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Within-subject reliability of motor unit number estimates and quantitative motor unit analysis in a distal and proximal upper limb muscle

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Abstract

Objective: To establish within-subject reliability of motor unit number estimates (MUNEs) and quantitative MU analysis using decomposition-based quantitative electromyography (DQEMG).

Methods: Following the acquisition of a maximum M-wave, needle and surface-detected EMG signals were collected during contractions of the first dorsal interrosseous (FDI) and biceps brachii (BB). DQEMG was used to extract motor unit potential (MUP) trains and surface-detected MUPs associated with each train, the mean size of which was divided into the maximum M-wave to obtain a MUNE. Retests were performed following the initial test to evaluate reliability.

Results: Subjects test-retest MUNEs were highly correlated ($r=0.72$ FDI; 0.97 BB) with no significant differences between test and retest MUNE values ($P>0.10$). Ninety-five percent confidence intervals were calculated to establish the range of expected retest MUNE variability and were ± 41 MUs for the FDI and BB. Quantitative information pertaining to MU size, complexity and firing rate were similar for both tests.

Conclusion: MUNEs and quantitative MU data can be obtained reliably from the BB and FDI using DQEMG in individual subjects.

Significance: Establishing within-subject reliability of MUNEs and quantitative MU analysis allow clinicians to longitudinally follow changes in the MU pool of individuals with disorders of the central or peripheral nervous system in addition to assessing their response to treatments.

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1. Introduction

Motor unit number estimates (MUNEs) and quantitative motor unit (MU) analysis performed using decomposition-based quantitative electromyography (DQEMG) provide clinically useful information pertaining to the physiological characteristics of individual MUs and the size of the underlying MU pool within a given muscle group. Taken together, this information may enable clinicians to better characterize the extent of MU loss and subsequent reorganization of the MU pool in response to disorders of the central and peripheral nervous systems, and to follow the natural history and response to treatment of these disorders.

To provide value as a clinical tool, the results obtained with DQEMG must be reliable so that changes from test to test may be interpreted as resulting from some process other than the imprecision of the technique. Previously, DQEMG derived MUNEs of the thenar muscle group were shown to be highly reliable across a population of healthy younger adults while other studies using these methods have attempted to characterize the numbers and characteristics of MUs within varying muscles (Boe et al., 2004; Doherty and Stashuk, 2003; McNeil et al., 2005a). While these studies have contributed significantly to the advancement of the technique, they have not adequately addressed clinical

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reliability as they have focused on the reliability of the method for a population, as opposed to the expected variability on test-retest for any one subject.

Consequently, it is essential to examine the reliability of the data obtained using DQEMG within individual subjects so that it may be used longitudinally to follow changes within a given subjects MU pool. In order to achieve this, a range of MUNE values must be determined that takes into account the degree of variability associated with the DQEMG technique. In a manner similar to those employed previously (Simmons et al., 2001), the calculation of confidence intervals for an individual subject's predicted MUNE value has been used to allow for the identification of those changes to the MU pool that fall outside of those expected to occur due to methodological variability, and therefore, result from disorders of the central or peripheral nervous system.

Thus, the purpose of the current study was to determine the within-subject reliability of MUNEs and quantitative MU analysis for the first dorsal interosseous (FDI) and biceps brachii (BB) muscles performed using DQEMG, and to establish 95% confidence intervals around the predicted MUNE values of individual subjects. The FDI and BB muscles were chosen due to their potential differences in force production strategies and the accessibility of their nerve supply, which allows for the acquisition of a maximum M-wave and subsequent calculation of a MUNE. Lastly, it was important to establish the reliability of MUNEs and quantitative MU analysis in muscles that represent different segments of the cervical cord, particularly for future studies of patients with motor neuron disease who may present with disease onset in different segments.

2. Methods

2.1. Subjects

Ten healthy subjects aged 27 ± 6 years volunteered to take part in the study. All gave informed consent and our institutional review board approved the study.

2.2. Force measurement

For the FDI muscle, subjects were seated during data collection with their right arm pronated and placed in a custom-made force dynamometer. In order to isolate the action of the FDI muscle, the thumb was stabilized with a metal brace at 90° extension and the lateral three digits separated from the second digit with a divider, and immobilized with a medium density sponge placed over the digits and secured with a Velcro strap. Additional straps placed just distal and proximal to the wrist joint line secured the forearm and hand position. The isometric abduction force exerted by the FDI was measured in Newtons (N) with a force transducer (Model FT-10; Grass-Telefactor, West-Warwick, RI) that was anchored to the device and aligned with the proximal interphalangeal joint of the second digit. The output from the force transducer was amplified (Model CP 122 AC/DC Amplifier; Grass-Telefactor, West-Warwick, RI) and converted to digital format by a 12-bit converter (CED model 1401 Plus, Cambridge Electronic Design, Cambridge, UK) at a sampling rate of 500 Hz and displayed on an analog oscilloscope (Model 5111A storage oscilloscope; Tektronix Inc., Beaverton, OR) placed in front of the subject.

The protocol used to measure BB force output is similar to a previously reported study (Klein et al., 2001). Subjects were supine on a padded table and the right arm placed in a custom-made force dynamometer. The legs were supported on a padded wooden box, with the hip and knee joints flexed to 90° and the right shoulder secured with a padded metal brace. The box and brace prevented the torso from sliding during contractions. The elbow joint was flexed 90° and placed in a padded cup with the forearm fully supinated. The wrist and fingers were prevented from flexing during contraction by a plastic splint that was strapped to the back of the wrist and hand. The ventral aspect of the wrist was secured with a strap to a padded curved bar (11×5.2 cm) that had a strain gauge attached (model SST-700-100A, ASTechnology, Haliburton, Ont., Canada). The output from the strain gauge was amplified (Neurolog, models NL 107, and NL 126, Digitimer, Welwyn Garden City, Hertfordshire, UK), and converted to digital format by a 12-bit converter (CED model 1401 Plus, Cambridge Electronic Design, Cambridge, UK) at a sampling rate of 500 Hz and displayed on an analog oscilloscope (Model 5111A storage oscilloscope; Tektronix Inc., Beaverton, OR) suspended above the subject.

The force signals for both the FDI and BB were analyzed off-line using a commercially available software package (Spike 2 v. 4.5; Cambridge Electronic Design, Cambridge, UK).

2.3. Electromyographic data collection

The DQEMG method and associated algorithms as described in detail elsewhere were used (Boe et al., 2004; Doherty and Stashuk, 2003). Electromyographic signals were acquired using DQEMG software on the Neuroscan Comperio (Compumedics Medical Systems, El Paso, TX). Intramuscular signals were recorded with a commercially available, disposable concentric needle electrode (Model N53153; Teca Corp., Hawthorne, NY) with a bandpass of 10 Hz–10 kHz, while surface signals were recorded with a bandpass of 5 Hz–5 kHz using self-adhering electrodes (Kendall-LTP, Chicopee, MA). For the FDI muscle, a full size electrode was cut in strips ($1 \text{ cm} \times 3 \text{ cm}$) and the active electrode located over the motor point of the muscle with the reference electrode located over the first metacarpophalangeal joint. For the BB muscle, full size electrodes ($2 \text{ cm} \times 3 \text{ cm}$) were used with the active electrode located

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