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CHAPTER 9

New techniques in surface electromyography

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INTRODUCTION

Surface detection of myoelectric signals offers a number of advantages over needle detection. Electrodes can be applied without discomfort, medical supervision and risks of infection, repeatability of measurements is higher (Jonsson and Komi, 1973) and long term monitoring and pediatric applications are far easier than with needles. Traditionally, surface detection has been preferred to needle detection for obtaining 'global' information about the time and/or intensity of superficial muscle activation such as in gait analysis, prosthetic control or biofeedback therapy. A number of recent reports have shown that the surface potentials generated by either voluntary or electrically elicited muscle activity contain much more information than is commonly detected by conventional non-invasive recording techniques.

Myoelectric activity detected with surface electrodes above a given muscle may be considered as a summation of tissue-filtered signals generated by a number of concurrently active motor units. Fig. 1 shows a simple model of surface myoelectric signal generation during voluntary and during electrically elicited contractions.

The application of signal processing techniques has recently extended the quantity and quality of information that can be extracted from surface detected signals. Frequency domain parameters have been found suitable to provide information about ongoing changes of muscle condition during sustained contractions (fatigue). Multipe electrode systems allow non-invasive estimation of average muscle fiber conduction velocity as well as extraction of individual motor unit firing rates. The growing number of non-invasive methods is providing greater insight into muscle physiology and pathology. Non-invasive muscle characterization appears to be possible in the near future. Some methods have been implemented on dedicated microprocessors that perform acquisition of the signal, on line computation and display of the parameters of interest (Fig. 2*A*). Other methods are mostly implemented via software and operate on personal computers or on minicomputers properly equipped with data acquisition circuitry and computer controlled stimulators (Fig. 2*B*). Some of these new techniques deserve particular attention while others must be applied with caution. Most are ready for clinical application. An overview of such techniques will be provided in this chapter.

SURFACE SIGNAL DETECTION AND AMPLIFICATION TECHNIQUES

In the last few years a number of 'active' electrodes have been developed and marketed. These devices include at least two detection surfaces and an impendance buffer or amplifier mounted directly on the electrodes, within the device itself. With respect to traditional electrodes they allow a better quality of detection with lower noise and movement artefacts, constant geometry, better repeatability of positioning, and lower power line interference.

Examples of such active probes are reported by Johnson et al. (1977), De Luca et al. (1979), Van der Locht and Van der Straaten (1980), Accornero (1981), Basmajian and De Luca (1985) and Hageman et al. (1985). A widely used configuration consists of two parallel bars 1 mm in diameter, 10 mm long and 10 mm apart. Bipolar concentric electrodes have been tested by Van Steenwijk (1986) who suggested that they would provide a higher local resolution than the standard bipolar linear type.

The effect of electrode geometry on spectral parameters has been theoretically approached by Lindström et al. (1970, 1977) and discussed by Zipp (1978) and by Lynn et al. (1978). The model developed by Lynn indicates that the precise shape of the muscle fiber membrane potential distribution and the electrode size have only minor effects on the surface detected signal while fiber depth has a high effect on signal amplitude and frequency content. Interelectrode spacing affects spectral shape. The results reported by Zipp (1978) indicate

