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AGE RELATED CHANGES IN SURFACE MYOELECTRIC SIGNALS

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ABSTRACT. The initial values and the time course of muscle fiber conduction velocity and of surface myoelectric signal spectral variables were studied during voluntary or electrically elicited contractions of the tibialis anterior muscle of 15 healthy elderly human subjects. Age ranged from 65 to 84 years. Isometric voluntary contractions were performed at 20% MVC (Maximal Voluntary Contraction) and 80% MVC for 20 s. Tetanic electrical stimulation was then applied to the main muscle motor point for 20 s with surface electrodes. Two stimulation rates (20 Hz and 40 Hz) and two stimulation amplitudes were used to induce different degrees of fatigue. One stimulation amplitude was supramaximal, the second was adjusted to induce a response (M-wave) about 30% of the maximal. Results were compared with those reported in previous work on healthy adults (age range 18 to 43 years).

The main findings of this work are: (*a*) when voluntary contraction level is increased from 20% MVC to 80% MVC conduction velocity and spectral variables increase; this increase is significantly smaller in elderly subjects, (*b*) during sustained contractions at 80% MVC the decrease of conduction velocity and spectral variables is significantly smaller in elderly subjects, (*c*) during contractions induced by supramaximal stimulation at 40 Hz the decrease of conduction velocity and spectral variables is not significantly different in the two age groups.

It is concluded that points *a* and *b* reflect the age related decrease of number and size of fast twitch fibers indicated by histological data. Point *c* is discussed and possible explanations are suggested.

Key words: human muscles, tibialis anterior, myoelectric signal, electrical stimulation, electromyography, conduction velocity, fatigue, aging.

Progressive muscular weakness is a feature of old age and is associated with a shrinking of muscle mass and muscle cross-section (43). The loss of muscle mass is associated to a decrease of fiber size and of fiber number. These decreases are more relevant for type II fibers than for type I fibers (16, 25, 27, 38). In addition, a loss of motoneurons, with consequent denervation and subsequent partial reinnervation of motor units, has been reported by many authors (10, 16, 24, 26). This loss becomes significant in the fifth or sixth decade of life and is more relevant for neurons with larger axons (1, 39, 40). The result of these changes is a shift toward a more uniform muscle fiber pattern with fewer and larger motor units and a higher percentage of type I fibers (17). As type II fibers become smaller, the statistical distribution of type I and type II fiber diameters become progressively more similar and by the age of 80 the residual type II fibers are smaller than type I fibers (18, 19, 24, 26). Therefore, maximal voluntary contraction (MVC) force should be expected to decrease with age while isometric endurance should be expected to increase with age (assuming constant efficiency of the contractile process) and myoelectric manifestations of muscle fatigue should be expected to decrease. These facts were indeed reported by Larsson in his 1978 review (24). Davies et al. (13) observed an increase of time to peak and of half-relaxation time of the twitch response of the triceps surae of elderly subjects, in agreement with the above observations. However, their elderly subjects showed a reduced mechanical endurance which was attributed to reduced blood flow but could also be attributed to an age related deterioration of the contractile process.

It is the purpose of this work to investigate to which degree age related changes of the neuromuscular system are reflected by surface myoelectric signal variables or parameters during voluntary or electrically elicited 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 (2, 3, 9, 19, 23, 30). This muscle is also particularly suitable for conduction velocity (CV) measurements because of a relatively long region between the motor point(s) and the lower tendon sufficient to accommodate the detection electrode (36).

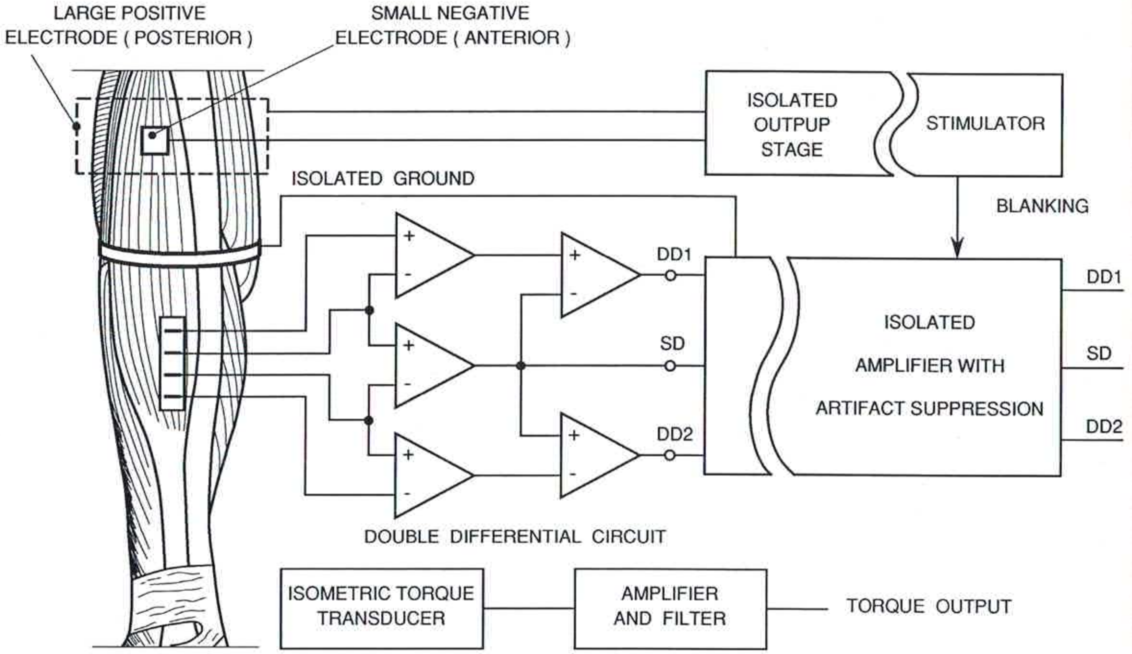

Fig. 1. Experimental set up and block diagram of the stimulation and signal conditioning system. SD: single differential myoelectric signal; DD1, DD2: double differential myoelectric signals.

