Comparing Upper Limb Intermuscular Coherence between Force- and Myoelectric-Control Tasks

Virginia Casasnovas Orus
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by

Virginia Casasnovas Orus

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Student number: 4622383
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Thesis committee: Dr. ir. A. C. Schouten, TU Delft, Supervisor
Dr. ir. M. van de Ruit, TU Delft, Supervisor
Prof. dr. ir. A. J. van der Veen, TU Delft, External committee member
Dr. W. Pan, TU Delft, External committee member

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Virginia
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Comparing Upper Limb Intermuscular Coherence between Force- and Myoelectric-Control Tasks

Virginia Casasnovas
Delft University of Technology

Synchronization between muscles caused by shared neural input has been proposed to have a functional significance in muscle coordination. In this study, intermuscular coherence (IMC), a measure of shared neural input to muscles, was used to explore the neural underpinnings of proximal arm muscle coordination in force- and myoelectric-control tasks. In force-control, the goal was to attain a force magnitude and direction, whereas in myoelectric-control, the goal was to have single or pairwise muscle activation. Muscle coordination was assumed to be more selective in myoelectric-control, being that this task imposed stricter requirements in muscle activation. Accordingly, the expected difference in muscle coordination would be reflected in IMC if shared neural input results from a neural strategy for motor control. The hypothesis was that when single muscle activation was required, IMC in myoelectric-control would be lower than in force-control. Opposite to this, when pairwise muscle activation was required, IMC would be higher in myoelectric-control. Therefore, the presence or not of shared neural input was thought to be the mechanism to correspondingly couple or decouple muscle activation. The main finding was that there was no significant difference in IMC between force- and myoelectric-control tasks for both single and pairwise muscle activation conditions. Furthermore, IMC in single muscle activation was higher than that of pairwise muscle activation conditions in both tasks. Finally, the bands in which IMC was increased were the alpha-band and lower part of the beta-band. These findings are interpreted as that neural constraints have a fundamental role in the distribution of shared neural input to proximal muscles, and thus to their coordination.

I. Introduction

Synchronization of activity reflects functional connections in the neuromuscular system [1, 2]. Evidence is that synchronized activity has been detected between motor cortex and muscle, entailing that cortical oscillations are effectively transmitted to muscles through the corticospinal tract. Moreover, this coupling is modulated with different stages of movement and so has been linked to a functional role in motor control [3–7]. Extending these findings, synchronization between muscles has also been found [3, 6, 8] and proposed to have direct implications in muscle coordination [9].

Neural input to muscles mainly reflects common input from the motoneuron pool, meaning that motoneuron activity is synchronized, which in turn leads to the observed correlated motor unit activity [10, 11]. The physiological origin of common neural input would be pre-synaptic neurons that branch and control different motoneurons at the spinal level [12]. These pre-synaptic neurons can be direct cortical projections to motoneurons, referred to as corticomotoneurons, or can be spinal interneurons. As well, it may be possible that pre-synaptic inputs are synchronized themselves without an underlying direct anatomical connection [6, 13]. In both cases, this would ultimately be reflected in motor unit synchronization caused by common neural input that can be given both within or between muscles.

In order to quantify the relationship between neural signals, coherence has been used as a correlation measure in the frequency domain [14, 15]. In particular, intermuscular coherence (IMC) measures the coupling between muscles, assumed to be caused by common neural input [6]. Regarding the frequency range at which IMC is typically detected, this corresponds to the alpha- (8-12 Hz), beta- (15-30 Hz) and gamma- (30-80 Hz) bands, and IMC has been used to demonstrate that common neural input at each band could be related to different neural origins [3, 6]. Significant IMC has been observed in a variety of tasks related to muscle coordination, involving either upper or lower limb muscles [16–21]. Modulation of common neural input with task has been found between finger muscles in the beta-band during precision-pincher [22, 23], as well as between homologous hand muscles in the alpha-band when high bimanual coordination was required [24]. This points towards a role of common neural input in synergistic coordination of distal upper limb muscles. Moreover, the increase of IMC in the gamma-band during motor adaptation to reaching in a force field indicates that neural synchrony could be a strategy for the optimization of muscle activation patterns [25]. Altogether, the presented studies concluded that common neural input between muscles, measured by IMC, was linked to a functional role in motor coordination. Thus, these findings point towards the importance of muscle synchronization which could result from a neural strategy for motor control.
In this study, an alternative way of exploring the neural underpinnings of muscle coordination by varying task feedback is proposed. The two types of visual feedback tested will involve force- and myoelectric-control (myoelectric refers to muscle electrical activity). In force-control, mechanical output, thus force magnitude and direction, will determine task success. In contrast, in myoelectric-control single or pairwise muscle activation will be necessary for task success. Both control methods will result in the activation of the same muscles. However, the required accuracy of muscle activation will be higher in the myoelectric-control task. Consequently, the muscle activation pattern in myoelectric-control will be more selective compared to force-control.

The idea behind this experiment is that the expected difference in muscle coordination between tasks will be reflected in IMC estimates if this is a neural strategy. The hypothesis is that when single muscle activation is required, IMC in myoelectric-control will be lower than in force-control, as decoupled activation is forced. Opposite to this, when pairwise muscle activation is required, IMC will be higher in myoelectric-control, as coupled muscle activation is sought. Thus, the presence or not of shared neural input may be the mechanism to correspondingly couple or decouple muscle activation when this is relevant for task success. Furthermore, it will be possible to explore differences in IMC between single and pairwise muscle activation conditions within tasks. Increased IMC for pairwise muscle activation is expected, again supporting that shared neural input is a mechanism for coupled muscle activation.

