



# Effects of spastic cerebral palsy on multi-finger coordination during isometric force production tasks

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## Abstract

In this study, we quantified changes in finger interdependence (enslaving), multi-finger synergies, and feedforward modulation of synergy properties (i.e., anticipatory synergy adjustment) during single- and multi-finger force production tasks in individuals with cerebral palsy (CP). Spastic diplegic CP and healthy control subjects performed sets of finger force production tasks by each of the hands, including maximal force production and submaximal quick pulse force production in an isometric condition. The framework of the uncontrolled manifold hypothesis was used to quantify the indices of multi-finger synergies and the anticipatory synergy adjustment (ASA). The CP group showed lower maximal forces and higher indices of finger interdependence (enslaving), while the indices of multi-finger synergies stabilizing total finger forces during stable force production were not different significantly compared to the controls. Further, the time of ASA for the CP group was not delayed. The CP group showed a significantly less drop in the synergy indices during the anticipatory and quick pulse phase compared to the control group, which was accompanied by larger co-contraction indices of the forearm muscles. These findings suggest that the function of assembling motor synergies for stable force production is not affected by CP, while the ability to modulate synergy properties may be impaired with CP partially due to spasticity. The spasticity presumably hampers the purposeful feedforward destabilization of the performance. The results suggest that quantification of multi-digit synergies may provide an alternative tool for quantitative assessment of impaired coordination in the CP individuals.

**Keywords** Cerebral palsy · Multi-finger synergy · Anticipatory synergy adjustment · Stability

## Introduction

Cerebral palsy (CP) is a permanent and non-progressive disorder which occurs due to damage to the immature brain of a fetus or infant. It has been known that antenatal processes and abnormal parturition resulting in incomplete

development of the brain are the primary cause of cerebral palsy, but the underlying mechanism for the broad spectrum of abnormalities caused by CP is not clearly known. However, clinical observations and research provide ample evidence that the dysfunction observed in CP is associated with physical weakness, reduced the passive range of motion, delayed motor and intellectual development (Graham and Selber 2003; Odding et al. 2006), and considerable variation of fiber size of wrist flexors as compared to wrist extensors (Ponten et al. 2005). CP is classified into several types depending on the severity and the range of malfunctioning body segments (Paneth 2008), and the most common type is spastic CP. In particular, the weakened and stiffened contractile properties of skeletal muscle (von Walden et al. 2017) and uncontrolled reflex muscle contraction, the so-called “*spasticity*”, are common clinical features in spastic CP (Odding et al. 2006). Besides, spastic CP is characterized by stiffened networks of extracellular macromolecules (i.e., extracellular matrix) and an increased sarcomere length in muscles (Smith et al. 2011). Therefore, the behavior patterns

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Joo Kong and Kitae Kim have equal contribution to this study; thus, they are co-first authors.

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of CP are assumed to be caused by various factors, including impairment of the nervous system, the abnormal formation of peripheral properties, and their interaction. For example, the spastic gait in CP individuals could be viewed as a decrease in the flexible alternation of the synchrony and asynchrony of the motion of the joints partially caused by coupled actions (i.e., less individuated actions) of the joints along with a less fluid pattern of muscle co-contraction (Crenna 1998; Tedroff et al. 2008). The coupled actions resulting in positive co-variation of elements are known to be attributable to peripheral and central reasons (Li et al. 1998, 2002; Schieber 1990; Schieber and Hibbard 1993). These phenomena are in line with the term, *abnormal synergies* used in clinics, which favors the stereotypical patterns of muscle activation shown in patients with neurological disorders (Neckel et al. 2006). However, the quantitative relationship between the peripheral properties and control strategies in patients with CP is rarely investigated. The majority of the attention has been given to the investigation of abnormal walking with CP (Lee et al. 2018). It has been reported that more than half of children with CP suffer from upper extremity dysfunction (Boyd et al. 2001; Brown et al. 1987). The investigation on the upper extremities in CP individuals has relied primarily on the incidence of hand preference (i.e., handedness) (Bansal et al. 2016), differences in physiological properties of pairs of muscles (i.e., flexors and extensors) (Ponten et al. 2005; Von Walden et al. 2017), and etc., while there is limited research investigating the changes in neural strategies of manual hand actions associated with the handedness and the functional activities of muscles (e.g., voluntary changes in movement direction) in CP individuals. We believe that the neural process to govern the aforementioned features (e.g., handedness, properties of muscles, etc.) in CP individuals may differ from those in the healthy controls.

The neural coordination involving multiple fingers has been objectively quantified and is associated (Kim et al. 2018; Park and Xu 2017) with the concept of motor abundance (Gelfand and Latash 1998). Based on the concept of motor abundance, successful motor performance in a specific environment could be achieved by the flexible organization of a redundant set of elements considering proper activation of internal elements such as muscles, motor units, etc. The number of actively involved elements for a particular movement is usually larger than the minimum number decided by the mechanics. Therefore, the design purpose of the redundancy in human body configuration may be to yield a prerequisite condition for the formation of flexible combinations of elements. In particular, the redundancy would be advantageous for stabilizing salient performance variables such as the variability of the overall performance and minimizing the net force or torque (Kim et al. 2018). The neural strategies related to the flexible organization of redundant

elements considering the required mechanics of the motor tasks have been termed as “*synergies*” (Latash et al. 2007). The organization of the solution families yields to the stabilization of performance, which is presumably the reflection of the function of the neuronal structure. In this regard, stability in the current context could refer to the function of the neural process that ensures the synergic actions of multi-elements by showing task-specific co-variation among the involved elements. On the contrary, *abnormal synergies* partially refer to less flexible patterns of actions of elements (e.g., stereotyped muscle activation) that hampers intentional movement.

Another important neural function for successful voluntary actions in humans involves the modulation of stability, which could be associated with the concept of agility. A general definition of agility is the ability of the body to change its mechanics in effective ways (Latash and Huang 2015; Palermo and Suppa 2018). Then, what would be effective ways to alter its mechanics? One possible and smart way is to turn off synergy formation since stability is no longer required and should be weakened (i.e., destabilized) in case of voluntary changes of the mechanical outcomes. The function of purposeful destabilization has been extensively studied and showed that the synergy strength is gradually reduced in a feedforward manner if one has information about the upcoming actions. The feedforward manner means that the stability properties start to change (actually, reduce) in advance of visible changes in salient performance variables. This phenomenon has been termed anticipatory synergy adjustment (ASA). The time of ASA ( $t_{ASA}$ ) observed in young adults is about 200–300 ms before the initiation of visible actions, and it was reported that the mechanism of ASA was independent of the muscle strength and movement direction of upcoming action (i.e., not muscle-specific mechanism) in healthy adults. On the contrary,  $t_{ASA}$  is delayed in the elderly (Olafsdottir et al. 2007) and various populations having neurological disorders, including Parkinson’s disease (Park et al. 2012), multi-element atrophy (Park et al. 2013), and in stroke survivors (Jo et al. 2016).

In this study, a set of multi-finger isometric force production tasks was employed to quantify the force production capability as well as the index of multi-finger synergy and its feedforward modulation in individuals with spastic cerebral palsy (CP) wherein the functions of the upper extremities including the hands and fingers remained intact. The experimental setup and procedures were similar to those of several earlier studies of Parkinson’s disease (Park et al. 2012) and in stroke survivors (Jo et al. 2016), while the individual finger forces during both flexion and extension effort and forearm muscular activities were additionally measured in the current experiment.

Based on previous knowledge and experimental outcomes with CP, we formulated the following hypotheses: (1) The

force production capability in CP individuals is smaller than that in age- and gender-matched controls. (2) The individual finger action (i.e., force production) in CP individuals is less independent compared to the controls. (3) The synergy index for CP individuals is significantly different from that of the controls, while we predict that the anticipatory synergy adjustment is altered in CP.

## Methods

### Subjects

Nine patients with CP who were diagnosed as spastic diplegia by a pediatric orthopedic specialist (age  $18.6 \pm 3.6$  years, three females, right-handers, GMFCS level II) were enrolled in the study. The symptoms of diplegic CP in the recruited subjects affected mostly symmetrical parts of their legs. In addition, age- and gender-matched ten control subjects with no history of neurological and peripheral problems and sensory deficit (age  $21.5 \pm 2.3$  years, three females, right-handers) were recruited. Prior to the experiment, we interviewed all the subjects and their parents to check if they were able to understand the experimental instruction and perform the current motor tasks using fingers. Additional inclusion criteria for CP subjects comprised (1) no history of severe symptoms and surgery on upper extremities including hands and fingers and (2) no sensory deficit in both hands and fingers confirmed by the monofilament test (monofilament

