Myoelectric signal conduction velocity and spectral parameters: influence of force and time

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BROMAN, HOLGER, GERARDO BILOTTO, AND CARLO J. DE LUCA. *Myoelectric signal conduction velocity and spectral parameters: influence of force and time.* J. Appl. Physiol. 58(5): 1428–1437, 1985.—Reports on measurement of muscle fiber conduction velocity in humans are scarce. Inferences on the behavior of conduction velocity have been drawn from the behavior of myoelectric spectral parameters. The present report contains information on conduction velocity and spectral parameters studied at various muscle contraction levels and during and after sustained contractions. The following results have been obtained from measurements on the tibialis anterior muscle. Conduction velocity demonstrated a positive correlation with limb circumference and with muscle force output. Thus we conclude that the diameters of the muscle fibers of high-threshold motor units are, on an average, larger than those of low-threshold motor units. The study of a sustained contraction and on the recovery after such a contraction revealed that conduction velocity consistently decreased during a strong contraction as did various myoelectric spectral parameters. However, the spectral parameters decreased approximately twice as much as did the conduction velocity, and we conclude that factors other than the conduction velocity along the muscle fibers affect the myoelectric signal during a high-level contraction. These other factors appertain to changes in the firing statistics of individual motor units as well as the correlation between the firings of different motor units.

human; muscle fatigue; motor unit firing pattern

MUSCLE FIBER CONDUCTION VELOCITY is a basic physiological parameter that greatly affects the myoelectric signal, whether detected by indwelling or surface electrodes. By muscle fiber conduction velocity, we refer to the velocity of propagation of action potentials along the muscle fibers. Early attempts to measure conduction velocity in humans were made using intramuscular electrodes (8). Later, conduction velocity of single muscle fibers was investigated in humans in situ (31). The first report on noninvasive estimation of conduction velocity was based on a frequency-domain technique, involving rather extensive signal processing times (22). Only a few reports exist (10, 20, 21) that give an interpretation of properties of myoelectric signals in terms of conduction velocities. Recently, several authors have described various methods to assess the myoelectric conduction velocity by noninvasive means (23, 24, 33).

In contrast to the scarcity of reports on muscle conduction velocities, the use of power spectral parameters to characterize myoelectric signals has received considerable attention, particularly in recent years (12, 15); for details see a review by De Luca (10). It is widely accepted that spectral shifts of the myoelectric signals occur during prolonged muscular activity, and spectral methods have been extensively used in ergonomics to quantify localized muscular fatigue (9, 15) and also in clinical research (2, 19, 29). In addition, a considerable amount of research has been directed toward the study of the dependency of myoelectric power spectrum and spectral parameters on factors such as muscle, electrode, contraction level, handedness, blood flow, and temperature (11, 25).

There is no unanimous interpretation of spectral changes observed during prolonged high-force-level contractions. Several researchers favor the hypothesis that changes in myoelectric conduction velocity can in itself explain most of the effects observed in the myoelectric signal (18, 22). In fact, it can be deduced from a mathematical model of the myoelectric power spectrum (21) that myoelectric spectral parameters are approximately proportional to conduction velocity if all other factors are constant. Others dispute this proposed leading role of the conduction velocity. They propose that other mechanisms may be important, such as changes in the firing patterns of motor units, including changes in the individual motor unit behavior, changes in synchronization between motor units, and recruitment of new motor units during constant-force fatiguing contractions (3, 4, 16). The possibility of fatigue in the neuromuscular junction has recently been ruled out (5). Also, it seems improbable that lack of energy for the metabolic machinery would be primarily responsible for changes in the myoelectric signal (1). Recently Sadoyama et al. (28) have suggested that conduction velocity changes account for only part of the spectral modifications seen in the myoelectric signal during fatiguing contractions.

The objective of the present study was bascially twofold. The first was to test the following hypotheses: *1*) muscle fiber conduction velocity increases with contraction level; 2) muscle fiber conduction velocity decreases with time during a sustained high-force-level contraction (fatigue); and *3*) muscle fiber conduction velocity increases with time after a sustained high-force-level contraction (recovery). The second objective was to compare the estimated conduction velocity values with spectral parameters evaluated from the same myoelectric signal and to compare the behavior of the spectral parameters among themselves. Such a comparison would be directed at an increased understanding of the factors that affect the myoelectric signal, for instance, in assessing if myoelectric spectral parameter changes observed during prolonged high-force-level contractions could be attributed entirely to conduction velocity changes.

METHODS

Equipment. To estimate myoelectric signal conduction velocity using a noninvasive technique, we have employed a specially constructed electrode and signal-processing scheme (unpublished observations). The method employed is based on a procedure of detecting the myoelectric signal at two different locations along the muscle fiber direction and of estimating the time delay between these two signals.

The electrode used consists of four recording surfaces in one unit. This electrode and the recording setup is depicted in Fig. 1. The rationale for using second differentials was the occurrence, in some subjects, of nondelayed activity in any two first differential channels (unpublished observations). This nondelayed activity appears as simultaneous deflections of the same polarity in both channels. Such activity will give rise to erroneously high conduction velocity estimates if included in the data used for velocity estimation. The two second differentials and the middle first differential were amplified with a pass-band of 15 Hz–1 kHz and stored on an FM tape recorder.

Myoelectric conduction velocity was estimated by a cross-correlation technique. The device that performs these calculations operates as follows: The two signals to be used are band-pass filtered between 80 and 160 Hz and amplified in a limiter to retain the signs of the original signals only. The rationale for the high value of the lower 3-dB frequency of the band-pass filter is as follows. For a conduction velocity of 4 m/s, the wavelengths of frequency components below the chosen frequency are at least 5 cm long. Such long waves would not be expected to travel undistorted along the muscle fibers. Naturally, all waves traveling along the muscle fibers will be more or less distorted due to the finite length of the fibers, longer waves being more distorted than shorter ones. The choice of 80 Hz represents a compromise between rejecting distorted waves and keeping a high signal-to-noise ratio of the myoelectric signals used for conduction velocity estimation. The sign signals are fed to a variable delay line, and one bit cross-correlation is performed for three different lag values. The difference between the two nonadjacent estimated cross-correlation values is integrated, and the integral is used to control the delays of the delay lines; for the principles of operation, see Fig. 2. The device thus locks at the maximal value of the sign cross-correlation function, and the delay to maximum is used to estimate the conduction velocity. In addition, the maximal cross-correlation value is available. The maximal cross-correlation value is corrected according to the arc sine law of correlation between the signs of two Gaussian stochastic processes in the later stages of the processing. Implicit in this method is the assumption that the conduction velocity is uniform for all frequencies of the myoelectric power spectrum, an assumption that is supported by observations of the power spectrum of single motor unit action potentials (6).

