The inter- and intraindividual variability of the frequency power density spectral and surface EMG amplitude parameters and of the muscle fiber conduction velocity (MFCV) is studied in 26 healthy volunteers during fatiguing isometric ischemic intermittent exercise of the m. biceps brachii at 80% of the maximal voluntary contraction level, with a contraction rate of 30/min. No significant age effects were found. Males were significantly stronger compared with females. The higher initial SEMG amplitude and the stronger shift of the frequency power density spectrum (PDS) to lower frequencies appear to be significantly correlated with males. Fatigue induces an almost proportional compression of the SEMG frequency content. The muscle fiber conduction velocity has the highest intraindividual reproducibility (r = 0.81). Despite the definite and strong influence of the MFCV on the PDS, the shift of the PDS can not be explained by a change of MFCV alone.
Key words: fatigue • muscle fiber conduction velocity • surface electromyography • power spectrum • reproducibility

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VARIABILITY AND INTERRELATIONSHIPS OF SURFACE EMG PARAMETERS DURING LOCAL MUSCLE FATIGUE

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Surface electromyography (SEMG) is an important tool in studying muscle fatigue. Usually (i) the SEMG amplitude, and (ii) the power density frequency spectrum (PDS) of the SEMG are analyzed. During fatiguing submaximal exercise the SEMG amplitude increases and the PDS shows a shift to lower frequencies.

From the electromyogram a third parameter can be derived: the muscle fiber conduction velocity (MFCV). The MFCV is regarded as a direct electrophysiological sign of the sarcolemmal excitability. The decrease of the MFCV is an important cause for the PDS to shift to lower frequencies during fatigue.[2,8,11,19,34,37] Many experiments[2–4,7,34,37] are based on few observations, not considering gender or age.

The present SEMG study has two aims: (i) to study the intraindividual reproducibility and the influence of gender, age, or both on the PDS parameters, the SEMG amplitude, and on the MFCV during fatiguing exercise in a relatively large group of healthy volunteers; and (ii) to analyze the relationship between the decrease of MFCV and the shift of the PDS, and the relationship between the decrease of MFCV and the SEMG amplitude.

MATERIALS AND METHODS

Twenty-six healthy, right-handed volunteers (12 males: mean age 32.9 ± 9.4 [SD] years, mean height 1.84 ± 0.08 m, mean weight 77.2 ± 6.9 kg; 14 females: mean age 31.9 ± 9.7 years, mean height 1.70 ± 0.04 m, mean weight 62.4 ± 7.4 kg) participated in the experiment. Informed consent was achieved. The Committee on Experiments in

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Humans of the Faculty of Medicine approved the experimental protocol. Three age categories were distinguished: below 30 (n = 10), between 30 and 40 (n = 10), and above 40 years (n = 6).

For use in the discussion some anthropometric measurements are presented. The subcutaneous fat layer (Fat) over the m. biceps brachii by means of skinfold measurements[10] was marginally lower in males (4.9 ± 1.5 mm) compared with females (6.5 ± 2.8 mm, P = 0.05). The mean upper arm muscle–bone area[20] (MBA) of males was 54.3 ± 9.9 cm^2, which is significantly higher compared with females (45.3 ± 3.2 cm^2, $P < 0.05$).

Experimental Set-Up. The protocol consists of isometric intermittent contractions of the biceps brachii muscle at an 80% maximal voluntary contraction (MVC) level under ischemic conditions. Thirty contractions per minute are made under acoustic control of a metronome. Ischemia is applied by inflating a small cuff (8 cm), placed around the upper arm as proximal as possible, over 200 mmHg. Subjects are seated, the right arm fixed in a horizontal position with the shoulder in abduction, and the elbow in a virtually right angle. The arm flexor dynamometer is supplied with strain gauges. Because supination causes pain to the volar side of the wrist pressing against the dynamometer, the hand is pronated.[24] The force generated thus results from a combined action of the biceps brachii and brachioradial muscles. The skin is abrased and cleaned with ethanol. The surface electrode array consists of four in-line placed gold electrodes (diameter 2 mm). Two bipolar EMG signals are derived from electrodes 1 and 2 and electrodes 3 and 4, respectively. The distance between the electrode pairs is 12 mm.[24] No electrode paste is used. The two SEMG signals are amplified and bandpass (5–500 Hz) filtered with amplifiers of our own design, A/D converted (12 bits), sampled (sampling frequency 1700 Hz), and stored on magnetic disk (PDP 11/10 configuration, Digital Equipment Corp.). The exerted force is also digitized and fed into the computer.

