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

Chairman:

David A. Winter
Department of Kinesiology
University of Waterloo
Waterloo, Ontario, Canada, N2L 3G1

Members:

Gunter Rau
Helmholtz-Institut for Biomedizinische Technik
D-51 Aachen
West Germany

Roland Kadefors
Projekt Lindholmen Center of Education
Production and Research
Box 8714
S-402 75 Goteborg
Sweden

Holger Broman
Department of Applied Electronics
Chalmers University of Technology
S-412 96 Goteborg
Sweden

Carlo J. de Luca
Children's Hospital Medical Center
300 Longwood Avenue
Boston, Massachusetts 02115
U.S.A.

Authors' Note

This report is the final report of an ISEK Ad Hoc Committee that was formed in 1977 to deal with the problems arising from inconsistent and erroneous terms and units in the reporting of EMG research. The Committee has addressed this problem at all levels, from the electrophysiological terminology right through to the more common processing techniques. Also, the Committee makes some recommendations regarding technical standards that should be aimed for by all researchers. This report has evolved from the First Interim Report presented at the 4th Congress of ISEK in Boston in August 1979, from meetings held with researchers at the Congress, and from many personal discussions held over the past few years. The document presents not only the fundamental theoretical and physiological relationships but also the total practical experience of the Committee and those consulted. Only a few references have been listed and are intended to represent or amplify certain issues because a complete bibliography of electromyography would occupy several hundred pages.

The Problem

It is axiomatic that all researchers in any scientific area should be able to communicate with clarity the results of their work. The area of electromyography is one where inconsistent and erroneous terms and units are the rule rather than the exception. Even veteran researchers are continuing to contribute to the confusion. The research in many major papers cannot be replicated because of lack of detail on the protocol, recording equipment or processing technique. For example, the term integrated EMG(IEMG) has been used to describe at least 4 different processing techniques, and the units employed can be mV, mV/sec., mV·sec, or just arbitrary units! No wonder there are conflicts and misunderstandings.

Form of Report

The report concentrates on four major aspects of the myoelectric signal and its subsequent recording and processing. Figure 1 shows the breakdown of the report: Part 1—The Neuromuscular Domain, Part II—The Recording System, Part III—Temporal Processing and Part IV—Frequency Processing. In addition, the Committee presents a Part V—General Experimental Details, which outlines important kinesiological and experimental information that should be reported. References made to the literature are by no means complete, rather they are cited as examples of erroneous or anomalous reporting. Finally, a checklist is given to summarize the recommendations of the Committee.

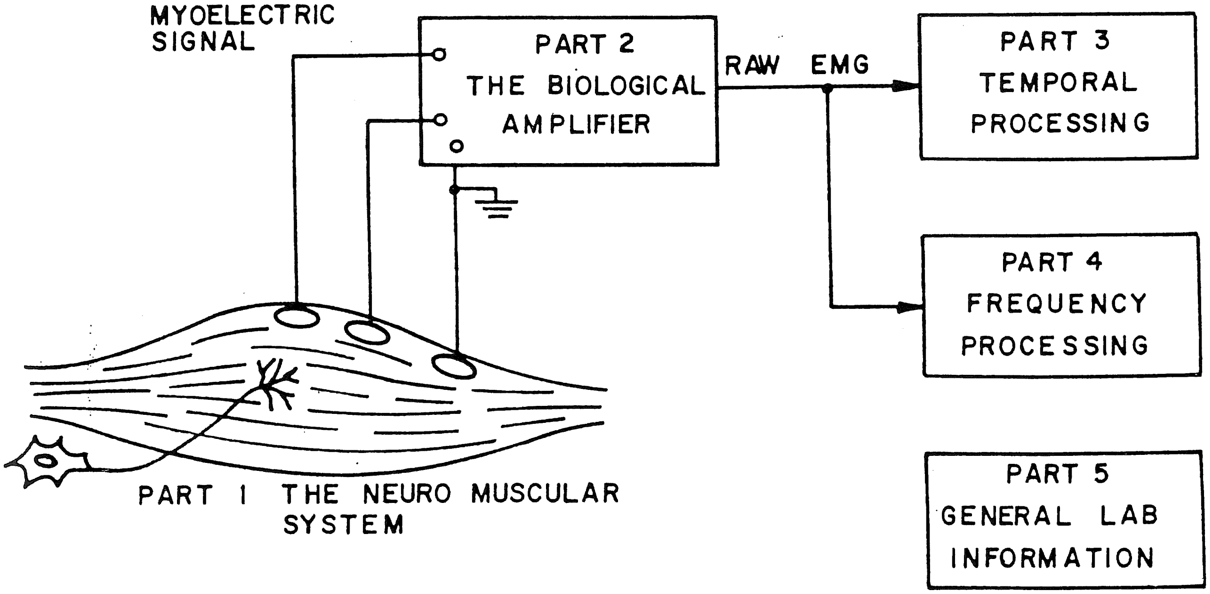

Figure 1. Schematic outline of scope of report.

Part I: Terminology Applied to the Neuromuscular Unit

References: DeLuca, 1979; Buchthal and Schmalbruck, 1980.

α motoneuron—is the neural structure whose cell body is located in the anterior horn of the spinal cord and through its relatively large diameter axon and terminal branches innervates a group of muscle fibres.

motor unit (MU)—is the term used to describe the single smallest controllable muscular unit. The motor unit consists of a single α motoneuron, its neuromuscular junction, and the muscle fibres it innervates (as few as 3, as many as 2000).

muscle fibre action potential or motor action potential (MAP)—is the name given to the detected waveform resulting from the depolarization wave as it propogates in both directions along each muscle fibre from its motor end plate. Without the use of special micro techniques it is generally not possible to isolate an individual MAP.

motor unit action potential (MUAP)—is the name given to the detected waveform consisting of the spatio-temporal summation of individual muscle fibre action potentials originating from muscle fibres in the vicinity of a given electrode or electrode pair. Its shape is a function of electrode type (recording contact area, inter-wire spacing, material, etc.), the location of the electrode with respect to the fibres of the active motor unit, the electrochemical properties of the muscle and connective tissue and the electrical characteristics of the recording equipment. A MUAP detected by a surface electrode will be quite different from the MUAP detected by an indwelling electrode within the muscle tissue. Each motor unit will generally produce a MUAP of characteristic shape and amplitude, as long as the geometric relationship between the electrode and active motor unit remains constant. However, when the MUAP consists of less than about five MAPs the waveform may vary randomly due to the "jitter" phenomenon of the neuromuscular junction. A given electrode will record the MUAPs of all active motor units within its pick-up area.

motor unit action potential train (MUAPT)—is the name given to a repetitive sequence of MUAPs from a given motor unit.

inter pulse interval (IPI)—is the time between adjacent discharges of a motor unit. The IPI depends on the level and duration of a contraction and even at an attempted constant tension the IPI is irregular. Its variation is conveniently seen in an IPI histogram.

motor unit firing rate—is the average firing rate of a motor unit over a given period of time. When a motor unit is first recruited it fires at an initial rate and generally increases as the muscle tension increases. Meaningful estimates of average firing rates should be calculated over at least six consecutive IPIs.

synchronization—is the term to describe the tendency for a motor unit to discharge at or near the time that another motor unit discharges. It therefore describes the interdependence or entrainment of two or more motor units.

myoelectric signal—is the name given to the total signal seen at an electrode or differentially between two electrodes. It is the algebraic summation of all MUAPTs from all active motor units within the pick-up area of those electrodes. The myoelectric signal must be amplified before it can be recorded (when it is called an electromyogram).