Fifteen experiments were performed on 15 elderly ambulatory and self-sufficient volunteer subjects (10 females and 5 males) with no evidence of metabolic, orthopaedic or neurological disorders. For each subject, conduction velocity and F wave latency of the peroneal nerve (stimulation at the ankle) were measured and found to be within the normal range (21). Mean and SD were 49.5 ± 5.6 m/s for conduction velocity and 48.4 ± 4.0 ms for F wave latency. Ages ranged from 65 to 84 years, with a mean and SD of 70.8 ± 5.1 years for the male group and 73.8 ± 8.0 years for the female group.

Stimulation technique

Stimulation was applied, as described in greater detail in previous papers (22, 23, 30), using a monopolar technique with a negative, rectangular, soaked sponge electrode (2 by 3 cm) placed on the most proximal motor point of the muscle and a large (8 by 12 cm) positive electrode placed on the gastrocnemius muscle. Therefore, the current lines would traverse a roughly conical space across the leg alongside the tibia providing a current distribution more uniform than that obtainable with two electrodes on the same side of the leg.

Avoiding or removing the large stimulation artifact present between the myoelectric (ME) signal detection electrodes is a common problem in this type of experiments. Estimates of myoelectric signal variables are detrimentally affected by stimulation artifacts. To avoid this problem the artifact suppression technique described by Knaflitz et al. was used (22).

A rectangular current pulse with a time width of 0.2 ms was selected. Stimulation rates were chosen at 20 Hz and 40 Hz. The 20 Hz value is near the threshold for fused contraction while above 45 Hz the M-waves are no longer clearly separable and begin to overlap leading to a truncation of the waveform and consequent alteration of any frequency domain variable.

Two stimulation levels were used. A supramaximal stimulation (about 15% above the level generating the maximal M-wave) was defined as the high-level stimulation (HLS). A stimulation level eliciting an M-wave with an average peak-to-peak amplitude of 25–30% of the maximal M-wave was defined as the low-level stimulation (LLS).

Myoelectric signal detection technique

The ME signal was detected with the four-bar electrode technique described by Broman et al. (8, 9). The single differential output was obtained from the two central bars and was used to compute the ME signal spectral variables. The detection technique provided two double-differential outputs from which the average muscle fiber CV was estimated (see Data Processing section). Common-mode signals (such as power line interferences) and differential signals (such as volume-conducted signals from neighboring muscles) simultaneously present on the three electrode pairs were rejected by the input circuit. The stimulation and ME signal detection circuits were fully isolated.

The three ME signals were amplified and low pass filtered with a cut-off frequency of 480 Hz (120 dB/decade rolloff). These three signals and the torque signal were sampled at 1 024 sample/s, converted in digital form with a 12 bit A/D converter and stored on the disk of a IBM-AT personal computer. The data were processed after each contraction. The table of ME signal variables and torque versus time was also stored on disk (see Data Processing).

Fig. 1 shows a schematic diagram of the experimental set-up and of the stimulation/detection system.

Experimental protocol

The experimental protocol consisted of a preparatory phase and an experimental phase. Each subject was lying on a bed with the knee fully extended and the ankle joint at 90 degrees. 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 three. The stimulation electrode was moved over the motor point area until a location was found that provided the greates muscle mechanical response and the highest M-wave within pain tolerance. The most proximal motor point was always found to be the best according to this criterion.

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 and was positioned, with an elastic strap, with the four bars perpendicular to the muscle fibers. Correct alignment of the probe was indicated by maximally delayed and highly correlated double-differential signals during voluntary and stimulated test contractions. The best probe position was usually parallel to the tibial crest. The test contractions were intense but lasted only a few seconds. A skin temperature sensor, with a resolution of 0.1°C was fixed on the skin, near the detection electrode, to verify that skin temperature would not change more than ±0.5°C during the experiment. About five minutes were allowed before beginning the experimental phase in order to avoid any fatigue effect.

The experimental phase started with three maximal voluntary contractions (100% MVC) lasting about five seconds and spaced four minutes apart. The highest of the three maximal torque values was taken as the 100% MVC value. A 20% MVC contraction followed by an 80% MVC contraction were then performed. These voluntary contractions lasted 20 s, were spaced four minutes apart and were performed with visual feedback which was used by the subject to match and maintain a torque target level shown on an oscilloscope screen. Stimulated contractions were then performed with the subject relaxed and physically passive. Such condition was indicated by the absence of voluntary ME signals. Previous experience had indicated an excellent repeatability of data within the same experimental session (30), therefore each contraction was performed only once in order to avoid cumulative fatigue effects. In each experiment a 20 Hz and a 40 Hz LLS contraction were performed and were followed by a 20 Hz and a 40 Hz HLS contraction. Each contraction lasted 20 s and a four minute interval was allowed for recovery between contractions. The most fatiguing contraction was left as the last one.

Data processing

Spectral variables and CV were computed using the periodogram algorithm on signal epochs of 1 s providing a spectral resolution of ±1 Hz. Power spectra of stimulated contractions were calculated with the same algorithm from the electrically-elicited responses averaged over a 1 s signal epoch. Zero padding of the averaged response up to 1 s was used to obtain the same frequency resolution of voluntary contractions.

CV was computed as e/d where $e = 10$ mm was the inter-electrode distance and d was the time delay between the two double-differential signals. This 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 & Dorfman (29). the correlation coefficient between the two aligned signals was computed using Parseval's theorem and CV values were rejected if the correlation coefficient was consistently below 0.8 during a contraction. Twenty values for each variable were obtained over the 20 s contraction time.

