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CONDUCTION VELOCITY IN HUMAN MUSCLE FIBERS - NORMAL VALUES AND TECHNICAL NOTES

Invasive and surface EMG determination techniques

J.H. van der Hoeven

ABSTRACT

Muscle fiber conduction velocity (MFCV) is studied in 51 healthy volunteers using an invasive (I-MFCV) and a surface (S-MFCV) EMG technique in the biceps brachii, in order to acquire normal values and to study reproducibility and effects of gender and age. In the invasive technique, muscle fibers are directly stimulated with needle electrodes, and the latencies of the evoked potentials are measured by a concentric needle electrode. Surface EMG is used to determine the S-MFCV, the median frequency of the power density spectrum and the integrated EMG (IEMG). A clear correlation between the mean MFCV of both methods is found. The invasive method gives information particularly about the variability of the MFCV, since the fibers are measured irrespective of innervation and recruitment order. The surface method is especially suited for examination of contraction-induced MFCV changes. Its main advantages are, beside the non-invasive character, the easy repeatability and the possibility to measure at different force levels or during continuous contractions. The S-MFCV increases with age. It is hypothesized that this is an effect of hypertrophy in the muscle fibers of the remaining MU’s, and therefore a consequence of the physiological motor unit loss during aging. We found a positive correlation between the neuromuscular efficiency (the quotient of force and IEMG) and the mean S-MFCV. It is suggested that this is a reflection of the higher neuromuscular efficiency of type II motor units.
INTRODUCTION

Studies on muscle fiber conduction velocity (MFCV) in human muscle in situ are rarely performed. Although Buchthal et al. (1955), and later other authors, performed studies on the subject (for a review see Arendt-Nielsen and Zwarts, 1989), determination of MFCV is seldom used routinely. This seems to be due mainly to technical difficulties in MFCV determination. Nevertheless, during the last decade there is increasing evidence that determination of the MFCV is not only useful in research, but can also aid in diagnosis in clinical electromyography (Troni et al. 1983b; Zwarts et al. 1988; Zwarts and van Weerden, 1989; van der Hoeven et al. 1993; Linssen et al. 1990). A major distinction in MFCV determination techniques can be made between invasive and surface methods. The purpose of this study is to determine normal values in a large control group, to study the effects of gender and age and the intraindividual reproducibility. MFCV measurements were performed in healthy subjects in biceps brachii muscle. An invasive and a surface EMG method were compared with respect to the differences, potential benefits and limitations.

METHODS

Subjects

Experiments were carried out on a group of 51 healthy individuals (30 men: mean age 36.3 SD 11.5 years, mean height 1.81 SD 0.07 m., mean weight 72.6 SD 7.9 kg; and 21 women: mean age 34.7 SD 13.1 years, mean height 1.69 SD 0.05 m., mean weight 61.1 SD 6.7 kg) with no complaints about their neuromuscular system. None of the subjects used medication. All measurements were performed after obtaining informed consent. Intraindividual reproducibility was studied in 10 subjects with at least a 2-week interval between both tests.

MFCV estimation

1. Invasive Method

Experiments were performed in the brachial biceps (short head) muscle at rest. We used a modified method of Troni et al. (1983a) on a Nicolet EMG apparatus (Viking I). Subjects were lying down, the upper part of the body slightly elevated, with the elbow slightly flexed. A stimulation needle electrode (Dantec 13L64, area of uninsulated tip: 2 mm²), was placed in the distal part of the muscle, 5-15 mm beyond the fascia. A silver surface electrode (the anode) was placed 10 to 15 mm distally (fig. 1). The muscle was stimulated with gradually increasing strength (suprathreshold), until a clear twitch was palpable (1 - 2 mA, 0.2 msec, 1 Hz). Guided by the twitch, a small concentric needle electrode (Dantec 13L58) was placed 50 - 60 mm proximal and manipulated until a reproducible polyphasic action potential was seen (amplitudes 20 - 500 µV). The signals were amplified and bandpass filtered, 500 Hz - 10 kHz; the time base varied between 5 - 10 msec per division. Care was taken to place the electrodes perpendicular to the skin. A 4-trace storage was used to ensure the reproducibility of the action potentials. Only spikes larger than 20 µV of a representative sample were used for calculations. All latencies were measured at the positive turning points and resulting invasive MFCVs (I-MFCV) were calculated. The following parameters
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were used: mean I-MFCV, fastest and slowest I-MFCV, number of spikes per insertion and ratio fastest/slowest I-MFCV (F/S ratio), as a measure of the scatter in conduction velocities.

