Immediate Effect of Wearable Balance Training Device on Muscle Co-Contraction and Postural Control During Standing

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***Abstract*— Postural control is one of the primary body functions for fall prevention. Unexpected perturbation-based balance training is effective for improving postural control. However, the effect of perturbation-based training using assistive devices on muscle activity and co-contraction for standing balance is still unclear. This training is also difficult to perform easily because it requires large instruments or expert guidance. The purpose of this study is to demonstrate the effect of perturbation-based balance training using a wearable balance training device (WBTD) on postural control. In this study, fourteen healthy young adult males were assigned to either a WBTD group or a sham group. In the intervention session, participants in the WBTD group were perturbed either left or right direction at random timing by the WBTD during tandem stance balance training. Participants in the Sham group did not receive external perturbation during tandem stance balance training. Before and after the intervention session, participants of both groups underwent unexpected lateral perturbation postural control testing (pre- and post-test). The normalized integral of electromyography (IEMG), co-contraction index (CCI), and center of pressure (COP) parameters were measured in the pre- and post-test. Experimental results showed that the WBTD group in the post-test significantly decreased left Gluteus Medius IEMG, CCI of both Gluteus Medius, and peak COP_{ML} velocity, compared to those of the pre-test ($p < 0.001$, $p = 0.024$, $p = 0.031$, respectively). We conclude that balance training using WBTD could improve flexible postural control adjustment via cooperative muscle activation.**

***Index Terms*— Balance training, co-contraction, postural control, unexpected perturbation, wearable device.**

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I. Introduction

FALLS and falls related injury critically affect older individuals healthy life span and activities of daily living. With an increased aging population, falls and falls related injuries are increasing in the world. Previous study has reported that approximately 40% of individuals older than 65 years fall at least once per year [1]. Falls and falls related injuries are a critical risk of long-term admissions into nursing home care [2]. Improvement of postural control during standing is important for fall prevention. Postural control is one of the primary body functions for stabilizing the center of mass (COM) against external perturbation and preventing falls [3]. Postural control is a complex system, consisting of multiple components such as reactive postural control, dynamic stability, and motor system integration [4]. Postural control impairment correlates to falls and falls related injury in older individuals and individuals with neurological disorders [5], [6]. A previous study has reported that relative risk of postural control impairment for falls is 1.2 - 2.4 times higher in community-living older individuals [7]. Therefore, improvement of balance function by balance training is important for fall prevention. In particular, postural stability response to lateral perturbation is important for fall prevention [8]. Mediolateral direction of center of pressure (COP) sway is one of the predictive parameters for fall risk in older individuals [9]. In order to decrease mediolateral postural disturbance, the central nervous system coordinates bilateral muscle activity [10]. Gluteus Medius (GM) and External Oblique muscle activity increases for minimizing mediolateral sway when perturbation occurs during standing balance task [11]. These muscle activities are important for mediolateral stability during single-leg stance and gait [12], [13]. Hip adductor muscles also contribute to mediolateral stability during standing balance task as well as hip abductor and trunk muscles [14], [15].

Regarding postural control, there is also evidence that co-contraction between agonistic and antagonistic muscles affects postural control skill [16], [17]. Previous studies have reported older individuals increase lower limb muscle co-contraction compared to younger individuals during standing and gait [18], [19]. This higher muscle co-contraction is a compensatory strategy for older individuals to maintain stability by stiffening their joint [20]. However, increasing joint stiffness and more rigid body movement by excessive muscle co-contraction could lead to a higher risk of instability during postural perturbations [16]. Thus, reducing co-contraction during challenging balance task is an important factor for stability. Effects of balance training are also known to change balance function immediately as well as long term. This immediate effect is elicited due to changes in multisensory integration by the central nervous system. Immediate multisensory integration change occurs via central nervous system correction of postural control based on the error between predicted input and actual input [21]. Previous studies have reported that a balance task during standing and gait affected brain activity and stability [22], [23].

