Journal of Neuroscience Methods 195 (2011) 10–14

Contents lists available at ScienceDirect

Journal of Neuroscience Methods

journal homepage: www.elsevier.com/locate/jneumeth

Coherence between surface electromyograms is influenced by electrode placement in hand muscles

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ARTICLE INFO

Article history:
Received 24 August 2010
Received in revised form 18 October 2010
Accepted 19 October 2010

Keywords:
EMG–EMG coherence
Motor unit
Common modulation
Cross-correlation
Hand
Pinch grip
Synchronization

ABSTRACT

We used multi-channel surface electromyograms (EMGs) to examine if electrode location influences coherence measures derived from pairs of EMGs recorded from two hand muscles during a pinch task. A linear probe of 16 electrodes was used to estimate the location of the innervation zone in first dorsal interosseous (FDI) and abductor pollicis brevis (APB). Four electrodes were then placed on the skin overlying each muscle and three bipolar electrode configurations were constructed with their center points directly over the innervation zone, and 15 mm distal and proximal to the innervation zone. Ten subjects performed two force-matching tasks for 120 s at 2 N and 3.5 N by pressing a force sensor held between the thumb and index finger. Coherence spectra were calculated from pairs of EMGs recorded from the two muscles. Maximal coherence from 1 to 15 Hz and 16 to 32 Hz was calculated at both force levels from the EMGs with electrodes centered over the innervation zones of FDI and APB. These values were compared to the maximal coherence from all other EMG comparisons across muscles recorded with electrodes that avoided the innervation zones. ANOVA revealed significant main effects only for electrode location, with a 58.1% increase ($p = 0.001$) in maximal coherence for EMGs detected from pairs of electrodes that avoided the innervation zone (from 0.11 ± 0.02 to 0.18 ± 0.03; mean ± 95% confidence interval). These results indicate that electrode location relative to the innervation zone influences EMG–EMG coherence and should be carefully considered when placing EMG electrodes on hand muscles.

1. Introduction

Coherent oscillations in the synaptic inputs to motor neurons are thought to result in the periodic modulation of motor unit discharge rates (i.e., motor unit coherence), though the functional significance of the rhythmic activity to motor control is not clear (Baker et al., 1999; Kilner et al., 2003; Semmler, 2002). One potential confound to determining the functional significance of the rhythmic activity is related to the technique commonly used to identify the rhythmic activity. Specifically, coherence analysis between two signals is routinely used in neurophysiology to estimate the magnitude of the linear correlation between specific frequency components in the two signals (Amjad et al., 1997; Christakos, 1997; Halliday et al., 1995). Although coherence measures in humans are most commonly derived from the discharge times of pairs of simultaneously active motor units (Farmer et al., 1993; Kakuda et al., 1999; Semmler et al., 2003), this may be problematic as pair-wise correlations between the discharge times of two neurons may not accurately reflect correlated activity across entire populations of neurons (Schneidman et al., 2006).

More recently, coherence measures derived from pairs of surface electromyograms (EMGs) recorded over synergist muscles (i.e., EMG–EMG coherence) have been used and are thought to provide a more representative measure of rhythmic activity across muscles (Baker et al., 1999; Farmer et al., 2007; Kilner et al., 1999). Two primary lines of evidence support the use of EMG–EMG coherence as a population measure of rhythmic activity. The first line of support is mathematically derived (Amjad et al., 1997; Halliday et al., 1995; Myers et al., 2003; Stegeman et al., 2010; Williams and Baker, 2009), including modeling approaches that suggest that the surface EMG can be used to reflect the common modulation in the discharge rates of motor units within and across muscles. The second line of support is derived from the positive associations demonstrated experimentally between measures of coherence calculated from different neurophysiologic signals, including EMGs, electroencephalograms (EEGs), magnetoencephalograms (MEGs), and cortical neuron and motor unit discharge times (Baker et al., 1999; Halliday et al., 1998; Hansen et al., 2002; Kilner et al., 1999). For example, Baker et al. (1999) found that coherence between cortical neurons and EMGs was modulated in a task-dependent manner from 20 to 30 Hz in primates performing precise hand tasks and that EMG–EMG coherence from intrinsic hand and forearm

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doi:10.1016/j.jneumeth.2010.10.018

muscles was modulated similarly in humans performing the same task.

Nonetheless, the surface EMG can be influenced by many different factors independent of neuromuscular activity (Farina et al., 2004b; Keenan et al., 2007) which will influence its sensitivity as a measure of motor unit activity. Although previous work has focused on how signal processing (e.g., rectification) may influence EMG-derived measures of coherence (Myers et al., 2003; Neto and Christou, 2010; Stegeman et al., 2010; Yao et al., 2007), the influence of electrode location on coherence measures has not been systematically addressed. In contrast to a number of studies that highlight the importance of avoiding the innervation zone when placing EMG electrodes (Beck et al., 2008; Merletti et al., 2003; Rainoldi et al., 2004; Roy et al., 1986), many studies report placement of electrodes on the muscle belly where the innervation zone is likely to be located (see review: Mesin et al., 2009). This problem is likely compounded in hand muscles, where short fiber lengths and an innervation zone usually located in the middle of the muscle (Keenan et al., 2005; Saitou et al., 2000) make it difficult to avoid the innervation zone when placing surface electrodes. The development of multi-channel EMG systems has allowed non-invasive methods to estimate the location of the innervation zone (Lapatki et al., 2006; Mesin et al., 2009; Saitou et al., 2000). The purpose of this study was to use multi-channel surface EMGs to examine if electrode location relative to the estimated location of the innervation zone influences EMG–EMG coherence measures during a pinch-grip task.

