EVALUATION OF CROSS-TALK IN ELECTROMYOGRAPHIC SIGNALS

by

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STATEMENT BY AUTHOR

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Abstract

Activity of skeletal muscles produce electrical signals that can be measured using electrodes placed on the skin surface over a target muscle or with electrodes inserted into the muscle. Such electromyographic (EMG) signals provide fundamental information about the intensity of the neural drive acting upon muscle. In addition, EMG signals are widely deployed as control sources for powered prosthetic limbs. One limitation related to recording EMG signals, however, is that signals arising from neighboring muscles may contribute significantly to the activity detected with electrodes placed over or within a given target muscle. Such unwanted contribution of signal from muscles other than the targeted muscle is referred to as cross-talk. Cross-talk was investigated in four neighboring muscles in the forearm with different electrode types and configurations: bipolar intramuscular, monopolar intramuscular, and bipolar surface EMG. Cross-correlation analysis was performed for every pairwise combination of EMG signal recorded. The peak correlation coefficient at near-zero time delay provided an index of the degree of cross-talk. Correlation coefficients dropped off exponentially with distance between recording electrodes. Bipolar intramuscular EMG had the narrowest pick-up range, with a length constant of 14.5 mm. Bipolar surface EMG had a longer length constant of 37.0 mm, whereas monopolar intramuscular EMG had the longest length constant of 64.5 mm. A second set of experiments indicated that correlation in EMG signals detected in different muscles was unlikely to have a neural basis. Therefore, because of their wide detection range, monopolar configurations including those involving intramuscular electrodes, should be avoided.
**Introduction**

Activity of skeletal muscles produce electrical signals that can be measured using electrodes placed on the skin surface over a target muscle or with electrodes inserted into the muscle. Such electromyographic (EMG) signals provide fundamental information about the intensity of the neural drive acting upon muscle. In addition to its importance for basic research, EMG signals are widely deployed as control sources for externally powered prosthetic limbs. Surface EMG is the most common method of recording electrical activity of muscles. Compared to intramuscular electromyography (iEMG), the electrodes are more convenient and non-invasive. However, when recording muscles deep within a limb, or when muscles reside in close proximity to one another, iEMG may be a better method. One limitation related to recording EMG signals, however, is that signals arising from neighboring muscles may contribute significantly to the activity detected with electrodes placed over or within a given target muscle. Such unwanted contribution of signal from muscles other than the targeted muscle is referred to as cross-talk.

Surface electrodes are generally considered to be more prone to cross-talk compared to intramuscular electrodes. However, when measuring EMG signals, more than the electrode type needs to be considered. The EMG configuration is also important as the EMG signal changes due to configuration. The two most common EMG configurations are monopolar and bipolar.

A monopolar configuration requires one electrode on (or in) the target muscle, and one reference electrode placed at a location with little or no electrical activity such as over a bone. Differential amplifiers then output the difference of the two signals in order
to reject noise that is common on both electrodes. A bipolar configuration typically involves two electrodes on (or in) the target muscle. The difference of the two signals in this case not only helps to remove common noise detected on the two electrodes but in theory, should also spatially restrict the detection range, and thereby should minimize crosstalk compared to monopolar configuration.

An additional practical consideration has to with the total number of electrodes involved. For example, in the case of chronically implanted electrodes used to control prosthetic limbs, with more electrodes, the surgical challenges are greater; there is a greater likelihood of electrode failure; and greater possibility for infection or other post-operative difficulties. A monopolar-configured electrode system would require half the number of electrodes (using a common reference electrode) as a bipolar system. However, there is very little information as to the extent of cross-talk for monopolar intramuscular EMG electrodes.

Therefore, the purpose of this project was to evaluate experimentally the degree of cross-talk for EMG signals detected with surface and intramuscular electrodes in both monopolar and bipolar configurations. To our surprise, monopolar intramuscular EMG signals were highly susceptible to detection of activity from muscles quite distant from the muscles in which the electrodes were placed. As such, the use of monopolar intramuscular electrodes to record EMG signals should be avoided.

