ankle was fixed at 0°, 45°, or 90°, thus simulating the muscle length changes during shortening contractions. The compound action potential was detected for each electrode location and for each ankle position. Two sets of measurements were taken. In one set, the electrode was attached to the skin in a normal fashion; in the other set, the electrode was held firmly against the skin with pressure applied against the muscle, thereby reducing the relative movement of the electrode and skin with respect to the muscle fibers.

A typical example of the results is displayed in Figure 2.19. Note that the shape of the compound action potential is susceptible to the location of the electrode along the muscle, and in the case where no pressure was applied to stabilize the electrode, the shape was altered by the position of the ankle. In both cases, the location which provided the most unreliable results was the one corresponding to the innervation zone.

These observations may be conveniently explained by using a schematic approach for the formulation of the MUAP, which in some sense has a comparable genesis to the compound action potential. (The reader is referred to Chapter 3 for a detailed explanation of this concept.) This schematic representation is presented in Figure 2.20. The MUAP is formed by linearly superimposing the action potentials from all the active muscle fibers in the vicinity of the detection electrode. In the *top panel*, the electrode is located near the innervation zone, in the *bottom panel* away from the zone. It is apparent in the top panel that a slight displacement of the electrode results in a drastically different waveform for the MUAP, whereas in the bottom panel an equivalent displacement does not result in such drastic alterations of the waveform.

It is also apparent from Figure 2.20, and may be noted in the empirical results of Figure 2.19, that the relative movement of the detection electrode with respect to the active fibers (as may easily occur when a muscle shortens and the electrode remains stationary on the skin) may, by itself, alter the characteristics of the detected signal. This modification of the EMG signal would not be related to physiological aspects of the contracting muscle. A situation of this nature is found in investigations which acquire EMG signals during anisometric contraction, such as gait. The modulation of the amplitude of the EMG signals obtained from lower limb muscles during gait should be interpreted with great caution.

*We suggest that the preferred location of an electrode is in the region halfway between the center of the innervation zone and the further tendon.*

CHAPTER 3

# Description and Analysis of the EMG Signal

The EMG signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle. It is an exceedingly complicated signal which is affected by the anatomical and physiological properties of muscles, the control scheme of the peripheral nervous system, as well as the characteristics of the instrumentation that is used to detect and observe it. Most of the relationships between the EMG signal and the properties of a contracting muscle which are presently employed have evolved serendipitously. The lack of a proper description of the EMG signal is probably the greatest single factor which has hampered the development of electromyography into a precise discipline.

This chapter will present two main concepts. The first is a discussion of a structured approach for interpreting the information content of the EMG signal. The mathematical model which is developed is based on current knowledge of the properties of contracting human muscles. These properties are discussed in Chapter 5. The extent to which the model contributes to the understanding of the signal is restricted to the limited amount of physiological knowledge currently available. However, even in its present form, the modeling approach supplies an enlightening insight into the composition of the EMG signal.

The second concept in this chapter concerns a discussion of methodologies that are useful for processing and analyzing the signal.

## REVIEW OF NOMENCLATURE

Throughout this chapter, specialized terms will be used to describe distinct aspects of signals. These terms will be defined now so as to eliminate possible confusion.

*Waveform*—The term which describes all aspects of the excursion of the potential, voltage, or current associated with a signal or a function of time. It incorporates all the notions of shape, amplitude, and time duration.
*Amplitude*—That quantity which expresses the level of signal activity.
*Time duration*—The amount of time over which a waveform presents detectable energy.
*Phase*—In electromyography, this term refers to the net excursion of the amplitude of a signal in either the positive or negative direction.
*Shape*—The characteristics of a signal which remains unaltered with linear scaling in either the amplitude or time domains. An example of such characteristics is the phases of an action potential.

The distinction between the concept of shape and waveform is depicted in Figure 3.1. In Figure 3.1*A* the amplitude is scaled linearly by 2 and by 0.5, but the shape remains unaltered. In Figure 3.1*B* the time is scaled linearly by 2 and by 0.5, but the shape remains unaltered. In Figure 3.1*C* the amplitude and time are scaled nonlinearly, and the number of phases is altered. In this latter case the shape changes.

## THE MOTOR UNIT ACTION POTENTIAL

Under normal conditions, an action potential propagating down a motoneuron activates all the branches of the motoneuron; these in turn activate all the muscle fibers of a motor unit (Krnjević and Miledi, 1958a;

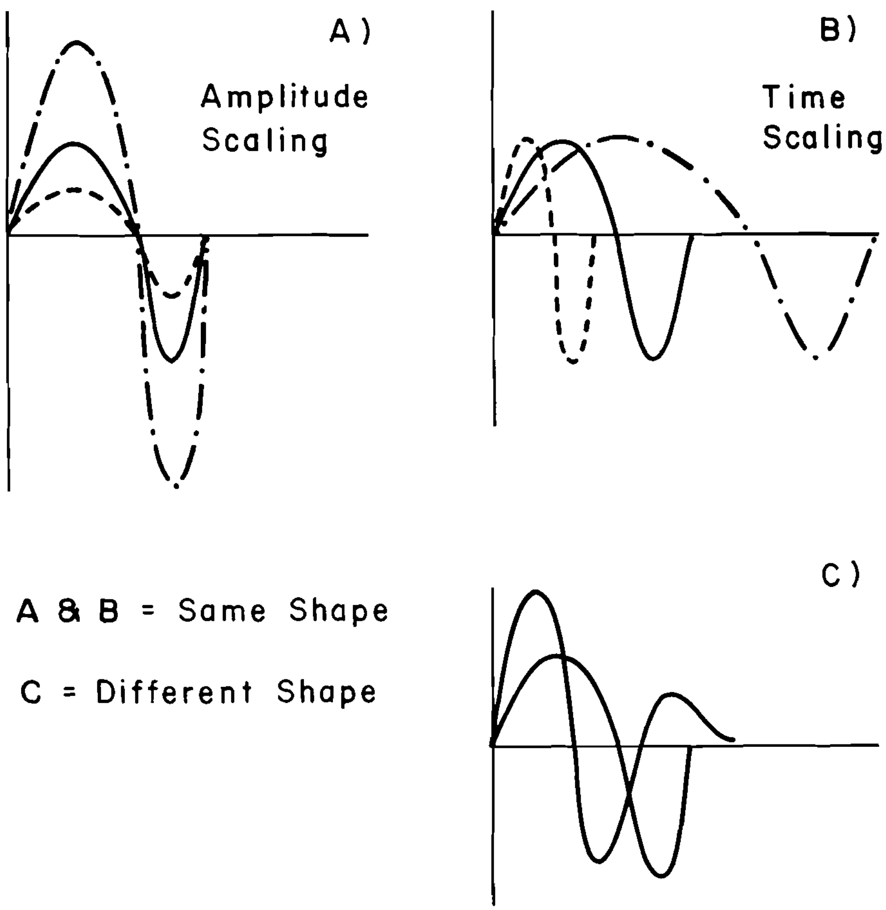



**Figure 3.1.** Distinction between the concept of shape and waveform. In *A* and *B* the shape remains unaltered through linear scaling (0.5 and 2) of the amplitude and time. In *C* the shape is modified through nonlinear scaling and an additional phase. In all three examples, the waveform has been altered.



Paton and Wand, 1967). When the postsynaptic membrane of a muscle fiber is depolarized, the depolarization propagates in both directions along the fiber. The membrane depolarization, accompanied by a movement of ions, generates an electromagnetic field in the vicinity of the muscle fibers. An electrode located in this field will detect the potential or voltage (with respect to ground), whose time excursion is known as an action potential. A schematic representation of this situation is presented in Figure 3.2. In the diagram, the integer *n* represents the total number of muscle fibers of one motor unit that are sufficiently near the recording electrode for their action potentials to be detected by the electrode. For the sake of simplicity, only the muscle fibers from one motor unit are depicted. The action potentials associated with each muscle fiber are presented on the right side of Figure 3.2. The individual muscle fiber action potentials represent the contribution that each active muscle fiber makes to the signal detected at the electrode site.

For technical reasons, the detection electrode is typically bipolar, and the signal is amplified differentially. The waveform of the observed action potential will depend on the orientation of the detection electrode contacts with respect to the active fibers. For simplicity, in Figure 3.2 the detection surfaces of the electrode are aligned parallel to the muscle fibers. With this arrangement, the observed action potentials of the muscle fibers will have a biphasic shape, and the sign of the phases will depend upon the direction from which the muscle membrane depolarization approaches the detection site. To clarify the relative position of the neuromuscular junction of each muscle fiber and the recording site in Figure 3.2, lines have been drawn between the nearest point on each muscle fiber and the detection surfaces. In the diagram, a depolarization approaching from the right side is reflected as a negative phase in the action potential and *vice versa.* Note that when the depolarization of the muscle fiber membranes reaches the point marked by the two lines, the corresponding muscle fiber action potential will have a zero interphasic value.

In human muscle tissue, the amplitude of the action potentials is dependent on the diameter of the muscle fiber, the distance between the active muscle fiber and the detection site, and the filtering properties of the electrode. The amplitude increases as $V = ka^{1.7}$, where $a$ is the radius of the muscle fiber and $k$ is a constant (Rosenfalck, 1969); it decreases approximately inversely proportional to the distance between the active fiber and the detection site. The filtering properties of a bipolar electrode are a function of the size of the detection surfaces, the distance between the contacts, and the chemical properties of the metal-electrolyte interface. For details see discussion in previous chapter.

The duration of the action potentials will be inversely related to the

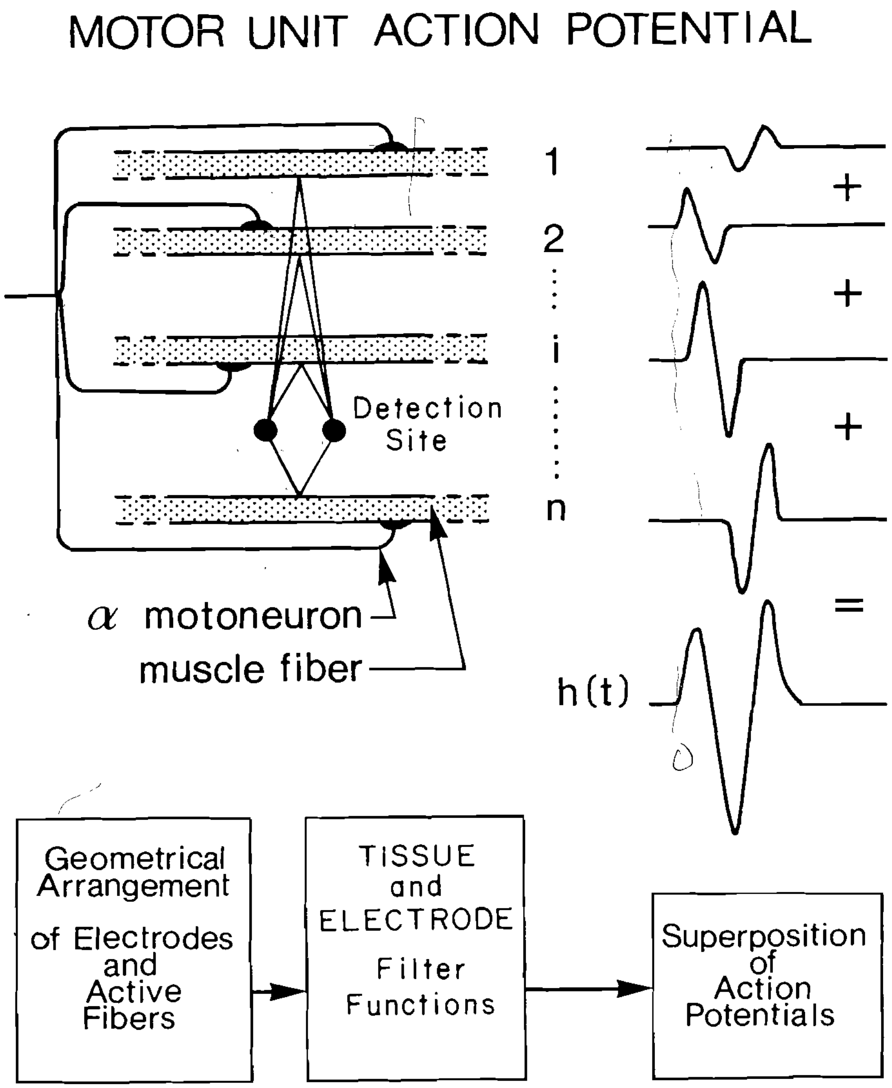



**Figure 3.2.** Schematic representation of the generation of the motor unit action potential.

conduction velocity of the muscle fiber, which ranges from 3 to 6 m/s. The relative time of initiation of each action potential is directly proportional to the difference in the length of the nerve branches, and the distance the depolarizations must propagate along the muscle fibers before they approach the detectable range (pickup area) of the electrode. This relative time of initiation is also inversely proportional to the



conduction velocities of the nerve branch and the muscle fiber. The time delay caused by propagation along the muscle fibers is an order of magnitude greater than that caused by the nerve branch because of the much faster alpha-motoneuron conduction velocity (in the order of 50 to 90 m/s).