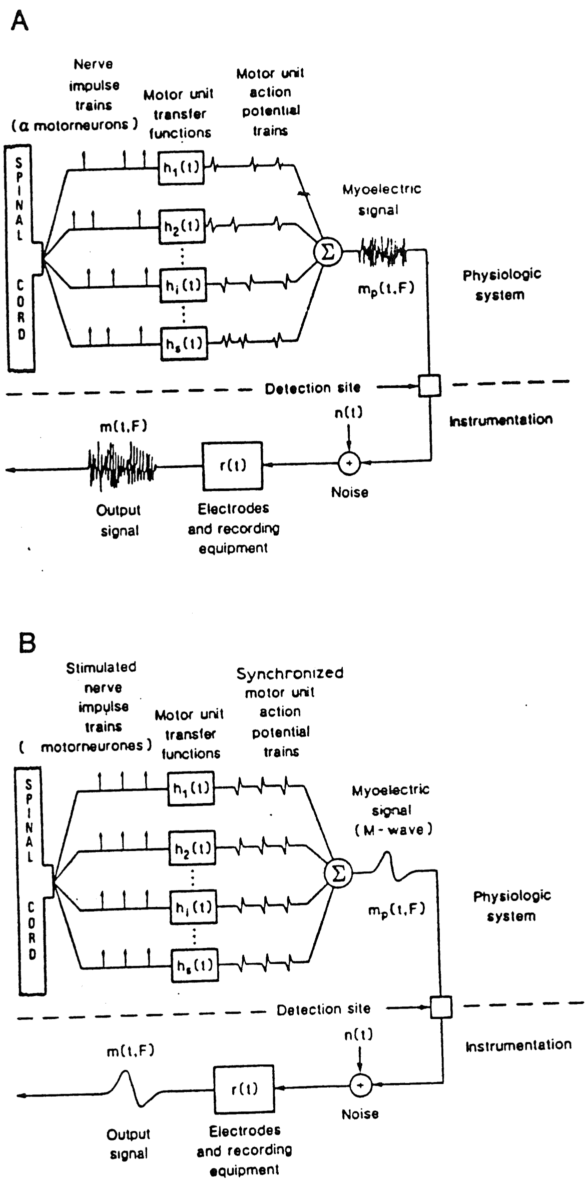

Fig. 1. Mathematical model for the generation of myoelectric signals. *A*: model for generation of myoelectric signal during voluntary contractions. The motor axons fire at different rates. Each motor unit transfer function relates the surface potential contributed by that motor unit to the motoneuron pulse. The asynchronous contributions generate the interference pattern. *B*: model for generation of myoelectric signal during electrically elicited contractions. The motor axons are synchronously triggered by electrical stimuli. The synchronous contributions generate the M wave. The surface signal is assumed to be detected with one couple of electrodes and it is function of contraction level and of time. See also Fig. 5.

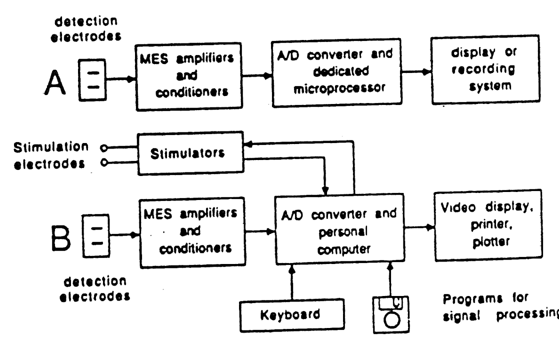

Fig. 2. Block diagrams of instrumentation systems for myoelectric signal processing. *A*: block diagram of a microcomputer based dedicated system. Such systems are developed for monitoring a specific parameter in either clinical or research environment and are relatively unflexible and not user programmable. Processing is usually performed on-line. *B*: block diagram of a personal or minicomputer based system. Such systems are more versatile and suitable for research environments. Appropriate programs may be developed for computing the desired parameters. Signals are sampled and stored in memory. Processing is usually performed off line.

(as predicted by theory) narrower spectra for greater interelectrode distances.

The clinician should be aware that the appraisal of clinical data published on myoelectric signal spectral density or on spectral parameters requires the knowledge of interelectrode spacing. The need for standardization of interelectrode distance for clinical applications is evident.

Electrode arrays offer a number of advantages with respect to the simple bipolar technique. Two sets of bipolar electrodes next to each other allow an estimate of average action potential propagation delay and therefore of muscle fiber conduction velocity. However, as shown by Broman et al. (1985a), this technique may lead to falsely high values of conduction velocity due to the simultaneous presence of differential signals on the two electrode sets probably due to crosstalk and to the effect of tissue anisotropy and non-homogeneity. A simple spatial filtering approach greatly reduces the problem leading to more reliable estimates (Figs. 3 and 4).

More complex spatial mapping and/or filtering techniques have been proposed by other authors. For example, Monster et al. (1980) used a 50 surface electrode array and one triggering indwelling electrode to map the spatial distribution of a motor unit potential over the tibialis anterior surface. More recently, Reucher et al. (1987a,b) used the weighted summation of surface signals to localize end plate regions, to identify single motor unit potentials and firing rates and to measure muscle fiber conduction velocity. These techniques appear to be very promising for clinical applications.