Mean and median frequency (MNF, MDF), average rectified value and root mean square value (ARV, RMS), CV and correlation coefficient and torque were then tabulated and plotted versus time for each contraction. MNF and MDF will be jointly referred to as spectral variables. Only the behavior of spectral variables and CV will be discussed in this paper. The time course of these variables was fitted with a least-square regression line ($y = mt+n$) or a least-square exponential curve of the type $y = ae^{-bt}+c = ae^{-t/\tau}+c$ with $\tau = 1/b$. The curve that showed the lowest residual standard deviation and the highest correlation coefficient was selected and its intercept with the y-axis was taken as the initial value. Most electrically-elicited contractions showed a short transient of the variables during the first second, probably due to the movement of the muscle below the electrodes. When this was evident, the first experimental point was not included in the curve fitting algorithm (see Fig. 2A).

During high level voluntary or stimulated contractions (80% MVC and HLS) the ME signal variables showed a decreasing behavior reflecting muscle fatigue. The issue of fatigue quantification has been more extensively discussed in a previous paper (31). The term "fatigue index" is used in this work to describe an indicator of time related changes shown by a specific myoelectric signal variable.

Two indices were used to quantify the observed decrements. They are described in Fig. 2. The first index is either the slope m of the linear regression or the initial slope of the exponential regression $-a/\tau$. Both are taken as positive for decreasing patterns and negative for increasing patterns. To allow comparisons between different variables, this index was normalized with respect to the initial value n or $a+c$ and expressed in %/s. This index will be referred to as the 'normalized slope index'. The second index has been suggested by Merletti et al. (31), it is referred to as the "area ratio index" and is defined in Fig. 2B and by the following equation:

$$F = 1 - \frac{1}{2(N-1)\,y_r}\sum_{i=1}^{N-1}(y_{i-1}+y_i)$$

where y_i is the sequence of values of the variable considered (e.g. CV, MNF, MDF) sampled at N equally spaced points and y_r is a reference value, usually taken as $y_r = y_1$.

Such index is the ratio of two areas, is a pure number between 0 and 1 for decreasing behavior of the variable and is negative for increasing behavior of the variable. In addition this index is not related to any curve fitting of the data and is a function of the duration of the contraction. To avoid the effect of the initial transient, due to muscle movement, and to

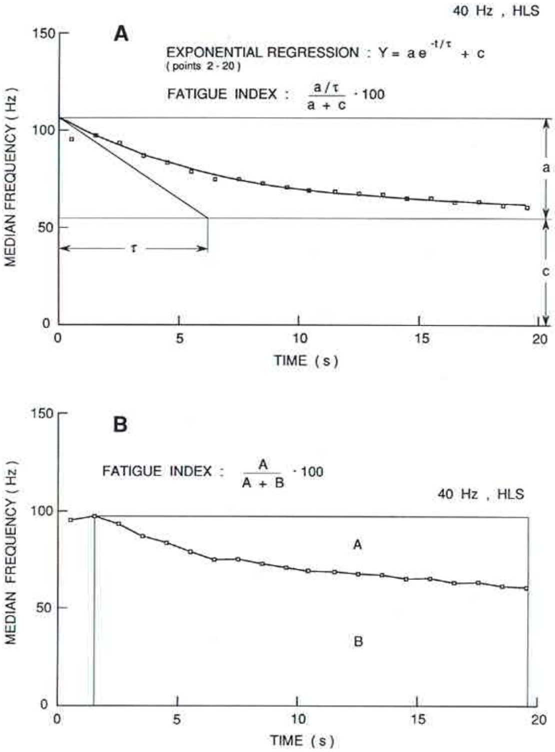

Fig. 2. (A) Definition of the normalized initial slope index. For linear regression $y = mt+n$ the normalized initial slope is m/n. For exponential regression $y = ae^{-bt}+c$, with $\tau = 1/b$, the normalized initial slope is $(a/\tau)/(a+c)$. (B) definition of the area ratio index as the ratio of the A and $A+B$ areas. Alternatively, the index may be defined as $1-B/R$ where R is the area of the reference rectangle and B is the area under the curve. See text for a more detailed definition. Regression curves and areas are computed starting from the second experimental point in order to avoid the effect of the muscle movement artifact often observed during the first second of contraction.

standardize the contraction duration, this index was always computed from the second to the last value of each contraction, as indicated in Fig. 2B. This fatigue indicator will be referred to as 'area ratio index'.

Caution must be used in extending the interpretation of these indices to an actual "measure" of muscle fatigue. Fatigue is a complex phenomenon reflected by both the amount and speed of change of myoelectric signal variables. Its quantification requires certain conditions (stable motor unit pool, etc.) and the reduction of this concept to a single value or curve is not always possible or correct (31). Therefore, the initial slope index or the area index must be considered as descriptors of the time behavior of specific myoelectric signal variables. They are useful for comparing this behavior in different experimental conditions or subject groups. Although they are related to muscle fatigue, they should not be considered as direct quantitative expressions of the fatigue phenomenon (31).

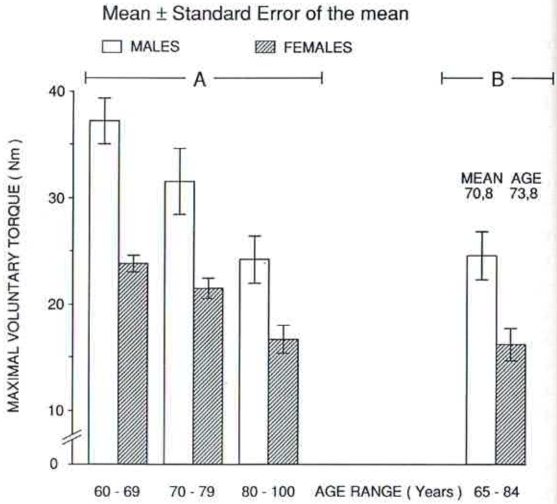

Fig. 3. Maximum isometric voluntary torque of dorsal flexion of the foot in males and females of different ages. Mean values and standard error of each mean are reported. (A) Data from Vandervoort & McComas (41). Measurements performed with the ankle joint at 30 degrees of plantar flexion. (B) Data from the sample of this study. Measurements performed with the ankle joint at 90 degrees.

