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Effect of Vertical Force Sensation on Fingertips by Grasping a Balloon on Muscle Activity

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***Abstract*— Postural sway and muscle co–contraction are known to increase during upright standing with aging. A previous study demonstrated that grasping a helium–filled balloon stabilized postural. In this study, we measured the center of pressure (COP) and electromyography (EMG) of the trunk and ankle muscles in 17 healthy young adults while they grasped a balloon during standing and analyzed sway and muscle activity characteristics. The results indicated that the sway–reducing effect was observed not only with upward force sensation but also with downward force sensation. These findings suggest that sensory feedback processed at the spinal cord level primarily modulates trunk muscle activity, enhancing postural control and thereby improving stability.**

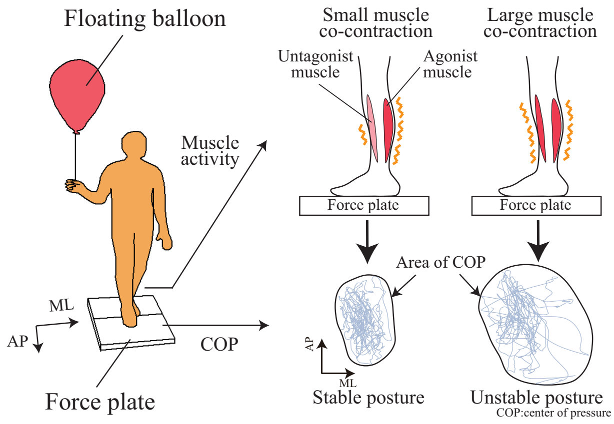

Fig. 1. Study concept

I. INTRODUCTION

Falls among the elderly have become a major concern, with survey results indicating that 28–35% of individuals aged 65 years or older experience at least one fall annually [1]. Approximately 10% of these falls result in serious injuries, such as fractures, making falls the third leading cause of long–term care needs in Japan [2]. These findings highlight the urgent need to provide effective strategies to support postural control.

One promising approach involves enhancing balance control by applying sensory feedback to the fingertips. Shimatani et al. reported that body sway was reduced when participants grasped a helium–filled balloon [3]. This effect was attributed to enhanced position sense arising from sensory feedback at the fingertips; however, the mechanism behind this phenomenon has yet to be discussed.

A related phenomenon, known as light touch contact, has also been shown to improve stability. This effect occurs when a person lightly touches a fixed object with a force of less than 1 N, thereby eliciting somatosensory feedback and reducing postural sway [4]. Prior studies on light touch have demonstrated that it decreases the simultaneous activation of multiple ankle muscles [5]. This reduces joint stiffness and decreases reliance on a feedforward co–contraction strategy, leading to more effective postural control. Excessive dependence on this strategy can impair responses to unexpected perturbations [6] and lead to unnecessary fatigue in situations requiring continuous fine postural adjustments. Thus, light touch is considered to suppress excessive feedforward control and promote more efficient postural regulation. It is reasonable to speculate that fingertip feedback from grasping a balloon may similarly reduce reliance on feedforward control.

Furthermore, the stabilizing effect of a light touch has been observed across various directions of contact. Prior research has demonstrated that this stabilization is achieved both when touching a horizontal surface (e.g., a table), generating an upward force on the fingertip, and when touching a vertical surface (e.g., a hanging piece of paper), generating a horizontal one [4] [7]. This suggests that the stabilizing effect of balloon grasping may not be limited to upward force sensations, but could also arise from force sensations in other directions.

Based on this perspective, the present study aimed to elucidate the mechanisms underlying the stabilization effect of balloon grasping by analyzing electromyographic (EMG) activity. Furthermore, to determine whether this effect is dependent on the direction of the force, postural sway and EMG were compared between conditions of grasping a helium–filled balloon (generating an upward force) and an air–filled balloon (generating a downward force).

II. METHODS

A. Experiment

17 healthy young men participated in the experiment (mean age: 22.8 ± 1.4 years). Two force plates (TF–3040, Tech Giken; sampling frequency: 1 kHz) and an 8–channel EMG system (pico: 4 ch, miniwave: 4 ch; Cometa; sampling frequency: 2 kHz) were used for data collection.