2. Surface method

The experiments were performed on the biceps brachii muscle. The subject was seated in a chair, the arm fixed in a horizontal semiflexed position at an angle of 120 degrees, supported at the elbow and the supine wrist. The isometric force of the elbow flexion was measured at the wrist. The exerted force was displayed before the subject on a voltmeter (fig. 2). Three silver electrodes (diameter 2 mm) were placed in a rigid bipolar array with a common center electrode, interelectrode distance 10 mm. The skin was abraded and cleaned with ethanol. No electrode paste was used. The localization of the electrodes was parallel to the fiber direction, nearly halfway between the innervation zone and the distal tendon. The two EMG signals were differentially amplified (Disa EMG amplifier type 14C13) and bandpass filtered (20 - 500 Hz). The EMG signals were synchronously digitized by a 12-bit A/D converter with two different sample rates: 6024 Hz (velocity estimation) and 2048 Hz (power spectra) over two connected signal periods of 0.34 and 1.025 sec respectively. The surface MFCV (S-MFCV) was calculated from the delay of the two signals by the cross correlation method (Lynn, 1979; Naeije and Zorn, 1983). Only correlation coefficients higher than 0.85 were accepted. The power spectrum was computed over the frequency range of 5-250 Hz by application of the fast Fourier transform over the digitized signal. The median frequency (Fmed) was calculated. All measurements were performed on different force levels in duplicate: 20-30-50-75 and 100% MVC. The mean S-MFCV and the mean Fmed were calculated from the summarized data at the different force levels.

Procedure

Figure 1. The invasive EMG experimental setup.
All subjects were first examined using the invasive method. Afterwards the surface method was performed. Knowledge of the fiber direction facilitated the positioning of the surface electrodes.

**Statistics**
Differences between groups were analyzed by means of the Students $t$-test, unpaired samples, 2-tailed. The intraindividual reproducibility was calculated from measurement-remeasurement correlations (Pearson). The differences between the first and second experiment were tested for each parameter (Students $t$-test, paired samples, 2-tailed). The SD of the absolute differences was used to determine the absolute intraindividual measurement error ($SD/\sqrt{2}$). Linear regression analysis was used to test changes with age. Statistical significance was accepted at a level of 5%.

**RESULTS**

*Figure 2. The surface EMG experimental setup.*
In all subjects an invasive MFCV measurement could be made. In three subjects, surface MFCV estimation was impossible because of correlation coefficients below 0.85. For a summary of the results see table I. Ten subjects performed the experiment twice. Parameter values from the first and second experiment were not significantly different. The measurement-remeasurement correlation coefficients and the absolute measurement errors are presented in table I. All MFCV values were higher in males, however only the I-MFCV differences between males and females were significant. In fig. 3 the relation between the mean values of the invasive and surface MFCV estimation are given. Note that in all cases the surface measurement gives the highest value. The clear correlation between the MFCV results of both methods was found to be significant (linear regression analysis, $r=0.43$) for both males and females. The ratio between fastest and slowest I-MFCV result (F/S ratio) and the number of spikes per insertion showed no significant differences between the sexes. Plotting the slowest I-MFCV vs. the fastest I-MFCV result (fig. 4), shows a significant positive correlation ($r=0.40$) between the two factors. Table I shows the mean maximal force at elbow flexion for the different groups. The force in the male patient group was significantly higher. The IEMG data showed clearly higher values in the male group as well. When the S-MFCV is plotted against the neuromuscular efficiency (NME), defined as the quotient of force and IEMG, a significant positive correlation is found in males (fig. 5). The S-MFCV increased significantly with increasing age. This was not found with respect to the mean I-MFCV (fig. 6). The maximum force decreased with increasing age. This effect was only significant in males (fig. 7).
DISCUSSION

Invasive MFCV estimation

With the aid of a simple technique it was possible to perform invasive MFCV estimations in all subjects. The measurements were performed within 5 to 10 min, and, despite the invasive nature, without much discomfort. The method is essentially based on the one described by Troni et al. (1983a). However, there are some differences, to minimize the discomfort and to simplify the method. A surface electrode replaced one of the stimulation monopolar needles (the anode). Additionally, instead of a single-fiber EMG needle, a small concentric needle was used. This simplifies the finding of the activated fibers. One drawback might be some loss of selectivity. However, since it is likely that (nearly) all fibers in the stimulated bundle are depolarized simultaneously due to the local stimulation technique, most of the selectivity is achieved by the high pass filtering of the signal (Payan, 1978). Superposition of single fiber potentials is a common finding, irrespective of the needle type used. Buchthal et al. and Meadows also demonstrated the possibility of measuring I-MFCV with a concentric needle electrode, although in a different experimental set-up (Buchthal et al. 1955; Meadows, 1971).