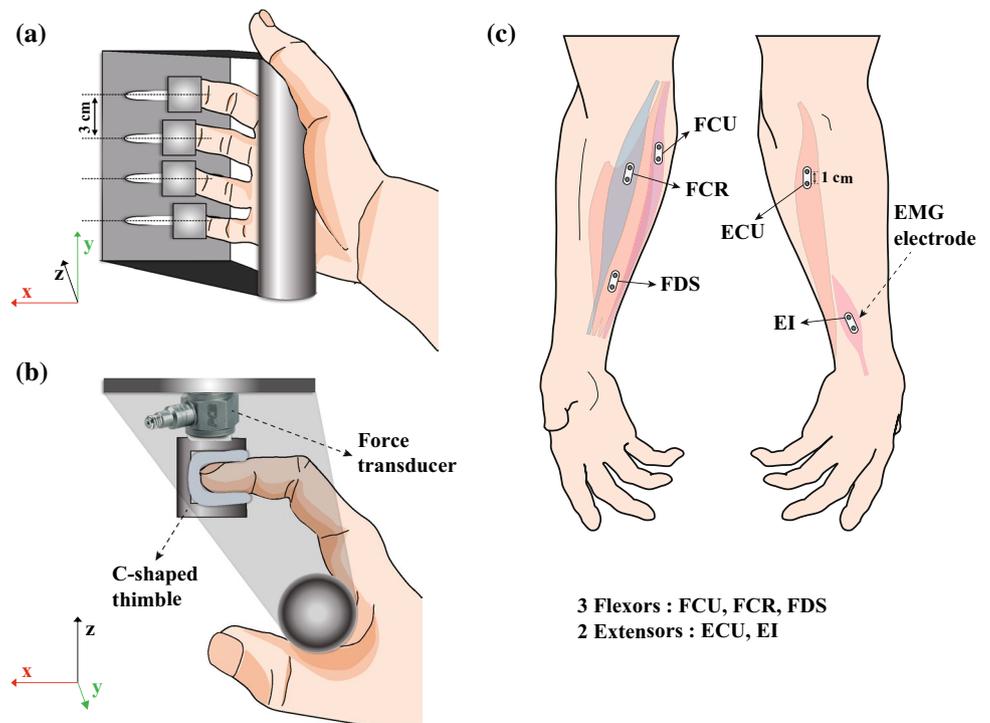
Test, North Coast Medical Inc. USA). All the participants signed a consent form, and parental consents were obtained for the participants under 18 years of age. The consent form indicated the procedures and potential risks involved during the experiment in accordance with the ethical standards set by the Institutional Review Board of Seoul National University Bundang hospital (IRB No. B-1707-408-302).

### Apparatus

#### Finger force measurement

Four unidirectional force transducers (Model 208A03, PCB Piezotronics Inc., Depew, NY, USA) with amplifiers were used to measure individual finger forces during isometric finger flexion and extension efforts. The transducers were attached to the customized aluminum panel (Fig. 1a). The C-shaped thimbles were affixed to the individual transducers (Fig. 1b) where the distal phalanges of the fingers were supposed to be inserted. The vertical distance (y-axis) between the transducers was fixed at 3 cm, while the position along the mediolateral direction (x-axis) and the angle along the anteroposterior axis (z-axis) of the transducers could be adjusted according to the hand anatomy of individual subjects. The panel with the transducers was aligned vertically and mechanically fixed to the immovable table so that the direction of isometric finger forces (e.g., flexion and extension forces) was perpendicular to the direction of the gravity. The sampling rate of the force signals was set at 500 Hz

**Fig. 1** The illustrations of the experimental equipment and condition. **a** The experimental frame was mechanically fixed to the table and the sensors were attached to the customized experimental frame (size:  $90 \times 140 \times 250$  mm). The distance between adjacent transducers was set at 3 cm along the y-axis. A cylinder was placed underneath the palm to ensure a consistent configuration of the hand and fingers during the tasks. **b** The four fingers were naturally flexed at the proximal interphalangeal joint for about  $10^\circ$ – $20^\circ$  and inserted in corresponding C-shape thimbles. **c** The configuration of EMG sensors. A total of five EMG sensors was attached to five muscles on the forearm including three finger flexors (FDS, FCU, FCR) and two finger extensors (EI, ECU)



using a customized LabVIEW program (National Instrument, Austin, TX, USA).

### Electromyography (EMG) measurement

The surface electromyography (EMG) system (Trigno IM, Delsys Inc., Boston, MA, USA) was used to measure the activities of the forearm muscles during finger force production. The surface EMG (sEMG) sensors were attached to five muscles (Fig. 1c) on the forearm including three flexors (flexor digitorum superficialis, FDS; flexor carpi ulnaris, FCU; flexor carpi radialis, FCR) and two extensors (extensor indicis, EI; extensor carpi ulnaris, ECU) to measure the muscle activity involved in the finger flexion and extension during the task. The sampling rate of the EMG data was set at 2000 Hz.

### Procedures

Prior to the main experiment, the participants performed the Grooved Pegboard test (Model 32025, Lafayette Instr., Hawthorn Woods, IL, USA) for the purpose of a clinical test as measures of manual hand dexterity (Ruff and Parker 1993). For the Grooved Pegboard test, the time in sec to place 25 pegs in the holes and the number of dropped pegs were measured for the non-dominant and dominant hand, separately. The main experiment consisted of (1) maximal voluntary contraction (MVC) task, (2) single-finger ramp force production task, and (3) multi-finger quick pulse force production tasks (Park and Xu 2017). All participants of both groups were instructed to maintain a fixed arm posture: approximately 45° flexion of shoulder and elbow, 45° abduction and 10° internal rotation of the shoulder, and the wrist was placed neutrally between abduction and adduction while sitting in a height-adjustable chair for all experimental tasks. Real-time force feedback was provided from the monitor (27-inches 1920 × 1080 resolution at 60 Hz) mounted in front of 50 cm at the participant's eye level during the experiments. Two CP subjects were able to complete a part of experimental conditions; thus, the data of these two subjects were excluded for the analysis.

### Maximal voluntary contraction (MVC) task

The participants were asked to produce isometric maximal voluntary contraction (MVC) forces with all four fingers. The MVCs of the dominant and non-dominant hand for the flexion and extension were measured separately. The real-time feedback of the total force ( $F_{TOT}$ ) was provided during the task. The participants performed two trials for each of two directions (flexion and extension) and two hands (dominant and non-dominant). For a single trial, 8-s was given to reach the maximum isometric forces. After each MVC

trial, the participants had a minimum of 3 min' rest, and the additional rest was provided upon the request to avoid the effect of fatigue. The surface EMGs of the five muscles (e.g., finger flexors and extensors) were also collected during the trials. The data from two trials with higher  $F_{TOT}$  value were selected as  $MVC_{TOT,k,l}$  ( $k = \{\text{flexion, extension}\}$ , and  $l = \{\text{dominant, non-dominant}\}$ ). The individual finger forces ( $MVC_{i,k,l}$ ;  $i = \{\text{index, middle, ring, little}\}$ ,  $k = \{\text{flexion, extension}\}$ ,  $l = \{\text{dominant, non-dominant}\}$ ). The  $MVC_{i,k,l}$  (MVC of individual fingers) and  $MVC_{TOT,k,l}$  (MVC of all four fingers) values were also used to determine the target force values for the next two tasks.

### Single-finger ramp force production task

The participant performed a set of single-finger ramp force production tasks that required to make either extension or flexion effort with a single task-finger of one of the hands while matching with its force template shown in the computer screen as accurately as possible. The template on the computer screen consisted of three phases including a 4 s horizontal line at zero force, 12-s slanted line starting from 0 to 40% of  $MVC_{i,k,l}$  ( $i = \{\text{index, middle, ring, little}\}$ ,  $k = \{\text{flexion, extension}\}$ , and  $l = \{\text{dominant, non-dominant}\}$ ), and the last 4 s horizontal line at 40% of  $MVC_{i,k,l}$ . The important instruction to the subjects was that “keep all the fingers on the corresponding sensors and do not pay attention to unintended force production by non-task fingers”. We collected both task and non-task finger forces, albeit the feedback screen showed only the force values of the single task-finger.

### Multi-finger quick pulse force production task

In this task, the participants were instructed to produce a multi-finger steady-state force followed by a quick pulse force while changing force direction and magnitude simultaneously. There were two sub-conditions that included (1) flexion–extension (FE) and (2) extension–flexion (EF). The first letter (i.e., F or E) indicates the direction of finger force at steady-state, while the second letter represents the direction of finger force for the quick pulse. The flexion and extension forces were assigned to negative and positive, respectively, in the y-axis of the feedback screen. A single trial lasted 10 s, which composed of the steady-state and quick pulse phases. The participants were instructed to produce the steady-state flexion (FE condition) or extension (EF condition) forces for at least 3–4 s within the first 5 s and to produce a quick pulse force to the target value in a self-paced manner within the next 5 s. Note the quick pulse force production was initiated by a self-selected time (i.e., anticipated manner), not by external cue. The magnitudes of the steady-state force was set at 5% of  $(|MVC_{TOT,flexion,k}| + |MVC_{TOT,extension,k}|)/2$ , where  $k = \{\text{dominant, non-dominant}\}$ .

The change in force magnitude from the steady-state to the quick pulse was set at 20% of  $(IMVC_{TOT,flexion,k}^l + IMVC_{TOT,extension,k}^l)/2$ . We purposefully designed symmetrical task space in the constrained flexion and extension force magnitudes at the steady-state and their magnitude changes for both the FE and EF conditions. About 20 min practice time was provided before the experiment and the participants performed 25 trials for each condition and each hand (25 trials  $\times$  2 directions  $\times$  2 hands = 100 trials in total). After the completion of each trial, we had a visual inspection of time series  $F_{TOT}$  to detect erratic trials. If the time profiles of  $F_{TOT}$  for particular trials showed major mistakes such as multiple peaks or serious fluctuation of  $F_{TOT}$  (i.e., exceeding error margin set at  $\pm 5\%$  of the target  $F_{TOT}$  value), the trial was excluded, and additional trials were given. More than 10 s break between every two trials was provided and no fatigues were reported for all participants.