Part II: The Recording System

Introduction

With respect to recording electromyographic signals the most important property is the distribution of signal energy in the signal frequency band. The signal frequency spectrum picked up by electrodes depends on:

(a) the type of muscle fiber since the dynamic course of depolarization/repolarization is specific to the muscles (e.g. heart muscle cells, skeletal muscle or smooth muscle cells),

(b) the characteristics of the volume conductor: the electrical field is influenced by the shape, conductivity and permitivity of tissues and the shape of the boundaries,

(c) the location and physical structure of the electrodes, especially the distance from the cell surface.

Theoretically, the signal source frequency can only be determined by microelectrode techniques at the cellular level. In practice different types of macro-electrodes are utilized: (a) needle electrodes, (b) wire electrodes, (c) surface electrodes. Needle and wire electrodes are invasive, surface electrodes are non-invasive. In clinical diagnosis, needle and wire electrodes are indispensible but the progress in surface electrode methodology has progressed very rapidly mainly because of its wide use. Unfortunately, there are many difficulties which have to be taken into account, and which are now discussed in more detail.

The Problem

For surface electrode recording, two aspects have to be considered: (1) The electrophysiological sources within the body as a volume conductor, resulting in an electrical field at the skin surface (the upper boundary of the volume). Each motor unit contributes independently of each other, and the separation of each of these different sources is increasingly difficult as their distance to the electrode increases. (2) The detection of electrophysiological signals at the skin surface has to take into account the electrical properties of the skin, the electrodes, as well as the signal characteristics. The distortion and the disturbances can be reduced by a proper design.

Typical amplitude and frequency range of the signals in question are shown in Table 1. However, the actual ranges depend greatly on the electrode used.

Table 1

Signal	Amplitude Range, mV	Signal Frequency Range, Hz	Electrode Type
Indwelling EMG	0.05 - 5	0.1 - 10,000	Needle/Wire
Surface EMG	0.01 - 5	1 - 3,000	Surface
Nerve Potentials	0.005-5	0.1 - 10,000	Needle/Wire

a. Surface Electromyography

The electrodes are electrically coupled to the motor action potentials propagating within the muscle tissue. The stages of this coupling are depicted in Figure 2. These different stages are considered separately and their properties related in order to find out which are first order effects and which can be neglected; please consider Figure 3.

Signal source and body tissues

Within the body a relatively high conductance exists due to the concentration of freely moving ions. The specific resistance is, in the signal frequency band up to 1 KHz, ohmic, and in the order of magnitude of 100-1000 ohm·cm; this value depends on the nature of the tissue in consideration (fat, lung, blood, human trunk). At about 1 KHz the capacitive current and the resistive current are of equal value. Therefore, when using needle or wire electrodes the input impedance of the amplifier should be not less than 1 Megohm, if signal frequencies above 2 KHz have to be detected.

Skin resistance

The resistance of the unprepared skin is complex and varies over a wide range dependent on (a) skin site, (b) subject and (c) time and (d) skin preparation. In worst cases we see values up to several megohms at low frequencies. In order to minimize the influence of this complex resistance on the signal, high input resistance amplifiers can be achieved very easily by means of modern FET-technology.

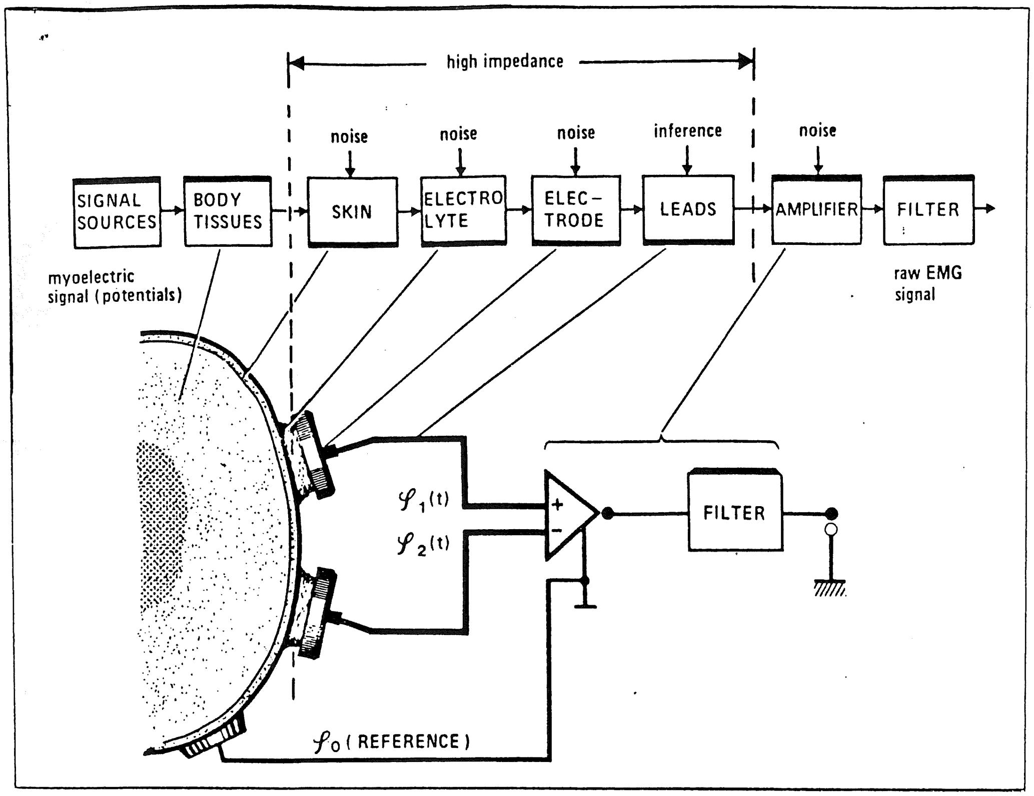

Figure 2. Measuring set up for surface electromyography.

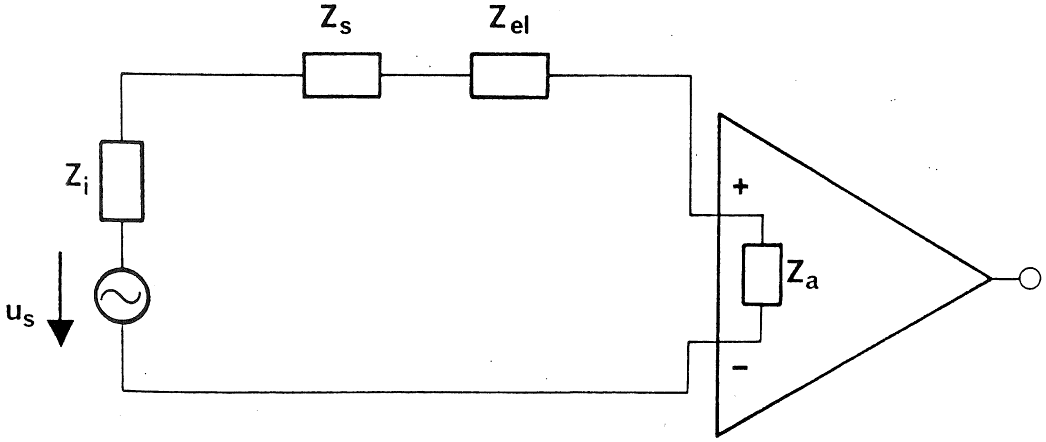

Figure 3. Simplified circuit with lumped elements describing the most important electrical properties. Zs = complex skin resistance, Zel = complex resistance of electrodes and electrode/electrolyte transition, Za = complex input resistance of the amplifier, Zi = complex resistance of the body tissues.