The fact that measured action potentials are the consequence of direct muscle fiber stimulation is supported by the following arguments: (1) Anatomical experiments (Christensen, 1959; Aquilonius et al. 1984) and multi-electrode surface EMG registrations (Masuda et al. 1983) show a well-defined end-plate zone in biceps brachii muscle of 5-10 mm width nearly halfway the muscle. (2) During stimulation with decreasing strength a gradual disappearance of the spikes is

Figure 4. Relation between the slowest and fastest I-MFCV result for both sexes. Solid line: linear regression analysis, r=0.40, p=0.0027.
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found, which argues against direct nerve excitation ("all or nothing" effect). (3) Troni et al. (1983a) have shown that curarization did not influence the results. (4) Measured MFCVs are well within the range as found by others who use different techniques (Arendt-Nielsen and Zwarts, 1989).

Direct muscle activation has some important advantages. The measurements are performed in resting muscle, so only minimal patient cooperation is required. The fibers in a given area are depolarized irrespective of the state of innervation (van der Hoeven et al. 1993) and independent of the level of voluntary recruitment. Most of the recorded muscle fibers lie within a semicircular territory of 0.5 mm radius around the tip of the electrode. In the normal biceps, this area could contain up to 50 muscle fibers, belonging to different MU's (Nandedkar et al. 1988) individual MU's being represented normally by 3-5 muscle fibers within this area, which suggests that the activity captured belongs to maximally 10-15 different MU's.

The I-MFCV and the F/S ratio have the best reproducibility (see table I). However, some sources of error have to be taken into account. (1) A misdirection of the needle electrodes leads to a non-systematic wrong MFCV estimate. For example, a misdirection of $\theta^0$ of both needles in opposite directions 15 mm beneath the skin results in an error of 5%-8%, see also table I. (2) The risk of stimulating a nerve fiber cannot be excluded completely. In my experience, this is always accompanied by a gross muscle twitch without exact relation to the stimulation needle, in combination with a sudden latency shortening. An abrupt disappearance of these fast components is usually seen following a slight reduction of the stimulus strength or a reposition of the stimulation needle. (3) Variations in stimulus strength can result in variations in the point of depolarization.

**Figure 5.** Relation between mean S-MFCV and neuromuscular efficiency (NME), both sexes. Linear regression analysis. Solid line: men, significantly different from zero, $r=0.54$, $p=0.0003$. Dotted line: women, not significantly different from zero.
along a muscle fiber. Increasing the stimulus results in a shift of the depolarization point towards the uptake electrode, and therefore in a shortening of the latency. Since the inverse relation between conduction velocity (CV) and latency (CV = distance/latency), higher MFCV estimates will result. With respect to this source of error, the stimulus strength was standardized in a narrow range, between 1-2 mA. (4) The high-pass filter of 500 Hz could result in small time shifts of the spikes and consequently a lower MFCV. (5) The occurrence of "late" potentials, which are often found in muscle pathology, can vary locally within the muscle, due to non-homogeneous muscle involvement. However, the finding of these potentials is strongly facilitated by the greater uptake area of the concentric needle and slight needle repositions. Theoretically, this will result in a slight bias to a higher F/S ratio.

The results are nearly identical to those of Zwarts (1989), which were found in a pilot study in a small number of subjects. The effects of age (between 20 and 74 years) are relatively low. Although the mean force clearly declines, especially in males (fig. 7), the mean I-MFCV did not significantly change in the age groups tested (fig. 6).

The low F/S ratio is remarkable in all age groups, reflecting the small differences between conduction velocities within a healthy subject. These differences are supposed to be related mainly to fiber diameter (Håkansson, 1956). Nandedkar et al. (1985) used a formula for simulation purposes; the propagation velocity is calculated for each muscle fiber as a linear function of the diameter by the formula

\[
\text{propagation velocity (m.s}^{-1}) = 2.2 + 0.05 (\text{diameter - 25}) \quad (\mu M)
\]

Applying this formula to a fiber diameter range found in a normal male biceps muscle of 30 to 70 \(\mu M\) (Polgar et al. 1973), this results in a MFCV range of 2.5-4.5 m.s\(^{-1}\), F/S ratio 1.8, hence somewhat higher than the experimental results. Buchthal et al. (1955) suggested as cause of the low scatter a synchronizing effect of the simultaneous activation of a muscle fiber bundle. However, there is as yet no evidence for this hypothesis. Alternatively, a nonlinear relation between fiber diameter and conduction velocity must be considered.

