Myoelectric manifestations of fatigue in voluntary and electrically elicited contractions

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MERLETTI, ROBERTO, MARCO KNAFLITZ, AND CARLO J. DE LUCA. *Myoelectric manifestations of fatigue in voluntary and electrically elicited contractions.* J. Appl. Physiol. 69(5): 1810–1820, 1990.—The time course of muscle fiber conduction velocity and surface myoelectric signal spectral (mean and median frequency of the power spectrum) and amplitude (average rectified and root-mean-square value) parameters was studied in 20 experiments on the tibialis anterior muscle of 10 healthy human subjects during sustained isometric voluntary or electrically elicited contractions. Voluntary contractions at 20% maximal voluntary contraction (MVC) and at 80% MVC with duration of 20 s were performed at the beginning of each experiment. Tetanic electrical stimulation was then applied to the main muscle motor point for 20 s with surface electrodes at five stimulation frequencies (20, 25, 30, 35, and 40 Hz). All subjects showed myoelectric manifestations of muscle fatigue consisting of negative trends of spectral variables and conduction velocity and positive trends of amplitude variables. The main findings of this work are *1)* myoelectric signal variables obtained from electrically elicited contractions show fluctuations smaller than those observed in voluntary contractions, *2)* spectral variables are more sensitive to fatigue than conduction velocity and the average rectified value is more sensitive to fatigue than the root-mean-square value, *3)* conduction velocity is not the only physiological factor affecting spectral variables, and *4)* contractions elicited at supramaximal stimulation and frequencies >30 Hz demonstrate myoelectric manifestations of muscle fatigue greater than those observed at 80% MVC sustained for the same time.

human muscle; tibialis anterior; myoelectric signal; electrical stimulation; electromyography; conduction velocity

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IT IS WELL KNOWN that the power spectral density of the myoelectric signal undergoes frequency compression during sustained muscle contractions long before the muscle becomes unable to sustain a desired force. Such changes are referred to as myoelectric manifestations of localized muscular fatigue (4, 8) and are attracting increasing interest because of their potential applications for monitoring muscle condition during functional electrical stimulation of muscles (32) as well as for noninvasive muscle analysis and assessment (31, 33).

Muscle fiber conduction velocity (CV) is defined as the propagation velocity of the depolarization along muscle fibers and is known to decrease during a strong sustained contraction. CV is a basic physiological parameter that affects the myoelectric signal spectral density contributing to its compression during the fatigue process. Two spectral indexes that have been studied intensively and found to be appropriate indicators of spectral compression are the mean (MNF) and the median (MDF) frequency. They will be jointly referred to as spectral variables. Their correlation to CV and to muscle fiber type distribution has been reported by many authors (1, 3–5, 7, 8, 10, 18, 26, 30, 34). The myoelectric signal amplitude is also known to change during sustained contractions, reflecting the fatigue process. The average rectified value (ARV) and the root-mean-square value (RMS) are respectively related to the area under the rectified signal and to the mean power of the signal within a specified time window (see APPENDIX). They are commonly used to describe amplitude variations and will be jointly referred to as amplitude variables.

Electrical stimulation provides interesting experimental paradigms to study muscular properties and fatigue because it gives the experimenter control of motor unit firing frequency and recruitment and, if selectively applied, eliminates the problem of cross talk from nearby muscles. The issue of fatigue during electrically elicited contractions is of paramount importance in functional electrical stimulation techniques for external control of paralyzed extremities. The relevance of noninvasive techniques capable of monitoring muscle fatigue during functional electrical stimulation is obvious, especially in closed-loop system applications (32). Furthermore, electrical stimulation may be expected to allow *1)* degrees of muscle fatigue greater and more repeatable than during voluntary contractions and *2)* standardization of muscle testing procedures in ways that would be independent of the subject's ability or willingness to perform voluntary efforts. On the other hand electrical stimulation may be limited in its applications by discomfort or pain and by the unpredictability of motor unit recruitment order (21).

This study was undertaken with the objective to investigate the myoelectric manifestations of localized muscular fatigue during sustained electrically elicited isometric contractions and to compare them with those observed during voluntary contractions.

METHODS

The tibialis anterior muscle was selected for this study because of the relatively extensive body of data available on its structure and behavior (1, 2, 7, 16, 17, 19, 21, 28). This muscle is also particularly suitable for CV measurements because it contains a relatively long region between the motor point(s) and a lower tendon that is sufficient to accommodate the detection electrode and to ensure reliable estimates of CV (28).

Twenty experiments were performed on 10 volunteer subjects (9 males, 1 female) with no history of orthopedic or neurological disorders. Ages ranged from 18 to 41 yr, with a mean of 29 ± 6.9 (SD) yr. The dominant-side leg was always used. Two experiments were performed on each subject on different days. The experimental procedure has been described in greater detail previously (21) and will be only summarized here.

Stimulation Technique

A monopolar stimulation technique was chosen so as to improve selectivity and field uniformity, especially in the deeper part of the muscle. A negative rectangular sponge electrode (2 × 3 cm) was placed on the most proximal motor point of the muscle, and a larger (8 × 12 cm) positive sponge electrode was placed on the gastrocnemius muscle. The current lines would therefore traverse a roughly conical space across the leg alongside the tibia (Fig. 1). Both stimulation electrodes were damped with tap water.

Avoiding or removing the stimulation artifact affecting the myoelectric signal is a common problem in this type of experiment. Estimates of CV are detrimentally affected by stimulation artifacts simultaneously present on both the myoelectric signals used for the estimate. Estimates of spectral and amplitude variables are also affected. To avoid this problem, an artifact suppression technique was implemented. The complete technique has been described elsewhere (20). A schematic description of the system is provided in Fig. 1.

Pulse duration has been shown to affect the selectivity of stimulation (12). Narrow pulses allow a more gradual recruitment of nerve fibers. A pulse width of 0.1 ms was selected as a compromise between such requirement and the rise time and slew rate of the stimulator's output stage. To study the effect of frequency on MDF, MNF, and CV during fatigue, stimulation was applied at 20, 25, 30, 35, and 40 Hz. The lowest frequency was selected as the near minimum required for tetanic contractions; the highest was selected as the maximum that allowed clearly separated compound action potentials (M waves).

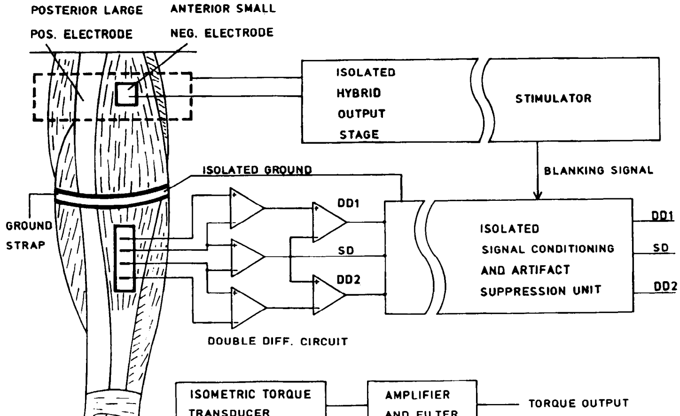

FIG. 1. Stimulation and detection system. Hardware is contained in a single device that provides constant current stimuli and artifact-free single- and double-differential myoelectric signals. System is described in detail by Knaflitz and Merletti (20). SD, single-differential myoelectric signal; DD1 and DD2, double-differential myoelectric signals.