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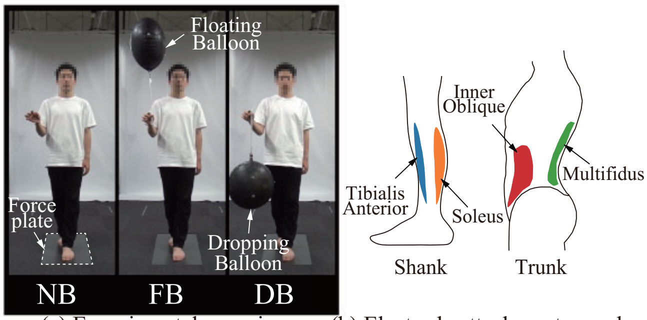


(a) Experimental overview (b) Electrode attachment muscles

Fig. 2. Experimental view and data processing

The experimental setup is illustrated in Fig. 2. Participants were instructed to stand in a tandem stance–defined as placing the first toe of the dominant foot against the heel of the non–dominant foot–on two force plates positioned side by side, with no gap between them. All trials were performed with the participants' eyes closed. Three experimental conditions were tested: (1) standing without a balloon (No Balloon: NB), (2) grasping a helium–filled balloon that generated an upward force of 0.05 N (Floating Balloon: FB), and (3) grasping an air–filled balloon weighted to generate a downward force of the same magnitude (0.05 N) (Dropping Balloon: DB). The force of the floating balloon was adjusted to balance a 5 g weight, while that of the dropping balloon was set to match the force of the floating balloon. Participants were instructed to grasp the balloon using their dominant hand, holding it between the thumb and index finger. The grasping arm was kept in a comfortable position, and the same posture was maintained in the NB condition. The contralateral arm was allowed to hang naturally alongside the body. Based on a previous study reporting that directing attention to the point of contact enhances stabilization [9], participants were instructed to stand as still as possible and to focus on the sensation at their fingertips. Each trial lasted 30 seconds, with the initial 5 seconds excluded from the analysis. Participants completed all three conditions (NB, FB, and DB) in a randomized order, followed by a 3–minute rest period. This entire sequence constituted one set. The first 11 participants performed six such sets and the remaining six performed eight sets. The first set was treated as a practice session and excluded from the analysis.

B. *Analysis method*

Postural stability was evaluated using two primary measures: the center of pressure (COP) and surface electromyography (sEMG) signals recorded from eight muscles. COP–based sway assessment has been widely applied in previous studies and is regarded as an effective method for evaluating postural stability and balance ability. Similarly, analyses of EMG signals have been employed to assess standing function [6]. The tandem stance, due to its narrow base of support, induces postural sway primarily in the frontal plane. In this posture, the tibialis anterior (TA) contributes to frontal–plane balance control, while the soleus (SO) plays a key role in maintaining upright posture [10]. Additionally, younger adults have been reported to exhibit greater co–contraction of trunk muscles such as the multifidus (MF) and internal oblique (IO) compared with older adults [8]. Based on these considerations, electrodes were attached bilaterally to the ankle muscles (SO and TA) and trunk muscles (IO and MF). Muscle activity during each task was recorded, and inter–task differences in EMG characteristics were analyzed.

1) COP analysis: COP displacement measured by the force plate was preprocessed with a 20 Hz low–pass filter. Following this, the 95% prediction ellipse area was calculated. The 95% prediction ellipse area represents the region containing 95% of the COP distribution, under the assumption that the COP scatter follows a chi–squared distribution [11]. This index is widely used as a measure of postural stability and was calculated as:

$$95\% \text{ PRED Area} = \pi ab, \tag{1}$$

where a and b denote the semimajor and semiminor axes of the prediction ellipse, that were estimated according to Eq.(2):

$$a = \sqrt{\chi^2_{2,0.95}\,\lambda_1}, \quad b = \sqrt{\chi^2_{2,0.95}\,\lambda_2}. \tag{2}$$

In Eq.(2), λ_1 and λ_2 represent the eigenvalues of the COP covariance matrix, and $\chi^2_{2,0.95}$ is the critical value of the chi–squared distribution with two degrees of freedom at a significance level of 0.95. Larger values of this index indicate greater postural sway.