The waveform and, therefore, the frequency spectrum of the action potentials will be affected by the tissue beween the muscle fiber and the detection site. As described in the previous chapter, the presence of this tissue creates a low-pass filtering effect whose bandwidth decreases as the distance increases. This filtering effect of the tissue is much more pronounced for surface electrode recordings than for indwelling electrode recordings because indwelling electrodes are located closer to the active muscle fibers.

Thus far, muscle fiber action potentials have been considered as distinguishable individual events. However, since the depolarizations of the muscle fibers of one motor unit overlap in time, the resultant signal present at the detection site will constitute a spatial-temporal superposition of the contributions of the individual action potentials. The resultant signal is called the motor unit action potential (MUAP) and will be designated as $h(t)$. A graphic representation of the superposition is shown on the *right side* of Figure 3.2. This particular example presents a triphasic MUAP. The shape and the amplitude of the MUAP are dependent on the geometric arrangement of the active muscle fibers with respect to the electrode site as well as all the previously mentioned factors which affect the action potentials (Refer to discussion in previous chapter for details.)

If muscle fibers belonging to other motor units in the detectable vicinity of the electrode are excited, their MUAPs will also be detected. However, the shape of each MUAP will generally vary due to the unique geometric arrangement of the fibers of each motor unit with respect to the detection site. MUAPs from different motor units may have similar amplitude and shape when the muscle fibers of each motor unit in the detectable vicinity of the electrode have a similar spatial arrangement. Even slight movements of indwelling electrodes will significantly alter the geometric arrangement and, consequently, the amplitude and shape of the MUAP.

Given the various factors that affect the shape of an observed MUAP, it is not surprising to find variations in the amplitude, number of phases, and duration of MUAPs detected by one electrode, and even larger variations if MUAPs are detected with different electrodes. In normal muscle, the peak-to-peak amplitude of a MUAP detected with indwelling electrodes (needle or wire) may range from a few microvolts to 5 mV, with a typical value of 500 $\mu$V. According to Buchthal et al (1954), the number of phases of MUAPs detected with bipolar needle electrodes may range from one to four with the following distribution: 3% monophasic, 49% biphasic, 37% triphasic, and 11% quadriphasic. MUAPs having more than four phases are rare in normal muscle tissue but do appear in abnormal muscle tissue (Marinacci, 1968). The time duration of MUAPs may also vary greatly, ranging from less than 1 to 13 ms (Stålberg et al, 1975; Basmajian and Cross, 1971).

Petersén and Kugelberg (1949) first reported a slight prolongation of the MUAP with advancing age. Later, Sacco et al (1962) proved, in a systematic study of abductor digiti quinti, biceps brachii, and tibialis anterior of normal infants (3 months of age) and adults, that the duration of the MUAPs was significantly shorter in the muscles of infants (Fig. 3.3). This they explained in terms of the increase in width of the endplate zone with growth. In persons from 20 to 70 years of age, the mean duration of the MUAPs increased 25% further in the biceps brachii, but they remained unaltered in the abductor digiti quinti. Whenever the duration of the MUAPs increased with age, there was an increase in mean amplitude. The Russian physiologists Fudel-Osipura and Grishko (1962) reported that in aged persons the amplitude of the MUAP decreased and the time duration increased. A similar but more detailed

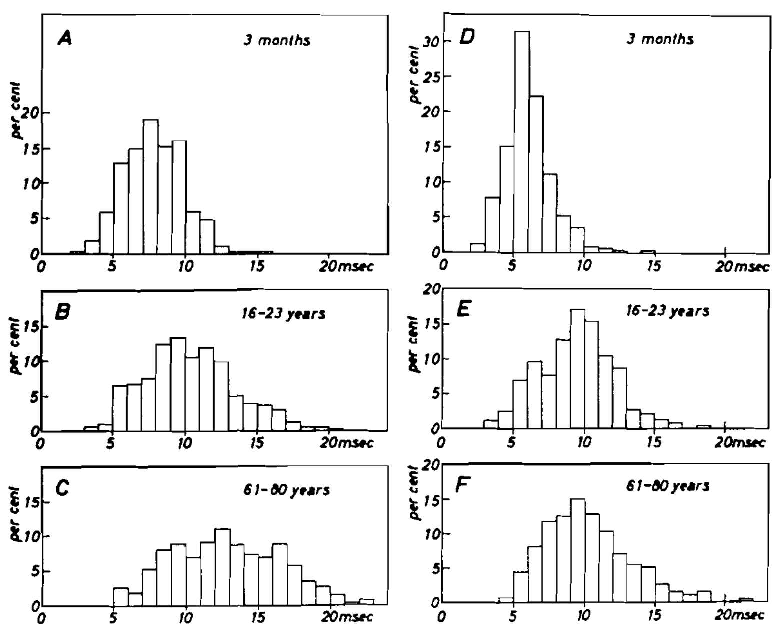



**Figure 3.3.** Histograms of duration of motor unit potentials recorded with concentric electrodes from biceps brachii (*A–C*) and abductor digiti quinti (*D–F*) of subjects of different ages. (From G. Sacco et al, © 1962.)



study was reported by Carlson et al (1964). They found a considerable number of highly complex and long duration MUAPs in more than half of the older age group (Fig. 3.4), suggesting that such a deviation from normal motor unit activity is a characteristic of aged skeletal muscle. (No polyphasic or long-duration potentials were noted in the young normal individuals who made up their control group.) In view of the absence of denervation (fibrillation) potentials in all aged subjects and the finding of normal motoneuron conduction velocities, Carlson et al could not relate the presence of complex potentials to a neurogenic disturbance. They thought this could be explained on the basis of physiological alteration of the muscle fiber with an associated delay in fiber response.

*It should be emphasized that the amplitude and shape of an observed MUAP are a function of the geometrical properties of the motor unit, muscle tissue, and detection electrode properties.* The filtering properties of the electrode (and possibly the cable connecting the electrode to the preamplifiers, as well as the preamplifiers themselves) can cause the observed MUAPs to

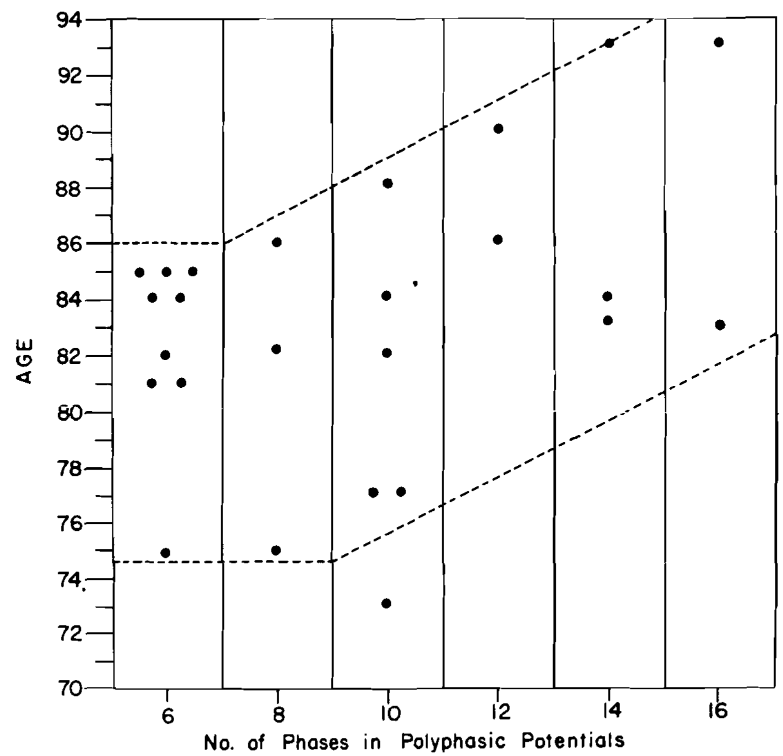



**Figure 3.4.** Increase in polyphasic activity as a function of age. (From K.E. Carlson et al, 1964, *American Journal of Physical Medicine*)

have additional phases and/or longer durations. This is an inevitable behavior of most filter networks. These effects were known empirically when Petersén and Kugelberg (1949) reported that the configuration of the electrode affected the duration and amplitude of the detected action potentials.

## THE MOTOR UNIT ACTION POTENTIAL TRAINS

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

The muscle fibers of a motor unit are randomly distributed throughout a subsection of the normal muscle and are intermingled with fibers belonging to different motor units. Evidence for this anatomical arrangement in the rat and cat has been presented by Edström and Kugelberg (1968), Doyle and Mayer (1969) and Burke and Tsairis (1973). There is also indirect electromyographic evidence suggesting that a similar arrangement occurs in human muscle (Stålberg and Ekstedt, 1973; Stålberg et al, 1976). The cross-sectional area of a motor unit territory ranges from 10 to 30 times the cross-sectional area of the muscle fibers of the motor unit (Buchthal et al, 1959; Brandstater and Lambert, 1973). This admixture implies that any portion of the muscle may contain fibers belonging to 20 to 50 motor units. Therefore, a single MUAPT is observed when the fibers of only one motor unit in the vicinity of the electrode are active. Such a situation occurs only during a very weak muscle contraction. As the force output of a muscle increases, motor units having fibers in the vicinity of the electrode become activated, and several MUAPTs will be detected simultaneously. This is the case even for highly selective electrodes which detect action potentials of single muscle fibers. As the number of simultaneously detected MUAPTs increases, it becomes more difficult to identify all the MUAPs of any particular MUAPT due to the increasing probability of overlap between MUAPs of different MUAPTs.

### A Model for the Motor Unit Action Potential Train

Several investigators have attempted to formulate mathematical expressions for the MUAPT (Bernshtein, 1967; Libkind, 1968; De Luca, 1968; Coggshall and Bekey, 1970; Stern, 1971; Brody et al, 1974; Gath, 1974; De Luca, 1975). Of these investigators, only Libkind (1968), De Luca (1975), and Gath (1974) have employed empirically derived information to construct the model.



The MUAPT may be completely described by its IPIs and the waveform of the MUAP. From a mathematical point of view, it is convenient to describe the MUAPT as a random process in which the waveform of the MUAP is present at random intervals of time. Considerable support for this approach is presented in the remainder of this chapter. This approach requires that we describe mathematically the waveform and the rate of occurrence of the MUAPs in the train, i.e., the firing rate.