The procedures for spectral parameter estimation have been described in previous publications (7, 32). In short, all spectral parameters were obtained via analog filtering of the myoelectric signals. The following parameters were estimated: mean frequency (the ratio between the first and the zeroth spectral moments), median frequency (the frequency which divides the spectrum in areas of equal power), the intensity of zero crossings (the square root of the ratio between the second and the zeroth spectral

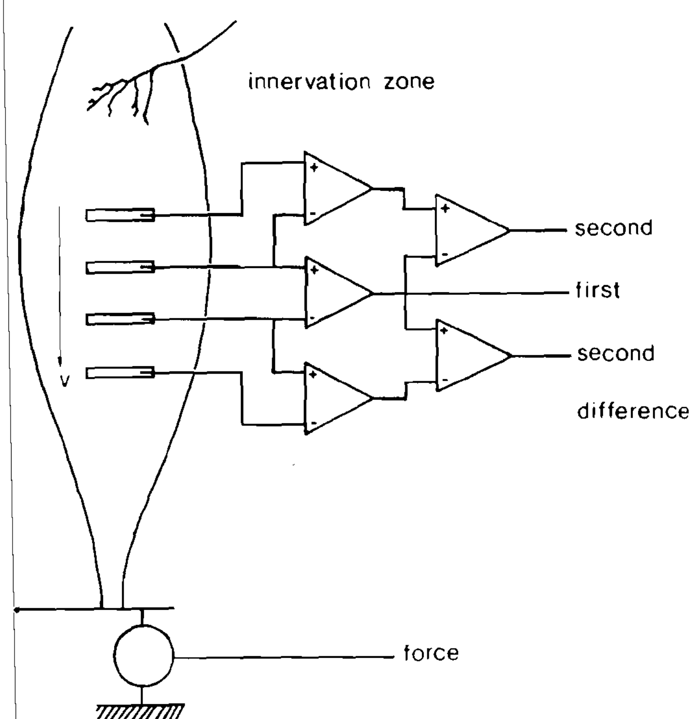

FIG. 1. Schematic of detection setup. Electrode unit with 4 bars, separated by 1 cm, is placed on 1 side of innervation zone. 3 myoelectric signals, 1 first differential and 2 second differentials, are recorded for further off-line processing. Force transducer output is also recorded.

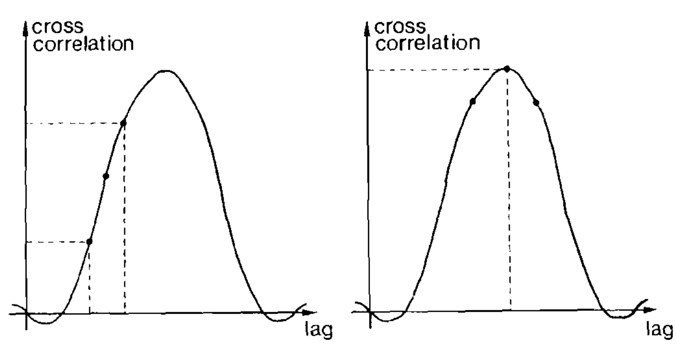

FIG. 2. Illustration of conduction velocity estimation method. Cross-correlation function of 2 myoelectric derivations and initial lags of cross-correlation estimation (*left*). Difference of cross-correlation values at 2 nonadjacent lags is used in a feedback loop to control delay lines of conduction velocity estimation. After convergence, lags are properly positioned (*right*), and time to middle lag is used as time to maximum of cross-correlation function.

moments), and the myoelectric signal root-mean-square (rms) value. From these parameters the bandwidth and relative bandwidth were calculated. In addition to measurements involving the myoelectric signal, muscle output was monitored by a force transducer.

Subjects. Eight male subjects with no known history of disease of the neuromuscular system participated in the present study after signing an informed consent form. Their ages ranged from 23 to 39 yr. All had a right-hand dominance. No selection of subjects was made based on physical condition, such as stature or level of athletic exercise.

Procedures. The experiments were performed on the right tibialis anterior muscle. The skin area above the muscle was shaved, abraded with fine sandpaper, and moistened with a conductive gel that was massaged into the skin. Care was taken not to form a superficial layer of gel that would shunt the myoelectric signals. Correct electrode placement was achieved using one or both of the following two methods. First, the motor point of the muscle was localized using superficial electrical stimulation. Second, the electrode was manipulated in the range from the motor point to the distal tendinous attachment of the muscle until the monitored myoelectric signals were similar in appearance but time delayed. The detection setup is depicted in Fig. 1.

Subjects were then seated comfortably, and the foot and ankle joint was secured to a device that was equipped with a stiff force transducer. Thus all contractions were as isometric as possible. The maximal voluntary contraction (MVC) was estimated by several attempted maximal dorsiflexions of the foot, with adequate rest periods between each contraction. The hip, knee, and ankle joints were held at approximately right angles, and the foot was inverted 15°.

The experimental protocol consisted of three studies. The first, which will be referred to as the "force study," involved a series of 10-s contractions at various force levels with appropriate rest periods in between. The following series of contractions and rest periods were used: 10% MVC, 2-min rest, 20% MVC, 2-min rest, 40% MVC, 2-min rest, 50% MVC, 3-min rest, 60% MVC, 5-min rest, 80% MVC, 5-min rest, 100% MVC, and 5-min rest. This entire sequence of events was repeated twice. The second study involved performing a sustained contraction at 80% MVC. This study, referred to as the "fatigue study," was initiated after a 10-min rest period following the force study. The sustained contraction was maintained by the subject until the experimenter instructed the subject to stop. Throughout the contraction the subject was verbally encouraged to maintain the force level as close as possible to 80% MVC. When the muscle force output was observed by the experimenter to have decreased to ~50% MVC, instructions were given to terminate the contraction. The third study, referred to as the "recovery study," consisted of a series of short contractions at 1-min intervals starting at 30 s after termination of the sustained contraction of the fatigue study. Each recording during the recovery phase consisted of a 5-s contraction at 20% MVC followed by a 5-s contraction at 80% MVC. The recovery phase was continued for at least 10 min.