**Methods**

*Experiment 1*

Five experiments were performed in which EMG activity from the right brachioradialis (BR), pronator teres (PRO), flexor carpi radialis (FCR), and palmaris
longus (PL) were recorded in healthy human volunteers (4 men, 1 woman, ages 23–55 yr). The experimental procedures were approved by the Human Investigation Committee at the University of Arizona. All subjects gave their informed consent to participate in the study.

Subjects were seated in a dental chair with their right elbow and wrist supported on a horizontal platform. The activities of each of the four muscles were recorded simultaneously using three types of electrode configurations: bipolar intramuscular (BIM), monopolar intramuscular (MIM), and bipolar surface (BSF) (Figure 1). For the intramuscular recordings, low impedance tungsten electrodes were used (Frederick Haer and Co., Bowdoinham, ME, 250-µm shaft diameter, 2 – 4 mm of uninsulated length at the electrode tip). Starting from the PL, an intramuscular electrode was gently advanced through the skin, and placed in the muscle. Weak electrical stimulation from an isolation unit was used in order to test for appropriate placement (1 Hz pulse of 0.05 ms, 0.1 - 3.0 mA; Grass Instruments model SIU7). Once the intramuscular electrode placement was verified based on visual inspection of evoked response and palpation of identified distal tendons, a second electrode was placed just distal to the first electrode (~1.5 cm apart). The placement of the second electrode was verified as well. This process was repeated for the other target muscles, working radially, ending with the BR. A single reference electrode, placed on medial epicondyle of the humerus, was used for the monopolar intramuscular. Surface electrodes (4 mm diam Ag-AgCl) were attached to the skin near every intramuscular electrode site. The proximal electrode for each BSF channel was placed just proximal to the proximal intramuscular electrode in the same target muscle. Similarly, the distal electrode for each BSF channel was placed just distal to the distal
intramuscular electrode in the same target muscle (~3.0 cm apart). A Ag-lined grounding strap was moistened and attached to the subject, around the biceps, and connected to the common ground.

Electrodes were then connected to a differential amplifier system (Grass Model 8) through a multichannel electrode board. The electrode-selector panel of the amplifier system enabled routing specified electrodes as inputs to designated amplifiers. For monopolar recordings, the proximal intramuscular electrode in each muscle was paired with the single surface reference electrode as the two inputs to a particular amplifier (Figure 1). EMG signals were amplified (x1,000), band-pass filtered (30–1000 Hz), and digitally sampled (4000 samples/channel) (Cambridge Electronics Design, 1401 plus)

After all electrodes were in place, the hand was stabilized in a position of full supination by a padded horizontal handle attached to the platform. Subjects were then instructed to hold a continuous isometric contraction (at about 20% of maximum voluntary contraction) of all target muscles together. This was accomplished by instructing the subject, while holding onto the handle mounted across the palmar surface of the hand, to flex at the wrist, pronate at the wrist, and flex at the elbow. Subjects were instructed to hold the contraction steady for 60 s while EMG signals were recorded from the 12 channels (3 configurations x 4 muscles). The subject was given a few minutes of rest and then the process was repeated. Once the trials were recorded, locations of each electrode were marked, and distances between electrodes were measured.

Analysis. Data analysis was carried out off-line using custom-designed scripts in MATLAB. Waveform cross-correlation was performed to quantify the degree of cross-talk between muscle pairs (Winter et al 1994). Cross correlation analysis was performed
using 60 s of concurrently recorded pairs of unprocessed EMG signals, with time-shift increments of 0.25 ms for a maximum time shift of ± 40 ms. The peak correlation coefficient (r) of the resulting cross-correlation functions were identified for every pair of muscles recorded. This was done separately for each type of electrode configuration.

The peak correlation coefficient was then plotted as a function of distance (mm) between electrodes for each muscle pair and for each type of electrode configuration. These data (correlation coefficient vs. distance) were then fit with a falling exponential function of the form:

\[ r(d) = e^{-d/\lambda} \]

where \( r \) is the correlation coefficient, \( d \) is the distance between electrodes, \( e \) is the base of the natural logarithm, and \( \lambda \) is the length constant. From these fits, the length constants were determined for the three types of electrode configuration. The length constant provides a simple metric for representing the distance from the electrode configuration over which significant signal would likely be detected.

