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Communication

Computing motor unit number index of the first dorsal interosseous muscle with two different contraction tasks

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ABSTRACT

Motor unit number index (MUNIX) is a recently developed novel neurophysiological technique providing an index proportional to the number of motor units in a muscle. The MUNIX is derived from maximum M wave and voluntary surface electromyogram (EMG) recordings. The objective of this study was to address a practical question for computing MUNIX in the first dorsal interosseous (FDI), a multifunctional muscle that generates torque about the second metacarpophalangeal joint, i.e., how will different lines of muscle activation influence its MUNIX estimates? To address this question, the MUNIX technique was applied in the FDI muscle of 15 neurologically intact subjects, using surface EMG signals from index finger abduction and flexion, respectively, while the maximum M wave remained the same. Across all subjects, the average MUNIX value of the FDI muscle was 228 ± 45 for index finger abduction, slightly smaller than the MUNIX estimate of 251 ± 56 for index finger flexion. Different FDI muscle activation patterns resulted in an approximately 10% difference in MUNIX estimates. The findings from this study suggest that appropriate definition of voluntary activation of the FDI muscle should be kept to ensure consistency in measurements and avoid source of error. The current study is limited by only assessing neurologically intact muscles. It is important to perform a similar analysis for patients with amyotrophic lateral sclerosis (ALS), given that ALS is the primary intention of the MUNIX method as a potential follow-up measurement for motor unit loss.

1. Introduction

Since the introduction of the motor unit number estimation (MUNE) [10], the technique and its various forms of improvement have been used as an important tool in basic and clinical neurophysiology [2]. The traditional MUNE methods involve estimates of single motor unit action potential size using either incremental nerve stimulation or spike triggered averaging techniques, both potentially laborious and time-consuming. Thus, a motor unit number index (MUNIX) technique was developed [11]. The MUNIX measurement is based on surface electromyogram (EMG) and maximum M wave or compound muscle action potential (CMAP) recordings, which induces minimal discomfort and can be performed quickly. Because of these practical conveniences, the MUNIX measurement has attracted increasing attention. A number of its applications have been reported recently, focusing on detecting motoneuron loss and measuring disease progression in amyotrophic lateral sclerosis (ALS) and other neuromuscular disorders [1,6,7,12–17].

The MUNIX technique requires recording of voluntary surface EMG signals for different muscle contraction levels. While "simple" muscles exert forces about a single degree of freedom joint, the body has many multifunctional muscles capable of producing joint torques in more than one direction. For example, biceps brachii, deltoid, and the interossei muscles can exert forces in multiple directions about their respective joints. Although the MUNIX measurement is convenient to implement and technically no major difficulties are anticipated, there is a practical question when computing MUNIX in multifunctional muscles, i.e., how will different lines of force generation influence the MUNIX estimates?

To answer this question, the objective of the current study was to assess directional dependence of MUNIX calculation in the first dorsal interosseous (FDI), a multifunctional muscle that generates torque about the 2nd metacarpophalangeal (MCP) joint. The plane of force production about the MCP joint consists of linear combinations of flexion, extension, abduction, and adduction of the index finger. The FDI muscle is a primary abductor and synergistic flexor about the MCP joint. These two lines of muscle action were used

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to determine whether there are systematic differences in MUNIX estimates of the FDI muscle according to the direction of force generation at the index finger.

2. Methods

2.1. Subjects

Fifteen neurologically intact subjects (7 males, 8 females, 42.0 ± 13.5 years) participated in this study. The study was approved by the Institutional Review Board of Northwestern University (Chicago, IL, USA). All the subjects gave their written consent before the experiment.

2.2. Experimental protocol

The experiments were performed in the FDI muscle of the dominant hand in all the subjects. Subjects were seated comfortably in a chair with the examined forearm placed in its natural, resting position on a height-adjustable table. They were instructed to relax, and the hand and forearm were held in a vertical half supinated position.

Prior to the recording, each subject's skin over the ulnar aspect of the wrist, the back of the hand, and the index finger were slightly abraded and cleaned with rubbing alcohol. To begin, the maximum M wave or CMAP was first recorded. The primary equipment used for this recording was a Sierra Wave EMG system (Cadwell Lab Inc., Kennewick, WA, USA). Electrode placement was similar to that for standard ulnar motor studies. The active surface electrode (10 mm silver/silver chloride disc) was positioned over the motor point of the FDI muscle with the reference surface electrode positioned over the second MCP joint. An adhesive ground electrode was placed on the back of the hand.

