Decrease in force control among older adults under unpredictable conditions

Ebisu Shunsuke[a], Kasahara Satoshi[b*], Saito Hiroshi[c], Ishida Tomoya[b]

[a]Department of Physical Therapy, Division of the Medical Technology, Tokachi Rehabilitation Center

[b]Department of Rehabilitation Sciences, Faculty of Health Sciences, Hokkaido University, Sapporo, Japan

[c]Department of Physical Therapy, School of Rehabilitation, Tokyo Kasei University, Sayama, Japan

***Corresponding author**

Kasahara Satoshi

Department of Rehabilitation Sciences, Faculty of Health Sciences, Hokkaido University

West 5, North 12, Kita-ku, Sapporo 060-0812, Japan

Tel: +81-11-706-3390

Fax: +81-11-706-3390

E-mail: kasahara@hs.hokudai.ac.jp

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[1]LR: low-range; HR: high-range; RMSE: root mean square error; CCI: co-contraction index; PFs: plantar flexors; ROM: range of motion; FRT: functional reach test; TA: tibialis anterior; GA: gastrocnemius; sEMG: electromyography system

Abstract

Objectives: Falls in older adults generally occur during unpredictable situations. Controlling posture through fine-tuned muscle force before and after falls is necessary to avoid serious injuries. However, details regarding force control among older adults during unpredictable situations are unclear. This study determined the features of force control in a random force-tracking task among older adults.

Methods: Ten healthy older adults (67–76 years) and eight healthy young adults (20–23 years) participated in three force-tracking tasks with ankle plantar flexion: low-range (LR), high-range (HR), and pseudo-random (PR) force tasks. Force control ability was assessed using the root mean square error (RMSE) between the target and muscle forces produced by the participants. Muscle activities from the lateral head of the gastrocnemius and the tibialis anterior during each task were measured using surface electromyography to calculate the co-contraction index (CCI).

Results: In all tasks, older adults (RMSEs: 1.09–3.70, CCIs: 29.4–56.4) had a significantly greater RMSEs and CCIs than young adults (RMSEs: 0.49–1.83, CCIs: 11.7–20.6; all, $p < 0.05$). The RMSEs during force generation were significantly greater than those during force release (LR: $p < 0.01$, HR: $p < 0.05$), except for the random force-tracking task in older adults. CCIs during the force release phase in both groups (older adults: 27.8–56.4, young adults: 15.0–20.6) were consistently greater than those during force generation (older adults: 24.5–50.4, young adults: 11.7–17.2). CCIs in force-tracing tasks differed in older adults, whereas those in the random force-tracing task increased. RMSEs and CCIs in the random and LR force-tracing tasks were significantly negatively correlated with the functional reach test (all: $r > 0.5$, $p < 0.05$).

Conclusion: Force control in older adults declines in low-band and random muscle force output. Moreover, increased CCIs in older adults are particularly pronounced during unpredictable situations.

Keywords: Aging; Ankle muscles; Electromyography; Muscle strength; Motor control

1. Introduction

When performing daily activities and work, physical movement is achieved by fine-tuned muscle strength (i.e., muscle output) and coordinated muscle contractions between agonists and antagonists. Postural and motor control requires continually fluctuating force levels because most functional activities are performed in varying situations (Perraton et al., 2017; Ward et al., 2019; Williams et al., 2003). However, the sensorimotor system gradually degenerates with advancing age, and motor errors increase (Hortobágyi et al., 2001; Park et al., 2016).

The decline in motor performance with aging is mainly attributed to changes in muscle strength, muscle structure, and muscle mass (i.e., sarcopenia, which is a peripheral change). However, maximal strength is not necessary for daily activities (Ward et al., 2019). It has been reported that improvement in maximal strength does not always lead to improvements in functional capacity (Hortobágyi et al., 2001). Recently, in addition to peripheral factors, age-related changes in the neuro-muscle-skeletal system (i.e., dynapenia in the central and peripheral nervous system) have been suggested to influence the accuracy and error of force control in response to task requirements (Knol et al., 2019; Manini and Clark, 2012). The variability of motor output under submaximal force increased in older adults compared to that in young adults, suggesting that the decline in the sensorimotor system with advancing age influences the feedback mechanism, which is important for muscle strength control (Hortobágyi et al., 2001).

The force control task is often used to assess coordination among muscles of motor control in the central nervous system (Knol et al., 2019). In most previous studies, the force control tasks were discrete movements that match pinch or grip force to a certain target force; thus, information regarding the ability to control muscle strength in the lower limbs is limited (Ward et al., 2019). Considering daily activities (i.e., gait and stair lifts) and postural maintenance (i.e., standing balance), force control of the lower limbs is important, particularly the ankle joint, which plays a key role in postural control (i.e., the ankle strategy) (Kasahara et al., 2015). Previous force tasks comprise a constant or fixed isometric task and make the participants perform those tasks randomly (Lauzière et al., 2012; Marchini et al., 2017; Yang et al., 2019). In addition, traditional force-tracking tasks with a ramp (Choi et al., 2019; Park et al., 2016; Patel et al., 2020; Spiegel et al., 1996) or sinusoidal (Berger et al., 2020; Knol et al., 2019; Perraton et al., 2017; Telianidis et al., 2014; Ward et al., 2019) waves comprise force repetition with similar constant amplitude. However, the risk of falls in older adults increases during unpredictable situations (Gerards et al., 2021).