**Experiment 2**

It is possible that common signals detected using waveform cross-correlation analysis might arise in part due to actual shared neural signals acting upon different muscles rather than as a consequence of cross-talk. A widely used method to assess across-muscle neural coupling is to quantify spike-time correlations of motor unit pairs residing in different muscles (Bremner et al. 1991; Keen & Fuglevand, 2004; Hockensmith et al. 2005). The presence of nearly-coincident spiking (also referred to as short-term synchrony) that occurs more often than that expected due to chance for a pair of concurrently active motor units is taken as evidence that the motor units receive
common last-order neural inputs (Sears & Stagg 1976; Kirkwood & Sears, 1978). Therefore, to determine whether such neural coupling might partially contribute to correlated EMG signals recorded in Experiment 1, we carried out an additional set of experiments to characterize the degree of short-term motor unit synchrony within and across the four test muscles.

Subjects for these experiments (4 men, ages 23-55 yr) were seated in the same arrangement as described above. Motor-unit action potentials were recorded with high impedance (and selective) tungsten microelectrodes (Frederick Haer and Co., Bowdoinham, ME, 1- to 5-um tip diameter, 5- to 10-um uninsulated length, 250-um shaft diameter, ~200 kΩ impedance at 1,000 Hz after insertion). Two microelectrodes were inserted either into the same muscle or in two different muscles to record the activity of separate motor units on each electrode. Weak electrical stimulation was used initially to verify microelectrode placement in target muscle based on the evoked motor response, as described above. After electrical stimulation, electrodes were connected to differential amplifiers and the intramuscular EMG signals were amplified (x1,000), band-pass filtered (0.3–3 kHz; Grass Instruments), and displayed on oscilloscopes.

The intramuscular electrodes were gently manipulated during a weak contraction until action potentials of motor units could be clearly identified on each electrode. Once different motor units were identified on the two electrodes, subjects sustained weak contractions of the target muscles such that both units remained active. During the contractions, the forces exerted were not specified; rather, subjects were instructed to maintain the discharge of the two units at low rates. Intramuscular EMG signals were recorded for 5 minutes or until the motor unit action potentials could no longer be clearly
discriminated. Subjects received visual and auditory feedback on the discharge of the motor units and 1–2 minutes of rest between recordings. After each recording, the position of at least one and often both of the microelectrodes was adjusted until the action potentials of a presumably different motor unit could be identified. This occasionally involved removal of an intramuscular electrode and reinsertion at a new site. Electrical stimulation was performed when the electrode position was changed to verify electrode placement. Successive trials were performed for ~2 hours. Intramuscular EMG signals were digitally sampled at 2 and 18.5 kHz, respectively, using the data-acquisition and analysis system (Cambridge Electronics Design, Cambridge, UK).

**Analysis.** Spike times of individual motor units were identified based on motor-unit discrimination using a template-matching algorithm. An event channel representing the timing of discharges of accepted action potentials for a motor unit was generated. The discharge times of one unit, termed the event unit, were plotted relative to the discharge times of a second unit, termed the reference unit, to generate a cross-correlation histogram. In situations where more than one unit was detected on a given electrode, multiple cross-correlations could be performed with units detected on the other electrode. Cross-correlation histograms had 1-ms bin widths and included periods 100 ms before and 100 ms after the discharge of the reference unit.