The SEMG electrode array is positioned parallel to the biceps brachii muscle fiber direction, distal to the motor point. The maximum cross-correlation between the two SEMG signals is determined at a considerable force level, taking muscle shortening into account. Although stable estimates of MFCV are obtained for lower values, signals are only accepted if the maximum cross-correlation at a moderate force level exceeds 0.7.[2,7]

The maximal voluntary contraction (MVC) level is determined twice. Five seconds prior to the test the cuff is inflated (>200 mmHg). During the experiment, visual feedback on the performance (force) is realized on a paper recorder. The subjects are verbally encouraged to reach the 80% MVC level during each contraction. Failure point is defined as the moment when the 80% MVC level can no longer be reached. Data analysis is performed over the performance time (PT), until failure point. Intraindividual reproducibility is studied in 13 volunteers with at least a 1-week interval between both tests.

Signal Analysis. The data are computer-analyzed off-line. First, force peaks are identified with a periodicity of approximately 2 s. The force (N) in a peak is defined as the mean value in this peak in an interval of 0.25 s around the local absolute maximum. Almost coincidental EMG bursts are localized by determining their local burst maximum in a period of 0.5 s after rectification. Following these preprocessing steps, the frequency spectra of the subsequent EMG bursts in both channels are calculated (Fast Fourier Transform algorithm). Five parameters are derived. The (i) ninetieth (P90), (ii) fiftieth (Fmed), and (iii) tenth (P10) percentiles [Hz] are determined from the frequency spectrum. The (iv) EMG amplitude [μV] is derived from the root mean squared (RMS) value of the burst center. The (v) conduction velocity along the sarcolemma (MFCV [$m \cdot s^{-1}$]) is calculated from the phase difference spectrum of the two channels. Assuming a constant time delay between the two channels for all frequencies in the signal, this phase difference should be proportional to frequency. The proportionality factor can be determined by a linear least-squares fit to the phase-difference versus frequency relation. In this regression, the phase difference at a specific frequency is linearly weighted by the power at that frequency in both power density spectra. This method is an alternative from that proposed by McGill and Dorfman.[28] The MFCV is determined by relating the time delay to the interchannel distance. In order to apply the SEMG amplitude as a measure of the neuromuscular efficiency, an extra parameter is defined as "normalized amplitude" in which the RMS-amplitude of each EMG burst is divided by the mean force generated in that burst (norm.ampl [$\mu V \cdot N^{-1}$]). Neuromuscular efficiency should be considered as the reciprocal of this normalized EMG value (NME = 1/norm.-ampl).

Derived Parameter Analysis. The change of the five parameters as a function of time can be ex-

pressed accurately by a straight line characterized by the two linear regression parameters (intercept [int] and slope). The SEMG amplitude and the normalized amplitude are better characterized by one intercept and two slope parameters: the intercept belongs to a straight line over the first half of PT, and the slope parameters are calculated separately over the first half (slope1) and over the second half (slope2) of the performance time period. Details can be found in Linssen.[27]

Statistical Analysis. Mean values with (±) standard deviations are presented. The intraindividual reproducibility is calculated from measurement–remeasurement correlations (Pearson). Systematic errors are tested by calculating the difference values between the first and second experiment for each parameter (paired *t*-test). The standard deviation of the absolute differences between subjects are used to determine the absolute intraindividual measurement error (SD/$\sqrt{2}$). Some parameters are nonnormally distributed. Logarithm transformation normalizes these parameters (maximal force, slope of MFCV, and initial SEMG amplitude). The corresponding geometric means and the coefficients of variation are presented. A two-way ANOVA is used to test possible influences of gender and age on the parameters. Only correlations with a $P < 0.01$ significance level are discussed.

