Sensitivity of EMG-EMG coherence to detect the common oscillatory drive to hand muscles in young and older adults

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Keenan KG, Massey WV, Walters TJ, Collins JD. Sensitivity of EMG-EMG coherence to detect the common oscillatory drive to hand muscles in young and older adults. J Neurophysiol 107: 2866–2875, 2012. First published February 29, 2012; doi:10.1152/jn.01011.2011.—Multichannel surface electromyograms (EMGs) were used to examine the sensitivity of EMG-EMG coherence to infer changes in common oscillatory drive to hand muscles in young and older adults. Previous research has shown that measures of coherence calculated from different neurophysiological signals are influenced by the age of the subject, the visual feedback provided to the subject, and the task being performed. The change in the magnitude of EMG-EMG coherence across experimental conditions is often interpreted as a change in the oscillatory drive to motoneuron pools of a pair of muscles. However, signal processing (e.g., full-wave rectification) and electrode location are also reported to influence EMG-EMG coherence, which could decrease the sensitivity of EMG-EMG coherence to infer a change in common oscillatory drive to motoneurons. In this study, multichannel EMGs were used to compare EMG-EMG coherence in young (n = 11) and older (n = 10) adults during index finger abduction and pinch grip tasks performed at 2 and 3.5 N with a low and a high visual feedback gain. We found that, across all conditions, EMG-EMG coherence was influenced by electrode location (P < 0.001) but not by subject age, visual feedback gain, task, or signal processing. These results suggest that EMG-EMG coherence is most sensitive to electrode location. The results are discussed in terms of the potential issues related to inferring a common oscillatory drive to hand muscles with surface EMGs.

motor unit; synchronization; cross-correlation; electromyogram

DIRECT AND INDIRECT PATHWAYS from the motor cortex to the neurons innervating the muscles of the hand are necessary for the precise control of hand function (Lemon 2008). However, the neural strategies through which control across muscles is achieved are still largely unknown. One candidate strategy is a common oscillatory drive to populations of spinal motoneurons. Coherent oscillations in the synaptic inputs to motoneurons result in the periodic modulation of motor unit discharge rates, and can thus influence performance of motor tasks (Farmer 1998). However, the functional significance of this rhythmic neural activity in motor control, especially during healthy aging, remains largely unknown (Baker et al. 1999; Kilner et al. 2002; Semmler 2002).

One difficulty in determining the functional significance of the oscillatory drive in motor control may be related to the sensitivity of the techniques commonly used to identify rhythmic neural activity in humans performing voluntary contractions. Specifically, the oscillatory drive to motoneurons is routinely measured by using coherence analysis between two signals 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). The coupling of individual motor units, in particular, is traditionally measured with coherence analysis derived from the discharge times of a pair of concurrently active motor units, termed motor unit coherence (Farmer et al. 1993; Kakuda et al. 1999; Kilner et al. 2002). This approach may be problematic, however, as pairwise correlations between the discharges of two neurons may significantly underestimate correlated activity across entire populations of neurons (Negro and Farina 2011; Santello and Fuglevand 2004; Schneidman et al. 2006). Because surface electromyograms (EMGs) reflect activity from many motor units, coherence measures derived from pairs of surface EMGs recorded over synergist muscles (i.e., EMG-EMG coherence) have been used to provide a more representative measure of rhythmic activity across muscles (Baker et al. 1999; Farmer et al. 2007; Kilner et al. 1999). These studies find significant coherence in the frequency band from 16 to 32 Hz that appears related to specific motor parameters (e.g., digit displacement). Additional experimental support is derived from corroboration of EMG-EMG coherence at 16–32 Hz by other measures of coherence calculated from neurophysiological signals, including 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; Negro and Farina 2011).

Nevertheless, the sensitivity of EMG-EMG coherence to detect oscillatory drive to the motoneuron pools of synergist muscles is largely unknown. The surface EMG is influenced by many different factors that are independent of muscle activity (Farina et al. 2004; Keenan et al. 2007), and these factors may influence the accuracy of any measure derived from the EMG to infer activity in populations of motor units. For example, EMG-EMG coherence is sensitive to the location of the recording electrodes overlying first dorsal intersosseus (FDI) and abductor pollicis brevis (APB) during a pinch grip task (Keenan et al. 2011). Specifically, placing electrodes near the innervation zone, defined as the region of muscle tissue in which action potentials propagate bidirectionally outward toward the tendons, decreased the magnitude of EMG-EMG coherence, which could decrease its sensitivity as a measure of common oscillatory drive. The change in EMG-EMG coherence with electrode location likely results from alterations in EMG amplitude (Beck et al. 2008; Rainoldi et al. 2004) and spectral frequency content (Mesin et al. 2009; Roy et al. 1986) when electrodes are placed over the innervation zone. In addition, the methods used to process an EMG signal—for
example, rectifying the EMG signal prior to conducting a coherence analysis—are reported both to improve (Boonstra and Breakspear 2012; Halliday and Farmer 2010; Myers et al. 2003; Yao et al. 2007) and to impair (Farina et al. 2004; McClelland et al. 2011; Neto and Christou 2010; Stegeman et al. 2010) characterization of motor unit behavior.

Regardless of the above limitations, coherence between different neurophysiological signals, including EMGs, is reported to be influenced by factors of interest to motor control researchers and clinicians. We investigated three of these factors in the present study, including the age of the subject, the gain of the visual feedback signal provided to the subject, and the task being performed. These factors are discussed in more detail below.