For the purpose of our discussion, the firing rate will be considered to be only a function of time ($t$) and force ($F$) and will be denoted as $\lambda(t, F)$. This restriction in the notation is adopted for convenience. However, it should be clearly understood that the derivations which follow apply for any general description of the firing rate. If future investigations reveal definitive relationships between the firing rate and the force rate, velocity, and acceleration of a contraction, they can be readily incorporated into the ensuing model with no loss of generality. A systematic way of obtaining a mathematical expression for $\lambda(t, F)$ is to fit the IPI histogram with a probability distribution function, $p_x(x, t, F)$. The inverse of the mean value of $p_x(x, t, F)$ will be the firing rate, or

$$\lambda(t, F) = \left[\int_{-\infty}^{\infty} x p_x(x, t, F)\, dx\right]^{-1}$$

Alternatively, a mathematical expression for $\lambda(t, F)$ could be obtained by performing a regression analysis of the IPIs as a function of time and force.

On the other hand, it would be extremely difficult to give a unique mathematical description of the MUAP because there are many possible shapes. However, if a MUAPT is isolated and the MUAP can be identified, it would be possible to make a piecewise approximation of the shape. Refer to De Luca (1975) for additional information on the mathematical representations.

It is further convenient to decompose the MUAPT into a sequence of Dirac delta impules, $\delta_i(t)$, which are passed through a filter (black box) whose impulse response is $h_i(t)$. The impulse response of the filter may be constructed to be time variant in order to reflect any change in the waveform of the MUAP during a sustained contraction. This concept is not included in this part of the development of the model so that the equations do not become unnecessarily cumbersome.

If each Dirac delta impulse marks the time occurrence of a MUAP in a MUAPT, the output of the filter will be the MUAPT or $u_i(t)$. The integer $i$ denotes a particular MUAPT. This decomposition, shown in Figure 3.5, allows us to treat the two characteristics of the MUAPT separately.

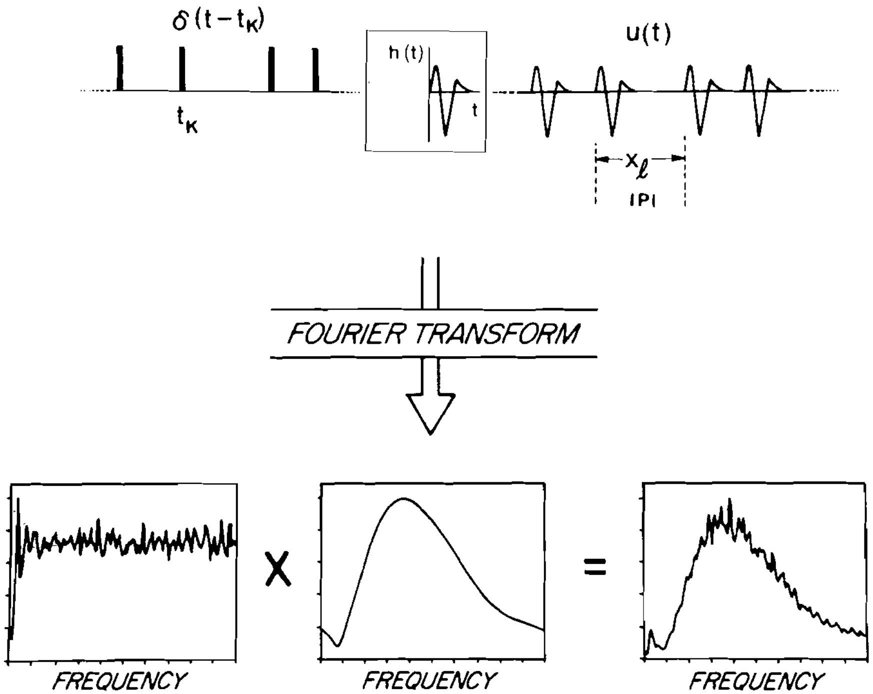



**Figure 3.5.** Model for a motor unit action potential train (MUAPT) and the corresponding Fourier transform of the interpulse intervals (IPIs), the motor unit action potentials (MUAP), and the MUAPT.

The Dirac delta impulse train can be described by

$$\delta_i(t) = \sum_{k=1}^{n} \delta(t - t_k)$$

It follows that the MUAPT, $u_i(t)$, can be expressed as

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

where $t_k = \sum_{l=1}^{k} x_l$ for $k, l = 1, 2, 3, \ldots, n$. In the above expressions, $t$ is a real continuous random variable, $t_k$ represents the time locations of the MUAPs. $x$ represents the IPIs, $n$ is the total number of IPIs in a MUAPT, and $i$, $k$, and $l$ are integers which denote specific events.

Reports have appeared in the literature which argue in favor of the existence of minimal (if any) dependence among the IPIs of a particular MUAPT (Masland et al, 1969; De Luca and Forrest, 1973a; Kranz and Baumgartner, 1974; and others). It could be argued that these reports do not firmly conclude that the IPIs of a MUAPT are statistically independent. In fact, it is easy to envisage specific circumstances and situations where the Ia-alpha motoneuron reflex loop may inject some



dependence. However, the overwhelming majority of the available data supports an approach for modeling the IPI train as a renewal process which provides considerable mathematical and practical convenience.

It is now possible to write expressions for two time-domain parameters of signals, i.e., the mean rectified value and the mean-squared value (whose square root is the root-mean-squared value, or rms) by invoking one restriction, that is, the waveform of the MUAP remains invariant throughout the train. Then, and only then, it follows that:

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

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

where $\hat{t}$ is a dummy variable and $E$ is the mathematical symbol for the expectation or the mean. Although the above equations can be solved, the computation requires the execution of a convolution. De Luca (1975) has shown that since $\lambda(t, F)$ is slowly time varying, the above expressions can be greatly simplified to:

$$E\{|u_i(t, F)|\} = |h_i(t)|\,\lambda_i(t, F)$$

$$MS\{u_i(t, F)\} = h_i^2(t)\lambda_i(t, F)$$

In most cases, this approximation introduces an error of less than 0.001%. The bar denotes an integration from zero to infinity as a function of time. These mathematical computations are displayed in Figure 3.6. It should be noted that the first term on the right side of the equations is now a scaling value and is independent of time. Hence, these MUAPT parameters are reduced to the expression of the firing rate multiplied by a scaling factor.

To compute the expression for the power density spectrum (frequency content) of a MUAPT it is necessary to consider additional statistics of the IPIs and the actual MUAP waveform. The IPIs can be described as a real, continuous, random variable. Only minimal (if any) dependence exists among the IPIs of a particular MUAPT. Therefore, the MUAPT may be represented as a renewal pulse process. A renewal pulse process is one in which each IPI is independent of all the other IPIs.

The power density spectrum of a MUAPT was derived from the above formulation by Le Fever and De Luca (1976) and independently by Lago and Jones (1977). It can be expressed as:

$$S_{u_i}(\omega, t, F) = S_{\delta_i}(\omega, t, F)\,|H_i(j\omega)|^2$$

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

for $\omega \neq 0$

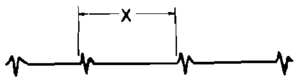
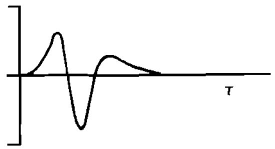
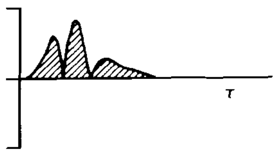
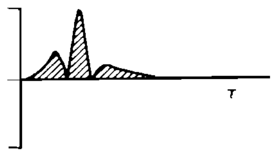
| TERM | DIAGRAM | EXPRESSION |
|---|---|---|
| a) GENERALIZED FIRING RATE OF A TYPICAL MOTOR UNIT |  MOTOR UNIT ACTION POTENTIAL TRAIN | $\lambda(\tau,\varphi) = \frac{1}{E(X)}$ |
| b) MOTOR UNIT ACTION POTENTIAL |  | $h_i(\tau)$ |
| c) AREA UNDER THE RECTIFIED MOTOR UNIT ACTION POTENTIAL |  | $\underline{\lvert h_i(\tau)\rvert} = \int_0^\infty \lvert h_i(\tau)\rvert d\tau$ |
| d) AREA UNDER THE SQUARE OF A MOTOR UNIT ACTION POTENTIAL |  | $\underline{h_i^2(\tau)} = \int_0^\infty h_i^2(\tau) d\tau$ |

**Figure 3.6.** Explanation of some of the terms in the expressions in the text.

where

$\omega$ = the frequency in radians
$H_i(j)$ = the Fourier transform of $h_i(t)$
$M(j\omega, t, F)$ = the Fourier transform of the probability distribution function, $p_x(x, t, F)$ of the IPIs.

More recently, Blinowska et al (1979) have derived similar expressions. The above expressions clearly indicate that the power density spectrum may be constructed from the energy spectra of the MUAPs of the motor units and the firing statistics of the motor unit. In addition, the contribution of the firing statistics has a multiplicative effect on the energy spectrum of the MUAP.

By representing $h_i(t)$ by a Fourier series, LeFever and De Luca (1976) were able to show that in the frequency range of 0 to 40 Hz the power density spectrum is affected primarily by the IPI statistics. A noticeable peak appears in the power density spectrum at the frequency corresponding to the firing rate and progressively lower peaks at harmonics of the firing rate. The amplitude of the peaks increases as the IPIs become



more regular. Beyond 40 Hz, the power density spectrum is essentially determined by the shape of $h_i(t)$.

This point has been verified empirically. By using the computer-assisted decomposition technique developed by LeFever and De Luca (1982) and Mambrito and De Luca (1983), it is possible to obtain highly accurate IPI measurements of MUAPT records many seconds long. The details of this technique are described in Chapter 4. The Fourier transform of the IPIs may then be computed directly. Figure 3.7 presents the magnitude of such Fourier transforms for MUAPTs that were all detected during two separate isometric constant-force contractions maintained at 50% of maximal voluntary force in the first dorsal interosseous muscle. The time duration of the MUAPT segment that was analyzed was 5 s. The function with the solid line represents the average. The histograms present the IPI distribution of each motor unit, the one on the left corresponding to the function with the broken line and the middle histogram to the function with the dash-dot line. Some statistics of the IPIs are presented in Table 3.1. The coefficient of variation, which is the ratio of the standard deviation to the mean value, is a measure of the regularity with which the motor unit is discharging. The smaller the coefficient of variation, the sharper and higher will be the peak corresponding to the firing rate in the magnitude of the Fourier transform. When the coefficient of variation is higher (0.26, 0.28), then the peak is less sharp and has lower amplitude (see Figure 3.7*b*).

### Concepts of Normalization and Generalized Firing Rate

A generalized representation of the EMG signal must contain a formulation which allows a comparison of the signal between different muscles and individuals. This is not a problem in some contractions, such as those involving ballistic movements. However, it is a requirement in isometric and anisometric contractions. The formulation for comparison may be obtained by normalizing the variables of the EMG signal with respect to their maximal measurable value in the particular experimental procedure. For example, in a constant-force isometric contraction, the time is normalized with respect to the duration that the individual can maintain the designated force level. The contraction force is normalized with respect to the force value of a MVC. The *normalized contraction-time* will be denoted by $\tau$, the *normalized force* by $\phi$, and their maximal value is 1.