**Data analysis**

Customized program codes (MATLAB, MathWorks, Natick, MA, USA) were written to process the measured force and EMG data. The raw force signals (i.e., forces data of individual fingers) were digitally low-pass filtered at 10 Hz cutoff with zero-lag, 4th-order Butterworth filter. The processing of the raw EMG data for data analysis was as follows. First, a notch filter at 60 Hz was applied to eliminate a possible noise generated by the external electronic supply. Then, the EMG signals were rectified and low-pass filtered at 50 Hz (2nd-order, zero-lag Butterworth filter). The rectified EMG data were integrated ( $iEMG$ ) over time of the steady-state force production phase ( $-900$  to  $-400$  ms with respect to  $t_0$ ). For the comparisons of the  $iEMG$  across muscles and participants,  $iEMG$  values were normalized by  $iEMG_{MVC}$  values of corresponding conditions (e.g., two hands and directions). The  $iEMG_{MVC}$  of a particular condition was computed in such a way that the EMG value of 400 ms (the same time interval as the steady-state period) around the time of reaching maximal  $F_{TOT}$  ( $MVC_{TOT,k,l}$ ) were integrated.

**Single-finger ramp force production task**

Enslaving matrix ( $E$ ) and enslaving index ( $EN$ ) were computed from the data acquired during the single-finger ramp force production tasks. For each participant, a set of four  $E$ s and  $EN$ s was computed including the  $E$  and  $EN$  of the combinations of two directions (e.g., flexion and extension) and two hands (e.g., dominant and non-dominant hand). For each combination of directions ( $k = \{flexion, extension\}$ ) and hands ( $l = \{dominant, non-dominant\}$ ), the elements in  $E_{k,l}$  represents the magnitudes of unintended forces by non-task fingers ( $j$ ) with respect to the total force ( $F_{TOT}$ ) when the task finger ( $i$ ) produced the constrained forces (i.e.,

ramp force template) voluntarily. Linear regression analyses were used to extract the regression coefficients ( $e_{i,j,k,l}$  in Eq. 1) between individual finger force ( $F_i$ ) and  $F_{TOT}$ . Then, all these coefficients configured the matrix of  $E_{k,l}$  (Eq. 2). The enslaving indices ( $EN_{k,l}$ ) was calculated by averaging all non-diagonal elements in the  $E_{k,l}$

$$F_{i,j,k,l} = f_{i,k,l}^0 + e_{i,j,k,l} F_{TOT,i,k,l} \tag{1}$$

$$E_{k,l} = \begin{bmatrix} e_{I,I,k,l} & e_{I,M,k,l} & e_{I,R,k,l} & e_{I,L,k,l} \\ e_{M,I,k,l} & e_{M,M,k,l} & e_{M,R,k,l} & e_{M,L,k,l} \\ e_{R,I,k,l} & e_{R,M,k,l} & e_{R,R,k,l} & e_{R,L,k,l} \\ e_{L,I,k,l} & e_{L,M,k,l} & e_{L,R,k,l} & e_{L,L,k,l} \end{bmatrix} \tag{2}$$

**Multi-finger quick pulse force production task**

A set of the following variables for each participant was computed for the combinations of the two directions ( $k = \{FE, EF\}$ ) and two hands ( $l = \{dominant, non-dominant\}$ ), separately.

Timing indices of time-series force data: we computed the time initiation of quick pulse force ( $t_0$ ) using time profile of  $F_{TOT}$  for individual trials, which was defined as the time when the first derivative of  $F_{TOT}$  ( $dF_{TOT}/dt$ ) reached 5% of its positive (FE condition) or negative peak (EF condition) values after the steady-state force production. The time of force peak ( $t_{PEAK}$ ) was defined as the time from  $t_0$  to the time moment at which the maximum (FE condition) or minimum (EF condition)  $F_{TOT}$  was detected. The average and standard deviation (SD) of  $t_{PEAK}$  across repetitive trials for each experimental condition were computed, separately. For the repetitive trials, the data were aligned with respect to  $t_0$  for the following variance analysis. Since  $t_{PEAK}$  was slightly different across repetitive trials, we resampled to 150 data points using cubic spline interpolation for the data from  $t_0$  to  $t_{PEAK}$ .

Two components of variances and force-stabilizing synergy index: the framework of the uncontrolled manifold analysis (Scholz and Schöner 1999; Scholz et al. 2000) were used to compute the two-component of variances,  $V_{UCM}(t)$  and  $V_{ORT}(t)$ , and the indices of force stabilizing synergy in a time-series,  $\Delta V(t)$ . This computation was performed for each subject under the four experimental conditions (e.g., EF and EF condition with dominant and non-dominant hand), separately. The elemental variables in the current analysis were the vector of finger mode ( $\mathbf{m}$ ) in the mode space, which refers to the independent hypothetical commands to four fingers. The vector of four finger forces is assumed to be a product of the mode vector ( $\mathbf{m}$ ) and enslaving matrix ( $E$ ) of corresponding condition (Eq. 3)

$$\mathbf{m}_{k,l} = [E_{k,l}]^{-1} \cdot F_{k,l}; \quad F = [f_I, f_M, f_R, f_L]^T, \tag{3}$$

where  $\mathbf{E}$  is enslaving matrix,  $\mathbf{F}$  and  $\mathbf{m}$  are force and mode vector, respectively.  $f_i$  refers to individual finger force. Again,  $k = \{\text{FE}, \text{EF}\}$  or  $\{\text{flexion}, \text{extension}\}$  and  $l = \{\text{dominant}, \text{non-dominant}\}$ . Note that a single enslaving matrix (e.g.,  $\mathbf{E}_{\text{flexion},l}$  or  $\mathbf{E}_{\text{extension},l}$ ) that corresponded to the direction of steady-state force was used to compute the mode vectors for each subject and hand. For example,  $\mathbf{E}_{\text{flexion},l}$  was employed to compute  $\mathbf{m}_{\text{FE},l}$ , and  $\mathbf{E}_{\text{extension},l}$  for  $\mathbf{m}_{\text{EF},l}$  computation.

First, a set of time-aligned data set was used to quantify the variances of two linear sub-spaces within a 4-dimensional mode ( $\mathbf{m}$ ) space across trials for each time sample. The first sub-space, uncontrolled manifold (UCM) space, was obtained by calculating a null space of Jacobian ( $\mathbf{J}$  in Eq. 4) in which the change of the element variables (dEV in Eq. 4) corresponds to no changes in  $F_{\text{TOT}}$  as a performance variable (i.e., dPV = 0). On the contrary, the ORT space refers to the space orthogonal to the UCM space in which the change of the element variables refers to the variability of performance variable across trials. The second sub-space was the orthogonal complement (ORT) to the UCM where the variance within ORT changed  $F_{\text{TOT}}$  (i.e., dPV  $\neq$  0)

$$dPV = \mathbf{J} \cdot dEV. \quad (4)$$

The multi-finger synergy indices ( $\Delta V$ s) for the stabilization of  $F_{\text{TOT}}$  in time-series were quantified as the relative amount of variances across trials within two sub-spaces, UCM space ( $V_{\text{UCM}}$ ) and the orthogonal (ORT) space ( $V_{\text{ORT}}$ ) shown in Eq. (5). Note that the sum of  $V_{\text{UCM}}$  and  $V_{\text{ORT}}$  was equal to the total variance ( $V_{\text{TOT}}$ ). Further, the variances were normalized by the degrees of freedom (DOFs) of the corresponding sub-spaces (e.g.,  $\text{DOF}_{\text{UCM}}$ ,  $\text{DOF}_{\text{ORT}}$ ,  $\text{DOF}_{\text{TOT}}$  in Eq. 5) (Arpinar-Avsar et al. 2013; Friedman et al. 2009; Latash et al. 2001)

$$\Delta V_{k,l}(t) = \frac{V_{\text{UCM},k,l}(t)/\text{DOF}_{\text{UCM}} - V_{\text{ORT},k,l}(t)/\text{DOF}_{\text{ORT}}}{V_{\text{TOT},k,l}(t)/\text{DOF}_{\text{TOT}}}. \quad (5)$$

The values in  $\Delta V$  were limited to the computational boundaries. Thus, the log transform (Fisher's z-transformation) was applied to avoid the ceiling effect (Kim et al. 2018). The steady-state period of  $\Delta V(t)$  was set between  $-900$  and  $-400$  ms before  $t_0$  to have 0.5 s of the steady-state time interval prior to a possible anticipatory change in  $\Delta V(t)$  (Park and Xu 2017), and the average values of the log-transformed  $\Delta V$  (i.e.,  $\Delta V_{\text{SS}}$ ),  $V_{\text{UCM}}$ , and  $V_{\text{ORT}}$  were computed for the statistical analysis.