Myoelectric Signal Detection Technique

The myoelectric signal was detected with the four-bar electrode technique described by Broman et al. (6, 7). The single differential output was obtained from the two central bars and was used to compute the myoelectric signal spectral and amplitude variables. The detection technique provided two double-differential outputs from which the average muscle fiber CV was estimated (see *Signal Processing*). A hybrid output stage, slew rate limiting, and time windowing (signal blanking) were used to eliminate the stimulation artifact (20). The stimulation and myoelectric signal detection circuits were fully isolated. The three myoelectric signals were low-pass filtered with a cutoff frequency of 480 Hz (120 dB/decade roll off), recorded on FM analog magnetic tape together with the force signal and a timing signal from the stimulator, and processed off-line.

Experimental Protocol

The experimental protocol consisted of a preparatory phase and an experimental phase. Each subject was seated comfortably in a dental chair with the ankle joint at ~90°. The foot was bound in an isometric brace equipped with a torque transducer. The motor points of the muscle were identified as those with the lowest stimulation threshold. The number of motor points ranged from one to five, with the largest and most sensitive usually located in the proximal half of the muscle. The stimulation electrode was moved over the motor point area until a location providing the highest muscle contraction with tolerable sensation was found.

The detection electrode was applied on previously shaved skin cleaned with alcohol. No conductive paste was necessary. The electrode was moved over the muscle in the area between the most distal motor point and the tendon. The four bars were perpendicular to the muscle fibers. The electrode was fixed with an elastic strap in the position that showed highly correlated double-differential signals during voluntary and stimulated test contractions (21). Such test contractions were only a few seconds long. To avoid any effect of the test stimulation on the subsequent trials, 3–4 min were allowed before the experimental phase was begun. A skin temperature sensor, with a resolution of 0.1°C, was fixed on the skin near the detection electrodes.

The experimental phase consisted of three maximal voluntary contractions (100% MVC) lasting 5–10 s and spaced 3 min apart. The largest of the three values was taken as the 100% MVC value. After a resting period of 5 min, a 20% MVC contraction followed by an 80% MVC contraction were performed. These voluntary contractions lasted 20 s, were spaced 3 min apart, and were performed with visual feedback wherein the subject was asked to match and maintain a force target level shown on an oscilloscope screen. The 20-s time duration was selected as the approximate maximal interval for which a contraction effort at 80% MVC could be sustained.

Stimulated contractions were then performed with the subject relaxed and physically passive. Such condition was indicated by the absence of voluntary myoelectric signals. Two stimulation levels were used. A supramaximal stimulation (~10–15% above the level generating the maximal M wave) was defined as the high-level stimulation. One high-level contraction lasting 15 s was performed to verify that neighboring and antagonist muscles were not activated. The stimulation frequency selected for this contraction was 20 Hz. The subjects had no feeling of either stimulation or muscle contraction under the large positive electrode, and no contraction was detected by palpation. No myoelectric activity was detected over the plantar flexors and peroneal muscles. A stimulation level eliciting a M wave with an average peak-to-peak amplitude of 25–30% of the maximal M wave was defined as the low-level stimulation. At low-level stimulation, the M-wave amplitude showed some pulse-to-pulse instability; attempts to stabilize it were unsuccessful. After a resting period lasting 10 min two 20-s contractions were performed at low-level stimulation at 3-min intervals at each stimulation frequency (20, 25, 30, 35, and 40 Hz). They were followed by two high-level stimulation contractions at the same stimulation frequencies and 5 min apart.

Signal Processing

Myoelectric and force signals played back from analog tape were sampled at 1,024 Hz for each myoelectric channel and at 85 Hz for the force channel. The samples were digitized by a 12-bit analog-to-digital converter and stored on the disk of a micro PDP 11/23 computer. Spectral variables and CV were then computed with numerical algorithms. Power spectra of voluntary contractions were calculated from the single differential myoelectric signal over two 0.5-s adjacent subepochs and averaged providing a frequency resolution of ±2 Hz. This procedure reduced the estimate standard deviation with respect to that resulting from choosing a 1-s epoch duration (24). Power spectra of stimulated contractions were calculated from the electrically elicited responses averaged over a 1-s signal epoch. Zero padding up to 1 s in the time domain was used to interpolate spectra obtaining lines 1 Hz apart. Spectral and amplitude variables were then computed. The time epoch was then shifted by 1 s, and the process was repeated so that 20 values for each variable were obtained over the 20-s contraction time.

CV was computed as $d/\Delta t$, where $d = 10$ mm was the interelectrode distance and Δt was the delay between the two double-differential signals. The delay was obtained by identifying the time shift required to minimize the mean square error between the two double-differential signal Fourier transforms using the method outlined by McGill and Dorfman (23).

It is known that spectral variables and CV are affected by intramuscular temperature. Edwards and Hill (11) found that the intramuscular temperature of the human abductor pollicis electrically stimulated at 30–50 Hz increased by 0.91°C in 1 min. A temperature change of ~0.3°C should therefore be expected in our 20-s contractions, leading to negligible errors of only ~1.2% in CV, MDF, and MNF estimates. Skin temperature variations of approximately ±1.5°C were observed during our experiments and attributed to ambient and limb temperature fluctuations. Therefore the values of MDF, MNF, and CV were corrected for skin temperature variations taking place during the experiment. The correction factor was 3.5°C obtained from our previous work (25).

Torque values were averaged over 1-s intervals and normalized with respect to the maximal voluntary torque. Spectral and amplitude variables, CV, and normalized torque were then tabulated and plotted vs. time for each contraction.

Curve Fitting and Statistical Processing

Figure 2 shows an example of the time course of the myoelectric signal variables obtained from one typical experiment. Curvilinear behavior with upward concavity was clearly evident in the most fatiguing contractions, whereas linear behavior was common at low-level, low-frequency, electrically elicited contractions and at 20% MVC contractions.

The data sets that showed curvilinear behavior appeared to be well fitted by a least-square exponential curve of the type $y = A \cdot e^{-Bt} + C$ with time constant $\tau = 1/B$ and y-axis intercept (initial value) given by $A + C$. Either A or τ $(1/B)$ could be used as measures of fatigue. However, it is intuitive that the muscle fatigue would be greater for greater A and greater B values, corresponding to greater and faster (smaller τ values) changes of the specific variable. Therefore a measure of fatigue was defined as the product $A \cdot B$ equal to A/τ, which is the absolute value of the initial slope of the regression exponential curve. The data sets showing linear behavior (for which the exponential regression algorithm was not suitable) were fitted with a least-square regression line whose slope was taken as measure of fatigue.

An alternative way of defining a measure of fatigue would be to consider the slope of a least-square regression line fitted to the first n seconds of each data set.

Although the first approach may overestimate the initial slope, the second may underestimate it (see RESULTS and DISCUSSION). Both methods were adopted in the analysis of the data sets and 5 s ($n = 5$) where chosen for the second method. Examples of the two methods are given in Fig. 3.