This study aimed to investigate the deficits in force control in unpredictable situations with plantar flexors (PFs) in older adults. Thus, we divided the force control into two controls (generation and release) using a force-tracking task and compared them between young adults and older adults. Concurrently, age-related changes in muscle coordination (i.e., increased co-contraction) were examined using electromyography. Furthermore, we examined whether the ability of force control is related to individual balance ability. We hypothesized that: force control would decline in older adults compared to young adults, the co-contraction index (CCI) would significantly increase in older adults, and the balance ability would be associated with force control and CCI in the ankle joint, particularly in unpredictable situations.

2. Methods

2.1. Participants

Eighteen healthy adults—specifically, eight young men (21.8 ± 1.0 [standard deviation] years, 174.4 ± 3.6 cm, 68.5 ± 7.0 kg) and 10 older men (70.6 ± 2.9 years; 166.7 ± 5.4 cm; 64.4 ± 10.0 kg)—participated in this study (Table 1). A dominant foot was defined as the leg kicking a ball; all participants reported the right foot as their dominant foot. Young adults were college students who volunteered, and older adults aged >65 years were randomly selected from the community-dwelling elderly who were registered in an employment agency. All participants independently lived without problems regarding activities of daily living in their community and had no disorders or injuries as well as any neurological, vestibular, orthopedic, or cognitive conditions that could interfere with their balance. Additionally, older adults experienced no falls for at least six months prior to their enrollment in this study. Participants were excluded if their visual acuity was below 1.0, as determined using the Landolt ring chart (Kasahara and Saito, 2015). All participants provided written informed consent for their participation, and the procedures were approved by the ethics committee of Hokkaido University School of Medicine (approval no. 11-03).

	Young group	Older group	p-value
Age (years)	21.8 ± 1.0	70.6 ± 2.9	<0.001
Height (cm)	174.4 ± 3.6	166.7 ± 5.4	<0.01
Weight (kg)	68.5 ± 7.0	64.4 ± 10.0	0.364
ROM of ankle dorsiflexion (degree)	16.4 ± 6.0	13.0 ± 4.0	0.192
PF_{max} (Nm)	106.2 ± 51.0	59.5 ± 16.3	<0.01
FRT (cm)	41.6 ± 3.2	32.9 ± 4.5	<0.01

Table 1 Participant characteristics.

Values are presented as mean ± standard deviation. ROM, range of motion; PF, plantar flexion; PF_{max}, maximum force of plantar flexors (PF) during maximum voluntary isometric contraction; FRT, functional reach test.

2.2. Experimental approach

Prior to the experiment, two experienced physical therapists conducted a functional reach test (FRT) (Robinovitch and Cronin, 1999) to assess the equilibrium ability of dynamic balance. The FRT was performed with both hands and without knee flexion using a reach test device (TKK5802; Takei Scientific Instruments, Niigata, Japan). The range of motion (ROM) of the ankle dorsiflexion during the knee extension was measured to assess flexibility. The maximum force of PFs during maximum voluntary isometric contraction (PF_{max}) was measured using a Biodex dynamometer (model 3 dynamometer; Biodex Medical Systems Inc., Shirley, NY, USA) before the force-tracking tasks. All the tests were performed in triplicate.

All force-tracking tasks were performed with the dominant leg on the Biodex, and the sitting position on the Biodex was fixed in 90° hip flexion, 0° knee extension, and the neutral position of the ankle joint by the strap over the pelvis and chest through all tasks (Fig. 1A). The computer monitor was placed 60 cm in front of the participants to provide visual information regarding both the target force and the force produced by pushing the attachment in the ankle plantarflexion direction at the neutral ankle position (Fig. 1A). If the participants pushed the attachment strongly, the actual force line on the monitor moved upward. The visual gain across the three tasks was always constant. Participants were instructed to match their produced forces to the onscreen target force line as precisely and rapidly as possible and were not given specific instructions about how to perform the ankle movement.

(Figure 1)

