The spread of muscle fiber conduction velocity

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Chapter 3

A new surface electromyography analysis method to determine spread of muscle fiber conduction velocities

Abstract

Muscle fiber conduction velocity (MFCV) estimation from surface signals is widely used to study muscle function, e.g., in neuromuscular disease and in fatigue studies. However, most analysis methods do not yield information about the velocity distribution of the various motor unit action potentials. We have developed a new method - the interpeak latency method (IPL) - to calculate both the mean MFCV and the spread of conduction velocities in vivo, from bipolar surface electromyogram (sEMG) during isometric contractions. sEMG was analyzed in the biceps brachii muscle in 15 young male volunteers. The motor unit action potential peaks are automatically detected with a computer program. Associated peaks are used to calculate a mean MFCV and the SD. The SD is taken as a measure of the MFCV spread. The main finding is that the IPL method can derive a measure of MFCV spread at different contraction levels. In conclusion, the IPL method provides accurate values for the MFCV and additionally gives information about the scatter of conduction velocities.

Introduction

The determination of the muscle fiber conduction velocity (MFCV) is widely used in studies of fatigue (van der Hoeven and Lange, 1994) and neuromuscular diseases (van der Hoeven et al., 1993b). The MFCV can be determined by surface as well as invasive methods (Arendt-Nielsen and Zwarts, 1989), which differ in a number of ways. Surface electromyogram (EMG) determination methods are generally based on the determination of the latency of recognizable parts of the motor unit (MU) action potential (MUAP) during voluntary contraction, recorded at different points along the muscle. These latencies can be determined by various methods, e.g., the cross-correlation (CC) method (Nishizono et al., 1979). This method reveals a global value of the MFCV but yields no information about the spread of the conduction velocities (CVs). Some invasive MFCV determination techniques have the advantage to estimate the spread of CVs within the muscle. Despite these advantages, the invasive character is a drawback of these methods. Invasive MFCV
measurements are generally performed in resting muscle: contractions during needle insertion are uncomfortable, and reproducible detection of potentials of the same muscle fiber(s) at different locations is difficult. Determining the CVs of individual MUAPs provides information about aspects of neuronal control, such as recruitment and the relation between firing rate and CV (Morimoto and Masuda, 1984). An increase of CVs can be found in pathological conditions like myopathic and neurogenic lesions (van der Hoeven et al., 1993b, 1994). Hence, detection of the spread of CVs can be helpful in diagnosis. In general, standard surface EMG has a low spatial resolution in detecting potentials on the skin. Rau and co-workers (1997a; 1997) developed a high-resolution surface EMG technique and were able to demonstrate the physiology of individual MUAPs with respect to activation patterns. Prutchi (1995) was able to obtain the spread of MFCV by high-resolution, large-array surface EMG system. Drost et al. (2001) demonstrated localized conduction disturbances of single MUs in myotonia congenita using multichannel surface EMG. However, these techniques require sophisticated and expensive instrumentation and complicated analysis techniques. We designed an analysis method that is based on a two-channel EMG, acquired by inexpensive and common instrumentation. The technique is computational, uncomplicated and based on the estimation of the latencies between defined peaks of the surface EMG. The EMG is recorded in the biceps brachii muscle by two bipolar electrodes, located parallel to the muscle fibers, at different levels of force. Local spread of MFCVs of different MUAPs can be determined by calculating multiple MFCVs. The results were compared with the CC method, calculated on the same signals.

Material and Methods

The experiments were performed on 15 healthy men (aged 25-42 yr, mean 28.5 yr, SD 4.7 yr) who gave their informed consent. None of them used medication. All measurements were performed during isometric contractions of the left biceps brachii muscle.
EMG Recording

The subjects sat in a chair, with their arm fixed in a horizontal, semiflexed position at an angle of 120°, supported at the elbow, with the wrist in the supine position. Three silver electrodes (diameter 2 mm) were placed in a rigid bipolar array with a common center electrode; the interelectrode distance was 10 mm. The skin was abraded and cleaned with ethanol. A small amount of 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. After optimal positioning, the rigid electrode array was fixed at the skin with adhesive tape. The isometric force of the elbow flexion was measured at the wrist. The exerted force was displayed in front of the subject on a light-emitting diode bar and simultaneously digitized and recorded. After filtering from 1 to 300 Hz, the two bipolar EMG signals were amplified. The EMG signals were synchronously digitized by a 16-bit analog-to-digital converter at 20 and 1 kHz over signal periods of 0.205 and 1 s, respectively. The interpeak latency (IPL) method and CC method were applied to the short sampling period.