To allow comparison between rates of change of different variables the initial slope was normalized with respect to the initial value defined as the intercept of the regression curve or line with the y-axis. The resulting normalized initial slope, or initial rate of change, is therefore expressed in percent per second.

The one- or two-sample (paired when appropriate) Wilcoxon tests were used (because of their independence from the probability distribution function of the sample) to estimate the statistical significance of differences between parameters of the fitted curves (29).

RESULTS

The values of the myoelectric signal variables obtained during voluntary or electrically elicited contractions were

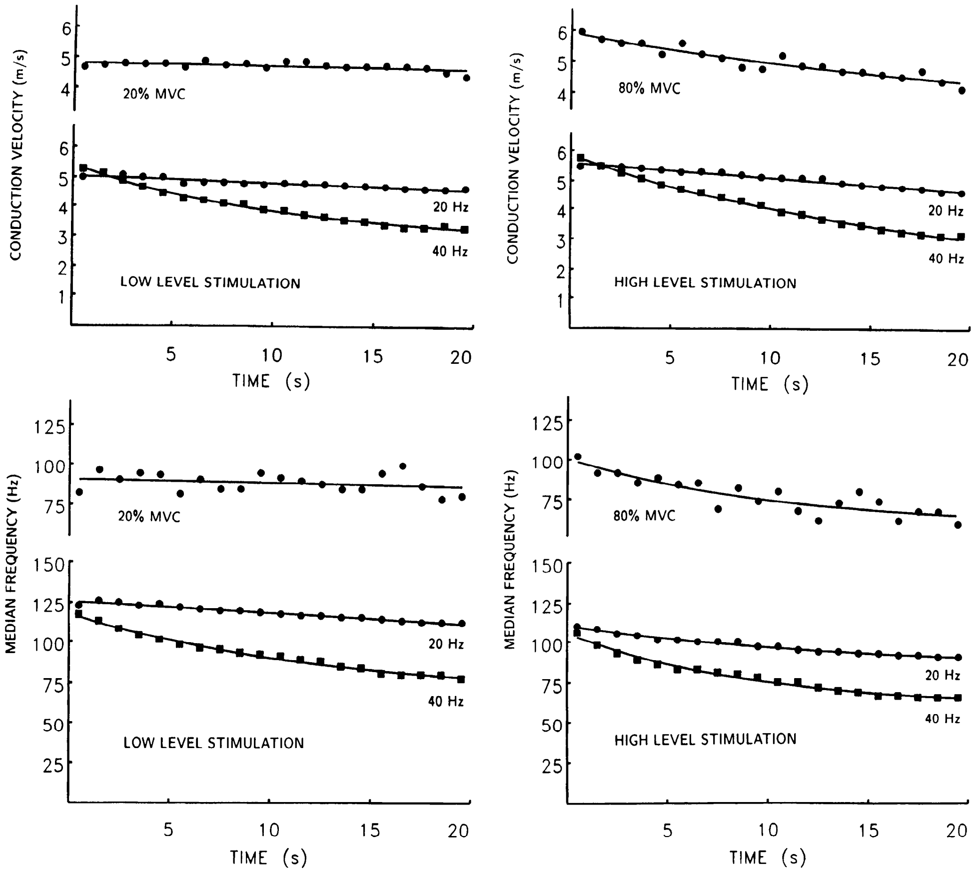

FIG. 2. Examples of plots of median frequency and conduction velocity vs. time during voluntary (20 and 80% MVC) and electrically elicited contractions at low and high stimulation levels and at 20 and 40 Hz. See text for details. Regression lines and regression exponential curves are shown. Similar plots are obtained for mean frequency. All contractions are from same subject.

similar and in agreement with previous findings reported by Kranz et al. (22). In all experiments, electrically elicited contractions showed relatively small fluctuations of the myoelectric signal variables about their regression curve or line. Greater fluctuations were observed during voluntary contractions, as shown in Figs. 2 and 3. All myoelectric signal variables showed a linear or moderately curvilinear behavior vs. time at low stimulation level and frequencies, whereas their behavior was more curvilinear at high stimulation level and frequencies. In electrically elicited contractions the exponential regression, when found to be appropriate to use, showed a correlation coefficient above 0.9 (range 0.90–0.99) and residual standard deviations of the order of a few percentage points of the initial value. During 80% MVC the exponential regression, when appropriate, showed lower correlation coefficients (range 0.75–0.90) and greater residual standard deviations than during electrically elicited contractions. Figures 3 and 4 show typical results.

Figure 4 shows the effect of the level and frequency of stimulation on MDF and CV. Greater fluctuations are evident at low-level stimulation due to the instability of the M wave. It is evident that changing the stimulation frequency from 20 to 40 Hz yields a much greater effect than changing the stimulation from low to high levels. High-level stimulation produces more fatigue than low-level stimulation probably because of the greater muscle portion activated and the greater intramuscular pressure resulting in greater ischemia. Figure 4 also shows different initial values of the myoelectric signal variables in different stimulation conditions. This issue has been discussed previously (21).

MDF and MNF showed a more curvilinear behavior, shorter time constants, and greater rate of change than CV. The time constant of MDF was 85% that of CV (mean of the ratios across all stimulated contractions admitting exponential regression) with a standard deviation of 37% and was lower than that of CV at $P \leq 0.01$ (paired Wilcoxon test on 70 pairs). Figure 5 shows cumulative results from the contractions that would admit exponential regression. The large standard deviations reflect intersubject and interexperiment variability. At high-level stimulation and at frequencies above 25 Hz (fatiguing contractions), the average estimate of the asymptotic value is between 60 and 67% of the initial value for CV, 57 and 61% of the initial value for MDF, and 60 and 62% of the initial value for MNF. The average ratio of the normalized asymptotic value of MDF with respect to that of CV is 93 ± 11%, and the difference between the MDF and CV normalized asymptotic values is significant at the $P \leq 0.05$ level (paired Wilcoxon test on 45 pairs). This observation suggests that, in highly fatiguing contractions, MDF and MNF would eventually decrease slightly more than CV with shorter time constants and greater initial slopes.