In most existing instrumentation an input impedance of 10 Megohms or even 1 Megohm is common, thus the skin has to be prepared (rubbed, abraded) until the resistance is down to less than 100 KΩ or even 10 KΩ respectively. Of course, then the impedance between the electrodes should be measured over the whole frequency band. However, the use of a high performance amplifier eliminates the preparation of the skin and the need to check the resistance, thus simplifying the measuring procedure.

Electrode, electrolyte and transition

The electrode resistance and the electrode/electrolyte transition are dependent on the electrode material and the electrolyte (paste or cream) in use. Here not only the electrical but also the mechanical and physiological properties have to be taken into account. Low polarization voltage is observed at Ag-AgCl-Electrodes; unfortunately the Ag-content varies dependent on the manufacturer resulting in differing properties. Stainless steel electrodes are adequate when using them on unprepared skin and for a frequency range above 10 Hz. Long term stability is excellent with black platina.

The electrode and electrode/electrolyte transition can be neglected with respect to the unprepared skin resistance over the entire frequency range; both have similar electrical characteristics. The impedance of both unprepared and prepared skin can be neglected as long as the input impedance of the amplifier is sufficiently high (at least 100 times the skin impedance).

The amplifier

Modern technology has resulted in amplifier specifications to overcome the measuring problems. However, old instrumentation gives rise to problems of input impedance, input current and noise. The desired and recommended specifications for newly designed amplifiers are as follows:

CMRR ⟩ 90 dB

Input resistance ⟩ 10^{10} ohms for dc coupled; ⟩ 10^{8} ohms at 100 Hz for ac coupled.

Input current ⟨ 50 nA for directly coupled amplifiers

Noise level ⟨ 5 μVrms with a source resistance of 100 KΩ and a frequency bandwidth from 0.1-1000 Hz.

The low input current is to be desired because of artifacts which may be caused by modulation of the skin resistance. It is known that the value Zs (Figure 3) varies with mechanical pressure. Time varying pressure occurs when movement is induced to the electrodes via cable motion. With a change △Zs = 50 KΩ and an input current of 50 nA the generated voltage is 2.5 mV.

In addition, pressure applied to the skin produces an artifact voltage which cannot be separated from the signal voltage, except by high pass filtering above the frequency of the movement artifact (about 10 Hz). In a recently performed study (Silny, et al., 1979) on cable properties some cables produced movement artifacts of several millivolts, even when using "special ECG cable." There are cable types now available which produce 100 times less voltage artifact than other cables.

Regarding the noise, the amplifier must not be tested with a short circuit at the input since the amplifier noise is insignificant compared to when using a high resistance source. Therefore, a high resistance test circuit has to be used when specifying the relevant amplifier noise. Of course, the noise situation is improved when the skin resistance has been reduced by skin preparation. Then an amplifier can be used which is of lower input impedance while the current noise becomes less important. Also, a reduction of signal frequency bandwidth reduces the noise level, and such a reduction should be done whenever possible.

b. Wire and Needle Electrodes

When using needle and wire electrodes the skin resistance will not impair the signal acquisition as long as an amplifier has a 10 MΩ input impedance. Even fine wire electrodes with a small area of active tip show a relatively low impedance. The properties are mainly defined by the electrode-electrolyte transition and if there are no noise problems in specific applications, one can use the same amplifier as in surface electromyographs.

Summary and Recommendations

a. Amplifier:

(i) The frequency range of the amplifier channel should be chosen according to Table 1. If in surface electromyography the lower cut-off frequency has been selected to be 20 Hz for suppression of movement artifacts it has to be reported.

Report: upper cut-off frequency
lower cut-off frequency
type of filter, slope

(ii) If a DC coupled amplifier is used, a higher input impedance and low input current are desired. AC coupling by a capacitor in each of the two differential electrode leads is commonly used but may give rise to large movement artifacts and polarization voltages if the input impedance is too high. Thus a somewhat lower input impedance is suggested.

Report: input impedance
input current, if dc coupled
noise with 100 KΩ at the input, $1 < f < 1000$ Hz
CMRR

b. Electrodes:

Report: type
spacing between recording contacts
material
stability, if important
off set voltage, if important

c. Skin:

Report: site
complex resistance in the whole signal frequency range if a low resistance input amplifier is used
preparation of skin

d. Electrode paste:

Report: type
manufacturer
electrochemical properties

Note #1: New instrumentation should be battery operated, at least in the first differential amplifier stage. This will result in a marked suppression of hum interference and when coupled to a computer, or other recording equipment the necessary isolation is achieved.

Note #2: The amplifier can be miniaturized and attached close to the electrodes. The shorter the electrode leads the smaller the picked up disturbances (hum and other interferences). In addition, the CMRR value is not decreased by unnecessary unsymmetries.

Note #3: The best procedure for adjusting the filter characteristics of the complete channel is to have the frequency band unrestricted, perform a frequency analysis and then adapt the filter bandwidth to the signal bandwidth.

Part III: Temporal Processing

Raw EMG

Visual inspection of the raw EMG is the most common way of examining muscle activity as it changes with time. Correlation of such phasic activity with other biomechanical variables (joint angles, acceleration, moments of force, etc.) or physiological variables has added to the understanding of normal muscle function as well as special motor functions in pathologies, in ergonomic situations and in athletic events. The amplitude of the raw EMG when reported should be that seen at the electrodes, in mV or μV, and should not reflect the gain of any amplifiers in the recording system.

Detectors

The quantification of the "amount of activity" is necessary so that researchers can compare results, not only within their own laboratories, but between laboratories. It is important to know not only when a muscle turns on or off, but how much it is on at all times during a given contraction. The basis for most of this quantification comes from a detector. A linear detector is nothing more than a full-wave rectifier which reverses the sign of all negative voltages and yields the absolute value of the raw EMG. Non-linear detectors can also be used. For example, a square law detector is the basis of the root mean square (rms) value of the EMG. The important point is that the details of the detector must always be reported because the results of subsequent processing and conclusions regarding muscle function are strongly influenced by the detector (Kadefors, 1973).

Types of Averages

(a) Average or mean

The mean EMG is the time average of the full-wave rectified EMG over a specified period of time. It is therefore, important for the researcher to specify the time over which the average was taken. Is it the duration of the contraction, the stride period, or the total exercise period? The mean value should be reported in mV (or μV), and for a period of $(t_2 - t_1)$ seconds,

$$\text{Mean} = \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} |\text{EMG}| \, dt \qquad \text{mV}$$

(b) Moving averages

It is often valuable to see how the EMG activity changes with time over the period of contraction, and a moving average is usually the answer. Several common processing techniques are employed with the detected signal (usually full-wave rectified). All moving averages are in mV or μV.