2) EMG analysis: The EMG data were preprocessed using a band–pass filter with a frequency range of 10–500 Hz and rectified using a moving root mean square algorithm, employing a 0.2 second window. The signals were then normalized to the peak EMG amplitude obtained during a maximum voluntary contraction (MVC) task, that was performed after the completion of all balance trials. From the processed data, the mean muscle activity, coefficient of variation (CV), and co–contraction index (CI) were calculated. Intermuscular coherence was computed from the raw EMG signals.

Mean muscle activity, calculated using Eq.(3), reflects the magnitude of muscle output required to maintain posture, while the CV, calculated using Eq.(4), represents the temporal consistency of this activity, serving as an index of stability. Thus, higher mean muscle activity suggests greater muscular effort to maintain posture, whereas a higher CV indicates reduced stability in postural control.

$$\text{Mean Muscle Activity} = \frac{1}{T}\int_0^T \text{EMG}(t)dt, \tag{3}$$

$$CV = \frac{\sigma}{\text{Mean Muscle Activity}} \times 100\%, \tag{4}$$

where, σ represents the standard deviation.

The CI was calculated from the activity of antagonistic muscle pairs as an index of co–contraction, evaluating the extent to that the joint was fixed during the task [8].

$$CI = 2I_{ant}/I_{total} \times 100\%, \tag{5}$$

where I_{ant} represents the integrated activity of antagonistic muscles, calculated as:

$$I_{ant} = \int_{t_1}^{t_2} \mathrm{EMG}_1(t)dt + \int_{t_2}^{t_3} \mathrm{EMG}_2(t)dt, \tag{6}$$

where $t_1 - t_2$ denotes the period when EMG_1 is smaller than EMG_2, and $t_2 - t_3$ represents the opposite case. I_{total} is the integral of the sum of EMG_1 and EMG_2 during task execution:

$$I_{total} = \int_{t_1}^{t_3} [\mathrm{EMG}_1(t) + \mathrm{EMG}_2(t)]dt. \tag{7}$$

In this study, three muscle pairs were analyzed: TA and SO as an ankle pair; the dominant–side and non–dominant–side MF as a trunk pair; and the dominant–side and non–dominant–side IO as another trunk pair. Since the tandem stance is asymmetrical across the midline, muscle activation patterns differ between the dominant and non–dominant sides. Accordingly, CI values were derived separately for the dominant and non–dominant sides only for the ankle muscle pair TA–SO.

Intermuscular coherence between agonist and antagonist muscles was also computed to evaluate the neural pathways driving muscle activation [12]. The calculation is given by:

$$C(f) = \frac{S_{xy}(f)}{\sqrt{S_{xx}(f)S_{yy}(f)}}, \tag{8}$$

where $S_{xy}(f)$ is the cross–spectrum between two muscles, and $S_{xx}(f)$ and $S_{yy}(f)$ are their respective auto–spectra. Coherence values range from 0 to 1, with values closer to 1 indicating greater similarity in frequency components between the signals. The 8–14 Hz band is considered to reflect proprioceptive feedback from Ia afferents, while the 15–30 Hz band is thought to contain information related to cortical control [12]. Moreover, coherence in the 15–30 Hz and 40 Hz ranges has been reported to increase, accompanied by greater difficulty in balance tasks. Based on these findings, the frequency spectrum was divided into alpha (8–14 Hz), beta (14–30 Hz), and gamma (30–60 Hz) bands, and the mean coherence within each band was calculated for evaluation. Analyses were performed using the same muscle pairs as those used for CI.

C. Statistical analysis

For each index calculated from COP and EMG data, 25 seconds of measurement data (excluding the initial 5 seconds) were analyzed. Normality was assessed using the Shapiro–Wilk test. When normality was confirmed, comparisons among the three conditions were performed using one–way analysis of variance (ANOVA); when normality was not satisfied, the Kruskal–Wallis test was applied. To control for inflated false–positive rates due to multiple comparisons,