The considered muscles will be upper limb ones that span the shoulder and elbow joints. Most studies that have evaluated IMC between arm muscles have focused on hand or finger ones [22–24, 26], motivated by the fact that distal muscles can be independently controlled to a higher extent [27], which can be exploited to probe neural input in different synergistic muscle activation patterns. Therefore, this study adds novelty by assessing neural input to proximal muscles. Thus, the main aim will be to investigate changes in IMC between force- and myoelectric-control tasks, assessing whether there is a difference in common neural input reaching proximal upper limb muscles linked to different muscle coordination schemes.

II. Materials and Methods

A. Participants

Seven healthy voluntary participants were included in the study (aged 22–27 years, 5 females, 6 right-handed) with no self-reported history of neurological or orthopaedic condition or injury (a total of 12 participants performed the experiment but 5 were excluded due to changes in task protocol). The experimental protocol was approved by the Human Research Ethics Committee (HREC) of the Delft University of Technology. All of the participants gave written consent prior to the experiment.

B. Experimental setup

Participants were seated in an automobile chair in front of a screen (Fig. 1). Their right arm was placed on a rectangular surface, part of a weight compensation device (Saebo, Charlotte, NC, USA), which was adjusted until the shoulder was abducted at approximately 90 degrees. This was done so that the arm was relaxed and the subject did not have to actively hold it in the abducted position, which would require unwanted shoulder muscle activation. The shoulder was flexed at approximately 45 degrees and the elbow was flexed at 90 degrees, so that the wrist was in line with the shoulder. The subject wore a wrist brace that attached magnetically to a fixed force sensor (ATI mini 45, calibration SI-145-5, sensitivity 1/16 N, Apex, NC, USA). Thus, the arm and force sensor attachment were in a horizontal (xy) plane at shoulder level and subjects were instructed to keep their arm in this plane while performing the experiment.

Static force readings were made with a 6-axis force sensor and collected at a rate of 2048 Hz through a digital acquisition system (USB-6361, National Instruments, Austin, TX, USA). The force sensor was rotated to have its y-axis in line with the subject forearm, and its z-axis in line with the gravity vector. This positioning was selected so that shoulder torques generated forces in the sensor y-direction and elbow torques generated forces in the sensor x-direction. More specifically, shoulder flexion was responsible for positive y-direction forces and extension for negative ones. Moreover, elbow extension was responsible for positive x-direction forces and flexion for negative ones.

Electromyography (EMG) was recorded with differential surface electrodes in bipolar configuration (Delsys, Boston, MA, USA) from 7 arm muscles: flexor carpi radialis (FCR), extensor carpi radialis brevis (ECRB), biceps brachii (BB), triceps lateralis (TLat), triceps long head (TLH), deltoïd anterior (DA) and deltoïd posterior (DP). The choice of muscles was motivated by the need of 2 antagonistic shoulder muscles (flexor-DA and extensor-DP) and 2 antagonistic elbow muscles (flexor-BB and extensor-TLH) that would be used to control the myoelectric task. An additional elbow extensor (TLat) was included as an alternative control muscle in substitution for TLH. Two wrist muscles (flexor-FCR and extensor-ECRB) were included to check for the use of the wrist in force application, given that only shoulder and elbow torques should contribute to this. EMG signals were analog low-pass filtered at 277 Hz and digitized at 1024 Hz.
C. Experimental protocol

The behavioral task consisted on a center-out-hold scheme in which subjects had to move a central green circular cursor to a given red circular target in the periphery and hold the cursor within it. After the hold period, subjects were instructed to relax and bring the cursor back to the center, finishing the trial (Fig. 2a,b). Control signals for the cursor varied depending on task type: static force for force-control and EMG from a pair of muscles for myoelectric-control. Accordingly, the target value for the force-control task was a force level in two directions, and the target value for the myoelectric-control task were certain EMG levels of two muscles. For both tasks, the subject had to remain in a static position, so cursor movement was done through isometric muscle contraction. There was a correspondence in target direction between tasks, and the target directions shown on screen were the same ones for both tasks. This entailed that for each target, the same muscles were activated in both tasks. Each target represented the line-of-action of a single muscle or the combined action of two muscles (Fig. 2c, Table I), correspondingly referred to as single-muscle or joint-muscle targets. Here, the line-of-action of a muscle will be defined as the force direction at the force sensor that would result from the contraction of that muscle. Before starting the experiment, resting EMG levels were measured to make sure the subject had his arm relaxed, else the setup was readjusted. Subjects first performed the force-control task followed by the myoelectric-control one. Breaks between blocks of trials were of at least 1 min to prevent muscle fatigue.
cursor radius increased by three halves of its initial value following current z-force value. The resting z-force value, namely shoulder adduction and abduction. In this way, unwanted arm movements outside the horizontal plane, in the vertical direction, which could be caused by each block, the resting z-force caused by arm weight.

To compute cursor position, force readings were low pass-filtered at 3 Hz (Butterworth, 2nd order) and a moving average over a window of 500 ms was implemented. As the reference axes of the force sensor were rotated 45 deg counterclockwise compared to the screen reference, the cursor position was rotated by this amount. The refresh rate for cursor position was 5 Hz. Four blocks of 60 trials each were performed where target direction was varied randomly. Before starting each block, the resting z-force caused by arm weight was measured. This was done to limit force production in the vertical direction, which could be caused by unwanted arm movements outside the horizontal plane, namely shoulder adduction and abduction. In this way, the resting z-force value \( F_{z,0} \) was used to scale the current z-force value \( F_z \) to adjust the cursor radius following \( r_{\text{cursor}} = r_{\text{cursor},0}(1 + |F_z/F_{z,0}|) \). When the cursor radius increased by three halves of its initial value \( (r_{\text{cursor},0}) \), the trial was considered a fail. After finishing all blocks, EMG data was pre-analyzed to set the target levels and target tolerances for the myoelectric-control task.