First, older adults (mean ± SD age: 70.4 ± 5.9 yr) demonstrate greater motor unit coherence than young adults (24.1 ± 4.1 yr) from 2 to 20 Hz during the performance of an index finger abduction task (Semmler et al. 2003). Although few studies have examined the role of an altered oscillatory drive in healthy older adults (>65 yr), this finding is consistent with the finding of increased interhemispheric EEG-EEG coherence (Maurits et al. 2006; ten Caat et al. 2008) and increased EEG- and MEG-EMG coherence (Johnson and Shinohara 2012; Kamp et al. 2011) in older adults during the performance of neurocognitive and motor tests. Increased oscillatory drive measured with motor unit recordings has also been linked with impaired performance during hand tasks in younger adults (Halliday et al. 1999; Kakuda et al. 1999), and older adults frequently perform worse on motor tasks (Enoka et al. 2003). Thus the decreased performance of older adults on motor tasks may be linked with increased oscillatory drive. Second, coherence between EEG and EMG during a finger flexion task increased in the 15–30 Hz frequency range when the visual feedback and precision demands of the pressing task were altered to require more precision (Kristeva-Feige et al. 2002). This observation is consistent with the finding that attention-demanding behavioral tasks augment motor unit synchronization (time-domain correlation between motor unit discharge times) in humans (Schmied et al. 2000) and oscillatory activity in the motor cortex of monkeys (Murthy and Fetz 1996; Sanes and Donoghue 1993). Third, EMG-EMG coherence is sensitive to the task being performed, as shown with decreased EMG-EMG coherence between hand muscles during dynamic tasks relative to static isometric grip force tasks (Baker et al. 2001; Kilner et al. 1999). Although coherence between different neurophysiological signals has been reported to vary across the three conditions listed above, it is unclear what functional significance these changes have, or if these previous findings can be extended to different aging populations and different motor tasks.

Although previous studies have found that many different factors can influence the measure of EMG-EMG coherence, it is unclear what factor or set of factors the measure is most sensitive to. For example, although we previously found that electrode location may influence EMG-EMG coherence (Keenan et al. 2011), we varied nothing else in that study, so it is not clear whether electrode location actually influences the sensitivity of EMG-EMG coherence to detect a change in coherence for parameters of interest to motor control researchers. Our experimental approach in the present study was to vary five different factors previously reported to alter coherence measures derived from different neurophysiological signals to determine which factors had the greatest influence on EMG-EMG coherence. The factors included subject age, visual feedback gain, task being performed, signal processing (i.e., full-wave rectification), and electrode location. On the basis of previous research, it was expected that the combination of older age, increased visual feedback gain, and greater coactivity of FDI and APB during a pinch grip task compared with an index finger abduction task would lead to increased levels of EMG-EMG coherence. It was also expected that the increase in EMG-EMG coherence under the above conditions would be altered by signal rectification and the location of the EMG electrodes.

MATERIALS AND METHODS

Ethical approval. The experiments were approved by the Institutional Review Board at the University of Wisconsin-Milwaukee. All participants gave their written formal consent before participating in the study.

Subjects. Eleven young (age 23.6 ± 1.7 yr, range 21–26 yr; 7 men, 4 women) and 10 older (age 71.1 ± 6.3 yr, range 65–86 yr; 4 men, 6 women) adults with no known neuromuscular disorders volunteered to participate in this study. Additionally, data from 10 healthy, young adults [5 men, 5 women; 24.4 ± 1.3 yr (mean ± SD), range 23–27 yr] from a previously published study (Keenan et al. 2011) were used as a comparison with older adults on the pinch grip task in the present study. All subjects were right-hand dominant as assessed by the Edinburgh Handedness Inventory (Oldfield 1971).

Experimental setup and procedure. The experimental approach was similar to our previously published approach (Keenan et al. 2011) that involved having young adults perform a pinch grip task at 2 and 3.5 N. In the present study, older adults performed both index finger abduction and pinch grip tasks for 120 s at absolute force levels of 2 and 3.5 N. Because data for the pinch grip task had already been collected (Keenan et al. 2011), young adults performed only the index finger abduction task for 120 s at 2 and 3.5 N in this study. The performance of older adults on the pinch grip task was compared with the data collected previously from the young adults (Keenan et al. 2011). These low force levels were chosen to replicate previous studies finding significant EMG-EMG coherence during a pinch task (Kilner et al. 1999), 2) to collect sufficient data (120 s) for the coherence analysis while limiting the possibility of fatigue, and 3) to maximize differences in performance between young and older adults, which is often greatest at low force magnitudes (Enoka et al. 2003). Also, in the present study, the visual gain of the force feedback signal was manipulated (i.e., low and high feedback gain conditions; detailed below), and EMGs were recorded from FDI, APB, extensor digitorum communis (EDC), and flexor digitorum superficialis (FDS). The order of the two force-matching trials and visual feedback gain conditions was randomized across subjects. In addition, older adults performed the index finger abduction and pinch grip task within the same experimental session in a counterbalanced order to control for a practice effect. Experimental tasks were separated by at least 20 min to minimize the possibility of fatigue. Each subject was seated on a chair with the right arm resting on a vacuum foam pad (VersaForm pillow; Tumble Forms) inflated to prevent movement of the forearm. Key methods are discussed below, highlighting those procedures that varied from the previous approach (Keenan et al. 2011).