RESULTS

Age related contractile changes in the dorsiflexor muscles of the human leg have been reported by Vandervoort & McComas (41). Fig. 3 shows the mean and the standard error of the mean of their (A) and our (B) estimates of maximal voluntary torque of dorsal flexion of the foot. Our values are somewhat lower because maximal efforts were performed with the ankle joint at 90 degrees while the data from Vandervoort & McComas are for an optimal ankle position at 30 degrees of plantar flexion. The torque values are in the expected range.

Torque values during electrically elicited contractions were always below 25% MVC, the highest values being obtained at 40 Hz HLS. This fact indicates that only a portion of the muscle was activated during stimulation, whereas the entire anterior compartment was activated during voluntary contractions.

Although amplitude variables (ARV, RMS) were recorded during the experiments, only CV and spectral variables will be discussed in this paper.

Initial values

Knaflitz et al. (23) reported the initial values of myoelectric signal spectral variables and CV for voluntary and electrically elicited contractions performed with

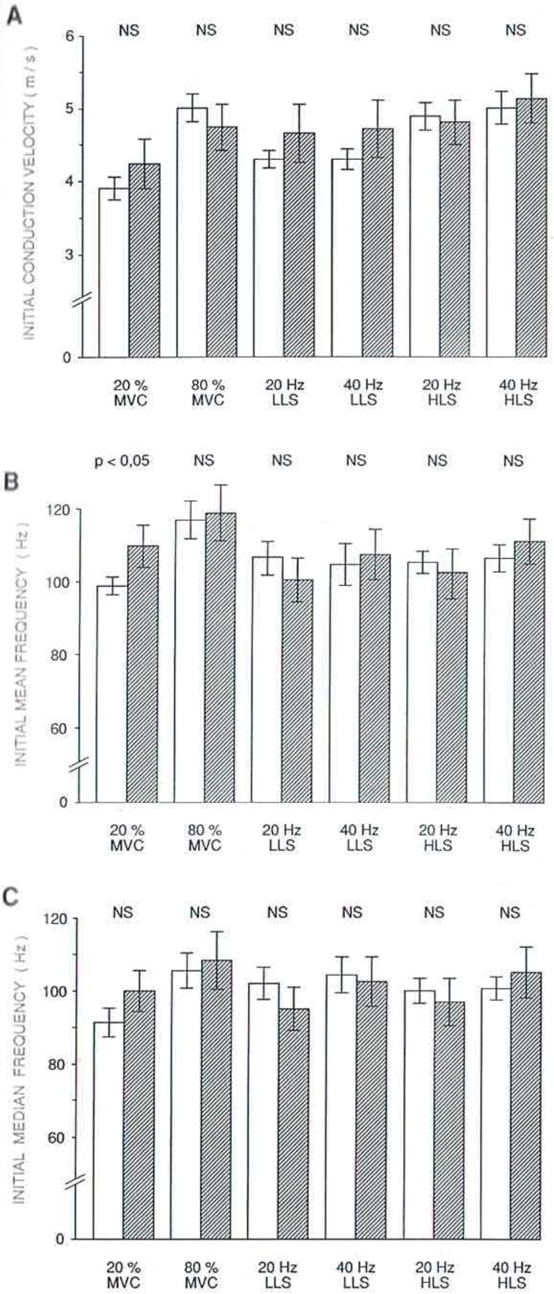

Fig. 4. Initial values of conduction velocity (A), mean frequency (B) and median frequency (C), in different experimental conditions in a group of 20 adults (18 to 43 years) and 15 elderly subjects (65 to 84 years). Data from the adult group are from Knaflitz et al. (23). Mean values and standard error of each mean are reported. □, Young adult subjects (18–43 years); ▨, Elderly subjects (65–84 years).

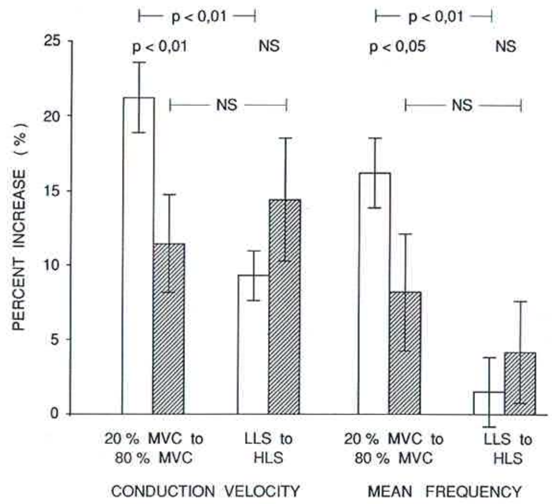

Fig. 5. Percent increment of the initial value of median frequency, mean frequency and conduction velocity when voluntary contraction is increased from 20% MVC to 80% MVC in a group of 20 adults (18 to 43 years) and 15 elderly subjects (65 to 84 years). Data from the adult group are from Knaflitz et al. (23). Mean values and standard error of each mean are reported. □, Young adult subjects (18–43 years); ▨, Elderly subjects (65–84 years).

the same protocol used in this work. Fig. 4 shows the comparison between the values obtained from their sample of 20 young adults and our sample of 15 elderly subjects. With the exception of MNF at 20% MVC the initial values of the myoelectric signal variables do not show statistically significant differences. The initial value of MNF at 20% MVC is higher in the elderly group ($p<0.05$, t-test). It is interesting to observe that CV at 20% MVC is higher in the elderly group while at 80% MVC is lower in the elderly group. Although this difference is not statistically significant, it suggests a smaller range of values of CV in the elderly group. This suggestion is supported by the results reported in Fig. 5.

