The relation between force magnitude, force steadiness, and muscle co-contraction in the thumb during precision grip

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Abstract

The aim of this study was to investigate the problem of agonist–antagonist co-contraction during a precision force task performed at different force levels. Using a precision grip, seven young adults performed a constant force matching task (10, 22.5, 35, 47.5, and 60% maximum) as accurately as possible (10 trials per force level). Muscle co-contraction in the thumb was monitored by the surface EMG activity of the extensor pollicis longus (EPL) and the flexor pollicis brevis (FPB), and the ratio between those EMG activities (EPL/FPB). Results showed that both EMG activities increased as grip force increased, but the EPL/FPB ratio decreased over the range of force investigated. Force steadiness (as expressed by the coefficient of variation, CV) appeared as a U-shape function of the force level (with maximal steadiness at 22.5%). Separate analyses at each force level showed no correlation between CV and EMG indices. In addition, the contrast between trials with high and low CV revealed no significant difference in terms of our EMG indices. We conclude that muscle co-contraction and grip force steadiness depend on grip force magnitude, but grip force steadiness does not depend on muscle co-contraction.

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Force steadiness during constant force production by the fingers has been studied widely [2,7,13,24]. A recurrent observation was that force steadiness, as expressed by the coefficient of variation (CV = S.D./mean), depended strongly on the force level for both single- [2,19,20] and multi-finger tasks [23,24]. Typically, the CV appeared as a U-shaped function of the force level. For instance, Slifkin and Newell [19] reported minimal value of CV for intermediate forces ranging at about 35% of maximal voluntary contraction (MVC). The authors proposed that 30–40% MVC is optimal because force can be adjusted either by varying the number of motor units, or by modulating their discharge frequency [8]. By contrast, above or below this region, a single strategy could be adopted. Although this provides an attractive explanation, Slifkin and Newell [19] reasoned as if a single muscle was engaged in force production, and thus obliterated the redundant nature of finger musculature, in particular the essential contribution of antagonist muscles (for a review see [21]). This possibility receives credit from a study in which better force control was accompanied by reduced antagonist co-contraction [16]. However, the fact that subjects were older adults and that force production was performed by ankle dorsiflexion is a limitation for further comparison with finger studies. On the other hand, another study showed that the reduced force steadiness exhibited by old adults during finger abduction was not associated with differences in coactivation of the antagonist muscle [2]. However, this study did not investigate whether, for each group of subject, co-contraction could contribute significantly to force steadiness. To summarize, the issue of whether co-contraction has positive, negative, or no effects on force steadiness is still unclear. The goal of the present study was to investigate in healthy young adults the association between muscle co-contraction, force...
magnitude, and force steadiness during a precision grip task. We addressed this issue by monitoring the electromyographic (EMG) activity of two antagonist thumb muscles during a force-matching task at various force levels.

Seven unpaid healthy volunteers, six males and one female, took part as subjects in the experiments. All of them were right-handed according to their preferential use of the right hand during writing and eating. The age of the subjects was 25.7 ± 3.9 years. Their weight was 73.3 ± 14.5 kg, and their height 1.79 ± 0.09 m. All the subjects gave informed consent according to the procedures approved by the Mediterranean University.

During testing, the subject was seated in a chair with his/her right forearm resting on a table and hand in a neutral position. Throughout the experiment the wrist-hand configuration was secured by a cast (see Fig. 1A). Participants produced force by contacting all the fingertips on a specially designed wooden device (see Fig. 1B). A piezoelectric sensor (model 208A03, Piezotronic, Inc.) was placed under the thumb for grip force measurement. The analog output signal from the sensor was connected to separate AC/DC conditioners (M482M66, Piezotronic, Inc.). The system was operating in a DC-coupled mode, utilizing the sensor’s discharge time constant as established by the built-in microelectronic circuit within the sensor. As such, the sensor’s time constant was theoretically infinite.

Bipolar electromyographic recordings from the thumb muscles were obtained from pairs of Ag–AgCl surface electrodes. Electrodes were positioned over the flexor pollicis brevis (FPB), and the extensor pollicis longus (EPL). The ground reference electrode was placed on the subject’s styloid process of the ulna. During the acquisition, EMG signals were amplified by tunable amplifiers, band-pass filtered (30–300 Hz), prior to digitization. Both force and EMG signals were sampled at 500 Hz. Increasing the sampling rate to 1000 Hz did not lead to appreciable improvement of the EMG signals for the purpose of this study (see another example in [17]).

A large set of thumb muscles have been shown to exhibit significant activity during pinch tasks [14]. The FPB and EPL muscles were selected for the following reasons (1) they were accessible to surface EMG [6,15]; (2) their implication into precision grip was clearly demonstrated [14]; (3) their maximal contraction result in moment of force at the carpometacarpal (CMC) and metacarpophalangeal (MP) joints that have rather similar magnitude but opposite direction (see Fig. 2 in [22]). Overall, the FPB/EPL pair appeared as the most appropriate pair to investigate the issue of muscle co-contraction during a precision grip task.

