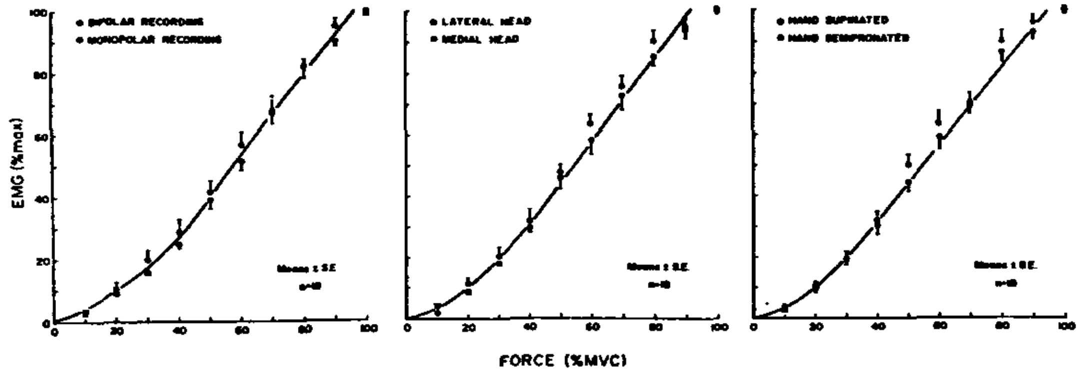

FIGURE 5-2
Mean values (±SE) for EMG-force relations for biceps brachii muscle using three different recording procedures. (Left) Bipolar and monopolar recording configuration. (Center) Lateral and medial head electrode placement. (Right) Supinated and semipronated hand position.

Reprinted with permission from Woods JJ, Bigland-Ritchie B: Linear and nonlinear surface EMG-force relationships in human muscles. Am J Phys Med 62:287-299, 1983, Fig. 3, p 293.

5. The reference contraction should reflect the activity being studied.[21-24,34]
6. Calculations can be made to remove the effect of noise from the signal, if desired.
7. The activity under investigation may also need to be normalized.[2,24,27,28]

OVERVIEW OF METHODS

Many different methods are used to reduce the data contained in the electrical signal and to present it in numerical form. Which method to use depends on why the information is needed, that is, the purpose of the study. The interpretation of the EMG signal plays an important role in determining the relationship of muscle activity to task performance. The most basic information obtained from a myoelectric signal is 1) whether or not the muscle is active and 2) the relative amount of activity of the muscle. By using the appropriate process of normalization, a reasonable estimate of muscle function can be obtained by the ergonomist. This information can be combined with an observation system or simply an event marker of some type to determine 1) when the muscle is active; 2) when a peak of activity occurs; 3) what the pattern of muscle activity is during a movement. position, or force production; and 4) whether fatigue has occurred. The instrumentation used is presented in Chapter 4, and other factors that can affect the values obtained are discussed in Chapter 6.

Raw Signal

The raw, or unprocessed, EMG signal is the basis of all methods of interpreting the myoelectric activity from muscles. The ergonomist should monitor the raw signal, even though other signal processing may be used, so that artifacts can be detected and controlled as necessary.

In the past, probably the most common way to interpret EMG was by visual inspection of the raw signal. With training, experience, and the use of multiple gains and oscilloscope sweep velration, the observer should be able to evaluate the raw EMG signal visually and effectively. The observer should be able to identify when the raw signal indicates that a muscle is active and when it is relaxed. The relative amount of activity may be classified either by words, such as nil, negligible, slight, moderate, marked, or very marked, or by numerical values, such as 0–5, with 0 being no activity and 5 being maximal activity.

Such visual observations are based on signal amplitude and frequency. An example is provided in the work of Sofranek and associates when they visually examined the raw myoelectric signal to determine the duration of the interference pattern (Figure 5-7).[35] The distance from the first action potential spike (on) to the last spike (off) of the interference pattern adjusted to the paper speed provided a measure of duration.

The raw myoelectric signal was also used by Maton

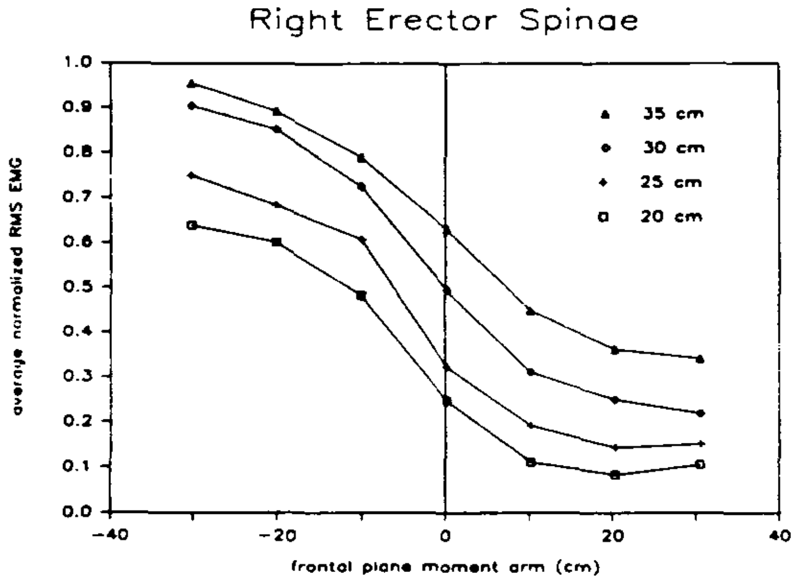

FIGURE 5-3 A
Right erector spinae EMG.

Reprinted with permission from Seroussi RE, Pope MH: The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. J Biomech 20:135-146, 1987, Figure 4a, p 141.

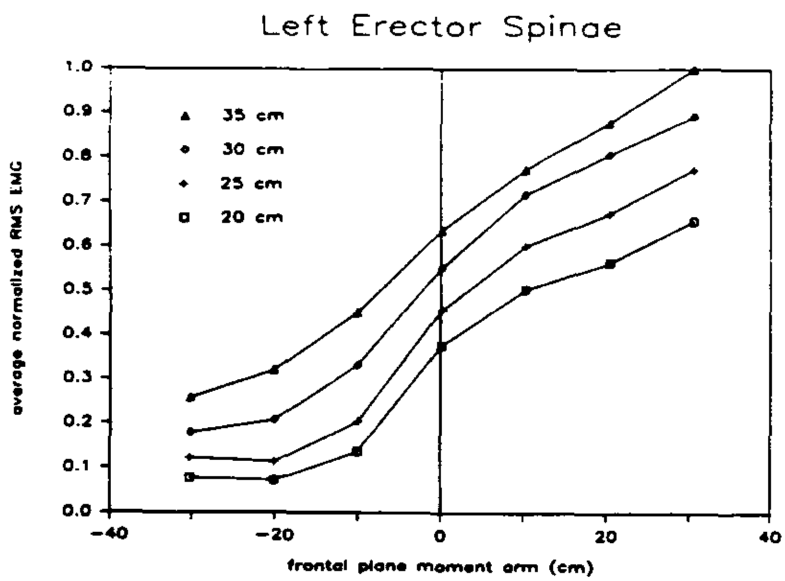

FIGURE 5-3 B
Left Erector Spinae EMG.

Reprinted with permission from Seroussi RE, Popse MH: The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. J Biomech 20:135-146, 1987, Figure 4b, p 141.

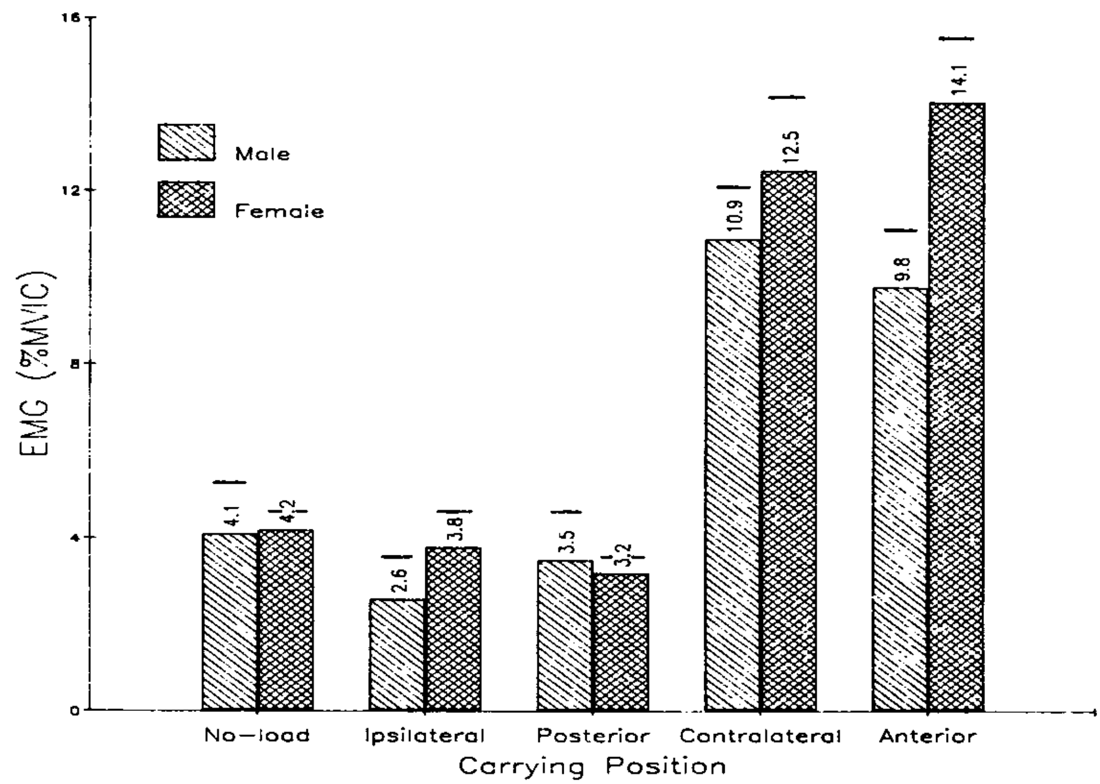

FIGURE 5-4

Erector spinae EMG mean values (% MVIC) for all four carrying positions and the no-load condition for males and females (both load sizes combined). The short horizontal bars indicate standard errors.

Reprinted with permission from Cook TM, Neumann DA: The effects of load placement on the EMG activity of the low back muscles during load carrying by men and women. Ergonomics 30:1413-1423, 1987, Figure 2, p 1419.

and associates to study the synergy of elbow extensor muscles during the deceleration phase of elbow flexion movements.[36] The authors observed the timing of the muscle bursts during the task. The amplitude values could not be quantified accurately, and only terms of weak and slight amplitude were used. The raw signal of the three heads of the triceps, the anconeus, and the biceps brachii muscles for the slow and the fast movements are shown in Figure 5-8.

The problems of using on-off information from raw EMG signals was illustrated by Winter.[27] He used three different threshold levels as six subjects walked normally. The differences in phasic patterns are shown in Figure 5-9. Different threshold levels give different on-off patterns, and, therefore, the on-off information can be misleading. Similar differences would be expected should similar techniques be used in ergonomic studies.

How other conditions influence activities evaluated in ergonomics was shown in a study by Janda and associates in which the raw myoelectric signal was used to study the role of the forearm and hand muscles during various phases of prehensile activity.[23] Surface myoelectric signals were recorded during a sustained grip of 10 kg at each of three different handle spacings. Each data set was compared to the value obtained from the widest handle spacing position. The results revealed that the extrinsic flexor muscles were active throughout the test range, but the intrinsic muscle group was active only at the narrower handle spacings. Sample results are shown in Figure 5.10 and are based on the millivolt values of the signal thickness. In these cases, the investigator would have difficulty making interpretive statements.

Perry and associates used a more involved approach to quantify the raw signal.[37] They devised an eight-point scale that accounted for both amplitude and frequency components of the signal (Figure 5-11). The amplitude measured in millimeters was divided into a four-point

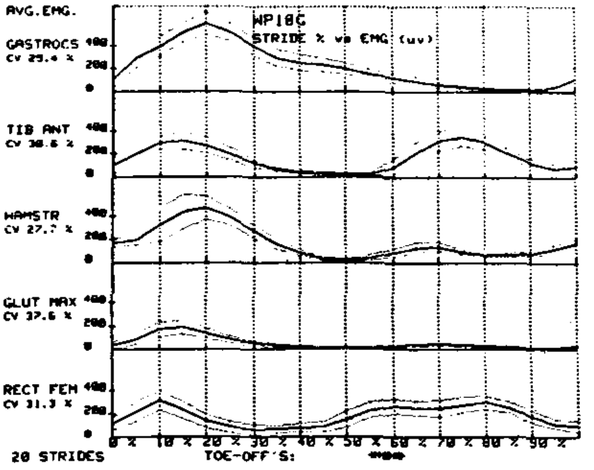

FIGURE 5-5

Ensemble average of linear envelope of EMG from five muscles of a typical patient. Solid line indicates the mean EMG signal in microvolts and the dotted lines represent one SD of the EMG profiles from the 20 strides.

Reprinted with permission from Winter D: Pathologic gait diagnosis with computer-averaged electromyographic profiles. Arch Phys Med Rehabil 65:393-395, 1984, Figure 6, p 395.

scale such as 1 mm = 0.5 point, 2 mm 1.0 point, 3 mm = 1.5 points. The values of 4 mm and 5 mm were both given a score of 2, and the values of 7 mm and 8 mm were given a score of 3. They also used a four-point scale to assess the density of the signal that would reflect the signal frequency. No signal would be zero density points, 50% of a signal would equal 2 points, and maximal activity showing a darkening record would equal 4 points. The maximum combined score from a maximal contraction would be 8. The final rating of the muscle activity was presented as a percentage of the myoelectric signal during maximal manual muscle testing. The percentage was obtained by adding the amplitude and density scores and dividing this value by 8. Their quantitative evaluation of the myoelectric signal reveals the onset and duration of muscle firing, provides a quantitative value of the amount of muscle firing, and identifies the interval phase (time) during which the muscle firing was greatest. In general, similar techniques could be applied to data obtained from ergonomic settings.

In summary, the raw EMG signal is a random signal obtained from the surface electrodes and then amplified. The raw signal should be monitored for all investigations, because the investigator can pick out major artifacts and eliminate that area or part of the signal.[13,38] The ergonomist should be aware that on-off information is nominal data but the relative amount of activity is ordinal data. Interpretation that can be made in ergonomic studies are dependent on the amplifier gain and the sensitivity settings of recording instruments. No current standards exist for instrument settings or interpretive rules; therefore, considerable judgement needs to be exercised in evaluating EMG records of raw data. Such a data form is of limited value when findings are to be related to force or fatigue. Thus, the signal is processed to attain a quantitative estimate that can be used for statistical or higher order analysis.[13]

Demodulation

The raw signal provides limited information and can actually provide inaccurate information if it is not processed into another form. Inman and associates stated that the raw waveform of the myoelectric signal is sufficiently complex that simple comparisons of peak-to-peak amplitudes are inaccurate.[17] If true, then other types of comparisons would also be inappropriate. Given that the raw signal is of high quality, however, further management of the data can be desirable. Instrumentation, as discussed in Chapter 4, has been designed that provides a number of outputs representing the average amplitude of the input.[17] The interpretation of the results of these techniques is included in the next section of this chapter.