**Surface MFCV estimation**

The surface EMG method results in systematically higher MFCV estimates than does the invasive method (fig. 3). The systematic difference between the two methods is also found in literature (Arendt-Nielsen and Zwarts, 1989). This is due to a combination of physiological factors and sources of error, both of which result in higher MFCV estimates. Most important physiological factors: (1) The S-MFCV values are biased by the fastest conducting fibers at a specific force level. At higher force levels, type II MU's, containing higher conducting fibers (Andreassen and Arendt-Nielsen, 1987), are also being recruited. (2) A non-homogeneous fiber type distribution, based on a relatively high number of type II fibers in the superficial areas (Jennekens et al. 1971), adds to this effect. (3) MFCV is stimulus-frequency dependent (Nishizono et al. 1989; Mihelin et al. 1991), at physiological firing rates of 10 to 20 Hz the MFCV increases 10-20% (Stålberg, 1966). (4) The surface measurements are performed during a slight shortening of the biceps muscle. Since MFCV is muscle-length dependent (Arendt-Nielsen et al. 1992; Trontelj, 1993), a higher
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value will result. Moreover, a slight shortening of the muscle during isometric contractions cannot be excluded.

The reproducibility of the S-MFCV is clearly lower than that of the I-MFCV (see table I). The most important sources of error include: (1) A less than optimal electrode location, not parallel to the muscle fibers, always results in an overestimation of the S-MFCV (Sollie et al. 1985). (2) The distance between the active fibers and the electrodes is variable. An increase in radial observation distance results in a strong decrease of signal/noise ratio (Stegeman and Linssen, 1992; De la Barrera and Milner, 1994). Since noise has no phase shift, overestimation of MFCV will result in the extreme cases. (3) Electrical inhomogeneous tissue between the muscle fibers and the recording electrodes sometimes results in signals without a clear time delay, resulting in an

Figure 6. Relation between age and MFCV. Linear regression analysis. Lower trace: mean I-MFCV values. Not significantly different from zero. Upper trace: mean S-MFCV values, significantly different from zero, \( r=0.28, p=0.02 \).
overestimation of the MFCV (Broman et al. 1985; Schneider and Rau, 1991). (4) A finite fiber length results in the appearance of a positive, constant latency peak, which will result in decrease or absence of latency. However, in bipolar recordings these non-moving potentials are largely cancelled (Gootzen et al. 1991).

The S-MFCV increases with increasing age (fig. 6). This could be an effect of hypertrophy in the muscle fibers of the remaining MU’s (van der Hoeven et al. 1993), and hence a consequence of the physiological motor unit loss during aging (Brown et al. 1988; McComas, 1991). The finding of increased MU amplitudes in elderly subjects supports this hypothesis (Campbell et al. 1973). This phenomenon could also possibly explain the sporadic finding of a solitary slow conducting, and probably atrophic fiber in healthy persons (Cruz Martinez, 1989).

The significant positive correlation in men between the NME and the mean S-MFCV, both at maximum voluntary contraction, is remarkable (fig. 5). We suggest that this is a reflection of differences in NME between type I and type II motor units. In biceps brachii the type II motor units have the highest values with respect to muscle fiber diameters, especially in males (Edström and Nyström, 1969; Jennekens et al. 1971; MacDougall et al. 1980) and consequently MFCV (Håkansson, 1956). This is also reflected in the positive correlation between force and MFCV (Andreassen and Arendt-Nielsen, 1987; Zwarts et al. 1988). Additionally, the maximal force of type II fibers may be about four times that of type I fibers (Linssen et al. 1991). On the other hand

![Figure 7. Relation between age and force at maximum voluntary contraction. Linear regression analysis. Solid line: men, significantly different from zero, r=0.32, p=0.04. Dotted line: women, not significantly different from zero](image-url)
there is no indication that the (extracellular) single fiber action potential, which forms the basis of the IEMG, differs between the two fiber types to such an extent (Wallinga-de Jonge et al. 1985). This implies that the highest NME values should be found in type II fibers.

In conclusion, it is clear that the invasive as well as the surface method result in reliable estimates of the MFCV in situ. However, each method seems to have a different area of potential use. The main advantage of the invasive method is the information it gives about the variability of the MFCV in a given muscle. The possibility to measure even non-innervated muscle fibers particularly contributes to its potential usefulness, not only with respect to research applications, but probably also for clinical use. The surface method, on the other hand, is especially suited for research purposes. Its main advantages are, beside its non-invasive character, easy repeatability and the possibility to measure at different force levels or during continuous contractions.

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