2. Myoelectric-control task

In the second task cursor position was determined with the activity of a pair of muscles. Three pairs of control muscles were tested in separate blocks: TLH-DA, DA-BB and BB-DP. One muscle of the pair acted on the elbow and the other on the shoulder, so that their joint activation did not entail co-contraction. There were 3 target directions for each block, thus for each control muscle pair. Two of these corresponded to the two perpendicular lines-of-action of the two muscles, and the targets were referred to as single-muscle targets. The third direction was in between these two and corresponded to the joint muscle action, referred to as joint-muscle target (Fig. 2c). The target EMG levels were set based on the maximum average EMG level over the hold period of the force-control task across target directions (Fig. 3a). This was calculated for each muscle individually, and this maximum typically coincided with the target direction that matched the muscle line-of-action. Given a control muscle pair, for the single-muscle target in line with the muscle line-of-action, the target EMG level of that control muscle was set to the calculated force-control maximum, whereas the target EMG level of the complementary control muscle was set to zero plus minus a tolerance. Similarly, for joint-muscle targets the target EMG level for each muscle in the pair was set to 70% of their force-control maximum EMG level. This was selected so that both muscles had equal contribution in reaching the joint-muscle target.

Target tolerance depended on target direction (Fig. 3b), although it was higher than for the force-control task due to increased cursor noise. For the single-muscle targets, and for a given control muscle, the target tolerance was set to 20% of the target level in the direction of the control muscle. The tolerance in the direction of the complementary control muscle was set according to the minimum average EMG level over the hold period of the force-control task across targets, that is, the minimum activation level during the hold period possible for the complementary control muscle. Taking this into account, if the tolerance in the direction of the complementary control muscle was higher than 20% of the target level, the target had an elliptical shape, allowing more variability in the direction of the complementary control muscle. The allowed movement time until the cursor reached the target was of 3 s and the hold time the cursor had to remain in the target was of 5 s. The cursor could leave the target for 40% of the hold time, which mainly occurred because of EMG noise. This permission was incorporated so that the trial...
was successful when factors unrelated to the subject degraded their performance.

D. Data analysis

Successful trials from all blocks were grouped per target for each task. To discard trials in which the subjects were possibly learning the task, the first 10 trials of force-control and first 5 of myoelectric-control of each block were not considered for analysis. Unless otherwise stated, analysis was done on the hold period data, excluding the first second of hold time to avoid including any fine-tuning or overshoot correction of the cursor position at the target. For time-domain analysis, pre-processed data was trial-averaged for each target, while for frequency-domain analysis, pre-processed data from each trial was appended for each target. As well, targets that did not have a minimum number of trials were not included in the frequency-domain analysis as this could bias the results (see Appendix A for a table of subject trial numbers and corresponding target inclusion).

1. Pre-processing

Force data was down-sampled to 1024 Hz to match the EMG sampling rate, and force and EMG data were synchronized using the trigger pulse. Force measurements were low-pass filtered at 5 Hz (Butterworth, 2nd order). EMG data was high-pass filtered at 10 Hz (Butterworth, 2nd order) and band-stop filtered at 50 Hz to remove power line noise. After this, rectification was applied as this step is thought to enhance motor unit firing frequency information over that of motor unit action potential shape [29, 30]. Finally, the rectified EMG data was low-pass filtered at 500 Hz (Butterworth, 2nd order). In addition, when considering EMG for time-domain analysis, each muscle signal was normalized by the maximum average EMG value of that muscle over the hold period across targets in force-control, similar to how single-muscle targets were defined for the myoelectric-control task.

2. Intermuscular coherence

IMC was used to assess the linear relation between a pair of muscle signals in frequency. The coherence measure is bounded, where a value of 0 indicates no linear relation between signals and a value of 1 would represent a perfect linear relationship in a noiseless situation [14]. As well, coherence is a normalized measure, entailing that despite differences in activation level between muscles it is possible to compare across muscle pairs. High IMC would be an indication of common input to motor units of two muscles. IMC was estimated with windowed overlapping hold period segments of the trial-appended pre-processed data (50% overlap, Hamming window), ensuring that segments belonged to a single trial. The segments were of 1 s each, which resulted in a frequency resolution of 1 Hz. The auto-spectral densities $(S_{xx}$ and $S_{yy}$) and cross-spectral density $(S_{xy})$ were calculated for each segment and averaged to calculate the coherence estimate (Eq. 1). This estimate was smoothed with a 3-point moving average.

$$C_{xy}(f) = \frac{|S_{xy}(f)|^2}{S_{xx}(f) S_{yy}(f)}$$

The threshold over which coherence was significant was
determined through \( CL = 1 - \alpha^{1/(L-1)} \), where \( \alpha \), the confidence level, was set to 0.05 and \( L \) was the number of overlapping segments used [14].

Additionally, coherence values were converted into z-scores by applying the Fisher transformation (Eq. 2), so that the values were normally distributed with a standard deviation of approximately 1. This transformation was applied to be able to compare values across participants [14, 31]. As well, in order to have a one-sided confidence level of 95%, the significance threshold was set to the corresponding z-score of 1.65. Lastly, z-scores were averaged across subjects for each muscle pair. In this case, the significance thresholds for each target were set with a composite score of 1.65 depending on the number of subjects that attained the given target according to Stouffer’s z-score method [3, 22].

\[
Z_{xy}(f) = \sqrt{2L} \tanh^{-1}\left(\sqrt{C_{xy}(f)}\right) \tag{2}
\]

The area of significant z-score was calculated for the alpha-, beta- and gamma-frequency bands. The corresponding areas were normalized by the length of the frequency band. For all estimates, the muscle pairs that were evaluated were the three control muscle pairs, and for each of these the evaluated targets were the two single-muscle targets of each muscle and their joint-muscle target.