RESULTS

Figure 1 presents an illustration of the parameter value analysis versus time. Descriptive statistics of the results are presented in the first column of Table 1. Correlations of most parameters with the Fmed and with the MFCV are presented in Table 2.

Reproducibility of SEMG Data. Thirteen volunteers performed the experiment twice. The measurement–remeasurement correlation coefficients and the absolute measurement errors are presented in Table 1. Parameter values from the first and the second experiment are not significantly different (all $P > 0.05$, paired *t*-test).

Age and Sex. A two-way ANOVA on gender/age shows no effect of age on any of the parameters studied. Gender is found to be related to the maximal force (males: 348 ± 80 N, females: 191 ± 45 N; $P < 0.001$), to the slope of Fmed (males: -0.98 ± 0.4 Hz · s^{-1}, females: -0.64 ± 0.4 Hz · s^{-1}; $P < 0.05$), and to the initial SEMG amplitude (males: 530 ± 231 μV, females: 319 ± 91 μV; $P < 0.01$).

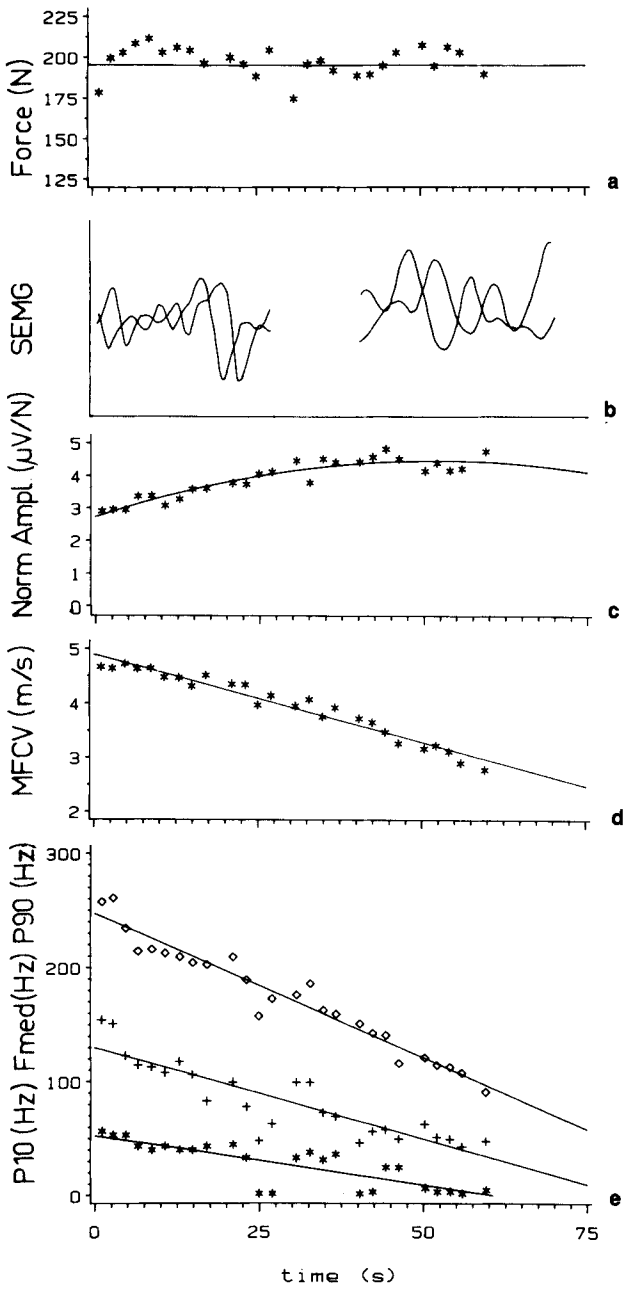

FIGURE 1. The experimental changes occurring in a healthy volunteer during a surface electromyographic (SEMG) recording of ischemic intermittent submaximal exercise over 60 s. **(a)** Force; **(b)** 30-ms bipolar SEMG recordings at the start and at the end of the performance time period; **(c)** the normalized amplitude of SEMG; **(d)** the muscle fiber conduction velocity; and **(e)** the power density spectrum percentiles (P90, P50 [Fmed], P10).