In addition to the experimental data, calibration signals were recorded in conjunction with each session. Furthermore, maximal leg circumference was measured while the subject stood upright.

Processing. All equipment was calibrated using sinusoids of known frequency and amplitude. Calibrations and experimental data were played back from the tape recorder to the conduction velocity estimator, the spectral moment analyzer (7), and the median frequency monitor (32). The outputs of these devices, as well as the force signal from the tape recorder, were sampled at the rate of 10 Hz. System gains were estimated from the calibration signals. Corrections were made for time shifts caused by the instrumentation. The time constants of the analog equipment were chosen as a compromise between accuracy of estimation and sluggishness of response. All time constants were in the order of 0.5 s. Data files were processed interactively in the following way. All records pertaining to nonfatiguing contractions were subjected to a linear regression fit vs. time for each parameter. The estimated regression line was used to calculate the initial value of the parameter as well as its mean value during the contraction. Refer to Fig. 3 for details. The rationale for the initial value estimation was that even during a 10-s contraction, muscle fatigue will occur at high contraction levels, thus obscuring the data. The fatiguing contractions were processed by sectioning the data into 2-s segments and calculating mean parameter values in each segment. Base-line correction, i.e., subtraction of offset estimated during relaxation, was performed on the

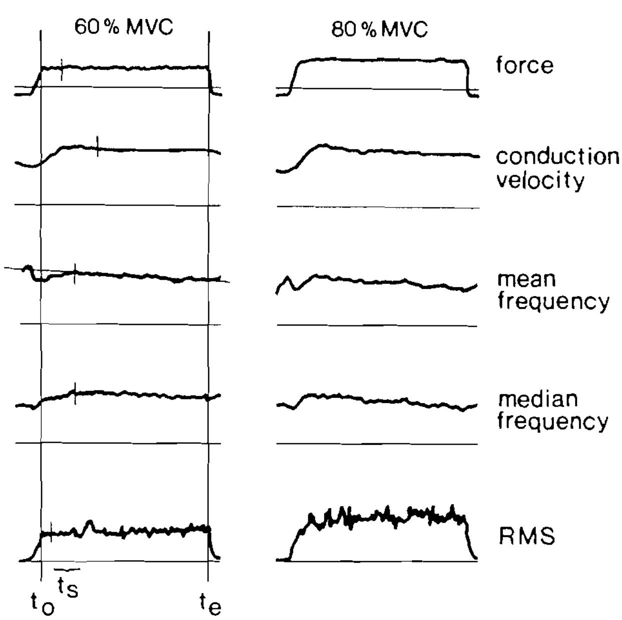

FIG. 3. Example of digitized data. From *top* to *bottom*: force transducer output, conduction velocity, mean frequency, median frequency, and myoelectric signal root-mean-square (rms) value. Data are from force study and contain records from 60 and 80% maximal voluntary contraction (MVC). Procedure for analyzing data is illustrated. Operator sets time zero (t_0) and end of record (t_e) for each epoch and for each channel start time (t_s). Times t_0 and t_e are shown by 2 *vertical lines* (*left*). t_s are shown by smaller *vertical bars* superimposed on data of each channel. Linear regression is performed between times t_s and t_e. Initial value is estimated as intercept of regression line with t_0 and mean value as mean between t_s and t_e. These parameter values are transferred to next generation of data files.

force signal and on the rms signal. An example of digitized data from our study of the influence of muscle force is shown in Fig. 3.

Evaluation of data across subjects. To compare data across subjects, a mean parameter value was estimated for each subject and each parameter studied. This mean parameter value was obtained as the average of the parameter values obtained at 40, 50, and 60% MVC for each subject. Relationships between mean parameter values were tested by the Spearman rank correlation test (30).

Evaluation of data of the force study. To study the influence of force on the parameters studied, each parameter value for each subject was normalized with respect to its estimated mean value. Thus the variability stemming from interindividual differences was reduced. The data thus treated will be referred to as normalized parameter values.

In addition, some further data manipulations were performed to test specific hypotheses. First, all normalized spectral parameters were divided by normalized conduction velocity to allow testing of the hypothesis that spectral parameters track myoelectric conduction velocity. Second, normalized rms values were divided by normalized force values to allow testing of the hypothesis that myoelectric signal output is proportional to contraction level (force).

To test the influence of muscle force on the measured and derived parameters, a low-force region was defined by the average values at 20 and 40% MVC and a high-force region by the average values at 60 and 80% MVC for each parameter and subject. A sign test was used on the difference between the high- and low-force data points to determine statistical significance. If, for any parameter, the signs for all eight subjects are the same, the high-force and low-force populations differ with a significance level of 0.5% (one-tailed comparison); if seven out of eight signs are the same, they differ with a significance level of 5% (one-tailed comparison). This result is obtained by a simple combinatorial argument. If it is assumed as a null hypothesis that there is no difference between the two populations, i.e., the probability of either sign is 0.5, then the probability of all eight signs being equal and, a priori, either positive or negative (one-tailed comparisons) is 0.5 to the power 8, which equals 1 in 256. The cumulated probability of eight or seven signs being, a priori, either positive or negative is nine times as great.

Evaluation of data in the fatigue and recovery studies. Since data in this part of our study were obtained at 20 and 80% MVC, normalization of parameters was obtained by dividing parameter values by the corresponding unfatigued values as obtained in the force study. The values thus obtained will from now on be referred to as normalized parameter values, keeping in mind that the normalization procedure differs from that of the force study.

Again, further data manipulation was performed. As in the force study, normalized spectral parameters were divided by normalized conduction velocity. Also, normalized rms values were divided by normalized force output values. In addition, normalized rms values were divided by normalized force output values and multiplied by normalized conduction velocity values to test the hypothesis suggested by Lindstrom and Magnusson (21) that myoelectric output at constant force is inversely proportional to conduction velocity.

In the fatigue study, where the use of paired differences is not adequate, we selected two populations of data points for each parameter and each subject. The populations were selected as the first third of the values and the last third of the values obtained during the fatiguing contractions. The Mann-Whitney U test (30) was applied to test the statistical significance of any conclusion that the two samples were drawn from different populations. The recovery data were not subjected to any statistical analysis due to the relatively small number of data points for each subject.