For the index finger abduction task, the participant’s right hand was placed in a manipulandum that isolated the index finger from the thumb and the other fingers. Subjects pressed into a small wooden dowel securely fastened to a force sensor (Nano 17; ATI Industrial Automation) that was positioned at the proximal interphalangeal joint of the index finger. For the pinch grip task, participants held a uniaxial force sensor (model ELFS-B3–10; Entrant) between the thumb-tip and
radial side of the distal phalanx of the index finger (i.e., key pinch). Further instrument particulars are provided in Keenan et al. (2011).

To provide visual feedback of force, subjects were seated facing a 24-in. LCD monitor located 1 m away. The target force for all conditions was displayed as a horizontal dashed black line located in the center of the screen, and the force produced by the subject during the 120-s trial was displayed as a horizontal solid green line. Subjects were instructed to match their actual force with the target force as closely as possible, with the solid green line moving up and down on the monitor with increasing and decreasing force, respectively. The low and high visual feedback gain conditions were associated with a large and a small range of forces being displayed to the participant, respectively. For example, if participants were performing the 2-N force task with low visual feedback gain, the target line was drawn at the center of the monitor and the bottom and top of the screen corresponded to 0 and 4 N, respectively. In contrast, for the high visual feedback gain, the bottom and top of the screen corresponded to 1.9 and 2.1 N, respectively, with the target line still drawn at the center of the monitor. Thus, for the high feedback gain condition, similar deviations of the force trace around the target line would appear 10 times greater than those for the low visual feedback gain condition. Visual gains varied from 180.0 to 1,800 pixels/N for the 2-N force level and from 102.9 to 1,029.4 pixels/N for the 3.5-N force level. These gains were selected to match the 10-fold increase in visual gain reported by Schmied et al. (2000) to increase motor unit synchronization.

**EMG methods.** First, the innervation zone and muscle fiber direction were identified with multichannel surface EMGs (EMG-USB; OT Bioelettronica, Torino, Italy). Surface EMG signals were recorded from FDI, APB, EDC, and FDS muscles of the dominant right hand. A ground electrode was strapped around the right wrist, and the subject’s skin was prepared with abrasive paste (Spes Medica). An electrode array consisting of 16 silver electrodes (2.5-mm interelectrode distance) was used to identify the innervation zone by visual analysis of the 15 bipolar EMG signals generated from consecutive electrodes during a series of submaximal test contractions of all muscles (Fig. 1A). As is commonly done (Merletti et al. 2003), the electrode array was positioned over each muscle and the orientation of the array was adjusted to clearly identify propagation of the motor unit action potentials in opposite directions from the innervation zone (Fig. 1B). The innervation zone was then identified to be the EMG signal that was closest to the location where motor unit action potentials began to propagate in opposite 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 overlying all muscles (Fig. 1, A and C). The direction of the array was also marked on the skin with a permanent marker to estimate muscle fiber direction (Fig. 1, A and C). For electrode placement over EDC and FDS, test contractions were performed with both the index and middle fingers, and electrodes were positioned to optimize EMG amplitudes when the index finger compartment of the two muscles was most active.

After estimation of the location of the innervation zone with the electrode array, surface EMG electrodes (4-mm diameter, silver-silver chloride; 15-mm interelectrode distance) were placed on the skin (shown only for FDI in Fig. 1C). Three bipolar EMG recording configurations were constructed from consecutive pairs of electrodes placed in line with muscle fiber direction over both FDI and EDC. Two electrodes (Fig. 1C, electrodes 1 and 2) and proximal (Fig. 1C, electrodes 3 and 4) to the innervation zone. For both APB and FDS, similar procedures were used to identify innervation zones and fiber directions; however, only single bipolar electrode configurations were used on each muscle (positioned 15 mm distal to the innervation zone), as the experimental equipment was limited to eight EMG amplifiers (Coulbourn Instruments, Whitehall, PA). 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 (13–1,000 Hz) with an isolated bioamplifier. EMG signal quality was checked before and after each experiment by checking the baseline level of noise at rest, as well as by having subjects perform a 3- to 5-s maximal contraction of each muscle while the experimenter provided manual resistance (Keenan et al. 2009).

![Fig. 1. Representative example of electromyogram (EMG) methods. A: an EMG array with 16 electrodes was placed over the abductor pollicis brevis (APB) muscle, and the electrode array was oriented to identify the location of the innervation zone and muscle fiber direction (see EMG methods). B: to identify the innervation zone, a total of 15 bipolar EMGs were recorded between each consecutive electrode during test contractions. 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 (identified with an asterisk). The dashed line is provided as reference to identify the propagation of the muscle fiber action potentials. This process was used in all muscles investigated in the present study. C: after identification of the innervation zone in the first dorsal interosseous (FDI) muscle, 4 electrodes were placed in line with the muscle fiber direction and 15 mm apart. Three bipolar EMG signals were recorded from consecutive electrodes. Electrodes 2 and 3 were centered over the estimated location of the innervation zone, and electrodes 1 and 2 and electrodes 3 and 4 were centered 15 mm distal and 15 mm proximal to the innervation zone, respectively.](http://jn.physiology.org/doi/10.1152/jn.01011.2011)
Data analysis. The primary dependent variable was the peak in coherence within the frequencies from 1 to 15 Hz and 16 to 32 Hz. 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 with 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 experimental condition was calculated within 1–15 Hz and 16–32 Hz. To examine the effect of signal processing, all comparisons were performed with both rectified and interference EMGs.