To increase standing balance ability against external perturbation, compensatory postural adjustments (CPAs) are made by the central nervous system. CPAs are used as corrective postural control functions after unexpected perturbations to recover the body and COP position [24], [25]. Postural control during a CPA phase reflects a person's stability of posture against unexpected external perturbation [11]. Unexpected perturbation-based balance training is one of the effective treatments for postural control. Compared to younger individuals, older individuals struggle to stabilize their posture during standing with external perturbation. Greater instability in response to unexpected perturbation results in increased COP displacement, velocity and muscle activity [26], [27], [28]. Compared to older individuals without experience of falls, older individuals with experience of falls demonstrate lower stability during unexpected perturbation balance tasks [29]. A previous study has reported that unexpected perturbation-based balance training improved postural control related to fall risk factors [30]. In another study, a balance training robot improved balance function and gait performance on older individuals with frail or prefrail [31]. Although perturbation-based balance training is effective for fall prevention and improving postural control, this training is difficult to perform because it requires expensive and large instruments or the guidance of a medical expert. Therefore, a perturbation-based training device that makes it easy for users to practice at their home is necessary.

Recently, wearable devices have been reported to contribute to user's health and improve rehabilitation [32], [33]. These devices could improve human movements, postural control, and daily activity. A wearable device for perturbation-based balance training has been developed which also allows users to practice this training at home [34]. This device is lightweight, flexible, and easy to use at home. It generates small, unexpected perturbation by pneumatic artificial muscles (PAM). Previous studies have reported that this device can be used for unexpected perturbation-based balance training, suggesting that it improves the static and reactive postural control immediately [34], [35]. However, the effect of perturbation-based training on muscle activity and muscle co-contraction for standing balance is still unclear. These indicators are important factors that contribute to postural control. If perturbation-based balance training using a wearable device could improve postural control and muscle co-contraction, the user may acquire more adjustable postural control at their home. This, in turn, could contribute to falls prevention. Although transition from a study in the lab-based setting to training at home requires a few steps including useability, to verify the effect of training on wearable devices is a critical initial step in the process. Therefore, the purpose of this study is to investigate effects of perturbation-based balance training using a wearable device designed to improve postural control function. We measured COP parameters and surface electromyography (EMG) of hip muscles for postural control during standing balance against unexpected external perturbation.

Fig. 1. Wearable balance training device (WBTD). Lateral PAMs were used for unexpected perturbation.

II. Methods

A. Participants

Fourteen healthy young adult males (age: 22.7 ± 0.9, height: 1.68 ± 0.04 m, weight: 59.1 ± 6.6 kg) participated in this study. Exclusion criteria were less than 20 years old, trunk or lower limb pain during standing or gait, having history of cardiac or neurological disorders, and surgery in the year before study participation. All procedures of this study were approved by the ethics committee of the Tokyo University of Science (21021). All participants were fully explained before the experiment, and the experiment was started after written informed consent.

B. Wearable Balance Training Device (WBTD)

A WBTD was used for unexpected perturbation-based balance training (Fig. 1). The WBTD consisted of four McKibben-type PAMs, solenoid valves (SYJ300, SMC, Tokyo, Japan), a soft shoulder supporter, a soft pelvic supporter, and CO_2 small tank (mini gas cylinder, NTG, Tokyo, Japan). This device was designed to be portable and

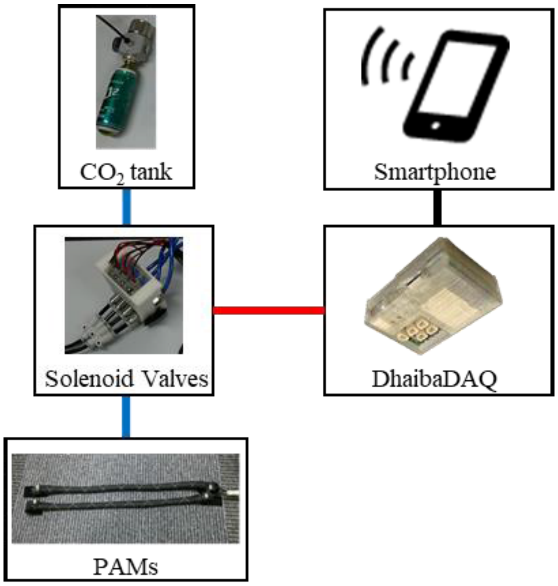

Fig. 2. Mechanism for PAM contraction. Wi-fi connection is indicated by the black line. Electronic circuit is indicated by the red line. Pneumatic flow for PAMs is indicated by the blue line.