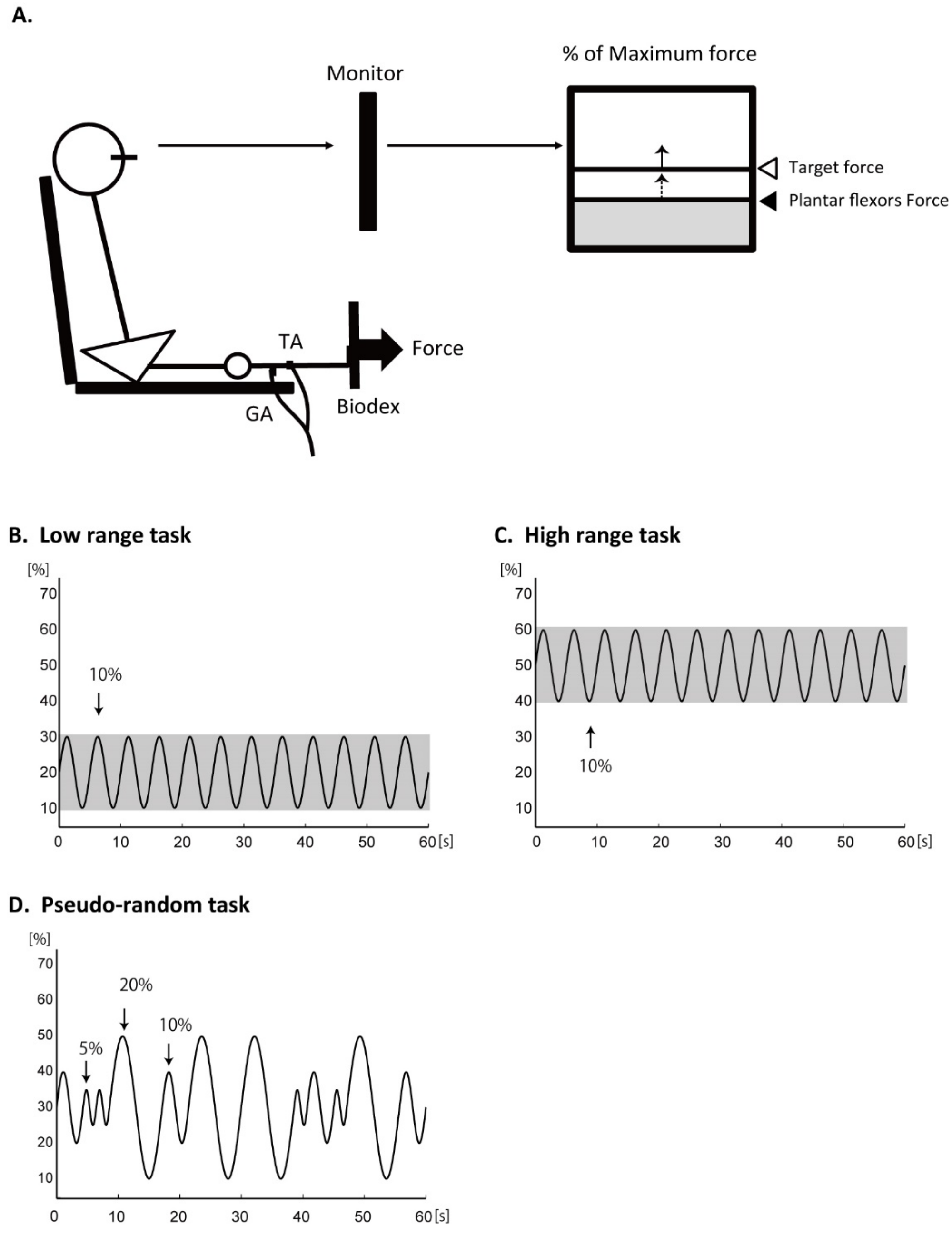

Fig. 1. A schematic representation of the experimental setup (A) and Target force trajectories (B-D). TA: tibialis anterior; GA: medial head of gastrocnemius.

The sinusoidal motion of the target force was controlled with a customized program using the LabView software (LabView 2009; National Instruments, Austin, TX, USA). The following three tasks, which comprised two constant tasks and one random task, were performed (Fig. 1B–D): (1) low-range (LR) task (Fig. 1B), with velocity at 0.2 Hz (12.6%/s), LR from 10% to 30% (Perraton et al., 2017) of PF_{max}, and amplitude of 10% PF_{max}; (2) high-range (HR) task (Fig. 1C), with velocity at 0.2 Hz, HR from 40% to 60% of PF_{max}, and amplitude of 10% PF_{max}; and (3) pseudo-random (PR) force task (Fig. 1D), with a middle range between 20% and 50% of PF_{max} and PR amplitude of 5%, 10%, and 20% PF_{max}. The maximum target velocity was constant at 0.2 Hz for all tasks because force variability is dependent on speed (Christou and Carlton, 2002; Park et al., 2016), and this sinusoidal frequency (approximately 0.2 Hz) was used for the tracking task in a previous study (Kasahara and Saito, 2015). Note that the PR task used several amplitudes with a planned random order and that the maximum velocity during each cycle in the PR task was dependent on each amplitude. Three practice trials were performed to familiarize the participants with the force-tracking task (Perraton et al., 2017; Telianidis et al., 2014). Participants were not provided with information regarding the random order of the target force in the PR task. The order of the force-tracking tasks was random across all participants. Each task time was 60 s, and rest was allowed between the tasks, if necessary.

Muscle activities of the tibialis anterior (TA) and the medial head of the gastrocnemius (GA) were concurrently recorded using a surface electromyography system (sEMG) (Bagnoli-2EMG System; DELSYS, Boston, MA, USA) for the force-tracking task (Fig. 1A). The electrodes were placed in line with each muscle fiber, and an indifferent electrode was placed on the head of the fibula according to our previous study (Kasahara et al., 2015). The sEMG signal was rectified after amplification (×1,000) and bandpass filtered from 10 to 500 Hz with a fourth-order Butterworth filter (Kasahara et al., 2015).

2.3. Data processing and analysis

All data processing and analysis were performed with a customized program using MATLAB (MathWorks Inc., Natick, Massachusetts, USA). The force signal from the Biodex was low-pass filtered with a zero-lag, second-order Butterworth filter at a cut-off frequency of 10 Hz, and sEMG signals were low-pass filtered in the same way at 6 Hz (Kasahara et al., 2015). For the constant force-tracking task, the force signal was divided for each cycle except for the first force trajectory; subsequently, using the MATLAB customized program, three randomly selected force trajectories were signal-averaged to obtain a representative force trajectory for each subject. For the random force-tracking task, three comparable force trajectory data at 10% amplitude, except for the first force trajectory, were chosen, and signal-averaging was used as representative data for each subject. Similarly, sEMG data were processed for each task, for each subject. Furthermore, data were divided into the force generation phase (i.e., increasing phase) and force release phase (i.e., decreasing phase).