Linear Envelope

A linear envelope can be used to provide an envelope that represents a profile of the myoelectric activity of the muscle over time. The electronic process involved includes rectification of the raw signal and then a passing of the signal through a low pass filter that follows the peaks and valleys of the rectified signal. The combination of the full-wave rectifier followed by a low pass filter is often called a linear envelope detector. Different filters provide different information.[39] The differences in the type of filter (eg, first order or Paynter) are discussed by Gottlieb and Agarwal.[40] The effects of different window lengths for smoothing are presented by Herschler and Milner.[39] The ergonomist should be warned that the specific devices used may affect the interpretation. The characteristics of the filter, therefore, should always be stated.

Inman and associates referred to this process as integration for "lack of a better name" [17] Note should be made, however, that the linear envelope process, although related to integration, is not true integration.[17,39]

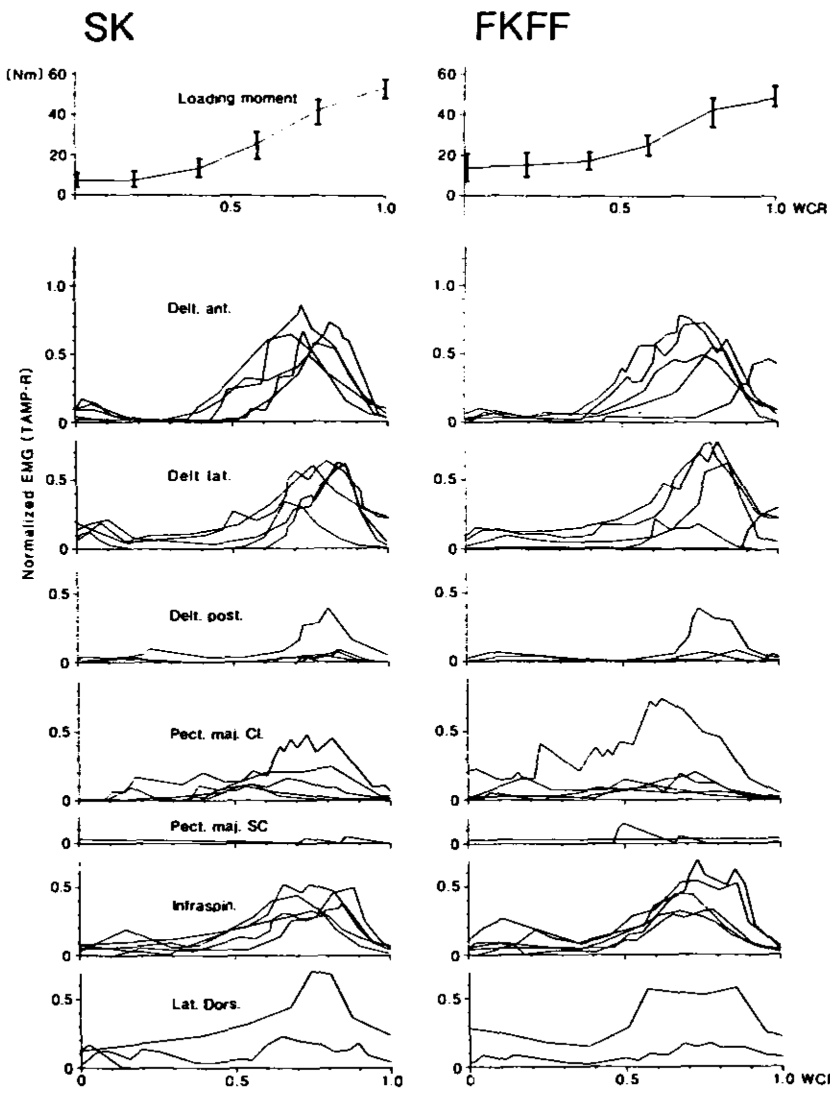

FIGURE 5-6

The upper left and right graphs show the loading moment of force (mean with 95% confidence intervals) for the SK and FKFF lifts. Time is expressed as a working cycle ration (WCR). The other graphs show individual muscular activity curves from five individuals from seven muscles: anterior, lateral (middle) and posterior parts of clavicular and sternocostal portions of pectoralis, infraspinatus, and latissimus dorsi. Activity norm TAMP-R. "Segmentation" of curves is a result of process. "Missing" curves indicate absence of act.

Reprinted with permission from Arborelius UP et al: Shoulder joint load and muscular activity during lifting. Scand J Rehab Med 18:71-82, 1986, Figure 2, p 74.

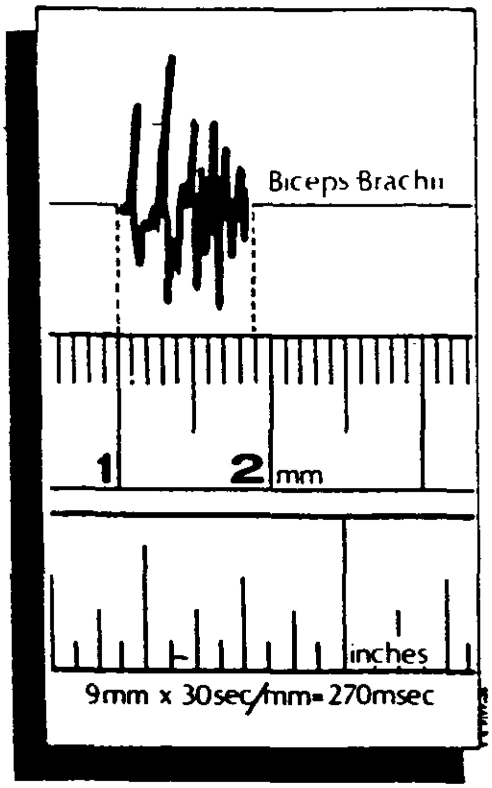

FIGURE 5-7
Measurement of movement time from a recording of an EMG interference pattern.

Reprinted with permission from Sofranek MG et al: Effect of auditory rhythm on muscle activity. Phys Ther 62:161-168, 1982, Figure 4, p 164.

The processed signal leaves the baseline when the myoelectric signal is greater than zero. The analog representation rises with the increase in muscle activity and falls with the decrease of muscle activity. The curve returns to zero, or baseline, when the muscle is relaxed. Thus, the linear envelope displays a pattern that reflects changes in the state of the muscle contraction (Figure 5-12). The evaluation of this signal provides an instantaneous muscle activity in millivolts that can then be related to a specific motion or position.[41] The myoelectric signal has been found to lag behind the production of tension by about 60 to 100 ms. Further delays can result from the instrumentation features (see Chapter 4). The signal, therefore, does not perfectly identify the mechanical event, although it is a good reflection. An event marker of some kind is recommended as an indicator of the mechanical event.

Numerous researchers have used the linear envelope to describe the myoelectric activity that occurs during various activities. Studies have been performed to determine upper limb muscle activity during various tasks,[3,4,17,18,28,29,31,42-44] trunk muscle activity,[6,24,45-47] and muscle activity of the lower limbs during locomotion.[27,31,39,42,48-50] Figures 5-5, 5-6, 5-13, and 5-14 show different ways of presenting this information.

The information obtained from the linear envelope includes the onset and duration of the muscle activity, the instantaneous muscle activity, and the pattern of muscle contraction. The technique is widely applicable in studies of periodic activity such as work-rest cycles and activities where repetitions could be averaged over an interval of time. An EMG from either the upper or lower extremities or the trunk can be subjected to this form of analysis.

Root Mean Square

The root-mean-square (RMS) voltage is the effective value of the quantity of an alternating current. The true RMS value of a myoelectric signal measures the electrical power in the signal.[51] The method of obtaining this measure is presented in Chapter 4. It gives a linear envelope of the voltage, or a moving average over time. Therefore, the RMS wave form is similar to the linear envelope (Figure 5-15).[34] In combination with a positive or time indicator, it provides an instantaneous measure of the power output of the myoelectric signal. The RMS value depends on the number of motor units firing, the firing rates of the motor units, the area of the motor unit, the motor unit duration, the propagation velocity of the electric signal, the electrode configuration, and the instrumentation characteristics.[10,52]

DeVries determined the efficiency of electrical activity as a physiological measure of the functional state of muscle tissue, using the RMS values as an indication of myoelectric activity.[53] The force and RMS values were linearly related, but the slopes of the lines were different for subjects of different strengths. Some investigators have continued with this application of EMG, but generally the methods have not been widely used. Some application in ergonometry probably exists. Another basic study was completed by Lawrence and DeLuca, who investigated whether the normalized EMG from surface electrodes versus normalized force relationship varies in different human muscles and whether this relationship depends on training and rate of force production.[1] They used RMS values because the RMS more completely represents motor unit behavior during muscle contraction.

Lind and Petrofsky studied the myoelectric amplitude during fatiguing isometric contractions, a condition potentially occurring at various work sites.[54] The

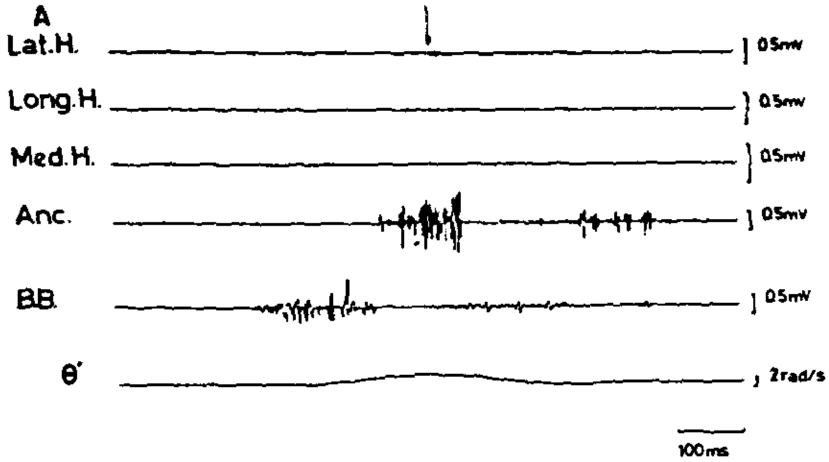

FIGURE 5-8 A

Typical records of flexion movements performed against one inertial load ($I_0 = 0.021$ kg•m^2) slow movement. Lat. H: surface EMG of lateral head of the triceps muscle. Long. H: surface EMG of long head of the triceps muscle. Med. H: surface EMG of medial head of the triceps muscle. Anc.: surface EMG of anconeous muscle. B.B.: surface EMG of biceps brachii. O': angular velocity. ↑: the arrow represents the onset of activity of the lateral head. (1) first burst; (2) second burst.

Reprinted with permission from Maton, et al: The synergy of elbow extensor muscles during dynamic work in man: II. Braking elbow flexion. Eur J Appl Physiol 44:279-289, 1980, Figure 1 A.

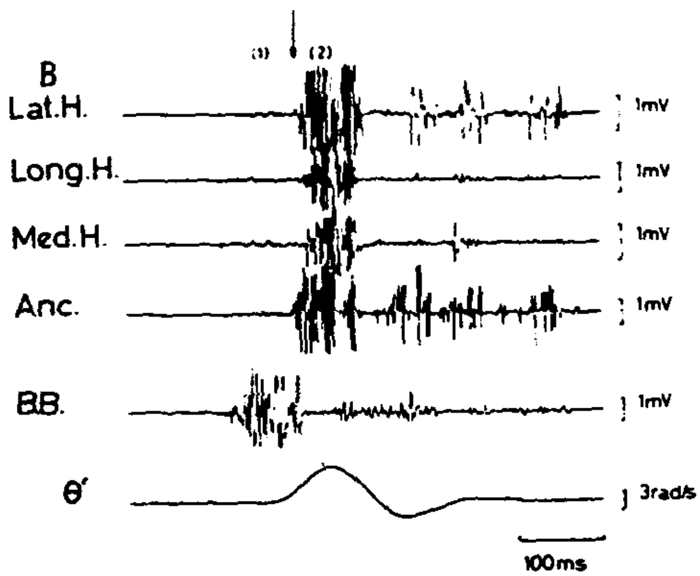

FIGURE 5-8 B

Typical records of flexion movements performed against one inertial load ($I_0 = 0.021$ kg•m^2) fast movement. Lat. H: surface EMG of lateral head of the triceps muscle. Long. H: surface EMG of long head of the triceps muscle. Med. H: surface EMG of medial head of the triceps muscle. Anc.: surface EMG of anconeous muscle. B.B.: surface EMG of biceps brachii. O': angular velocity. ↑: the arrow represents the onset of activity of the lateral head. (1) first burst; (2) second burst.

Reprinted with permission from Maton, et al: The synergy of elbow extensor muscles during dynamic work in man: II. Braking elbow flexion. Eur J Appl Physiol 44:279-289, 1980, Figure 1 B.

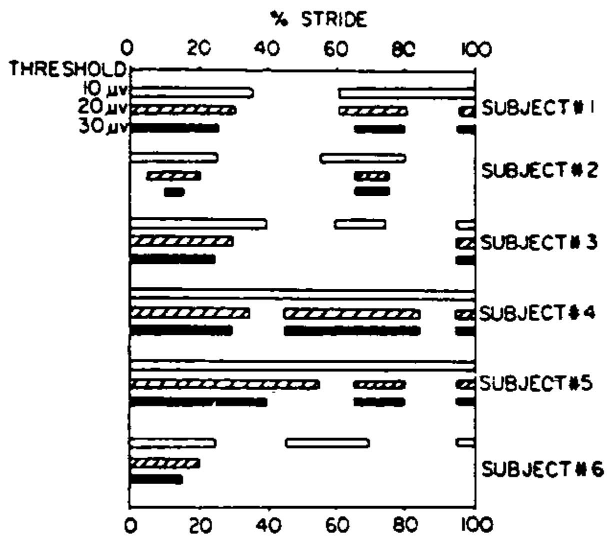

FIGURE 5-9

Phasic patterns as derived from profiles for the rectus femoris muscle with three arbitrarily chosen thresholds: 10 μV, 20 μV, and 30 μV. Depending on the subject and the threshold, considerable discrepancy would result in defining a "normal" pattern against which to compare patients.

Reprinted with permission from Winter D: Pathologic gait diagnosis with computer-averaged electromyographic profiles. Arch Phys Med Rehabil 65:393-395, 1984, Figure 3, p 394.