Changes in synergy properties: the anticipatory synergy adjustment (ASA) is defined as the phenomenon that the  $\Delta V$  value is reduced prior to  $t_0$  (i.e., before the predictable perturbation without an overt change of performance variables). We computed the changes in a set of two synergy properties during the anticipatory synergy adjustment including the time initiation and changes in the magnitude of  $\Delta V$  that

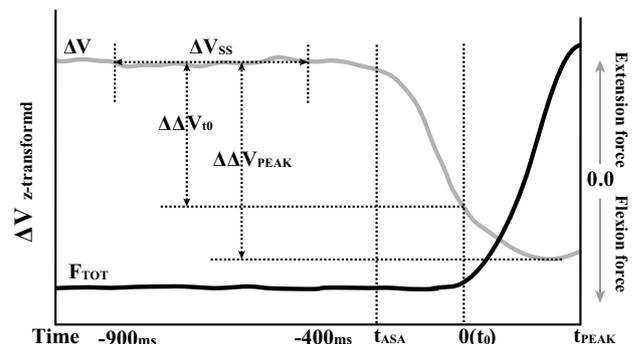
were represented by  $t_{\text{ASA}}$  and  $\Delta\Delta V_{t_0}$ , respectively (Olafsdottir et al. 2005; Park and Xu 2017). The  $t_{\text{ASA}}$  was defined as the time moment when the  $\Delta V(t)$  started to drop by more than 2 standard deviations of average  $\Delta V$  over the steady-state period (i.e.,  $\Delta V_{\text{SS}}$ ). The  $\Delta\Delta V_{t_0}$  was defined as the difference in  $\Delta V$  magnitude between the  $\Delta V_{\text{SS}}$  and  $\Delta V(t)$  at  $t_0$ . Further, the changes in  $\Delta V$  magnitude between the  $\Delta V_{\text{SS}}$  and the negative peak value of  $\Delta V$  after  $t_0$  ( $\Delta\Delta V_{\text{PEAK}}$ ) was also quantified (Fig. 2).

Co-contraction index (CCI): The Co-contraction index (CCI) indicates the relative muscle activity of the antagonists ( $i\text{EMG}_{\text{ANT}}$ ) to the overall muscle activities ( $i\text{EMG}_{\text{TOT}}$ ). The CCIs during the steady-state force production ( $\text{CCI}_{\text{SS}}$ ) and anticipatory phase ( $\text{CCI}_{\text{ASA}}$ ) were computed separately using Eq. (6) (Kellis et al. 2003). Since the required force direction at the steady-state was either flexion or extension, thereby, a group of antagonist muscles depended on the experimental condition. For example, the  $i\text{EMG}_{\text{ANT}}$  was computed as the sum of  $i\text{EMG}$  of the extensor muscles (e.g., EI and ECU) for the FE condition, while the  $i\text{EMG}$ s of the flexor muscles (e.g., FDS, FCU, and FCR) were supposed to be  $i\text{EMG}_{\text{ANT}}$  for the EF condition. The  $i\text{EMG}_{\text{TOT}}$  was calculated as the sum of  $i\text{EMG}$  of overall muscles

$$\text{CCI}_{k,l} = 2 \frac{i\text{EMG}_{\text{ANT},k,l}}{i\text{EMG}_{\text{TOT},k,l}}. \quad (6)$$

## Statistics

Standard descriptive statistics and mixed-design repeated measures (RM) of ANOVAs were performed to explore



**Fig. 2** The representative data of total force (black line,  $F_{\text{TOT}}$ ) and synergy index (grey line,  $\Delta V$ ) during quick pulse force production tasks. Flexion and extension forces are presented as negative and positive along the y-axis, respectively.  $t_{\text{ASA}}$ ,  $t_0$ , and  $t_{\text{PEAK}}$  represent the time of anticipatory synergy adjustment (ASA), the time of initiation of total force ( $F_{\text{TOT}}$ ) change, and time of  $F_{\text{TOT}}$  peak, respectively.  $\Delta V_{\text{SS}}$  represents the mean value of  $\Delta V(t)$  during the steady-state force production (i.e.,  $-900$  to  $-400$  ms with respect to  $t_0$ ).  $\Delta\Delta V_{t_0}$  and  $\Delta\Delta V_{\text{PEAK}}$  are the magnitude changes in  $\Delta V$  from the steady-state to  $t_0$  ( $\Delta\Delta V_{t_0}$ ) and  $t_{\text{PEAK}}$  ( $\Delta\Delta V_{\text{PEAK}}$ ), respectively

how the main outcome variables were affected by the factors including *Group* (two levels: CP and control), *Hand-Dominance* (two levels: dominant and non-dominant), and *Direction* (two levels: flexion and extension or EF and FE). The outcome variables included the measures of the Pegboard test, MVC finger force, enslaving index (EN), average  $t_{PEAK}$ , SD of  $t_{PEAK}$ ,  $t_{ASA}$ ,  $V_{UCM}$ ,  $V_{ORT}$ ,  $\Delta V_{SS}$ ,  $\Delta \Delta V_{i0}$ ,  $\Delta \Delta V_{PEAK}$ ,  $CCI_{SS}$ , and  $CCI_{ASA}$ . The factors were selected for particular statistical tests, and the maximum number of the selected factors was limited (i.e., two-way RM ANOVAs) due to small sample size in the current study. The assumption of sphericity was tested by Mauchly’s sphericity test, and the Greenhouse–Geisser corrections were used in case of rejecting the sphericity assumption. For the repetitive measures of the force and EMG, the intra-class correlation coefficients (ICC) as an index of test–retest reliability were estimated for the CP and controls, separately (Table 1). Conservative non-parametric Mann–Whitney tests with Bonferroni  $p$  value correction were conducted as post hoc pairwise comparisons. The statistical power for all comparisons was computed. It was confirmed that the powers of most of the comparisons were larger than 0.7, and the results of post calculation of the effect size (partial eta-squared,  $\eta_p^2$ ) for the comparisons were presented. The level of significance for all statistical tests was set at  $p < 0.05$ .

## Results

### Pegboard test

We quantified two indices from the Grooved Pegboard test including the time (i.e., the time to place 25 pegs in the holes) and the number of dropped pegs. On average, the time for the CP group ( $205.9 \pm 31.9$  s, mean  $\pm$  SD) was

**Table 1** ICC of finger forces and iEMG values

	Total finger force		Average value of each muscle EMG	
	CP	Controls	CP	Controls
<b>Dominant</b>				
FE	0.968*	0.998*	0.940*	0.987*
EF	0.969*	0.999*	0.942*	0.977*
<b>Non-dominant</b>				
FE	0.995*	0.998*	0.972*	0.987*
EF	0.955*	0.999*	0.892*	0.991*

The intra-class correlation coefficients (ICC) as an index of test–retest reliability were estimated for the CP and control group separately. The ICC value for force is the value for the  $F_{TOT}$ , and the ICC value for the EMG is the average of the values for the five muscles measured

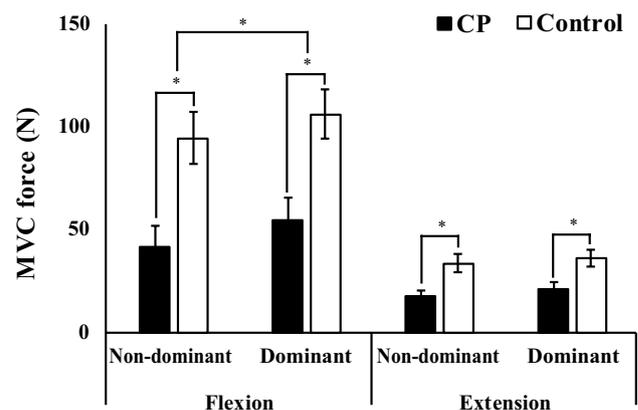
\* $p < 0.05$

longer than that for the controls ( $81.5 \pm 11.8$  s) with no significant differences between the hands for both groups, which was confirmed by a significant main effect of *Group* ( $F_{[1,15]} = 32.74, p < 0.001, \eta_p^2 = 0.69$ ) only, with no *Group*  $\times$  *Hand-Dominance*. Similarly, the number of dropped pegs for the CP group ( $4.5 \pm 5.6$ ) was larger than that for the controls ( $0.4 \pm 0.5$ ), and the group difference of the number of dropped pegs was larger in the dominant hand than in the non-dominant hand. These results were supported by a two-way RM ANOVA with factors *Group* (two levels: CP and control) and *Hand-Dominance* (two levels: dominant and non-dominant), which showed significant main effects of *Group* ( $F_{[1,15]} = 6.14, p = 0.02, \eta_p^2 = 0.29$ ) and the significant *Group*  $\times$  *Hand-Dominance* ( $F_{[1,15]} = 6.67, p = 0.03, \eta_p^2 = 0.31$ ).