2. Materials and methods

2.1. Subjects

Ten healthy, young adults (5 women, 5 men; 24.4 ± 1.3 years (mean ± SD); range: 23–27 years old) with no known neuromuscular disorders volunteered to participate in this study. All subjects were right-hand dominant as assessed by the Edinburgh Handedness Inventory (Oldfield, 1971). The Institutional Review Board at the University of Wisconsin-Milwaukee approved all procedures and subjects gave their written formal consent before participating in the study.

2.2. Experimental procedure

We first identified the innervation zone and muscle fiber direction using multi-channel surface EMG (EMG-USB; OT Bioelettronica, Torino, Italy). Surface EMG signals were recorded from first dorsal interosseous (FDI) and abductor pollicis brevis (APB) muscles of the non-dominant left hand. A reference electrode was placed on the left wrist and the subject's skin was prepared using abrasive paste (Spes Medica, Italy). A probe consisting of 16 silver electrodes (2.5-mm inter-electrode distance) (SA16/5, OT Bioelettronica, Torino, Italy) was used to identify the innervation zone by visual analysis of the 15 bipolar EMG signals generated from consecutive electrodes during a series of test contractions of the FDI and APB muscles (Fig. 1A). As commonly done (Merletti et al., 2003), the probe was positioned over each muscle and the orientation of the probe was adjusted to clearly identify propagation of the motor unit action potentials in both directions from the innervation zone (Fig. 1B). The innervation zone was then estimated as the EMG signal that was closest to the location where motor unit action potentials started to propagate in both directions and a clear change in the polarity of the phases of the potential was present (Mesin et al., 2009; Saitou et al., 2000); this location was marked on the skin over both muscles (Fig. 1A and C). The direction of the probe was also marked on the skin to estimate muscle fiber direction (Fig. 1A and C).

After estimating the location of the innervation zone, four surface EMG electrodes (4-mm diameter, silver–silver chloride; 15 mm inter-electrode distance) were placed on the skin overlying both the FDI and the APB muscles (shown only for APB in Fig. 1C). Three bipolar EMG recording configurations were constructed from consecutive pairs of electrodes placed in line with muscle fiber direction. Two electrodes (Fig. 1C, electrodes 2 and 3) were placed with their center point over the innervation zone. The other two electrode configurations had their center points positioned 15 mm distal (Fig. 1C, electrodes 1 and 2) and proximal (Fig. 1C, electrodes 3 and 4) to the innervation zone. A common ground electrode (4-mm diameter, silver–silver chloride) was placed on the head of the ulna on the dorsal surface of the hand. The surface EMG signals were amplified (1 K) and band-pass filtered (13–1 kHz) using an isolated bio-amplifier (Coulbourn Instruments). EMG signal quality was checked before and after each experiment by checking the baseline level of noise at rest, as well as having subjects perform a brief 3–5 s maximal contraction of each muscle (index finger abduction for FDI and thumb abduction for APB) while the experimenter provided manual resistance (Keenan et al., 2009).

The subjects were seated on a chair with the left arm resting on a vacuum foam pad (VersaForm pillow, Tumble Forms). We asked participants to perform force matching tasks for 120 s at 2.0 and 3.5 N while holding a uniaxial force sensor (model ELFS-B3-10; Entran) between the thumb-tip and radial side of the distal phalanx of the index finger (i.e., key pinch). These low force levels were chosen to replicate previous work using pinch grip tasks (Kilner et al., 1999). Subjects were asked to slightly extend the other three fingers not involved in the task so that they could not assist the index finger during the task. Two round metal caps (18 mm diameter) were covered with fine sandpaper (180 grit) and securely fastened onto both sides of the force sensor to provide a rigid, high-friction interface between the fingers and the sensor. To provide visual feedback of force, subjects were seated and facing a 24-in LCD monitor located 1.25 m away. The target force was displayed as a horizontal dashed line located in the middle of the screen, and the force produced by the subject during the 120 s trial was displayed as a horizontal solid line. Subjects were instructed to match their actual force with the target force as closely as possible. Due to hardware limitations related to significant cable noise in the Coulbourn bio-amplifiers when recording six EMG signals simultaneously, two force matching trials were performed at each force level. One trial involved recording two bipolar EMGs from the proximal and distal pairs of electrodes from each muscle (i.e., from electrode pairs that did not overlap the innervation zone of each muscle). The other trial involved an identical force matching task that was performed with the electrode connections reconfigured to record from only the pair of electrodes centered over the innervation zone of each muscle. The order of the two force-matching trials and force levels was randomized across subjects.

2.3. Analysis

The procedures for calculation of coherence between two signals have been described in detail in previous publications (Amjad et al., 1997; Halliday et al., 1995). Briefly, spectral analysis of the EMG signals was performed using custom scripts written in Matlab and based predominantly on software by Neurospec (www.neurospec.org) (Halliday et al., 1995). As EMG–EMG coherence during hand pressing tasks is reported to be in specific frequency bins (Baker et al., 1999; Farmer et al., 2007; Kilner et al., 1999), maximal EMG–EMG coherence for each force trial was calculated within the frequencies from 1 to 15 Hz and 16 to 32 Hz. For the main comparison of interest in the current study, coherence was calculated between rectified EMGs detected with electrodes centered over the innervation zones of FDI and APB, and this value was