Rationale for selection of statistical tests. All statistical tests used are described elsewhere (30). Nonparametric tests were selected because such tests do not involve any assumption of the distribution of the data points. Parametric statistical tests do employ specific assumptions of the distribution of the input data. For example, the commonly used Student's t test for differences of means of two populations and various types of analysis of variance schemes assume a Gaussian distribution of data points. The conclusions drawn from such tests are not valid unless the basic conditions are fulfilled. Such assumptions of normality of data are often difficult to justify, as is the case in the present investigation.

RESULTS

Before presenting the detailed results, we would like to point out that the first and second differentially derived myoelectric signals generally displayed a similar behavior across subjects in all experimental situations, as may be seen in Tables 1 and 2. Although some details of the results differ between the two derivatives, our main results could be based on either set. As may be deduced from Table 2, the parameters of the second differential myoelectric signal contain similar information as do those of the first differential myoelectric signal but generally with lower magnitudes of the correlation coefficients. This finding is in accordance with the fact that the second differential myoelectric signal is more influenced by measurement noise than is the first. Thus, in Figs. 1–12 and Tables 3 and 4, we present data from the first differential myoelectric signal only, and tests for statistically significant relationships were not performed on the second differential myoelectric signal. As stated in METHODS, the need for using the second differential myoelectric signal arises from the conduction velocity estimation procedure. Furthermore, as was expected, the information carried by the intensity of zero-crossings parameter was equivalent to that of the mean and median frequencies but more influenced by measurement noise (14). Consequently, the intensity of zero crossings is not depicted in any figure.

Our main results are summarized in Figs. 4–12 and Tables 1–4. They are as follows.

Dependency on subject. Table 1 presents the mean values (over 40, 50, and 60% MVC) of all parameters for

TABLE 1. *Mean conduction velocity and spectral parameters for all subjects*

Subj	Circ, cm	MVC, N	F, % MVC	v, m/s	c	First Differential EMG						Second Differential EMG					
						I_0, s^{-1}	E[f], Hz	s, Hz	e	f_{50}, Hz	rms, mV	I_0, s^{-1}	E[f], Hz	s, Hz	e	f_{50}, Hz	rms, mV
BM	40.4	743	48.2	4.87	0.86	221.2	93.3	59.3	0.64	79.7	0.085	264.1	114.5	65.6	0.57	109.1	0.092
CD	34.5	484	48.7	4.10	0.90	189.5	84.6	42.7	0.51	77.2	0.140	248.6	113.4	50.9	0.45	109.1	0.156
SB	39.2	655	48.1	4.55	0.77	239.0	101.0	63.8	0.63	87.3	0.065	243.1	101.2	67.3	0.67	91.8	0.063
HB	35.5	506	50.2	3.77	0.93	229.5	98.8	58.4	0.59	97.8	0.210	252.3	114.4	53.0	0.46	112.3	0.303
JB	33.6	481	49.2	3.37	0.87	184.5	80.5	45.0	0.56	73.0	0.094	206.3	89.5	51.2	0.57	83.9	0.095
JC	34.5	586	48.9	4.13	0.97	190.4	81.3	49.5	0.61	80.7	0.142	220.0	98.8	48.2	0.49	95.5	0.165
MS	36.7	655	49.0	4.63	0.93	235.3	99.3	63.0	0.64	96.4	0.216	254.7	112.4	59.5	0.53	111.5	0.320
SR	37.5	654	49.9	3.70	0.86	213.0	92.6	52.6	0.57	82.4	0.128	259.2	112.9	63.6	0.56	103.6	0.161

Circ, maximal calf circumference; MVC, maximal voluntary contraction of subject; F, contraction level; v, conduction velocity; c, maximal cross-correlation between 2 myoelectric derivations; I_0, intensity of zero crossings; E[f], mean frequency; s, bandwidth; e, relative bandwidth (s/E[f]); f_{50}, median frequency; rms, root-mean-square value of myoelectric signal; EMG, electromyogram.

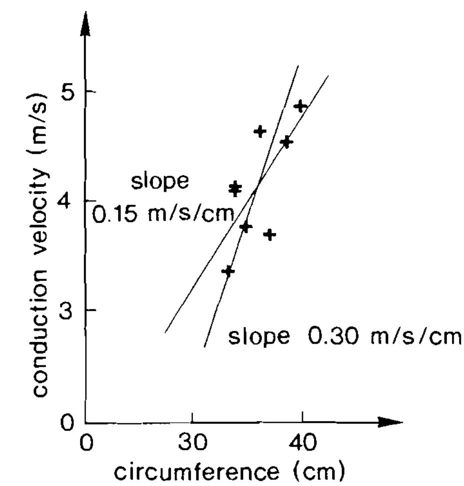

FIG. 4. Graphical representation of relationship between limb circumference and mean conduction velocity (see text for estimation procedure) for 8 subjects. Slope of relationship is in range 0.15–0.30 $m \cdot s^{-1} \cdot cm^{-1}$, see text for details.

the first and second differential myoelectric signals. Figure 4 graphically depicts the relation between limb circumference and estimated mean conduction velocity. Since neither conduction velocity nor limb circumference was free of estimation errors, two linear relations were estimated, using both variables as "error free," so that the intermediate range of slopes would be indicative of the magnitude of the covariation between circumference and conduction velocity. The range estimated from these data is 0.15–0.30 $m \cdot s^{-1} \cdot cm^{-1}$.

Table 2 presents the correlation coefficients between all parameters. Note the high correlations (statistically significant on the 5% level) between subject MVC, limb circumference, and myoelectric conduction velocity and within the myoelectric spectral parameters themselves (all but the correlation coefficient between the median frequency and the relative bandwidth are statistically significant on the 5% level) and the apparent lack of correlation between the rms value of the myoelectric signal and the above-mentioned parameters. As 21 tests for significance have been performed on the 5% level, we would expect approximately one false positive conclusion (if all variables were uncorrelated), a fact that does not affect the biological significance of our results.