### Maximal voluntary contraction (MVC) and single-finger ramp force production task

#### MVC force

The MVC force of the CP group was lower than that of the control group by about 52% and 45% for the flexion and extension direction, respectively. On average, the dominant hand showed larger MVC force than the non-dominant hand for both the CP and control group, especially during the flexion effort. In addition, the flexion MVC force was larger than the extension MVC force (Fig. 3). These observations were supported by separate two-way RM ANOVAs with the factors *Group* (two levels: CP and control) and *Direction* (two levels: flexion and extension) on the MVC forces of the dominant and non-dominant hand, which showed significant main effects of *Group* ( $F_{[1,15]} = 24.59, p < 0.001, \eta_p^2 = 0.63$  for dominant;  $F_{[1,15]} = 10.46, p = 0.006, \eta_p^2 = 0.41$  for non-dominant) and

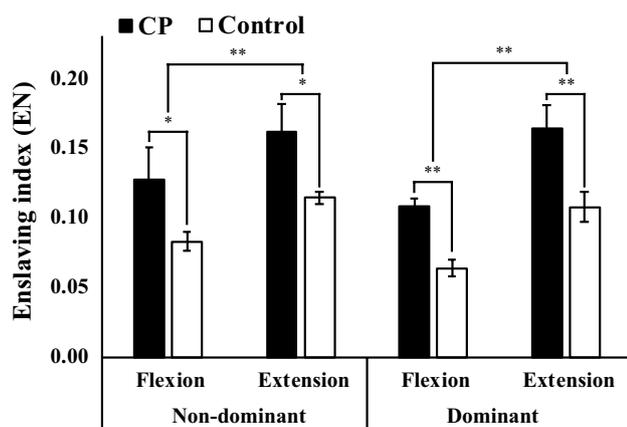


**Fig. 3** Maximal voluntary contraction (MVC) forces during flexion and extension effort for the dominant and non-dominant hands for the CP (CP; black bars) and control group (control; white bars). Values are means  $\pm$  standard errors across the subjects. Single asterisks (\*) indicate  $p < 0.05$

*Direction* ( $F_{[1,15]}=70.41, p<0.001, \eta_p^2=0.82$  for dominant;  $F_{[1,15]}=44.77, p<0.001, \eta_p^2=0.75$  for dominant) with significant interactions between the factors for both hands ( $F_{[1,15]}=8.36, p=0.01, \eta_p^2=0.36$  for dominant;  $F_{[1,15]}=8.34, p=0.01, \eta_p^2=0.36$  for dominant). The significant factor interactions reflected the fact that the effect of *Group* (i.e., group differences, CP < control) on the MVC forces were stronger in the flexion than in the extension for both hands. Further, pairwise comparisons confirmed that the MVCs of the dominant hand were larger than those of the non-dominant hand for each group for the flexion condition ( $p<0.05$ ).

### Enslaving index (EN)

In general, unintended force production by non-task fingers was obvious for both the CP and control group. The enslaving index (EN) is assumed to be the measure of finger independency that was quantified by the data from the single finger ramp force production task. On average, the ENs were smaller in the control group than in the CP group and were larger at the extension than at the flexion for both groups (Fig. 4). However, there were no significant differences in the enslaving indices between the hands for both groups. Two-way RM ANOVAs with factors *Group* and *Direction* were performed separately on the ENs of the dominant and non-dominant hand, which confirmed the significant main effects of *Group* ( $F_{[1,15]}=24.59, p<0.001, \eta_p^2=0.62$  for dominant;  $F_{[1,15]}=7.92, p=0.013, \eta_p^2=0.35$  for non-dominant) and *Direction* ( $F_{[1,15]}=29.93, p<0.001, \eta_p^2=0.67$  for dominant;  $F_{[1,15]}=24.02, p<0.001, \eta_p^2=0.62$  for non-dominant) with no factor interactions. Pairwise comparisons between the ENs of



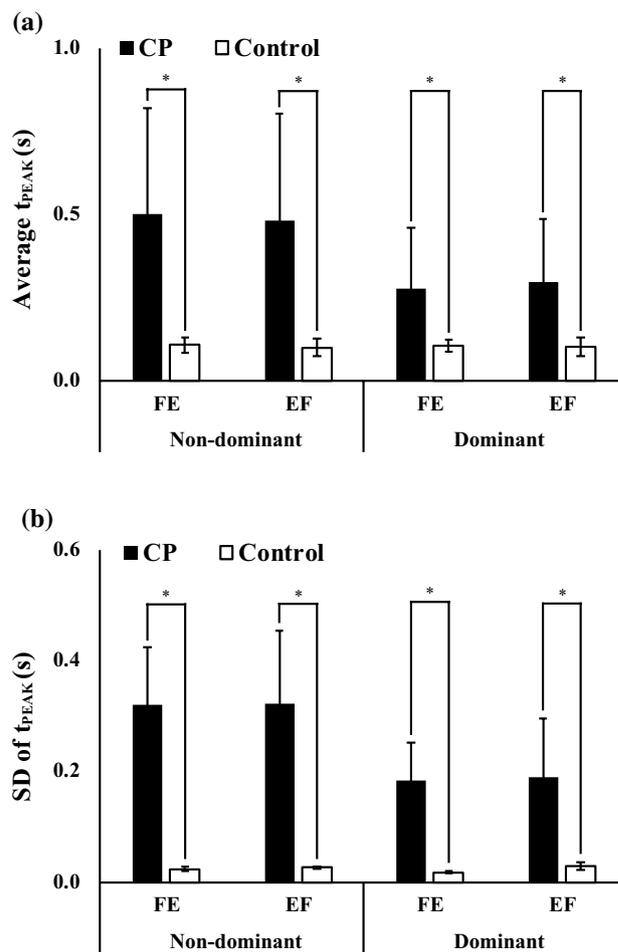
**Fig. 4** Index of finger force enslaving during the flexion and extension effort for the dominant and non-dominant hands for the CP (CP; black bars) and control group (control; white bars). Average values for CP and control group are presented with bars representing standard error (SE). Single asterisks (\*) indicate  $p<0.05$ ; double asterisks (\*\*) indicate  $p<0.001$

the dominant and non-dominant hand showed no significant differences for each group and direction.

### Multi-finger quick pulse force production task

#### Timing variables

The average and standard deviation (SD) of time to reach peak pulse force ( $t_{PEAK}$ ) with respect to  $t_0$  (i.e., time onset of detectable force change for the quick pulse force) across repetitive trials was computed. Both the average  $t_{PEAK}$  (Fig. 5a) and SD of  $t_{PEAK}$  (Fig. 5b) of the controls (average  $t_{PEAK}$ : 0.104 s; SD of  $t_{PEAK}$ : 0.024 s) was smaller than those



**Fig. 5** Average **a** and standard deviation (SD) **b** of the time of peak total force ( $t_{PEAK}$ ) across repetitive trials during the quick pulse force production tasks for the CP (CP; black bars) and control group (control; white bars). FE and EF represent the experimental condition where the first letter stands for the force direction at the steady-state and the second letter indicates the direction of quick pulse force. “F” and “E” indicate flexion and extension, respectively, and the results include both the dominant and non-dominant hands. Averaged across subjects data are presented with standard error bars. Single asterisks (\*) indicate  $p<0.05$

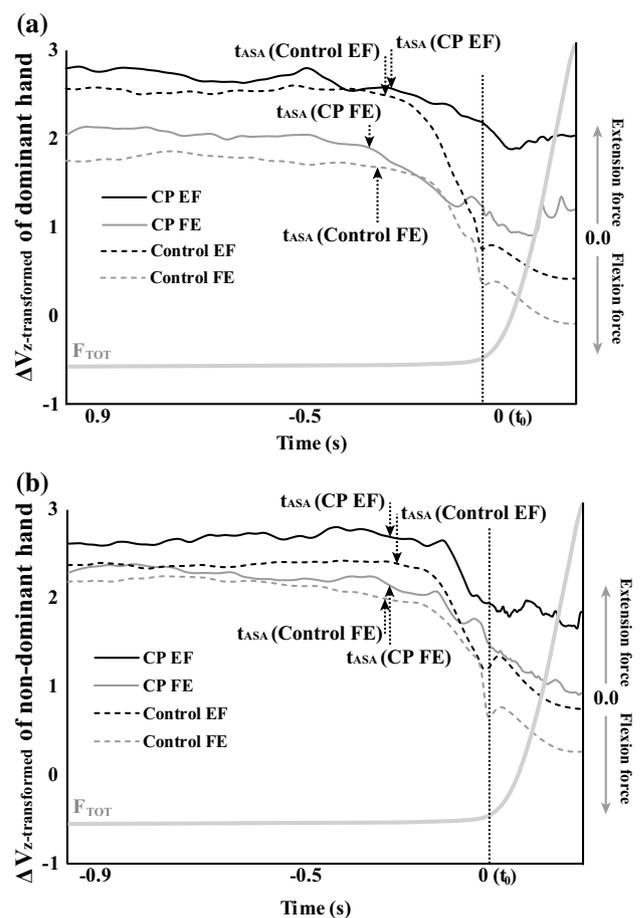
of the CP (average  $t_{PEAK}$ : 0.389 s; SD of  $t_{PEAK}$ : 0.173 s), respectively. In addition, there were no significant differences on both the average and SD of  $t_{PEAK}$  between the hands and directions for both groups. These findings were supported by two-way RM ANOVAs with factors *Group* and *Direction* for each hand, which confirmed the significant main effects of *Group* on the average  $t_{PEAK}$  ( $F_{[1,15]} = 7.56$ ,  $p = 0.015$ ,  $\eta_p^2 = 0.34$  for dominant,  $F_{[1,15]} = 7.49$ ,  $p = 0.015$ ,  $\eta_p^2 = 0.33$  for non-dominant) and the SD of  $t_{PEAK}$  ( $F_{[1,15]} = 6.82$ ,  $p = 0.02$ ,  $\eta_p^2 = 0.32$  for dominant,  $F_{[1,15]} = 12.24$ ,  $p = 0.003$ ,  $\eta_p^2 = 0.45$  for non-dominant) with no significant factor interactions.