Each experiment started with a series of maximal voluntary contraction. Each subject performed three trials, and the trial with the highest force produced was kept as a reference to adjust the target forces in the remaining trials. Over the group, the average MVC was 101.8 ± 13.4 N. For later off-line normalization of the EMG signals, subjects also performed maximal isometric flexion/extension contraction by the thumb; those contractions were obtained using manual resistance as described by Kendall et al. [9] Subsequently, subjects were asked to perform the force-matching task, that is to maintain constant forces as accurately as possible. During the task, they received continuous visual feedback by means of a monitor that displayed the thumb grip force. The target force consisted in a red horizontal line on the screen, and the actual thumb force consisted in a yellow line. The subject’s task was to match the yellow line to the red line (see a typical trial in Fig. 2). The duration of each trial was 10 s. We examined five levels of force (10, 22.5, 35, 47.5 and 60% MVC). For each level of force, subjects performed a block of 10 trials. The order of the blocks was randomized across subjects. Several practice trials were given before the experiment (typically five). Subjects were verbally encouraged by the experimenter, and they received feedback on their performance after each trial. The time interval between two successive trials was 30 s.

The trials were processed as follows. The first 5 s of each trial were discarded from the analysis because reaching the target, and stabilizing the force, required a few seconds (see also [23]). Based on the last 5 s, the following dependent measures were extracted. Force steadiness was assessed by the coefficient of variation of the force signal (CV = S.D./mean). Prior to this computation, force signals were low pass filtered with a second-order Butterworth filter (cut-off frequency = 30 Hz). Concerning EMG signals, we first removed any possible offset by subtracting the mean, and then rectified the signals. Additionally, EMGs were low pass filtered with a second-order Butterworth filter (cut-off frequency = 100 Hz; see [14]), and averaged over the time period. Afterwards, each EMG value was normalized with respect to its

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**Fig. 1.** Schematic representation of the grip device. (A) The forearm rested on a wooden board, and was secured by a cast screwed on the board. (B) A force sensor was mounted on a vertical piece of wood to measure grip force. The figure illustrates the relative position of the fingers.
maximal value measured during maximal flexion/extension of the thumb. Finally, we computed the relative activity between the two muscles, that is, we divided the normalized value obtained for the EPL by the normalized value obtained for FPB. This EMG ratio was taken as an index of muscle co-contraction (for other examples see [1,3]). A value close from zero means that EPL was poorly activated with respect to FPB (low co-contraction), and a value of one means that both muscles were activated to a similar degree (high co-contraction). The EMG activity of the EPL was also taken as an index of muscle co-contraction [2,16]. However, since the EPL activity was expected to increase with the level of force [14], this index was mostly devoted to comparisons that refer to similar force levels.

A first objective was to assess the effect of force magnitude on dependent variables (CV, EMGs, and EMG ratio). This was achieved using one-way repeated measure ANOVA with FORCE (five levels) as within-subject factor. For those ANOVAs, each data entry was taken as the average of 10 trials. As largely expected, ANOVA showed a main effect of FORCE for both EMGs ($F(4,24) > 17.37, P < 0.001$). They both increased as function of the force level ($R > 0.985, P < 0.01$). A main effect of FORCE was also found for the EMG ratio ($F(4,24) = 3.45, P < 0.05$). Fig. 3A shows that the EMG ratio decreased as the level of force increased ($R = −0.93, P < 0.05$). Additional Newman–Keuls post hoc tests showed that the EMG ratio was larger at 10% MVC compared to 60% MVC. Analysis of force steadiness is presented in Fig. 3B. ANOVA showed a main effect of FORCE ($F(4,24) = 10.19, P < 0.001$). This effect was due to the fact that CV appeared as a U-shape function of the force level: subjects were more accurate for intermediate forces. Newman–Keuls post hoc tests confirmed that CV at 22.5% MVC was significantly smaller than CVs at any other force levels ($P < 0.05$).

A second objective was to determine whether, within each level of force, there could be any specific relationship between force steadiness and muscle co-contraction. To achieve this goal we took advantage of intra-subject variability: within each block of 10 trials, some trials had larger CV than others. A first analysis consisted in computing, for each subject, and at each force level, the coefficient of correlation between CV and co-contraction indices over the 10 trials.
Results showed that, over the 35 coefficients of correlation computed between CV and EMG ratio, only one was significant. After Z-score transformation, the average R-value over the group was 0.08 ± 0.33. Very similar results were obtained for EMG values of EPL (only two coefficients of correlation were significant; average R-value = 0.05 ± 0.34). A second analysis consisted in examining whether co-contraction levels could be different in trials with high and low CV. For each subject, and at each force level, we only kept the trial with the lowest CV and the trial with the highest CV. Two-way repeated-measure ANOVAs with FORCE and PERF (high CV versus low CV) were conducted for each dependant variable. Overall, the comparison between these two sets of trials revealed that fluctuations of force were about 75% larger in trials with high CV (CV$_{high}$ = 1.05%, and CV$_{low}$ = 1.83%, F(1,6) = 11.5, P < 0.001). However, further comparisons between these two sets showed no significant differences in terms of co-contraction indices. Two-way ANOVA on EMG ratios showed no main effect of PERF (F(1,6) = 2.54, P > 0.05). In agreement with earlier analyses, we found a main effect of FORCE (F(4,24) = 2.77, P < 0.05), but there was no significant interaction between PERF and FORCE (F(4,24) = 0.91, P > 0.05). Similar conclusions were obtained when this procedure was repeated for the EMG values of EPL.