Performance Time. The mean performance time (PT) of males (69.6 ± 18.4 s) is not different from females (73.9 ± 17.8 s) (Tables 1 and 2). The PT is positively correlated with the slope of Fmed and with the slope of MFCV. Because of the negative sign of the slope parameters this means that an

Table 1. Results and reproducibility.

	Mean ± SD (n = 26)	Reproducibility (n = 13) Meas.–Remeas. (corr. coeff.)	Reproducibility (n = 13) Abs. meas. error (SD/$\sqrt{2}$)
Max. force* (N)	246.0 ± 38%	0.94	25.7
Performance time (s)	71.9 ± 17.9	0.42	11.9
Int. Fmed (Hz)	108.1 ± 31.8	0.65	16.8
Slope Fmed (Hz · s^{-1})	−0.80 ± 0.43	0.67	0.27
Int. P90 (Hz)	229.6 ± 66.1	0.80	24.6
Slope P90 (Hz · s^{-1})	−1.79 ± 0.96	0.66	0.54
Int. P10 (Hz)	48.1 ± 9.8	0.56	6.3
Slope P10 (Hz · s^{-1})	−0.31 ± 0.16	0.55	0.10
Int. MFCV (m · s^{-1})	4.85 ± 0.79	0.81	0.37
Slope MFCV* (10^{-2} m · s^{-2})	−2.31 ± 54%	0.80	0.62
Int. amplitude* (μV)	374 ± 43%	0.73	10.3
Int. norm.ampl (μV · N^{-1})	2.00 ± 0.75	0.77	0.8
Slope1 norm.ampl (10^{-2} s^{-1})	0.90 ± 1.13	0.21	1.1
Slope2 norm.ampl (10^{-2} s^{-1})	−0.30 ± 1.36	0.15	1.3

**Geometric mean and coefficient of variation (%) because of logarithm transformation.*

increase of PT correlates with less steep slopes of Fmed and MFCV.

SEMG Amplitude. The SEMG norm.ampl shows a typical curve during the test (Tables 1 and 2, Fig. 1c). Initially, the norm.ampl inclines. About halfway through the experiment a maximum is reached, whereafter the norm.ampl declines again. The intercept and the two slope values of the norm.ampl show a considerable interindividual spread. Although declining on average, the norm.ampl during the second part of the test does not differ significantly from zero. The intercept of the norm.ampl correlates negatively with the intercept of MFCV, suggesting that a fast MFCV is correlated with a good neuromuscular efficiency. The slope norm.ampl is correlated negatively with the slope MFCV during the first half of PT. This would mean that during the first half of the experiment the norm.ampl increases more when the decline of MFCV is stronger.

Table 2. Correlations.

	Int. Fmed	Slope Fmed	Int. MFCV	Slope MFCV
PT	−0.19	**0.53**	−0.22	**0.52**
Int. Fmed	—	**−0.80**	0.37	−0.17
Slope Fmed	**−0.80**	—	−0.37	**0.59**
Int. P10	**0.72**	**−0.77**	0.13	−0.38
Slope P10	−0.30	**0.68**	0.05	**0.61**
Int. P90	**0.91**	**−0.71**	**0.51**	−0.27
Slope P90	**−0.58**	**0.84**	**−0.51**	**0.79**
Int. norm.ampl	−0.05	0.13	**−0.53**	0.42
Slope1 norm.ampl	0.27	**−0.52**	**0.55**	**−0.65**
Slope2 norm.ampl	0.40	−0.34	0.42	−0.13

Correlations with $P < 0.01$ are presented in bold.