Determining the CVs and the Spread

IPL method: In both channels, all up-going (i.e., negative) peaks were identified and marked. We defined a negative peak as the derivative of the EMG signal, which has to be zero or to cross the zero line. Before and after this point, the derivative has to be > 0 and < 0, respectively, for at least 15 samples, whereas the EMG signal has to be above the zero line (i.e., negative) at EMG derivative = 0. For each negative and solitaire peak in channel 1, we searched for a corresponding negative and solitaire peak in channel 2 within the defined time window (Fig. 1 - fig. 3). We used an interelectrode distance of 10 mm, and, with the assumption of a physiological MFCV between 1.8 and 6.4 ms⁻¹ (Zwarts, 1989), the corresponding latencies were 5.55 and 1.56, respectively. The time window of 3.99 ms results in combination with the sampling time and frequency (0.205 s, 20 kHz) in a theoretical maximum number of detectable peaks of 51. After selection of peaks and control for assumptions, we determined the latency and resulting CV between each pair. The minimum number of
Figure 3.1: A: interpeak latency (IPL) method; illustration of peak detection. Step 1: recording 2 surface electromyogram (EMG) signals. A plot of these 2 surface EMG signals (represented by a solid and a dotted curve), as recorded by the surface electrode array, is shown. As an example, a time period of 0.1 s is shown, which is 50% of the total measurement time. The pair of electrodes near the end-plate zone generates the solid curve. The dotted curve is measured at 10 mm distal to the proximal pair of electrodes. Step 2a: the derivative of the 2 signals is calculated.
Figure 3.2: B: Step 3: finding the zero crossing of the derivative in both signals. The polarity of the signal is checked at all zero crossings of the derivative (indicated as step 2b). The zero crossing is marked as possible peak. Step 4: in case of a negative (i.e., up going) signal at this zero crossing, the duration of the peak is checked with the following criteria. Step 5: the period before the zero crossing has to be positive for at least 15 preceding samples and negative for at least 15 following samples. The minimal duration of a peak, therefore, is the sum of 15 preceding samples, 1 zero crossing, and 15 following samples. At a given sample frequency of 20 kHz, the duration of a peak is 1.55 ms. This period is indicated as a horizontal bar just beyond the x-axis. ∆, Change; arb, arbitrary.
Figure 3.3: C: IPL method; determination of multiple interpeak latencies after identification of peaks in signals 1 and 2. Solid triangles above 1 signal (▼) and below signal 2 (▲) indicate the peaks. For every peak in signal 1, a solitaire peak in signal 2 is searched within a window. The 2 horizontal lines at the bottom of the shaded triangle indicate the length of the window and its relation to the peak in signal 1. In this example, peaks 1-3 and peaks 6-8 were accepted as solitaire peaks. Peaks 4 and 5 were not solitaire within the window and were, therefore, excluded.

Paired peaks had to be three per measurement to facilitate the calculation of a SD. The mean CV [IPL CV (CV_{IPL})] and the SD were calculated for each measurement based on all acquired CVs per measurement.