First, we computed the variances of two sub-spaces (e.g.,  $V_{UCM}$  and  $V_{ORT}$ ) across repetitive trials. Generally,  $V_{UCM}$  (i.e., the variance in the null space of elemental variables) was larger than  $V_{ORT}$  (i.e., the variance in the orthogonal sub-space to the  $V_{UCM}$ ) for most of the conditions and subjects, thus, the description and statistical analysis performed separately on  $V_{UCM}$  and  $V_{ORT}$  as well as the synergy index,  $\Delta V$ . On average, both  $V_{UCM}$  and the  $V_{ORT}$  for the controls were smaller than those for the CP by about 91% and 78%, respectively. However, the statistical difference between the CP and the controls has been supported only for the  $V_{UCM}$ , especially at the EF condition. A two-way RM ANOVA with factors *Group* and *Hand-Dominance* for the EF condition confirmed a significant main effect of *Group* on the  $V_{UCM}$  ( $F_{[1,15]} = 26.46$ ,  $p < 0.001$ ,  $\eta_p^2 = 0.64$ ), with no factor interaction.

The magnitude of synergy indices during the steady-state force production ( $\Delta V_{SS}$ ) was not significantly different between the CP and controls, while  $\Delta V_{SS}$  for both the CP and controls at the EF condition (i.e., steady-state extension force production) was larger than that at the FE (i.e., steady-state flexion force production) condition for the dominant hand (Fig. 6b). These observations were confirmed by separate two-way RM ANOVAs with factors of *Group* and *Directions* at the dominant and non-dominant hands, which showed a significant main effect of *Direction* for the dominant hand ( $F_{[1,9]} = 16.16$ ,  $p = 0.001$ ,  $\eta_p^2 = 0.52$ ) without factor interactions. In addition, the hand difference on  $\Delta V_{SS}$  was observed in the controls (Fig. 6), especially at the FE condition ( $p < 0.05$ ), which was confirmed by a significant *Group* × *Hand-Dominance* ( $F_{[1,9]} = 8.39$ ,  $p = 0.018$ ,  $\eta_p^2 = 0.48$ ). There was no hand difference in the CP group for both the FE and EF conditions.

### Anticipatory synergy adjustment

We quantified the feedforward changes in the synergy properties (i.e., anticipatory synergy adjustment, ASA) including time ( $t_{ASA}$ ) and magnitude ( $\Delta \Delta V_{t_0}$ , i.e., the difference in  $\Delta V$  at  $t_0$  with respect to  $\Delta V$  at the steady-state in Fig. 2). It was commonly observed that the synergy



**Fig. 6** The total force ( $F_{TOT}$ , gray line) and synergy index of the CP group (solid lines) and control group (dashed lines) for the EF (black color) and FE condition (gray color) at the dominant (a) and non-dominant hand (b). Average values across subjects are presented. The times of ASA initiation ( $t_{ASA}$ ) are shown with the arrows

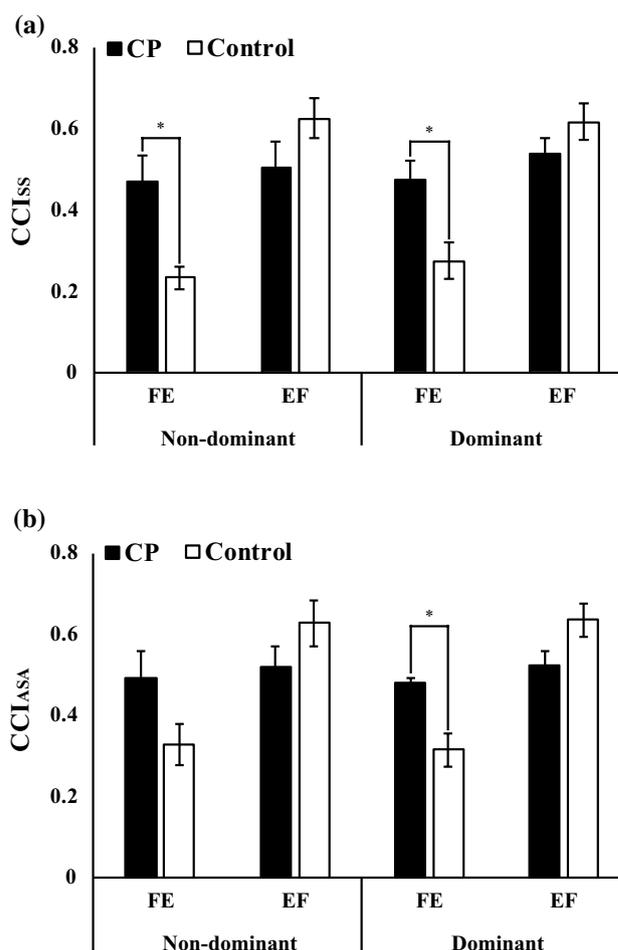
index dropped while no detectable changes in  $F_{TOT}$  prior to the initiation of the force pulse for all experimental conditions and groups (Fig. 6). Also, the  $\Delta V$  continued to decrease and reached its negative peak before the peak  $F_{TOT}$  (i.e., pulse force) was produced. On average,  $t_{ASA}$  was about 0.31 s, and there were no significant differences in  $t_{ASA}$  between the CP and controls for all experimental conditions. However,  $\Delta \Delta V_{t_0}$  was smaller in the CP group than in the controls by about 53% (i.e., smaller drop in  $\Delta V$  at  $t_0$  in the CP), while no significant difference between the directions (FE and EF) for both the dominant and non-dominant hand. The two-way RM ANOVAs with factors *Group* and *Hand-Dominance* showed significant main effects of *Group* only ( $F_{[1,15]} = 10.91$ ,  $p = 0.005$ ,  $\eta_p^2 = 0.42$  for FE;  $F_{[1,15]} = 8.73$ ,  $p = 0.01$ ,  $\eta_p^2 = 0.37$  for EF), with a significant *Group* × *Hand-Dominance* for the EF condition ( $F_{[1,15]} = 5.08$ ,  $p = 0.04$ ,  $\eta_p^2 = 0.26$ ). A significant interaction for the EF condition reflected the fact that the

group difference at the EF condition was stronger in the dominant hand than in the non-dominant hand. In addition, pairwise comparisons confirmed that  $\Delta\Delta V_{t_0}$  of the dominant and non-dominant hand showed no significant differences for each group and direction.

We further computed the  $\Delta\Delta V_{PEAK}$  that is  $\Delta V$  changes in its magnitude with respect to the negative peak value of  $\Delta V$  after  $t_0$  (Fig. 2). Similarly,  $\Delta\Delta V_{PEAK}$  was smaller in the CP group than in the controls, and the difference between the two groups was stronger in the dominant hand conditions (58% and 19% for the dominant and non-dominant hand, respectively) especially at the EF condition. Two-way RM ANOVAs with factors *Group* and *Hand-Dominance* confirmed the significant main effects of *Group* ( $F_{[1,15]} = 6.42$ ,  $p = 0.02$ ,  $\eta_p^2 = 0.30$  for EF) with significant *Group*  $\times$  *Hand-Dominance* ( $F_{[1,15]} = 7.04$ ,  $p = 0.018$ ,  $\eta_p^2 = 0.30$  for EF) only at the EF condition. The significant *Group*  $\times$  *Hand-Dominance* reflected the fact that the group difference on  $\Delta\Delta V_{PEAK}$  was stronger in the dominant hand ( $p < 0.05$ ) than the non-dominant hand. Seemingly, the  $\Delta\Delta V_{PEAK}$  was smaller in the CP than in the controls for the FE condition, whereas the difference did not reach statistical significance ( $p = 0.07$ ).

### Co-contraction of forearm muscles

The co-contraction index (CCI) during the steady-state force production ( $CCI_{SS}$ ) and anticipatory phase ( $CCI_{ASA}$ ) were computed. There were no significant differences on both  $CCI_{SS}$  and  $CCI_{ASA}$  between the dominant and non-dominant hands, while the both  $CCI_{SS}$  (Fig. 7a) and  $CCI_{ASA}$  (Fig. 7b) were larger in the CP group than in the controls especially at the FE condition by about 85% and 23%, respectively. For both the  $CCI_{SS}$  and  $CCI_{ASA}$ , two-way RM ANOVAs with factors *Group* and *Direction* were performed separately on the CCIs of the dominant and non-dominant hand, which confirmed the significant main effects of *Group* ( $F_{[1,15]} = 5.23$ ,  $p = 0.037$ ,  $\eta_p^2 = 0.26$  for dominant,  $F_{[1,15]} = 6.4$ ,  $p = 0.023$ ,  $\eta_p^2 = 0.3$  for non-dominant) and *Direction* ( $F_{[1,15]} = 15.97$ ,  $p = 0.001$ ,  $\eta_p^2 = 0.52$  for dominant,  $F_{[1,15]} = 9.78$ ,  $p = 0.007$ ,  $\eta_p^2 = 0.4$  for non-dominant) with significant *Group*  $\times$  *Direction* ( $F_{[1,15]} = 7.5$ ,  $p = 0.015$ ,  $\eta_p^2 = 0.33$  for dominant,  $F_{[1,15]} = 7.06$ ,  $p = 0.018$ ,  $\eta_p^2 = 0.32$  for non-dominant) for  $CCI_{SS}$ , and the significant main effect of *Direction* ( $F_{[1,15]} = 19.02$ ,  $p = 0.001$ ,  $\eta_p^2 = 0.56$  for dominant,  $F_{[1,15]} = 5.16$ ,  $p = 0.038$ ,  $\eta_p^2 = 0.26$  for non-dominant) with significant *Group*  $\times$  *Direction* ( $F_{[1,15]} = 11.01$ ,  $p = 0.005$ ,  $\eta_p^2 = 0.42$  for dominant) for  $CCI_{ASA}$ . Pairwise comparisons confirmed both  $CCI_{SS}$  and  $CCI_{ASA}$  for the CP > the controls at the FE for both the dominant and non-dominant hand ( $p < 0.05$ ) except  $CCI_{ASA}$  for the non-dominant hand ( $p = 0.09$ ).