The force-tracking performance was estimated as the temporal and spatial error, and the root mean square error (RMSE) between the actual and target forces was computed for every task using the following equation (1) (Knol et al., 2019):

$$RMSE = \sqrt{\frac{1}{n-1}\sum_{k=1}^{n}(T_i - F_i)^2} \qquad (1)$$

where n is the number of data points in the time series, T_i is the force target, and F_i is the actual force. The CCI between the TA and GA was assessed to estimate the ability of muscle coordination during the force-tracking task using the following equation (2) (Falconer and Winter, 1985):

$$CCI = \frac{2 \times \int sEMG\ GA}{\int sEMG\ GA + \int sEMG\ TA} \times 100 \qquad (2)$$

2.4. Statistical analysis

The adequacy of the sample size and significance level was confirmed by G*Power, with the effect size set at 0.6, the alpha at 0.05, and the power at 0.8 (Faul et al., 2007), according to Cohen's criteria (Cohen, 1988). Statistical analysis was conducted using SPSS version 18.0 (IBM Ireland Ltd., Dublin, Ireland). All data are shown as mean ± standard deviation. Normality of distribution was examined using the Shapiro-Wilk test. For unpaired comparisons between groups (i.e., demographic data, RMSE, and CCI), Student's t-test or the Mann-Whitney U test was performed. A repeated-measures two-way analysis of variance (ANOVA) (phase × task) was used to assess the differences in the RMSE and CCI in each phase. Mauchly's test of sphericity was used to test sphericity, and the Greenhouse-Geisser correction was used in the case of lack of sphericity (Pradhan et al., 2010). In addition, if there was a significant interaction, a simple effects analysis was performed with Bonferroni adjustment. The effect size was calculated as partial eta squared values (η^2) (Cohen, 1988). The correlation analysis between the RMSE or CCI and balance ability (i.e., FRT) was performed using Spearman's rank-correlation coefficient (ρ). All significance levels were set at $p < 0.05$.

3. Results

Participant characteristics are shown in Table 1. Height in the young group was significantly higher than that in the older group ($p = 0.003$). Measurements of PF_{max} and FRT were significantly greater than those in the older group (all $p < 0.01$). There was no significant difference in the ROM of the PF between the groups.

3.1. Force-tracking performance

In all the tasks, RMSEs were significantly greater in the older group than in the young group (Table 2). On the ANOVA (phase × task) analysis, in the young group, there was no significant interactive effect for the RMSE ($F_{(1.15, 8.12)} = 0.661$, $p = 0.463$, $\eta^2 = 0.086$), whereas significant main effects for phase ($F_{(1, 7)} = 14.212$, $p = 0.007$, $\eta^2 = 0.670$) and task ($F_{(2, 14)} = 54.007$, $p < 0.001$, $\eta^2 = 0.885$) were observed. The post hoc test showed that in the young group, RMSEs during the force generation phase were significantly greater than those during the force release phase across all tasks (all, $p < 0.01$); furthermore, RMSEs in the PR task were significantly greater than those in the LR (p

< 0.01) and HR (p = 0.001) tasks (Fig. 2A). In the older group, a significant interactive effect between phases and tasks for RMSE ($F_{(2, 18)} = 8.647$, $p = 0.002$, $\eta^2 = 0.490$) and a significant main effect for task ($F_{(1.12, 10.14)} = 8.162$, $p = 0.015$, $\eta^2 = 0.476$) were observed. There was no main effect for phase ($F_{(1, 9)} = 4.864$, $p = 0.055$, $\eta^2 = 0.351$) in the older group. Additionally, in the older group, the RMSEs during the force generation phase were significantly greater than those during the force release phase in the LR (p = 0.001) and HR (p = 0.017) tasks; however, no significant difference was identified between the force generation and release phases in the PR task (p = 0.407) (Fig. 2B). Furthermore, the post hoc test showed that only in the force release phase, the RMSE in the PR task was significantly greater than that in the LR (p = 0.041) and HR (p = 0.032) tasks (Fig. 2B).

Task	Force phase	Young group	Older group	T value	p-value
Low-range task	Generation	0.84 ± 0.31	2.12 ± 0.34	8.149	<0.001
	Release	0.45 ± 0.17	1.09 ± 0.45	3.788	0.002
High-range task	Generation	1.10 ± 0.27	2.54 ± 1.46	2.746	0.014
	Release	0.49 ± 0.23	1.20 ± 0.64	3.268	0.007
Pseudo-random task	Generation	1.83 ± 0.66	3.30 ± 1.84	2.349	0.037
	Release	1.12 ± 0.54	3.70 ± 2.81	2.838	0.018

Table 2 Comparisons of RMSEs between the groups.

Values are presented as mean ± standard deviation.

(Figure 2)