The goal of this experiment was to investigate the relationships between muscle co-contraction, force steadiness, and force magnitude. At this stage we showed (1) that force steadiness and muscle co-contraction varied as a function of the force level, but in different ways and (2) for a given force level, force steadiness did not depend on muscle co-contraction. Let us now discuss in more details those results and their implications.

The observation that intermediate forces were better stabilized than high and low forces is consistent with earlier studies [2,4,19,20]. However, we would like to emphasize that a large number of fingers seems beneficial to force steadiness. Indeed, a quick review of the literature indicates that CVs ranged between 2.3 and 5.2% for a single-finger task [19], between 1.2 and 5.5% for a thumb-index precision grip task [24], and between 1.1 and 1.6% when all five fingers were engaged in the task (this study). Those observations fit well the opinion of Latash et al. [12], that is “an increase in the number of fingers should be viewed not as an additional load for the central nervous system, but as a helpful modification allowing more flexibility for error compensation”.

The observation that EMG activities of the FPB and EPL increased as a function of precision grip force is consistent with the study of Maier and Hepp-Robinson [14]. Following their terminology, the positive correlation found between the FPB and EPL activity was classified as a muscle synergy of the coactivation type. Although the present study supports this view, it also points out that co-contraction (as reflected by the EMG ratio) is not set to a constant level, but instead decreases as a function of the force level. Other data collected during index finger abduction demonstrates a similar trend (see the ratio between the EMG activities of the first dorsal interosseous and the second palmar interosseus in [2]). In other words, as force increases, muscle activation becomes more and more specific of the agonist muscle. Could this change in muscle activation account for changes in force steadiness? The answer is likely to be no. Indeed, we have collected no direct evidence that muscle co-activation could influence grip force steadiness. First, across force levels, very different functions were obtained for the EMG ratio–force relationship (Fig. 3A), and for the CV–force relationship (Fig. 3B). Second, at each force level, we found no significant correlation between EMG parameters and force steadiness. And third, the opposition between the trials with high and low CV was unable to demonstrate any significant differences in terms of EMG parameters. Based on those observations, we conclude that muscle co-contraction is not a critical factor for force steadiness during a precision grip task.

Before addressing in further details the implications of a possible independence between muscle co-contraction and force steadiness, we felt reasonable to briefly expose the limitations inherent to our methodology. First, we have monitored the activity of only one flexor and one extensor muscle. It is conceivable that, given the redundant musculature of the thumb (nine muscles, see [22]), finer inter-muscle coordination may contribute to force steadiness. Second, when examining variability, there is a limitation of using measures such as means, CVs and SDs, because they ignore the temporal and frequency structure of time series, and thus may neglect potentially interesting contributions to motor variability. Third, EMG activities are highly variable as compared to grip force steadiness. This makes our study prone to type II error since a subtle but existing relationship could not be detected. However, it is not obvious to see other methodological approaches except the direct measurement of muscles forces (although much more invasive).

Silfkin and Newell [20] proposed that 30–40% MVC is an optimal range for finger force production because muscle force can be adjusted either by varying the number of motor units, or by modulating their discharge frequency [8]. In the present study, we hypothesized that changes in muscle coordination could provide additional flexibility to adjust force steadiness. What are the implications of our results with respect to this hypothesis? On the one hand, our results show that it is a crude simplification to override the flexibility of neuromuscular synergies, since obvious changes in muscle coordination were demonstrated. On the other hand, we have been unable to demonstrate that these changes could account for changes in performance. In other words, the flexibility offered by agonist–antagonist synergies would be of little benefit to force steadiness during a force-matching task. Still, it remains intriguing that all experiments conducted with precision grip tasks found optimal force range below 30–40% MVC (22.5% in this study, 25% in [24]; 15% in [4]). We propose that, because fingertip forces are typically small when manipulating objects (about 2N according to [5]), grip force steadiness at low force level is encouraged by everyday situations. By contrast, other less frequent manual tasks, like
single-finger task (cf. [19]), would benefit less from everyday experience.

Given the redundant musculature of the fingers, muscle coordination during precision grip is a priori a very complex task. Based on our surface EMG analysis, it seems that the central nervous system relies on a more complex strategy to modulate grip force magnitude than grip force steadiness. Indeed, although modulations in grip force magnitude were accompanied by selective changes in the gain of the agonist and antagonist activity, modulations in grip force steadiness were achieved without any changes in those muscle gains. This view contrasts with a recent study in which notable changes in brain activity were induced by dexterous scaling of fingertip forces [11]. However, we certainly do not exclude that other EMG analyses, such as motor unit synchronization, or EMG–EEG synchronization, could better portray the challenge faced by the central nervous system during meticulous force production tasks [10,18].

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