The most common is a low-pass filter, which follows the peaks and valleys of the full-wave rectified signal. Thus the characteristics of the filter should be specified (i.e., 2nd order Butterworth low-pass filter with cut-off at 6 Hz). It is somewhat confusing and meaningless to report the averaging time constant of the filter especially if a 2nd order or higher order filter is used. The combination of a full-wave rectifier followed by a low-pass filter is commonly referred to as a linear envelope detector.

With the advent of digital filtering the processing of the EMG can be processed many novel ways. Analog low-pass filters, for example, introduce a phase lag in the output, whereas digital filters can have zero phase shift (by first filtering in the positive direction of time, then refiltering in the negative direction of time). If such processing is used, the net filter characteristics should be quoted (i.e., 4th order zero-lag, low pass Butterworth filter with cut-off at 10 Hz).

Probably the most common digital moving-average type is realized by a "window" which calculates the mean of the detected EMG over the period of the window. As the window moves forward in time a new average is calculated. It can be expressed as follows:

$$\text{Window average } (t) = \frac{1}{T} \int_{t-T/2}^{t+T/2} |\text{EMG}| \, dt \qquad \text{mV}$$

Its value is in mV, and all that is needed is to specify the window width, T. Normally the average is calculated for the middle of the window because it does not introduce a lag in its output. However, if the moving average is calculated only for past history the expression becomes:

$$\text{Window average}(t) = \frac{1}{T} \int_{0}^{t+T} |\text{EMG}|\, dt \qquad \text{mV}$$

Such an average introduces a phase lag which increases with T, thus if this type is used T should be clearly indicated. Other special forms of weighting (exponential, triangular, etc.) should be clearly described.

(c) Ensemble average

In any repetitive movement or evoked response it is often important to get the average pattern of EMG activity. An ensemble average is accomplished digitally in a general purpose computer or in special computers of average transients (C.A.T.). With evoked stimuli it is often possible to average the resultant compound action potentials. The time-averaged waveform has an amplitude in mV, and the number of averages is important to report. Also, the standard error at each point in time may be important. The expression for N time-averaged waveforms at any time t is:

$$\text{EMG}(t) = \frac{1}{N} \sum_{i=1}^{N} |\text{EMG}_i|(t) \qquad \text{mV}$$

where EMG_i is the ith repetition of the EMG waveform to be averaged. An example of such an averaged waveform is presented in Figure 4: the linear envelope of the soleus muscle was averaged over 10 strides. A complete stride is shown; the amplitude would normally be in mV or μV, but here it is reported as a percentage of the EMG at 100% maximum voluntary contraction. There is no consensus at present as to standard methods of elliciting a maximum contraction because of variations in muscle length with different limb positions and the inhibitary influences present in agonist and antagonist muscle groups. However, such normalization techniques are indispensible for comparisons between different subjects and for retrials on the same subject.

Integrated EMG

Probably the most widely used (and abused) term in electromyography today is integrated EMG (IEMG). Probably the first use of the term was by Inman and co-workers (1952) when they described a waveform which followed the rise and fall in tension in the muscle. The circuit they employed was a linear envelope detector, not an integrator. The correct interpretation of integration is purely mathematical, and means the "area under the curve." The units of IEMG have also been widely abused. For example, Komi (1973) reports IEMG in mV/s, and in 1976 scales the IEMG in mV. The correct units are mV·s or μV·s. It is suspected that many of these researchers who report IEMG in mV are really reporting the average over an unspecified period of time and not an integration over that period.

There are many versions of integrated EMG's. Figure 5 shows a diagram of 3 common versions, plus the linear envelope signal that is so often misrepresented as an IEMG. The raw and full-wave rectified signals are shown for several bursts of activity and have their amplitudes reported in mV. The linear envelope as shown employed a second order low-pass filter with cutoff at 6 Hz, its amplitude also appears in mV.

The simplest form of integration starts at some preset time and continues during the total time of muscle activity. Over any desired period of time the IEMG can be seen in mV·s. A second form of integrator involves a resetting of the integrated signal to zero at regular intervals of time (usually from 50 to 200 ms), and the time should be specified. Such a scheme yields a series of peaks which represent the trend of the EMG amplitude with time; in effect, something close to a moving average. Each peak has units of mV·s (or μV·s because the integrated value over these short times will not exceed 1 mV·s). The sum of all the peaks in any given contraction should equal the IEMG over that contraction. A third common form of integration

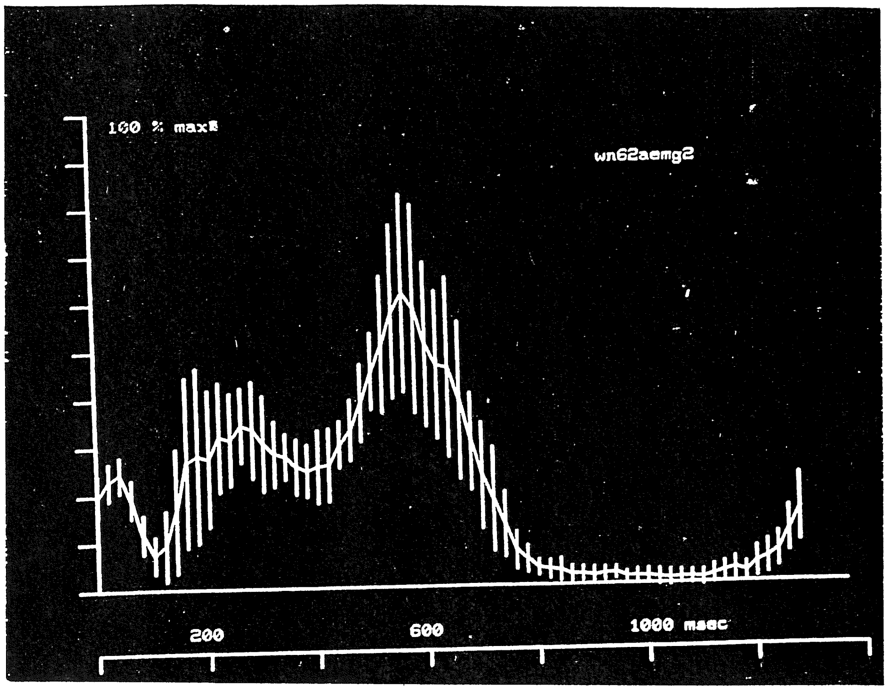

Figure 4. Average of linear envelope of soleus muscle over 10 walking strides. Contraction was normalized to 100% MVC, and standard deviation at each point in time is shown by vertical bars.

uses a voltage level reset. If the muscle activity is high, the integrator will rapidly charge up to the reset level, and if low activity occurs it will take longer to reach reset. Thus the activity level is reflected in the frequency of resets. High frequency of resets (sometimes called "pips") means high muscle activity, low frequency means low level activity, as seen by the lower trace of Figure 5. Each reset represents a value of integrated EMG and this should be specified (usually in μV·s). Again, the product of the number of resets times this calibration will yield the total IEMG over any given period of time.