Dependency on force. The force dependence of the myoelectric conduction velocity and the spectral parameters of the first differential myoelectric signal are shown graphically in Fig. 5. As may be deduced from Table 3, the following parameters increased consistently (0.5% level of significance) with force output: conduction velocity, intensity of zero crossings, mean frequency, and median frequency; whereas the relative bandwidth and the ratio between relative bandwidth and conduction velocity both decreased with force. Furthermore, the ratios between the spectral parameters and conduction velocity displayed a tendency to decrease with force. Two examples of the consistent behavior of these parameters can be found in Figs. 6 and 7, which show conduction velocity vs. force and the mean and median frequencies of the first differentially derived myoelectric signal vs. dorsiflexing force for all eight subjects, respectively.

Dependency on time during a high-force-level contraction (fatigue). Our study of the influence of time during a sustained high-force-level contraction (nominally 80% MVC) revealed statistically significant (0.5% level) decreases of the following parameters: force output, conduction velocity, all spectral parameters studied except the relative bandwidth, and the ratios between the intensity of zero crossings, the bandwidth, and the mean and median frequencies on the one hand and conduction velocity on the other. In addition, it was found that the ratio between the relative bandwidth and conduction velocity increased statistically significantly. The mean values across subjects for some of these parameters as a function of time are plotted in Fig. 8 for the first differential myoelectric signal. Figures 9 and 10 show the development over time of the conduction velocity and the mean and median frequencies for the first differential myoelectric signal, respectively. Note that the spectral parameters decrease more than the conduction velocity during the contraction. If conduction velocity changes alone were responsible for the observed changes in spectral parameters, one would expect the ratios of spectral parameters to conduction velocity to be approximately constant during the sustained high-force-level contraction. Table 4 gives the statistically significant changes of the parameters studied as observed during the high-force-level contraction.

Dependency on time after a high-force-level contraction

TABLE 2. *Correlation coefficients between mean parameters across subjects*

MVC, N	v, m/s	First Differential EMG I_0, s^{-1}	E[f], Hz	s, Hz	e	f_{50}, Hz	rms, mV	Second Differential EMG I_0, s^{-1}	E[f], Hz	s, Hz	e	f_{50}, Hz	rms, mV	
0.89	0.72	0.71	0.67	0.74	0.65	0.21	−0.38	0.68	0.42	0.94	0.62	0.25	−0.28	Circ, cm
	0.75	0.58	0.50	0.69	0.77	0.17	−0.26	0.57	0.33	0.83	0.55	0.21	−0.17	MVC, N
		0.56	0.49	0.64	0.68	0.27	−0.09	0.53	0.40	0.56	0.28	0.39	−0.04	v, m/s
			0.99	0.96	0.69	0.81	0.20	0.65	0.47	0.72	0.40	0.41	0.32	I_0, s^{-1}
				0.91	0.57	0.82	0.24	0.70	0.54	0.71	0.35	0.46	0.35	E[f], Hz
					0.86	0.75	0.13	0.53	0.33	0.71	0.48	0.30	0.25	s, Hz
						0.48	−0.01	0.23	0.05	0.55	0.49	0.07	0.09	e
							0.71	0.48	0.48	0.20	−0.11	0.56	0.78	f_{50}, Hz
								0.24	0.46	−0.40	−0.70	0.64	0.99	rms, mV
									0.94	0.63	−0.03	0.83	0.30	I_0, s^{-1}
										0.33	−0.37	0.96	0.48	E[f], Hz
											0.76	0.12	−0.28	s, Hz
												−0.54	−0.61	e
													0.66	f_{50}, Hz

MVC, maximal voluntary contraction of subject; v, conduction velocity; I_0, intensity of zero crossing; E[f], mean frequency; s, bandwidth; e, relative bandwidth (s/E[f]); f_{50}, median frequency; rms, root-mean-square value of myoelectric signal; circ, maximal calf circumference; EMG, electromyogram.

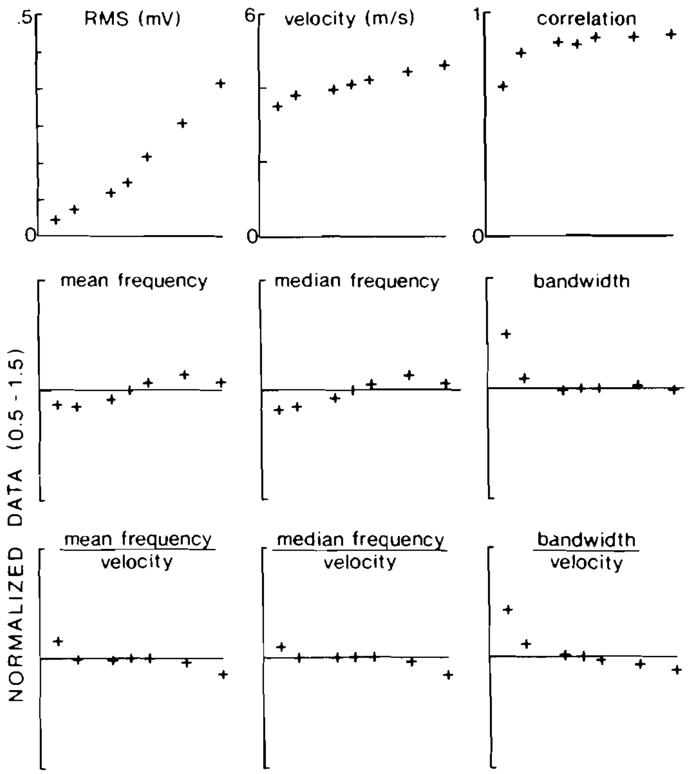

FIG. 5. Means across subjects of parameters from force study. Spectral parameters are from first differential myoelectric signal. Parameters shown are as follows: *First row*: root-mean-square (rms) value, conduction velocity, and maximal correlation; *Second row*: Normalized mean frequency, normalized median frequency, and normalized bandwidth; *Bottom row*: parameters of second row divided by normalized velocity.