**CC:** We used the method as it was introduced by Nishizono et al. (Nishizono et al., 1979). In short, the two EMG signals (20-kHz sampling frequency) were cross-correlated. The peak in the correlogram represents the displacement of the two signals in relation to each other, indicating the time shift between the two signals. Based on the time shift and the interelectrode distance, the MFCV is calculated.
<table>
<thead>
<tr>
<th>Authors</th>
<th>Year</th>
<th>EMG Activation</th>
<th>MFCV, m/s</th>
<th>SD</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prutchi (10)</td>
<td>1995</td>
<td>Surface MUAP detection</td>
<td>3.7</td>
<td>0.7</td>
<td>1.93-5.79</td>
</tr>
<tr>
<td>Prutchi (10)</td>
<td>1995</td>
<td>Surface MUAP detection</td>
<td>4.1</td>
<td>0.7</td>
<td>2.35-6.05</td>
</tr>
<tr>
<td>van der Hoeven et al. (19)</td>
<td>1993</td>
<td>Needle Stimulated</td>
<td>3.5</td>
<td>0.0</td>
<td>2.93-5.25</td>
</tr>
<tr>
<td>Zwarts et al. (21)</td>
<td>1988</td>
<td>Needle Consensus</td>
<td>3.99</td>
<td>0.1</td>
<td>2.8-4.1</td>
</tr>
<tr>
<td>Stålberg (14)</td>
<td>1966</td>
<td>Needle</td>
<td>3.25</td>
<td>0.23</td>
<td>2.83-3.72</td>
</tr>
<tr>
<td>Trontelj and Stålberg (16)</td>
<td>1983</td>
<td>Needle Stimulated</td>
<td>3.02</td>
<td>0.1</td>
<td>2.93-4.25</td>
</tr>
<tr>
<td>Present study (20% MVC)</td>
<td>2001</td>
<td>Surface IPL Voluntary</td>
<td>4.1</td>
<td>0.32</td>
<td>3.53-4.81</td>
</tr>
<tr>
<td>Present study (50% MVC)</td>
<td>2001</td>
<td>Surface IPL Voluntary</td>
<td>3.15</td>
<td>0.38</td>
<td>2.85-4.15</td>
</tr>
</tbody>
</table>

Table 3.1: MFCV in human biceps brachii muscle: individual measurements and standard deviations.
Procedure

All measurements were performed during contractions of 1.5 s. Each subject performed a maximum voluntary contraction (MVC) twice. The highest value was accepted as the maximum isometric force, which formed the basis to calculate the levels of relative force. Afterward, the subjects were asked to perform two contractions at each level of force; the CV was calculated as the average of both measurements. The contraction sequence was 100, 50, 10, 90, 20, 80, 30, 70, 40, and 60% of MVC.

Statistics

We used the ANOVA repeated-measurement analysis to determine the significance of changes in mean MFCV and the spread of the MFCV (i.e., the SD of the mean MFCV). The relation between the CC method and IPL method was investigated by linear and nonlinear regression analysis. Also, we calculated the correlation coefficient between the spread and the number of peaks per measurement. The changes of MFCV, spread, and count of peaks at different levels of force were summarized and quantified by linear and nonlinear regression analysis. Statistical significance was accepted at a level of 5%.

Results

MFCV Estimation

It was possible to identify a sufficient number of paired peaks to calculate the mean MFCV and its spread in all subjects. However, in 2 of the 300 measurements (0.7%), the IPL method could not identify three paired peaks at a low level of force because of a combination of low-amplitude signal and superposition of noise. The number of peaks was increasing gradually from 10 to 50% of MVC (7.6 ±5.4 to 19.2 ±5.4 per measurement). From 60% up to MVC, the number of peaks remained constant (Fig. 4). The CV_{IPL} and CC method CV (CV_{CC}) correlated well.

Nonlinear regression analysis revealed a positive correlation, with a slightly better fit than linear regression analysis (Fig. 5). The MFCV
Figure 3.4: Number (n) of detected peaks at different levels of force. Values are means SE (symbols, mean values; bars, SEs); n = 298 measurements. Note gradual increase of the no. of peaks with force up to 40% maximum voluntary contraction (MVC) and stable no. of peaks beyond this level of force. Linear regression analysis: goodness of fit, $r^2 = 0.697$; slope, $0.12 \pm 0.028$. The slope is different from 0 ($P < 0.05$).
Figure 3.5: Conduction velocities (CVs): cross-correlation (CC) vs. IPL method. Nonlinear regression analysis; \( n = 298 \) measurements; goodness of fit, 0.774; doubling time, 3.131. Note deviation of the curve from the \( x = y \) line at low CVs, where the CC method reveals higher values compared with the IPL method. The slope of the linear regression line was 0.93 (95% confidence interval: 0.87-0.99). Note influence of 4 measurements at low CVs on nonlinear as well as linear regression. MFCV, muscle fiber CV. Solid line, nonlinear regression; dashed line, equality \( x = y \).
Figure 3.6: CV [CC method (CV_{CC}) and IPL method (CV_{IPL})] vs. force. Values are means ±SE; n = 298 measurements. Note the difference (not significant) between the CC method and the IPL method at MVC. F_{max}, maximum force production.