A peak in the cross-correlation histogram around time 0 represents the synchronous firing of the two units greater than expected due to chance. The magnitude of the synchronous peak is thought to reflect the extent of shared last-order inputs to the two neurons (Sears & Stagg 1976). The cumulative sum procedure (cusum) was used to identify a synchronous peak in the cross-correlogram and was calculated by adding the
successive differences between the count of each bin and the baseline mean (Ellaway 1978). The baseline mean was calculated as the mean count in the first and last 60 ms of the cross-correlogram. A rise in the cusum near time 0 was used to delineate the peak in the cross-correlation histogram. Specifically, peak boundaries were determined as the bins corresponding to 10 and 90% of the difference between the minimum and maximum cusum values (Schmied et al. 1993). A peak in a cross-correlogram was considered to be statistically significant if the z-score (representing the number of SDs the mean count in the peak exceeded that in the baseline region) was $\leq 1.96$, associated with $p \leq 0.05$. The magnitudes of the peaks in the cross-correlograms were quantified using a synchronization index referred to as $k'$. The $k'$ index was calculated as the ratio of the mean number of counts in the peak to the baseline mean (Ellaway and Murthy 1985). When cross-correlograms did not exhibit clear peaks, the method described above for identifying the region of the histogram for the calculation of $k'$ was not reliable. Therefore, when cross-correlograms did not exhibit statistically significant peaks, $k'$ was automatically calculated for an 11-ms region of the cross-correlogram centered at time 0 (Semmler and Nordstrom 1995). All $k'$ values, regardless of the method used for calculation, were included in the analysis. A Kruskal-Wallis test was performed on $k'$ values for three different combinations of motor unit pairs: motor units residing in the same muscle, motor units residing in adjacent (neighboring) muscles, and motor units residing in non-adjacent muscles. Kruskal–Wallis is a nonparametric test based on the sum of the ranks and is used to compare three or more unpaired groups with different sample sizes. Values are reported as means $\pm$ SD with a probability of 0.05 selected as the level of statistical significance.
Results

Figure 2 shows an example recording from one subject. This recording shows a high signal-to-noise ratio, as the baseline electric potential of the recording is orders of magnitude smaller than the electric potential during the isometric contraction. The surface EMG signals had amplitudes in the range of ~ ± 1 mV, while intramuscular EMG signals varied between ~ ± 2.5 mV.

Figure 3 shows example waveform cross correlations obtained for different combinations of muscles. The first row shows cross-correlation analysis from the bipolar intramuscular (BIM) EMG signals, the second row shows cross-correlation analysis from the bipolar surface (BSF) EMG signals, and the bottom row shows cross-correlation analysis from the monopolar intramuscular (MIM) EMG signals. The columns are arranged by increasing numbers of intervening muscles. This example shows two trends. First, from top to bottom (i.e., from BIM to MIM), the amplitude of the peak cross-correlation value increases. And second, from left to right (i.e. with increased spacing), the amplitude of the peak cross-correlation value decreases. There was one exception to this latter trend. The peak correlation increased when going from the pronator-brachioradialis (PRO-BR, 1 muscle apart) combination to the palmaris longus-flexor carpi radialis (PL-FCR, 2 muscles apart) combination. This likely occurred because the actual physical distance between PRO and BR was greater than that between PL and FCR (see Figure 1).

Figure 4 shows peak correlation coefficients as a function of distance between electrodes for all subjects and trials. The black circles represent data from the bipolar
intramuscular channels, the green triangle represent data from the bipolar surface channels, and the red circles represent data from the monopolar intramuscular channels. For each type of electrode configuration, the data were fit using nonlinear regression (see Methods). There was a significant exponential relation between correlation coefficient and distance for all three configurations: bipolar intramuscular ($R^2=0.97$; $p \leq 0.0001$), bipolar surface ($R^2=0.87; p \leq 0.0001$), and monopolar intramuscular ($R^2=0.37; p \leq 0.0001$). The length constant values, determined for each electrode configuration were 14.5 mm for bipolar intramuscular, 37.0 mm for bipolar surface, and 64.5 mm for monopolar intramuscular. Therefore, a monopolar intramuscular recording arrangement will likely detect signal over a relatively large volume, and therefore, would highly susceptible to cross-talk.

As mentioned in the Methods, it is possible that neural coupling across muscles could have contributed to the waveform correlations like that shown in Figure 3. To evaluate this possibility, we quantified the degree of short-term synchronization for pairs of motor units recorded in the same and in different muscles. Figure 5 shows a short time segment (~10 s) from a representative 5-min recording in which one unit (Unit 1) was recorded from the FCR muscle and the other motor unit (Unit 2) was detected in the PRO muscle.