3. Outcome measures

The time-domain measures that were analyzed across subjects included force magnitude mean and force magnitude coefficient of variation (CV), as well as normalized EMG mean. Force magnitude CV was calculated to assess its consistency across trials, thus consistency in task performance, as it would reflect the variation in force magnitude across trials for each target. Moreover, force traces in the xy-plane and EMG traces in time were visualized for a representative subject. Analysis of time-domain measures was done in order to assess task performance and compare across tasks.

The frequency-domain measures included were IMC spectra and the normalized areas of significant IMC in the alpha-, beta- and gamma-bands. Both of these were expressed in z-scores and averaged across subjects. These measures are the primary outcome of this study, as they will reflect common neural input, which will be compared across tasks, targets and frequency bands.

4. Statistical tests

Statistical analysis was performed on force magnitude mean and CV, normalized EMG mean of control muscles and normalized area of z-score values (SPSS Statistics, version 25, IBM, Armonk, NY, USA). In order to select the test, first normality of the data was assessed with a Shapiro-Wilk test (\( \alpha = 0.05 \)). Given that normality was violated in more than one case, the Wilcoxon signed-rank test (\( \alpha = 0.05 \)) was selected. This is a non-parametric test that accounts for non-normality of data and is equivalent to the dependent t-test. Difference between tasks was probed for force magnitude mean and CV, normalized EMG mean of control muscles and normalized area of z-score values. As well, difference between the single-muscle targets and the joint-muscle target within each task for each control-muscle pair was evaluated for the normalized area of z-score measure.

III. Results

A. Time-domain analysis

1. Force data

Trial-averaged force traces from trial start to the end of the hold period in the xy-plane were inspected for each subject to assess task performance (a representative subject is shown in Fig. 4). Force traces were close to straight with some deviation during the hold period, expected for the center-out-hold scheme.

![Trial-averaged force trajectories from trial start to hold end in the xy-plane.](image)

FIG. 4. Trial-averaged force trajectories from trial start to hold end in the xy-plane. Traces correspond to a representative subject. Blue lines correspond to force-control (FC) and red lines to myoelectric-control (MC). Green circles indicate trial start and magenta circles indicate the end of the hold time. Magenta circles are labeled with the corresponding target direction. Force magnitude is higher in MC at targets 45, 90 and 135 deg. For all of the targets except 90 deg, the force directions in MC are rotated with respect to FC.
Force magnitude mean and CV during the hold period across subjects was examined for each target (Fig. 5). Force magnitude mean was similar or higher in myoelectric-control than in force-control depending on the target, and for targets 45 and 225 deg, which correspond to joint-muscle targets, this difference was statistically significant (target 45 deg: $Z = -2.366, p = 0.018$; target 225 deg: $Z = -2.197, p = 0.028$). Force magnitude CV was higher in myoelectric-control for all targets and this difference between tasks was statistically significant for all targets (target 270 deg: $Z = -2.023, p = 0.043$; rest: $Z = -2.366, p = 0.018$).

FIG. 5. Polar plots of force magnitude mean and force magnitude CV across subjects. Blue circles correspond to force-control (FC) and red circles to myoelectric-control (MC). Error bars indicate SEM. Statistical significance denoted by *p<0.05. The difference between force magnitude between tasks is significant for joint-muscle targets 45 and 225 deg. For all targets force magnitude CV is significantly higher for MC than FC.

2. **EMG data**

Trial-averaged smoothed EMG traces from trial start to the end of the hold period are shown for a representative subject (Fig. 6). Normalized EMG traces were very similar across subjects. EMG traces match a center-out-hold scheme, starting at a low level and increasing to the target level during hold period for the targets of interest of each muscle.

The average of the EMG traces over the hold period was computed and was again averaged across subjects (Fig. 7). All control muscles showed tuning in activation level across targets, peaking at the target direction that corresponded to the muscle line-of-action, so the equivalent single-muscle target. This muscle line-of-action was different for each muscle, except for TLH and DP, which accordingly were not tested as a control pair as they both give rise to shoulder extension. Comparing normalized EMG values between tasks, three different cases could be observed. For the single-muscle target corresponding to the muscle line-of-action there was no difference in EMG value between tasks. For joint-muscle targets, a significant difference was found for DP (target 225 deg: $Z = -2.366, p = 0.018$). Finally, significant differences were also found for DA and BB at the single-muscle target of the complementary control muscle (DA, target 0 deg: $Z = -2.197, p = 0.028$; BB, target 270 deg: $Z = -2.023, p = 0.043$).

FIG. 6. Trial-averaged smoothed EMG traces from trial start to hold end. Traces correspond to a representative subject. Smoothing has been done with a 200-point moving average. EMG traces are normalized by the single-muscle target level, so the maximum average EMG value for the hold period across target directions in the force-control task. Rows represent target directions. Columns represent control muscles. Blue traces correspond to force-control (FC) and red traces correspond to myoelectric-control (MC). For each muscle, EMG traces start at a low level and increase to the target level for its single- and joint-muscle target directions, following the center-out-hold scheme.

Furthermore, to assess the possible contribution of wrist movements to task performance, the normalized EMG values of non-control wrist muscles (ECRB and FCR) across subjects were visualized (Fig. 8). For these muscles, tuning across targets was not marked, making their possible contribution distributed across targets. Regarding myoelectric-control, the activity of these muscles did not control the cursor, so their activity was irrelevant for task performance.