increased with force up to 60% MVC for both methods and reached a plateau between 60 and 90% MVC. At MVC, the MFCV decreased (Fig. 6). We found no significant differences in calculated MFCV between both methods at specific levels of force.

Spread of MFCV

The spread has a normal (i.e., Gaussian) distribution (Fig. 7). We found an exponential association of the scatter in relation to the force level. From 10 to 50% of MVC, the spread (expressed in SD of MFCV) increased quickly, and from 60 to 100% MVC, it remained stable (Fig. 8). Additionally, the relation between the number of detected peaks and the
Figure 3.7: Distribution of spread. The spread is the SD of the mean MFCV. All measurements at all levels of force were included. The class width (x-axis) is 0.1 ms$^{-1}$. Note Gaussian (i.e., normal) distribution of the spread.
Figure 3.8: Spread of CV at different levels of force; nonlinear regression analysis (curve). Values are means ±SE; \( n = 298 \) measurements. Goodness of fit: \( r^2 = 0.99 \); half-life = 12.
SD of MFCVs of each measurement was investigated. We found no correlation between the spread and the number of detected peaks [correlation coefficient (Spearman r) = -0.19; goodness of fit ($r^2$) < 0.05], indicating that the spread (in SDs) is independent of the number of detected peaks.

**Discussion**

The aim of the present study was to find a parameter that reflects the spread of the muscle fiber CVs from a surface EMG signal. The first step was to design a method of analysis that is capable of calculating CVs of individual MUAPs. The peaks in the surface EMG signal were chosen as an easily recognizable element, suited for automated detection. In all subjects, it was possible to calculate the mean and the spread of MFCV by the IPL method. The $CV_{IPL}$ correlated well with the $CV_{CC}$ at all force levels (Fig. 6). We, therefore, conclude that the IPL method is an accurate method to calculate the CV within the physiological range from 1.8 to 6.5 m/s. A methodological difference between the IPL and CC methods is the quantity of signal used for calculations. In the CC method, the whole signal period is used, whereas the IPL method takes into account only small parts at a time (i.e., the peaks). Theoretically, the IPL method could be more susceptible to random errors because of a low similarity between the proximal and distal signal. In the present study, we found no correlation between the CC coefficient (i.e., similarity of the signals) and the difference, $CV_{IPL} - CV_{CC}$. This suggests that the random error for both methods is the same. The spread of MFCV (surface EMG) during voluntary contractions can be expressed by reporting lowest and highest values, as a percentage of the mean MFCV or as the SD of the mean MFCV. In the present study, we have chosen to use the SD of MFCV to cancel out possible outliers (single extreme high or low values), which are not representative of the global spread. Invasive methods of MFCV determination [e.g., after direct muscle fiber stimulation (Arendt-Nielsen and Zwarts, 1989; van der Hoeven et al., 1993b) are based on detection of MFCV differences between single muscle fibers. In these cases, even conduction differences between innervated and denervated muscle fibers can be detected. However, the use of surface EMG (and surface MFCV
detection) is limited to voluntary recruited muscle fibers. Hence, only conduction differences between single MUAPs can be detected. Nevertheless, comparing with invasive determination techniques seems useful, because the limits of CV scatter in resting muscle are set.