From records like that shown in Figure 5, we generated cross-correlation histograms of the firing times of pairs of motor units. Figure 6 shows example cross-correlation histograms for four pairs of motor units. In this example, each histogram involves one unit recorded in the FCR muscle and one motor unit recorded in each of the
other muscles (BR, PRO, and PL), including the FCR. The label above each histogram indicates the muscle combination, and the $k'$ value for each histogram is indicated. Keep in mind, a $k'$ value of 1.0 indicates no synchrony. There was a reasonably high level of synchrony for the pair of motor units residing in the same muscle (FCR-FCR) whereas little detectable synchrony was seen for any of the across muscle combinations.

Overall, we recorded a total 64 motor unit pairs. Of these, 31 were from within the same muscle, 27 were from adjacent muscles, and 6 were from non-adjacent muscles. Figure 7 shows the mean (SD) $k'$ values for each of these muscle combinations. Average $k'$ was 1.59 ± 0.42 for pairs of motor units within the same muscle, 1.05±0.09 for motor unit pairs in adjacent muscles ($n=27$), and 1.06±0.07 for motor units in non-adjacent ($n=6$) muscles. ($n=27$). Using a Kruskal-Wallis test, there was a statistically significant effect of muscle combination on $k'$ ($P<0.001$). Post hoc analysis indicated a significant ($p \leq 0.001$) difference in $k'$ between motor unit pairs within the same muscle and motor units pairs in adjacent and non-adjacent muscles. However, no statistically significant difference was found in $k'$ ($p \leq 0.001$) between motor unit pairs in adjacent muscles and non-adjacent muscles.

**Discussion**

We used waveform cross-correlation analysis to determine the amount of electrical cross-talk present in four muscles across the palmar side of the forearm. We found the bipolar intramuscular EMG had the smallest detection range, containing the least amount of cross-talk, while the monopolar intramuscular had the largest detection range, containing the most amount of cross-talk. In order to verify the waveform cross-correlation results were attributed to electrical cross-talk, and not neural coupling, we
performed motor unit synchrony experiments. We used event cross-correlation analysis of the discharge times of a pair of motor units to estimate the extent of neural coupling, attributable to divergence of descending pathways providing common input across the pair of motor units. Only motor unit pairs within the same muscle showed such synchrony. This suggests only electrical cross-talk accounts for high cross-correlation values.

While a higher degree of cross-talk was expected for monopolar compared to bipolar configurations, such a large detection distance for monopolar intramuscular electrodes (length constant of ~ 65 mm) was surprising. Indeed, the monopolar intramuscular electrodes exhibited a substantially larger detection range than that provided by surface bipolar electrodes. As such, monopolar electrodes, even those implanted well within the core of a muscle, are highly likely to pick up activity from neighboring muscles.

Assessment of neural coupling based on short-term synchronization of motor unit activity indicated little such coupling across the muscles in the forearm. Therefore, the relatively high degree of waveform cross-correlation seen for interference EMG signals (Figures 3 and 4) recorded in separate muscles is most likely due to electrical cross-talk.

**Conclusion**

When deciding on which type of EMG-recording approach to use, monopolar recording should be avoided in most circumstances. If an EMG recording is to be recorded from muscles whose nearest neighbors are at least ~37 mm away from where electrodes might be situated, and the muscles sit just below the skin surface, then a bipolar surface electrodes would seem to provide relatively cross-talk free recordings.
Otherwise, intramuscular bipolar EMG electrodes should be used. While it has yet to be shown that cross-talk can be completely eliminated, using the proper EMG can reduce cross-talk. An EMG setup can be decided based on distances of target muscles, desired invasiveness and ease of the experiment. While bipolar intramuscular EMG is clearly the best option for minimizing cross-talk, it is still very invasive. BIM EMG also requires weak electrical stimulation for muscle verification, which adds time and work to the given experiment. Depending on the details of the experiment, surface EMG can still be a good option, as it is minimally invasive, and easier to use.
Figures