FIG. 8.
FIG. 7. Average normalized EMG values over the hold period across subjects. EMG values are normalized for each subject by the single-muscle target level, so the maximum average EMG value for the hold period across target directions in the force-control task. Rows represent control muscles. Blue circles correspond to force-control (FC) and red circles to myoelectric-control (MC). Error bars indicate SEM. Statistical significance denoted by *p<0.05. Dotted lines denote a normalized EMG value of 1 and 0.7 to indicate the EMG target level in MC. Light gray areas mark the range of targets of interest for each muscle. Muscle activation is tuned with target direction and is highest at the target corresponding to the muscle line-of-action for both tasks, so the equivalent single-muscle target. For each control muscle, differences between FC and MC are greatest for joint-muscle targets and for the single-muscle target of the complementary control muscle.

B. Frequency-domain analysis

Averages of z-score spectra across subjects are reported for the three control muscle pairs (TLH-DA, DA-BB and BB-DP) (Fig. 9). All three of these muscle pairs showed significant coherence across the frequency spectrum for the corresponding targets of interest. Furthermore, a peak of IMC for both tasks between 8-20 Hz was present for all control muscle pairs. Three cases were found regarding the relation in average z-score between tasks, one for each control muscle pair.

TLH-DA showed higher IMC in myoelectric-control for both single-muscle targets, and no difference in IMC between tasks for the joint-muscle target. DA-BB showed no difference in IMC between tasks for either single- or joint-muscle targets. Lastly, BB-DP showed higher IMC in force-control for both single-muscle targets. Regarding the joint-muscle target of this pair, there was higher IMC in myoelectric-control in the alpha- and beta-bands and no difference between tasks in the gamma-band.

To quantify the relation between tasks, the area of significant z-score in each frequency band was averaged across subjects (Fig. 10). Moreover, in order to better visualize joint-muscle targets, areas for these are separately plotted (Fig. 11). The relation of IMC between tasks followed the same trend as the average z-score spectra in each frequency band. However, IMC differences between tasks were not significant in any case. Differences between the area of significant z-score of single-muscle and joint-muscle targets within each task were also assessed. Single-muscle targets had a significantly greater area than the joint-muscle target in at least one frequency band for TLH-DA in the force-control and myoelectric-control task (TLH-DA, alpha,
FIG. 9. **Average z-score spectra across subjects.** Targets for which the subject had not attained the minimum number of trials for coherence analysis are not included in the average to avoid biasing the results. Columns represent control muscle pairs and rows represent target directions. For each control muscle pair only relevant target directions are shown. Blue traces correspond to force-control (FC) and red traces correspond to myoelectric-control (MC). Dashed lines represent the confidence level above which coherence is significant; blue lines correspond to FC and red lines to MC. Light gray areas mark the alpha-band (8-12 Hz) and dark gray areas mark the beta-band (15-30 Hz). All three control muscle pairs show significant coherence for the corresponding three targets of interest. For TLH-DA and BB-DP there appears to be an IMC peak between 8-20 Hz for both tasks. For DA-BB this peak is less clear. TLH-DA shows higher IMC in MC for both single-muscle targets, and no difference in IMC between tasks for the joint-muscle target. DA-BB shows no difference in IMC between tasks for either single- or joint-muscle targets. BB-DP shows higher IMC in FC for both single-muscle targets, and higher IMC in MC for the joint-muscle target in the alpha- and beta-bands.

![Graphs showing z-score spectra for different targets and tasks.](image)

**target 45-90 deg, FC:** \(Z = -2.366, p = 0.018\), beta and gamma; **target 45-90 deg, FC:** \(Z = -2.197, p = 0.028\), MC: \(Z = -2.023, p = 0.043\).

**IV. Discussion**

The purpose of this study was to compare shared neural input, measured by IMC, between proximal arm muscles in order to investigate neural coupling in a force- and myoelectric-control task. The experiment was designed to have task-dependent muscle coordination, so subjects had to follow stricter requirements in muscle activation in the myoelectric-control task compared to the force-control one. Results showed that significant coherence was found for each of the muscle pairs in both tasks, confirming the existence of shared neural input between elbow and shoulder muscle pairs. However, the main finding was that there was no significant difference in IMC between force- and myoelectric-control tasks for both single- and joint-muscle targets. Furthermore, IMC in single-muscle targets was higher than that of joint-muscle targets in both tasks. Finally, the bands in which IMC was increased were the alpha-band and lower part of the beta-band.

**A. Force and EMG task modulation**

Force output was investigated and force magnitude across subjects was significantly higher in myoelectric-control for two of the three joint-muscle targets. The reason behind this result are differences in target EMG levels of joint-muscle targets, since the EMG levels acquired for these in force-control were not set as the target EMG levels in myoelectric-control. As for force directions, when viewing the force traces in the horizontal
**FIG. 10. Average normalized area of significant z-score across subjects.** Areas are normalized by the length of the frequency band. Targets for which the subject had not attained the minimum number of trials for coherence analysis are not included in the average to avoid biasing the results. Columns represent control muscle pairs and rows represent frequency bands over which the area was calculated. For each control muscle pair only relevant target directions are plotted. Blue circles correspond to force-control (FC) and red circles correspond to myoelectric-control (MC). Error bars indicate SEM. Statistical significance denoted by *p<0.05; blue stars correspond to FC and red stars to MC. The relation of the area between tasks varies with the muscle control pair, but there is no significant difference in any case. The trend over frequency bands is that TLH-DA shows higher IMC in MC for both single-muscle targets, and no difference in IMC between tasks for the joint-muscle target. DA-BB shows no trend in difference in IMC between tasks for either single- or joint-muscle targets. BB-DP shows higher IMC in FC for both single-muscle targets, and higher IMC in MC for the joint-muscle target in the alpha- and beta-bands. Comparing single- and joint-muscle targets within tasks, TLH-DA and BB-DP shows higher area for single-muscle targets than for joint-muscle targets; DA-BB shows no trend in the relation between targets.