The amplifiers used for surface myoelectric signal conditioning must have the following general specifications:

(a) sufficient gain to produce an output of the order of ± 1 V;
(b) input impedance of the order of 100 – 1000 MΩ;
(c) input bias and offset currents less than 50 pA;

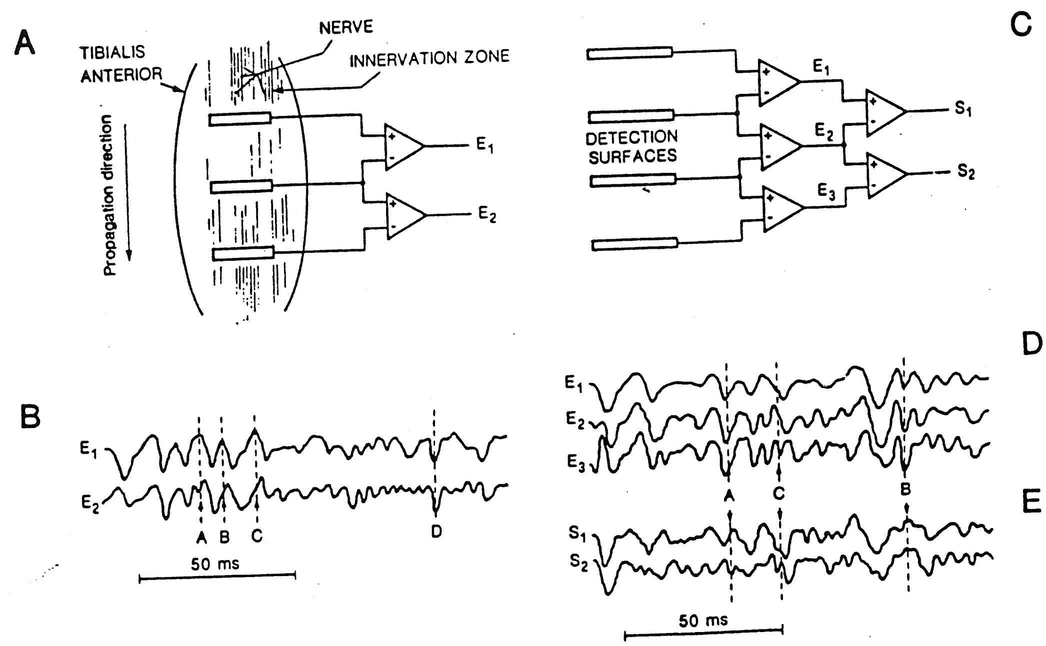

Fig. 3. Demonstration of delayed and non-delayed surface myoelectric signals obtained with two differential channels. (Redrawn from Broman et al., 1985a.) *A*: detection with two differential amplifiers. *B*: examples of delayed signals (A, B, C) and example of non-delayed signals (D). *C*: double differential detection system. *D*: outputs from the first stages showing delayed and non-delayed signals. *E*: outputs from the second stages showing cancellation of non-delayed signals.

(d) input equivalent noise less than 2 μV rms;
(e) bandwidth of 5 – 500 Hz (3 dB points);
(f) optical isolation between input and output is a preferred feature and it is mandatory if the output is to be connected to data acquisition and computer systems that do not satisfy the safety requirements for medical equipment.

Although only a few muscles have spectral components of the surface myoelectric signal beyond 250 Hz, a bandwidth of at least 500 Hz is required if accurate waveform reproduction is desired. It is important to realize that at half the cutoff frequency a single-pole low pass filter still applies a phase rotation of 26° to the input signal, thus modifying the output waveform by shifting its frequency components.