An increase of MDF, MNF, CV can be observed in both groups when the voluntary contraction level is increased from 20% MVC to 80% MVC. Such increase is statistically significant ($p<0.01$ paired Wilcoxon test) for MNF and CV in the elderly group and for MDF, MNF and CV in the adult group. While all subjects of the adult group show such increase for MDF, MNF and CV, only 12 out of 15 subjects of the elderly group show it for MNF and CV and only 8 show it for MDF. Two elderly subjects show a decrease of initial values of MNF, MDF and CV when

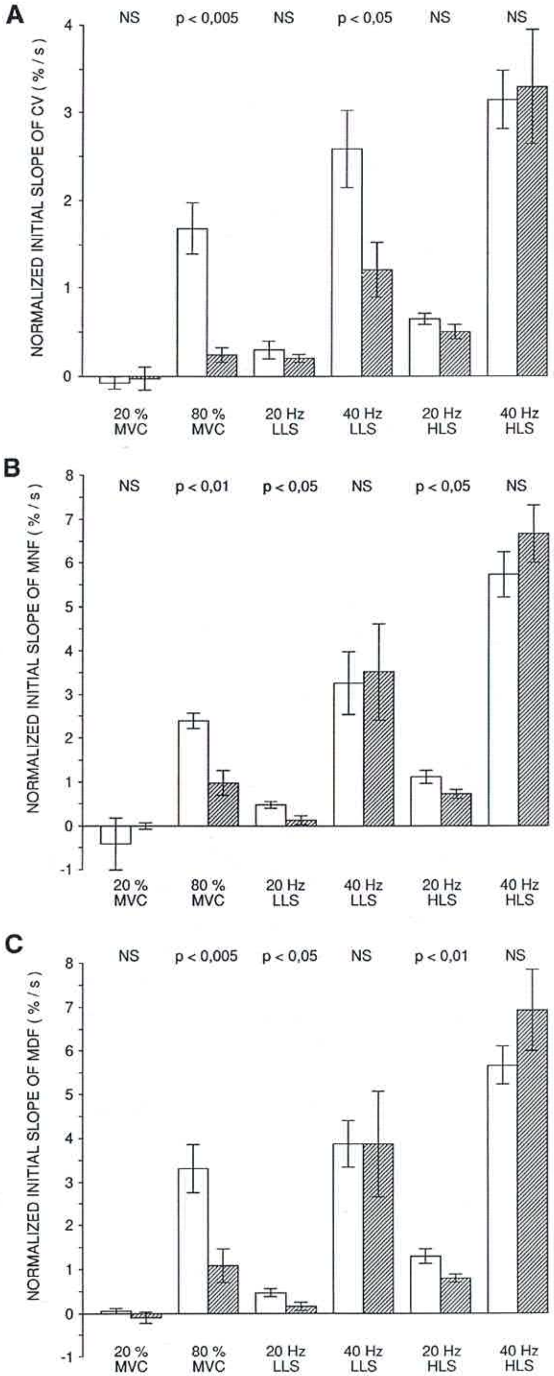

Fig. 6. Normalized initial slope index of conduction velocity (A), of mean frequency (B), and median frequency (C), in different experimental conditions in a group of 20 adults (18 to 43 years) and 15 elderly subjects (65 to 84 years). Data from the adult group are from Merletti et al. (30). Mean values and standard error of each mean are reported. □, Young adult subjects (18–43 years); ▨, Elderly subjects (65–84 years).

voluntary effort is increased from 20% MVC to 80% MVC. Fig. 5 shows that the percent increase of these three parameters is smaller in the elderly group. The difference is statistically significant for CV and MNF.

During electrically elicited contractions, reliable values of CV could be obtained from 12 out 15 subjects. In the remaining three subjects unphysiological or highly unstable values of CV were associated to low correlation coefficients between the double differential signals and were discarded.

When stimulation was increased from LLS to HLS (either at 20 Hz or 40 Hz) CV increased in 9 out of 12 cases and decreased in three cases. MDF and MNF increased in 7 cases and decreased in 5, leading to the observation of the four types of behavior reported by Knaflitz et al. (23):

Type 1 (increasing MDF-MNF, increasing CV): 5 cases

Type 2 (decreasing MDF-MNF, increasing CV): 4 cases

Type 3 (decreasing MDF-MNF, decreasing CV): 1 case

Type 4 (increasing MDF-MNF, decreasing CV): 2 cases

Overall results are reported in Fig. 5. When stimulation level is increased from LLS to HLS, the increase of initial value of CV, MDF, MNF is not significantly different in young or elderly subjects. Furthermore, elderly subjects do not show increases of CV, MDF, MNF initial values that are significantly different when voluntary effort increases from 20% MVC to 80% MVC or when stimulation level increases from LLS to HLS.

Fatigue indices

As mentioned above, the expression "fatigue index" is used here in a rather loose sense, meaning "attributes of the time course of myoelectric signal variables that are affected by fatigue". Fatigue indices calculated from our group of elderly subjects are compared with those obtained by Merletti et al. (30) from 20 adults.