Figures 6 and 7 show the normalized initial slope (initial rate of change in %/s) of CV, MDF, and MNF averaged across subjects and computed with the two algorithms described in METHODS. All values are significantly different from zero, except at 20% MVC, if the 20-s (linear or exponential) regression is used. All values are significantly different from zero, except at 20% MVC, 20- and 25-Hz low-level stimulation, if the 5-s linear regression is used. The difference between the initial rate

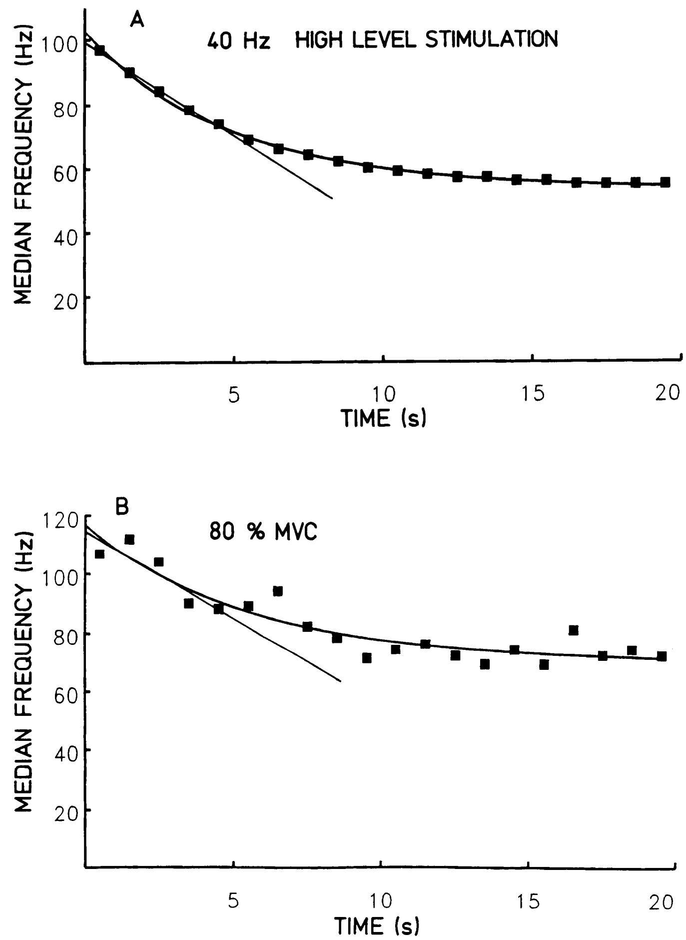

FIG. 3. Examples of time course of median frequency during an electrically elicited contraction (high-level stimulation, 40 Hz) and a voluntary contraction (80% MVC) showing curvilinear behavior. Values of parameters obtained from exponential regression and a linear regression over first 5 points are listed below. Initial values are affected little by regression algorithm chosen but absolute and normalized slopes are more affected. *A*: exponential regression: $y = 48.9e^{-t/4.8} + 53.7$, initial value = 102.6 Hz, initial slope = −10.13 Hz/s, norm initial slope = −9.85%/s, corr coeff = 99.9%; linear regression (5 s): $y = 99.1 - 5.8t$, initial value = 99.1 Hz, initial slope = −5.8 Hz/s, norm initial slope = −5.8%/s, corr coeff = 99.8%. *B*: exponential regression: $y = 47.1e^{-t/5.5} + 69.9$, initial value = 117.0 Hz, initial slope = −8.61 Hz/s, norm initial slope = −7.36%/s, corr coeff = 93.5%; linear regression (5 s): $y = 115.2 - 6.0t$, initial value = 115.2 Hz, initial slope = −6.0 Hz/s, norm initial slope = −5.2%/s, corr coeff = 89.2%.

of change of CV and that of either MDF or MNF is highly significant regardless of the definition of slope adopted ($P \leq 0.01$, paired Wilcoxon test), again indicating that spectral variables demonstrate an initial rate of change greater than that of CV. In general MDF shows an initial rate of change slightly higher than that of MNF. During 80% MVC contractions this difference is significant ($P \leq 0.05$, paired Wilcoxon test) regardless of the definition of slope adopted. This difference is significant in all other cases (except at 20% MVC, 20- and 40-Hz high-level stimulation) only if the linear regression over the first 5 s is adopted to define the slope. Therefore this method may be more sensitive to small initial slope differences.

The average initial rate of change of CV, MDF, or MNF, observed during 80% MVC, is near that observed at 30-Hz high-level stimulation regardless of the definition of slope adopted. Such a rate is higher than that observed at 25 Hz and lower than that observed at 35-Hz high-level stimulation. The estimated asymptotic value at 80% MVC is also slightly higher than that estimated at high-level stimulation and frequencies >25 Hz. These two observations indicate that a muscle stimulated at supramaximal levels and at frequencies >30 Hz shows, on average, myoelectric manifestations of fatigue greater than those observable during 80% MVC of the same duration.

Although CV, MDF, and MNF showed decrease during sustained stimulated contractions, amplitude variables (ARV, RMS) showed increase. Typical patterns for a voluntary and an electrically elicited contraction are reported in Fig. 8 in normalized form (normalization with respect to the intercept of the regression line or curve with the y-axis) to allow comparisons among different variables.

Each contraction was repeated twice in each experiment, and each experiment was repeated twice on each subject. Figure 9 shows an example of results from one subject. Intraexperiment repeatability of the data was usually within a few percentage points. Interexperiment variation in the same subject was often similar to intersubject variation. Two experiments per subject do not provide sufficient data for statistically significant conclusions; however, qualitative observation of the results suggests that the data are more repeatable for voluntary than for stimulated contractions.

Such intrasubject variability is due to the critical effects of stimulation and detection electrode location and to the possibly different muscle portions activated in different experiments. Better repeatability is observed for 80% MVC (Fig. 9). In such a case the whole muscle is activated and the main element of interexperiment variability is the location of the detection electrode. Better repeatability of electrically elicited contractions might be obtained by nerve trunk, rather than motor point, stimulation, although in such a case subject discomfort and possible cross talk between muscles could create additional problems.

DISCUSSION

There are four main findings in this paper. The finding presented first is the least relevant from a physiological point of view but is important because it deals with the quality of the data, which in turn supports the following three findings.

First Finding

The myoelectric signal variables obtained from electrically elicited contractions show fluctuations that are much smaller than those observed in voluntary contractions. There are three factors that may cause this behavior.

Different nature of signal. Spectral estimates of random signals are intrinsically more noisy than those of deterministic signals. It has been shown that the estimates of MDF and MNF of a simulated stationary random signal, having a power spectral density similar to that of the

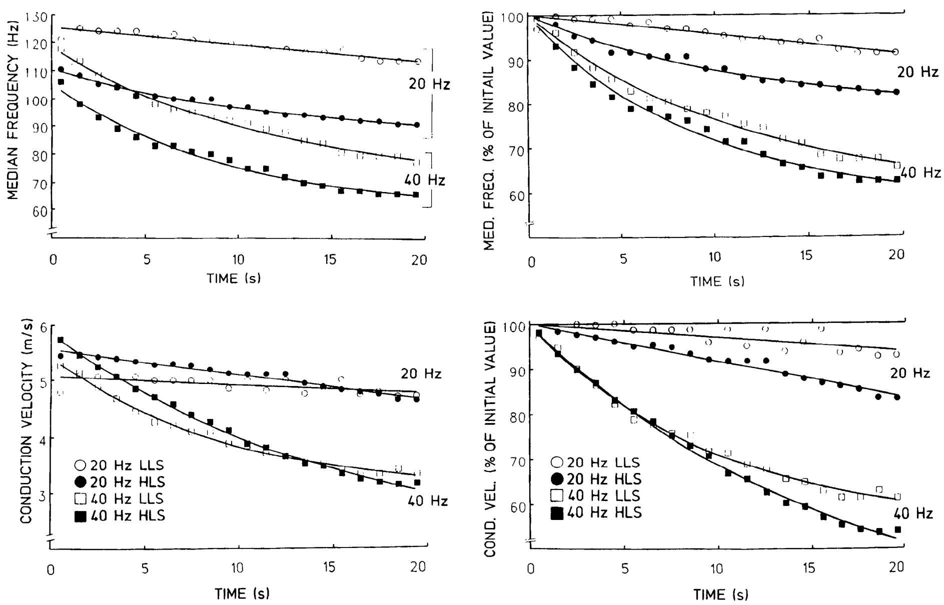

FIG. 4. Example of effect of stimulation level and frequency on time course of absolute and normalized values of median frequency and conduction velocity. Similar results were obtained for mean frequency. Linear or quasi-linear behavior is evident at low stimulation frequencies, whereas curvilinear behavior is evident at high stimulation frequencies. LLS, low level stimulation; HLS, high-level stimulation.