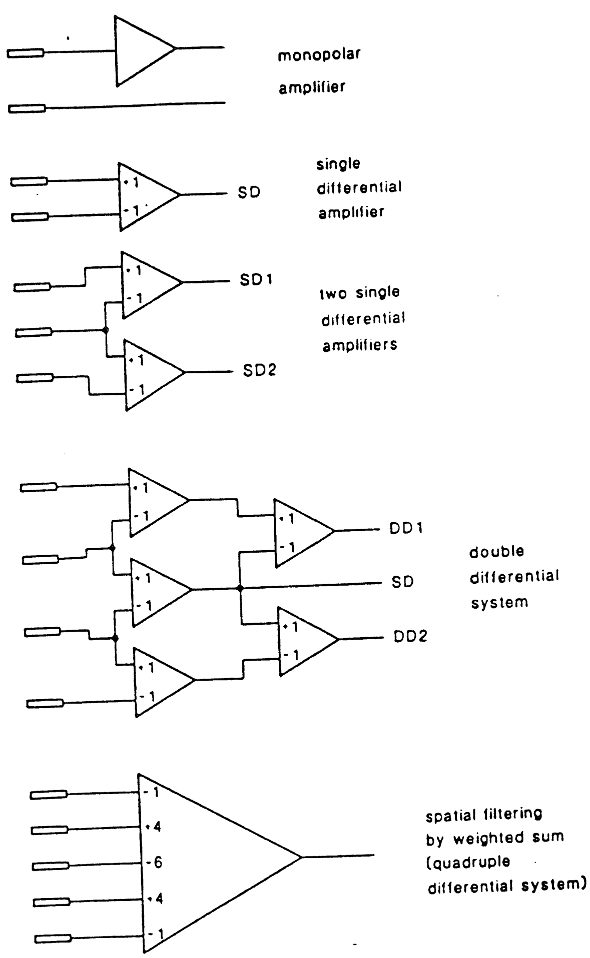

Fig. 4. Examples of surface myoelectric signals detection techniques.

SURFACE SIGNAL PROCESSING TECHNIQUES

A number of time domain and/or frequency domain parameters of either the voluntary or of the electrically elicited surface myoelectric signal carry information useful to estimate muscle properties and to quantify muscle performance. The main issues investigated by means of surface myoelectric signal processing are the following:

(a) relationship between myoelectric signal amplitude and force during isometric or dynamic contractions;
(b) relationship between spectral parameters, muscle fiber conduction velocity, tissue filtering function and muscle fatigue;
(c) diagnostic classification based on signal patterns or parameters;
(d) non-invasive fiber typing and muscle characterization.

The state of the art of such applications will be analyzed in the following pages.

Myoelectric signal to force relationship

The issue of myoelectric signal to force relationship during isometric contractions has been investigated by a number of authors whose work is reviewed in a chapter of 'muscles alive' (Basmajian and De Luca, 1985). Parameters which have been used as force indicators include the root mean square value, the low pass filtered or the integrated rectified signal with either linear or non-linear rectification function.

Although the results are affected by different detection and processing procedures and by lack of normalization, small muscles (first dorsal interosseous, adductor pollicis, etc.) have a quasi linear relationship between signal amplitude and force while larger muscles (biceps and triceps brachii, deltoid, soleus, etc.) present a non-linear relationship with the myoelectric signal increasing faster than force. Considerable differences in fiber type constituency do not significantly alter the relationship of the normalized signal amplitude to force.

Some points need further investigation before force estimation by surface myoelectric signal can receive widespread clinical application. For example, agonist-antagonist coactivation and crosstalk may contaminate the measurements of force output and of myoelectric signal amplitude. The relevance of these factors has not yet been discussed in the existing literature and deserves more attention.

The issue of myoelectric signal to force relationship during dynamic contractions is even more open to discussion. In this case the information of interest concerns the 'modulating' signal, presumably related to force, which defines the amplitude-time pattern of the myoelectric signal as indicated in the following equation.

$$E(t) = [W(t)]^a n(t)$$

where: $E(t)$ = surface myoelectric signal; $W(t)$ = modulating signal related to the degree of muscle activation; a = exponent indicating a non-linear relationship between $E(t)$ and $W(t)$ (usually it is assumed $1 \leq a \leq 2$); $n(t)$ = stochastic non-stationary process (noise) with gaussian distribution, unit variance and zero mean and whose spectrum is affected by tissue filtering function, fatigue and force level.

Traditional analog 'processors' such as the rectifier followed by a low pass filter, the periodically reset integrator, etc., do not provide satisfactory estimates of $W(t)$.

Recent work in this field has been focused on adaptive non-causal digital algorithms (not suitable for on-line processing) as reported by Shwedyk et al. (1977), Hogan and Mann (1980), D'Alessio (1984, 1985) and by Filligoi and Mandarini (1984). The effect of crosstalk, electrode position and interelectrode distance upon the estimate of $W(t)$ still needs investigation. Clinical applications of these estimation techniques are promising but not immediate.

Spectral parameters and conduction velocity

The relationship between surface myoelectric spectral parameters and muscle fatigue has been investigated by many authors. As shown theoretically by Lindström et al. (1970, 1977) and by Stulen and De Luca (1981), and experimentally by Bigland-Ritchie (1981), Naeje and Zorn (1982), Kereshi et al. (1983), Sadojama et al. (1983, 1985), Arendt-Nilsen (1984, 1985), De Luca (1984), Broman et al. (1985b), Eberstein and Beattie (1985) and many others, the mean and median frequencies of the surface power spectrum are related to muscle fiber conduction velocity.