At 80% MVC the normalized slope index of MDF, MNF, CV is significantly smaller in the elderly group. MDF and MNF show significantly smaller values of this index also at 20 Hz LLS and HLS contractions while they show similar values at 40 Hz LLS and HLS contractions. Fig. 6 shows the results for this index. The normalized slope index is significantly greater for MDF and MNF than for CV ($p<0.01$, paired Wil-

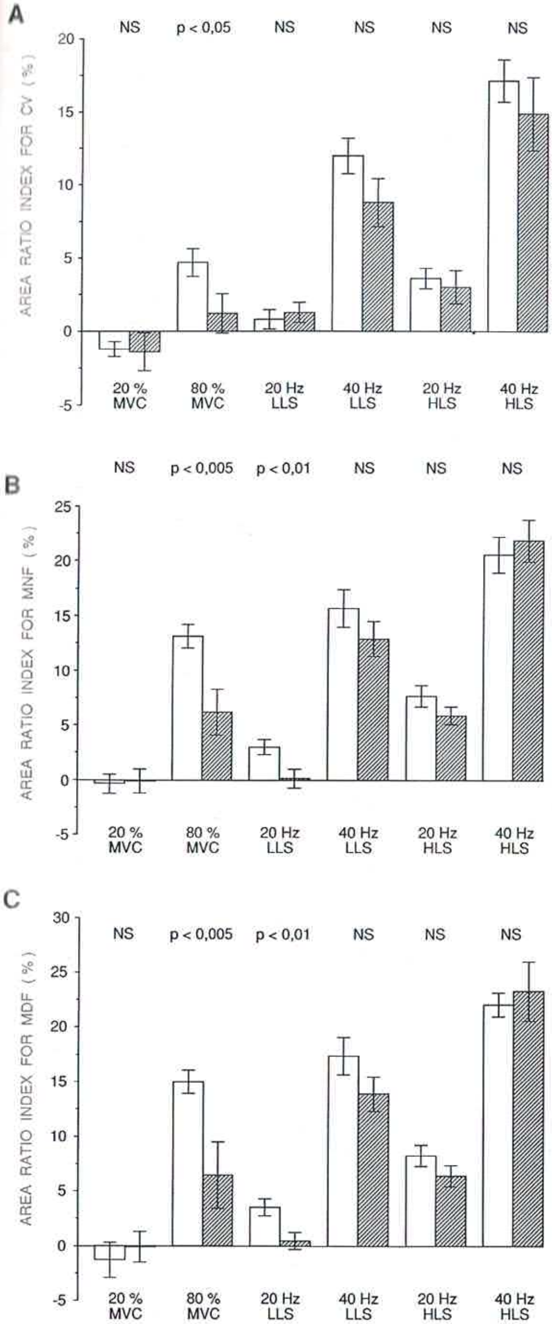

Fig. 7. Area ratio index of conduction velocity (A), mean frequency (B), and median frequency (C) in different experimental conditions in a group of 20 adults (18 to 43 years) and 15 elderly subjects (65 to 84 years). Data from the adult group are from Merletti et al. (30). Mean values and standard error of each mean are reported. □, Young adult subjects (18–43 years); ▨, Elderly subjects (65–84 years).

coxon test) in all conditions except the 'non-fatiguing' contractions at 20% MVC and 20 Hz LLS.

At 80% MVC the area ratio index of MNF, MDF, CV is significantly smaller in the elderly group than in the young group. The area ratio index for MDF in the elderly group shows significantly smaller values also at 20 Hz and 40 Hz LLS contractions. Fig. 7 shows the results for this index. The area ratio index is significantly greater for MDF and MNF than for CV ($p<0.01$, paired Wilcoxon test) in all conditions except the 'non-fatiguing' contractions at 20% MVC and 20 Hz LLS.

DISCUSSION

We are aware of the disagreement between estimates of CV performed with surface and with needle electrodes (44). In part this disagreement may be due to the structure of the muscle under consideration. In particular, the lower portion of the tibialis anterior may not necessarily consist of parallel fibers. Misalignment of some fibers with respect to the detection probe leads to an incorrect estimate of CV. However, it can be shown that the error is proportional to the cosine of the angle between the fibers and the probe and it becomes relevant only for misalignment of the order or 15–20 degrees. Individual differences in fiber alignment certainly contribute to the variability of CV estimates. Since no consistent difference in the "optimal" probe orientation was observed between the two groups (see Experimental protocol), we believe that there is no evidence of different fiber orientation in the lower tibialis anterior of young adults and elderly subjects. Individual differences would lead to variance but not to bias of the data.

Voluntary contractions

The work of Andreassen (3) has demonstrated that muscle fiber CV can be considered a "new size principle parameter" during voluntary contractions. This fact is confirmed by the work of Arendt-Nielsen & Mills, Broman et al. and Knaflitz et al. (5, 9, 23) and finds additional support in our results. However, in our elderly subjects, the increase of CV due to increased voluntary effort from 20% MVC to 80% MVC is smaller than that in the young adult group, as shown in Fig. 5. The difference is statistically significant at the 0.01 level (t-test). The fact that three out of 15 elderly subjects show a decrease of CV with increasing effort suggests either an alteration of the "size principle" or a marked decrease of CV of the motor units recruited last.

The smaller increase of CV may be attributed to a generally higher CV value shown at 20% MVC and a generally lower CV value shown at 80% MVC by the elderly subjects with respect to the young adults. Although these differences are not statistically significant, they suggest a smaller proportion of muscle fibers with very low and very high CV in the elderly population, that is, a more limited range of CV values. This suggestion finds support in the distribution of type I and type II fiber size in young and elderly subjects presented by Larsson (25). Larsson's data show a loss of large diameter type II fibers with age. This should be expected to lead to the observed lower mean value of CV in elderly subjects. Other authors have observed an age related process of denervation and partial reinnervation resulting in a decrease of motor unit number and an increase of average motor unit size (1, 12, 35, 38). This phenomenon would lead to a reduction of the number of small, low CV, type I motor units. Indeed, with age, some of these motor units may become part of larger units that would be recruited at low contraction levels therefore leading to an increase of the average CV value at 20% MVC. Further support to this hypothesis is provided by Gutmann & Hanzlikova (17) who reported a de-differentiation effect of age on muscles and a shift from a mixed pattern to a more uniform structure of muscle fibers and motor units. Although some of the literature does not concern specifically the tibialis anterior muscle, there is no indication that this muscle might undergo age related changes different from those of other muscles.