Part IV: Frequency Domain Analyses

Frequency domain methods have been used for more than a century. They have proven a powerful tool in that solutions to a linear differential equation of a function of time, say, are most easily obtained using the Fourier transform or Laplace transform. A second attractive property of Fourier transforms of functions of time (such as myoelectric signals) is that the function is described as a function of frequency (not to be confused, for example, with repetition rate of a succession of motor unit action potentials). A signal having finite energy content, such as a single motor unit action potential, can be described by its energy spectrum, which gives the distribution of energy as a function of frequency. A signal having infinite energy content, such as a hypothetical infinite succession of action potentials or an infinitely long interference pattern of the activity of several motor units, can similarly be characterized by its power spectrum. In practical work, a time-limited stretch of data is often regarded as periodically repeated (from long before Moses till after the end of time and thus is for infinite duration). The concept of power spectrum is consequently used unless the data is explicitly time-limited as for instance when stress is on one action potential. The square-root of the power spectrum and the square-root of the energy spectrum are both referred to as the amplitude spectrum. Figure 6 shows examples of two motor-unit potentials (modelled as differentiated Gauss pulses) with their amplitude and energy spectra.

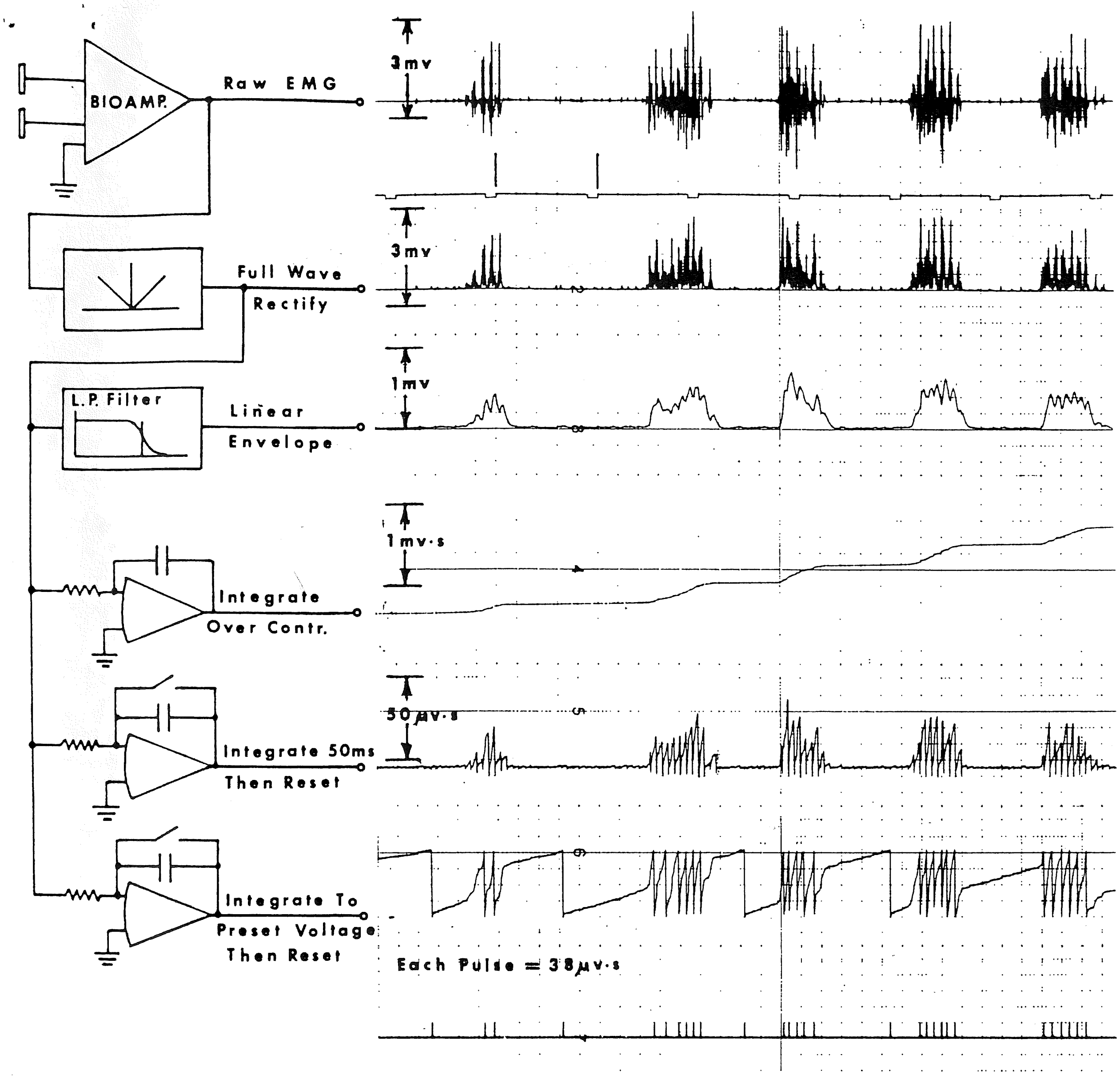

Figure 5. Example of several common types of temporal processing of the EMG.

Units of Measurement

Frequency is measured in Hz (Hertz), formerly in English literature in cycles per second. The name energy spectrum was originally devised for measures of electrical energy decomposed as a function of frequency. The unit for this quantity is joules[1] per hertz, abbreviated J/Hz or J Hz^{-1}. Similarly, the unit of the power spectrum is watts per hertz, W/Hz. Over the years, as frequency domain methods have become used more and more in the study of problems not related directly to energy and power, the names energy spectrum and power spectrum have been given a wider meaning. Thus, the units applied are not restricted to what is said above. Consider the example of EMG. The unit of power spectrum is the square of the unit of the amplitude of the myoelectric signal per hertz, that is volt squared per Hz, V^2/Hz. The unit of the power spectrum of the distance from earth to the moon (measured as a function of time) similarly is meters squared per Hz, m^2/Hz. The unit of the energy spectrum of an action potential is

---

[1] One joule was formerly referred to as one wattsecond.