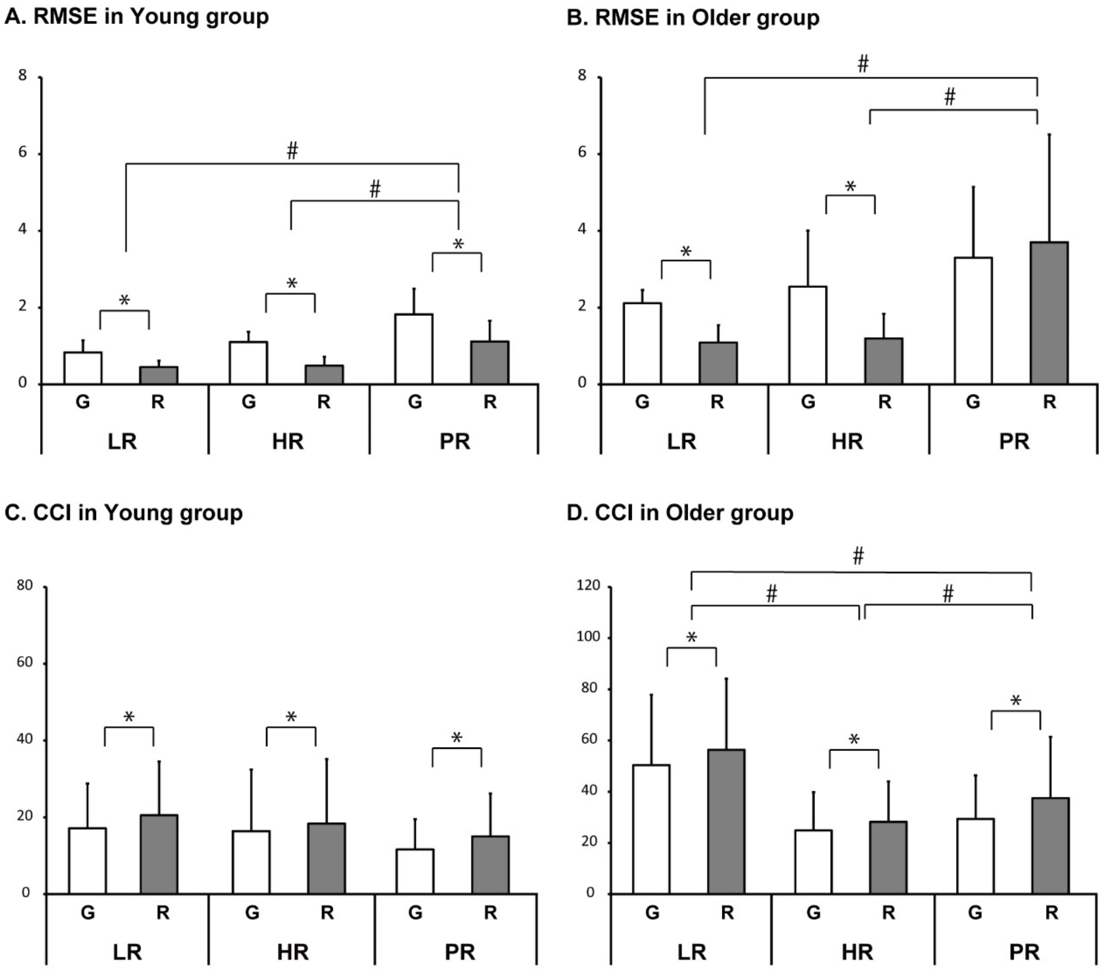

Fig. 2. Comparisons of the RMSE (A and B) and CCI (C and D) in each group. LR: Low range task;HR: High range task; PR: Pseudo-random task. G: Generation phase; R; Release phase. Asterisk (*) represents a significant difference between phases and hash (#) represents a significant difference between tasks.

3.2. CCI between the TA and GA

Except for the HR task, the CCI values in the LR and PR tasks were significantly greater in the older group than in the young group (Table 3). ANOVA for the young group indicated no significant interaction effect between phase and task for the CCIs ($F_{(2, 14)} = 3.621$, $p = 0.054$, $\eta^2 = 0.341$) or main effects for task ($F_{(1.12, 7.81)} = 0.632$, $p = 0.468$, $\eta^2 = 0.083$). There was a significant main effect for phase ($F_{(1, 7)} = 10.662$, $p = 0.014$, $\eta^2 = 0.604$). The post hoc test showed that the CCI during the force release phase was significantly greater than that during the force generation phase in the young group

(Fig. 2C). In the older group, there was no significant interaction effect between phase and task for the CCIs ($F_{(2, 18)} = 2.469$, $p = 0.113$, $\eta^2 = 0.215$), and a significant main effect for both task ($F_{(1.27, 11.45)} = 17.172$, $p = 0.001$, $\eta^2 = 0.656$) and phase ($F_{(1, 9)} = 23.657$, $p = 0.001$, $\eta^2 = 0.724$) in the older group. The post hoc test showed that in the older group, there were significant differences among all tasks, and CCIs were significantly greater in the order of LR, HR, and PR tasks (Fig. 2D). In addition, the CCIs during the force release phase were significantly greater than those during the force generation phase in all tasks.

Task	Force phase	Young group	Older group	T value	p-value
Low-range task	Generation	17.2 ± 11.7	50.4 ± 27.5	3.184	0.006
	Release	20.6 ± 14.0	56.4 ± 27.8	3.312	0.004
High-range task	Generation	16.4 ± 16.0	24.5 ± 14.5	1.155	0.265
	Release	18.4 ± 16.8	27.8 ± 15.2	1.284	0.217
Pseudo-random task	Generation	11.7 ± 7.9	29.4 ± 17.0	2.713	0.015
	Release	15.0 ± 11.2	37.5 ± 23.9	2.446	0.026

Table 3 Comparisons of CCIs between the groups

Values are presented as mean ± standard deviation.

3.3. Relationship between the FRT and the RMSE or CCI

Correlation analysis indicated a significantly negative correlation between the FRT and the RMSE (LR: $\rho = -0.583$ and $p = 0.011$, Fig. 3A; PR: $\rho = -0.531$ and $p = 0.023$, Fig. 3C, respectively) and CCI (LR: $\rho = -0.551$ and $p = 0.018$, Fig. 3D; PR: $\rho = -0.519$ and $p = 0.027$, Fig. 3F, respectively) in the LR and PR tasks. There was no significant correlation between the FRT ($\rho = -0.409$, $p = 0.092$,

Fig. 3B) and the RMSE or CCI (ρ = -0.346 and p = 0.159, Fig. 3E) in the HR task.