The particular influence of electrode location on EMG-EMG coherence was quantified by calculating coherence from EMGs formed with electrodes either centered over the innervation zone or placed away from the innervation zone. For the index finger abduction task, coherence was calculated between EMGs detected with electrodes centered over the innervation zones of FDI and EDC. This value was compared with the maximal EMG-EMG coherence calculated with pairs of EMGs from across the two muscles with electrodes that avoided the innervation zone. Specifically, a total of four comparisons across FDI and EDC were made by calculating coherence spectra between the EMGs detected proximal and distal to the innervation zone of FDI with the EMG recordings proximal and distal to the innervation zone of EDC. EMG-EMG coherence was calculated between EMGs from FDI and EDC because previous work has reported that the amplitude of muscle activity in both muscles is similar during index finger abduction tasks performed across a range of force levels (Valero-Cuevas et al. 1998).

A slightly different approach was necessary to examine the influence of electrode location when calculating coherence between EMGs from FDI and APB, because only one EMG signal was recorded from APB and that EMG recording was made with electrodes that avoided the innervation zone. Thus coherence was calculated between the EMG signal recorded over the innervation zone of FDI with the EMG from APB, and that value was compared with the maximal EMG-EMG coherence value from the other two EMGs with electrodes that avoided the innervation zone of FDI. To examine the influence of task on EMG-EMG coherence between the same muscle pair, coherence between APB and FDI was calculated for both the index finger abduction and pinch grip tasks in the older adults. This comparison was not possible for young adults because the population of subjects performing the abduction task in the present study was different from the population that previously performed the pinch grip task.

In addition to examining the magnitude of EMG-EMG coherence across different muscle pairs described above, we also characterized the percentage of trials across different conditions that had significant peaks in EMG-EMG coherence. This was a strategy employed by Farmer et al. (1993) to examine significant peaks in motor unit coherence within and across muscles and has frequently been used to characterize EMG- and EEG-EMG coherence across different muscles (Kilner et al. 2000; Raethjen et al. 2001). Specifically, we calculated the percentage of trials across conditions that showed significant EMG-EMG coherence above the 95% confidence interval between 1–15 Hz and 16–32 Hz for the following muscle pairs: FDI-EDC, FDI-APB, FDI-FDS, and EDC-FDS.

In addition to the analysis of EMG-EMG coherence, the motor performance of young and older adults on force-matching tasks was also analyzed by calculating the coefficient of variation in force (CV = standard deviation of force magnitude/mean force level × 100). Additionally, average full-wave rectified EMG amplitudes, normalized to the maximal EMG recorded during maximal contractions performed by each muscle, were calculated for all tasks to evaluate the influence of electrode location on EMG amplitude.

Statistical analysis. ANOVAs were conducted to test whether the peak in EMG-EMG coherence varied across the conditions examined in the present study. For all ANOVAs, the EMG-EMG coherence values were z-transformed (Rosenberg et al. 1989) to allow comparisons across subjects before performing the ANOVAs. For the index finger abduction task, a mixed between-within subjects ANOVA was conducted with repeated measures on visual feedback gain (low or high), frequency band (1–15 or 16–32 Hz), signal processing (rectified or interference EMG), and electrode location (i.e., electrodes centered over or away from the innervation zone) and with the between-subjects factor of age group (young and older adults). For the pinch grip task, a mixed between-within subjects ANOVA was conducted with repeated measures on frequency band, signal processing, and electrode location and with the between-subjects factor of age group. It was not possible to examine the influence of visual gain for the pinch grip task, as this data set had been collected previously from young adults (Keenan et al. 2011) who did not perform pinch grip under the high visual gain condition. To examine the influence of the two finger pressing tasks on EMG-EMG coherence from the same muscle pair (i.e., FDI and APB) in older adults, one-way ANOVA was performed with repeated measures on task (abduction or pinch), visual feedback gain, frequency band, signal processing, and electrode location. Again, this comparison was not possible in young adults as two different populations of young adults performed the pinch grip and index finger abduction tasks.

ANOVA was performed to test whether the coefficient of variation in force varied across conditions. For the index finger abduction task, a mixed between-within subjects ANOVA was conducted with repeated measures on visual feedback gain and force level and with the between-subjects factor of age group. For the pinch grip task, a mixed between-within subjects ANOVA was conducted with repeated measures on force level and with the between-subjects factor of age group. To examine the influence of different parameters on average EMG amplitudes, a mixed between-within subjects ANOVA was conducted with repeated measures on visual feedback gain, force level, muscle (FDI or EDC), and electrode location and with the between-subjects factor of age group.

Statistical significance for all tests was set at P < 0.05. The significance level for coherence functions was computed with α = 0.05. Results are presented as means ± SE in text and Tables 1 and 2. As the results were statistically similar for both force levels, only the results for the 2-N force level are presented unless otherwise stated.

RESULTS

The results from this study extended our previous approach (Keenan et al. 2011) by examining both young and older adults, assessing the role of the gain of the visual feedback signal, evaluating both pinch grip and index finger abduction tasks, and assessing the role of signal rectification in EMG-EMG coherence. Similar to previous studies (Laidlaw et al. 2000; Semmler et al. 2000), we found that the older adults were less steady (i.e., had a higher coefficient of variation in force magnitude) than the young adults when performing submaximal isometric contractions with the hand. However, the reduction in steadiness was not associated with a change in EMG-EMG coherence. Surprisingly, the only factor that influenced EMG-EMG coherence in young and older adults was the location of the recording electrodes.