myoelectric signal, are respectively affected by a coefficient of variation of 6 and 4%, leading to possible estimation errors of approximately ±12% for MDF and ±8% for MNF (24), whereas during electrically elicited contractions the myoelectric signal is quasi-deterministic and quasi-periodic because the muscle is driven by the stimulator. Moreover, thanks to spectral interpolation, if the M waves elicited by a stimulation train were identical, the error would be exclusively due to the truncation of the Fourier series, the discretization of the spectral lines, and the computational accuracy. All these factors are limited only by hardware, software, computational time, and cost rather than by the properties of the signal or by physiological factors.

Computational algorithm. The averaging process applied to the electrically elicited responses reduces background biological and instrumentation noise, whereas this cannot be done during voluntary contractions. This process leads to a higher signal-to-noise ratio and slightly improves the estimates of spectral variables during electrically elicited contractions.

Visual feedback control loop used to maintain constant force. In attempting to match the target, during voluntary contractions, each subject continuously adjusted the force generated by the anterior compartment of the leg, leading to torque fluctuations of the order of ±5–10%. To attain the goal the subject modulated the firing rate and the number of recruited motor units of the tibialis anterior and may also have modulated continuously the force generated by other synergistic and antagonistic

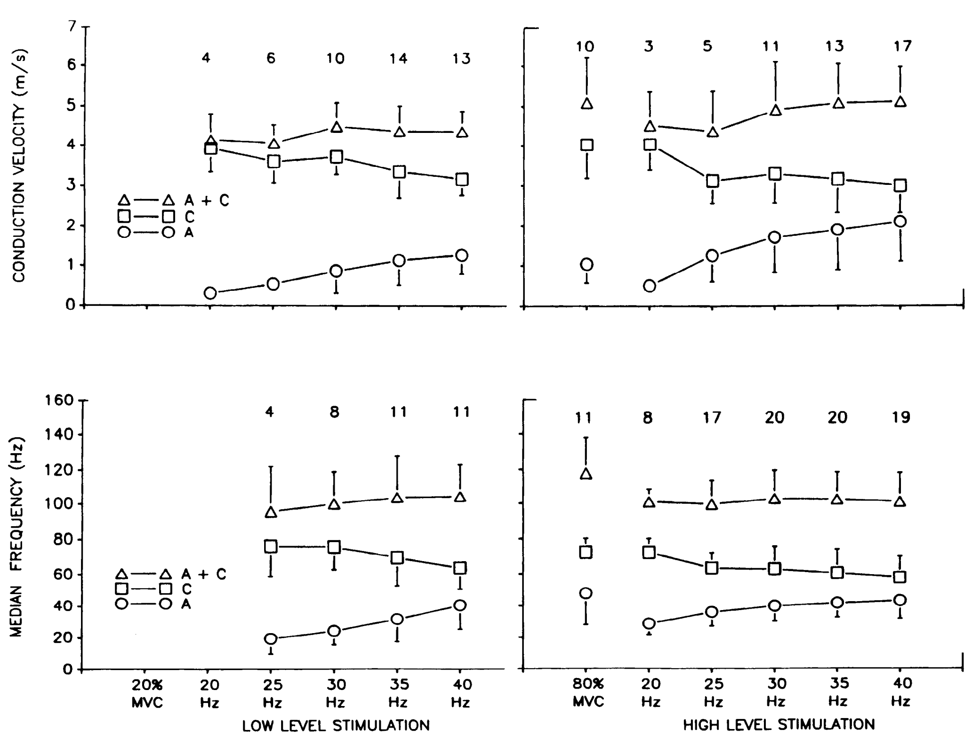

FIG. 5. Values of parameters $A + C$, C, and A of exponential regression $y = Ae^{-Bt} + C$. Numbers in each box indicate number of experiments whose contractions admitted exponential regression. Total number of experiments is 20. Exponential behavior is evident in almost all cases of fatiguing contractions (high-level stimulation at frequencies above 30 Hz) and is more common for median frequency than for conduction velocity. Results obtained for mean frequency were similar to those obtained for median frequency (see also Fig. 6). Means ± SD are shown. Large SD reflect interexperiment and intersubject variability.

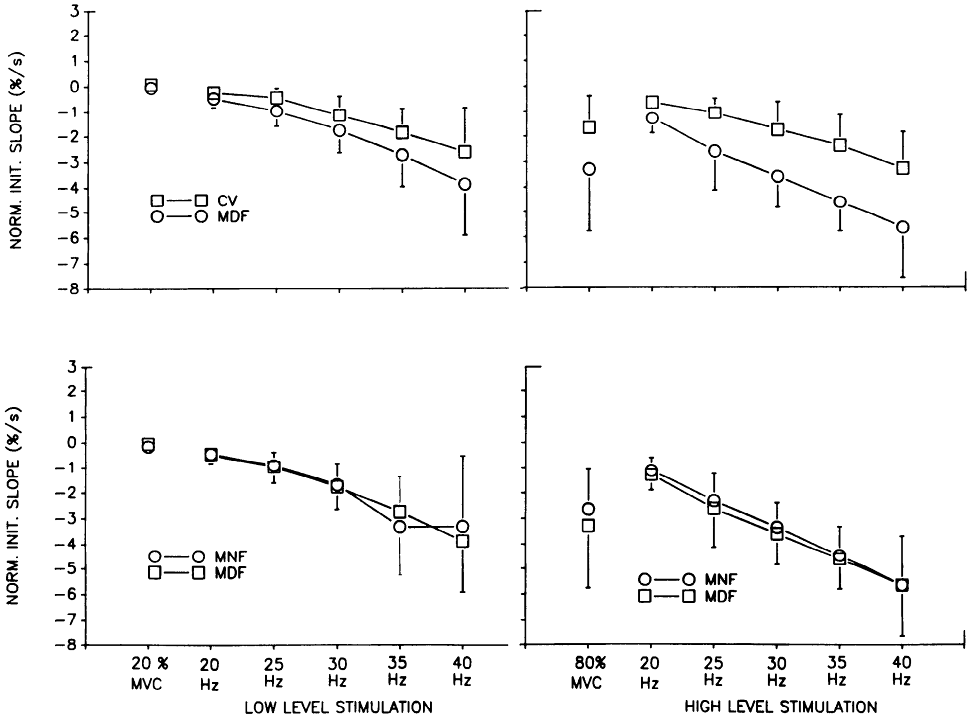

FIG. 6. Values of normalized initial slope of conduction velocity (CV) and median and mean frequencies (MDF and MNF, respectively) based on exponential regression model or linear regression model (for contractions not admitting exponential regression) over 20 s. Difference between slope of CV and that of either MNF or MDF is highly significant ($P \leq 0.01$, paired Wilcoxon test). Difference between slope of MNF and MDF is significant at $P \leq 0.05$ level only at 80% MVC and not in other conditions. Means ±SD of 20 values are shown.

muscles to compensate for the error. The resulting fluctuations of the control signal (and therefore of force), as well as any associated cross talk, may account for part of the high standard deviation of MDF, MNF, and CV that are known to be affected by force level (7, 21, 34).