The ulnar nerve was stimulated about 2 cm proximal to the wrist crease with a remote handheld stimulator. The intensity of the stimulation pulse (200 μs in duration) started around 15–20 mA. The intensity was further increased in increments of approximately 20% above that until the stimulation intensity eliciting the maximal response was reached. Then, the stimulation intensity was increased to 120% of the final intensity to confirm no further enlargement in the peak-to-peak amplitude of the M wave. To ensure that the M wave amplitude is maximized and meanwhile the positive take-off (which is often created in FDI muscle M wave recording [26] and makes it difficult to define onset latency) is minimized, the electrode placement was optimized by testing several different locations. In addition, during the experiment re-cleaning of the skin and re-application of the electrode cream were performed as necessary to guarantee the best recording quality. The maximum M wave was recorded only once, and this maximum M wave was used for calculation of MUNIX with different voluntary muscle contraction tasks.

After the maximum M wave recording, with the EMG electrode maintained at the same position, the voluntary surface EMG signals were recorded from the FDI muscle while the subject generated an isometric contraction in two different directions (i.e., index finger abduction or flexion). For each direction, the different force levels were recorded using a single trial with graded contraction consisting of 5–10 interference EMG epochs representing minimal to maximal effort. A resistant force was provided to subjects by the examiner to ensure isometric voluntary contraction of the FDI muscle. For each direction this protocol was performed two times. Subjects were allowed substantial rest to avoid muscle fatigue during the recording. For all the subjects, the surface EMG was sampled at 2 kHz, with a band-pass filter setting at 10–500 Hz.

2.3. Data analysis

The maximum M wave and different levels of surface interference pattern (SIP) EMG were used to compute the MUNIX for the FDI muscle. The MUNIX derivation was described in detail elsewhere [11,12] and its procedures are outlined in brief here.

The area and power of the maximum M wave and different levels of SIP signals were first computed. The M wave onset and offset were defined from the baseline, and the tiny positive take-off part was not included for calculating the maximum M wave area and power. The area and power of the maximum M wave and each level of SIP EMG were then used to compute the "ideal case motor unit count (ICMUC)" defined as the ratio of maximum M wave power to its area multiplied by the ratio of the SIP area to its power. Thus, each level of SIP gave two results: SIP area and ICMUC. Regression analysis was then used to define the relationship between the SIP area and the ICMUC by an exponential fitting: $\mathrm{ICMUC} = \beta(\mathrm{SIP\ Area})^{\alpha}$. The parameters β and α can be obtained from the regression using different levels of SIP. For each direction up to 20 different levels of SIPs (from combination of both trials) were used for this regression analysis. Finally, the SIP area of 20 mV ms was used to compute the MUNIX value from the established exponential fitting. The rationale for this selection was described by Nandedkar et al. [12]. To exclude the abnormally high MUNIX estimate induced by very low amplitude surface EMG signals (which may give very high ICMUC values), three criteria were imposed to accept an SIP epoch [11,12]: (1) SIP area > 20 mV ms; (2) ICMUC < 100; and (3) SIP area/CMAP area > 1.

We measured the MUNIX values in the FDI muscle using voluntary surface EMG signals from index finger abduction and flexion, respectively, while the maximum M wave remained the same. We examined whether the MUNIX estimates were significantly different from varying directions of muscle activation. A dependent Student's *t* test was used for statistical analysis. The significant level was defined as $p < 0.05$.

3. Results

Across all subjects, the maximum M wave amplitude from the FDI muscle was 14.1 ± 2.0 mV. Maximum M waves, in combination with voluntary surface EMG signals, were used to calculate the MUNIX measurement.

Fig. 1 demonstrates an example of the MUNIX calculation from the FDI muscle using index finger abduction and flexion, respectively. Fig. 1a shows the maximum M wave; Fig. 1b shows the surface EMG signals when muscle contraction force was increased from minimal to maximum during index finger abduction and flexion, respectively. The comparison of the MUNIX calculations from the two directions is presented in Fig. 1c. Analysis of SIP measurements (the individual data points in Fig. 1c) shows an excellent fit with the mathematical model used to calculate the MUNIX (the lines representing exponential fitting). It is noted that the voluntary surface EMG generated by the FDI muscle flexion was relatively small compared with that from the abduction, as indicated by the *x*-axis values of the individual data points used for the curve fitting. With the measured maximum M wave and different levels of SIP signals, this subject showed a MUNIX value of 228 for the FDI muscle abduction mode, and a MUNIX value of 265 for the flexion mode.

For all the subjects, exponential regression analysis showed a good fitting for the relationship between SIP area and ICMUC. As indicated in Fig. 2, across all subjects we observed an approximately 10% difference in MUNIX estimates using FDI muscle abduction or flexion mode. Fifteen subjects showed a MUNIX value of 228 ± 45 for the FDI abduction mode, which was slightly lower than the MUNIX value of 251 ± 56 for the FDI flexion mode ($p < 0.05$).