**FIG. 11. Average normalized area of significant z-score across subjects in each frequency band for joint-muscle targets.** Areas are normalized by the length of the frequency band. Targets for which the subject had not attained the minimum number of trials for coherence analysis are not included in the average to avoid biasing the results. Columns represent control muscle pairs. Blue circles correspond to force-control (FC) and red circles to myoelectric-control (MC). Error bars indicate SEM. Alpha-band z-score area for joint-muscle targets is highest among frequency bands. The relation of the area between tasks varies with the control muscle pair, but there is no significant difference in any case (p<0.05).
plane for each subject, myoelectric-control force traces were rotated with respect to the force-control ones in a number of cases. This may be because the line-of-action of the muscle did not exactly match the force target direction, and the EMG target was achieved faster in the rotated directions. Supporting this view, the force output in a horizontal plane generated by each of the control muscles used in this study have been estimated from EMG recordings, while the subject was in a similar position [32]. Results showed that the generated force directions did not exactly match the predefined force target directions of this study, so the observed rotation in force directions between tasks could be expected.

Moreover, force magnitude CV across subjects was significantly higher in myoelectric-control, meaning that there was higher variability in force magnitude between trials. This reveals that target acquisition in myoelectric-control was less restricted and that similar muscle activation could cause a range of force magnitude outputs, which is reasonable as arm muscle force output has been shown to be broad [33, 34]. In addition, it has been claimed that myoelectric-control tasks dissociate task constraints from natural limb biomechanics [35], as opposed to force-control, in which biomechanical constraints determine force output. Thus, this could also contribute to higher variability in force magnitude in the myoelectric-control task. Although discrepancies in force output could be perceived to alter IMC across tasks, it has been seen that changes in force level alone did not result in changes in IMC between hand muscles [23, 36, 37]. This implies that variations of force level within a range are not determinant when assessing common neural input to muscles in relation to muscle coordination.

Regarding muscle activation levels, all muscles showed broad tuning in activation in both tasks, peaking at the corresponding single-muscle target and remaining active for the rest of the targets, parallel to previous findings in a myoelectric-control task of hand and upper-arm muscles [35]. Single-muscle target EMG levels corresponding to the control muscle line-of-action matched between tasks. Minor differences were due to higher target tolerance in myoelectric-control, which allowed for a greater range in target EMG levels. However, for joint-muscle targets there was more deviation between tasks, although these were only statistically significant for DP. Following the same reasoning as with force magnitude, differences in EMG levels were found because the joint-muscle target EMG level was not set to the EMG levels of the individual muscles of the joint-muscle target acquired in force-control, but were set to have 70% of their single-muscle target EMG level. In this way, both muscles contributed equally to reaching the joint-muscle target, imposing coupled muscle activation. The higher EMG level in joint-muscle targets corresponded to either force- or myoelectric-control, depending on the muscle. The greatest difference was found for DP, which also had the largest range in EMG level, entailing that differences between target EMG levels in force-control were heightened. As well, higher target tolerance in myoelectric-control may have also contributed to the difference in joint-muscle EMG level between tasks. Finally, single-muscle target EMG levels corresponding to the complementary control muscle line-of-action were lower in myoelectric-control in all cases except for TLH. This could be expected as single-muscle targets required the activation of one control muscle and the relaxation of the other. Hence the relaxed muscle was imposed to have a lower EMG level than its EMG level at that target in force-control, enhancing decoupled muscle activation. The reason why TLH had a higher EMG level in myoelectric-control than in force-control could be the variability in resting EMG levels of its complementary control muscle DA. This resulted in difficulty to successfully attain the single-muscle target of this DA. Overall, it could be argued that differences in muscle activation level across tasks could be the cause of differences in IMC. However, differences in muscle activation level within a range have not been found to be predictive of IMC modulation, but this modulation was related to the type of muscle coordination [22].

### B. IMC task modulation

IMC of three control muscle pairs (TLH-DA, DA-BB and BB-DP) was evaluated in the target directions of interest for each pair, the single-muscle targets of each muscle and their joint-muscle target. Results show that there was no significant difference in IMC between tasks in any of these targets for any of the control muscle pairs. Focusing on TLH-DA, IMC in myoelectric-control was higher than in force-control for single-muscle targets, while the opposite happened for joint-muscle targets. For DA-BB, IMC in the alpha-band follows the same trend as for TLH-DA. However, in the beta- and gamma-bands there was higher or equal IMC between tasks for all targets, which leads to the absence of a trend in IMC task modulation among frequency bands. Finally, for BB-DP, there was higher IMC in the force-control task for single-muscle targets, and higher IMC in the myoelectric-control task for joint-muscle targets.

Regarding the functional role of shared neural input, our hypothesis was that joint-muscle targets would have higher IMC in myoelectric-control, as this would bind two muscles and functionally couple them to achieve the same goal. The opposite would happen in single-muscle targets, so neural input would be directed to each muscle individually in order to decouple their activity. This reasoning was proven in a study that linked strong positive EMG amplitude correlation with enhanced IMC for a pair of hand muscles, which
occurred when the task required muscle coordination to be synergistic [22]. Here, BB-DP followed the proposed argumentation but TLH-DA did not. An alternative interpretation derived from the IMC results of TLH-DA is that for joint-muscle targets in myoelectric-control, each muscle had to be independently controlled in order to reach the target accurately. Thus, although muscle activation was coupled, the underlying neural input was not [22], being directed to each muscle separately to control each of them more accurately. For single-muscle targets, the muscle activation level of one control muscle was similar to that of force-control, and the complementary control muscle had to remain relaxed, entailing that less neural input would be received to this complementary muscle overall. Hence, the total amount of shared neural input in the single-muscle target would be lower.