Second Finding

During electrically elicited contractions MDF, MNF, and CV decrease while ARV and RMS increase. ARV increases more than RMS. All variables are affected by stimulation level and frequency. Changing stimulation frequency from 20 to 40 Hz has a greater effect than changing stimulation level from low to high. During fatiguing contractions MDF, MNF, and CV show an asymptotic behavior.

These observations can be explained by considering the following. Muscle fiber CV is known to affect spectral and amplitude variables of the myoelectric signal (4, 8). It can be shown analytically that if CV changes by a factor k ($k \leq 1$ if CV decreases), the time scale of the myoelectric signal is expanded by the same factor,

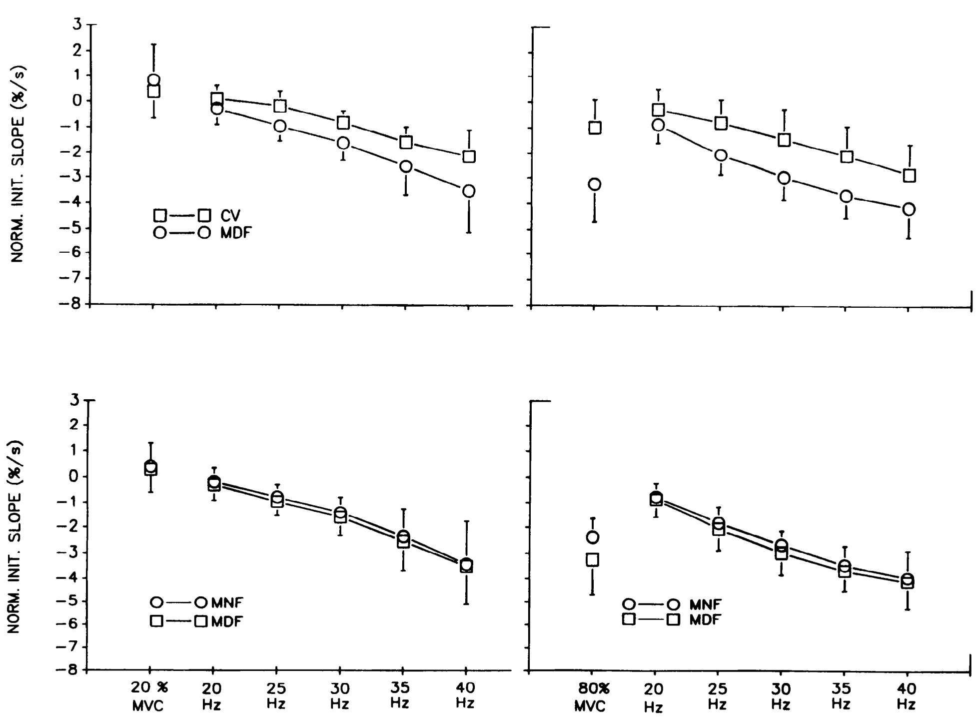

FIG. 7. Values of normalized initial slope of CV and MDF and MNF based on a linear regression model over first 5 s of contraction. Difference between slope of CV and that of either MNF or MDF is highly significant ($P \leq 0.01$). Difference between slope of MNF and that of MDF is significant at $P \leq 0.05$ level in all conditions except at 20% MVC and 20- and 40-Hz high-level stimulation. Means ± SD of 20 values are shown.

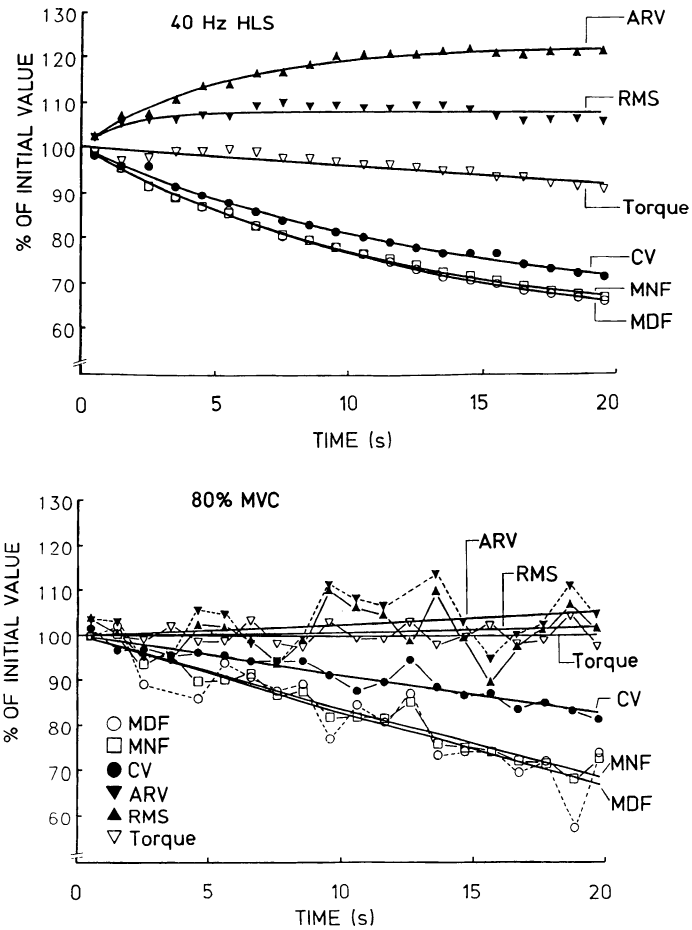

FIG. 8. Example of normalized values (normalization with respect to intercept of regression curve or line) of all variables during a 40-Hz high-level stimulated contraction and a 80% MVC from same subject. Much smaller fluctuations of estimates are evident during stimulated contractions. Similar decrements of MDF and MNF indicate spectral compression without significant spectral shape changes. Increase of average rectified value (ARV) is greater than that of root-mean-square value (RMS), as theoretically predicted, and is more evident during stimulated contraction. Force fluctuations are virtually absent during stimulated contraction, and mechanical fatigue is minimal in both contractions. Smaller rate of change of CV with respect to either MDF or MNF is more evident in the voluntary contraction.

whereas the frequency scale of the myoelectric signal power spectrum is compressed by the same factor. All characteristic frequencies (such as MNF and MDF) would change by a factor k, whereas the ARV would change by a factor $1/k$ and the RMS would change by a factor $1/\sqrt{k}$. It should therefore be expected that if only CV changes, CV, MNF, and MDF would show identical percent changes, but ARV would show changes in the opposite direction and with a magnitude of $1/\sqrt{k}$ times greater than that of RMS (for complete details see APPENDIX).