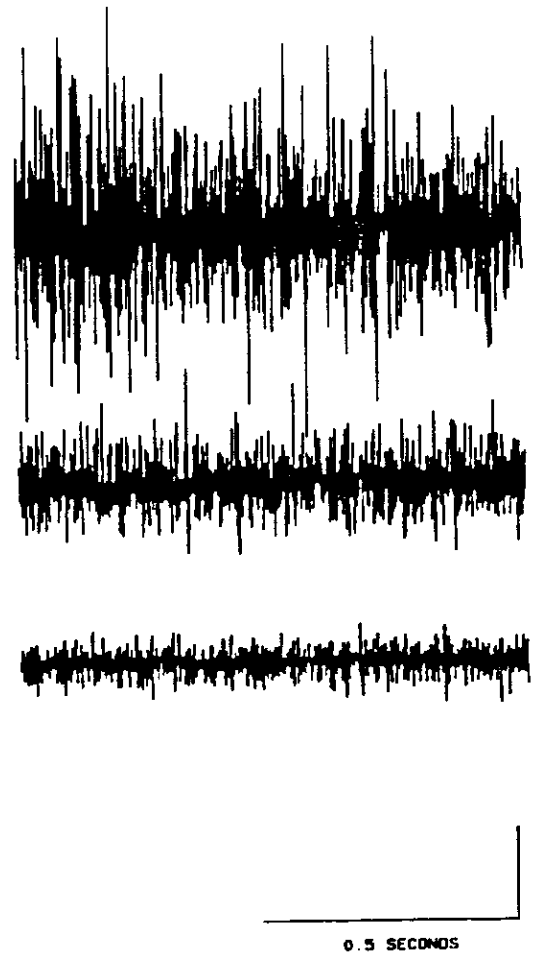

FIGURE 5-10

Surface EMG data of hand intrinsic muscles reveal maximum activity with narrowest (top tracing) dynamometer handle spacings with relatively little activity at widest handle spacing (bottom tracing).

Reprinted with permission from Janda DH et al: Objective evaluation of grip strength. J Occup Med 29:569-571, 1987, Figure 2, p 570.

RMS values were linearly related to the exerted force. With prolonged contractions of 25% MVC, the myoelectric amplitude decreased as the force decreased. They found that intrasubject values revealed linearity, but a large intersubject variation was present for the absolute amplitude. In a subsequent study, Petrofsky and Lind used RMS amplitude measures to determine the influence of different temperatures on the myoelectric signal during brief and fatiguing isometric contractions.[55] Hand gripping was the activity studied. The RMS amplitude was calculated over 1.5-second periods from digitized EMG. The normalized RMS amplitude during the brief isometric contractions showed a linear relationship with force after limb immersion in water of temperatures at 30° and 40°C, but demonstrated a curvilinear relationship after limb immersion in water of temperatures at 10° and 20°C (Figure 5-16). The normalized RMS amplitude of the EMG progressively increased during sustained contractions for all four water temperatures.

In considering the work site, Hagberg and Sundelin studied the load and discomfort of the upper trapezius for secretaries using a word processor.[56] They used an amplitude probability distribution function of RMS-detected signals for specific loads and for five-hour work periods. They found a static work level of approximately 3.0% of the MVC for the upper trapezius muscle over the length of the work period. Their results are displayed in Figures 5-17 and 5-18. Five other but separate studies of myoelectric changes with muscle fatigue have been presented by Jorgensen et al.[57] Root-mean-square determinations were used to evaluate changes in the myoelectric amplitude. Increases in amplitude were recorded for

FIGURE 5-11 A

Electromyographic activity of anterior tibialis muscle during tracelevel manual muscle test.

Reprinted with permission from Perry et al: Predictive values of manual muscle testing and gait analysis in normal ankles by dynamic electromyography. Foot Ankle 6:254-259, 1986, Figure 1, p 256.

FIGURE 5-11 B

Electromyographic activity of anterior tibialis muscle during fair level manual muscle test.

Reprinted with permission from Perry et al: Predictive values of manual muscle testing and gait analysis in normal ankles by dynamic electromyography. Foot Ankle 6:254-259, 1986, Figure 2, p 256.

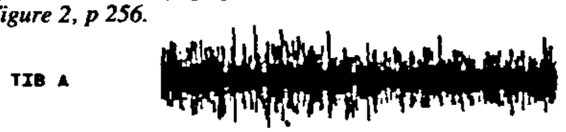

FIGURE 5-11 C

Electromyographic activity of anterior tibialis muscle during maximum level manual muscle test.

Reprinted with permission from Perry et al: Predictive values of manual muscle testing and gait analysis in normal ankles by dynamic electromyography. Foot Ankle 6:254-259, 1986, Figure 3, p 256.

both continuous and intermittent work. The studies revealed that different muscles respond differently in that the change in myoelectric amplitude for the triceps brachii was more pronounced than the change in amplitude of the biceps brachii.

The concept of **static** load was studied by Jonsson.[58] He looked at examples of actual work situations using RMS analyses to determine their load levels. The static component of muscle activation was considered to be the lowest level of muscular activation recorded during the work period. The **dynamic** component was considered to be the load levels above the static level (Figure 5-19). He discussed the possibility of job rotation to reduce the static load level during a full work day (Figure 5-20).

A number of other studies including ergonomic elements have been performed using RMS analysis. A few examples are given for reference. Winkel and Bendix[20] studied muscular performance during seated work, Hagberg and Sundelin[56] studied the use of a word processor, and Andersson et al[21] the activity of trunk muscles during desk work. Given these examples and the work now appearing in the literature, this is a widely accepted form of EMG processing used for ergonomic environments. The output form is similar to the linear envelope detector, but it represents a somewhat better mathematical representation of the original.

Integration

The total amount of muscle activity occurring during any given time interval is represented by the area under the curve during that time interval. The process for determining this area is called integration. Integration may be done manually or electronically. A simple way to determine the area under the curve is to trace the curve on paper, cut out the curve, and weigh the enclosed area.[38] In other cases, researchers have used a planimeter to evaluate the area under the curve.[7,22] Bigland and Lippold[8] found that electrical processing and planimetry give similar results. Most researchers are now using electronic integration (see Chapter 4).

Integrated electromyography (IEMG), evaluating the area under the curve, is a continuous evaluation of that area. The IEMG signal, therefore, increases as long as any myoelectric activity is present and decreases in slope as there is less myoelectric activity. The amplitude measure at any time along the curve represents the total electrical energy summed from the beginning of the activity. Because the IEMG curve keeps increasing, the curve may need to be reset to zero for practical purposes. This reset can be done either at fixed time intervals (time reset) or at a predetermined amplitude (level reset). Appendix B, Figure 5, contains a comparison of these various techniques. Because IEMG depends on the amplitude, duration, and frequency of the action potentials, it represents the number of active motor units.

The continuous increase in the integrated signal as the raw myoelectric signal remains constant as shown in Figure 5-21,[9] where the integrated and raw myoelectric signals are shown in relation to a static force. Nelson and associates recorded the myoelectric signal of the soleus muscle during isokinetic movements of ankle plantar flexion and dorsiflexion.[59] The change in the raw myoelectric signal is reflected by a change in the slope of the integrated signal. When the muscle activity is high, the slope of the integrated signal is steep. At lower levels of muscle activity, the signal tends to plateau (Figure 5-22). Note that as the contractions continue, the integrated signal continues to move away from the baseline. If the task has a long duration, this can lead to confusion in the

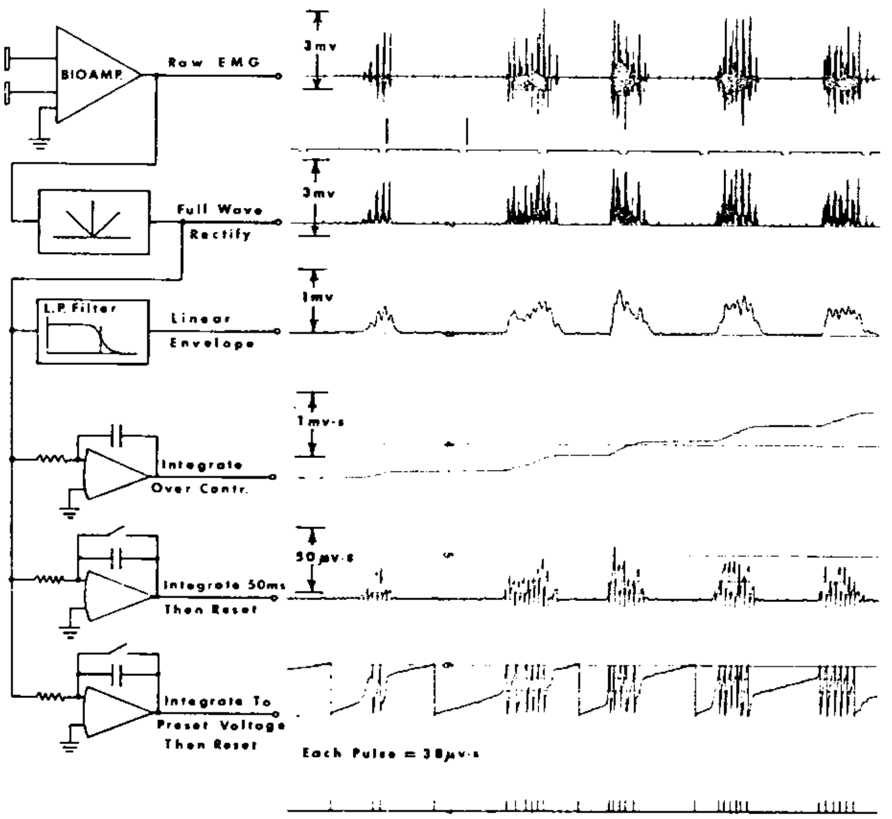

FIGURE 5-12

Example of several common types of temporal processing of the EMG.

Reprinted with permission from: Units, Terms and Standards in the Reporting of EMG Research. Report by the Ad Hoc Committee of the International Society of Electrophysiological Kinesiology, August, 1980, Figure 5, p 9.

recordings, that is, the lines from the integrated signal may cross into the line of the raw signal. An example of level reset is illustrated in Figure 5-23.[60] Numerous other researchers have used time or level reset integration to study muscle activity.[19,60–64] To obtain the integrated values from time reset, the investigator sums the value of the peaks over the desired contraction time. To obtain the integrated values from level reset, the investigator counts the number of reses times the level value for the period of the contraction.

The integrated myoelectric signals may also be collected over a short time span during a cyclic activity, such as occurs during the performance of jobs. Such a procedure was used by Jorge and Hull and is shown in Figure 5-24.[61] The rectified signal was integrated over 75 ms segments during the period of cycle, or 360 degrees. This method provided 10 integrated myoelectric values representing 36 degree intervals for the period. The average normalized electrical activity for the interval of interest can then be evaluated and compared with other intervals, muscles, or tasks. Figure 5-25 shows how Gregor and associates treated similar information.[25]

Integration of the myoelectric signal provides a measure of the number of active motor units and their rate of firing. It can provide information concerning the on-off time and the relative myoelectrical activity of the muscle over a set time period. Thus, as a form of output, the data can be analyzed in ways similar to those used for either the linear envelope detector or the RMS. Often, the selection is based on available equipment simply because there is neither a prescribed technique nor a prescribed standard that must be met for studies in the area of ergonomics.

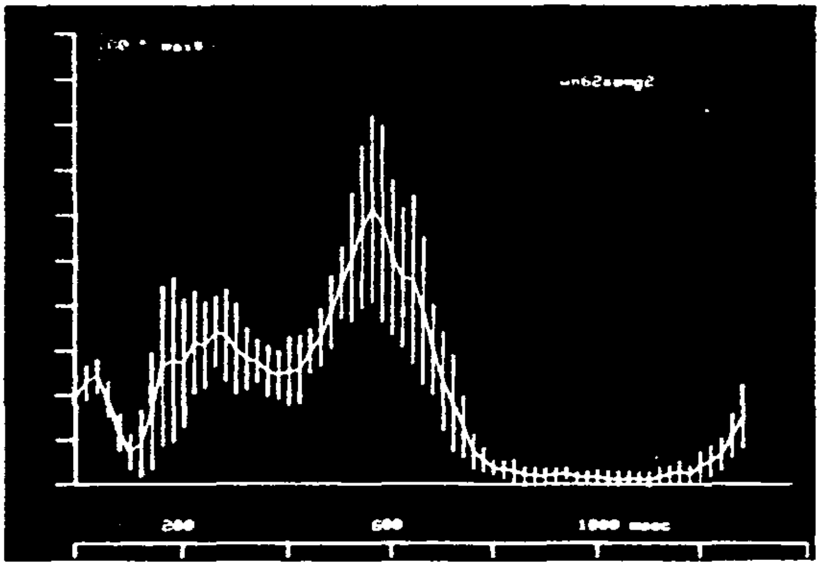

FIGURE 5-13

Average of linear envelope of soleus muscle over 10 walking strides. Contraction was normalized to 100% MVC, and standard deviation at each point in time is shown by vertical bars.

Reprinted with permission from Dainty D, Norma R: Standardizing Biomechanical Testing in Sport. Champaign, IL, Human Kinetic Publ Inc, 1987, Figure C.4, p 120.

Frequency Analysis

The myoelectric signal consists of a series of action potentials firing at certain frequencies. Frequency analysis (spectral, harmonic, Fourier) decomposes the myoelectric signal into sinusoidal components of different frequencies. As described in Chapter 4, frequency analysis can be done either by passing the raw myoelectric signal through a series of electronic filters and plotting the result or by digitizing the data and using a computer to analyze the data and present it in a smooth spectrum over a given frequency range. This frequency analysis gives the energy distribution of the signal as a function of frequency. It detects the amplitude of common frequencies of the signal. The power spectrum of the interference pattern, thus, essentially reflects the properties of the individual components. A lengthy and detailed description of the power spectrum and its analysis has been written by Lindstrom and Petersen.[65]

The power spectrum may be presented in linear, logarithmic linear, or double logarithmic scales. The power spectrum with linear scales is measured in volts squared per hertz (V^2/Hz). The decibel (dB) unit is used if the scale for power, energy, or amplitude is logarithmic.

The power spectrum of the total signal reveals the component individual motor unit properties. The area under the power spectral curve equals the signal power. The frequency power spectrum shows only smaller upward shifts in the frequency spectrum as the force of the contraction increases. This increase occurs at low levels of tension, but after about 50% of the MVC, the frequency values no longer increase.

A common use of power spectrum analysis has been the evaluation of local muscle fatigue. With a sustained muscle contraction, the high frequency components of the signal decrease, but the low frequency components gradually increase. This change results in a shift in the power spectrum toward the lower frequencies.[66]

The two most reliable measures of the power spectrum are the mean frequency and the median frequency (Figure 5-26). The mean frequency is the average of all frequencies. The median frequency is that frequency having 50% of the frequency distribution on each side. The median frequency appears to be less sensitive to noise than the mean frequency.