Figure 1. Experimental set up. A) schematic diagram depicting overall EMG recording arrangement. 2 intramuscular electrodes (black dots) and 2 surface electrodes (green circles) are on each muscle: BR – brachioradialis, PRO – pronator teres, FCR – flexor carpi radialis, and PL – palmaris longus. The proximal intramuscular electrode of each muscle and the reference electrode (blue rectangle) comprise the monopolar intramuscular (MIM) channels. Both intramuscular electrodes in each muscle comprise the bipolar intramuscular (BIM) channels, and both surface electrodes on each muscle comprise the bipolar surface (BSF) channels. B) photograph showing placement of electrodes in different muscles of one subject. BR – brachioradialis, PRO – pronator teres, FCR – flexor carpi radialis, and PL – palmaris longus.
Figure 2. An example recording of ~60 s of all 12 channels simultaneously in one subject. From bottom to top, the EMG configurations BIM - bipolar intramuscular, MIM – monopolar intramuscular, and BSF – bipolar surface are shown. Within the configurations, the channels are arranged by anatomy, from the radial to ulnar side: BR – brachioradialis, PRO – pronator teres, FCR – flexor carpi radialis, and PL – palmaris longus. The amplitude of the BIM and MIM channels are ~ ±2.5mV. The amplitude of the BSF channels are ~ ±1.0mV.
Figure 3. An example cross-correlation plot of every pairwise combination of muscles for each configuration for one subject. From top to bottom, the rows are arranged by configuration: BIM – bipolar intramuscular, MIM – monopolar intramuscular, and BSF – bipolar surface EMG. The columns are arranging by spacing of the muscles, BR – brachioradialis, PRO – pronator teres, FCR – flexor carpi radialis, and PL – palmaris longus, from left to right, by the closest together (1 apart; PL-FCR, FCR-PRO, and PRO-BR) to the furthest apart (2 apart; PL-PRO and FCR-BR. 3 apart; PL-BR). From top to bottom, the amplitude of the peak cross-correlation value increases and from left to right, the amplitude of the peak cross-correlation value decreases (with the exception of PRO-BR and PL-FCR; this is due to PRO-BR distance being greater than PL-FCR distance).
Figure 4. A peak cross-correlation vs. distance plot, with all subject data, and exponential fits. The black markers show bipolar intramuscular (BIM) data, the red markers show the monopolar intramuscular (MIM) data, and the green markers show the bipolar surface (BSF) data. The exponential fits from regression analysis are represented by solid lines, corresponding to the color of each configuration. The MIM data show high cross-correlation coefficient values, even at far distances, indicating a high degree of cross-talk. The BIM data show the lowest cross-correlation values, indicating the lowest degree of cross-talk.
Figure 5. An example recording of a pair of motor units in the brachioradialis (BR). This 10 s sample was taken from a trial lasting ~ 5 min. From top to bottom, the third and sixth trace show the intramuscular electromyographic (IEMG) signals detected by the two intramuscular electrodes in the target muscle. Shown above each IEMG signal are discriminated motor units (spikes) from the IEMG signals, from template matching software. Shown above the spike channels are the associated instantaneous discharge rates of the motor units detected in the IEMG signal. The insets shown to the right of the signal recordings show all the motor unit action potentials identified for each channel for this 10 s segment overlaid on a brief time scale. The discharge times of one unit relative to the other were then used to generate cross-correlation histograms.
Figure 6. Example cross-correlation histograms for pairs of motor units recorded in 4 different muscles. A measure of synchrony magnitude (k’) is indicated for each histogram. BR – brachioradialis, PRO – pronator teres, FCR – flexor carpi radialis, and PL – palmaris longus. Example histograms show cross-correlations from two units in the same muscle (FCR-FCR), one muscle apart (PRO-FCR and FCR-PL), and two muscles apart (BR-FCR). The highest k’ value is from pair of motor units within the same muscle. The adjacent and non-adjacent muscle pairs have k’ values close to 1.0, the magnitude of the baseline value.
Figure 7. Mean synchrony index for motor unit pairs in 4 different muscles. The motor unit pairs were characterized by their relative target muscle locations: within the same muscle, in two adjacent muscles, and in two non-adjacent muscles. Error bars shown represent 1 SD. Overall, pairs of motor units within the same muscle exhibit far greater synchrony than pairs of motor units in adjacent muscles or non-adjacent muscles.
References