Muscle Fiber Conduction Velocity. The relationship between the MFCV and the spectral EMG parameters is studied (Tables 1 and 2, Figs. 1d and 2). Of the spectral parameters studied, the intercept of MFCV is only significantly correlated with the P90 parameters of the PDS. The slope of MFCV is significantly correlated with all spectral slope parameters. There is a linear, usually nonproportional relationship between the decay of the MFCV and the shift of the PDS to lower frequencies (Fig. 2). Figure 2 consists of 26 lines which are drawn between the coordinates ([$MFCV_{int}$, $Fmed_{int}$] and [$MFCV_{PT}$, $Fmed_{PT}$]) for each subject (solid lines). The lines are extrapolated (dotted lines). During the experiment, the solid lines are followed from right to left. In 80% of the cases the relative Fmed decay is found to be steeper than the relative decay of MFCV (Fig. 2).

Power Density Frequency Spectrum. The intercepts and slopes of the P90 and, to a lesser degree, of the P10 percentiles of the PDS are highly correlated with the intercept and with the slope of Fmed, respectively (Tables 1 and 2, Fig. 1e). The relative (slope/int) PDS parameters do not differ largely between the three percentile parameters (slope/int of P10 = 6.4 s^{-1}, of Fmed = 7.4 s^{-1}, of P90 = 7.8 s^{-1}). This indicates an almost proportional compression of the PDS during fatiguing exercise, somewhat stronger for the higher frequencies. In general, the intercept and slope of Fmed and MFCV are correlated higher with the intercept and slope P90, compared with the correlations with the P10 parameters.

DISCUSSION

Signal Analysis. The intermittent character of the obtained data requires that the analysis is only per-

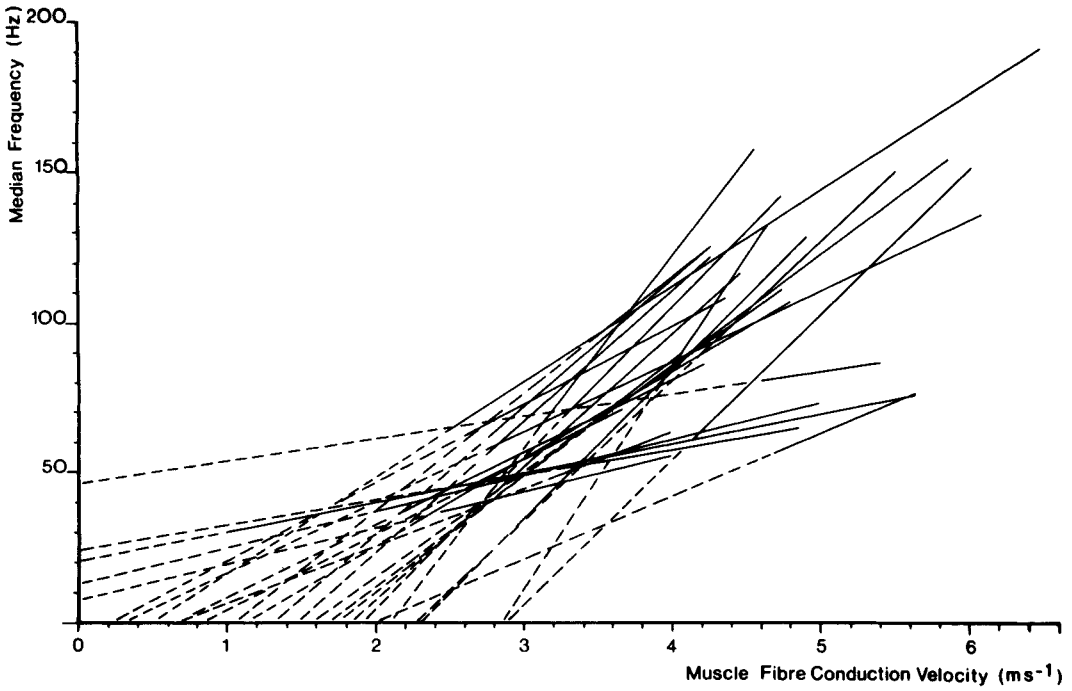

FIGURE 2. The relationship between the MFCV and the power density frequency spectrum shows that the slope of Fmed usually is steeper compared with the slope of MFCV. For further explanation see text.