An additional factor that may affect the order of recruitment of motor units in the elderly population is a possible age-related alteration of the feedback information provided to the α-motoneuron pool by the peripheral sensors (spindles, Golgi tendon organs, proprioceptive sensors).

Morimoto & Masuda (32) reported that CV of the fibers of motor units of the human vastus lateralis increased by about 15% when the firing rate increased from 6 pps to 17 pps. Nishizono et al. (34) observed a CV increase of 31% when the firing rate of motor units of the biceps brachii was increased from 1 pps to 40 pps. An age-related smaller increase of firing rate may explain the smaller increase of CV observed in elderly subjects when voluntary contraction level increased from 20% MVC to 80% MVC.

Considerations similar to those presented for CV apply for MDF and MNF. When voluntary effort is increased from 20% MVC to 80% MVC the increase of these spectral variables is smaller in the elderly population as shown in Fig. 5 (statistical significance at the 0.05 level is shown by MNF only). The initial values of MDF and MNF at 20% MVC are higher in the elderly group ($p \leq 0.05$ for MNF only) but are almost identical in the two groups at 80% MVC (Fig. 4), indicating that they do not consistently reflect CV values. Individual differences of tissue filtering function and the effect of other factors may easily mask the relationship between initial values of spectral variables and CV when values are averaged across subjects.

A second highly significant difference between young and elderly adults concerns the normalized slope index and the area ratio index during voluntary contractions at 80% MVC. Previous work on adults has been reported by Merletti et al. (30, 31) whose findings are compared with ours. Both indices are significantly lower in the elderly subjects for MDF, MNF and CV, as indicated in Fig. 6 and Fig. 7. This observation is consistent with the age related decrease of number and size of type II fibers reported by many authors on both humans and animals (16, 17, 24, 25, 26, 27).

Karlsson et al. (20) reported a higher lactate concentration in muscles with higher percentage of fast twitch fibers. Lactate concentration affects pH which is known to affect CV (7, 14). Therefore, the significantly smaller decrease of CV observed in elderly subjects with respect to the young adults at 80% MVC may reflect the more limited pH changes due to the lower percentage of fast twitch (type II) fibers. This interpretation of our results is supported by the findings of Linssen et al. (28) who recently reported no decrement of CV in the quadriceps femoris during exercise performed by subjects with a high type I fiber predominance (>90%).

Our findings are consistent with the literature data and indicate that age related histological changes are reflected by electrophysiological variables.

Previous work has indicated that MDF and MNF decrease more than CV during sustained 80% MVC contractions (5, 4, 9, 15, 30, 33). In particular Linssen et al. (28) observed a decrease of spectral variables associated to constant CV in subjects with a high type I fiber predominance. The same conclusion emerges from our data. Both fatigue indices show a sensitivity of spectral variables to fatigue greater than that of

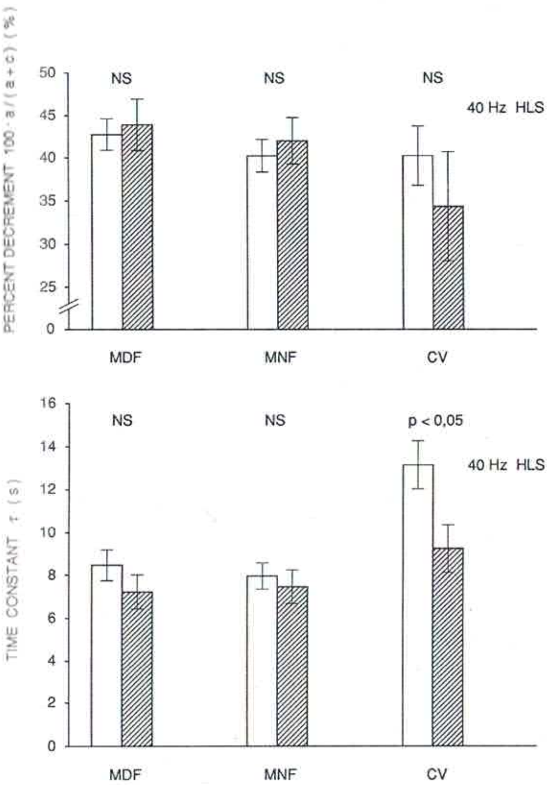

Fig. 8. Percent decrement $100a/(a+c)$ and time constant τ of the exponential regression for MDF, MNF and CV during electrically elicited contractions at 40 Hz HLS (see also Fig. 6). Data from the adult group are from Merletti et al. (30). Mean values and standard error of each mean are reported. □, Young adult subjects (18–43 years); ▨, Elderly subjects (65–84 years).

CV, indicating that other factors, beside CV, affect MDF and MNF.

Electrically elicited contractions

When stimulation intensity was increased from LLS to HLS, initial CV increased in 9 cases out of 12, suggesting recruitment in ascending order of CV, while reverse order of recruitment is suggested by the other three cases. This observation, as well as the four types of behavior of spectral variables and CV, agrees with the findings of Knaflitz et al. (23). The explanation provided by these authors relates to the interplay of electrical excitability of the terminal axonal branches and the location of these branches in the current field. This explanation may apply to our findings as well. When stimulation rate was increased from 20 Hz to 40 Hz a non-significant increase of CV was observed (see Fig. 4a), in agreement with the results of Knaflitz et al. (23). Nishizono et al. (34) observed a CV increase of about 10% when the firing rate of individual motor units of the biceps brachii was increased from 20 Hz to 40 Hz. Such change may be smaller in the tibialis anterior or might have been masked by individual variability and insufficient sample size in our experiments.

As observed during voluntary contractions and in our previous work, both fatigue indices show that, during electrically elicited contractions, the sensitivity of spectral variables to fatigue is greater than that of CV, providing additional evidence to the existence of factors, other than CV, that affect spectral variables (7, 30).