Such relationship is obvious from Fig. 5. If the depolarized zone will propagate at higher speed, the signal detected by the electrodes and available from the amplifier will have more rapid changes, higher frequency content and its spectrum will exhibit a higher mean and median frequency. However, spectral parameters are markedly affected by quantities that have less influence on conduction velocity estimates, such as the depth of the fibers, the length and the spatial spreading of the depolarized zones. On the other hand, conduction velocity estimates are more critical than spectral parameter estimates, require accurate positioning of multiple electrode systems and may only be obtained in relatively long muscles. An overview of the problems and of the boundary conditions related to muscle fiber conduction velocity estimates is reported in

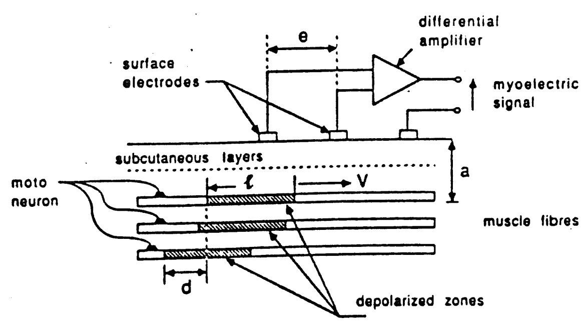

Fig. 5. Physical model for generation of surface myoelectric signal. The motor unit action potential detected between the electrodes is affected by the propagation velocity V of the depolarized zones, by the number of depolarized zones (number of fibers of the motor unit), by their length (l), by their spatial spreading (d), by the depth of each fiber (a), and by the interelectrode distance (e).

two papers by Sollie et al. (1985a,b). They showed that an electrode misalignment of 15° with respect to muscle fiber direction generates a 6.7% error in conduction velocity estimates. Roy et al. (1986) have shown that the position of the electrodes with respect to the innervation zone and to the tendon is a crucial factor for meaningful measurements of conduction velocity. For additional details refer to chapter 16 in this volume.

For most short or pennated muscles, conduction velocity measurements are not possible because a sufficiently long zone of unidirectional propagation of action potentials cannot be identified. In such cases mean or median frequency may be preferred as indicators of fatigue.

Intra-experiment, inter-experiment and inter-subject repeatability of initial mean and median frequency readings at four contraction levels have been assessed by Merletti et al. (1985) who found slightly better repeatability for mean frequency with standard error values of 4.8% for intra-experiment, 6.7% for inter-experiment and 16% for inter-subject repeatability at maximal voluntary contraction levels of tibialis anterior muscles. Hary et al. (1982) and Balestra et al. (1988) also found the mean (or centroid) frequency estimate to be less variable than the median estimate. The higher standard deviation of the median frequency estimaty is likely due to the greater variability of the lower end of the spectrum which is more heavily weighted in the calculation of the median frequency. This variability can be reduced by high-pass filtering of the myoelectric signal or by low pass filtering of the median frequency estimates with some loss of time resolution. The median frequency estimate is less affected by noise (Stulen and De Luca, 1981) and more sensitive to fatigue; thus, on these grounds it should be preferred to the mean.

Clinically usable on-line monitors of 'muscle fatigue' that provide plots of median frequency as function of time are available as prototypes and have been described in the literature (Stulen and De Luca, 1982; Gilmore and De Luca, 1985; Merletti et al., 1985).

Muscle fiber conduction velocity may be estimated as $V = e/t$ where V is the conduction velocity, e is the inter-electrode distance and t is the delay between two myoelectric signals detected with two differential or with a double differential system (Figs. 3 and 4). Estimation of such delay may be performed using different algorithms. Zero crossing and cross-correlation techniques have been used by many authors (Lynn, 1979; Nishizono et al., 1979; Masuda et al., 1982; Naeje and Zorn, 1982, 1983; Sadoyama et al., 1983, 1985; Zorn and Naeje, 1983; Broman et al., 1985b). A more efficient and accurate algorithm can be derived from the alignment technique proposed by McGill and Dorfman (1984). Very recently a new technique has been proposed by Hunter et al. (1987) which provides information about the distribution of conduction velocities within the volume of detection of the electrodes.

The available techniques for monitoring of spectral parameters and/or muscle fiber conduction velocity appear to be sufficiently established and ready for wider field testing and clinical applications.