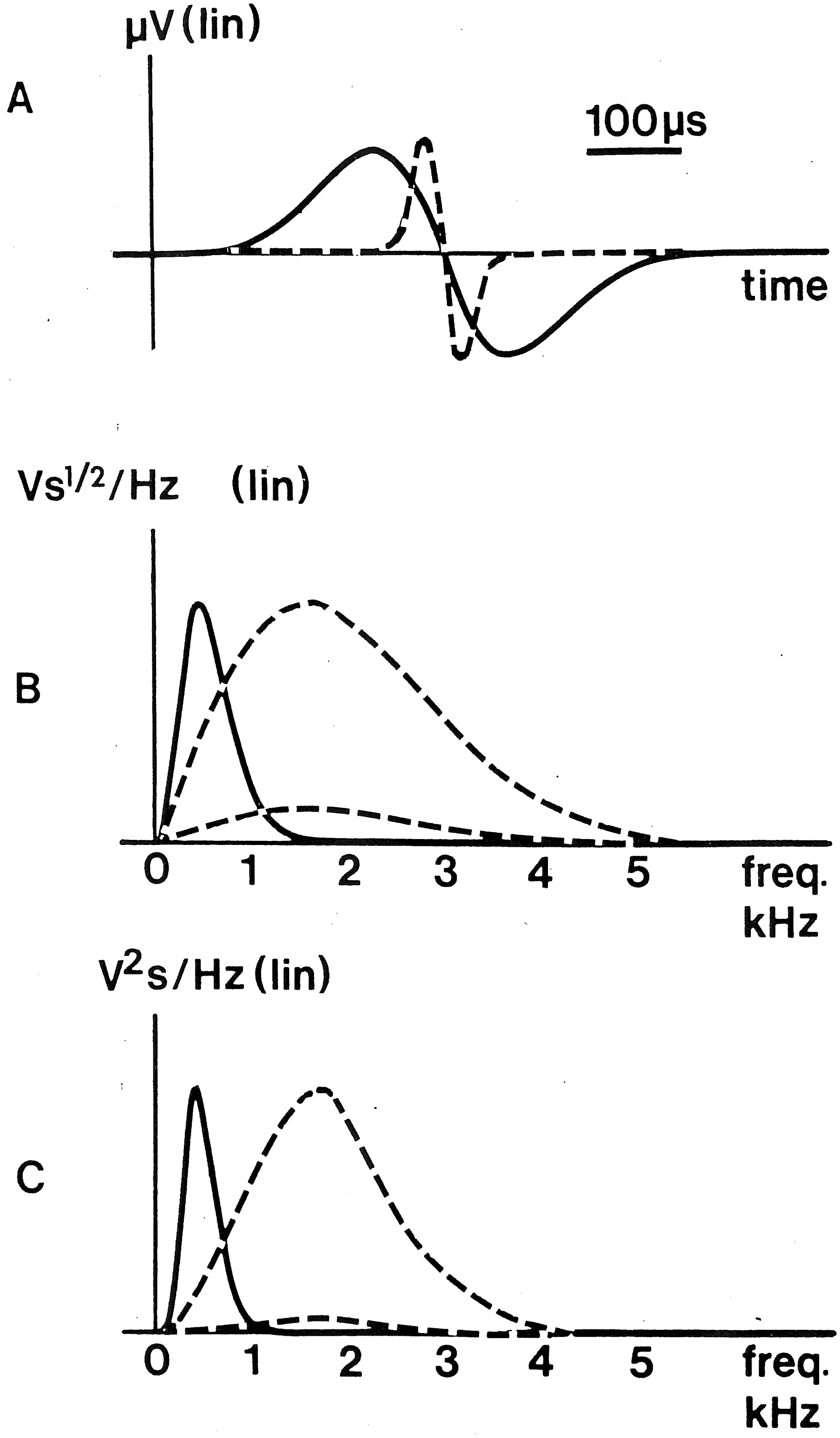

Figure 6. (a) Two examples of action potentials, modelled as differentiated Gauss pulses. The duration of the wider potential is four times that of the shorter one. (b) Amplitude spectra of the two pulses. (c) Energy spectra of the two pulses. The lower dashed curves indicate the factual spectra of the short action potential; the upper dashed curves have been normalized to the same spectral peak magnitudes as those of the wide action potential. Linear scales, arbitrary magnitudes.

V^2s/Hz. The unit for the amplitude spectrum of MEG is thus either V/Hz, if one starts from the power spectrum, or $Vs^{1/2}/Hz$, if one starts from the energy spectrum.

Logarithmic Scales and Normalization

In visualizing spectra in graphs, valuable information is often lost due to the limited dynamic range of linear scales. For instance, if the maximum value of a power spectrum is 100 V^2/Hz and interesting phenomena occur at a level of 1 or 0.1 V^2/Hz, they will obviously be lost to the eye if the plot is on linear scales. The cure is to plot the spectrum on double logarithm scales or on linear-logarithmic scales. In order to take the logarithm of a quantity, it is necessary that the quantity have no dimension. For example, the logarithm of 2 volts is not defined. By normalizing 2 volts, by referring it to for instance 1 volt (the unit of measurement), one obtains the dimensionless quantity 2, which has a logarithm. The level of reference may be chosen freely as long as it has the same unit as the variable of interest. The concept of decibel,[2] dB, is used for logarithmically scaled power, energy, and amplitude spectra. One may thus plot a spectrum on log-lin scales in the form of "decibels vs logarithmically scaled frequency." Figure 7 illustrates the effect of logarithmizing the spectra of Figure 6. Caution should be exerted to ensure that the plot should not extend below the noise level of the system.

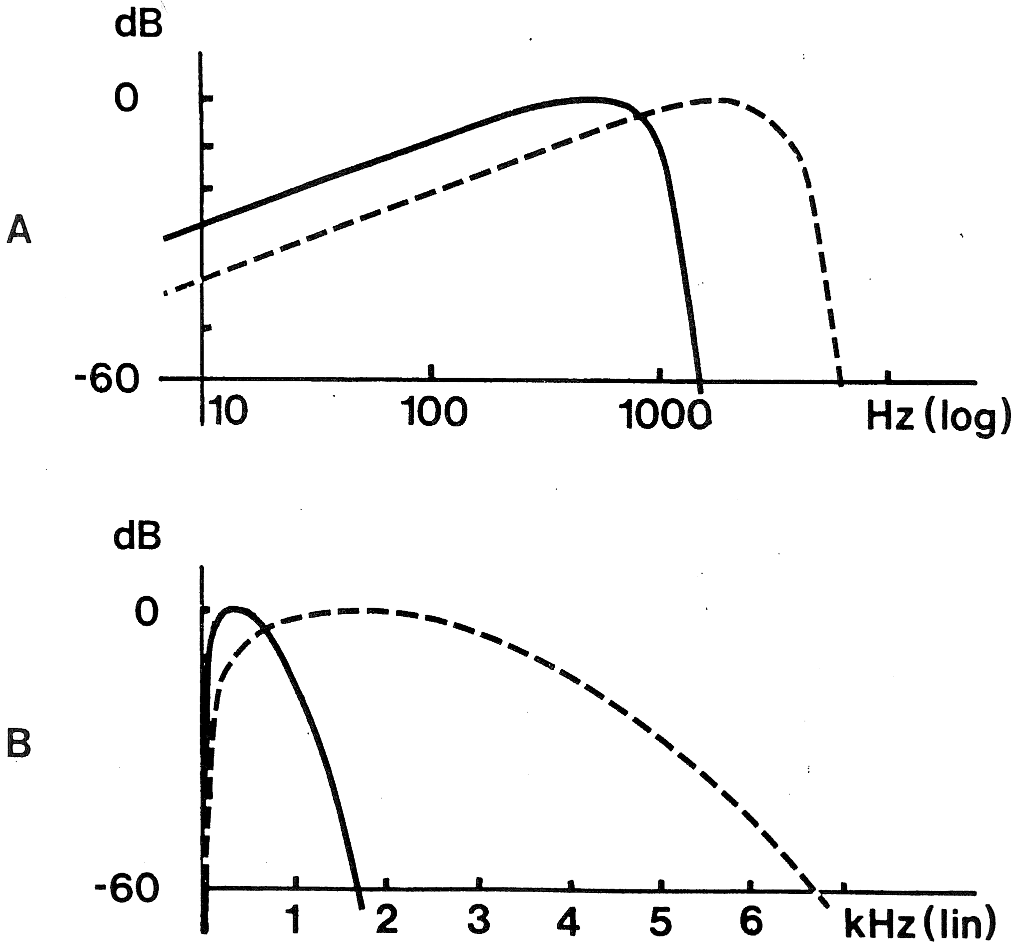

Figure 7. Effect of logarithmizing the spectra of Figure 6. (a) Double logarithmic, and (b) log/lin scales. The spectra have been normalized to the same peak magnitude.