The shift of frequency spectrum may be caused by such factors as follow:

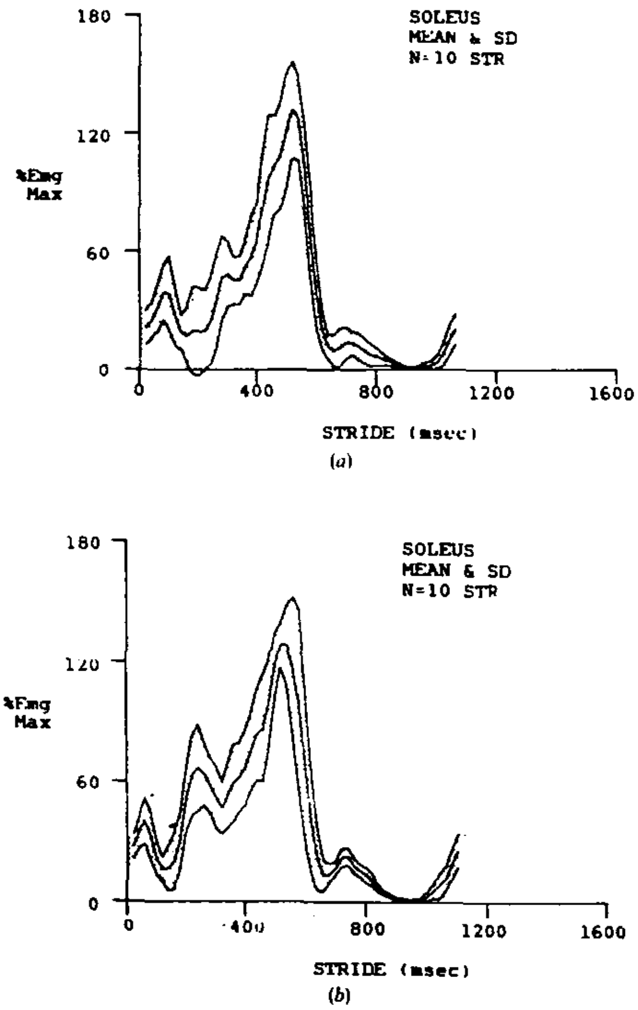

FIGURE 5-14

Summary of the left soleus muscle activity for 10 strides of walking on (a) walkway and (b) treadmill for one subject. The 0-0 msec point on the abscissa corresponds to heel strike (HS). The ensemble average of 10 strides along with plus and minus one standard deviation (SD) for each sample of ten points are represented (sampling frequency at 50 Hz).

Reprinted with permission from Arsenault, et al: Treadmill versus walking locomotion in humans: An EMG study. Ergonomics 29:665-676, 1986, Figure 1, p 669.

1. Synchronization of motor units.[65,67]
2. Increase and decrease in recruitment of motor units.[65]
3. Dysfunction of muscle spindles.[65]
4. Combination of synchronization and desynchronization.[65]
5. Change in shape of the motor unit signals.[65]
6. Propagation velocity changes.[65,68]
7. Intramuscular pressure changes.[69]

Evaluation of spectral shifts of the myoelectric signal has allowed for the study of fatigue during a variety of job conditions. Chaffin presented the results of the change in the myoelectric frequency following exhausting contractions.[33] The shift in the frequency during a prolonged contraction occurs because of the increase in the low-frequency power from about 17% in the rested condition to over 60%. Figure 5-27 reveals this shift in center frequency from above 40 Hz to below 30 Hz.[33]

Several other researchers have used frequency analysis to determine the presence of fatigue in a muscle following a specific task.[32,55,70-80] Frequency analysis was also used by Lindstrom et al to determine changes in muscular fatigue and action potential conduction velocity.[81] The output signals were recorded on a logarithmic scale in decibels versus the center frequency of the filter bands that gives the power spectrum of the myoelectric signal (Figure 5-28). They noted a decrease in the center frequency, a finding that may be of interest during worksite analyses.

The use of spectral median frequency has recently been questioned as an indicator of muscular fatigue. Matthijsse and associates examined power spectral median frequency and mean power by means of Fourier analysis in relation to plantar flexor muscle contraction during a prolonged task.[82] They found that although some subjects demonstrated a median frequency shift, no significant difference was found across subjects. They suggest that care be taken when applying median frequency analysis to determine fatigue. These warnings may be related to frequency change found by some researchers to occur as a consequence of a change in load[83-85] or a change in muscle length.[86]

Hogan and Mann studied changes in the myoelectrical signal power spectrum of the bicep brachii muscle during different muscle force levels, in different subjects, with different electrode locations and electrode configurations.[83] An illustration of the double logarithmic presentation of the results is shown in Figure 5-29. Gander and Hudgins used the power spectrum process to study the effect of increasing load on the biceps brachii.[84] The characteristics of median frequency and relative power were compared with torque values as shown in Figures 5-30 and 5-31.

Bazzy and associates studied the effect of a change

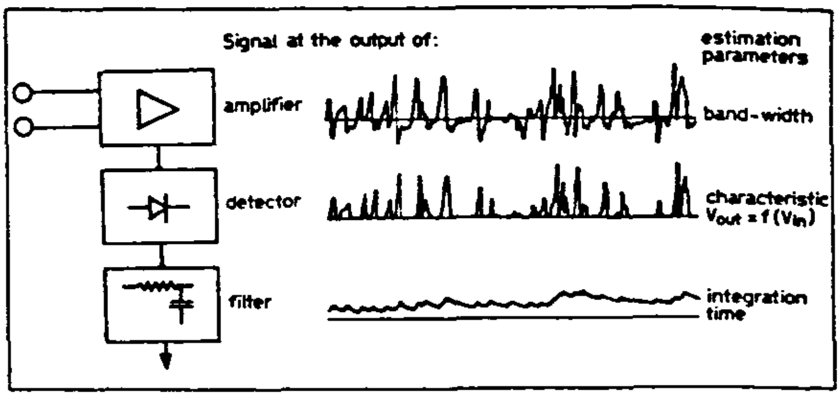

FIGURE 5-15
The "single channel"—an elementary signal processing array (left). Example of signal processing (right). Of importance, to arrive at improved estimates of the myo-electric signal level is the signal band-width, the detector characteristics, and the postdetector band-width (related to averaging time).

Reprinted with permission from Kadefors R: Myo-electric signal processing as an estimating problem. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1, pp 519-532, Figure 1, p 522.

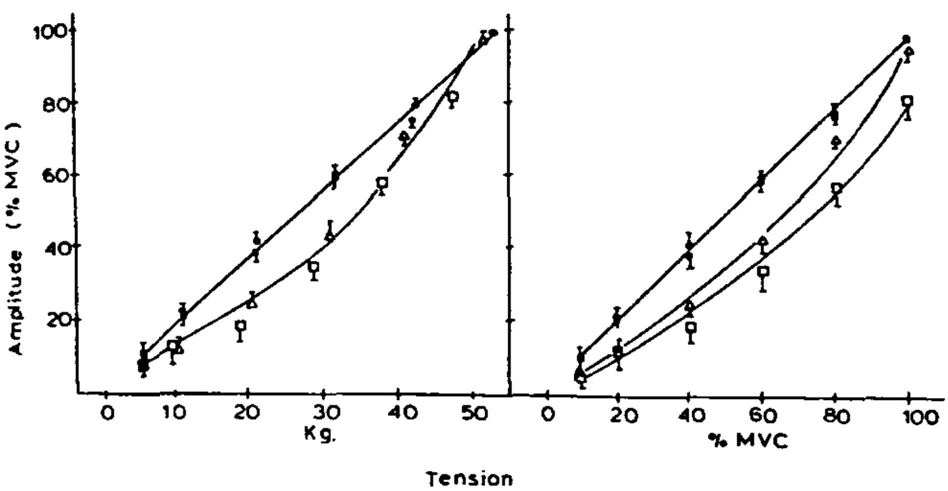

FIGURE 5-16
The RMS amplitude of the power spectra of the surface EMG during brief isometric contractions at each of four bath temperatures, 10 (□), 20 (Δ), 30 (O), and 40 (•) °C compared with both the relative and absolute tension developed by the muscles. Each point illustrates the mean of two experiments on each of 10 subjects ± SD.

Reprinted with permission from Petrofsky JS, Lind AR: The influence of temperature on the amplitude and frequency components of the GM during brief and sustained isometric contractions. Eur J Appl Physiol 44:189-200, 1980, Figure 1, p 193.

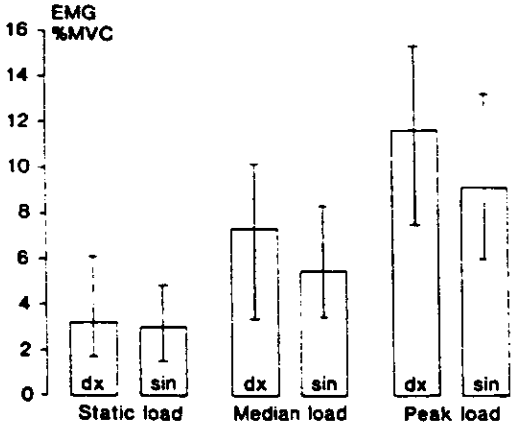

FIGURE 5-17

Median values and the first and third quartile of load levels on the right (dx) and left (sin) sides when operating a word-processor. Both the 3-hour and the 5-hour work periods are in the calculation (n = 12).

Reprinted with permission from Hagberg M, Sundelin G: Discomfort and load on the upper trapezuis muscle when operating a word processor. Ergonomics 29:1637-1645, 1986, Figure 3, p 1642.

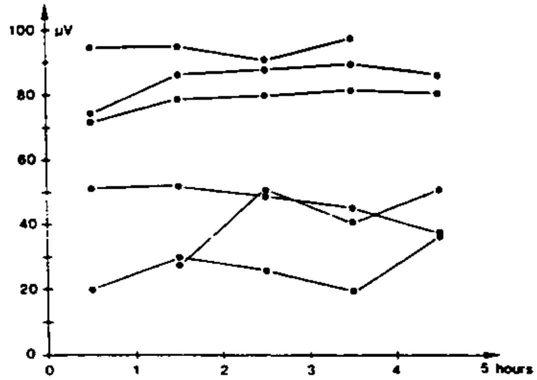

FIGURE 5-18

Hour-by-hour analysis of the 5-hour work period. Values are given in time mean RMS microvolts. The values for each operator are connected by a line.

Reprinted with permission from Hagberg M, Sundelin G: Discomfort and load on the upper trapezuis muscle when operating a word processor. Ergonomics 29:1637-1645, 1986, Figure 4, p 1642.

in the muscle length upon the frequency content of the myoelectric signal.[86] They found that the length at which a muscle isometrically contracts can alter the mean centroid frequency of the signal. The results of one of their subjects is presented in Figures 5-32 and 5-33.

In summary, some investigators have demonstrated changes in the EMG with fatigue. Most of these experiments have been published within the last years, and there is now effort being made to clarify the relationships and meaning. Any ergonomist proposing to use these techniques needs to be well grounded in signal analysis techniques to use these methods and comprehend the volume of literature that will result from other studies. Because of the importance of fatigue to ergonomics, this area promises to be of significance in analyses proposed to be used in the worksite.

Zero Crossings

The number of times the raw EMG signal crosses the baseline (zero value) appears to be related to muscle contraction force. Within limits, as the muscle activity increases, the frequency increases, resulting in more zero crossings. The frequency of zero crossings can easily be counted electronically. As with the earlier mentioned frequency values, zero crossing values do not increase at high levels of muscular effort. At about 60% of the maximal voluntary contraction the zero crossings count levels off (Figure 5-34).[87] Interpretation of quantitative data, therefore, is not always simple.

Because the use of spectral changes in the myoelectric signal as a valid indicator of muscle fatigue have been questioned, Hagg and associates[88] and Suurkula and Hagg[89] have used a frequency analysis based on zero crossings, to study shoulder and neck disorders in assembly line workers. They believe that the results are promising and suggest that the technique is valuable for ergonomic studies at the workplace. Given the availability of other techniques, however, there may not be much emphasis forthcoming on this form of analysis.

Spike Countings

Bergstrom manually counted the number of positive and negative spikes, or peaks, of the raw myoelectric signal.[90] Spikes can also be counted by electronic methods. Spikes of low amplitude are given equal value to spikes of high amplitude. The total count appears to be related to the amount of muscle activity. The number of spikes increases linearly with increasing contraction force to about 70% of MVC and then levels off.

Robertson and Grabiner compared two methods of counting spikes of the myoelectric signal values obtained by an integrated process: digital spike counting

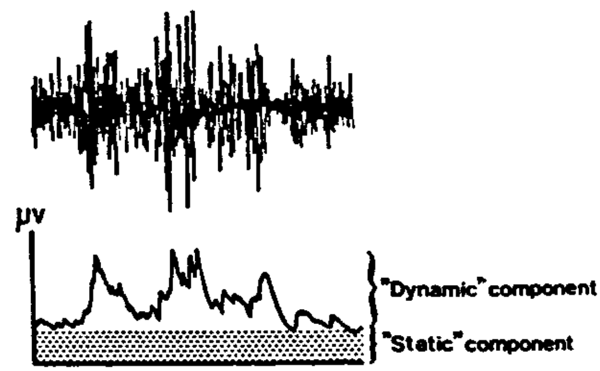

FIGURE 5-19
Raw EMG and the corresponding RMS detected and low pass filtered EMG signal for approximately 6s of activity. The muscular load may be subdivided into a "static" component and a "dynamic" component.

Reprinted with permission from Jonsson B: The static load component in muscle work. Eur J Appl Physiol 57:305-310, 1988, Figure 1, p 305.

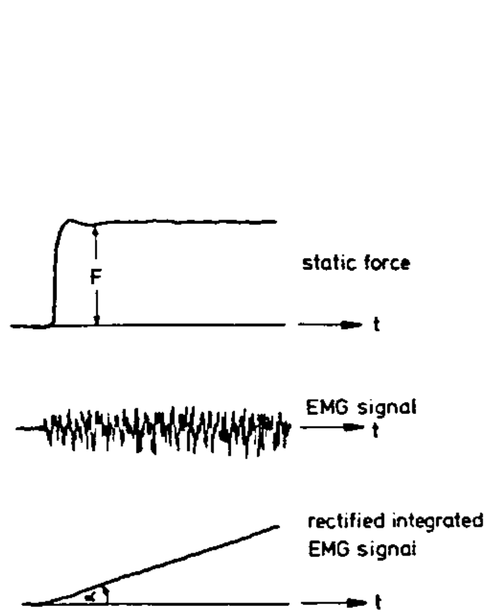

FIGURE 5-21
Registration of the force, the EMG signal, and the rectified and integrated EMG signal (schematically). The angle was evaluated as a measure of the total EMG activity.

Reprinted with permission from Rau G, Vredenbregt J: EMG-force relationship during voluntary static contractions (m. biceps). In Cerquiglini S, et al (eds): Biomechanics: III. Medicine and Sport. Baltimore, MD, University Park Press, 1973, vol 8, pp 270-274, Figure 1, p 272.