The spread of CV that we found at force levels up to 60% MVC was very close to the data found by Schneider et al. (multichannel surface EMG, voluntary contraction) (1989), Graham et al. (surface EMG, zero-crossing technique) (1984), and Stålberg (invasive EMG, voluntary contraction) (1966) (Table 1). The values during voluntary contractions are mostly derived at levels of force when clear identification of single MUAPs is possible (Stålberg and Ekstedt, 1968). In practice, that means almost always low force levels. At higher levels of force, the spread is difficult to measure because of the interference pattern, which makes detection of individual MUAPs impossible. A way of solving this problem is by using high-spatial filtering, e.g., with a small interelectrode distance in combination with highpass filtering. Prutchi (1995) described such a method and found in one subject a mean MFCV of 4.37 ms$^{-1}$ (range: 3.3-6.1 ms$^{-1}$ at MVC) in biceps brachii muscle. In comparison, we found a mean MFCV of 4.26 ms$^{-1}$ (SD: 0.6, range: 3.06-5.46 ms$^{-1}$). Huppertz et al. (1997b) found a range of velocities of 2.8-4.1 ms$^{-1}$, using high-resolution surface EMG recordings at MVC in small hand muscles. These values are relatively low compared with our findings, which is probably due to a combination of the muscle type (small hand muscle with generally smaller muscle fibers) and the level of force.

A comparison with invasive studies can be based on experiments with the use of voluntary contractions and measurements with the use of direct stimulation of muscle fibers. Stålberg (1966) found, during voluntary contractions, a MFCV of 3.69 ms$^{-1}$ (SD: 0.71 ms$^{-1}$). The SD that he found is slightly higher than what we estimated. This could be caused by the higher representation of (fastconducting) type II fibers at the surface layers of the biceps brachii muscle, which will influence surface EMG more than invasive EMG techniques. Van der Hoeven et al. (1993a) studied MFCV using direct muscle stimulation (1 Hz). They found a mean MFCV of 3.23 ms$^{-1}$ (SD 0.21 ms$^{-1}$, range 2.81-3.65 ms$^{-1}$), which is clearly lower than our results during voluntary muscle activation. This difference is probably due to the dependency between MFCV and stimulation
rate (Sadoyama and Masuda, 1987; Trontelj and Stålberg, 1983). However, because not only the absolute MFCV but also the scatter increases, a nonlinear relation between CV and stimulation rate must be assumed, in which fast-conducting fibers have a higher increase with increasing firing rate. Force production is caused by the simultaneous effect of recruitment and rate coding. On lower force levels (< 50% MVC), recruitment is the most important factor. On higher force levels, after recruitment of all MUs, rate coding is the remaining cause of additional force. The change of spread can be interpreted according to these basic principles. We suggest that the steep increase between 10 and 40-50% MVC is (mainly) caused by the effect of recruitment of succeeding larger MUs (size principle). The later recruited, larger MUs are composed of muscle fibers with higher diameters and consequently higher MFCV values (Arendt-Nielsen and Zwarts, 1989; Jennekens et al., 1971a,b). At these force levels, the slower conducting MUs are still detectable as peaks in the surface EMG signal, adding to the measured spread. From 50-60% up to MVC, we found no further increase in the spread. We suggest that the reason for the stable spread beyond 50% MVC is the fact that (nearly) all MUs have been recruited. In that case, a substantial further increase in scatter is unlikely.

Theoretically, the mentioned dependency between firing rate and MFCV could also influence the scatter in CVs. Assuming a combination of coexisting physiological firing rates (between 8 and 25 Hz), an early recruited fast firing (slow) MU will be recruited together with a slow firing (fast) MU, resulting in a decreased scatter.

The same tendency could result from high firing rates, which will result in the repeated detection of the same MUAP, depending on firing rate and acquisition time. For example, a 25 Hz firing rate during 0.205 s could theoretically result in a fivefold detection of the same MUAP, which, consequently, will influence the scatter (SD of MFCV IPL). Although this would argue for a reduction in sampling time, an advantage of a longer sampling time could be an increased chance of detecting low-amplitude MUAPs within the interference pattern, assuming unsynchronized firing.

An alternative approach to detect conduction differences out of the surface EMG could be based on repeated CC estimation on EMG segments in a moving window of fixed length (Duchene and Hogrel, 1997).
This, theoretically, would result in a statistical estimate of the scatter. However, the IPL technique measures the peaks in the signal, formed by the propagating MUAPs, which connect it more directly to the underlying physiological phenomena.

In conclusion, we found that the surface EMG IPL method provides accurate values for the MFCV and, additionally, gives information about the spread of CVs, even at higher levels of force.
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