TABLE 3. *Influence of contraction level on conduction velocity and spectral parameters for first differential EMG*

Subj	v	I_0	E[f]	s	e	f_{50}	I_0/v	E[f]/v	s/v	e/v	f_{50}/v	rms/F
BM	+	+	+	−	−	+	−	−	−	−	−	−
CD	+	+	+	−	−	+	−	+	−	−	+	+
SB	+	+	+	−	−	+	−	+	−	−	+	−
HB	+	+	+	+	−	+	+	+	−	−	−	+
JB	+	+	+	−	−	+	−	−	−	−	−	+
JC	+	+	+	−	−	+	−	−	−	−	−	+
MS	+	+	+	+	−	+	+	+	+	−	+	+
SR	+	+	+	−	−	+	−	+	−	−	+	−
Total	8	8	8	2	0	8	2	5	1	0	4	5

+, Increase with contraction; −, decrease with contraction; v, conduction velocity; I_0, intensity of zero crossing; E[f], mean frequency; s, bandwidth; e, relative bandwidth (s/E[f]); f_{50}, median frequency; rms, root mean square; F, contraction level; EMG, electromyogram.

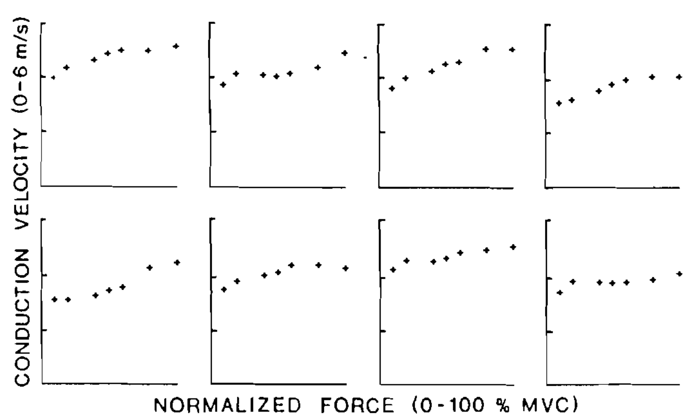

FIG. 6. Conduction velocity vs. force for all 8 subjects.

(recovery). The mean across subjects of conduction velocity and various spectral parameters during recovery are shown in Figs. 11 and 12 for the first differential myoelectric signal obtained at 20 and 80% MVC, respectively. Note the apparent flatness of the ratios between the spectral parameters and the conduction velocity.

DISCUSSION

Variation across subjects. The observed correlations (Table 2) between the subject MVC, limb circumference, the myoelectric conduction velocity, and spectral parameters are in all likelihood an expression of the influence of muscle fiber diameter on action potential conduction

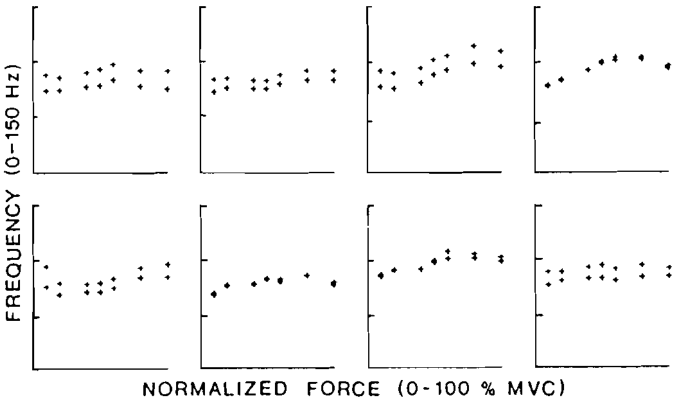

FIG. 7. Mean (*top*) and median (*bottom*) frequencies vs. force for all 8 subjects of first differential myoelectric signal.

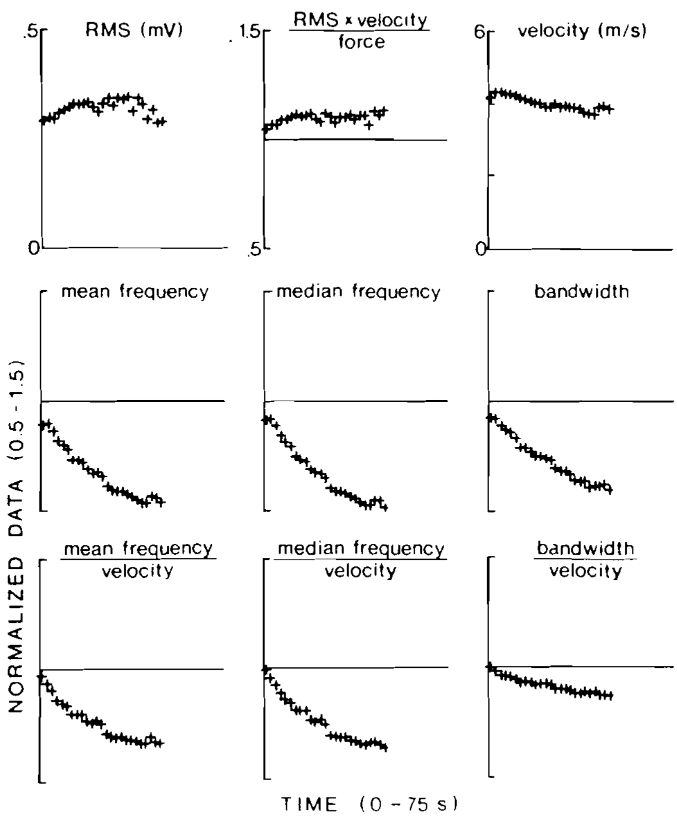

FIG. 8. Means across subjects of parameters from fatigue study. As endurance time varied across subjects, plotted data represent means across a varying number of subjects. Spectral parameters are from first differential myoelectric signal. Parameters shown are as follows: *First row*: root-mean-square (rms) value, normalized rms divided by normalized force and multiplied by normalized conduction velocity, and conduction velocity; *Second row*: normalized mean frequency, normalized median frequency, and normalized bandwidth; *Bottom row*: parameters of second row divided by normalized velocity.

velocity (13). The lack of correlation of the above-mentioned parameters and the rms value of the myoelectric signal is explained by one or more of the following arguments. First, the influence of skin thickness on the rms value is of major importance and may well mask other weaker influences. Second, a stronger subject is supposedly recruiting a larger muscle mass at a given relative contraction level, giving a positive contribution to the correlation between MVC and the rms value of

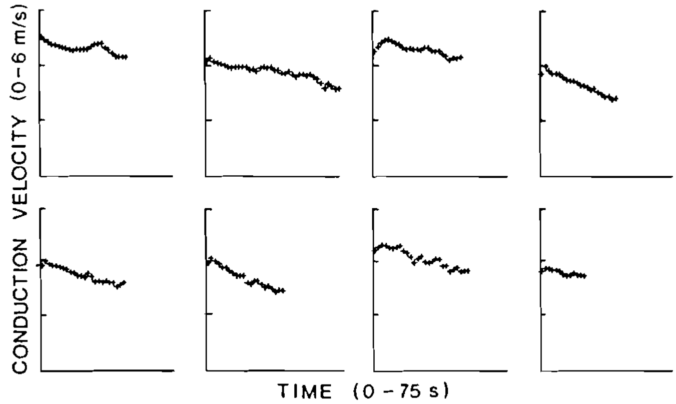

FIG. 9. Conduction velocity vs. time during a sustained high-force-level contraction (nominally 80% MVC) for all 8 subjects.