The reported discrepancy in IMC task modulation with muscle pair could be due to neural constraints, thus there is an existing distribution of neural pathways that allow for the connection between motoneurons of different muscles or not. This leads to the idea that neural input will be directed in different ways depending on the muscle and in turn may have different functional roles. Moreover, the muscles forming a control pair acted on different joints, and these were proximal, not naturally requiring such fine control as distal muscles [27]. Evidence of this reduced flexibility in forming coordination patterns has been seen in a similar myoelectric-control task, which concluded that neuromuscular constraints had a greater effect on proximal muscles compared to distal ones [35]. In this way, the absence of significant differences in coherence across tasks could be explained by the reduced ability to modulate independent neural input to proximal muscles. Altogether, it can be interpreted that shared neural input depends on the underlying neural pathways.

C. IMC target modulation

Regarding IMC differences between single- and joint-muscle targets, for two of the muscle pairs (TLH-DA and BB-DP), IMC for joint-muscle targets was lower than for single-muscle targets in most of the frequency bands for both tasks. For DA-BB, IMC target modulation depended on the frequency band. Note that IMC was low overall for this pair and showed no consistent task modulation, so results from DA-BB can be considered inconclusive.

Higher IMC in single-muscle targets seems to be counter-intuitive, as no activation coupling should be necessary. As well, higher IMC for single-muscle targets appears to be in contrast with studies that have found significant coherence between muscles that were required to act in synergy to achieve a goal [22, 23, 26]. However, it is important to mention that these studies focused on distal muscles that allow for independent control, so shared neural input may have served as a binding mechanism for coupled muscle activation. This may not be the case for proximal muscles, for which the neural input may be already widespread in single-muscle target directions. Following this line of reasoning, the required high activation level of one of the control muscles could have caused its increased neural input to spill over to the complementary one [38], causing higher IMC in single-muscle targets. Another theory is derived from the fact that muscles also receive afferent input that can serve as a mechanism to inhibit their contraction [39]. In this way, for single-muscle targets IMC may be detecting this inhibitory input to the complementary control muscle in addition to the increased excitatory input to the other control muscle. As well, connecting to the task modulation interpretation, it may be possible that joint-muscle targets required independent muscle control to gain accuracy, which would be in line with IMC target modulation results. Again, IMC target modulation can be attributed to neural constraints, so depending on the muscle pair, coupled muscle activation does not necessarily arise from shared neural input.

D. IMC frequency band modulation

Comparing IMC estimates between frequency bands has been done to relate neural input to different neural sources. In this experiment, IMC was highest between 8-20 Hz, comprising the alpha-band and the lower part of the beta-band. The observed alpha-band IMC could be due to spinal contribution [40, 41], which has been related to the generation of gross movements as the ones required in this study. Furthermore, the tasks required a degree of independent muscle control to achieve the force or EMG target accurately. Direct corticospinal pathways are responsible for fractionated muscle control, mostly innervating distal muscles [27], but also projecting to proximal ones [42]. Thus, the presence of lower beta-band IMC found in this study, linked to a cortical origin [4, 5], could be the result of corticospinal pathways trying to control proximal muscles in an independent way. As for the gamma-band, this has been linked to function instead of a particular neural source. IMC in this band did not give clear information about possible shared neural input, as no consistent peaks over this band across targets were seen. This appears reasonable as it is known that corticospinal binding shifts to the gamma band during near maximal force production [43] or during dynamic force production [44], which was not the case in this experiment as low static forces were required. Altogether, it can be inferred that parallel neural pathways act in the control of proximal muscles. Nevertheless, future work including corticospinal coherence analysis, quantifying neural coupling between cortex and muscle, would provide more information on the presumed
cortical contribution.

E. Muscle coordination strategies

Within the context of muscle coordination, muscle synergies have been proposed to explain the low-dimensionality of muscle activation patterns [45], defined as a group of muscles that share a function and would simplify the selection of motor commands from the central nervous system. However, the neural underpinnings of muscle synergies remain unresolved. Most studies on muscle synergies have used dimensionality reduction techniques to extract the low-dimensionality of muscle activation patterns [46], but this method does not directly reflect any deliberate strategy of the neural system. Thus, the use of IMC as a measure of shared neural input could bring light to the origin of muscle synergies, as shared neural input could be their underlying mechanism. Based on the found differences in IMC task modulation between muscle pairs, it can be inferred that proximal muscles are part of fixed muscle synergies arising from neural connectivity. Moreover, the emergence of neural synergies through a myoelectric-control task was previously found for distal muscles only [26]. Therefore, it may be that the task-dependent synergies may be restricted to distal muscles that have greater flexibility in their control.

F. Limitations

To begin discussing the limitations of this study, one of these was the small sample size, as only 7 subjects were included for analysis. Consequently, this had an effect on significance tests, perhaps being one of the reasons why no significant difference in IMC was found between tasks. Experimental issues that were encountered include variability in resting EMG values, as these were sensitive to changes in posture. This could impair performance in the myoelectric-control task, being that resting EMG values were used for cursor position calculation. Although posture should have remained invariant throughout the experiment, this was not always the case and minor changes between blocks were possible. However, when this happened, the subject was re-positioned until the resting EMG values were close to zero again. Heightened sensitivity to posture was characteristic of DA, leading to increased difficulty to acquire its targets in the myoelectric-control task and the resulting impairment in force and EMG data between tasks. Inappropriate electrode placement on DA could be one of the causes. Another reason could derive from the chosen subject positioning, namely abducting the shoulder 90 degrees could cause activation of surrounding stabilizing muscles that affected DA EMG, even though subjects perceived they were in a relaxed position with their arm being held by the weight compensation platform.