Clinical and diagnostic applications

The availability of new signal processing techniques and prototype devices makes it possible to use surface myoelectric spectral parameters and conduction velocity in the clinical environment for diagnostic classification and for monitoring of therapy effectiveness. Surface electromyography, where properly applied, can replace some traditional needle techniques in clinical applications. Inbar and Noujaim (1984) compared classifications of normal/abnormal cases obtained with needle and with surface techniques based on spectral analysis. In seven out of eight cases the two techniques led to the same classification. Troni et al. (1983), using needle techniques, found reduced muscle fiber conduction velocity in the interictal period in subjects affected by hypokalemic periodic paralysis; similar observations can now be performed with surface techniques.

A 30 electrode array placed on the biceps brachii was used by Hilfiker and Meyer (1984) to quantify altered propagation behavior in five muscular dystrophy patients and to detect reduced conduction velocity and pathological spreading of end plates in five Duchenne dystrophy subjects. The multiple electrode technique proposed by Masuda et al. (1985a,b) and by Reucher et al. (1987a,b) allow the location and mapping of the innervation zones, the identification of motor unit firings and the estimation of motor unit territory and action potential shape. The clinical relevance of this information cannot be underestimated.

The duration of the motor unit potential is commonly used by clinicians to discriminate between different types of neuromuscular pathologies. Such parameter is commonly evaluated from needle detected myoelectric signal. If a dominant shape exists for the motor unit action potentials, and the power spectrum shows dips related to MUAP duration, the estimate of such parameter may be based on surface signals, as shown by Maranzana (1978).

Muro et al. (1982) observed an increase of mean frequency with increasing force in the biceps brachii of five healthy subjects and five myogenic subjects while a decrease of mean frequency with increasing force was observed in five neurogenic subjects.

Although diagnostic applications of surface electromyography are still limited, the potential for its use is rapidly growing both in the neurological field as well as in physical, occupational and sports medicine. The reader is referred to a review by De Luca (1984) and to chapter 16 in this volume for additional information concerning the application of spectral parameters. The section of this chapter on electromyography during electrical stimulation indicates further specialized clinical applications.

Muscle characterization

Some interesting applications of surface myoelectric signals concern the investigation of muscle architecture and the non-invasive estimation of fiber type distribution. Rosenthal et al. (1981) observed different rates of median frequency time decay during sustained isometric contractions of human soleus, medial gastrocnemius, vastus lateralis, vastus medialis, first dorsal interosseous. Preliminary results indicate a correlation between the time constant of the decay and the percentage of type II fibers reported in the literature for those muscles. Similar results are reported by Sato (1982). The possibility of non-invasive estimation of fiber type distribution appears very promising. In fact De Luca et al. (1983) reported that during sustained contractions the median frequency decreases in proportion to the ratio of type II to type I fibers. Further insight into muscle architecture may be achieved by comparing data obtained during voluntary and electrically elicited contractions as described in the following section.

SURFACE ELECTROMYOGRAPHY IN ELECTRICALLY ELICITED CONTRACTIONS

Electrical stimulation of a nerve trunk or of the motoneurons terminal branches (over a muscle motor point) provides an alternative way to elicit muscle contractions. Comparison between myoelectric signal parameters obtained during voluntary and electrically elicited contractions provides useful information about the neuromuscular system. Cioni et al. (1985) demonstrated that the RMS value of the compound potential (M wave) elicited in the human tibialis anterior by supramaximal stimulation of the peroneal nerve is three times larger than the RMS of the myoelectric signal detected during a maximal voluntary contraction. This finding indicates that, in the interference pattern, algebraic summation of the MUAPs accounts for a loss of about 2/3 of the RMS. These authors also found that this ratio increases to about 4.5 and to about 7.5 respectively in the non-affected and in the affected limb of hemiparetic subjects due to a decrease of the voluntary signal.

Electrical stimulation of the nerve fibers innervating a muscle provides for the execution of interesting experimental paradigms in which muscle properties may be investigated independently of central drive mechanisms. In particular, the effect of motor unit synchronization and of firing frequency may be studied and correlated to muscle performance and to central and/or peripheral fatigue mechanisms. The issue is relevant from the standpoint of clinical research as well as from the standpoint of optimization of neuromuscular stimulation techniques (such as for paraplegic ambulation) where the issue of fatigue reduction is very important.