(Figure 3)

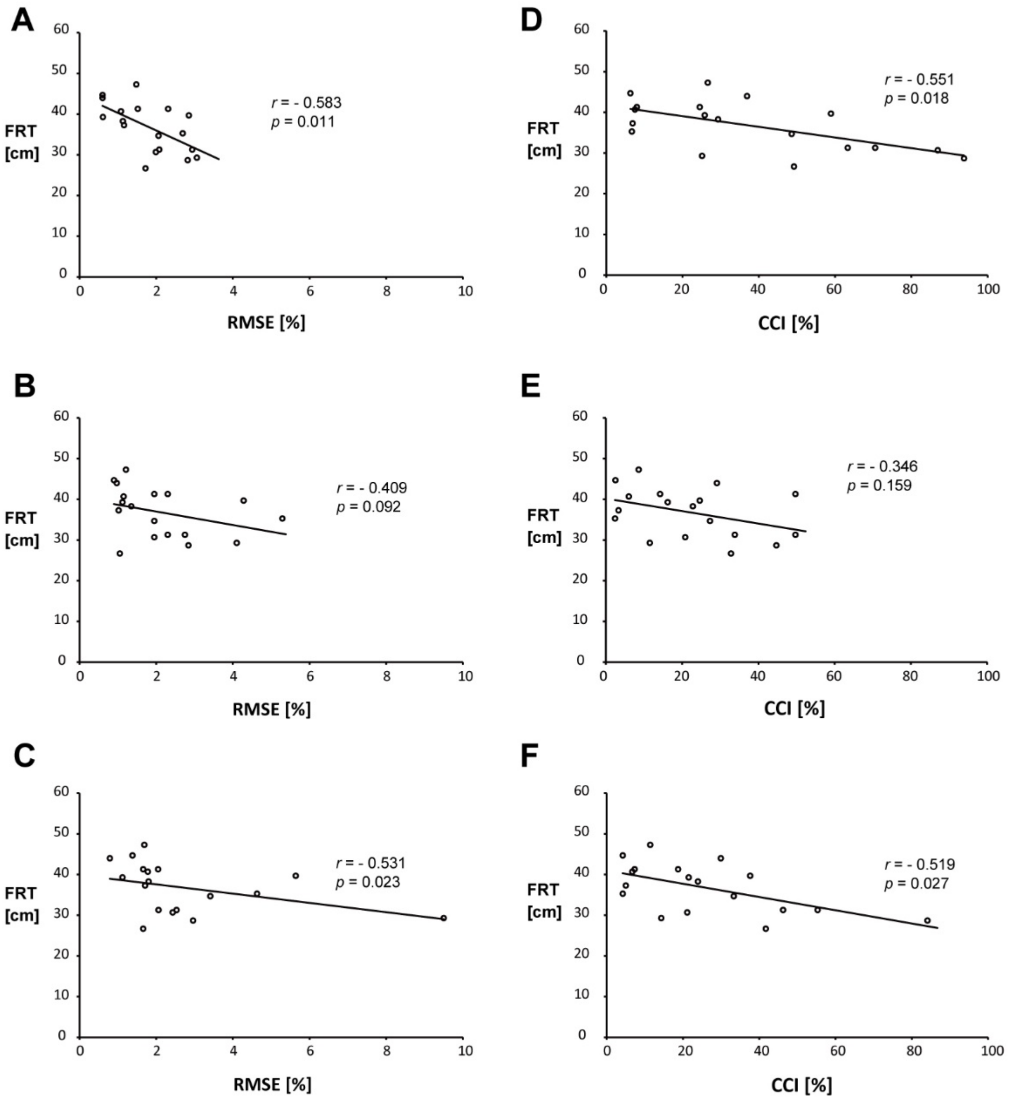

Fig. 3. Correlations between FRT and RMSE (A, B, C) or CCI (D, E, F) in each task. (A) and (D) represent simple liner regressions between the FRT and the RMSE and the CCI during the low range task, respectively. (B) and (E) represent simple liner regressions between the FRT and the RMSE and the CCI during the high range task, respectively. (C) and (F) represent simple liner regressions between the FRT and the RMSE and the CCI during the pseudo-random task, respectively.

4. Discussion

As expected, in the current study, an increase in motor error was identified across tasks in the older group compared to those in the young group, and our results are in accordance with previous findings with the pinch (Knol et al., 2019) or hand grip tests (Berger et al., 2020). Although poor visibility or impaired vision in older adults may influence the force accuracy in the force-matching task using visual feedback, it is confirmed that visual acuity is not related to the amplitude of force errors or lower performance because the participants were within the normal range of visual acuity (Hortobágyi et al., 2001; Kasahara and Saito, 2019). An interesting finding in this study was that the decline in force control in the random force-tracking task differed between the young and older groups from intragroup comparisons of RMSE. This finding may provide evidence that older adults are not able to relax their force well in unpredicted situations. Furthermore, the co-contraction between agonist and antagonist muscles in the LR and PR force tasks increased in the older group compared to that in the young group. Finally, this study indicated the relationship between balance and force control, particularly in the LR or PR force tasks under submaximal force.

4.1. Force control in predictable and unpredictable situations

Previous studies on force control have reported that force errors increased during force release (Lindberg et al., 2012; Ohtaka and Fujiwara, 2019, 2016; Park et al., 2016; Voelcker-Rehage and Alberts, 2005). Contrary to expectations, in this study, the RMSE of force tracking through plantar flexion was always greater during force generation than during force release in the HR and LR tasks (i.e., the constant task) in both groups. One of the possible explanations for this difference may be the different methods used, for example, the joint used. Most previous studies used the pinch, grip, or elbow flexion to assess the force control (Choi et al., 2019; Lindberg et al., 2012; Naik et al., 2011; Ohtaka and Fujiwara, 2016; Park et al., 2016; Voelcker-Rehage and Alberts, 2005). We assessed the force control using the ankle plantar flexion because we were interested in the postural and motor control during standing. Ohtaka (2016, 2019) examined the accuracy of force control with both upper and lower limbs at several force levels. Their results with the upper limb were consistent with those in previous studies; however, those with the lower limb were not. Conversely, their results with the lower limb were close to our results. Vaillancourt (2003) suggested that the difference in force control between the upper and lower limbs was based on the use and disuse of the neuromuscular system at different joints in daily activities. Therefore, the difference in the body part (or joint) used could explain why our results regarding force errors differed from those of previous studies.