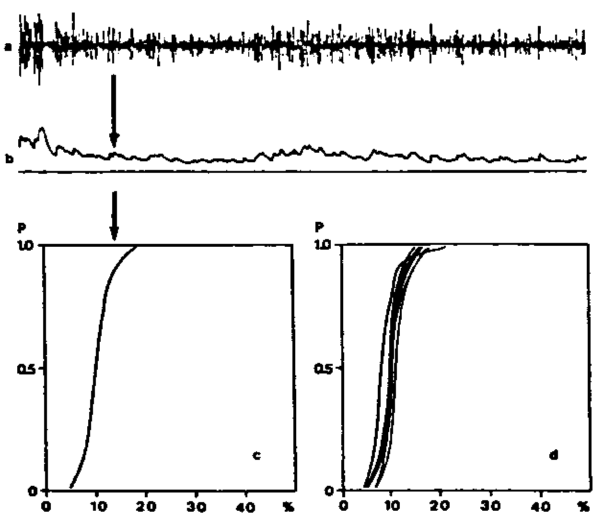

FIGURE 5-20
Raw EMG and the upper portion of the right trapezius (a), with the corresponding RMS detected and low pass filtered EMG (b), and the amplitude probability distribution curve (c), for one assembly task in an electronic industry (assembling telephone jacks), as well as the amplitude probability distribution curves for all six tasks involved in the job rotation (d).

Reprinted with permission from Jonsson B: The static load component in muscle work. Eur J Appl Physiol 57:305-310, 1988, Figure 2, p 307.

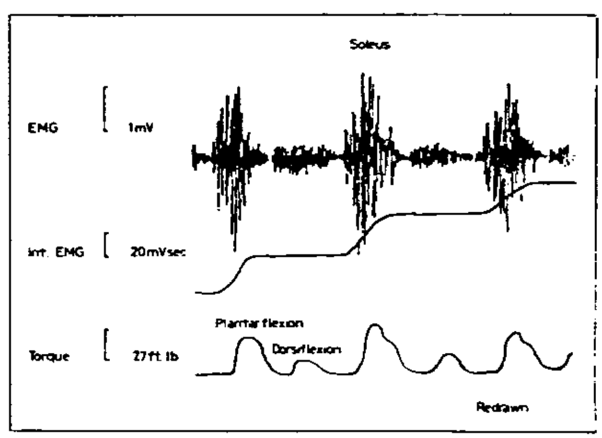

FIGURE 5-22
Recording of isokinetic movement at 216°/s. From above downward, soleus EMG, integrated EMG, torque, of ankle plantar flexion and dorsiflexion.

Reprinted with permission from Nelson AJ, et al: The relationship of integrated electromyographic discharge to isokinetic contractions. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1, pp 584-595, Figure 1, p 586.

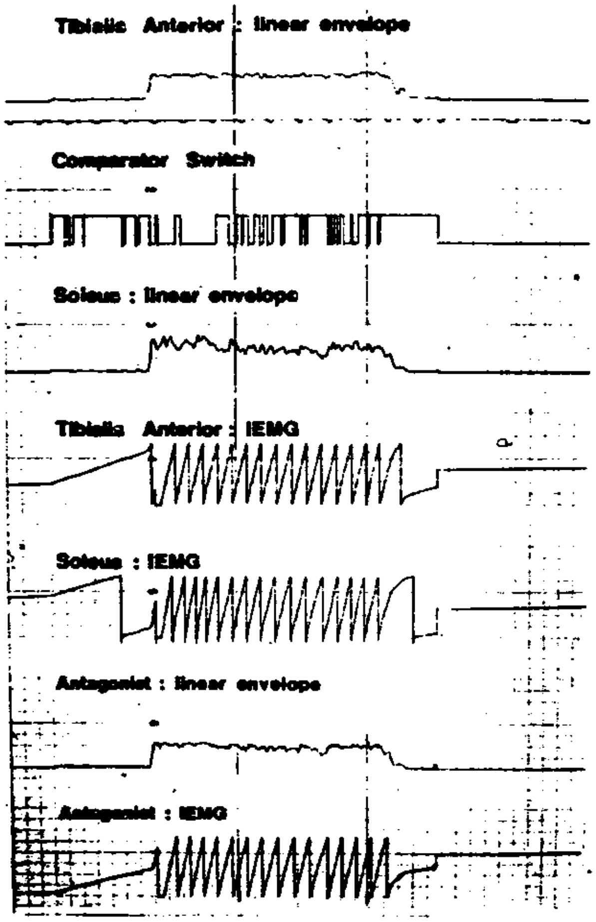

FIGURE 5-23

Typical processed EMG signals required for the co-contraction calibration. If linear envelope of tibialis anterior and soleus are identically equal during this voluntary isometric co-contraction, the IEMG of each muscle will be equal and will be equal to the antagonist IEMG. However, 100% co-contraction is not quantified because of the momentary imbalance in muscle activity, and the normal noise present in the linear envelope signal; levels of CC are 90% or higher.

Reprinted with permission from Falconer K, Winter DA: Quantitative assessment of co-contraction at the ankle joint in walking. Electromyogr Clin Neurophysiol 25:135-149, 1985, Figure 3, p 141.

(Figure 5-35 A) and manual spike counting (Figure 5-35 B). Four levels of muscle force productions (25%, 50%, 75%, 100%) were analyzed. Digital spike counting was done for three separate levels of amplitude (25%, 50%, 75%), each providing different counts. Manual spike counting showed little ability to discriminate among the four force levels and had a nonsignificant relationship with IEMG (Figure 5-36). The digital spike counting increased its level of correlation with IEMG, with increases in the amplitude level to 75% ($r = .37$). Although at 50% and 75% levels a significant relationship was found between digital spike counting, this relationship has little practical meaning for the ergonomist.

Turns

The number of times the myoelectric signal changes direction also is related to the frequency of the raw signal. Several turns may occur without the signal crossing the baseline. A turn is defined as that point where the direction of the signal changes following an amplitude difference of more than 100 mV. The number of turns increases rapidly as muscle force of low levels increases, but increases very slowly at high levels of muscle force. The number of turns reaches its maximum before the maximum muscle force is reached. Turns analysis discriminates well between low level muscle forces, but it discriminates poorly at high levels.[92]

This method is used more often in clinical studies with needle electrodes than in kinesiological studies. Thus, the application to ergonomics is very limited.

APPLICATIONS

In this section, the methods that can be used to evaluate the myoelectric signal are summarized. The instrumentation used to obtain each type of signal is detailed in Chapter 4, and the factors that may affect the interpretation are discussed in more detail in Chapter 6.

Linear Envelope

The linear envelope is obtained after full wave rectification of the raw signal followed by filtering (using a linear envelope detector). The filter must have a sufficiently short time constant to follow changes in the myoelectric changes and must be long enough to produce effective averaging (25–300 msec).[13] The resistive-capacitive (RC) network should have a time constant greater than the spike duration.[95] The RC network has a slow roll-off for frequencies above its cutoff time constant. Thus, long time constants must be used to provide sufficient smoothing of brief, high amplitude peaks.[52] A Paynter or Butterworth filter may be used, but such filters may have difficulty providing a variable time constant.[52]

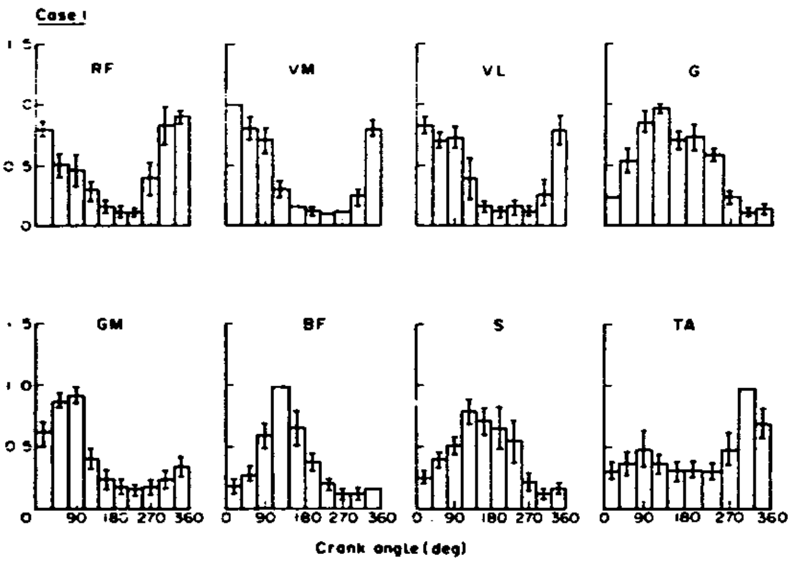

FIGURE 5-24

Average IEMG results for the eight muscles. Vertical lines indicate ± ½ SD.

Reprinted with permission from Jorge M, Hull ML: Analysis of EMG measurements during bicycle pedaling. J Biomech 19:683-694, 1986, Figure 5, p 689.

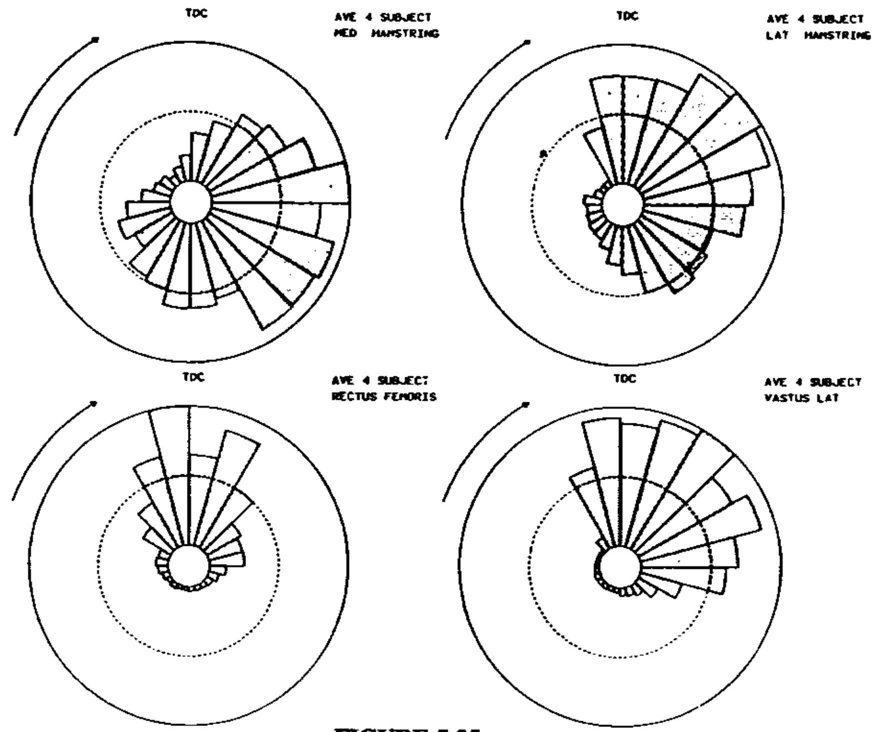

FIGURE 5-25

Average integrated EMG patterns for the medial hamstring, lateral hamstring, rectus femoris, and vastus lateralis muscles. Each shaded section represents percentage of maximum activity over 15° of the pedaling cycle in four of the five subjects. The dashed line represents 50% of maximum activity observed in that muscle.

Reprinted with permission from Gregor RJ et al: Knee flexor moments during propulsion in cycling: A creative solution to Lombard's Paradox. J Biomech 18:307-316, 1985, Figure 6, p 113.

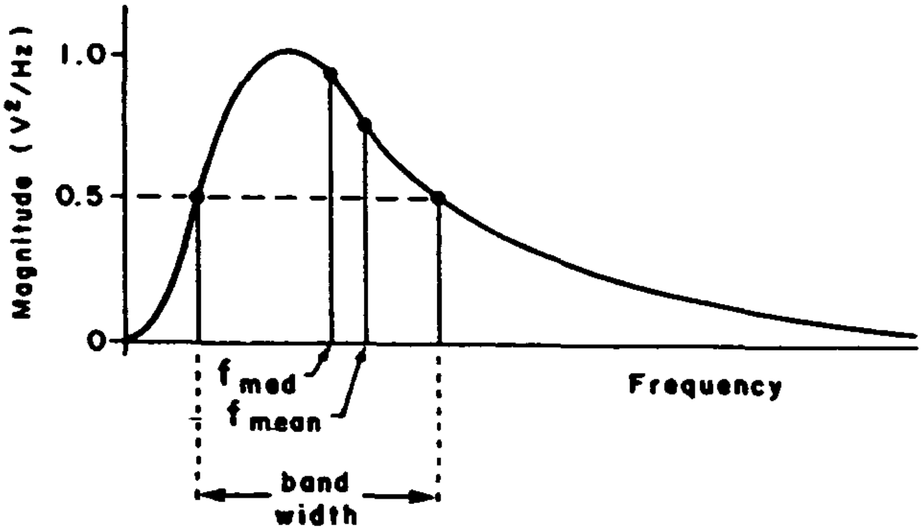

FIGURE 5-26
An idealized version of the frequency spectrum of the EMG signals. Three convenient and useful variables: the median frequency, f_{med}; the mean frequency, f_{mean}; and the bandwidth are indicated.

Reprinted with permission from Basmajian JV, DeLuca CJ: Muscles Alive: Their Functions Revealed by Electromyography, ed 5. Baltimore, MD, Williams & Wilkins, 1985, Figure 3.16, p 99.

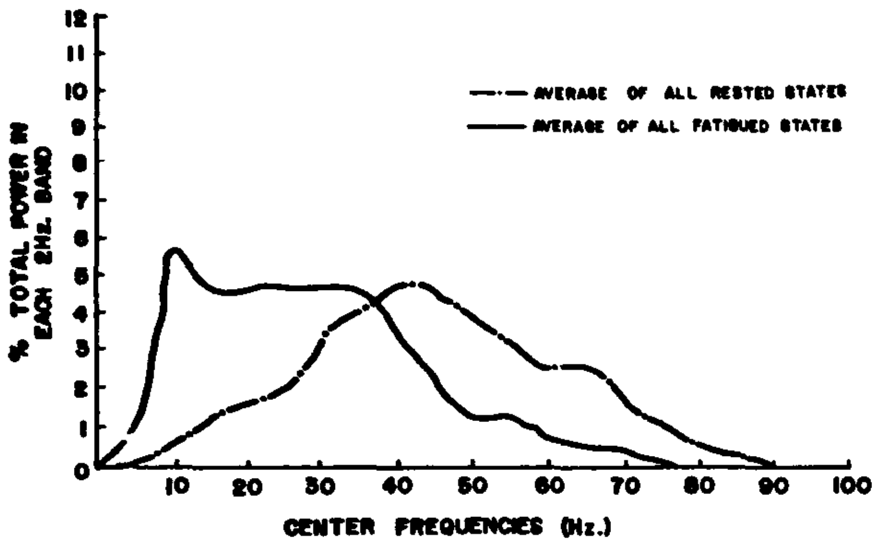

FIGURE 5-27
Average EMG spectra with reference to fatigue level.