Index finger abduction. To characterize the extent of coherence between muscle pairs, we calculated the percentage of trials with significant EMG-EMG coherence across the different conditions in the study. For index finger abduction tasks there were significant peaks in coherence between EMGs from
FDI and EDC in 91.7% of all trials. Although there were no systematic differences based on age, force level, frequency range, or visual gain conditions, when electrodes were positioned to avoid the innervation zone 100% of all trials exceeded the 95% confidence interval. The percentage of trials with significant coherence between EMGs from FDI and EDC decreased to 83.75% when coherence was computed from EMGs with electrodes positioned over the innervation zones. This result was similar across all muscle pairings. Specifically, when recording EMGs that avoided the innervation zone relative to EMGs detected over the innervation zone, the percentage of trials with significant coherence decreased from 92.5% to 76.25% between EMGs from FDI and FDS, respectively; from 100% to 95% between EMGs from FDI and APB, respectively; and from 97.5% to 77.5% between EMGs from FDS and EDC, respectively. Thus there was substantial coherence across all muscle pairs, with the percentage of trials with significant coherence being influenced by electrode location.

Similarly, there were no significant differences in the magnitude of the peak coherence between EMGs from FDI and EDC during index finger abduction tasks (Table 1) for age (P < 0.43), frequency range (P = 0.127), signal processing (P = 0.121), and visual gain (P = 0.28). Also, none of the interaction effects was significant (P > 0.05). However, the magnitude of coherence was significantly less (P < 0.001) when EMGs were recorded with electrodes located over the innervation zones of FDI and EDC relative to when electrodes were positioned to avoid the innervation zone (Table 1). The coefficient of variation in force during index finger abduction was significantly (P < 0.001) decreased for young versus older adults (1.7 ± 0.2% vs. 3.2 ± 0.2%) and for the 2.0- and 3.5-N forces (2.4 ± 0.2% vs. 2.0 ± 0.2%; P < 0.003). Thus the change in the coefficient of variation in force during pinch grip tasks across young and older adults was not associated with an altered EMG-EMG coherence.

**Pinch grip.** We also calculated the percentage of trials with significant EMG-EMG coherence across the different conditions involving pinch grip. This analysis was restricted to include only the older adults. Similar to index finger abduction, when recording EMGs that avoided the innervation zone relative to EMGs detected over the innervation zone, the percentage of cases of significant coherence decreased from 100% to 97.5% between EMGs from FDI and APB, respectively; from 100% to 65% between EMGs from FDI and EDC, respectively; from 93.75% to 76.25% between EMGs from FDI and FDS, respectively; and from 92.5% to 80% between EMGs from FDS and EDC, respectively. Thus, as with index finger abduction tasks, there is substantial coherence across all muscle pairs during pinch grip tasks, with the percentage of cases of significant coherence being influenced by electrode location.

There were no significant differences in the magnitude of the peak coherence between EMGs from FDI and APB during a pinch task (Table 1) for age (P = 0.55), frequency range (P = 0.36), or signal processing (P = 0.111). Also, none of the interaction effects was significant (P > 0.05). However, the magnitude of coherence was significantly less (P < 0.001) when EMGs were recorded with electrodes located over the innervation zones of FDI and APB relative to when electrodes were positioned to avoid the innervation zone (Table 1). The coefficient of variation in force during pinch grip was significantly (P < 0.001) decreased for young versus older adults (1.7 ± 0.2% vs. 3.2 ± 0.2%) and for the 2.0- and 3.5-N forces (2.4 ± 0.2% vs. 2.0 ± 0.2%; P < 0.003). Thus the change in the coefficient of variation in force during pinch grip tasks across young and older adults was not associated with an altered EMG-EMG coherence.

**EMG-EMG coherence for two different tasks.** When EMG-EMG coherence between FDI and APB in the subsample of older adults (Table 2) during the abduction and pinch grip tasks was examined, no significant differences were found for the two different finger pressing tasks (P = 0.332). Also, none of the interaction effects was significant (P > 0.05). However, consistent with the findings above, a significant decrease in EMG-EMG coherence between FDI and APB in the subsample of older adults was found with EMG electrodes positioned over the innervation zone with respect to electrodes positioned away from the innervation zone (P = 0.001; see Table 2).

**Electrode location.** For all comparisons detailed above, EMG-EMG coherence was significantly higher when electrodes were positioned to avoid the innervation zone (Figs. 2 and 3). Representative data are shown in Fig. 2. Full-wave rectified EMGs are shown in Fig. 2A for 120 s during the index finger abduction task at 2 N. EMG signals were detected concurrently from two different locations over FDI and EDC, and each electrode pair was positioned either to be directly over (Fig. 2A, left) or away from the innervation zone (Fig. 2A, right). Three different EMG-EMG coherence spectra (Fig. 2B) were calculated from pairs of EMGs across muscles. EMG pairs used for the coherence analysis are connected with arrows and each electrode pair was positioned either to be directly over (Fig. 2A, left) or away from the innervation zone (Fig. 2A, right). Three different EMG-EMG coherence spectra (Fig. 2B) were calculated from pairs of EMGs across muscles. EMG pairs used for the coherence analysis are connected with arrows.