In the model, expressions for the parameters of the EMG signal are formed by a superposition of the equations of the MUAPT derived in the previous section. Such an approach requires that the mathematical relationship of the firing rates of all the individual MUAPTs be first definable and then known. Such information is difficult to obtain. Even if it could be obtained it may prove to be useless due to the difficulty in

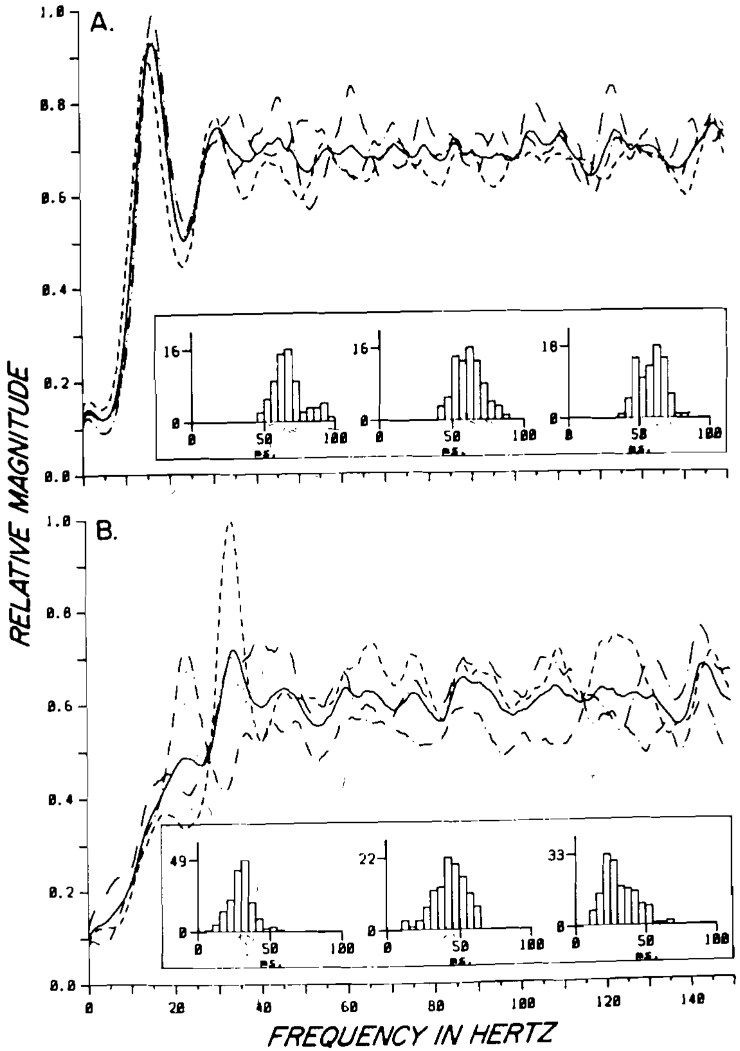



**Figure 3.7.** Magnitude of the Fourier transforms and histograms of the interpulse intervals (IPI) of the motor unit action potential trains obtained from the first dorsal interosseous muscle during two separate constant-force isometric contractions performed at 50% maximal force. The function plotted with the *solid line* represents the average. The histograms present the IPI distribution, with the one on the *left* corresponding to the function with the *broken line* and with the middle histogram corresponding to the function with the *dash-dot* line. Note that the peak corresponding to the average firing rate value is present in some functions.



**Table 3.1**
**Interpulse Statistics of MUAPTs Obtained During Isometric Constant-Force Contractions Sustained at 50% of Maximal Force in the First-Dorsal Interosseous Muscle[a]**

| | $\mu$[b] (ms) | SD (ms) | CV |
|---|---|---|---|
| | 58.5 | 9.3 | 0.20 |
| Fig. 3.7*A* | 61.8 | 10.0 | 0.16 |
| | 69.3 | 13.7 | 0.16 |
| | 29.7 | 8.3 | 0.28 |
| Fig. 3.7*B* | 31.3 | 11.8 | 0.38 |
| | 43.4 | 11.2 | 0.26 |

[a] This is the detailed data of the functions and histograms plotted in Figure 3.5.
[b] $\mu$, mean; SD, standard deviation; CV, coefficient of variation.

defining this as a function of time and force. There is considerable evidence (which will be described in Chapter 5) that the firing rate is not necessarily a monotonic function of either time or force, and the relationship between firing rate and force is also dependent on the particular muscle. To overcome this barrier, De Luca (1968) introduced the concept of the *generalized firing rate* and defined it as the mean value of the firing rates of the MUAPTs detected during a contraction. For a detailed description of the calculation of the generalized firing rate, refer to De Luca and Forrest (1973a).

It must be emphasized that the generalized firing rate is a mathematical concept which may only be properly mathematically described with extreme difficulty. It will be dependent on many factors which may effect the occurrence of the IPIs, such as recruitment of a new motor unit, minute force perturbations in the intended force output of a muscle, reflex activity, etc. One example of the mathematical formulation has been described by De Luca and Forrest (1973a) and is presented here:

$$\lambda(\tau, \phi) = \frac{1000}{\beta(\tau, \phi)\Gamma[1 + 1/\kappa(\tau, \phi)] + \alpha} \text{ pulses per second}$$

$$\kappa(\tau, \phi) = 1.16 - 0.19\tau + 0.18\phi$$
$$\beta(\tau, \phi) = \exp\,(4.60 + 0.67\tau - 1.16\phi) \text{ ms}$$
$$\alpha = 3.9 \text{ ms}$$
$$\text{for } 0 < \tau < 1,\ 0 < \phi < 1$$

The above values are valid for the middle fibers of the deltoid muscle during a constant force isometric contraction. Other relationships will exist for other muscles. The above equation is plotted in Figure 3.8.

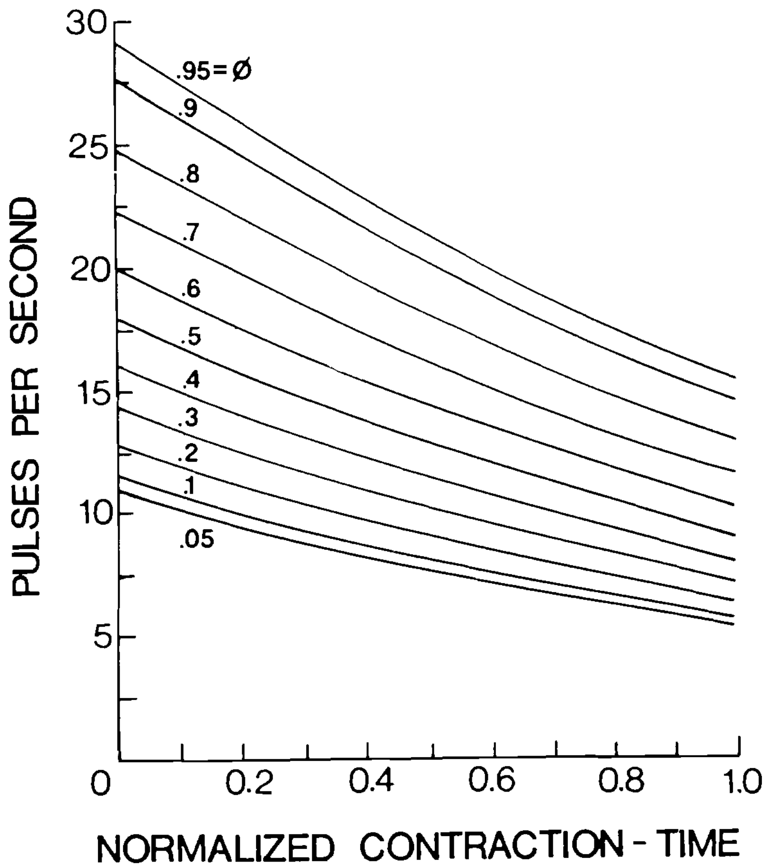



**Figure 3.8.** Generalized firing rate of motor unit action potential trains as a function of normalized contraction-time at various normalized constant-force levels. The force was normalized with respect to the maximal isometric contraction.

### A Model for the EMG Signal

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

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

and is displayed in Figure 3.9, where 25 mathematically generated MUAPTs are added to yield the signal at the bottom. This composite signal bears striking similarity to real EMG signals.

Biro and Partridge (1971) obtained empirical evidence which justified this approach. The modeling approach was later expanded by De Luca



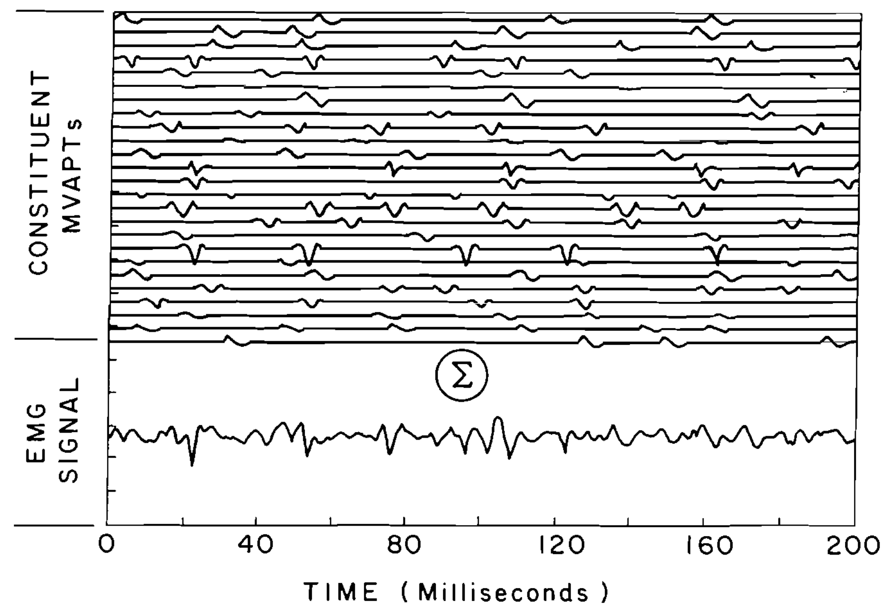



**Figure 3.9.** An EMG signal formed by adding (superimposing) 25 mathematically generated MUAPTs.

and van Dyk (1975) and Meijers et al (1976). A schematic representation of the model is shown in Figure 3.10. The interger $p$ represents the total number of MUAPTs which contribute to the potential field at the recording site. Each of the MUAPTs can be modeled according to the approaches presented in Figure 3.2 and 3.5. The superposition at the recording site forms the physiological EMG signal, $m_p(t, F)$. This signal is not observable. When the signal is detected, an electrical noise, $n(t)$, is introduced. The detected signal will also be affected by the filtering properties of the recording electrode, $r(t)$, and possibly other instrumentation. The resulting signal, $m(t, F)$, is the observable EMG signal. The location of the recording site with respect to the active motor units determines the waveform of $h(t)$, as described at the beginning of this chapter.

### A Comparison of the Time-Dependent Parameters of the EMG Signal

From this concept it is possible to derive expressions for the mean rectified value, the root-mean-squared value, and the variance of the rectified EMG signal. The expressions are presented in Figure 3.11. Their derivation can be found in the article by De Luca and van Dyk (1975). In Figure 3.11, each of the terms of the expressions are associated with five physiological correlates which affect the properties of the EMG signal.