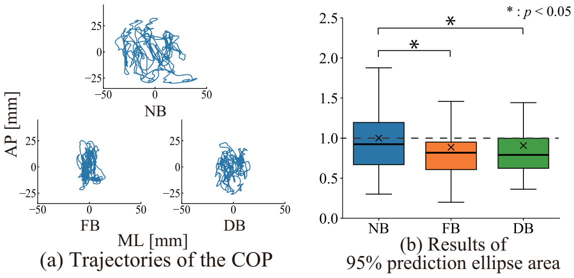


Fig. 3. Example and result of COP analysis

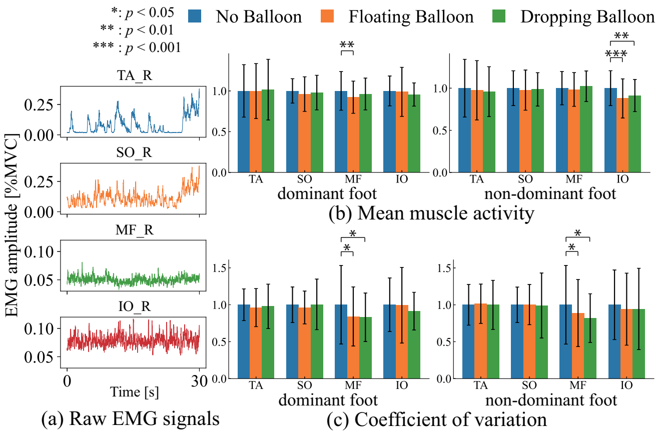


Fig. 4. EMG activation magnitude and variability

Holm's method was used, and the level of statistical significance was set at $p < 0.05$. In addition, one–sample t–tests were conducted to examine whether normalized indices under the NB condition deviated from the reference value of 1. When the assumption of normality was not met, the Wilcoxon signed–ank test was applied instead. The statistical analysis was performed using JASP (version 0.19.3, Netherlands). For all statistical analyses, confidence intervals were set at 95%.

III. RESULTS

A. COP analysis

Figure 3 (a) presents an example of COP trajectories under each condition. The trajectories indicate that sway in the frontal plane was smaller in the FB and DB conditions compared to the NB condition. To quantify sway, the 95% prediction ellipse area for each subject was normalized by the mean value of the NB condition, and the average across all trials was calculated (Fig. 3 (b)). The results demonstrated that sway was significantly reduced in both the FB and DB conditions relative to NB (both $p < 0.05$; Table 1). However, no significant difference was observed between FB and DB.

B. EMG analysis

Figure 4 (a) shows representative EMG signals recorded from the dominant foot during the task. The calculated mean muscle activity and coefficient of variation (CV) are presented in Fig. 4 (b) and (c). As shown in Fig. 4 (b), mean

TABLE I

SUMMARY OF COP AND EMG ANALYSIS RESULTS FOR THE THREE TASK CONDITIONS (NB, FB, DB).