<table>
<thead>
<tr>
<th>Age</th>
<th>Frequency Range</th>
<th>Electrode Location</th>
<th>Signal Processing</th>
<th>Visual Gain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Young</td>
<td>1-15 Hz</td>
<td>Avoid IZ</td>
<td>Interference</td>
<td>Low</td>
</tr>
<tr>
<td>Older</td>
<td>16-32 Hz</td>
<td>Over IZ</td>
<td>Rectified</td>
<td>High</td>
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<tr>
<td>Abduction</td>
<td></td>
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<tr>
<td>FDI-EDC</td>
<td>F(1,19) = 0.65</td>
<td>F(1,19) = 2.54</td>
<td>F(1,19) = 0.29</td>
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<td></td>
<td>P &lt; 0.43</td>
<td>P &lt; 0.127</td>
<td>P &lt; 0.001*</td>
<td>F(1,19) = 1.24</td>
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<td>Pinch</td>
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<td>4.22 ± 0.3</td>
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<td>FDI-APB</td>
<td>F(1,19) = 0.39</td>
<td>F(1,19) = 2.63</td>
<td>F(1,19) = 2.80</td>
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<tr>
<td></td>
<td>P &lt; 0.55</td>
<td>P &lt; 0.001*</td>
<td>P &lt; 0.111</td>
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</table>

Values are means ± SE. EMG-EMG coherence values were z-transformed to allow comparisons across subjects. Note that the ANOVA for pinch grip was performed without examining visual gain (n/a) because comparisons were made to previously published data from young adults who did not perform the pinch grip task during a high visual gain condition (Keenan et al. 2011). There were no significant interaction effects. IZ, innervation zone; FDI, first dorsal interosseus; EDC, extensor digitorum communis; APB, abductor pollicis brevis. *P < 0.001.

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in Fig. 2A, and the line type connecting the two EMGs is identical to the line type used to represent the coherence spectra shown in Fig. 2B. Peak EMG-EMG coherence was near 24 Hz for all three electrode locations; however, the amplitude of the peak in coherence was less when EMGs were detected with electrodes positioned over the innervation zone with respect to electrodes placed away from the innervation zone (Fig. 2). When avoiding only one of the two innervation zones (Fig. 2B), peak EMG-EMG coherence was at a value in between the two other coherence spectra. Individual subject data are shown in Fig. 3 to demonstrate the influence of electrode location on peak EMG-EMG coherence in older and young adults for both the pinch grip and index finger abduction tasks. For this representative figure (Fig. 3), the peak in EMG-EMG coherence was calculated between 16 and 32 Hz during the low gain visual feedback condition after the EMG signals had been rectified. As with the representative data in Fig. 2 and the data reported above, electrode location clearly influences peak EMG-EMG coherence during the pinch grip and index finger abduction tasks for older and young adults.

The decrease in EMG-EMG coherence when detected with electrodes positioned over the innervation zone is at least partly due to the decreased amplitude of the EMG signal. We calculated EMG amplitudes from FDI and EDC for the abduction task in young and older adults. Full-wave rectified EMG amplitudes were averaged for the 120-s task duration and were compared across the two force levels and two visual gain conditions. There was a significant decrease ($P = 0.021$) in average EMG amplitude when electrodes were positioned over the innervation zone (3.9 ± 0.4%) rather than away from the innervation zone (6.1 ± 1.2%). As expected, EMG amplitude was significantly different for the 2.0- and 3.5-N forces (4.2 ± 0.01% vs. 5.8 ± 0.01%; $P < 0.002$). None of the interaction effects was significant ($P > 0.05$).
DISCUSSION

The purpose of this study was to use multichannel surface EMGs to examine the sensitivity of EMG-EMG coherence to infer changes in common oscillatory drive to hand muscles of young and older adults. In contrast to our expected findings, EMG-EMG coherence was not influenced by the age of the participant, the gain of the visual feedback signal, the task being performed, or signal rectification (Tables 1 and 2). EMG-EMG coherence was influenced only by the location of the recording electrodes. These results suggest a number of sensitivity issues related to the use of surface EMGs to detect the common modulation of motor unit activity. Before the potential EMG sensitivity issues are addressed, however, differences between our experimental approach and the previous work are discussed. The significance of our findings, along with limitations of the experimental approach, is also discussed below.

Subject age. Although we expected an increase in EMG-EMG coherence in older adults, no study to date has assessed coherence across muscles in healthy older adults (≥65 yr), either by motor unit coherence or by EMG-EMG coherence. We found that older adults did not have an increased EMG-EMG coherence relative to young adults for either the pinch grip or the index finger abduction task (Table 1).

No study to date has assessed whether EMG-EMG coherence, or motor unit coherence across muscles, is altered in healthy older adults, which confounds comparison of the results obtained from a pair of motor units in a single muscle (Semmler et al. 2003) to synergist muscles. As Semmler et al. (2003) report, motor unit coherence within a single muscle (e.g., FDI) can be changed with advancing age during isometric finger pressing tasks, but it is unclear whether across-muscle motor unit coherence is altered with advancing age. Without a change in the common oscillatory drive that is shared across muscles, there would be no significant change in EMG-EMG coherence with advancing age. Thus increases in EEG-EEG (Maurits et al. 2006; ten Caat et al. 2008) or EMG-EMG (Johnson and Shinohara 2012) coherence with advancing age do not necessarily indicate that EMG-EMG coherence would correspondingly increase in older adults.