formed on that part of the signal where the muscle is maximally active. A first arbitrary recognition is done on the force peaks. The length of the time segments for analysis (0.25 s and 0.5 s for force and EMG, respectively) are chosen based on sufficient stationarity of the signals within that time period. The choice of method determining the MFCV needs clarification. Considering the maximum frequency content of the EMG signals, the sample rate used to feed the EMG signals into the computer (1700 samples per second) is adequate. However, the corresponding sample interval (0.6 ms) is far too long to resolve the time shift between both signals (increasing from 3 ms to about 5 ms in a typical experiment). For that reason we use the described phase-difference versus frequency method. This method is capable of obtaining a resolution as fine as necessary independent of the sampling interval used. Aligning waveforms via phase shifting in the frequency domain, as done by McGill and Dorfman,[28] is an analogous procedure which is applied by Merletti et al.[30] in determining the MFCV. The minimization procedure needed in that approach is avoided by the linear least-squares fit method applied here. A weighing of the individual phase differences per frequency component is necessary to make both methods fully compatible. The larger the power of a specific component of the frequency power density spectrum is, the more importance the related phase-difference has in the fit of a straight line through the phase-difference spectrum. The procedure gives consistent and reproducible MFCV values, even for correlations lower than those accepted in our analysis ($r > 0.7$). For a comparison, the MFCV was always determined on basis of the cross-correlation technique as well. Although far too rough, as stated before, the outcome never differed systematically from that of the "phase-difference" method.

Reproducibility. The intercept and slope MFCV and the intercept P90 have the best reproducibility (see Table 2). Reliable MFCV measurements depend on a careful positioning of the surface electrodes parallel to the muscle fiber alignment and not too close to the motor-point or the muscle tendon. High cross-correlation coefficients do not guarantee a correct MFCV measurement.[6] The MFCV is considered to be a sensitive electrophysiological sign of integrity at the sarcolemmal level during fatigue.[24] The SEMG amplitude and most of the power density spectral parameters are influenced by a variety of other physiological factors. This may be the reason for the higher MFCV reproducibility.

Gender Influence. The sex-related force differences are most likely caused by the larger upper arm muscle cross-sectional area (MBA) and the

larger diameters of the type II muscle fibers in men.[9] The sex-related differences in slope Fmed and in intercept of amplitude appear to be caused by the thicker subcutaneous fat layer (Fat) (see Methods) in women, which determines the average distance between the surface electrodes and the active motor units, and has a low electrical conductivity. The fat layer influences the low-pass characteristics of volume-conduction in the tissue[12] and may reduce the higher frequencies in the spectral frequency distribution of women, thus resulting in a relative dominance of the lower frequencies. The apparent dependency of the Fmed on the subcutaneous layer further follows from the absence of a significant dependence of the PDS parameters to the MBA, indicating that the sex-related PDS differences are not primarily caused by the different upper arm muscle cross-sectional areas. Support for the above explanation comes from the absence of gender-related differences when comparing the relative slopes (slope/int) of Fmed (and P90). The results from simulation studies also indicate a constant ratio between the initial value and the slope of Fmed when correcting for the volume-conductor as the varying factor.[35]

SEMG Amplitude. The correlation between the initial neuromuscular efficiency and the initial MFCV indicates that, at an 80% MVC force level, muscle fibers with a stronger force-generating capacity have a higher MFCV. The initial increase of the SEMG amplitude during submaximal exercise was expected. Several factors responsible for this increase can be mentioned, such as changes in the firing frequency of individual motor units (MUs),[4,5,15] recruitment of additional MUs,[5,8,15,34] and synchronization between MU firing patterns.[5,7,8,17,21,31] Also, local changes of the sarcolemmal excitability, reflected in MFCV[5,8] and other properties of the intracellular action potential characteristics (duration, amplitude) influence the EMG amplitude.[18,23,32,33,35]