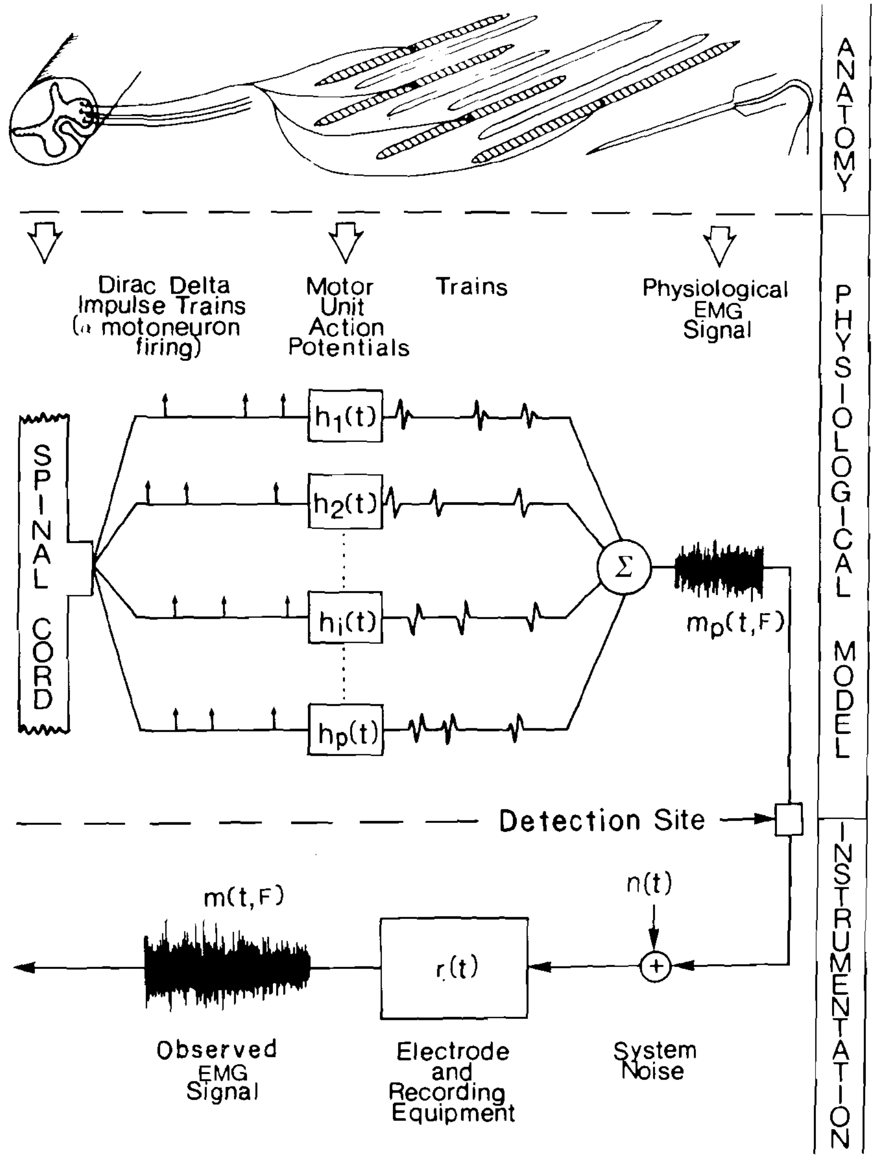



**Figure 3.10.** Schematic representation of the model for the generation of the EMG signal.



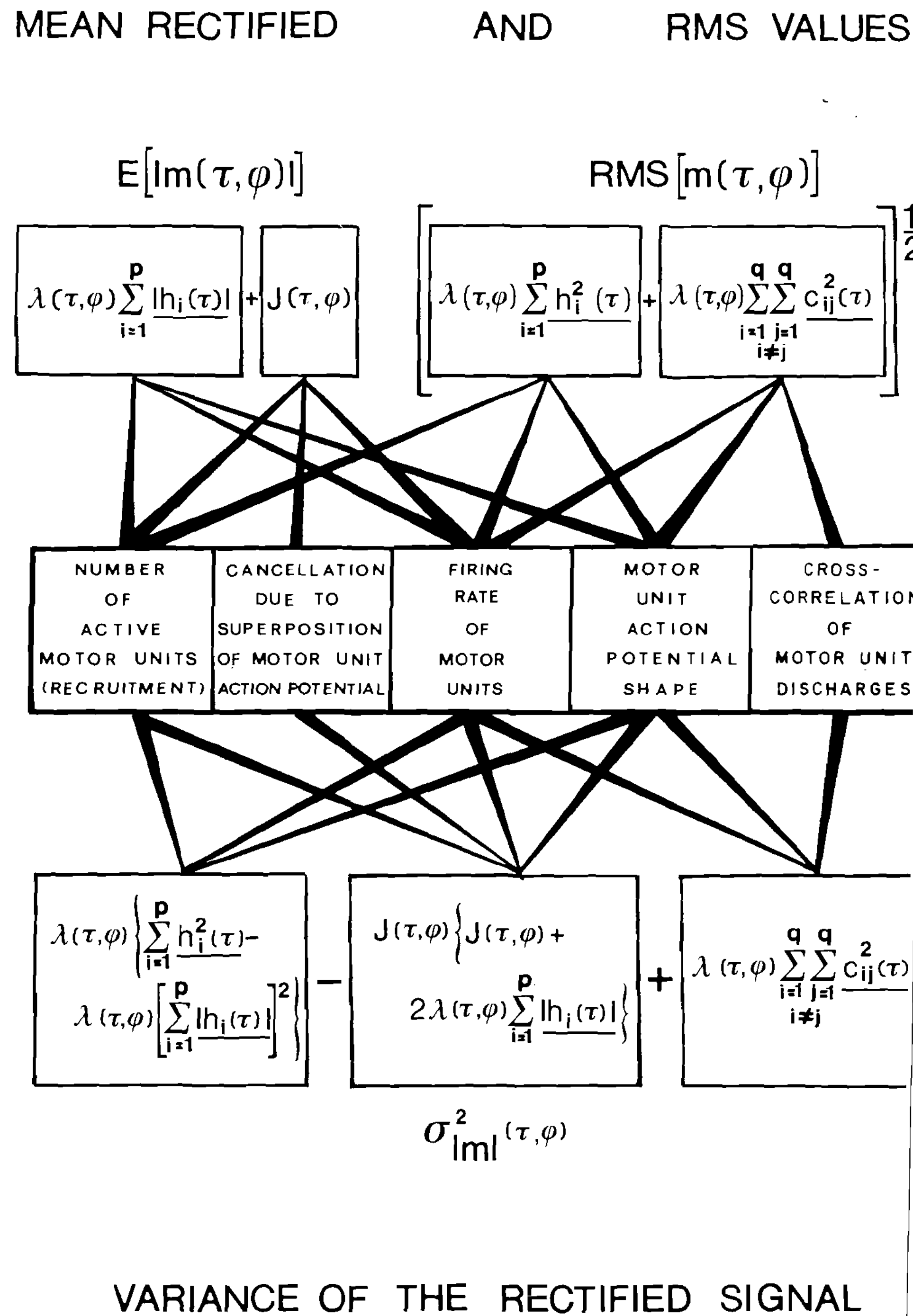



**Figure 3.11.** Theoretical expressions for parameters of the EMG signal and their relation to physiological correlates of a contracting muscle.

In the equation of the mean rectified value, the term $J(t, F)$ is a nonpositive term which accounts for the cancellation in the signal due to the superposition of opposite phases of the MUAPTs. In a sense, the *superposition term* represents the EMG activity which is generated by the muscle but is not available in the observed EMG signal. The expression for the mean rectified value confirms that this parameter of the EMG signal is dependent on the number and firing rates of the MUAPTs detected by the electrode, the area of the MUAPs, and the amount of cancellation occurring from the superposition of the MUAPTs.

The integral of the mean rectified value is a commonly used parameter in electromyography. By definition, it will be dependent on the same physiological correlates as the mean rectified value. It is often used to obtain a relationship between the EMG signal and the force output of the muscle. A linear relationship has been reported often. Considering all the physiological correlates involved, the nonlinearity of the motor unit behavior, and the viscoelastic properties of muscle tissue, a linear relationhip should be considered as accidental.

The root-mean-squared value is also dependent on the number and firing rates of the MUAPTs and the area of the MUAPTs but is not affected by the cancellation due to the MUAPT superposition. However, it is affected by the cross-correlation between the MUAPTs, represented by the $c_{ij}^2(\tau)$ terms. In the corresponding equation in Figure 3.11, the integer $v$ denotes the number of MUAPTs that are cross-correlated. Note that any two MUAPTs can still be synchronized even if their cross-correlation term is zero, since the lack of cross-correlation is not sufficient to prove independence. Under this condition, their synchronization has no effect on the root-mean-squared value.

The expression for the variance of the rectified signal is more complicated, containing all the terms which are present in the previous two parameters. Therefore, it reflects the combined effect of all the physiological correlates. This parameter represents the AC power of the rectified EMG signal and should prove to be useful in analyzing the EMG signal. However, it has been used sparingly. Most of the past investigations have dealt with the DC level of the EMG signal.

The approach used thus far has been directed at relating the measurable parameters of the EMG signal to the behavior of the individual MUAPTs. However, when the electrode detects a large number of MUAPTs (greater than 15), such as would typically be the case for a surface electrode, the law of large numbers can be invoked to consider a simpler, more limited approach. In such cases, the EMG signal can be effectively represented as a band-limited signal with a Gaussian distributed amplitude. It is also possible to represent the amplitude by a carrier signal consisting of a random signal whose statistical properties are those of the signal during an isometric contraction, and a modulation signal reflecting the force and time-dependent properties of the signal. Such a representation was first publicized by Kreifeldt and Yao (1974) and was later expanded by Shwedyk et al (1977) and Hogan and Mann (1980).



## PROPERTIES OF THE POWER DENSITY SPECTRUM OF THE EMG SIGNAL

According to the model in Figures 3.10 and 3.11, the power density spectrum of the EMG signal may be formed by summing all the auto- and cross-spectra of the individual MUAPTs, as indicated in this expression:

$$S_m(\omega) = \sum_{i=1}^{p} S_{u_i}(\omega) + \sum_{\substack{i,j=1 \\ i \neq j}}^{q} S_{u_i u_j}(\omega)$$

where $S_{u_i}(\omega)$ = the power density of the MUAPT, $u_i(t)$; and $S_{u_i u_j}(\omega)$ = the cross-power density spectrum of MUAPTs $u_i(t)$ and $u_j(t)$. This spectrum will be nonzero if the firing rates of any two active motor units are correlated. Finally, $p$ = the total number of MUAPTs that comprise the signal; $q$ = the number of MUAPTs with correlated discharges. For details of this mathematical approach, refer to De Luca and van Dyk (1975).

De Luca et al (1982b) have shown that many of the concurrently active motor units have, during an isometric muscle contraction, firing rates which are greatly correlated. It is not yet possible to state that all concurrently active motor units are correlated (although all our observations to date support this point). Therefore, $q$ is not necessarily equal to $p$, which represents the total number of MUAPTs in the EMG signal. The above equation may be expanded to consider the following facts:

1. During a sustained contraction, the characteristics of the MUAP shape may change as a function of time ($t$). For example, De Luca and Forrest (1973a), Broman (1973, 1977), Kranz et al (1981), and Mills (1982) have all reported an increase in the time duration of the MUAP.
2. The number of MUAPTs present in the EMG signal will be dependent on the force of the contraction ($F$).
3. The detected EMG signal will be filtered by the electrode before it can be observed. This electrode filtering function will be represented by $R(\omega, d)$, where $d$ is the distance between the detection surfaces of a bipolar electrode.

Note that the recruitment of motor units as a function of time during a constant force has not been considered; however, the required modification to the equation is trivial, and the concept may easily be accommodated. The concept of "motor unit rotation" during a constant force contraction (i.e., newly recruited motor units replacing previously active motor units) which has, at times, been speculated to exist, has also not been included. No account may be found in the literature which has provided evidence of this phenomenon by definitively excluding the likelihood that the indwelling electrode has moved relative to the active muscle fibers and, in fact, records from a new motor unit territory in the muscle. For a review of these details consult the material in Chapter 5. Hence:

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

The notation used in this equation may appear to be awkward in that it contains functions of time and frequency, when the frequency variable in turn represents a transformation from the time domain. This notation is used to describe functions that have time-dependent frequency properties. From a practical point of view, such equations are useful to describe the properties of "slow" nonstationary stochastic processes.