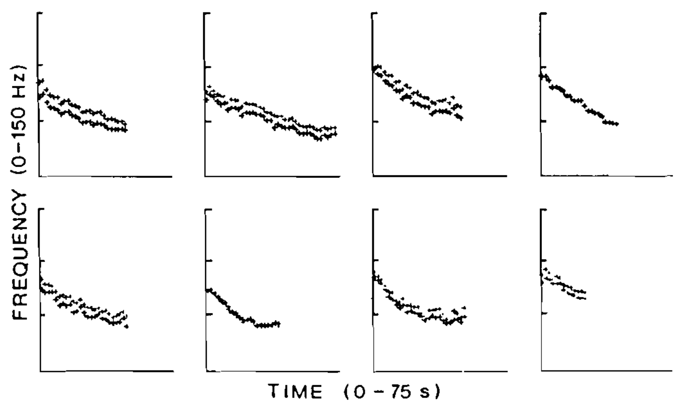

FIG. 10. Means across subjects of mean (*top*) and median (*bottom*) frequencies vs. time during a sustained high-force-level contraction (nominally 80% MVC). As endurance time varied across subjects, plotted data represent means across a varying number of subjects. Data were obtained from first differential myoelectric signal.

the myoelectric signal. On the other hand, the higher conduction velocity of the myoelectric signal of this stronger subject will give a negative contribution to this relationship. Third, a stronger subject with a larger limb circumference may have a greater distance from the recording electrode to the active muscle tissue, giving a negative contribution to the correlation between the rms value of the myoelectric signal and the maximal voluntary contraction.

Influence of force. The above-mentioned positive influence of fiber diameter on action potential conduction velocity (13) provides a simple explanation for our observations that average conduction velocity and the mean and median frequencies increase with contraction level (Figs. 5–7). In fact, our results may be interpreted as an indication that the muscle fibers in later recruited motor units are systematically larger.

The ratios between the spectral parameters (mean and median frequencies) and the conduction velocity are almost constant in the force range of 20–80% MVC. This is in agreement with the approximate proportionality between these parameters predicted from mathematical modeling (21) and empirical observations (25). The relatively low value of the spectral parameters at 100% MVC, finally, may reflect an alteration of the firing pattern of the motor units at this high level of effort.

TABLE 4. *Influence of time during a sustained strong contraction on muscle output, conduction velocity, and spectral parameters for first differential EMG*

Subj	F	v	I_0	E[f]	s	e	f_{50}	rms	I_0/v	E[f]/v	s/v	e/v	f_{50}/v	rms*v/F
BM	–	–	–	–	–	0	–	0	–	–	–	+	–	0
CD	–	–	–	–	–	0	–	+	–	–	–	+	–	+
SB	–	–	–	–	–	+	–	0	–	–	–	+	–	0
HB	–	–	–	–	–	+	–	0	–	–	–	+	–	–
JB	–	–	–	–	–	0	–	0	–	–	–	+	–	–
JC	–	–	–	–	–	0	–	0	–	–	–	+	–	0
MS	–	–	–	–	–	0	–	–	–	–	–	+	–	0
SR	–	–	–	–	–	+	–	0	–	–	–	+	–	0
Total	8	8	8	8	8	2.5	8	4	8	8	8	0	8	4.5

Contraction was nominally 80% MVC. F, contraction level; v, conduction velocity; I_0, intensity of zero crossing; E[f], mean frequency; s, bandwidth; e, relative bandwidth (s/E[f]); f_{50}, median frequency; rms, root mean square; EMG, electromyogram.

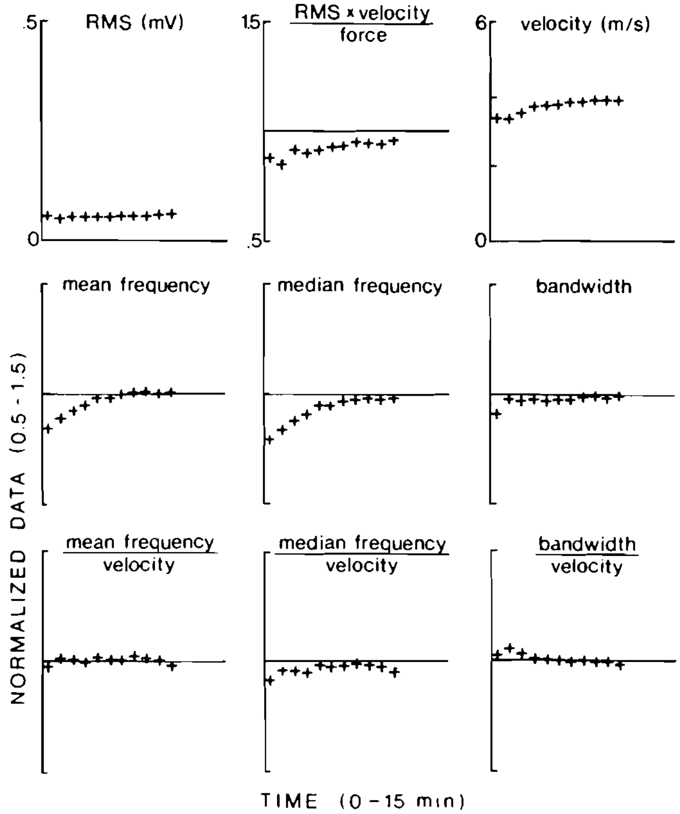

FIG. 11. Means across subjects of parameters from recovery study. Spectral parameters are from first differential myoelectric signal at 20% MVC. Parameters shown are as follows: *First row*: root-mean-square value, normalized rms divided by normalized force and multiplied by normalized conduction velocity, and conduction velocity; *Second row*: normalized mean frequency, normalized median frequency, and normalized bandwidth; *Bottom row*: parameters of second row divided by normalized velocity.