Further confounding our understanding of the role of oscillatory drive in older adults, and in contrast to their earlier study (Semmler et al. 2003), Semmler et al. (2006) reported that motor unit coherence was not significantly different for young and older adults, except from 6 to 9 Hz, where coherence was less in older adults. The failure to find an increased coherence with age in the later study could be related to performance of the task under dynamic conditions (postural hold and isometric contractions), which is different from the isometric force pressing tasks performed in the present study and by subjects in the earlier Semmler et al. (2003) study. Performing finger tasks under dynamic conditions is known to significantly decrease EMG-EMG and motor unit coherence (Baker et al. 2001; Kilner et al. 1999) and could have masked age-associated differences. These contradictory findings do not clarify the role of oscillatory drive in older adults. Moreover, the use of pairwise comparisons between the discharges of single motor units, as in the studies above, may significantly underestimate oscillatory motor unit activity (Negro and Farina 2011) and influence the sensitivity of the measure of motor unit coherence to detect differences in oscillatory drive.

The potential change in common oscillatory drive across muscles with advancing age is likely influenced by many factors, confounding simple predictions of an increase or decrease in oscillatory drive with advancing age. For example, previous work has shown that EMG-EMG coherence is very sensitive to the loss of afferent feedback (Fisher et al. 2002; Riddle and Baker 2005). As there are somatosensory impairments with aging (see Johansson 1996), it is possible that oscillatory synaptic drive may increase from supraspinal centers with advancing age but is offset by the loss of peripheral sensory input with advancing age. The result could be that EMG-EMG coherence and across-muscle motor unit coherence may not be modulated with advancing age, as the two effects cancel each other out.

Thus the failure to find a significant increase in EMG-EMG coherence with advancing age could be due to the fact that one does not exist. Alternatively, the failure to find a difference in EMG-EMG coherence with advancing age in the present study could be due to the fact that the measure is influenced by many different factors that limit its sensitivity as a measure of oscillatory drive. The high sensitivity of EMG-EMG coherence to electrode location raises this possibility and is discussed below. Future studies using signals from multiple sites throughout the neuromuscular system, for example, simultaneous recordings of EEGs, as well as motor unit activity and EMGs from synergist muscles, may help to improve our understanding of how the drive to motoneurons changes with advancing age, and also to clarify what factors influence the capability of the different recordings to detect oscillatory drive.

High and low visual gain. In contrast to previous findings (Kristeva-Feige et al. 2002) that report a change in EEG-EMG coherence by altering the precision demands of the finger pressing tasks, EMG-EMG coherence did not change with visual gain in the present study. This finding is reported for the young and older subjects performing the index finger abduction task in Table 1 and for the older subjects for both tasks in Table 2. Kristeva-Feige et al. (2002) used a low precision condition in which subjects were able to perform within a 20% window around their target force level, while for the high precision condition subjects were asked to maintain their force level at the target line as steadily as possible. The authors found that EEG-EMG coherence increased for the task performed during the high precision condition. In the present study, precision demands were varied by manipulating the gain of the visual feedback presented to the subjects while asking them to maintain their target force as steadily as possible in both conditions. This slightly different approach was used because increasing the gain of the visual feedback signal had previously been shown to effectively alter motor unit synchronization (Schmied et al. 2000) and to increase the coefficient of variation in force magnitude in older adults relative to young adults (Sosnoff and Newell 2006). As the common modulation of input received by active motor units has been identified to play a significant role in force fluctuations (Halliday et al. 1999; Kakuda et al. 1999), it was expected that including a high visual gain condition in older adults would further increase EMG-EMG coherence. However, although force variability was greater in older adults compared with young adults, there were no differences between visual gain conditions for force variability or EMG-
EMG coherence (Tables 1 and 2). Similar to the results related to subject age above, it is unclear whether the finding of no significant change in EMG-EMG coherence is due to the fact that altering precision demands only influences corticomuscular coherence (Kristeva-Feige et al. 2002) or that the measure of EMG-EMG coherence is not sensitive enough to detect the change in common oscillatory drive.

**Task.** Although oscillatory activity is a widespread feature of neural activity, its functional significance remains unclear. Of interest is whether the oscillatory behavior is simply an epiphenomenon or has direct functional significance for hand motor control. For example, Riddle and Baker (2006) reported that the level of EEG-EMG coherence covaries with the amount of digit displacement. This finding suggests that corticomuscular coherence is not an epiphenomenon but is directly related to specific parameters (i.e., digit displacement) needed for hand motor control. Similarly, we assessed whether coherence between EMGs from FDI and APB was different for two tasks that would require different levels of synergist muscle activity. Specifically, APB is expected to have little involvement during index finger abduction, as a manipulandum was used to isolate the index finger from the thumb and the other fingers. In contrast, APB is involved in pinch grip tasks (Johanson et al. 2001; Maier et al. 1993), although its primary action is abduction of the thumb. Nonetheless, EMG-EMG coherence between FDI and APB was not different between the two different tasks in the subpopulation of older adults (Table 2). This result suggests that EMG-EMG coherence is not sensitive to differences across these tasks, and that EMG-EMG coherence may be an epiphenomenon with little functional consequence for at least some motor tasks.