The condition showing the greatest fatigue indices is the 40 Hz HLS contraction. It is interesting to observe that in this condition no statistically significant difference can be detected between young and elderly subjects for either fatigue index and for either MDF, MNF or CV, as is evident from Figs. 6 and 7. It may be argued that equality of normalized slope indices does not necessarily imply equal time courses of the variable since the exponential regression, used for most 40 Hz HLS contractions, may show higher values of $a/(a+c)$ (see Fig. 2) associated to higher values of τ leading to similar values of $(a/\tau)/(a+c)$ (31). To check if this was indeed the case the values of τ and $a/(a+c)$ were compared in the two samples. Young adults and elderly subjects showed no significant difference between the two sets of $a/(a+c)$ values for either CV, MDF, MNF. No significant difference was found between the two sets of τ values for either MDF or MDF. CV showed a significantly smaller τ value compensated by a (non-significantly) smaller value of $a/(a+c)$, suggesting a somewhat faster but more limited decrease of CV in older subjects. Fig. 8 shows the results of these comparisons. The values of the area ratio index, reported in Fig. 7, provide further support to the observation that, at 40 Hz HLS, the time course of CV, MDF, MNF is not different in the two age groups.

This finding seems to suggest that in highly (unphysiological) fatiguing situations, such as at 40 Hz HLS, the myoelectric manifestations of muscle fatigue are no longer affected by age (and, therefore, by the proportion of type I and type II fibers).

It must be underlined that, during electrically elicited contractions, only a small portion of muscle mass is activated, most likely the most superficial, leading to torques below 25% MVC. The motor units of this portion are recruited at a fixed rate. During voluntary contractions at 80% MVC, the entire muscle is activated under control of the Central Nervous System and the different motor units are firing at a rate that is optimal in relation to their histochemical properties and endurance capabilities. Therefore, it may be speculated that two muscles, with different fiber type proportions, may show similar behaviors of CV (or spectral variables) when a portion of them is electrically activated while they may show different behaviors when their entire mass is voluntarily activated. In the second case, the CV behavior would reflect the activation strategy as well as the fiber type proportion whereas, in the first case, the fiber type proportion would be dominant. If the effect of fiber type proportion were less marked than the effect of activation strategy, the two muscles would show more similar behaviors of CV during electrically elicited contractions than during voluntary contractions. If this were the case, the differences between the fatigue indices of young and elderly subjects at 80% MVC would reflect different activation strategies rather than different fiber type constituency.

A second interpretation of the results obtained at 40 Hz HLS is based on the observation that motor units do not fire at such high rates in physiological conditions. A number of authors reported the highest firing rate of type I motor units in the range of 18–35 pps and that of type II motor units in the neighborhood of 65 pps. A comprehensive review of the literature on this subject is provided by Basmajian & De Luca (6) and by Burke (11). Using a sophisticated decomposition technique, De Luca (6) reported that, in individual muscles, motor units recruited at higher force levels fire at rates lower than motor units recruited at lower force levels, suggesting that, in physiological conditions, fast units with higher lactic acid production are activated at rates lower than fatigue resistant units. If both unit types were electrically activated at 40 Hz, as in our case, the lactic acid production would then be much higher than in voluntary contractions and might rapidly flood the activated portion of the muscle, leading to myoelectric manifestations of muscle fatigue relatively independent of fiber type constituency. Similar considerations might apply for ionic shifts across muscle fiber membranes. Therefore, it may be speculated that muscles with different fiber type proportions may show different manifestations of muscle fatigue during voluntary contractions and similar manifestations when electrically driven to unphysiological conditions.

A third possible interpretation is that a lower rate of metabolite removal, possibly due to age-related altered blood flow, compensates for the lower production rate of metabolites (consequent to a lower proportion of type II fibers in the elderly subjects) during stimulated contractions, resulting in muscle fatigue manifestations similar to those shown by muscles with a greater proportion of type II fibers and higher blood flow. This explanation was suggested by Davies et al. (13) who found higher mechanical manifestations of muscle fatigue in the triceps surae of elderly subjects and attributed it to an enhanced degree of ischemia. This interpretation of our results appears to be weak since clear differences of fatigue indices are observed at 80% MVC when the muscle is also operating in ischemic conditions.

Additional factors may relate to the effect of electrically elicited feedback upon α-motoneurons not directly excited by stimulation. The intense afferent activity might elicit reflexes that would modify the pool of activated neurons.

We are aware that the above considerations are quite speculative, however, they are supported by the observation that in less fatiguing, more physiological conditions (20 Hz LLS and HLS) the area ratio index shows consistently lower values for the elderly subjects, although statistically significant difference is shown only by spectral variables and not by CV at 20 Hz LLS (Fig. 7). The normalized slope index shows a similar behavior. In these less fatiguing conditions, the activated portion of the muscle may not be ischemic and firing rate is in the physiological range, leading to a behavior more similar to that observed during voluntary contractions.

CONCLUSIONS

There are four main findings in this work:

1. The increase of CV and spectral variables when voluntary contraction level is increased from 20% to 80% is smaller in elderly subjects than in young adults.
2. The decrease of CV and spectral variables during a sustained contraction at 80% MVC is smaller in elderly subjects than in young adults.
3. The decrease of CV and spectral variables during a sustained, fatiguing, electrically elicited contraction (such as at 40 Hz HLS) is not different in elderly subjects and in young adults.
4. The sensitivity of spectral variables to fatigue is greater than that of CV.

From point 1 and 2 it may be concluded that, during sustained voluntary contractions, CV and spectral variables reflect histological differences between muscles and may possibly be used for non-invasive muscle characterization (37, 42). Point 3 indicates that this is not the case for fatiguing electrically elicited contractions. This point requires further investigation. Point 4 provides additional support to the existing evidence that surface myoelectric signal spectral variables are affected not only by the average muscle fiber CV but by other factors as well as it is evident both in young adults and in elderly subjects.

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