**Fig. 7** Co-contraction index at the steady-state ( $CCI_{SS}$ ) **(a)** and the ASA phases ( $CCI_{ASA}$ ) **(b)** of the CP (CP; black bars) and control group (control; white bars) during the four experimental conditions (FE and EF for the dominant and non-dominant hands) of quick pulse force production tasks. Values are means  $\pm$  standard errors across the subjects. Single asterisks (\*) indicate  $p < 0.05$

### Discussion

The hypothesis formulated in the Introduction that the CP group would show significant differences compared to the control group in the indices of performances and a set of indices quantified by the uncontrolled manifold (UCM) computation has been partially supported. The CP group differed significantly from the control group in the indices of performance, including the MVC (smaller MVC in the CP) and the time to reach peak force ( $t_{PEAK}$  was longer in the CP). Further, the enslaving index (EN) of the CP was larger than that of the control group (i.e., less individuated finger force production in the CP group). On the contrary, the magnitudes of the synergy indices during the steady-state phase ( $\Delta V_{SS}$ ) showed no effect of CP, and the difference of  $\Delta V_{SS}$  between the flexion and extension effort ( $\Delta V_{SS}$

of the extension > flexion) was observed in both the CP and controls. Notably, there was no statistical difference on  $\Delta V_{SS}$  between the two hands for the CP group, while the handedness of the synergy index was observed in the control group (i.e.,  $\Delta V_{SS}$  of the dominant hand < non-dominant hand). In addition, the time of the anticipatory synergy adjustment ( $t_{ASA}$ ) was not significantly delayed in the CP group compared with that of the control group. Significant differences in the indices from the synergy computation between the groups were associated with a drop in the magnitudes of the synergy index ( $\Delta\Delta V_{t_0}$ ) during ASA ( $t_{ASA}$  to  $t_0$ ) as well as during quick pulse force production ( $\Delta\Delta V_{PEAK}$ ). The CP group showed a significantly less drop in the synergy indices during the anticipatory and quick pulse phase compared to the control group, which was accompanied by larger co-contraction indices of the forearm muscles (e.g.,  $CCI_{ASA}$  and  $CCI_{SS}$ ) as compared to the control group.

### Loss of strength and individuation in finger force production with cerebral palsy

A considerable number of studies regarding the phenomenon of finger actions have provided ample evidence that the actions of the individual fingers, such as moving or producing forces are not independent (Li et al. 2004). The lack of individuation resulting in the unintended coupled actions of the fingers (i.e., positive co-variation of finger actions) has been termed as enslaving (Li et al. 1998), and enslaving is known to be attributed by several factors including biomechanical (e.g., passive connection, multi-tendon effect of the extrinsic hand muscles) and central factors (e.g., overlapped cortical projection) (Li et al. 1998, 2002; Schieber 1990; Schieber and Hibbard 1993; Zatsiorsky et al. 2000). A series of previous studies showed that relatively high enslaving indices were observed in patients with neurological disorders such as Parkinson's disease (Park et al. 2012), cerebellar disorders (Park et al. 2013), and stroke (Jo et al. 2016). Additionally, the high enslaving of the finger force production in the patient mentioned earlier was associated with a weakened force production capability (e.g., smaller MVC forces). Similarly, the CP group in this study showed higher enslaving but smaller MVC force compared to the control group for both flexion and extension. These results are not surprising since it is well known that the skeletal muscle mass in individuals with CP is smaller compared to the age- and gender-matched control groups (Elder et al. 2003). However, given that the reduced MVC forces were accompanied by relatively smaller enslaving indices with healthy aging (Oliveira et al. 2008; Shinohara et al. 2003) and during the fatigue process (Danion et al. 2001), the muscle strength would not be a critical parameter to gauge finger individuation and enslaving.

A possible interpretation for this discrepancy is that peripheral weakness is not a direct cause of the less independent actions of the elements, but the abnormal supraspinal process may have a larger contribution to the less independent actions of the elements (i.e., finger forces in this study) for CP individuals as well as for other populations with neurological disorders. Note that the differences in the MVC forces between the dominant and non-dominant hands were not accompanied by the changes in the enslaving indices for the CP group in the current results (i.e., no significant effect of *Hand-Dominance* on the enslaving index). Seemingly, perfect individuation of the finger actions is probably the best situation for dexterous hand actions. However, experimental evidence suggests that the enslaving patterns of normal people are beneficial to some extent for the stabilization of rotational actions that require positive co-variation of the finger forces (Park et al. 2012; Zatsiorsky et al. 2000). Although there is no known criterion on the standard values of the enslaving for “good” performance using hand and fingers, it is highly probable that excessive enslaving hampers flexible involvement of fingers resulting in a detrimental effect on precise manipulation of the net actions by multiple fingers of CP individuals. Besides, the enslaving index for both the control and CP group was larger during the extension effort than during the flexion effort in the current study, while the extension of the MVC force was relatively large. These observations further support the idea that the causality between the peripheral strength and individuation of finger force production is not robust.

Lastly, for the quantification of the coordination index (i.e., synergy index) in the next section, the space of the elemental variables in the computation was not a force space, but a mode space where the variables of the hypothetical commands (i.e., mode vector) were defined while removing the effects of enslaving (Danion et al. 2003). Note that the mode vector ( $\mathbf{m}$  in Eq. 3) represents the intended involvement of all four fingers by the neural command (Zatsiorsky et al. 2000). Therefore, the UCM computation using mode vectors presumably describes how the neural commands to the fingers, not actual finger forces, are organized to execute a given motor task. Since CP individuals showed a high enslaving of finger force production, the analysis of the synergy index with the mode vector was more conservative than with the force vector (i.e., the difference was less inflated due to the effect of enslaving). Thus, comparison of the synergy indices in the mode space may provide a better insight into the changes in the stability properties by the neural commands to finger actions observed in CP individuals. Also, the outcomes of the mode space analysis in the current study would be comparable to the results of previous studies with stroke survivors and PD patients (Park et al. 2012; Jo et al. 2016).

## The intact ability of synergy formation, but impaired ability to modulate its properties with cerebral palsy

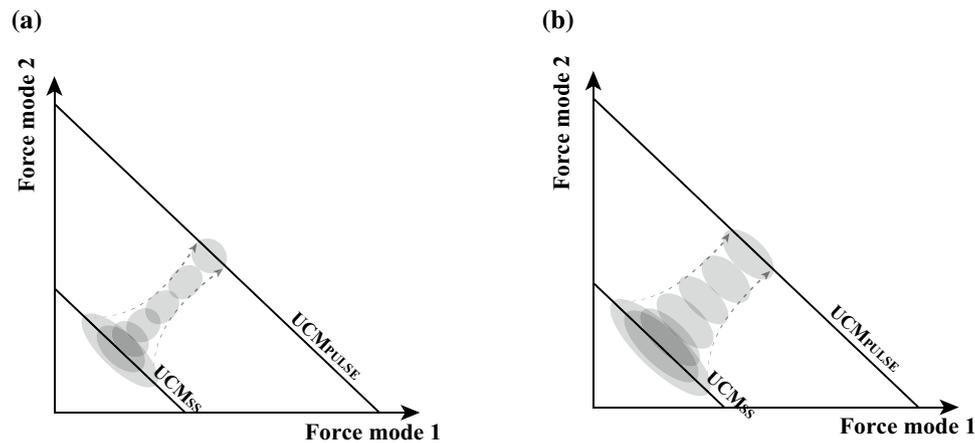
In the current study, the term *synergy*, describes the process of the organization of neural structures thus ensuring flexible patterns of actions of redundant elements to maintain the stability properties of voluntary movements (Latash et al. 2001; Li et al. 1998; Scholz et al. 2002). Thus, the redundant set of elements within a human movement system do not cause a situation that stretches the computational loads on the human controller for the formation of an optimal combination of elements. Rather, it may arrange a prerequisite condition in which flexible combinations of elements equally satisfy the task mechanics (i.e., synergic actions) by compensating for errors associated with performance (Kim et al. 2018; Latash et al. 2003, 2004; Li et al. 1998; Park et al. 2011). An operational definition of motor synergies has been proposed to quantify motor coordination with the help of the computational method of the uncontrolled manifold approach (Scholz and Schönner 1999; Scholz et al. 2000). Motor synergy indices have been defined as the neural organization of elemental variables to ensure stable performance (e.g., total force,  $F_{TOT}$ , in the current study). In other words, the synergy indices are indicative of the stability properties of the functional and mechanical outcomes performed by multiple elements.