It is apparent from the examples presented in Figure 3.7 that the individual MUAPT power density function $S_{u_i}(\omega, t, F)$ may or may not have a peak at the frequency value corresponding to the firing rate, depending on the regularity with which the motor unit discharges, i.e., the coefficient of variation of the IPIs. When more than one MUAPT is present, the presence of a peak will also depend on the amount of separation between the mean of the individual IPI histograms. (This latter parameter determines how the peaks and valleys in the initial part of the Fourier transform superimpose and cancel out; note the different results of Figures 3.7*A* and *B*.) In general, when many MUAPTs having a wide range of individual coefficients of variation are present, the peaks will be less pronounced, and the effect of the IPI statistics on the magnitude of the Fourier transform of the EMG signal will be negligible above 30 Hz. No peaks will be distinct in the low frequency region. Empirical evidence of this behavior has been reported by De Luca (1968) and Hogan (1976), who noted large peaks occurring between 8 and 30 Hz. Hogan (1976) further showed that as successively more motor units are detected during increasing force level contractions, the amplitude of the peak diminishes with respect to the remainder of the spectrum. Hogan's data are presented in Figure 3.12.

It should be noted that the two parameters which have been identified as affecting the presence of the firing rate peak are both related to synchronization. That is, the smaller the coefficient of variation and the closer the average firing rate values, the greater the probability of two or more motor units discharging during a specific time interval. It should also be immediately added that physiological events may also occur that may render the MUAPTs dependent and thereby introduce a third parameter to the concept of synchronization.

Now, a major question arises concerning the cross-power density term. Does it really vary as a function of time during a sustained contraction



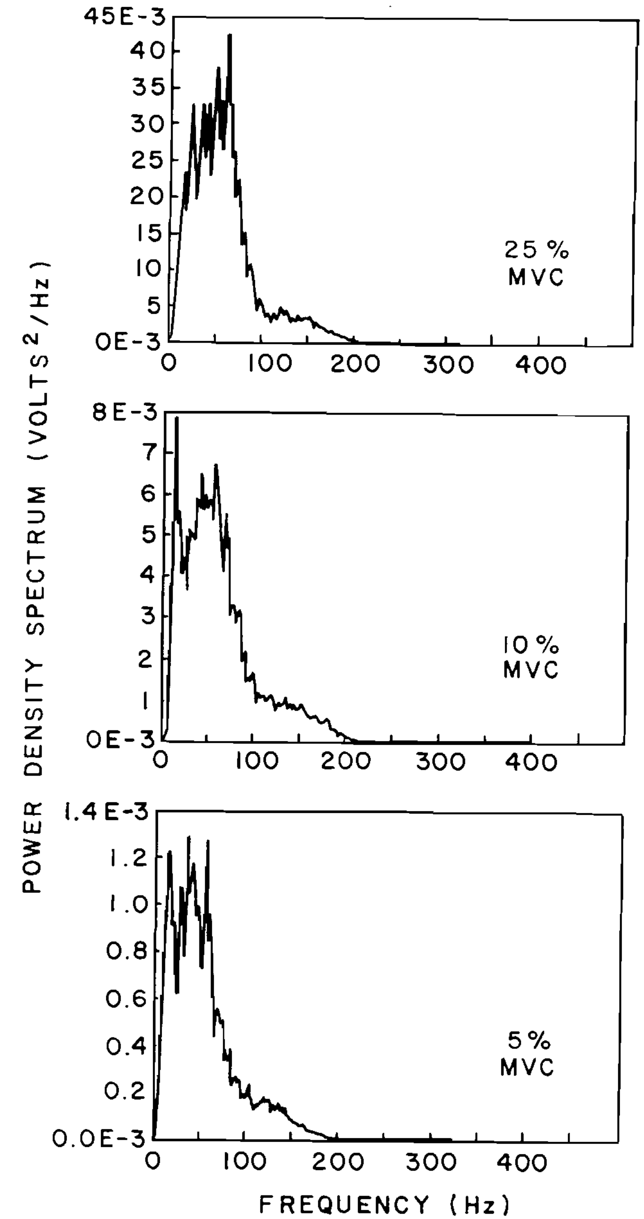



**Figure 3.12.** The power density spectrum when the EMG signal obtained from the biceps brachii during isometric constant-force contractions which are performed at 5%, 10%, and 25% of maximal level. Note that the peaks below 30 Hz are more pronounced during the 5% and 10% maximal level contractions. (From N.J. Hogan, © 1976, M.I.T., Cambridge, MA.)

so as to singly influence the power density spectrum of the EMG signal? There are three eventualities that may influence its time dependency: (1) the characteristics of the shape of the MUAP $u_i(t)$ and $u_j(t)$ change as a function of time; (2) the number of MUAPTs which are correlated varies as a function of time; (3) the degree of cross-correlation among the correlated MUAPTs varies. A change in the shape of the MUAP of $u_i(t)$ and $u_j(t)$ would not only cause an alteration in the cross-power density term but also would cause a more pronounced modification in the respective autopower density spectra. Hence, the power density spectrum of the EMG signal would be altered regardless of the modifications of the individual cross-power density spectra of the MUAPTs. There is to date no direct evidence to support the other two points. In fact, De Luca et al (1982*a* and *b*) have presented data which indicate that the cross-correlation of the firing rates of the concurrently active motor units does not appear to depend on either time during, or force of a contraction. This apparent lack of time-dependent cross-correlation of the firing rates is not inconsistent with previously mentioned observations, indicating that the synchronization of the motor unit discharges tends to increase with contraction time. These two properties can be unrelated.

Up to this point, the modeling approach has provided an explanation of the following aspects and behavior of the power density spectrum:

1. The amplitude increases with additionally recruited MUAPTs.
2. The IPI firing statistics influence the shape of the spectrum below 40 Hz, although this effect is not necessarily consistent, and is less evident at higher force when an increasing number of motor units are active.
3. The tendency for motor units to "synchronize" will affect the spectral characteristics but will be limited to the low frequency components.
4. Modification in the waveform of MUAPs within the duration of a train will effect most of the spectrum of the EMG signal. This is particularly worrisome in signals that are obtained during contractions that are anisometric, because in such cases the waveform of the MUAP may change in response to the modification of the relative distance between the active muscle fibers and the detection electrode.

The above associations do not fully explain the now well-documented property of the EMG signal, which manifests itself as a shift towards the low frequency end of the frequency spectrum during sustained contractions. It is apparent that modifications in the total spectral representation of the MUAPs can only result from a modification in the characteristics of the shape of the MUAP *per se*. During attempted isometric contraction, such modifications have their root cause in events that occur locally within the muscle. Broman (1973) and De Luca and Forrest (1973a) were the first to present evidence that the MUAP increases in time duration during a sustained contraction. More recently, Kranz et al (1981) and Mills (1982) have provided further support.



This approach was pursued by Lindström (1970), who derived a different expression for the power density spectrum of the EMG signal detected with bipolar electrodes. He approached the problem by representing the muscle fibers as cylinders in which a charge is propagated along their length. The simplified version of his results may be expressed as:

$$S_m(\omega) = R(\omega, d)\left[\frac{1}{v^2}\, G\left(\frac{\omega d}{2v}\right)\right]$$

where $v$ = the average conduction velocity of active muscle fibers contributing to the EMG signal; $G$ = the shape function which is implicitly dependent on many anatomical, physiological, and experimental factors; and $d$ = the distance between the detection surfaces of the bipolar electrode.

One of the factors which is incorporated in the $G$ function is the filtering effect of the tissue between the source (active muscle fibers) and the detection electrode. Note that this distance depends on the location (depth) of the muscle fibers within the muscle, plus the thickness of the fatty and skin tissues beneath the electrode. The form of this expression consists of modified Bessel functions of the second kind (Lindström, 1970). A plot of the expression for the "distance" or "tissue" filtering function has been presented in Figure 2.10. It has the characteristics of a low-pass filter with a cutoff frequency that is inversely related to the distance between the recording electrode and active muscle fibers.

The above expression may be generalized by introducing the time- and force-dependent effects on the EMG signal; then

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

This representation of the power density spectrum explicitly denotes the interconnection between the spectrum of the EMG signal and the conduction velocity of the muscle fibers. Such a relationship is implicit in the previously presented modeling approach because any change in the conduction velocity would directly manifest itself in a change in the time duration of $h(t)$ as seen by the two detection surfaces of a stationary bipolar electrode. If the conduction velocity were to decrease, the depolarization current would require more time to traverse the fixed distance along the fibers in the vicinity of the detection surfaces and the detected MUAPs would have longer time durations. Hence, the frequency spectrum of the MUAPs and the EMG signal, which they comprise, will have a relative increase in the lower-frequency components and a decrease in the higher-frequency components. In other words, a shift toward the low-frequency end would occur. The above equation also explains the amplitude increase of the EMG signal as the conduction velocity decreases.

As discussed in the previous chapter, Lindström (1970) was able to show that for a bipolar electrode located between the innervation zone and the tendon of a muscle, the magnitude of the electrode filter function,

$$R(\omega, d) = K sin^2(\omega d/2v)$$

where $d$ is the distance between the detection surfaces and $v$ is the conduction velocity along the muscle fibers. The $sin^2$ term has a zero value at regular frequency intervals. These frequency values are related to the interdetection surface spacing and the conduction velocity of the muscle fibers. (Refer to Chapter 2 for details.) This function has a multiplicative effect on the power density spectrum, therefore, it too will be zero at the particular frequency values. Thus, if the interdetection surface spacing is known, it may be possible to calculate the average conduction velocity of the muscle fibers whose action potentials contribute to the detected signal.

These "dips" in the power density spectrum have been reported by several investigators (Lindström et al, 1970; Broman, 1973; Agarwal and Gottlieb, 1975; Lindström and Magnusson, 1977; Trusgnich et al, 1979; Hogan and Mann, 1980; and others), but their presence is not always prominent. Consider the case in which the detected signal consists of several MUAPTs whose respective MUAPs all contribute substantial energy to the power density spectrum but whose muscle fibers have different conduction velocities. Then a filter function somewhat similar to that shown in Figure 2.16 would apply for every MUAP, but the location of the zero points of the function would be different in accordance with the filter equation. Hence, the sum of the electrode filter functions would constitute a function with no clearly demonstrable dips, and the power density spectrum would not contain a discernible dip.

The opposite is true when there is one MUAPT in the EMG signal which contributes considerably more energy to the EMG signal or when most of the motor units which contribute to the EMG signal have nearly similar conduction velocities.

In a sense, the presence of the dips in analogous to the presence of the "peaks" at the low frequency end of the spectrum. Both will be present when individual MUAPTs in the EMG signal contribute considerable energy to the signal. But, it is possible for only one of these indicators to be present, depending on the firing statistics of the concurrently active motor units that contribute to the detected EMG signal.

At this point, it is necessary to recollect some consistently observed properties of the motor units and the EMG signal as a function of time and force during a contraction. As the force output of the muscle increases, the number of active motor units increases, and the firing rates of all the active motor units generally increase or remain nearly constant. As the time of a sustained contraction increases, the time duration of MUAPs increases, and the spectrum of the ME signal shifts towards the low frequency end. The ramifications of these factors interpreted through the explanations provided by the mathematical models are all graphically illustrated in Figure 3.13.