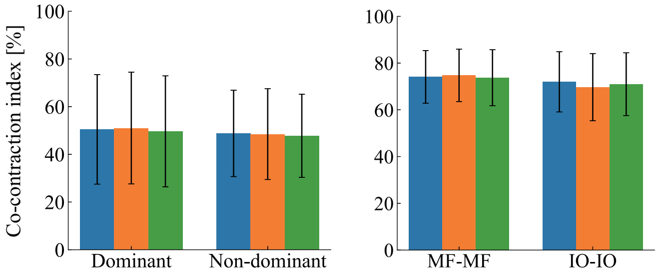
						ANOVA		Post-hoc/t-test		
Index	Side/Band	Target	NB	FB	DB	p	η^2	NB-FB	NB-DB	FB-DB
95% prediction ellipse area	–	–	0.93 [0.69–1.16]	0.80 [0.60–0.98]	0.80 [0.60–0.96]	**0.023**	0.019	**0.046**	**0.046**	0.929
Mean Muscle Activity	D	TA	0.97 [0.78–1.19]	0.91 [0.77–1.19]	0.96 [0.80–1.19]	0.793	0.000	1.000	1.000	1.000
		SO	1.01 [0.90–1.09]	0.97 [0.83–1.04]	1.00 [0.83–1.11]	0.146	0.006	0.150	0.556	0.556
		MF	0.96 [0.73–1.07]	0.89 [0.81–1.10]	0.94 [0.82–1.07]	**0.012**	0.024	**0.009**	0.190	0.196
		IO	0.96 [0.91–1.05]	0.92 [0.88–1.05]	0.96 [0.87–1.05]	0.285	0.002	0.340	0.793	0.793
	ND	TA	0.92 [0.80–1.14]	0.90 [0.79–1.16]	0.94 [0.82–1.12]	0.873	0.000	1.000	1.000	1.000
		SO	0.98 [0.84–1.09]	0.95 [0.81–1.12]	0.98 [0.84–1.09]	0.531	0.000	0.829	0.860	0.860
		MF	0.98 [0.89–1.09]	0.96 [0.86–1.04]	0.99 [0.90–1.12]	0.120	0.008	0.387	0.465	0.126
		IO	0.99 [0.88–1.09]	0.90 [0.73–0.99]	0.92 [0.79–1.02]	**<.001**	0.061	**<.001**	**0.006**	0.184
CV	D	TA	1.00 [0.85–1.13]	0.96 [0.78–1.12]	0.96 [0.82–1.11]	0.442	0.000	0.781	0.781	0.972
		SO	0.97 [0.85–1.09]	0.96 [0.80–1.07]	0.95 [0.80–1.10]	0.741	0.000	1.000	1.000	1.000
		MF	0.92 [0.74–1.08]	0.78 [0.59–0.98]	0.76 [0.61–0.98]	**0.006**	0.029	**0.015**	**0.015**	0.948
		IO	0.92 [0.79–1.06]	0.87 [0.73–1.06]	0.91 [0.75–1.02]	0.470	0.037	0.732	1.000	1.000
	ND	TA	0.96 [0.85–1.05]	1.00 [0.86–1.16]	0.99 [0.92–1.09]	0.585	0.000	0.958	0.958	0.958
		SO	0.96 [0.86–1.11]	0.98 [0.82–1.13]	0.95 [0.85–1.08]	0.714	0.000	1.000	1.000	1.000
		MF	0.89 [0.73–1.07]	0.79 [0.65–0.95]	0.80 [0.63–0.94]	**0.019**	0.021	**0.038**	**0.038**	0.914
		IO	0.91 [0.76–1.11]	0.88 [0.68–1.04]	0.85 [0.69–0.99]	0.136	0.007	0.240	0.187	0.758
CI	D	TA–SO	53.6 [25.3–69.6]	53.0 [24.1–71.1]	49.5 [24.1–68.5]	0.341	0.001	0.747	0.747	0.436
	ND	TA–SO	50.7 [31.6–61.7]	52.7 [30.5–61.7]	53.3 [32.7–60.1]	0.716	0.000	1.000	1.000	1.000
	–	MF	73.6 [65.8–83.1]	73.3 [66.1–84.7]	71.5 [65.5–83.7]	0.461	0.000	0.994	0.994	0.643
	–	IO	75.7 [62.3–83.4]	71.0 [56.8–83.7]	72.3 [64.4–83.7]	0.461	0.000	0.994	0.994	0.643
Intermuscular Coherence	Alpha 8–14 Hz	D TA–SO	1.00 [Ref]	0.84 [0.68–1.12]	0.86 [0.69–1.06]	–	–	**0.019**	**0.008**	0.738
		ND TA–SO	1.00 [Ref]	0.94 [0.76–1.23]	0.96 [0.76–1.21]	–	–	0.335	0.461	0.744
		MF	1.00 [Ref]	1.07 [0.90–1.37]	0.99 [0.77–1.19]	–	–	**0.008**	0.461	**0.048**
		IO	1.00 [Ref]	0.94 [0.68–1.16]	0.97 [0.78–1.19]	–	–	0.063	0.432	0.169
	Beta 15–30 Hz	D TA–SO	1.00 [Ref]	0.85 [0.70–1.05]	0.92 [0.76–1.17]	–	–	**0.009**	0.181	0.259
		ND TA–SO	1.00 [Ref]	0.92 [0.76–1.11]	0.85 [0.73–1.08]	–	–	**0.021**	**0.017**	0.339
		MF	1.00 [Ref]	0.92 [0.77–1.14]	0.95 [0.80–1.13]	–	–	**0.046**	0.165	0.738
		IO	1.00 [Ref]	0.94 [0.73–1.14]	0.85 [0.69–1.05]	–	–	**0.050**	**<.001**	0.056
	Gamma 30–60 Hz	D TA–SO	1.00 [Ref]	0.97 [0.79–1.11]	0.86 [0.77–1.07]	–	–	0.135	**0.011**	0.088
		ND TA–SO	1.00 [Ref]	0.92 [0.76–1.13]	0.81 [0.69–1.07]	–	–	0.062	**0.001**	**0.046**
		MF	1.00 [Ref]	0.96 [0.76–1.05]	0.94 [0.82–1.08]	–	–	**0.002**	**0.004**	0.516
		IO	1.00 [Ref]	0.86 [0.78–1.05]	0.85 [0.72–1.00]	–	–	**<.001**	**<.001**	0.158