Selective muscle stimulation may be used to investigate signal crosstalk among nearby muscles (De Luca and Merletti, 1988). Other interesting possibilities are offered by the reversed order of motor unit recruitment observed during voluntary contractions and during contractions electrically evoked with electrodes implanted on a nerve (Mortimer, 1981; Gorman and Mortimer, 1983). If this were the case also with surface stimulation it should be possible to describe the electric parameters of the extremes of the motor unit population by analyzing the myoelectric signal generated by low level voluntary contractions and the signal generated by low level stimulation. In fact, small motor units (small fibers with lower conduction velocity) would be selectively activated at low level voluntary contractions and lead to low conduction velocity readings while large units (large fibers with higher conduction velocity) would be selectively recruited by low level stimulation and lead to high values of conduction velocity readings. Results presented by Merletti et al. (1988b) show that indeed this pattern can be observed in some experiments but in most cases the motor unit recruitment order (as indicated by the initial value of muscle fiber conduction velocity) is the same as in voluntary contractions. Geometric factors related to the electric field distribution in the tissue and to the location of the nerve fibers play an important role in determining the recruitment order during surface stimulation. Muscle architecture may be further investigated with this technique.

Kranz et al. (1983) applied supramaximal stimuli to the median nerve at the wrist during maximal voluntary contractions of the thenar muscles in ten normal subjects. It was found that the power spectra of the voluntary myoelectric signal was similar to that of the averaged electrically elicited response (Fig. 6) both at the beginning and at the end of a maximal voluntary contraction sustained for 45 s. Median frequency increased with force and decreased with time. These authors conclude that the spectral changes observed during sustained voluntary contractions as well as with maximal stimuli reflect slowing of muscle action potential conduction velocity. Similar results had already been obtained by Mills in 1982. Intermittent maximal voluntary contractions presented myoelectric signal spectral parameters and spectral changes very similar to those of a maximal electrically elicited response. More recently Merletti and De Luca (unpublished data) found the spectra of voluntary myoelectric signal recorded at the beginning and at

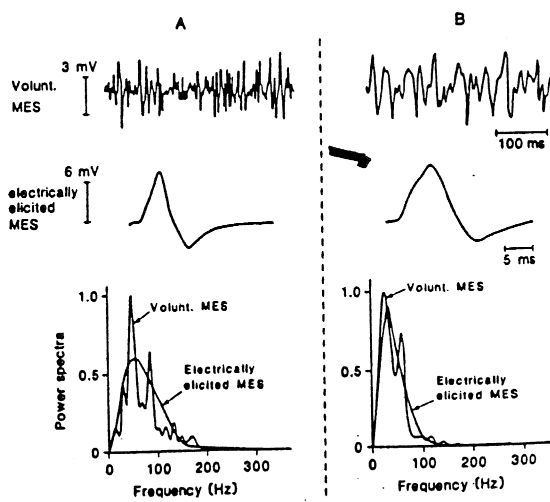

Fig. 6. Time course and power spectra of voluntary and electrically elicited myoelectric signals detected on the thenar muscles at the beginning (*A*) and at the end (*B*) of a 45 s maximal voluntary contraction. Stimulation delivered at the median nerve at the wrist. (Redrawn from Kranz et al., 1983.) The slowing of the voluntary signal and of the M wave is reflected by the compression of the spectra toward the lower frequencies. The spectra of the voluntary and of the averaged electrically elicited signal are similar and show the same compression.

the end of a 25 s 80% MVC contraction of tibialis anterior to be similar to those obtained during a tetanic maximal contraction with 20 Hz stimulation (see Fig. 7).

During electrically elicited contractions the myoelectric signal is a periodic sequence of M waves (see Fig. 1*B*). Such sequence can be studied by (a) analyzing a signal epoch of a few seconds or, (b) averaging *N* responses and analyzing the averaged response. In the first case the signal is periodic and, therefore, its power spectrum is a sequence of 'lines' representing the power contribution of each harmonic. In the second case the signal is not periodic; the spectrum is continuous