### A Test for the Time Domain Parameters of the Model

Before testing the parameters derived by the model, it is necessary to comment that the neuromuscular system is an extraordinary actuator. It is capable of generating and modulating force under a wide variety of

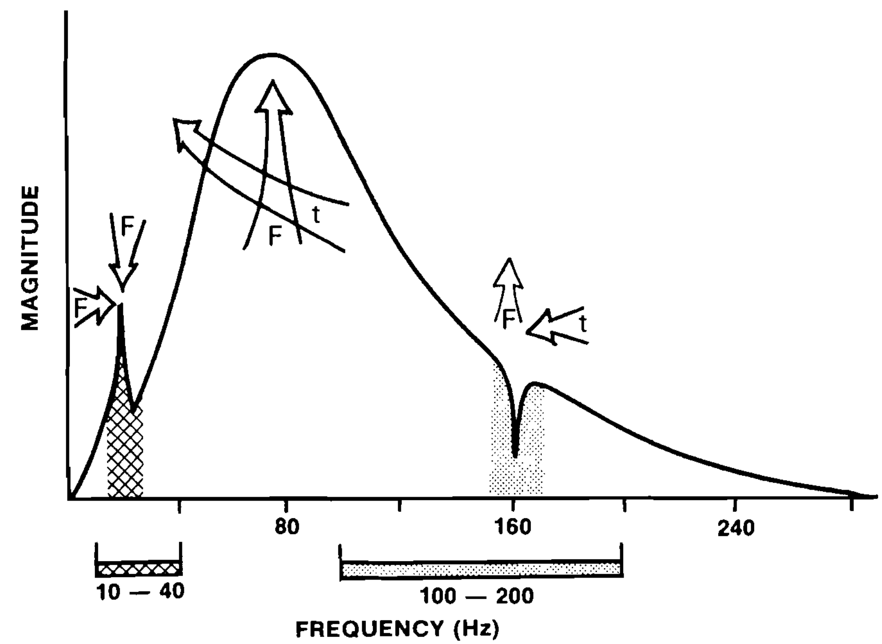



**Figure 3.13.** Diagramatic representation of the frequency spectrum of the differentially detected EMG signal with a graphic representation of the modifications which occur as a function of time and force of contraction. The shape has been purposefully exaggerated so as to accentuate interesting segments. The direction of the *arrows* indicates the direction of the modification on a particular segment of the spectrum caused by increasing either force or time. (For example, an arrow with the letter *F* indicates that as the force increases, the segment to which it is pointing will be modified in the direction of the *arrow*). The peak in the low frequency components is associated with the firing rates of motor units; the dip in the high frequency components is associated with the conduction velocity along the muscle fibers. The bars on the frequency axis indicate the range over which the peaks and dips may occur.

static (isometric) and dynamic (velocity, acceleration) conditions. It is known that the behavior of the MUAPTs varies for different types of contractions. However, it is possible to test the model for the EMG signal recorded during constant-force isometric contractions.

In a study performed by Stulen and De Luca (1978), EMG signals were simultaneously recorded differentially with bipolar surface and needle electrodes from the deltoid muscle while 11 subjects performed sustained constant-force isometric contractions at 25, 50, and 75% of maximal voluntary contraction (MVC). The empirical values of the parameters corresponding to those derived previously were calculated and compared. Let us consider the empirical root-mean-squared parameter which is plotted in Figure 3.14. The *solid lines* represent the average value for the 11 subjects. The *vertical lines* indicate 1 SD about the average. For convenience, the magnitude of the values has been normalized with respect to the largest value of the average.

Note that the amplitude of the root-mean-squared parameter increases as a function of time when the EMG signal is detected with surface electrodes, and decreases when detected with indwelling electrodes. Why? These signals were recorded simultaneously from the same area of the same muscles. To explain this apparent paradox we must turn our attention to the model.

It is possible to solve the equation for the root-mean-squared parameter in Figure 3.14 with the following restrictions: (1) no recruitment occurs during a constant-force contraction, (2) the areas of the MUAPs do not change; and (3) the MUAPTs are not cross-correlated. With these assumptions, the root-mean-squared parameter is directly proportional to the square root of the generalized firing rate. In fact, if the generalized firing rate of Figure 3.8 is normalized, it provides an exceptionally good fit to the mean value of the root-mean-squared parameter of the signal recorded with indwelling electrodes at 25 and 59% MVC but not at 75% MVC. It appears that the decrease in the signal from contractions executed at less than 50% MVC is due to the decrease in the firing rates of the motor units and that recruitment and synchronization do not play a significant role. But apparently at 75% MVC, other physiological correlates affect the EMG signal. Synchronization is a likely candidate to explain the different behavior of the root-mean-squared curves at 75% MVC. This indication is also implied in the behavior of the empirical mean rectified parameter and the variance of the rectified signal. However, it is necessary to emphasize that this result does not provide direct evidence of synchronization.

During a muscle contraction maintained at a constant force, if the firing rate of the motor units decreases and there is no significant recruitment, a complementary mechanism must occur to maintain the constant force output. One possible mechanism is the potentiation of



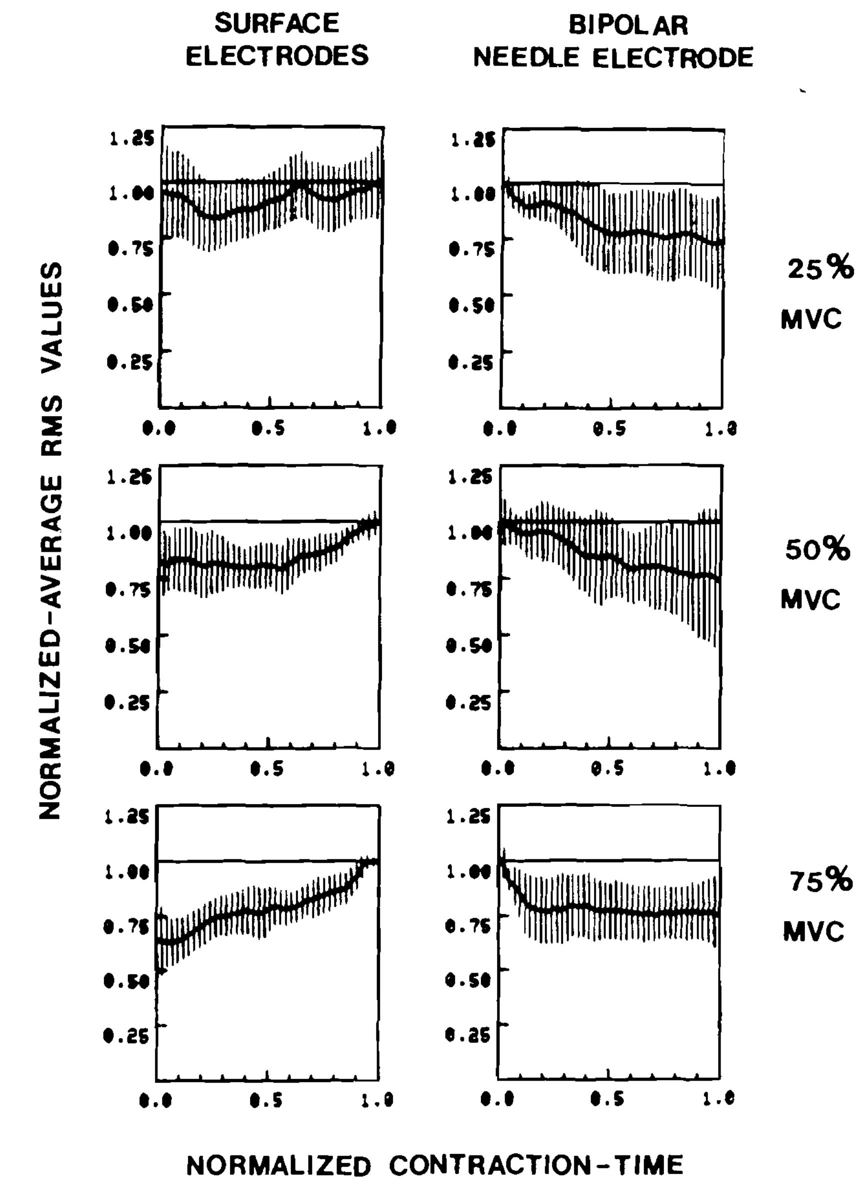



**Figure 3.14.** Average of the normalized root-mean-squared values from all subjects, plotted as a function of contraction time. Both the amplitude and time duration are normalized to their respective maxima. The *vertical lines* indicate 1 SD about the average.

twitch tension of the motor units as a contraction progresses. Evidence for potentiation of twitch tension caused by sustained repetitive stimulation has been presented by Gurfinkel' and Levik (1976) *in situ* in the human forearm flexors and by Burke et al (1976) *in vivo* in the cat gastrocnemius.

Now, let us consider the EMG signals recorded with surface electrodes. Why is the root-mean-squared value increasing during all three force levels when the firing rate is decreasing? The behavior of the firing rate is not affected by the type of electrodes used to record the signal. One possible explanation is as follows. During a sustained contraction, the conduction velocity along the muscle fibers decreases (Stålberg, 1966; Lindström et al, 1970). As a result, the time duration of the MUAPs increases (De Luca and Forrest, 1973a; Broman, 1973). This change in the shape of the MUAPs is reflected in the power density spectrum as a shift toward the lower frequencies which has been well documented. Hence, more signal energy passes through the tissue between the active fibers and the surface electrodes. Lindström et al (1970) have made theoretical calculations which show that the muscle tissue and differential electrodes act as low-pass filters. (Refer to Figs. 2.10 and 2.16 in the previous chapter for details.) As the distance between the active fibers and the electrodes increases, the bandwidth of the tissue-filter decreases. The increasing effect on the signal, due to the tissue and electrode filtering, overrides the simultaneously decreasing effect of the firing rate. The filtering effect is not seen in the signal recorded with the indwelling electrodes because the active fibers are much closer to the detection electrode.

Hence, the apparent paradox in the behavior of the EMG signal recorded with surface and indwelling electrodes is resolved by considering the detection arrangement.

It has been shown that the modeling approach is in agreement with the empirical results that could be tested. The model has also helped to resolve some ambiguities as well as to give insight into the information contained in the EMG signal. The limited discussion and arguments based on the describable known MUAPT behavior that have been presented are not sufficient to establish the generality of the model. However, as additional information describing the behavior of MUAPTs becomes available (especially the force dependence), the model should prove to be more useful and revealing.

## TIME DOMAIN ANALYSIS OF THE EMG SIGNAL

Most of the parameters and techniques that will be discussed in this section have found wide use over the past years. Some of the approaches are relatively new and have only received acclaim among some groups.

As may be seen from the previous sections, the EMG signal is a time- and force (and possibly other parameters)-dependent signal whose amplitude varies in a random nature above and below the zero value. Typically, the signal is recorded with AC coupled amplifiers. This guarantees that the average value (in this case the DC level) will be zero. This situation will prevail even if the amplifers are DC coupled, if the DC polarization potential is zero. In any case, it is apparent that simple averaging of the signal directly will not provide any useful information.



### Rectification

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

### Smoothing of the Rectified Signal

The rectified signal still expresses the random nature of the amplitude of the signal. A useful approach for extracting amplitude-related information from the signal is to smoothen the rectified signal. This procedure may be accomplished either by analog means or digital means. The concept of smoothing involves the suppression of the high-frequency fluctuations from a signal so that its deflections appear smoother. This may now be recognized as low-pass filtering procedure that has been discussed at the beginning of Chapter 2. The amount of smoothing performed on the signal will depend on the bandwidth of the low-pass filter that is used; the smaller the bandwidth, the greater the smoothing.

### Averages or Means of Rectified Signals

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

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

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

The above expression will provide only one value over the time window $T = t_j - t_i$. In order to obtain the time-varying average of a complete record of a signal, it is necessary to move the time window $T$ duration along the record; this operation is referred to as *moving average*. This may be accomplished in a variety of ways. For example, the window $T$ may be shifted forward one, or two, or any number of digitized time intervals up to the time ($T$) equivalent to the width of the window. In the latter case, each average value that is produced from the integral operation is unbiased, that is, calculated from data not common to the previous value. When the time window is shifted for a time less than its width, the calculated average values will be biased, that is, each average value will have been derived from some data that is common to the data used to calculate the previous average value. The moving average operation may be expressed as:

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

Like the equivalent operation in the analog sense, this operation introduces a lag, that is, $T$ time must pass before the value of the average of the $T$ time interval may be obtained. In most cases this outcome does not present a serious restriction, especially if the value of $T$ is chosen wisely. For typical applications we suggest values ranging from 100 to 200 ms. However, it should be noted that the smaller the time window $T$, the less smooth will be the time-dependent average (mean) of the rectified signal. The lag may be removed by calculating the average for the middle of the window.

$$\overline{|m(t)|} = \frac{1}{T}\int_{t-T/2}^{t+T/2} |m(t)|\, dt$$

But, in this case, fringe problems occur at the beginning and the end of a record when either side of the window is less than $T/2$.