Reprinted with permission from Chaffin DB: Localized muscle fatigue: Definition and measurement. J Occup Med 15:346-354, 1973, Figure 2, p 347.

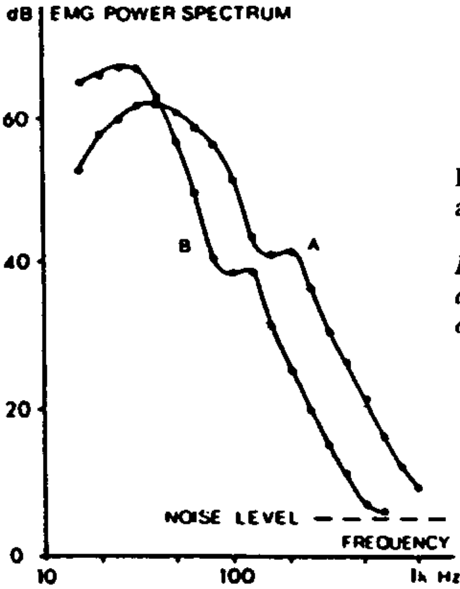

FIGURE 5-28

Electromyographic power spectra obtained at 2 kilopond loads. A) Before, and B) after 30-seconds maximum load.

Reprinted with permission from Lindstrom L et al: Muscular fatigue and action potential conduction velocity changes studies with frequency analysis of EMG signals. Electromyography 10:341-356, 1970, Figure 2, p 347.

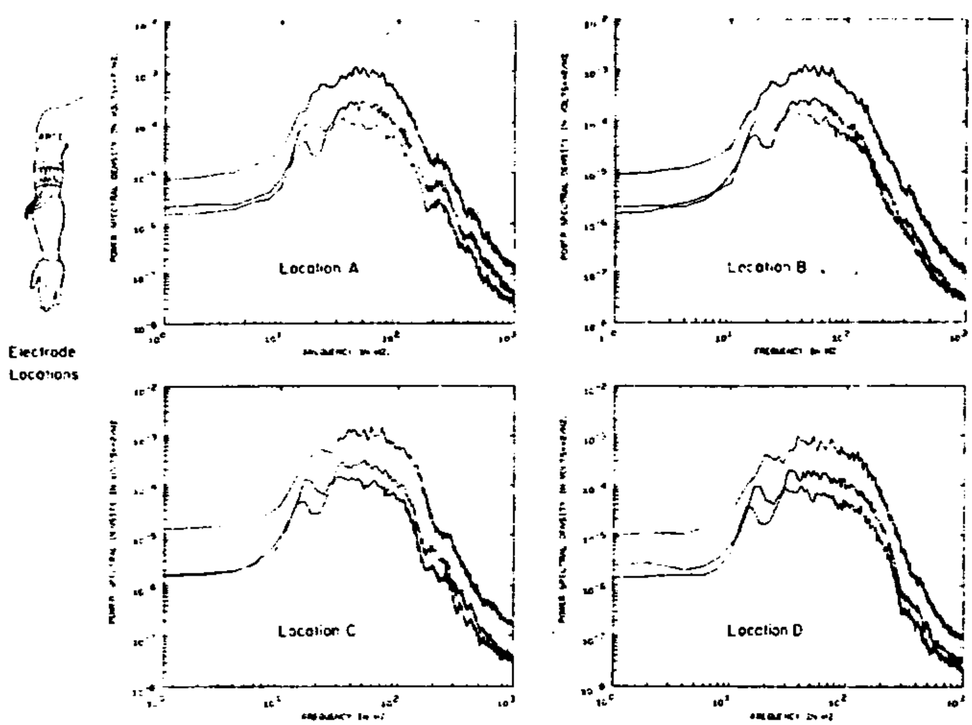

FIGURE 5-29

Logarithmic plots of myoelectric signal power spectra obtained at three contraction levels (5%, 10%, and 25% of maximum voluntary contraction) from four locations across the biceps brachii of able-bodied subject D.L.

Reprinted with permission from Hogan N, Mann RW: Myoelectric signal processing: Optimal estimation applied to electromyography: II. Experimental demonstration of optimal myoprocessor performance. IEEE Trans Biomed Eng 27:396-410, 1980, Figure 5, p 399.

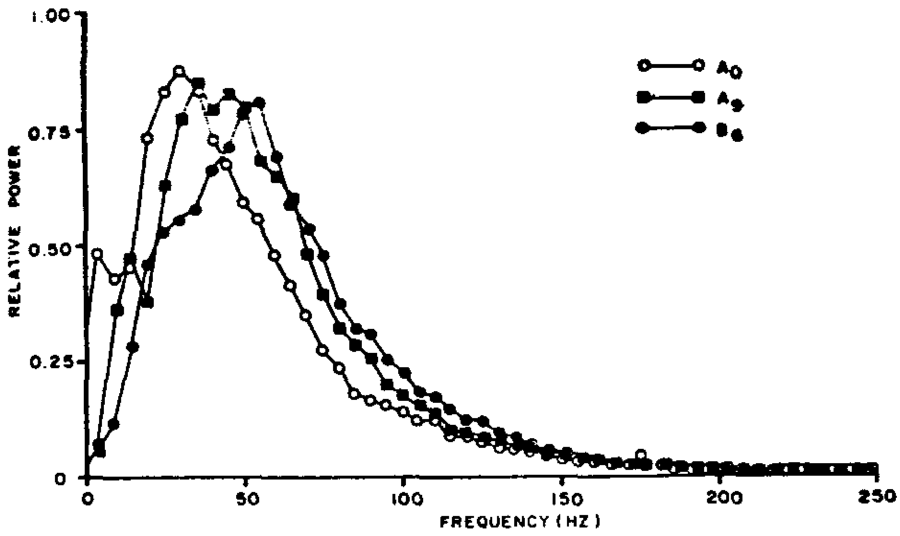

FIGURE 5-30

Average power spectra for all six subjects; each point is the mean of 22 values 0 – 0 0.2 Nm appled torque; ■ – ■ 9 Nm applied torque; • – • 35 Nm applied torque.

Reprinted with permission from Gander RE, Hudgins BS: Power spectral density of the surface myoelectric signal of the biceps brachii as a function of static load. Electromyogr Clin Neurophysiol 25:469-478, 1985, Figure 2, p 473.

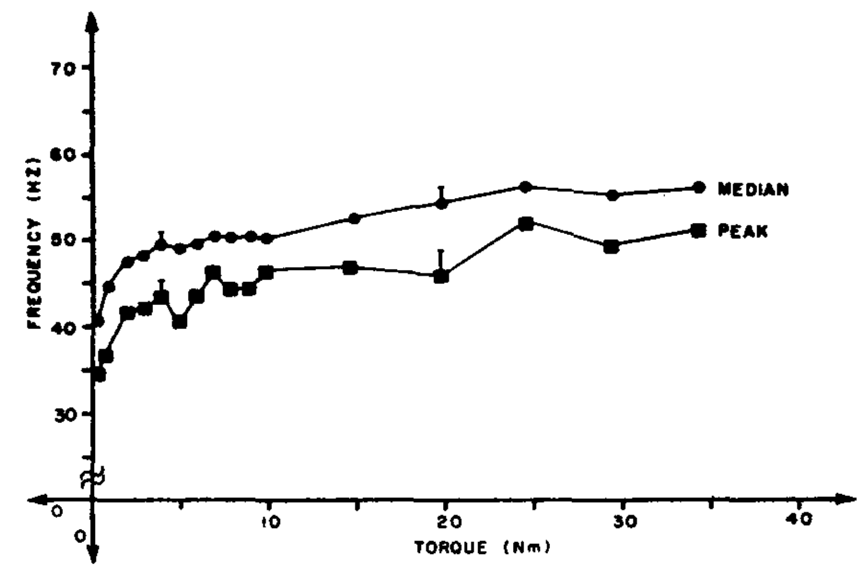

FIGURE 5-31

Median frequency (• – •) and frequency of the peak of the spectrum ■ – ■ as a function of applied torque. Each point is the mean of 22 values, and the vertical bars represent typical standard errors.

Reprinted with permission from Gander RE, Hudgins BS: Power spectral density of the surface myoelectric signal of the biceps brachii as a function of static load. Electromyogr Clin Neurophysiol 25:469-478, 1985, Figure 3, p 473.

The linear envelope represents the number of motor units firing their firing rate, the area of the motor unit, and the amount of cancellation from superposition.[94] The linear envelope is generally proportional to amplitude, duration, and firing rate of the motor units from the muscle being studied. It can provide relative amplitude and instantaneous amplitude information. Note that some filters may cause a lag in the envelope signal, which will affect the instantaneous signal information.[13,38,52]

Integration

Integration requires full wave rectification followed by filtering and the use of an integrator. Digital integration algorithms may be used.[94] Integration uses all parts of the signal and represents the total amount of energy of the signal.[38,65]

The processed signal represents the number of motor units firing, their firing rate, the area of the motor unit, and the amount of cancellation from superposition.[38,94] Integration is proportional to motor unit amplitude, duration, and rate of firing.[38] It is independent of propagation velocity.[65]

Integration fails to discriminate between artifacts and motor units.[94] Noise also may be a problem when recording low force level contractions. Large window (time) sampling durations may detect an unacceptably large proportion of noise when investigating low force level contractions.[95]

Root Mean Square

To obtain the RMS value, a ballistic galvanometer, thermocouple, strongly damped voltmeter, or digital computer may be used.[93] A nonlinear detector may be used instead of a linear detector.[70] The RMS may also be calculated from the power spectrum (moment over zero) or from the squared value of the signal in the time domain.[65] The RMS voltage determination and integration techniques are essentially equivalent.[96]

The RMS signal depends on the number of motor units firing, their firing rates, and the area of the motor units. The signal is affected by the cross-correlation between motor units; it does not appear to be affected by cancellation from motor unit superposition or by synchronization.[94] The signal amplitude is also inversely proportional to the propagation velocity.[65]

The RMS signal has immediate relationship to the power spectrum. The curve represents the power of the myoelectric signal. The power value is equal to the area under the spectral curve.[52] The signal amplitude is proportional to the square root of the total signal power.[65]

The RMS value is usually proportional to the mean amplitude value,[55,69] but is related to motor unit and firing rate by the square root.[10,54]

The RMS signal appears to have a linear relationship with tension for brief isometric contractions.[8,10,54,55] The RMS signal sensitivity to recruitment and firing rate is different from IEMG and yields a different relationship to the force produced.[65]

The RMS values may provide inaccurate information if using them to determine fatigue, because factors other than fatigue may be affecting the signals (see Chapter 6). Use of RMS is a preferred method of interpreting the myoelectric signal.[95]

Frequency Spectrum

Early investigation used octave band filters, but fast Fourier transformation (FFT) methods are now being used more frequently.[13] The square of the magnitude of FFT is what is being analyzed. This important element in a set of mathematical and statistical concepts is ideally suited both for evaluation of the myoelectric signal and the signal generating physiological and pathophysiological mechanisms. The power spectrum provides a detailed picture of total myoelectric power valid for all levels of contraction.[97]

The frequency spectrum is affected by changes in the duration and shape of the involved motor units, but not generally affected by the firing rate or amplitude. Its shape is independent of exerted force.[93] The mean frequency changes reflect basic physiological responses to muscular contraction and seem to be good for study of fatigue.[55]

The center frequency is not affected by force of contraction except at low levels where it slightly increases with an increase in force.[93] The center frequency is sensitive to changes in the conduction velocity.[30,51] Petrofsky warned that because of its response to changes in muscle temperature the center frequency may not be useful to quantify force and fatigue.[70] Baidya and Stevenson, however, state that center frequency is a reliable measure of local muscle fatigue in repetitive work.[73] The logarithmic scale means of presentation allows a wide dynamic range so that influences often lost in visual inspection may accurately be retained.[97]

Some advantages[65] of the spectrum analysis are as follows:

1. Insensitive to interference between motor unit contributions and responds to single motor units or whole muscle signals. This characteristic extends its range and scope over conventional methods.

2. Quantitative, so it allows comparisons between repeated investigations done over a period of time.
3. Easily done on computers.
4. Provides the possibility to relate the myoelectric signal to physiologic events.

Some of the disadvantages[65] include the following:

1. Use of the FFT gives an average of the myoelectric signal, and single details may be hidden.
2. Certain measures related to recruitment of motor units and their firing rates are not shown in the spectral description.
3. Certain motor unit contributions of high amplitude or those frequently repeated will dominate the shape of the power spectrum. (This may be controlled by electrode placement.)
4. The spectrum may collect low level frequencies from other muscles, which may be interpreted as part of the investigated muscle's signal.

Zero Crossings

The information from the myoelectric signal is processed by an analog filter followed by conversion from analog to digital values. The zero crossings then are obtained by appropriate software.[38]

Large numbers of action potentials of similar shape and size may saturate the zero count. The IEMG may be a better method to use in this situation. Lower numbers of recruited motor units may generate poorly fused and noisy integrals.[38] Zero crossing should be used with care at low levels. Background noise and low numbers of active motor units may invalidate the results. Appropriate levels of 20 dB above noise level should be used as test contractions.[66,68]

Spike Countings

Spike detection requires full wave rectification followed by manual or electronic counting.[88] This method is used less now than in the past. Spike detection discards much myoelectric information and concentrates on the rate random events. The peaks may be unduly influenced by large numbers of fast-twitch, fatigable motor units than by the small less fatigable motor units.[38] Spike counting is more suitable for low force levels. As the myoelectric signal increases, the spikes tend to interfere with each other, and the count becomes invalid.[13,93,98]

Turns

The number of turns are related to the number of motor units, firing rate of the motor units, duration of motor units, and number of polyphasic units.[65,93] The number of turns increases with the increase in force at low levels up to about 30% to 50%.[52,92,93] This method is an excellent means of discriminating between low levels of muscle activity, but it discriminates poorly at high levels.[95]

SUMMARY

There are numerous forms of EMG output available to the ergonomist. This chapter provides the basis for the use of the various forms, presents research data that has been developed regarding their use, and discusses applications of these techniques to ergonomic studies. Use of these methods are important because they form the basis for understanding the muscle activity of humans during their performance of activities required to participate in any functional activity.