In our test, the "normalized" SEMG amplitude slightly increases initially. Because of the negative correlation between the slope of MFCV and the slope1 of the norm.ampl during the first part of the test ($r = -0.65$), this finding suggests that the expected increase of the SEMG amplitude due to recruitment of MUs and increasing firing frequencies is intensified by the decline of MFCV. Simulation studies indicate that the decline of MFCV only tends to increase the SEMG amplitude if a simultaneous increase of the intracellular action potential duration is considered.[23,35] The occurrence of such an increase of the intracellular action potential duration during fatigue has been shown.[14,22,36]

During the second part of the experiment, a very slight decline of the norm.ampl occurred, equaling zero, although the force level is maintained. One might argue that the 80% MVC force level can only be maintained when assisted by an increase in central drive. As during the first part, the decrease of MFCV coupled to an increase of intracellular action potential duration may have an increasing influence on SEMG amplitude. During the second part of the test, the force level most probably reaches the maximal voluntary contraction level that is achievable for that moment. Considering the above, factors reducing the amplitude, such as decreasing MU firing rates during prolonged maximal effort, should be assumed.[4,13] Probably the observed, almost constant, SEMG amplitude during the second part of the test originates from the counteracting mechanisms described above; i.e., the decline of MFCV and the decreasing MU firing rates, which cause a decrease of the SEMG amplitude; and the central drive and the prolonged intracellular action potential duration, which cause an increase of the SEMG amplitude.

Because of these local fiber action potential changes, the often seen interpretation of EMG amplitude as a sign of MU recruitment, firing statistics, or both, should be made cautiously. More knowledge on the intracellular electrophysiological changes affecting the SEMG amplitude is required.[35]

Power Density Frequency Spectrum. The intercepts and slopes of the PDS parameters, Fmed, P90, and P10, show a strong interdependency. When we consider the relative slope of the PDS decay (slope/int), the three PDS parameters decrease proportionally. This proportionality indicates a compression of the spectral frequency distribution without shape deformation. In healthy volunteers, the study of the change of more than one PDS parameter thus does not offer additional information.

The significant correlation (Table 2) between P90 and MFCV suggests that the influence on the frequency distribution by parameters other than the MFCV predominantly affects the lower frequency area of the PDS. This is in accordance with the knowledge that the central driving mechanisms, e.g., increasing firing frequency and synchronization, predominantly influence the lower frequency bands of the frequency distribution.[16]

Influence of the MFCV on the Frequency Spectrum. The importance of MFCV for explaining the spectral content changes during fatigue is often discussed in the literature. We therefore give special attention to the relationship between MFCV and Fmed. Generally, the influence of volume conduction[23] and the increase of the duration of motor unit action potentials caused by the influence of MFCV should be regarded as one of the major causes of the shift of the PDS to lower frequencies.[3,18] Simulation studies incorporating volume conductor characteristics and structural motor unit properties point to an almost proportional relationship between MFCV and Fmed of the PDS.[35] Some authors report that the relative fall of Fmed always exceeds that of MFCV,[4,7,11] while others point to a proportional relationship between MFCV and Fmed.[1,2,8,11,29,30] In our study, the relative decrease of Fmed is usually steeper than the relative slope of MFCV (Fig. 2). In that case, the MFCV decline cannot be the sole determinant of the PDS shift to lower frequencies, which is supported by the large variability of the individual MFCV versus Fmed data. In case of a sole determination of the Fmed changes by the MFCV decay, the individual Fmed–MFCV relationship (Fig. 2) would have been more directed to the origin. Finally, in patients with myophosphorylase deficiency, and in patients with a strong type I fiber predominance, we already showed that the PDS shifts to lower frequencies even without changes of the MFCV.[24–27]

In conclusion, this SEMG study of fatigue shows that many SEMG parameters vary considerably interindividually. This result implies that conclusions based on a small number of observations, not unusual in this field, may be fallible. Although experimentally cumbersome, it appears that the MFCV has the highest intraindividual reproducibility. Besides a number of findings which seem self-evident, some SEMG parameters, for instance, the relation between MFCV and SEMG amplitude, are more difficult to interpret.

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