Notes: Values are median [IQR]. D: Dominant side, ND: Non-dominant side. All indices except CI were normalized to each participant's NB condition. Ref: reference value. Post-hoc p-values are adjusted using Holm's method. **Bold values** indicate statistical significance ($p < 0.05$).

muscle activity was significantly reduced in the dominant–side MF between NB and FB, and in the non–dominant–side IO between NB–FB and NB–DB (MF: $p < 0.01$; IO: NB–FB $p < 0.001$, NB–DB $p < 0.01$; Table 1). Figure 4 (c) demonstrates that CV was significantly reduced in MF on both sides compared to NB (dominant–side MF: NB–FB $p < 0.05$, NB–DB $p < 0.05$; non–dominant–side MF: NB–FB $p < 0.05$, NB–DB $p < 0.05$; Table 1).

The results of the CI for both trunk and ankle muscles are presented in Fig. 5. A slight tendency toward reduced CI was observed in both trunk muscle pairs, although no significant differences were confirmed (Table 1). Conversely, for the ankle muscles (TA–SO), an increasing trend in CI was observed under the FB condition, although it was not statistically significant (Table 1).

Finally, based on the 95% prediction ellipse area, 13 participants who showed greater stabilization in FB compared to NB were classified as the FB–stabilized group, while 11 participants who stabilized in DB were classified as the DB–stabilized group. Intermuscular coherence was then calculated for both trunk and ankle muscles in each group, with the results shown in Fig. 6. For the ankle muscle pair, alpha–band coherence on the dominant side was significantly lower than baseline values (FB $p < 0.05$, DB $p < 0.01$; Table 1). Additionally, significant decreases were confirmed across most frequency bands above 15 Hz (dominant–side beta–band: FB $p < 0.01$; non–dominant–side beta–band: FB $p < 0.05$, DB $p < 0.05$; dominant–side gamma–band: DB $p < 0.05$; non–dominant–side gamma–band: DB $p < 0.01$, FB–DB $p < 0.05$; Table 1). In MF, a significant increase in alpha–band coherence was observed under the FB condition but not under DB ($p < 0.01$, Table 1). A direct comparison between FB and DB further confirmed a significant increase in alpha–band activity in FB ($p < 0.05$, Table 1). A similar comparison for IO revealed no significant differences. For the trunk muscles overall, as with the ankle muscles, significant decreases were observed in most frequency bands above 15 Hz (MF gamma–band: FB $p < 0.01$, DB $p < 0.01$; IO beta–



(b) Results of TA-SO (b) Results of trunk

Fig. 5. Result of CI

band: FB $p < 0.05$, DB $p < 0.001$; IO gamma–band: FB $p < 0.001$, DB $p < 0.001$; Table 1).

IV. DISCUSSION

The COP sway analysis demonstrated that the 95% prediction ellipse area was significantly reduced under both FB and DB conditions compared to the NB condition. This finding indicates that the sway reduction effect was elicited not only by upward force sensation but also by downward force sensation. Previous studies have suggested that light touch contact, in that the fingertips lightly touch a fixed point, provides somatosensory input that feeds back information on body tilt [13]. Other studies have proposed that the stabilizing effect of light touch is mediated by enhanced body awareness of spatial orientation relative to the environment [5]. It is interesting to notice that although both approaches provide somatosensory feedback to the fingertips, balloon grasping differs in that the balloon moves only in response to body sway, without direct contact with the external environment. Thus, we speculated that instead of enhancing body–environment awareness, balloon grasping may stabilize posture by providing sensory feedback specific to body sway.

To our best knowledge, this is the first time that the muscular activities during postural stability when grasping a balloon were studied. Although no significant changes in mean muscle activity or CV were observed in the ankle muscles, significant reductions were found in the trunk muscles (MF and IO). This suggests that the effects of balloon grasping may manifest primarily in the upper body rather than the lower body. Our research findings provided first–hand material and revealed that there may be a specific pattern of muscle activities to maintain body balance when grasping a balloon.