In most cases the increase of stimulation level leads to an increase of initial value of conduction velocity (21) (Fig. 4 shows one such case), indicating recruitment of faster conducting fibers. The increase of stimulation level increases the current density across the muscle activating motoneurons in progressively deeper regions. Helliwell et al. (16) and Henriksson-Larsen et al. (17) have reported that a higher percentage of type II motor units is found in the deeper portions of the human tibialis anterior. Hopf et al. (18) have reported that higher conduction velocities are associated with fast-twitch motor units, presumably type II, a result supported by data provided by Sadoyama et al. (31). It is therefore reasonable to expect that a greater muscle portion, with a higher percentage of type II motor units, would be activated at high-level stimulation with respect to low-level stimulation. A greater portion of the muscle would become ischemic. This factor, associated with the greater percentage of type II fibers, would lead to the observed greater rate of decrease of MDF, MNF, and CV (25). Alternatively, if the stimulation frequency is increased, intramuscular pressure and the production of metabolites also would increase. Both factors lead to greater accumulation of metabolites, greater electrolyte shifts, and greater rate of decrease of MDF, MNF, and CV. The asymptotic behavior of the myoelectric signal variables observed in these conditions suggests a limit value of membrane parameter changes. Beyond this limit value, propagation of action potentials may no longer be possible, and fibers (or motor units) become progressively less excitable, leading to the decrease of amplitude variables and of force output observable at the end of some contractions (Fig. 8).

Third Finding

Both MDF and MNF show rates of change greater than those of CV with either of the two definitions adopted for the rate of change. MDF shows slightly greater rates of decrease than MNF (the level of significance of the difference depends on the definition of rate of change). The average initial rate of change of CV during voluntary or electrically elicited fatiguing contractions is only 31–72% that of MDF (depending on contraction conditions), suggesting that factors other than CV affect the power spectrum compression and shape.

The different rates of change shown by MDF, MNF, and CV may be explained by the following considerations.

Nonuniform decrease of muscle fiber CV. If some fibers of a motor unit decreased their CV more than others, the spatial distribution of the depolarization zones of the individual fibers would be altered. The electric potential distribution on the surface may increase or decrease in length depending on the relative change in the conduction velocity of the individual muscle fibers. If spatial spreading occurs, then the increased length of the surface potential distribution would cause an increase in M-wave duration. MDF and MNF would be affected by both M-wave duration increase and average CV decrease and therefore would change more than CV. For example, let d be the length of the depolarization zone of each fiber and let the ratio of the minimal to maximal CV of the fibers of a motor unit decrease from 1 to 0.8 with uniform distribution. Then, at a distance $2d$ away from the innervation point (assumed identical for all fibers), the depolarization zones would spread over a length $1.4d$. The signal source would therefore be 40% wider than in the case of identical CV for all fibers. This widening would cause a decrement of MDF and MNF in addition

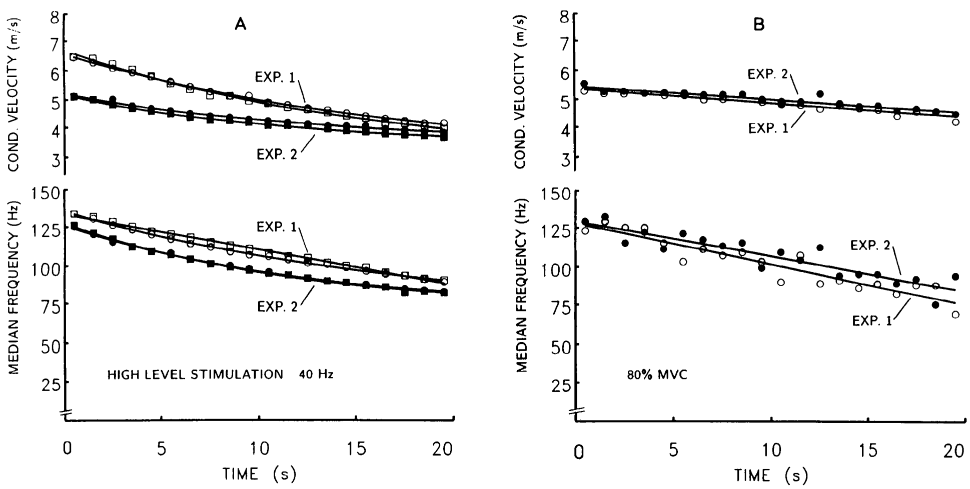

FIG. 9. *A*: example of median frequency and conduction velocity plots from two 40-Hz high-level stimulation contractions in first and second experiments on same subject. *B*: example of median frequency and conduction velocity plots from a 80% MVC contraction in first and second experiments on same subject. Similar results were obtained for mean frequency. Intraexperiment repeatability is better than interexperiment repeatability. Interexperiment repeatability is better during 80% MVC than during electrically elicited contractions.

to the 10% decrement due to the spectral compression consequent to the decrease of average CV. This behavior would be particularly evident during electrical stimulation, because in such a situation the recruited motor units act as a single motor unit with a wider range of CV values and CV rates of change.

Increase of length of depolarization zone. Changes in the potential distribution and the length of the depolarization zone of the individual activated fibers could affect MDF and MNF. This event has been discussed by Dimitrova (9) and by Gydikov et al. (14) in some experiments for which the methodology has not been well described. Normal length of depolarization zones of the voluntarily contracting human biceps was found by these authors in the range of 18.36 ± 0.48 mm, increasing to 26.64 ± 1.22 mm "after hypoxemy." During electrical stimulation, normal values were reported in the range of 30.5 ± 1.57 mm, increasing to 37.0 ± 1.2 mm "after fatigue" (14). Although these data need further support, they do suggest an interesting explanation for our findings.

Changes of spectral shape. The rate of change of MNF (averaged across subjects and based on the linear fit over the first 5 s of each contraction) varies between 85 and 96% of that of MDF, depending on the level and frequency of stimulation. This observation indicates that the power spectrum compresses with a change of shape leading to an increase of skewness. Changes of spectral shape may be due to changes of action potential shape, nonuniform decrease of CV, recruitment of new motor units, or derecruitment of previously active motor units.

Decrease of conduction velocity of motoneuron branches within muscle. Nerve branches are in the same biochemical environment as the muscle fibers and may therefore be affected by pH and ionic concentration changes due to muscle metabolism. Because motoneuron branches have different lengths, a decrease of their conduction velocity would increase the spatial spreading of the depolarization zones of the fibers of a motor unit resulting in a widening of the M wave. No information has been found in the literature for support or disproval of this hypothesis.

Fourth Finding

Contractions elicited at supramaximal stimulation and frequencies >25–30 Hz demonstrate myoelectric manifestations of muscle fatigue greater than those observed at 80% MVC sustained for the same time.

This observation can be explained by considering the following. Data about the firing rate of human tibialis anterior at high levels of contraction are scarce. Grimby et al. (13) and Hannerz (15) measured the firing rate of individual motor units of the human tibialis anterior at 100% MVC. They observed firing rates ranging from 30 to 65 pulses/s. During sustained maximal efforts the motor units with high firing rates ceased to fire while those with lower firing rates continued to fire at 15–20 pulses/s. Paul (27) found the highest firing rate of motor units of the human tibialis anterior at 80% MVC to be 33.4 pulses/s.