**Signal rectification.** Full-wave rectification of the EMG signal before calculation of coherence has been suggested as a method to improve characterization of the modulation of motor unit behavior (Boonstra and Breakspear 2012; Myers et al. 2003; Yao et al. 2007), and most research using EMGs includes this signal processing step (Baker et al. 1999; Keenan et al. 2011; Kilner et al. 1999; Negro and Farina 2011). However, both experimental and computational approaches have reported that rectification may actually impair the ability to identify oscillatory drive with surface EMGs (Farina et al. 2004; McClelland et al. 2011; Neto and Christou 2010; Stegeman et al. 2010). As related to the present study, full-wave rectification had no influence on the current findings (Tables 1 and 2). Although signal rectification has previously been shown to alter EMG-EMG coherence, our present results suggest that the measure of EMG-EMG coherence is more sensitive to another factor, that of electrode location.

**Electrode location.** Similar to our previous study in young adults performing a pinch grip task (Keenan et al. 2011), EMG-EMG coherence is sensitive to electrode location (Figs. 2 and 3). Moreover, in the present study we find that this effect appears to be independent of subject age, visual feedback gain, task, or signal rectification (Tables 1 and 2).

Placement of bipolar electrodes on the skin overlying the middle of a muscle (the so-called “belly” of a muscle), where the innervation zone is likely to be located, is the most common recording configuration (see for review Mesin et al. 2009). Nonetheless, many studies suggest placing electrodes away from this location (Beck et al. 2008; Merletti et al. 2003; Rainoldi et al. 2004; Roy et al. 1986). One complication with positioning electrodes over the innervation zone is that the amplitude of the EMG signal is frequently at its lowest value with regard to all other detection sites over the muscle, as clearly shown with multichannel EMGs (Beck et al. 2008; Rainoldi et al. 2004). The decrease in EMG amplitude in the present study with location of the electrodes over the innervation zone is consistent with this previous work. In addition, spectral estimates of the frequency content of a signal can be altered by electrode location relative to the innervation zone (Mesin et al. 2009; Roy et al. 1986), which may be problematic for coherence analysis, which involves cross-correlation between EMG signals in the frequency domain.

It would be convenient to dismiss the influence of the innervation zone on EMG-EMG coherence as easily avoided and insignificant, but three lines of reasoning make such a conclusion untenable. First, EMG-EMG coherence changed substantially with small changes (1.5 cm) in electrode placement, and this effect is likely not easily avoided. For example, recording EMG with bipolar electrodes over hand muscles necessarily involves placing electrodes on the muscle belly, because of the small size of the muscles. Although anatomical maps of innervation zones are often used to guide EMG electrode placement (e.g., see Rainoldi et al. 2004), not all muscles are uniform in the arrangement of their innervation zones (Rainoldi et al. 2004; Saitou et al. 2000), which presents practical limitations when trying to avoid the innervation zone. Similarly, although multichannel EMG systems are available to estimate the location of the innervation zone, their use is still not standard practice and may be cost-prohibitive for many users. Second, the location of the innervation zone relative to the recording electrodes can change during both isometric and anisometric contractions (Mesin et al. 2009; Piitulainen et al. 2009), which could result in unpredictable changes in EMG coherence that are unrelated to a change in rhythmic drive to the muscle. Third and finally, we were able to assess the influence of the electrode location on EMG-EMG coherence with multichannel EMGs, but there are many other factors that influence the EMG and measures derived from it that cannot routinely be assessed in vivo. These factors include the shape of the muscle and volume conductor, thickness and conductivities of tissue layers, tissue inhomogeneities, fiber length, pinnation angle, inclination of the electrodes relative to the muscle fibers, and cross talk from nearby muscles (Farina et al. 2004). For example, cross talk cannot be disregarded as a potential confound in the present study, particularly with EMGs detected with surface electrodes on the skin overlying EDC. Although surface EMG electrodes are routinely placed over the index finger compartment of EDC, and a multichannel EMG array was used in the present study to guide proper electrode placement, it is possible that cross talk from nearby muscles could still have influenced the present results of EMG-EMG coherence during index finger abduction.

The influence of electrode location on EMG-EMG coherence may be related to the change in shape of motor unit action potentials based on where over the skin the EMG signal is detected. Studies using high-density EMG arrays with up to 125 channels of EMG detected concurrently report a change in the shape of motor unit action potentials with electrode location (Kleine et al. 2007). The change in motor unit action potential shape with electrode location near the innervation zone may influence the frequency content of the EMG signal.
and thus influence the measure of EMG-EMG coherence. Similarly, in a previous study using a numerical model of surface EMGs, we assessed the sensitivity of using cross-correlation between surface EMGs to detect time-domain motor unit synchronization (Keenan et al. 2007). The cross-correlation index derived by cross-correlating surface EMGs is the time-domain equivalent of the frequency-domain analysis of EMG-EMG coherence. Although motor unit synchronization across muscles influenced the cross-correlation index, so did every other parameter varied in that study, including excitation level, muscle size, fat thickness, skin conductivity, and motor unit conduction velocity. Moreover, the change in the cross-correlation index in that modeling study was closely related to changes in the shapes of the constituent motor unit action potentials across the various conditions. A recent computational modeling study has also found that action potential shape influences EMG-EMG coherence (Boonstra and Breakspear 2012), although the authors report a differential effect based on signal rectification that our present study did not find. Nonetheless, it is possible that EMG-EMG coherence is more sensitive to factors influencing motor unit action potential shape than it is to changes in oscillatory drive to muscle pairs. These data, in addition to our present data, make it clear that to gain a better understanding of the functional role that common oscillatory drive has on motor performance, particularly in older adults, it is first necessary to further identify the sensitivity of the techniques commonly used to measure the amount of oscillatory drive.

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DISCLOSURES

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AUTHOR CONTRIBUTIONS

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