Muscle co–contraction did not differ significantly between conditions in either the trunk or ankle muscles. Whereas other fingertip–based somatosensory feedback techniques have been reported to reduce ankle co–contraction, balloon grasping did not produce this effect [5]. However, intermuscular coherence in the beta band—that reflects cortical input related to feedforward control—was reduced in the ankle muscles. This finding suggests that although the control strategy was modulated, a certain degree of ankle co–contraction was still required to maintain the challenging tandem stance, thereby masking changes in co–contraction.

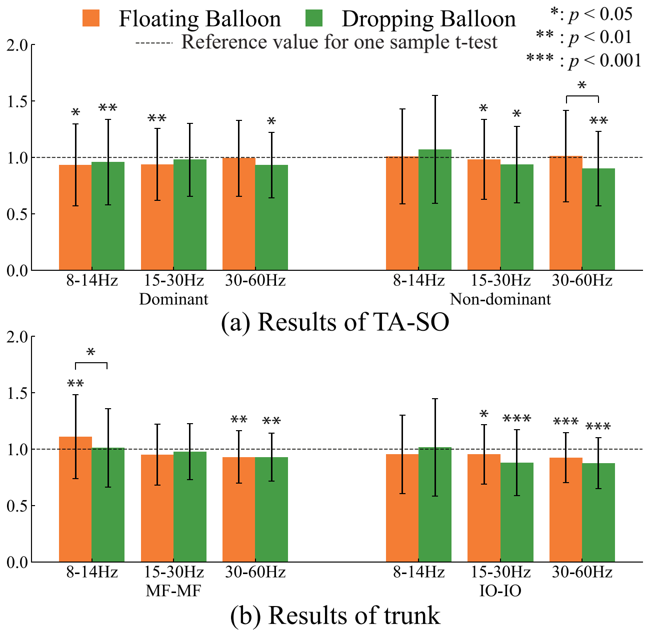


Fig. 6. Result of coherence

In terms of intermuscular coherence, a significant decrease in the alpha band was observed in the dominant–side ankle muscles. Previous studies have indicated that, during tandem stance, balance is primarily maintained using the rear leg [10]. The present results, however, suggest that feedback–mediated postural adjustments in the rear legs were diminished, indicating a shift toward relying on other body regions for postural control. For the trunk muscles, MF showed a significant increase in alpha–band coherence under FB, but not under DB. Direct comparison further revealed that FB elicited significantly greater alpha–band coherence than DB. This may be explained by an upward force sensation generating a backward shift of the upper body's center of gravity, thereby inducing an unconscious strategy that relied more on the back. Consistent with this, FB also produced significant decreases in mean muscle activity and CV in MF, suggesting that balloon grasping facilitated more precise balance control through feedback, leading to postural stabilization. If this interpretation is correct, downward balloon grasping would be expected to enhance abdominal feedback. However, while IO showed a slight increasing trend, no significant difference was observed. One possible explanation is that downward force sensation is more familiar in daily life compared to the less common sensation of upward force. As a result, the relatively small downward force applied in this study (0.05 N) may have been insufficient to alter control strategies.

This study has several limitations. The participants were a homogeneous group of young, healthy males, which restricts the generalizability of these findings. Moreover, the use of a challenging tandem stance with eyes closed may overestimate the intervention's stabilizing influence relative to more natural standing. Future validation under everyday postural conditions (eyes open, externally perturbed balance) is required to clarify practical applicability.

V. CONCLUSION

In this study, we focused on the effect of grasping a balloon on reducing postural sway while standing and investigated the changes in sway and muscle activity of the trunk and ankles when participants grasped a helium-filled balloon, exerting an upward force, and an air-filled balloon, exerting a downward force. The results confirmed that grasping the balloon reduced sway regardless of the direction of force. Although no effect on muscle co–contraction was observed, feedforward control of the ankle joint on the dominant leg decreased, while feedback of the trunk muscles increased. This suggests that grasping the balloon may enhance feedback in the trunk, contributing to postural stabilization. Future work will involve a more detailed investigation of how variations in the direction of force sensation influence the stabilizing effect and its reproducibility in more diverse populations (women, elderly individuals) and under more natural postural conditions (eyes open, balance perturbations).

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