## Integration

The most commonly used and abused data reduction procedure in electromyography is the concept of integration. One of the earliest (if not the earliest) references to this parameter was by Inman et al (1952). They erroneously applied the term. Their analysis procedure used a linear envelope detector to follow the envelope of the EMG signal as the force output of the muscle was varied. It is not surprising that with such an incorrect introduction the term should become improperly applied. The literature of the past three decades is swamped with improper usage of this term although, happily, within the past decade it is possible to find an increasing number of proper usages.

The term *integration*, when applied to a procedure for processing a signal, has a well-defined meaning which is expressed in a mathematical sense. It applies to a calculation which obtains the area under a signal or a curve. The units of this parameter are V·s or mV·ms.

It is apparent that a signal, such as the observed EMG signal which has



an average value of zero, will also have a total area (the integrated value) of zero. Therefore, the concept of integration may only be applied to the rectified value of the EMG signal. The operation is expressed as:

$$I\{|m(t)|\} = \int_{o}^{t} |m(t)|\, dt$$

Note that the operation is a subset of the procedure of obtaining the average rectified value. Since the rectified value is always positive, the integrated rectified value will increase continuously as a function of time. The only difference between the integrated rectified value and the average rectified value is that in the latter case the value is divided by $T$, the time over which the average is calculated. *There is no additional information in the integrated rectified value.*

As in the case with the average rectified value, the integrated rectified value may be more usefully applied by integrating over fixed time periods, thereby indicating any time-dependent modifications of the signal. In such cases, the operation may be expressed as

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

We suspect that there are two principal reasons which account for the wide spread usage of this operation. The first is simply its historical precedent. Regardless of the improper usage of the operation, its application has been continuously employed over the past three decades. The second is that if a sufficiently long integration time $T$ is chosen, the integrated rectified value will provide a smoothly varying measure of the signal or a function of time.

## The Root-Mean-Square (RMS) Value

The derivations of the mathematical expressions for the time- and force-dependent parameters presented in the earlier parts of this chapter indicated that the rms value provided more information than the previously described parameters. (Refer to Figure 3.11 for more details.) However, its use in electromyography has been sparse in the past. The recent increase in its use is possibly due to the availability of analog chips that perform the rms operation and to the increased awareness for technical competence in electromyography. The rms value is obtained by performing the operations described by the term, in reverse order, that is

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

*We recommend the use of this parameter above the others.*

A comparison of the analysis techniques operating on the same signal is presented in Figure 3.15. In this case the signal was obtained with wire electrodes in the biceps brachii during an isometric contraction.

### Zero Crossings and Turns Counting

This method consists of counting the number of times per unit time that the amplitude of the signal contains either a peak or crosses the zero

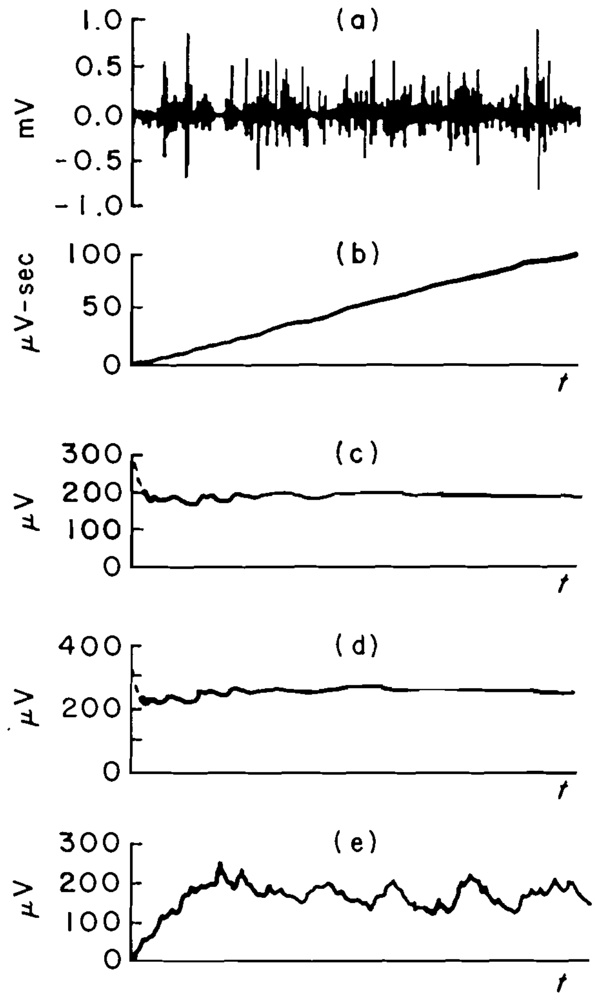



**Figure 3.15.** Comparison of four data reduction techniques: (a) raw EMG signal obtained with wire electrodes from biceps brachii during a constant-force isometric contraction; (*b*) the integrated rectified signal, (c) the average rectified signal; (*d*) the root-mean square signal; (e) smoothed rectified signal. In the latter case, the signal was processed by passing the rectified EMG signal through a simple RC filter having a time constant of 25 ms. The time base for each plot is 0.5 ms. (From J.V. Basmajian et al, 1975.)

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value of the signal. It was popularized in electromyography by Willison (1963). The relative ease with which these measurements could be obtained quickly made this technique popular among clinicians. Extensive clinical applications have been reported, some indicating that a discrimination may be made between myopathic and normal muscle. However, such distinctions are usually drawn on a statistical basis.

This technique is not recommended for measuring the behavior of the signal as a function of force (when recruitment or decruitment of motor units occurs) or as a function of time during a sustained contraction. Lindström et al (1973) have shown that the relationship between the turns or zeros and the number of MUAPTs is linear for low level contractions. But, as the contraction level increases, the additionally recruited motor units contribute MUAPTs to the EMG signal. When the signal amplitude attains the character of Gaussian random noise, the linear proportionality no longer holds.

## FREQUENCY DOMAIN ANALYSIS OF THE EMG SIGNAL

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

An idealized version of the power density spectrum of the EMG signal, along with various parameters of interest, is presented in Figure 3.16.

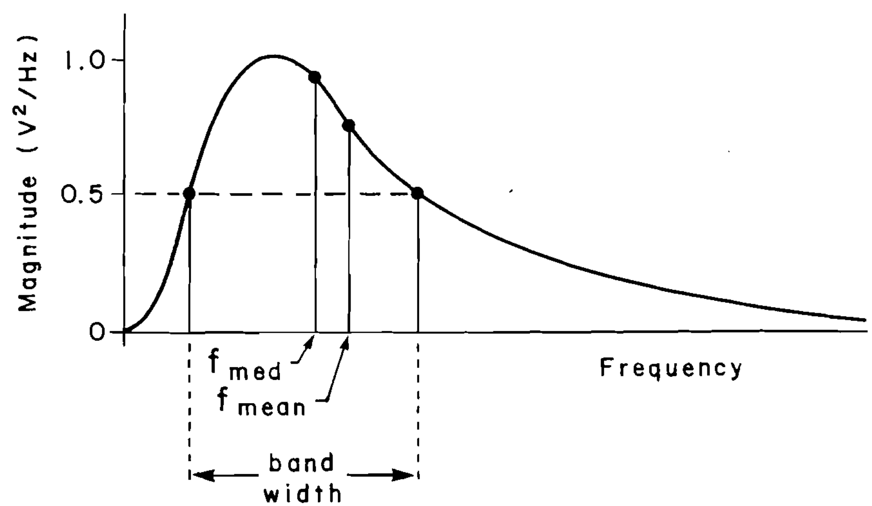


**Figure 3.16.** An idealized version of the frequency spectrum of the EMG signals. Three convenient and useful parameters: the median frequency, $f_{med}$; the mean frequency, $f_{mean}$; and the bandwidth are indicated.

Note that this plot has linear scales, because in our opinion such a representation provides a more direct expression of the power distribution. A logarithmic scale, which is the scale of preference in other disciplines such as acoustics, would compress the spectrum and unnecessarily distort the distribution.

Three parameters of the power density spectrum may be conveniently used to provide useful measures of the spectrum. They are: the median frequency, the mean frequency, and the bandwidth of the spectrum. Other parameters such as the mode frequency and ratios of segments of the power density spectrum have been used by some investigators but are not considered reliable measures, given the inevitably noisy nature of the spectrum (refer to Fig. 3.12). The bandwidth measure has been discussed in the previous chapter. Note that because the amplitude scale is in V/Hz, the 3 dB points or the corner frequencies are defined by a decrease of a factor of 0.5. The median frequency and the mean frequency are defined by the following equations:

$$\int_{o}^{f_{med}} S_m(f)\, df = \int_{f_{med}}^{\infty} S_m(f)\, df$$

$$f_{mean} = \frac{\int_{o}^{f} f\, S_m(f)\, df}{\int_{o}^{f} S_m(f)\, df}$$

where $S_m(f)$ is the power density spectrum of the EMG signal. Stulen and De Luca (1981) performed a mathematical analysis to investigate the restrictions in estimating various parameters of the power density spectrum. The median and mean frequency parameters were found to be the most reliable, and of these two, the median frequency was found to be less sensitive to noise. This quality is particularly useful when the signal is obtained during low level contractions when the signal to noise ratio may be less than 6.

CHAPTER 4

# Decomposition of the EMG Signal and Analysis of the Motor Unit Action Potential Trains

Decomposition of the EMG signal is the procedure by which the EMG signal is separated into its constituent motor unit action potential trains. This concept is illustrated in Figure 4.1. The development of a system to accomplish such a decomposition will be beneficial both to researchers interested in understanding motor unit properties and behavior, and to clinicians interested in assessing and monitoring the state of a muscle.

In the clinical environment, measurements of some characteristics of the motor unit action potential (MUAP) waveform (for example, shape, amplitude, and time duration) are currently used to assess the severity of a neuromuscular disease or, in some cases, to assist in making a diagnosis. Thus, the decomposition of the EMG signal is useful in two ways. First, a partial decomposition must be implicitly performed by the clinical investigator to ensure that what is actually observed is a MUAP and not a superposition of two or more MUAPs or some other ephemeral artifact. Second, averaging the MUAP waveforms present in the same train will produce a low noise representation of the MUAP and, hence, provide a more faithful representation of the events occurring within the muscle. Any decomposition scheme devised for such application (i.e., to extract only MUAP shape and amplitude) will have weak constraints on its performance. A useful technique should allow detection of some, but not necessarily all, firings of a single unit in a particular record.

For physiological investigation, both the statistic of the interpulse intervals (IPI, time between two successive firings of the same motor unit) and the MUAP waveform characteristics are used to study motor unit properties and motor control mechanisms of muscles. In these conditions, much stronger constraints are imposed on the performance of a decomposition technique. It is desirable, in fact, to monitor the simultaneous activities of as many motor units as possible. Furthermore, all the firings of the observed motor units should be detected. Shiavi and Negin (1973) showed that an error of 1% in the detection of a motor unit firing prevented the observation of some relevant motor unit behavioral phenomena. Statistical analysis of IPI also implies acquisition and processing of relatively long EMG signal records (in the order of dozens of seconds), thus increasing the time required for the decomposition.