REFERENCES

1. Lawrence JH, DeLuca CJH: Myoelectric signal versus force relationship in different human muscles. J Appl Physiol 54: 1653-1659, 1983
2. Ericson MO, Nisell R, Arborelius UP, et al: Muscular activity during ergometer cycling. Scand J Rehabil Med 17:53-61, 1985
3. Bagg SD, Forrest WJ: Electromyographic study of the scapular rotators during arm abduction in the scapular plane. Am J Phys Med 65:111-124, 1986
4. Woods JJ, Bigland-Ritchie B: Linear and nonlinear surface EMG/force relationships in human muscles. Am J Phys Med 62:287-299, 1983
5. Seroussi RE, Pope MH: The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. J Biomech 20:135-146, 1987
6. Cook TM, Neumann DA: The effects of load placement on the EMG activity of the low back muscles during load carrying by men and women. Ergonomics 30:1413-1423, 1987
7. Lippold OCJ: The relation between integrated action potentials in a human muscle and its isometric tension. J Physiol (London) 117:492-499, 1952
8. Bigland B, Lippold OCT: The relationships between forces, velocity and integrated electrical activity in human muscles. J Physiol (London) 123:214-224, 1954
9. Rau G, Vredenbregt J: EMG-force relationship during voluntary static contractions (m. biceps). In Cerquiglini S, et al (eds): Biomechanics: III. Medicine and Sport. Baltimore, MD, University Park Press, 1973, vol 8, pp 270-274
10. Milner-Brown HS, Stein RB: The relation between the surface electromyogram and muscular force. J Physiol (London) 246:549-569, 1975

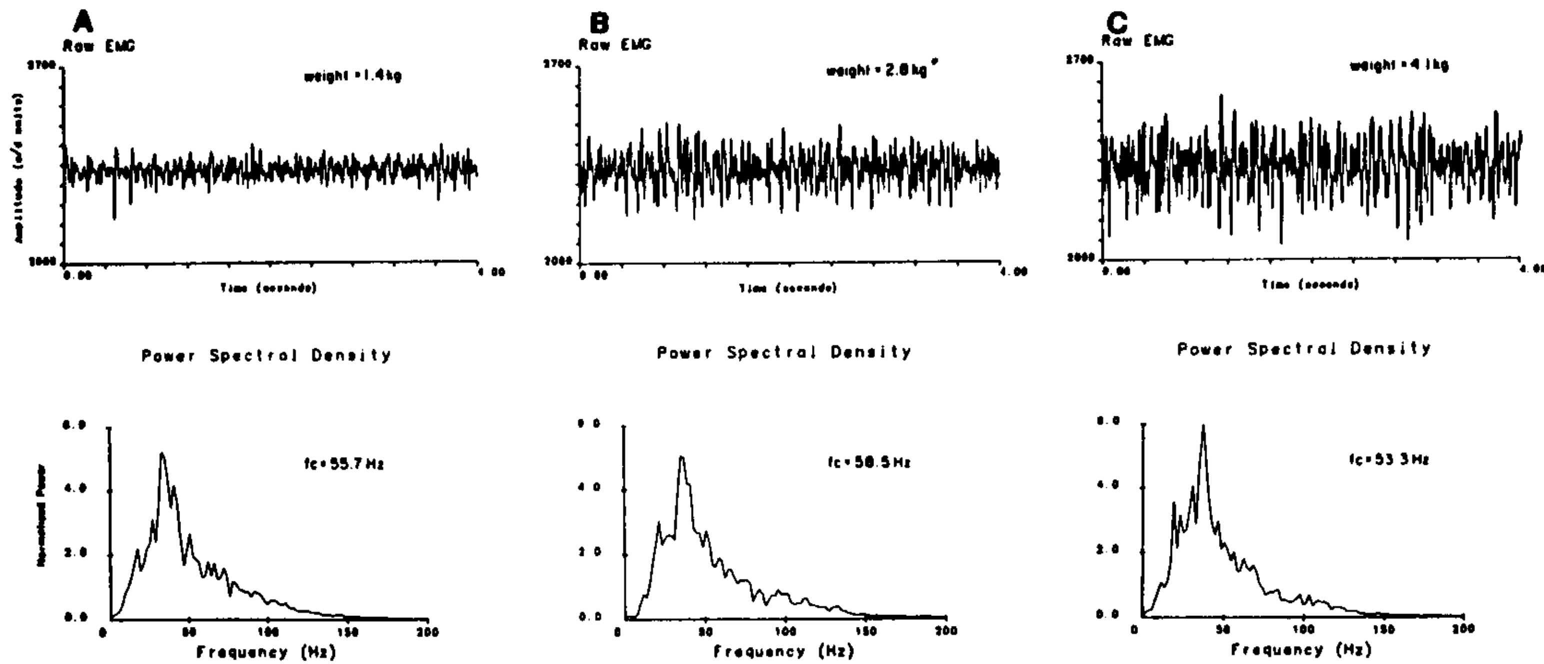

FIGURE 5-32

Raw electromyogram (EMG) data (top) and corresponding EMG power spectral density (bottom) in 1 subject with weight of 1.4 kg (A), 2.8 kg (B), and 4.1 kg (C) held at longer length *L*1. Note that amplitude of raw EMG signal increases with increasing weight, but centroid frequency (f_c) changes little and inconsistently as weight changes ($P > 0.1$).

Reprinted with permission from Bazzy AR, et al: Increase in electromyogram low-frequency power in nonfatigued contracting skeletal muscle. J Appl Physiol 61:1012-1017, 1986, Figure 3, p 1016.

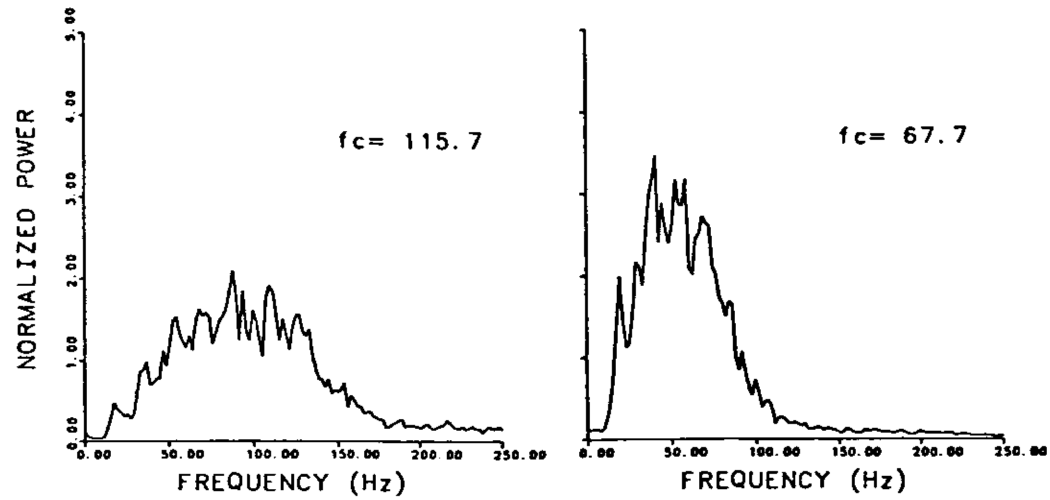

FIGURE 5-33

Power spectral density (PSD) of biceps brachii muscle electromyogram of 1 subject at long length $L1$ (right) and shorter length $L2$ (left). Note shift to left in PSD from $L2$ to $L1$ f_c (centroid frequency).

Reprinted with permission from Bazzy AR, et al: Increase in electromyogram low-frequency power in nonfatigued contracting skeletal muscle. J Appl Physiol 61:1012-1017, 1986, Figure 2, p 1014.

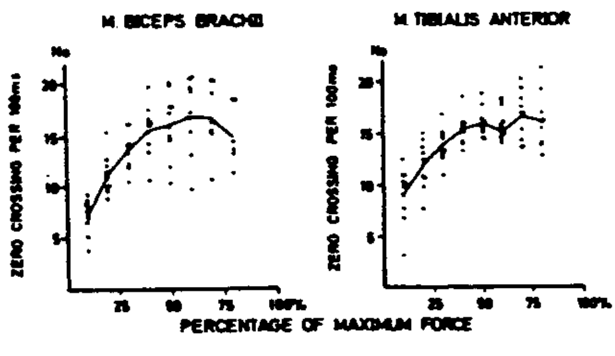

FIGURE 5-34

Number of zero crossings at 100 μV related to gradually increasing force. Each point is the mean of 3 recordings, and the lines combine the mean values (m. biceps brachii 10 subjects and m. tibialis anterior 10 subjects).

Reprinted with permission from Christensen H, Lo Monaco M, Dahl K, et al: Processing of electrical activity in human muscle during a gradual increase in force. Electroencephalogr Clin Neurophysiol 58:230-239, 1984, Figure 5, p 234.

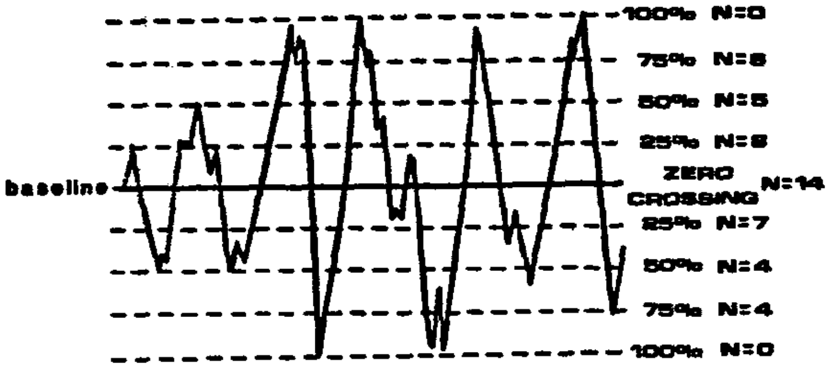

FIGURE 5-35 A
Spike counting by window.

Reprinted with permission from Robertson RN, Grabiner MD: Relationship of IEMG to two methods of counting surface spikes during different levels of isometric tension. Electromyogr Clin Neurophysiol 25:489-498, 1985, Figure 1, p 492.

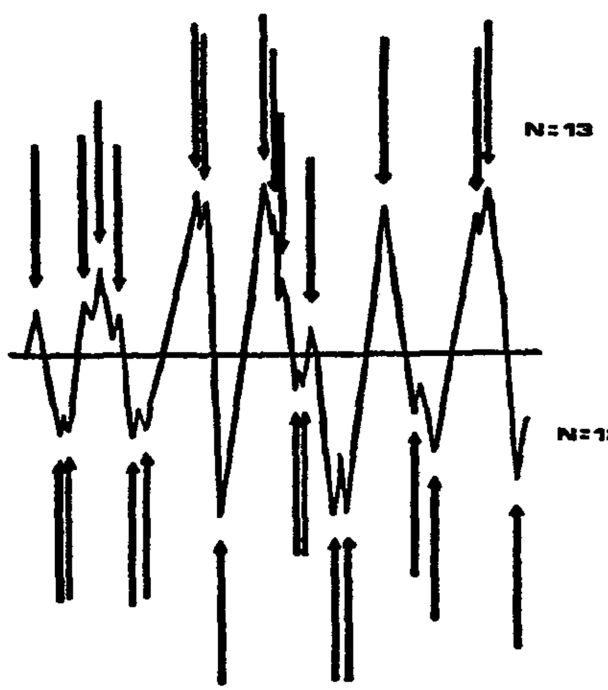

FIGURE 5-35 B
Manual spike counting of raw EMG signal.

Reprinted with permission from Robertson RN, Grabiner MD: Relationship of IEMG to two methods of counting surface spikes during different levels of isometric tension. Electromyogr Clin Neurophysiol 25:489-498, 1985, Figure 4, p 493.

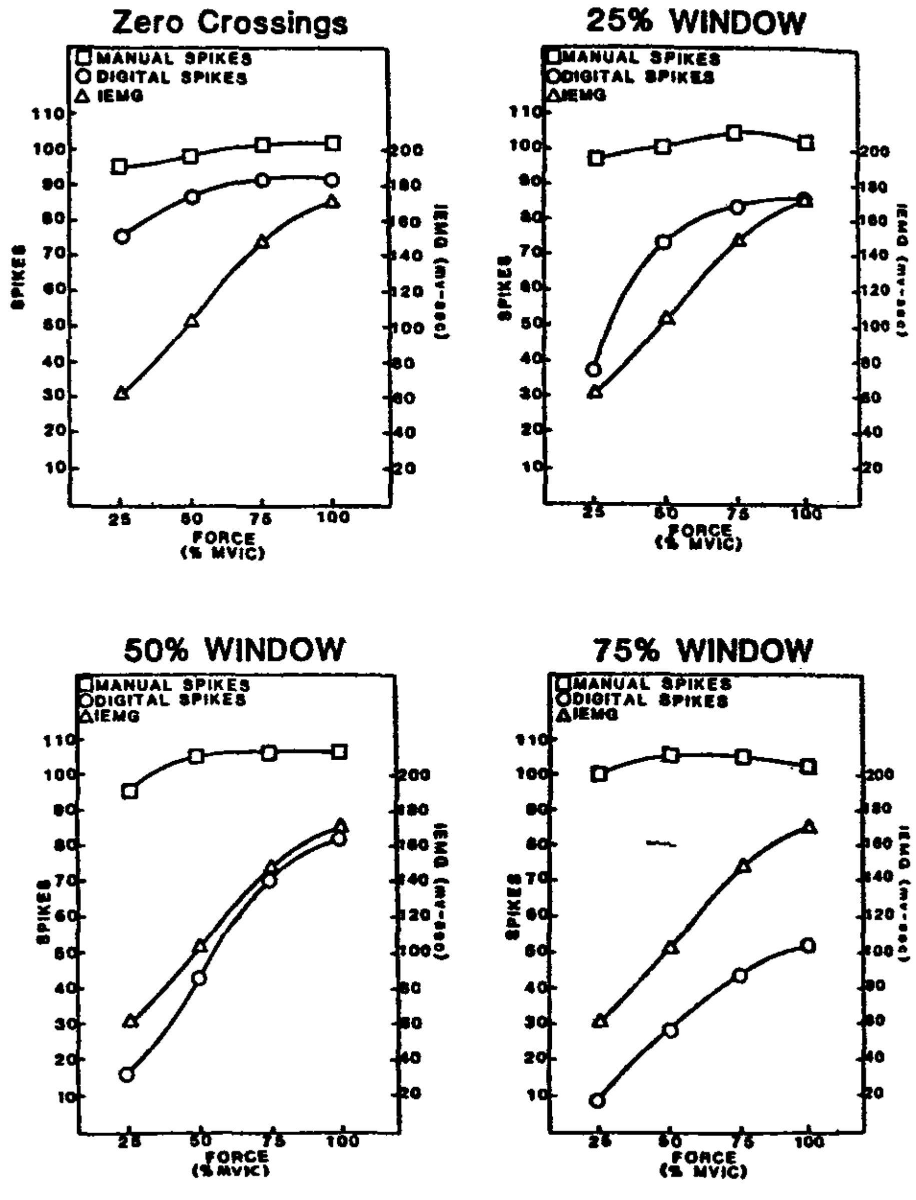

FIGURE 5-36
Relationship between IEMG and manual and digital spikes as a function of tension by window.