Another concern related to experimental design was the assumption that the two types of visual feedback led to different requirements in muscle coordination linked to different neural input distribution. As well, not only did the two tasks have different visual feedback but the constraints were different, being that in the myoelectric-control task biomechanical constraints were bypassed [35]. Accordingly, this would lead to think that the coordination strategy employed would be different. Nevertheless, muscle coordination was not explicitly measured, thus it is not possible to affirm that myoelectric-control led to selective muscle control as this was not quantified. In this manner, perhaps the lack of significant differences between tasks would stem from the lack of muscle coordination difference.

Finally, it is important to comment on the validity of IMC as a non-invasive measure of shared neural input. One concern is electrode placement, as this is important in order to maintain EMG consistency across subjects and avoid cross-talk. To account for these issues, electrodes were placed according to literature guidelines [47]. Furthermore, cross-talk was not apparent as for each muscle pair coherence was not high over the entire spectrum. Another issue is that coherence is highly sensitive to EMG pre-processing and the use of rectification in particular has been greatly debated. Whereas some studies argue that rectification is a non-linear transformation that distorts the frequency content of EMG [48, 49], others are in favour of its application as it has been seen that rectification can enhance motor unit firing information [29, 30, 50, 51]. In this study, rectification was applied because it has been seen that the effect of this step depends on the amount of amplitude cancellation of the EMG signal, which changes with muscle activation level [52, 53]. Consequently, as similar muscle activation levels were required across tasks, rectification alone would not be the cause of IMC modulation between tasks. Further potential bias in the coherence estimate could derive from the number of segments used for its calculation, which depended on the number of successful trials achieved. This bias was removed by not including target directions with an insufficient number of trials for each subject in the group analysis. Finally, the effect of task learning on IMC was accounted for by not including a number of initial trials from each block. For all of these reasons, it can be concluded that the estimation of coherence was made so as to avoid bias in the measure and preserve consistency across subjects.

V. Conclusions

The objective of this study was to compare proximal muscle IMC, a measure of shared neural input, between force- and myoelectric-control tasks. This would help to eval-
ulate whether neural input coupling between muscles has a role in muscle coordination strategies. The main conclusions are the following:

- No significant differences in force magnitude and muscle activation level were given for single-muscle targets. Conversely, significant differences in these measures were found for joint-muscle targets, derived from the difference in target EMG level between task for these.
- No significant difference in IMC across tasks was found. Differences in task modulation across muscle pairs points towards the role of neural constraints in the distribution of shared neural input to proximal muscles and its functional role.
- IMC target modulation translated in higher IMC for single-muscle than joint-muscle targets, which agrees with the need of independent neural input in joint-muscle targets to achieve accurate muscle control. Moreover, it can be said that coupled muscle activation does not necessarily arise from shared neural input.
- IMC was highest in the alpha-band and lower part of the beta-band, relating to subcortical and cortical contributions respectively. This supports the idea that parallel neural pathways control proximal muscles.
- Regarding muscle coordination strategies, differences in IMC task modulation across muscle pairs supports that neural connectivity imposes constraints in proximal muscle coordination. This links to the idea that neural pathways lead to fixed muscle synergies for motor control.

[19] Maurice Mohr, Marius Nann, Vinzenz von Tscharner,


[48] Verity M. McClelland, Zoran Cvetkovic, and Kerry R. Mills, “Rectification of the EMG is an unnecessary and inappropriate step in the calculation of Corticomuscular


A. Appendix: Figures

FIG. 12. EMG spectra for a representative subject. Trial-averaged EMG was Fourier transformed and then normalized by the number of transform points. Rows represent target directions. Columns represent control muscles. Blue traces correspond to force-control (FC) and red traces correspond to myoelectric-control (MC). EMG spectra in MC is lower than FC for DA and DP at the single-muscle target of their complementary control muscle (target 0 deg for DA and target 180 deg for DP). For the rest of spectra, differences between tasks are minor.

FIG. 13. Force magnitude across targets for both tasks for all subjects. Each color codes for a subject. Rows represent tasks. There is higher variability in force magnitude among subjects in MC.

FIG. 14. Force magnitude CV for each subject across targets for both tasks for all subjects. Each color codes for a subject. Rows represent tasks. There is higher variability in force magnitude CV among subjects in MC.
FIG. 15. Average normalized EMG values over the hold period for each target direction for all subjects. Each color codes for a subject. Rows represent control muscles. Columns represent tasks. Dotted lines denote a normalized EMG value of 1 and 0.7 to indicate the EMG target level in myoelectric-control. There is higher variability of muscle activation in force-control across subjects. Muscle activation is tuned to the corresponding single-muscle target.

FIG. 16. Average normalized area of significant z-score of TLH-DA for all subjects. Areas are normalized by the length of the frequency band. Each color codes for a subject. Columns represent tasks and rows represent frequency bands over which the area was calculated. Single-muscle targets have higher variability across subjects for both tasks. Single-muscle target areas are higher or equal than joint-muscle target areas in most cases.

FIG. 17. Average normalized area of significant z-score of DA-BB for all subjects. Areas are normalized by the length of the frequency band. Each color codes for a subject. Columns represent tasks. Rows represent frequency bands over which the area was calculated. There is no difference in area values for joint- and single-muscle targets, and there is a similar amount of variability in these across subjects.

FIG. 18. Average normalized area of significant z-score of BB-DP for all subjects. Areas are normalized by the length of the frequency band. Each color codes for a subject. Columns represent tasks. Rows represent frequency bands over which the area was calculated. Single-muscle targets have higher variability across subjects for both tasks.
TABLE II. Number of trials included for analysis for each subject. Bold trials were not included in frequency-domain analysis.

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