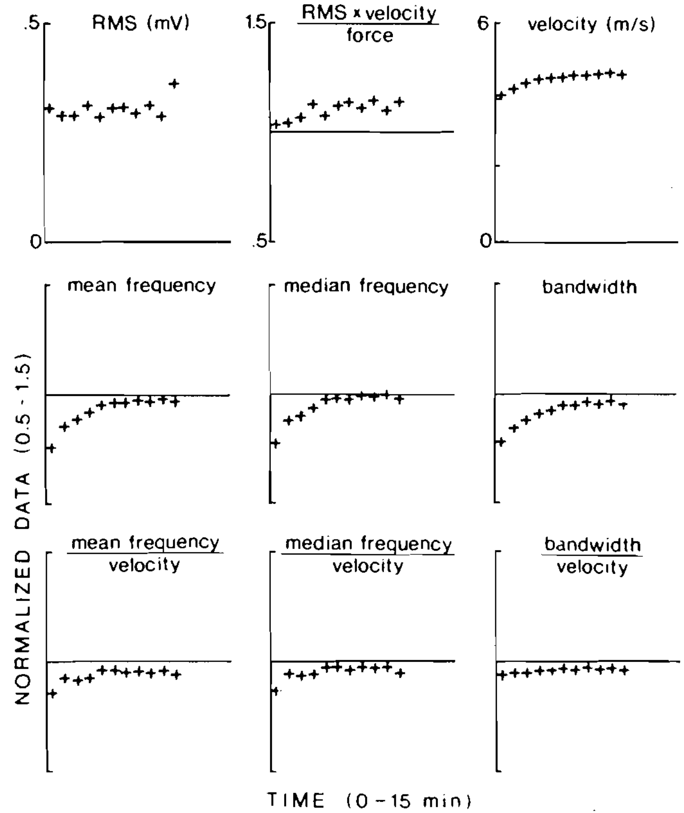

FIG. 12. Means across subjects of parameters from recovery study. Spectral parameters are from first differential myoelectric signal at 80% MVC. Parameters shown are as follows: *First row*: root-mean-square (rms) value, normalized rms divided by normalized force and multiplied by normalized conduction velocity, and conduction velocity; *Second row*: normalized mean frequency, normalized median frequency, and normalized bandwidth; *Bottom row*: parameters of second row divided by normalized velocity.

Influence of time during and after a high-force-level contraction. The information obtained on the behavior of conduction velocity and myoelectric spectral parameters during a fatiguing contraction in this study was consistent across all subjects studied (Figs. 8–10). The decreases observed in conduction velocity and spectral parameters provide confirmation of results reported by us and others but contrast with the result of Naeije and Zorn (27), who reported that myoelectric spectral parameters may decrease without any simultaneous decrease in conduction velocity. The larger decrease of the spectral parameters than that of the conduction velocity agrees with the reported conclusions of Bigland-Ritchie et al. (3). These authors, however, used different techniques to assess both myoelectric spectral parameters and conduction velocity. In addition, their conclusion is based on an experiment of muscle cooling where the effects of simultaneous nerve cooling have been neglected. Our observation of a larger decrease in spectral parameters than in myoelectric conduction velocity leaves us with the following two alternatives of interpretation. Either we are wrong in assuming proportionality between conduction velocity and spectral parameters, even if all other factors are stationary, or some additional phenomenon occurs, shifting the myoelectric signal power spectrum towards lower frequencies. This question will be resolved by studying the data obtained during the recovery from the effects of the high-force-level contraction.

The data obtained during the recovery phase were consistent across subjects. Conduction velocity and spectral parameters return towards precontraction values with a time course in close agreement with earlier reported results (6, 25); that is, recovery occurs with a time constant of a few minutes (Figs. 11 and 12). A novel observation is the apparent constancy of the ratios between mean and median frequencies with respect to the conduction velocity that is observed at all times during the recovery phase except 30 s after the cessation of the contractions.[1] This constancy is strongly suggestive of a proportionality between these spectral parameters and conduction velocity. Our interpretation of these results is that firing pattern changes contribute significantly to changes observed in the myoelectric spectral parameters during high-force-level contractions. Furthermore, this influence recovers more quickly than does the conduction velocity, although not within 30 s.

The decrease in conduction velocity during the high-force-level contraction is conveniently explained by the accumulation of metabolites that occur during a muscle contraction (26). Also, the recovery of the velocity agrees with the removal of metabolites after the contraction (17). This behavior of the conduction velocity explains part of the observed decreases of the myoelectric spectral parameters. It is not possible to indicate a cause-effect description to account for the additional decrease of the spectral parameters on an equally firm foundation. It is known (10) that the following changes in firing patterns may decrease the spectral parameters of the myoelectric signal: *1*) a decrease in mean firing rate, *2*) a change in the variability of the interpulse intervals of the individual motor units, and *3*) an increased tendency of grouped firings. All these effects have been claimed by various authors to occur during high-force-level attempted constant-force contractions; although they are not necessarily present simultaneously or even in all situations. A further possibility is the recruitment of new motor units during the fatiguing contraction. Most of these four events are poorly documented and their possible interaction even more so. In addition, the mechanisms that invoke these events, if indeed they occur, during fatiguing contractions are obscure.

In conclusion, our first main result is the demonstration of the relation between average muscle fiber conduction velocity and contraction level in the tibialis anterior muscle. This fact indicates that later-recruited motor units have larger muscle fibers than earlier-recruited ones. Our second main result is the demonstration of the consistent decrease of average myoelectric conduction velocity and myoelectric spectral parameters in the same muscle during a high-force-level contraction. Our third main result is the demonstration of the fact that no single process appears to account for the effects of a high-force-level contraction on the myoelectric signal: in addition to myoelectric conduction velocity decrease, changes in the firing patterns of the active motor units are suggested.

[1] This observation was not hypothesized before the experiment. Thus it is not possible to test this effect statistically. However, had the proper hypothesis been formulated in advance, it would have been concluded, on the 0.5% level of significance, that the ratios between the normalized mean and median frequencies and the normalized conduction velocity are less than unity at, and only at, time 30 s for the series of contractions at 80% MVC during recovery.

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