---

[2] The decibel concept is, to be strict, linked with power ratios; the gain in decibels of a system equals ten times the base ten logarithm of the ratio of output to input power.

Discrete Parameters of Spectra

To report research results as functions of one variable is difficult. Among the problems is the question of proper statistical evaluation. One solution is to reduce the information to what is carried by discrete parameters. There is of course a large number of such parameters (this is the very cause of the problem). A few that have been used in the analysis of EMG will be mentioned here.

Figure 8(a) shows an example of a spectrum plotted on logarithmic scales. In the case of a smooth and unimodal spectrum like the one shown, one can easily find the upper and lower 3-dB frequencies, defined by the frequencies at which the spectrum has fallen 3 dB from its maximum value. The 3-dB bandwidth is defined at the difference between the upper and lower 3-dB frequencies (fb and fa, respectively), and the center frequency fc is given by the geometric mean of the 3-dB frequencies. A 3-dB drop is equivalent to a decrease by 50% in a linear scale representation of the power spectrum (see Figure 8(b)), and are therefore referred to as half-power frequencies.

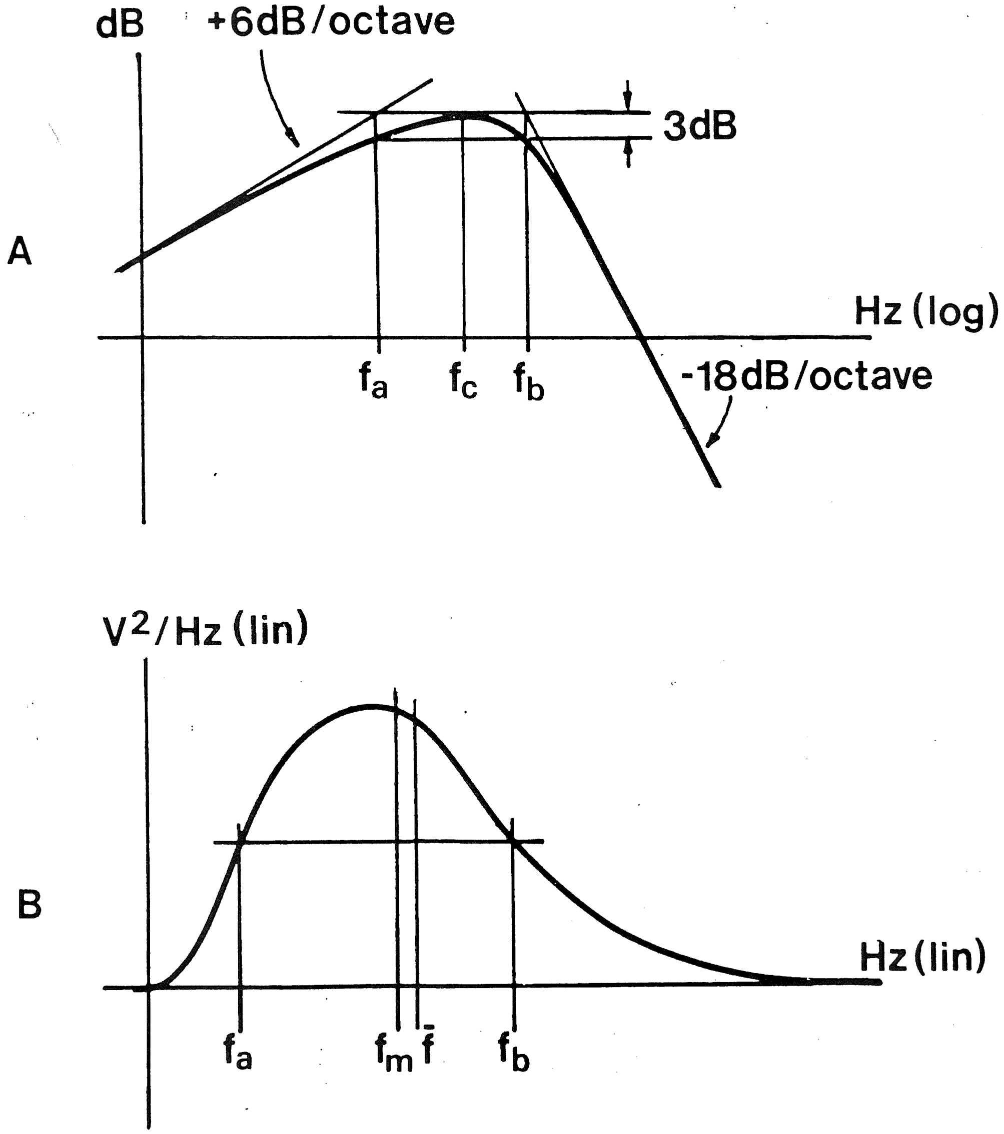

Figure 8. (a) An example of a power spectrum, with high-frequency and low-frequency asymptots indicated, and 3-dB limits given. Double logarithmic scales. (b) Same power spectrum as above, with linear scales, and with some parameters of interest given; fm and f denote the median and the mean frequencies, respectively. The locations are approximate. In principle, the areas under the curve on both sides of fm are identical, whereas f denotes the frequency where the area under the curve would balance if cut out with scissors.

In Figure 8(a) is shown a piecewise linear approximation, known as the asymptotic Bode diagram. This asymptotic diagram is entirely defined by the slopes of the lines, the frequencies where the slope changes, the breakpoints, and a scale factor.

The variables discussed so far are commonly used in engineering sciences. Now we will discuss several parameters that have parallels in statistics. In order to do so, the concepts of spectral movements will first be defined. The spectral moment of order n is given by:

$$m_n = \int_0^\infty f^n W(f)\, df$$

where f is frequency and W is the power or energy spectrum.

The mean frequency, $\bar{f}$, is the ratio between the spectral moments or orders one and zero (similar to the mean value in statistics). Thus

$$\bar{f} = \frac{m_1}{m_0} = \frac{\int_0^f f\, W(f)\, df}{\int_0^f W(f)\, df} \qquad \text{Hz}$$

The statistical bandwidth is the square-root of the difference between the ratio of the moment of order two to that of order zero and the square of the mean frequency (cf. the standard deviation in statistics). Rice has shown that, for Gaussian noise, the intensity of zero crossings equals the square-root of the ratio between the moments of orders two and zero, and the intensity of turning points equals the ratio between the fourth and second order moments.