Reprinted with permission from Robertson RN, Grabiner MD: Relationship of IEMG to two methods of counting surface spikes during different levels of isometric tension. Electromyogr Clin Neurophysiol 25:489-498, 1985, Figures 6-9, p 496.

11. Komi PV, Viitasalo JHT: Signal characteristics of EMG at different levels of muscle tension. Acta Physiol Scand 96:267-276, 1976
12. Moritani T, deVries HA: Reexamination of the relationship between the surface integrated electromyogram (IEMG) and force of isometric contraction. Am J Phys Med 57:263-277, 1978
13. Perry J, Bekey GA: EMG-force relationships in skeletal muscle. CRC Critical Review in Biomedical Engineering 7:22, 1981
14. Viitasalo JHT, Komi PV: Signal characteristics of EMG with special reference to reproducibility of measurements. Acta Physiol Scand 93:531-539, 1975
15. Solomonow M, Baratta R, Zhou BH, et al: Historical update and new developments on the EMG-force relationships of skeletal muscles. Orthopedics 9:154-1543, 1986
16. Yang JF, Winter DA: Electromyography reliability in maximal and submaximal contractions. Arch Phys Med Rehabil 64:417-420, 1983
17. Inman VT, Ralston JH, Saunders JB de CM, et al: Relation of human electromyogram to muscular tension. Electroencephalogr Clin Neurophysiol 4:187-194, 1952
18. Chaffin DB, Myun L, Freivalds A: Muscle strength assessment from EMG analysis. Med Sci Sports Exercise 12:205-211, 1980
19. Heckthorne CW, Childress DS: Relationships of the surface electromyogram to the force length, velocity and contraction rate of the cineplastic human biceps. Am J Phys Med 60:-19, 1981
20. Winkel J, Bendix T: Muscular performance during seated work evaluated by two different EMG methods. Eur J Appl Physiol 55:167-173, 1986
21. Andersson GBJ, Schultz AB, Ortengren R: Trunk muscle forces during desk work. Ergonomics 29:1113-1127, 1986
22. Jonsson B, Hagberg M: The effect of different working heights on the deltoid muscle. Scand J Rehabil Med [Suppl] 3:26-32, 1974
23. Janda DH, Geiringer SR, Hankin FM, et al: Objective evaluation of grip strength. J Occup Med 29:569-571, 1987
24. Bobet J, Norman RW: Effects of load placement on back muscle activity in load carriage. Eur J Appl Physiol 53:7-75, 1984
25. Gregor RJ, Gavanagh PR, LaFortune: Knee flexor moments during propulsion in cycling: A creative solution to Lombard's Paradox. J Biomech 18:307-316, 1985
26. Lyons K, Perry J, Gronely JK, et al: Timing and relative intensity of hip extensor and abductor muscle action during level and stair ambulation. Phys Ther 63:1597-1605, 1983
27. Winter D: Pathologic gait diagnosis with computer-averaged electromyographic profiles. Arch Phys Med Rehabil 65:393-395, 1984
28. Arborelius UP, Ekholm J, Nemeth G, et al: Shoulder joint load and muscular activity during lifting. Scand J Rehabil Med 18:71-82, 1986
29. Arborelius UP, Ekholm J, Nisell R, et al: Shoulder load during machine milking: An electromyographic and biomechanical study. Ergonomics 29:159-1607, 1986
30. Stulen FB, DeLuca CJ: Frequency parameters of the myoelectric signal as a measure of the conduction velocity. IEEE Trans Biomed Eng 7:515-523, 1981
31. Soderberg Gl, Cook TM: An electromyographic analysis of quadriceps femoris and muscle setting and straight leg raising. Phys Ther 63:1434-1438, 1983
32. Hagberg M: Muscular endurance and surface electromyogram in isometric and dynamic exercise: J Appl Physiol Respir Environ Exercise Physiol 51:1-7, 1981
33. Chaffin DB: Localized muscle fatigue: Definition and measurement. J Occup Med 15:346-354, 1973
34. Kadefors R: Myo-electric signal processing as an estimating problem. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1, pp 519-532
35. Sofranek MG, Koshland GF, Raymond G: Effect of auditory rhythm on muscle activity. Phys Ther 62:16-168, 1982
36. Maton B, LeBozec S, Cnockaert JC: The synergy of elbow extensor muscles during dynamic work in man: II. Braking elbow flexion. Eur J Appl Physiol 44:279-289, 1980
37. Perry J, Ireland ML, Gronely J, et al: Predictive values of manual muscle testing and gait analysis in normal ankles by dynamic electromyography. Foot Ankle 6:254-259, 1986
38. Loeb GE, Gans C: Signal processing and display. In Loeb GE, Gans C (eds): Electromyograph for Experimentalists. Chicago, IL, University of Chicago Press, 1986, pp 244-262
39. Hershler C, Milner M: An optimality criterion for processing electromyographic (EMG) signals relating to human locomotion. IEEE Trans Biomed Eng 25:413-420, 1978
40. Gottlieb GL, Agarwal GC: Filtering of electromyographic signals. Am J Phys Med 49:142-146, 1970
41. Close JR, Nickel ED, Todd FN: Motor-unit action-potential counts. J Bone Joint Surg [Am] 42:1207-1222, 1960
42. Arsenault AB, Winter DA, Marteniuk RG, et al: How many strides are required for the analysis of electromyographic data in gait? Scand J Rehabil Med 18:133-135, 1986
43. Schuldt K, Ekholm H, Harms-Ringdahl K, et al: Effects of changes in sitting work posture on static neck and shoulder muscle activity. Ergonomics 29:1525-1537, 1986
44. Komi PV, Rusko H: Quantitative evaluation of mechanical and electrical changes during fatigue loading of eccentric and concentric work. Scand J Rehabil Med [Suppl] 3:121-126, 1974
45. Grieve DW, Pheasant ST: Myoelectric activity, posture and isometric torque in man. Electromyogr Clin Neurophysiol 16:3-21, 1976
46. Harms-Ringdahl K, Ekholm H, Schuldt K, et al: Load moments and myoelectric activity when the cervical spine is held in full flexion and extension. Ergonomics 29:1539-1552, 1986
47. Nemeth G: On hip and lumbar biomechanics: A study of joint load and muscular activity. Scand J Rehabil Med [Suppl] 10:4-35, 1984
48. Arsenault AB, Winter DA, Marteniuk RG: Treadmill versus walking locomotion in humans: An EMG study. Ergonomics 29:665-676, 1986

49. Ericson MO, Nisell R, Ekholm J: Quantified electromyography of lower-limb muscles during level walking. Scand J Rehabil Med 18:159-163, 1986
50. Schuldt K, Ekholm H, Nemeth G, et al: Knee load and muscle activity during exercises in rising. Scand J Rehabil Med [Suppl] 9:174-188, 1983
51. Hermens HJ, Boon KL, Zilvold G: The clinical use of surface EMG. Acta Belg Med Phys 9:119-130, 1986
52. Lindstrom L, Petersen I: Power spectra of myoelectric signals: Motor activity and muscle fatigue. In Stalberg E, Young RR (eds): Neurology: I. Clinical Neurophysiology. Stoneham, MA, Butterworth Publishers, 1981, pp 66-87
53. DeVries HA: "Efficiency of electrical activity" as a physiological measure of the functional state of muscle tissue. Am J Phys Med 47:10-22, 1968
54. Lind AR, Petrofsky JS: Amplitude of the surface electromyogram during fatiguing isometric contractions. Muscle Nerve 2:257-264, 1979
55. Petrofsky JS, Lind AR: The influence of temperature on the amplitude and frequency components of the GM during brief and sustained isometric contractions. Eur J Appl Physiol 44:189-200, 1980
56. Hagberg M, Sundelin G: Discomfort and load on the upper trapezius muscle when operating a word processor. Ergonomics 29:1637-1645, 1986
57. Jorgensen K, Fallentin N, Krogh-Lund C, et al: Electromyography and fatigue during prolonged, low-level static contractions. Eur J Appl Physiol 57:316-321, 1988
58. Jonsson B: The static load component in muscle work. Eur J Appl Physiol 57:305-310, 1988
59. Nelson AJ, Moffroid M, Whipple R: The relationship of integrated electromyographic discharge to isokinetic contractions. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1, pp 584-595
60. Falconer K, Winter DA: Quantitative assessment of cocontraction at the ankle joint in walking. Electromyogr Clin Neurophysiol 25:135-149, 1985
61. Jorge M, Hull ML: Analysis of EMG measurements during bicycle pedaling. J Biomech 19:683-694, 1986
62. Gravel D, Arsenault AB, Lambert J: Soleus-gastrocnemius synergies in controlled contractions produced around the ankle and knee joints: An EMG study. Electromyogr Clin Neurophysiol 27:405-413, 1987
63. Richardson G, Bullock MI: Changes in muscle activity during fast, alternating flexion-extension movement of the knee. Scand J Rehabil Med 18:5-58, 1986
64. Miller D: Comparison of electromyographic activity in the lumbar paraspinal muscles of subjects with and without chronic low back pain. Phys Ther 65:1347-1354, 1985
65. Lindstrom L, Petersen I: Power spectrum analyis of EMG signals and its application. In Desmedt JE (ed): Computer-Aided Electromyopgraphy: Progress in Clinical Neurophysiology. Basel, Switzerland, Karger, 1983, vol 10, p 5
66. DeLuca CJ: Myoelectrical manifestations of localized muscular fatigue in humans. CRC Critical Review in Biomedical Engineering 11:251-279, 1985
67. Shochina M, Gonen B, Wolf E, et al: A study of the application of the Fourier series in the analysis of the electrophysiological pattern of development of muscle fatigue. Electromyogr Clin Neurophysiol 26:513-520, 1986
68. Sadoyama T, Masuda T, Miyano H: Relationships between muscle fiber conduction velocity and frequency parameters of surface EMG during sustained contraction. Eur J Appl Physiol 51:247-256, 1983
69. Kadefors R, Arvidsson A, Herberts P, et al: EMG data processing in ergonomics with emphasis on shoulder muscle load. In: Desmedt JE (ed):Computer-Aided Electromyography: Progress in clinical Neurophysiology. Basel, Switzerland, Karger, 1983, vol 10, pp 273-286
70. Petrofsky JS: Frequency and amplitude analysis of the EMG during exercise on the bicycle ergometer. Eur J Appl Physiol 41:1-15, 1979
71. Kuorinka I: Restitution of EMG spectrum after muscular fatigue. Eur J Appl Physiol 57:311-315, 1988
72. Komi PV, Tesch P: EMG frequency spectrum, muscle structure, and fatigue during dynamic contractions in man. Eur J Appl Physiol 42:4-50, 1979
73. Baidya KN, Stevenson MG: Local muscle fatigue in repetitive work. Ergonomics 31:227-239, 1988
74. Broman H, Magnusson R, Petersen I, et al: Vocational electromyography. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 1, pp 519-532
75. 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
76. Paiss O, Inbar GF: Autoregressive modeling of surface EMG and its spectrum with application to fatigue. IEEE Trans Biomed Eng 34:761-770, 1987
77. Okada M: Effect of muscle length on surface EMG wave forms in isometric contractions. Eur J Appl Physiol 56:482-486, 1987
78. Huijing PA, Adelerbof A, Giesberger R, et al: Triceps source EMG power spectrum changes during sustained submaximal isometric contractions at different muscle lengths. Electromyogr Clin Neurophysiol 26:181-192, 1986
79. Bigland-Ritchie B, Johansson R, Lippold OGJ, et al: Changes in motoneuron firing rates during sustained maximal contractions. J Physiol (London) 340:335-346, 1983
80. Hary D, Belman MJ, Propst J, et al: A statistical analysis of the spectral moments used in EMG test of endurance. J Appl Physiol 53:779-783, 1982
81. Lindstrom L, Magnusson R, Petersen I: Muscular fatigue and action potential conduction velocity changes studied with frequency analysis of EMG signals. Electromyography 10:341-356, 1970
82. Matthijsse PC, Hendrich K, Rijnsburger WH, et al: Ankle angle effects on endurance time, median frequency and mean power of gastroenemius EMG power spectrum: A comparison between individual and group analysis. Ergonomics 30:1149-1159, 1987
83. Hogan N, Mann RW: Myoelectric signal processing: Optimal estimation applied to electromyography: Part II. Experimental demonstration of optimal myoprocessor performance. IEEE Trans Biomed Eng 27:396-410, 1980

84. Gander RE, Hudgins BS: Power spectral density of the surface myoelectric signal of the biceps brachii as a function of static load. Electromyogr Clin Neurophysiol 25:469-478, 1985
85. Blinowska A, Verroust J: Low frequency power spectrum of the EMG signal. Electromyogr Clin Neurophysiol 27:349-353, 1987
86. Bazzy AR, Karten JB, Haddad GG: Increase in electromyogram low-frequency power in nonfatigued contracting skeletal muscle. J Appl Physiol 61:1012-1017, 1986
87. Christensen H, Lo Monaco M, Dahl K, et al: Processing of electrical activity in human muscle during a gradual increase in force. Electroencephalogr Clin Neurophysiol 58:230-239, 1984
88. Hagg GM, Suurkula J, Liew M: A worksite method for shoulder muscle fatigue measurements using EMG, test contractions and zero crossing technique. Ergonomics 30:1541-1551, 1987
89. Suurkula J, Hagg GM: Relations between shoulder/neck disorders and EMG zero crossing shifts in female assembly workers using the test contraction method. Ergonomics 30:1553-1569, 1987
90. Bergstrom RM: The relation between the number of impulses and the integrated electrical activity in the electromyogram. Acta Physiol Scand 45:97-101, 1959
91. Robertson RN, Grabiner MD: Relationship of IEMG to two methods of counting surface spikes during different levels of isometric tension. Electromyogr Clin Neurophysiol 25: 489-498, 1985
92. Gilai A: Continuous EMG interference pattern analysis. Electromyogr Clin Neurophysiol 27:339-348, 1987
93. Fuglsang-Frederiksen A: Electrical activity and force during voluntary contraction of normal and diseased muscle. Acta Neurol Scand [Suppl] 63:83, 1981
94. Basmajian JV, DeLuca CJ: Muscles Alive: Their Functions Revealed by Electromyography, ed 5. Baltimore, MD, Williams & Wilkins, 1985
95. Grieve DW, Davanagh PR: The quantitative analysis of phasic electromyograms. In Desmedt JE (ed): New Developments in Electromyography and Clinical Neurophysiology. Basel, Switzerland, Karger, 1973, vol 2, pp 489-496
96. Johnson JC: Comparison of analysis techniques for electromyographic data. Aviat Space Environ Med 49:14-18, 1978
97. Lindstrom L, Magnusson R, Petersen I: Muscle load influence on myoelectric signal characteristics. Scand J Rehabil Med [Suppl] 3:127-148, 1974
98. Maton B: Motor unit differentiation and integrated surface EMG in voluntary isometric contraction. Eur J Appl Physiol 35:149-157, 1976