Our results show that a stimulation frequency of 25–30 Hz induces decrements of MNF, MDF, and CV similar to those observed at 80% MVC. It is interesting to note that 30 Hz is the clinically preferred frequency for applications of functional electrical stimulation to the muscles of the anterior compartment of the leg.

Electrical stimulation at frequencies >30 Hz may therefore be used to induce myoelectric manifestations of muscular fatigue greater and faster than those observable during voluntary contractions. Furthermore, such manifestations are independent of the subject's ability or willingness to voluntarily sustain fatiguing contractions and may be measured with greater accuracy.

General Comments

The observation that MDF or MNF changes more than CV is in conflict with the results reported by Eberstein and Beattie (10), who found the same rate of decrease for CV and MNF during 60 and 70% MVC contractions of the biceps brachii in nine healthy subjects. On the other hand, our results are in agreement with the findings of Naeije and Zorn (26), who observed a decrease of spectral variables even without simultaneous decrease of conduction velocity. Broman et al. (7) also observed a decrease of spectral variables greater than that of CV during 80% MVC contractions of the tibialis anterior in eight healthy human subjects. It appears that CV is not the only factor leading to the widening of the M wave (or to a "slowing" of the voluntary myoelectric signal), which in turn results in a decrease of MNF and MDF.

Further consideration must be given to the most appropriate measure of myoelectric manifestations of muscular fatigue. Two measures have been used in our work. The first is represented by a least-square exponential regression, or a linear regression in case of unsuitability of the exponential algorithm, over the full duration of the contraction. The measure is defined as the ratio A/τ in the exponential case and as the regression coefficient in the linear case. This approach has the advantage of using all the available values of each data set but has the disadvantage of using two different fitting criteria whose choice is mainly a matter of algorithm. In addition, in the case of exponential fitting, the value A/τ is the slope of the regression curve at $t = 0$, which may not have a clear or relevant physiological meaning. Furthermore, the estimate of the initial slope is sensitive to the position of the first few data points that may be influenced by transients due to muscle movements underneath the electrodes at the beginning of the contraction.

The second measure of fatigue is defined as the slope of a least-square regression line of the data during the first n seconds. In our work $n = 5$ appeared to be a reasonable choice. This approach has the advantage of a more uniform mathematical definition of a measure of fatigue, perhaps with a clearer physiological meaning, but uses only a limited set of points, whose number is arbitrarily selected, and does not convey any information about the curvature of the data set. Furthermore, if the decreasing behavior of the specific variable is not obvious from the first few seconds, this measure may grossly underestimate the general decreasing trend of the data. This case is particularly common during voluntary contractions with large fluctuations of the data about the regression line or curve.

APPENDIX

Effect of Signal Time Scaling on Spectral and Amplitude Variables

Consider the real signals $x(t)$ and $x(kt)$ of Fig. 10. If $x(t)$ has a Fourier transform $X(f)$ and a power spectral density $P(f) = |X(f)|^2$, then $x(kt)$ has a Fourier transform $X_k(f) = (1/|k|)\ X(f/k)$ and a power spectral density $P_k(f) = 1/k^2 P(f/k)$.

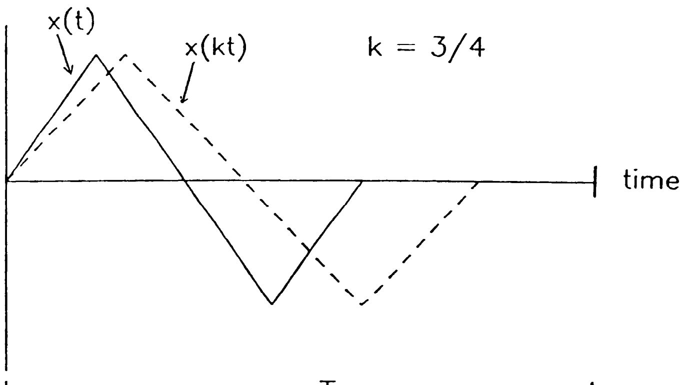

FIG. 10. Example of time scaling of a signal $x(t)$ by a factor $k = 3/4$.

The MNF of the power spectral density $P(f)$ is defined in *Eq. A1*

$$f_{\text{mean}} = \frac{\int_0^\infty f \cdot P(f)\,\mathrm{d}f}{\int_0^\infty P(f)\,\mathrm{d}f} \tag{A1}$$

the MDF of the power spectral density $P(f)$ is defined in *Eq. A2*

$$\int_0^{f_{\text{med}}} P(f)\,\mathrm{d}f = \int_{f_{\text{med}}}^\infty P(f)\,\mathrm{d}f = \frac{1}{2}\int_0^\infty P(f)\,\mathrm{d}f \tag{A2}$$

and the MNF and MDF of $P_k(f)$ are obtained by substituting $P(f)$ with $P_k(f)$ in *Eqs. A1* and *A2*. Let us define the MNF and MDF of $P_k(f)$ as $f_{\text{mean},k}$ and $f_{\text{med},k}$. By further substituting $f/k = f'$ and $\mathrm{d}f = k\mathrm{d}f'$ one obtains *Eqs. A3* and *A4*

$$f_{\text{mean},k} = \frac{\int_0^\infty \frac{f}{k^2} \cdot P\left(\frac{f}{k}\right)\mathrm{d}f}{\int_0^\infty \frac{1}{k^2} \cdot P\left(\frac{f}{k}\right)\mathrm{d}f} = \frac{\int_0^\infty f' \cdot P(f')\,\mathrm{d}f'}{k\int_0^\infty P(f')\,\mathrm{d}f'} = \frac{1}{k} f_{\text{mean}} \tag{A3}$$

$$\int_0^{\frac{f_{\text{med}}}{k}} P(f')k\,\mathrm{d}f' = \int_{\frac{f_{\text{med}}}{k}}^\infty P(f')k\,\mathrm{d}f' = \frac{1}{2}\int_0^\infty P(f')k\,\mathrm{d}f' \tag{A4}$$

which implies

$$f_{\text{med},k} = \frac{1}{k} f_{\text{med}}$$

Let us define A and R as the ARV and RMS values of $x(t)$ and A_k and R_k as the ARV and RMS values of $x(kt)$ over the time interval $0 - T$ as

$$A = \frac{1}{T}\int_0^T |x(t)|\ \mathrm{d}t$$

$$R = \sqrt{\frac{1}{T}\int_0^T x^2(t)\mathrm{d}t}$$

and

$$A_k = \frac{1}{T}\int_0^T |x(kt)|\ \mathrm{d}t$$

$$R_k = \sqrt{\frac{1}{T}\int_0^T x^2(kt)\mathrm{d}t}$$

by substituting $t' = kt$ and $\mathrm{d}t = \mathrm{d}t'/k$ and maintaining the same integration interval in the new time scale we obtain

$$A_k = \frac{1}{kT}\int_0^T |\ x(t')\ |\ \mathrm{d}t' = \frac{1}{k} A$$

$$R_k = \sqrt{\frac{1}{kT}\int_0^T x^2(t')\mathrm{d}t'} = \frac{1}{\sqrt{k}} R$$

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