Another difference in task parameters from previous studies (Knol et al., 2019; Lindberg et al., 2012; Naik et al., 2011; Ohtaka and Fujiwara, 2019, 2016; Park et al., 2016; Pedão et al., 2013; Vaillancourt and Newell, 2003; Vieluf et al., 2017; Voelcker-Rehage and Alberts, 2005) was the shape of the force target signal. In the ramp force tracking task (Lindberg et al., 2012; Naik et al., 2011; Ohtaka and Fujiwara, 2019, 2016) or step (Park et al., 2016), the force generation was started from approximately 0, while the force release was started from the maintained constant force level. These studies (Lindberg et al., 2012; Ohtaka and Fujiwara, 2019, 2016) have reported increased force error during the force release, while it is possible that the difficulty during the force release is not always consistent with that during the force generation. Since additional force control (i.e., force maintenance or braking) is required to avoid a rapid force release, the force release may be more challenging compared with the force generation (Knol et al., 2019; Lindberg et al., 2012; Pedão et al., 2013; Vaillancourt and Newell, 2003; Vieluf et al., 2017). In our study, we used the sinusoidal force tracking task, and our force generation started from 10%, 30%, and 50% of the maximum force, not from 0%. Therefore, our force tracking task required both force maintenance and modulation during the force generation, and it was speculated that the challenging level was the same between both phases and our force generation would be more challenging compared with force generations used in previous studies.

Finally, results of force errors in previous studies in healthy individuals may reflect the features or aspects of each methodology.

Compared with the other two tasks, the young group in this study had increased force errors during both generation and release phases in the random task, and the older group had increased force errors during the release phase. Traditional force control tasks required a stable force or holding and were randomly performed at several force amplitudes (Lauzière et al., 2012; Marchini et al., 2017; Yang et al., 2019). In addition, previous force-tracking tasks with a ramp (Choi et al., 2019; Park et al., 2016; Patel and Lodha, 2020; Spiegel et al., 1996) or sinusoidal (Berger et al., 2020; Knol et al., 2019; Perraton et al., 2017; Telianidis et al., 2014; Ward et al., 2019) waves comprised force repetition with the same constant amplitude and frequency. It was not necessary to predict the motion of the target force in force-maintaining tasks, and participants were able to predict the target position or motion easily through a pre-practice (i.e., familiarization) in force-tracking tasks. The random force task in the current study always changed the target force amplitude of the sinusoidal wave in every trial for our participants. Although the order in the random task in this study was pseudo-random, its order was an unexperienced event. Unpredictable behavior increases activation of cortical areas responsible for movement planning compared to predictable behavior (Dassonville et al., 1998; Legon and Staines, 2006), and more sustained attention to the unpredictable sensory input (i.e., the visual input) is necessary to properly execute the motor task (Legon and Staines, 2006). Therefore, it is necessary for participants to attend to the target motion frequently; the random task appears to be a challenging task with more loads regardless of age (Legon and Staines, 2006).

Our finding that the force accuracy during the random task declined in older adults compared with young adults is consistent with that of a previous study with an irregular sine wave pattern (Hübner et al., 2018). Furthermore, we found that the aspect of force errors differed between groups. In the young group, the RMSE during force generation was greater than that during force release across all tasks. In contrast, in the older group, the RMSE in the random task was equal between the force generation and release phases; in other words, the force error increased more during the force release. As mentioned above, the sensory input is essential to adjust and stabilize the motor control immediately during the unexpected situation (Dassonville et al., 1998; Legon and Staines, 2006). However, the pattern of the sensory input (e.g., tactile sense) differs between the force generation and release. In the current study, we speculate that the sensory input gradually increases during the force generation but gradually decreases during the force release. Barbosa et al. (2018) used some force levels in the ankle plantar flexion and suggested that the greater pressure in a higher force increases somatosensory information, culminating in greater force control, and concluded that older adults required a large amount of sensory input to maintain a good force control. Ohtaka and Fujiwara (2019) suggested that the decreased proprioceptive feedback of force during the force release was related to the increased force errors. We consider that the increased force errors during the force release in the random task in older adults may arise from the sensory input during the force release and decreased sensory system with aging, but a further study is required to investigate the age-related sensory system.

4.2. Co-contraction between agonists and antagonists during unpredictable situations

CCIs in the force-tracking task with the sine-wave pattern were consistently greater in the force release than in the force generation in both young and older groups; these findings support that the force control strategy differed between force generation and release (Lindberg et al., 2009; Ohtaka et al., 2016). The agonist muscles (i.e., GA) are required to maintain the baseline force and increase the force during the force generation phase, but the antagonist muscles (i.e., TA) are not always needed. In contrast, in the force release phase, the agonist muscles must reduce force gradually, and the co-contraction between the agonist and antagonist muscles plays a role in avoiding excessive reduction of force. Therefore, it is considered that the difference in the co-contraction between the force generation and release phases results from the task demand. Young adults in this study may employ co-contraction to decrease the variability of force control, particularly during the force release phase, compared with the force generation phase, and older adults attempted to use the same compensatory strategy.