On the contrary, we are often facing situations that need to change the performance variables (i.e., body mechanics) quickly resulting in intentional actions such as walking, moving a hand-held object, etc. This means that the performance variables (e.g., the center of mass of the whole body, net finger forces or torques) do not need to be constant but need to be changed quickly. In this sense, intentional changes in the performance variables hamper the preservation of the stability of the performance or vice versa; thus, the change in the performance variables may be considered error values of the performance values that are supposed to be maintained or stabilized at a specific moment. Therefore, the controller may need to destabilize the system strategically when it comes to the changes in body mechanics by means of switching off the synergy (Latash 2008). Indeed, it has been well shown that the human controller is capable of modulating the stability properties (i.e., synergy indices) even prior to visible changes in performance (Olafsdottir et al. 2005), which has been termed anticipatory synergy adjustment (ASA). Thus, the anticipatory synergy adjustment is assumed to be a part of the neuronal process regarding the feedforward mechanism to prepare future actions against predictable perturbations that may occur externally or due to voluntary changes. Given that the peripheral strength and directions of the net mechanics (e.g., net force or torque) seems not to influence the measures of ASA in

healthy adults (Park et al. 2015; Park and Xu 2017), the neural mechanism of the feedforward adjustment of the synergy properties may have a different origin from the process of organizing the co-variation patterns of the elements associated with a stable performance. Note that feedforward destabilization (i.e., anticipatory synergy adjustment) is not a consequence of the changes in the salient performance variables since the synergy strength is partially dependent on the magnitude of the variance of elements that do not affect the performance ( $V_{UCM}$ ) and the destabilization is the reflection of the reduced magnitude of the variance. Of course, the synergy indices are continually being reduced (i.e., destabilized) even after the salient performance variables begin to change (Fig. 8).

In the current study, we quantified the stability index during the steady-state force production ( $\Delta V_{SS}$  in the results) as well as the two indices of property modulation of the  $\Delta V_{SS}$  during the anticipatory adjustment including the information about “when” (i.e., time initiation of the drop in the synergy indices,  $t_{ASA}$ ) and “how much” (magnitude changes,  $\Delta \Delta V$ ). Recent studies have reported both parallel and non-parallel changes in the index of synergy strength and its feedforward modulation depending on the types of progressive neurological diseases. A group of previous studies has reported a parallel change in the strength of synergy (i.e., magnitude of  $\Delta V$ ) and the time-set of feedforward modulation (e.g.,  $t_{ASA}$ ), especially for the patients with neurological disorders where the subcortical structure is damaged. For example, the patients with Parkinson’s disease (Park et al. 2012) and cerebellum atrophy (Park et al. 2013) showed the decreased  $\Delta V$  for the stable force production along with the delayed time of ASA for the preparation of quick change of the performance variables (e.g., net end-effector force). Another important finding with the patients with the subcortical disorder was that the hand difference on the synergy strength, which was observed in healthy controls (i.e., larger synergy indices in the non-dominant hand), was preserved. The handedness on the synergy strength is in line with the dynamic dominance hypothesis (Sainburg 2002), which describes the role of lateralization in human limb control.

On the contrary, non-parallel changes of a set of synergy indices were observed in stroke survivors during an exercise involving both arms (Reisman and Scholz 2003) and multi-finger force production tasks (Jo et al. 2016). Cortical stroke survival showed a delayed ASA but no difference in the strength of synergy as compared to the control subjects (Jo et al. 2016). For CP subjects, in the current study, there was no statistical difference in the steady-state synergy index compared to the controls. This observation is compatible with the results involving stroke survivors whose cortical structure is damaged, albeit CP subjects showed a preserved ability to initiate feedforward adjustment such that the time of the anticipatory synergy (i.e.,  $t_{ASA}$  in the Results)



**Fig. 8** An illustration of changes in data distribution over repetitive trials for the control group (**a**) and CP group (**b**). In the space of finger mode {mode 1; mode 2}, solid lines represent the uncontrolled manifolds (UCMs) for the total force ( $F_{TOT}$ ) magnitudes of the steady-state (smaller magnitude of  $F_{TOT}$ ) and quick pulse (larger magnitude of  $F_{TOT}$ ). A series of data distributions and their shapes depict the changes in sharing pattern and two variances,  $V_{UCM}$  and  $V_{ORT}$ , in

a time-series during the quick pulse force production, which further illustrate the main difference of  $\Delta V$  modulation (i.e., destabilization) between the CP and control group during the anticipatory synergy adjustment and quick pulse force production phases. Note that the changes in data distribution in a time-series differ between the two groups, which shows that the shapes of the data distributions do not change dramatically in the CP group (**b**)

adjustment was not affected by muscle strength and movement direction (e.g., flexion and extension by flexors and extensor, respectively) as the control subjects did. On the other hands, the differences between the synergy indices in the non-dominant and dominant hand during steady-state force production were disappeared in CP subject in the current study. Note that the handedness is assumed to be associated with cortical mechanism. Thus, it is highly probable that CP is associated with detrimental changes in the cortical mechanism, not in the function of the sub-cortical structure. Thereby, a possible interpretation of the current findings is that, first; the formation of synergy for stable performance may be the function of the subcortical structures (i.e., trans-thalamic loop, Latash and Huang 2015; Rispal-Padel et al. 1981) and is relatively unaffected in cortical dysfunction and, second; the feedforward mechanism may compromise a wide range of neural networks including the loops of the corticospinal, basal ganglia, and cerebellum.

Another interesting finding was the significant dissociation between the modulations of the two synergy properties. CP subjects showed a preserved ability to initiate feedforward adjustment (i.e.,  $t_{ASA}$ ), whereas the ability to reduce the magnitude of synergy strength (i.e., ability to destabilize the performance) seems to be impaired for the CP subject in the current study. It might seem counterintuitive, but the dissociation observed in CP individuals may be related to spasticity. Spasticity is accompanied by increased viscoelastic stiffness of the peripheral structure and stretch reflex as well as less flexible shifting of the length-threshold ( $\lambda$ ) of the tonic stretch reflex (Katz and Rymer 1989). In particular, according to the equilibrium point hypothesis

(Feldman 1986), shifting of the threshold ( $\lambda$ ) of the tonic stretch reflex is one of the crucial factors associated with the regulation of the levels of muscle activation along with muscle length (i.e., force-length relationship). Due to the stiffened contractile properties of muscles, it may be possible that the range of the  $\lambda$ -shifting is limited for CP individuals resulting in a loss of muscle relaxation. Indeed, a retained muscle co-contraction is observed in spastic CP even during the unsupported phase of walking (Damiano et al. 2000) where the deactivation of an antagonist muscle group is necessary to swing the unsupported leg and to move forward (Frost et al. 1997). Recent data about the high co-contraction index observed during both steady-state and the anticipatory phase shown in the CP group are consistent with previous findings, and further support the possible mechanism of spasticity speculated by the equilibrium point hypothesis. In other words, spasticity that is partially quantified by the co-contraction index in the current study in the CP group may be associated with dysfunction of purposeful destabilization of the performance to make proper changes of body mechanics even during isometric muscle contraction (i.e., minimal changes in muscle length). Figure 8 shows the simplified illustration of the current results regarding the main difference of  $\Delta V$  modulation (i.e., destabilization) between the control (Fig. 8a) and CP group (Fig. 8b) in the space of the two force modes. The two slanted lines in the figure represent the uncontrolled manifolds (UCMs) for the total force ( $F_{TOT}$ ) magnitudes at the steady-state ( $UCM_{SS}$  in the figure) and quick pulse ( $UCM_{PULSE}$  in the figure). A series of data distributions in Fig. 8a, b and their shapes depict the changes in sharing patterns and two variances,

$V_{UCM}$  and  $V_{ORT}$ , in a time-series during quick pulse force production. The shape of the data distributions between the two groups is similar ( $V_{UCM} > V_{ORT}$ ) during the steady-state force production, while the relative amount of both  $V_{UCM}$  and  $V_{ORT}$  is larger for the CP group as compared to the control group. However, the changes in data distribution in a time-series differ between the two groups. Notably, the data distribution for the control group is circular ( $V_{UCM} < V_{ORT}$  or  $V_{UCM} \approx V_{ORT}$ ) during quick pulse force, while the shapes of the data distributions do not change dramatically in the CP group ( $V_{UCM} > V_{ORT}$ ). In turn, the neural function for the purposeful destabilization for a quick change in the net force may be impaired in CP.

The results of the current experiment provide a few intriguing outcomes related to the changes in the control strategy of CP individuals. However, the relatively small sample size and limited age-range of the recruited subjects in this study are apparent drawbacks; thus, the conclusion of this study should be viewed as tentative. CP is known to be a non-progressive disorder, but there is still some doubt as to whether abnormal control strategies for stabilization and destabilization of a particular performance persist throughout their lifetime or change with various treatments such as physical activities, surgery, anti-spastic medication (e.g., intrathecal baclofen), etc.

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