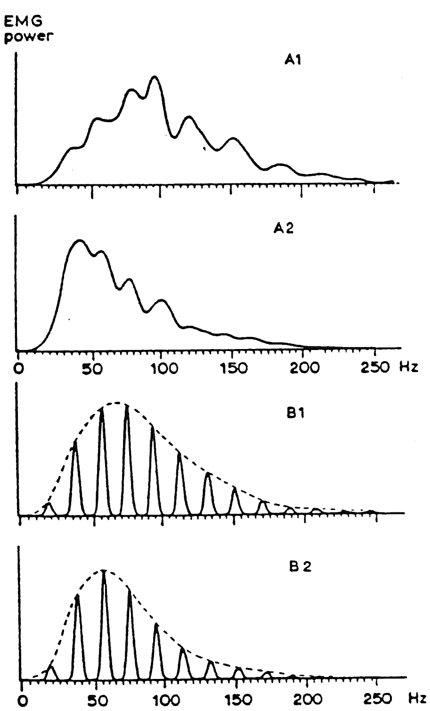

Fig. 7. Power density spectra of myoelectric signal from the tibialis anterior muscle. *A1* and *A2*: power spectra of surface myoelectric signals computed on 3 s epochs at the beginning and after 25 s of a 80% MVC. *B1* and *B2*: power spectra of surface myoelectric signals computed on 3 s epochs at the beginning and after 25 s of a contraction elicited with supramaximal stimuli at 20 Hz (0.2 ms duration). The electrically elicited M waves were not averaged thus resulting in a line spectrum which shows the effect of full synchronization of the motor units.

and approximately coincides with the envelope of the spectral lines of the first case.

Fig. 7 shows the spectra of voluntary (*A1* and *A2*) and electrically elicited (*B1* and *B2*) myoelectric signals detected on human tibialis anterior. The electrically evoked potentials were not averaged in order to generate a 'line spectrum' with lines spaced 20 Hz apart and representing the harmonics of the periodic M wave. A periodic M wave, and therefore a line spectrum, would describe a situation of full synchronization of all the recruited motor units. It is interesting to observe that such condition does not necessarily imply a spectral shift and a change of spectral parameters as suggested in numerous reports. Our work in progress is revealing that motor unit firing frequency and full synchronization have little influence on the shape of the surface signal power spectrum which is mostly affected by the shape of the motor unit action potentials as theoretically shown by De Luca (1979, 1984) and by other authors.

Results recently reported by Merletti et al. (1988a) also show that the estimates of spectral parameters and conduction velocity based on the averaged myoelectric response to stimulation have much smaller fluctuations that those obtained during voluntary contractions and therefore provide information of much better quality for muscle investigation.

Moritani et al. (1985a,b) studied the responses evoked in human soleus and gastrocnemius by electrical stimulation of the posterior tibial nerve. They found an amplitude decline and a waveform widening which progressed with time and increased with stimulation frequency. Such decline and widening were more marked in the gastrocnemius suggesting a correlation with fiber type dominance and indicating a possibility of non-invasive muscle fiber typing.

The study of electrically evoked myoelectric signals and muscle force allows the investigation of muscle fatigue mechanisms. From such studies Bigland-Ritchie et al. (1979) concluded that at stimulation frequencies above 50 Hz force decrease is due to failure of electrical propagation, an event avoided during voluntary contractions by progressive reduction of motor unit firing rate and by increased twitch duration. These authors were able to match the time patterns of force and of rectified myoelectric signal obtained during maximal voluntary contraction to those obtained during an electrically elicited contraction. The match was achieved by appropriately continuously decreasing the stimulation frequency. Such result may indicate the feasibility of a closed loop system that would minimize fatigue of electrically stimulated muscles by monitoring surface myoelectric signals while continuously adjusting stimulation parameters.

The detection and processing of electrically evoked signals require the use of appropriate techniques to remove the stimulation artifact from the detected signal. An artifact suppression system has been proposed by Knaflitz and Merletti (1988). The system consists of a signal conditioner and of a stimulator. The signal conditioner is an optically isolated amplifier with slow rate limiting and time windowing (blanking) of the residual artifact. The stimulator operates as a current generator during pulses and as a voltage generator between pulses. The latter mode assures an active short circuit between electrodes, a rapid discharge of the electrode capacitances and a short after stimulus transient. The residual artefact is limited to about 25 μV (peak-to-peak voltage referred to the input).

SUMMARY

Although clinical applications of surface myoelectric signals are still mostly limited to biofeedback and gait analysis, a much wider spectrum of applications is ready to be transferred from the research environment to the clinical field. Such applications are extensively based on sophisticated processing of the surface myoelectric signal made possible either by dedicated microprocessors or by specific software developed for personal computers. Most of the applications described require a working knowledge of their limits, a respectful use of instrumentation, and an awareness of signal generation, signal propagation and signal processing.

The potential clinical applications of the available techniques and the development of new methods provide a stimulating challenge for bioengineers and clinicians in the forthcoming years.

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