Age-related changes in CCIs were observed in the LR and PR tasks. In general, the increased co-contraction between agonists and antagonists is a well-known feature of postural and motor control in older adults (Craig et al., 2016; Iwamoto et al., 2017; Krishnamoorthy et al., 2004; Piscitelli et al., 2017; Woollacott et al., 1988). The positive effects of co-contraction are considered to contribute to sensory information (Craig et al., 2016; Lauzière et al., 2012) and joint stabilization (Benjuya et al., 2004). The large amount of sensory information in older adults is important for maintaining good control of force production in the force-matching task (Abrahamová and Hlavacka, 2008), and increased CCIs have been interpreted as a compensatory strategy for age-related decline in sensory acuity, particularly proprioceptive acuity (Craig et al., 2016). In the case of tracking during the low-band force level in older adults, high-level co-contractions may be required to improve sensitivity and amplify small sensory information (Christou, 2011; Christou and Carlton, 2002; Enoka et al., 2003). Moreover, older adults may increase joint stiffness and improve joint instability by increasing the co-contraction level to minimize joint instabilities compared with young adults (Benjuya et al., 2004; Enoka et al., 2003; Krishnamoorthy et al., 2004; Lauzière et al., 2012; Woollacott et al., 1988). Therefore, our findings suggest that older adults control the small force in the LR task using the co-contraction strategy (Choi et al., 2019; Craig et al., 2016; Voelcker-Rehage et al., 2005). In contrast to the lower force task, it seemed that greater CCIs would not be necessary because muscle contractions in the high force task were kept higher, and thus, CCIs in both groups were equal.

Our results corroborate previous evidence that co-contraction in older adults (Krishnamoorthy et al., 2004; Woollacott et al., 1988) and individuals with neurological disorders

(Pradhan et al., 2010) increased during unpredicted perturbations. Activation of cortical areas responsible for movement planning increases during an unpredictable behavior compared to a predictable one (Dassonville et al., 1998; Schubotz and von Cramon, 2002), and there is evidence that the age-associated differences in the coordination between the agonist and antagonist muscles reflect an altered motor plan (Casamento-Moran et al., 2017). The unpredictable (i.e., random) task requires both the execution of the motor task and the continuance of attention at equal and more than that during the prediction of the stimulus, and the feedback information from sensory input is important to guide the tracking task successfully during unpredictable situations (Abrahamová and Hlavacka, 2008; Legon and Staines, 2006). Older adults may compensate for age-related deficits in the sensorimotor system by increasing muscle contraction (Christou, 2011; Christou and Carlton, 2002; Enoka et al., 2003; Lauzière et al., 2012); therefore, increased CCIs in the random task in older adults may be the age-related change in the force control strategy. The deficit of force control in older adults is considered to be attributable to the age-related decline in the central and peripheral sensorimotor system or information processing in the central nervous system (Hübner et al., 2018). We provide new evidence that the force strategy in older adults is influenced by the ability to predict.

4.3. Relationship between balance and force control during unpredictable situations

The FRT was shown to be an effective assessment for detecting fall risk in older adults and patients, and this study used the FRT as the balance ability during standing posture (Robinovitch and Cronin, 1999). Recent studies have shown that maximum muscle strength was not associated with force control (Perraton et al., 2017) and that the variability of the force generation was not related to postural sway (Barbosa et al., 2018). Our findings indicated that the RMSE and CCI in the low-band and random force-tracking tasks were associated with the FRT, suggesting that motor performance during these tasks may become a clinical marker for motor control in older adults or patients with motor disorders.

4.4. Limitations

The present study has several limitations. The sample size of this study was relatively small and comprised only male patients, although we recruited patients regardless of sex. Nonetheless, our results are partially consistent with those of previous studies, although they require considerable attention before generalizing to the entire older generation. There are several complex factors that cause falls in older adults (i.e., biological, behavioral, environmental, and socioeconomic factors). Similar to motor function, the sensory and cognitive functions are important for force control in older adults; however, the sensation (e.g., somatic sensation and equilibrium sense) and attention tests (e.g., Trail Making Test and Stroop Test) were not performed. Considering the interactions and confounders among these factors, further studies are required to determine the relationship between motor control and perceptual acuity in the sensory or cognitive-motor systems in older adults.

5. Conclusion

The force control was lower in the older group than in the young group for all force tracking tasks, and the CCI in the older group was greater during the LR and PR tasks. Moreover, the magnitude of the age difference was greater in the release phase of the PR task than in the other tasks. Overall, our findings provide important information on the features of motor control among older adults with regard to fall risk in unpredictable situations. Force control and co-contraction during the random task may be useful adjuncts to conventional measurements for motor control, providing greater insight into age-related changes in the neuromuscular system in older adults. Further studies should investigate the relationship between motor control and sensation or cognitive function in older adults.

Authors Statement

Ebisu Shunsuke and Kasahara Satoshi contributed to the study concept, recruited subjects, collected data, performed data analysis, and prepared the initial manuscript. Dr. Ishida Tomoya and Saito Hiroshi contributed to the study concept, provided editorial support, and critically reviewed and revised the manuscript.

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Figure legends

Fig. 1. A schematic representation of the experimental setup (A) and Target force trajectories (B-D). TA: tibialis anterior; GA: medial head of gastrocnemius.

Fig. 2. Comparisons of the RMSE (A and B) and CCI (C and D) in each group. LR: Low range task; HR: High range task; PR: Pseudo-random task. G: Generation phase; R; Release phase. Asterisk (*) represents a significant difference between phases and hash (#) represents a significant difference between tasks.

Fig. 3. Correlations between FRT and RMSE (A, B, C) or CCI (D, E, F) in each task. (A) and (D) represent simple liner regressions between the FRT and the RMSE and the CCI during the low range task, respectively. (B) and (E) represent simple liner regressions between the FRT and the RMSE and the CCI during the high range task, respectively. (C) and (F) represent simple liner regressions between the FRT and the RMSE and the CCI during the pseudo-random task, respectively.