The frequency at the maximum of the power spectrum may be called the mode (most probable frequency), and the median frequency fm is the frequency which divides the spectrum into two parts of equal power (energy), and is defined by:

$$\int_0^{fm} W(f)\, df = \int_{fm}^\infty W(f)\, df$$

All these parameters, and several not mentioned here, have been used in reports on EMG-research results. The parameters have different properties as they emphasize different aspects of the spectrum. It would be premature to recommend the use of any specific subset, if such a recommendation should ever be made. It is, however, important to know of the various possibilities, and to be familiar with the properties of the parameters.

Spectral Estimation

Integrals over an infinite time period appear in the mathematical definition of the power spectrum of a noise signal. For obvious reasons then, spectra can never be evaluated exactly; rather any spectrum obtained is an estimate of the mathematical concept. The use of finite stretches of raw EMG data introduces estimation errors. These errors are often summarized as the bias of the estimate, the systematic error, and the variance of the estimate, the statistical uncertainty. Methods employed to reduce these errors include windowing of the raw data, averaging of successive spectral estimates and smoothing of spectral estimates. In reporting results based on spectral analysis, it is important to state the estimation procedure and, if possible, give figures on the bias and the variance of the final estimate. It is important also to report questions pertinent to other sources of errors. In particular, one should give some indication on the noise level of the experimental setup, and state the sampling rate if digital methods are used. The possibility of excessive line interference is easily checked by visual inspection of the spectra obtained.

Part V: General Experimental and Kinesiological Information

One major drawback which prevents a full understanding, comparison or replication of any EMG research is inadequate detail of the protocol itself especially related to the anatomy, physiology and biomechanics of the neuromusculo-skeletal system under test.

Types of Contraction

1. Isometric—muscle has an average fixed length or joint is at a fixed angle (specify length or angle).
2. Isotonic—a contraction which produces an average constant force or, for in vivo contractions at an average constant moment (torque) (specify force (N) or moment (N·m)). Remember, that lifting or lowering a mass is not isotonic unless it is moving at a constant velocity.
3. Isokinetic—muscle is contracting at a constant linear velocity, or constant angular velocity (specify velocity (m/s) or angular velocity (rad/s)).
4. Concentric—muscle is shortening under tension.
5. Eccentric—muscle is lengthening under tension.

Associated Biomechanical Terms

1. Mechanical Energy—is the energy state of any limb segment or total body system at an instant in time. It is measured in joules (J).
2. Mechanical Power—is the rate of doing work or rate of change of energy at an instant of time. It is measured in watts (W).
3. Mechanical Work—is the time integral of the mechanical power over a specified period of time. It is also equal to the change in energy of a system (segment or total body) over that same period of time. It is measured in joules (J).
4. Positive Work—is the work done by concentrically contracting muscles. Thus the time integral of mechanical power over a specified time is positive, or the net change in energy of the system is also positive.
5. Negative Work—is the work done by an eccentrically contracting muscle. Thus the time integral of mechanical power over the specified time is negative, or the net change in energy of the system is negative.
6. Moment of Force (Torque)—Product of a force and lever arm distance about a centre of rotation (usually a joint centre). The unit is Newton-meters (N·m).
7. Impulse—is the time integral of a force or moment curve, and is usually employed in ballistic movements to reflect changes in momentum of the associated limbs. Linear impulse is quantified in N·s, angular impulses in N·m·s. The impulse is a prerequisite to calculation of average force or moment over a given period of time.

Electrode and Anatomical Details

The type and position of the electrodes must be reported. If indwelling electrodes are used additional information is necessary (needle, wire, unipolar, bipolar, depth of electrode, etc.). If there are problems of cross-talk the exact positioning of electrodes is necessary along with details of any precautionary tests that were done to ensure minimal cross-talk. If co-contractions can nullify or modify your results some evidence is necessary to demonstrate that the antagonistic activity was negligible.

It is now quite common to quantify an EMG amplitude as a percentage of maximum voluntary contraction (MVC). The details as to how these were elicited are important. Also, the position of the body, adjacent limbs, etc., need to be described.

If electrical stimulation is being done, additional electrode data are necessary: position of anode and cathode, surface area of contact or, in case of indwelling electrodes, details of the exposed conductive surface. The strength, duration and frequency of the stimulating pulses is mandatory, and remember that the strength is usually in current units (ma) rather than voltage, because the net depolarization is a function of current leaving the electrodes. Without a knowledge of skin/electrode impedance, the voltage information is not too meaningful. Thus with a constant voltage stimulation it is important to monitor and report the current pulse waveform.

General Subject Information

In any population study it is often relevant to give details of age, sex, height and weight of normals that may sometimes influence the results of certain experiments. Also, in conditions of fatigue or special training appropriate measures should be specified. For athletic or ergonomic tasks the researcher must give sufficient information to ensure that other centres could replicate his experiments. In the assessment of pathological movements certain clinical and medical history details of each patient may be necessary (i.e. level of lesion, number of months since stroke, type of prosthesis).

Finally, A Checklist of Common Terms

TERMINOLOGY	UNITS	COMMENTS/RECOMMENDATION
Amplifier Gain	ratio or dB	
Input Resistance or Impedance	ohms	10^{10} (resistance) on new dc equipment, 10^{8} (impedance) on new ac amplifiers at 100 Hz Min. 100 times skin impedance
Common Mode Rejection Ratio (CMRR)	ratio or dB	90 dB or better
Filter cut-off or Bandwidth	Hz	type and order of filter
EMG (raw signal)	mV	
EMG (average)	mV	specify averaging period
EMG (F.W. Rect.)	mV	
EMG (non-linear detector)	mV	specify non-linearity (i.e. square law)
EMG (linear envelope)	mV	cut-off frequency and type of low-pass filter
Integrated EMG (iEMG)	mV·s	specify integration period
Integrated EMG and Reset every T	mV·s or μV·s	specify T (ms)
Integrated EMG to Threshold and Reset	mV·s or μV·s	specify threshold (mV·s)
Power Spectral Density Function (PSDF)	$\mu V^2/Hz$	
Mean Spectral Frequency (MSF)	Hz	
Median Frequency	Hz	

References

Buchthal, F. and Schmalbruch, H. Motor unit of mammalian muscle. Physiol. Rev. 60:90-142, 1980.

DeLuca, C. J. Physiology and mathematics of myoelectric signals. IEEE Trans. Biomed. Engng. BME—26:313-325, 1979.

Inman, V. T. et al. Relation of human EMG to muscle tension. EEG and Clinical Neurophysiology. 4:187-194, 1952.

Kadefors, R. Myoelectric signal processing as an estimation problem. New Developments in EMG and Clinical Neurophysiology. Vol. 1, 519-532 (Karger, Basel, 1973).

Komi, P. V. and Viitasalo, J. H. T. Signal characteristics of EMG at different levels of muscle tension. Acta Physiol. Scand. 96:267-276, 1976.

Komi, P. V. Relationship between muscle, tension, EMG and velocity of contraction under concentric and eccentric work. New Developments in EMG and Clinical Neurophysiology. Vol. 1, 596-606 (Karger, Basel, 1973).

Silny, J., Hinsen, R., Rau, G., vonEssen, R., Merx, W. and Effert, S. Multiple-electrode measuring system for electrocardiographic evaluation of the course of acute myocardial infarction. Biomedizinische Technik 24:106-112, 1979.