lightweight to facilitate use at user's home and at clinical sites. The WBTD weighed 0.9 kg, in total. The natural length of the PAMs was 250 mm, and PAMs were extended to 270 mm. In this study, two PAMs were attached to left and right sides of the WBTD. The PAMs generated external force to induce lateral trunk bending. Solenoid valves for the PAMs were controlled by Dhaiba DAQ (wireless modules) [36]. Details of the WBTD have been described in a previous study [35]. The PAMs contract by compressed air flow from the CO_2 small tank when the solenoid valves are opened. From initial testing, it was shown that a PAM generated 40 N in response to air pressure condition of 0.2 MPa. Fig.2 shows contraction system configuration for the PAM in this device.

C. Experimental Setup and Procedure

The participants were randomly assigned to either the WBTD group or the Sham group. This randomization was achieved by computer-based random-number sequence generation. The experiment was conducted with a pre-test session, an intervention session, and a post-test session in the same day. Fig. 3 shows procedure of the experiment in this study. The participants were instructed to perform tandem stance with the dominant leg placed behind the non-dominant leg during a minute in each session. The leg that participants' preferred to use when they kick a ball was defined as the dominant leg [37]. In this study, the dominant leg of all participants was the right side. The participants were asked to keep their balance as consistent as possible for a minute and looked at a sign 2 m away, which was placed at their eyes level during the tandem stance test.

In the pre-test and post-test session, both group participants underwent an unexpected lateral perturbation postural control test on a force plate (Tech Gihan, Kyoto, Japan) during one-minute tandem stance, as in previous studies [35], [38]. Two air cylinders (CJ2E16-200AZ, SMC, Tokyo, Japan) were set on each side of the participants for the pre-test and post-test session. The height of the air cylinders was set at the height of iliac crests of each participant. During the tandem stance, participants were laterally perturbed at random timing and in random direction by the two air cylinders pushing the pelvis of the participants. Therefore, participants could not expect the perturbation timing and direction. This lateral perturbation force by the air cylinders was approximately 88 N. The one-minute tandem stance test was repeated three times in the pre-test and post-test sessions. The average number of perturbations was four times during one-minute of tandem stance. Although sometimes the perturbation for each side was experienced a different number of times during each one-minute tandem stance session due to randomization used to prevent prediction. Overall, the number of perturbations to each side was similar on average for the total tandem stance test in this study.

During the intervention session, participants of the WBTD group underwent unexpected perturbation in the mediolateral direction induced by the PAMs during tandem stance. In contrast, participants of the Sham group were not perturbed by the PAMs, but instead just wore the device during tandem stance. The participants of both groups performed 16 tandem stance tasks for one-minute each. After the intervention session, the post-test was performed in the same method as the pre-test. Rest time was set between the sessions for the participants. In addition, participants could sit and rest during each session at any time if they asked.

D. Data Analysis

During pre-test and post-test session, EMG of lower limb muscles and COP data from a force plate were measured. The air cylinders, EMG, and force plate were synchronized by electrical trigger. The EMG data were recorded by Delsys Trigno Research+ System (Delsys Inc., MA, USA) with a sampling frequency of 2000 Hz. The EMG sensors were attached bilaterally over the GM and Adductor Longus (AL) according to surface EMG for non-invasive assessment of muscles (SENIAM) recommendations [39]. These EMG data were filtered with a fourth-order bandpass Butterworth filter (20 – 450 Hz). Then, EMG data were full wave rectification and liner envelopes were created with 20 Hz cut-off frequency using a fourth-order low-pass Butterworth filter. Timing of the perturbations (t_0; time $= 0$) in the both test session was determined using the timing of electrical trigger that opens the solenoid valve of air cylinder and the time until the air cylinder contacted the participants' pelvis. From mechanical tests of this device, mean absolute error was approximately 10 ms. To evaluate postural control ability against unexpected perturbation, EMG data from 50 to 350 ms after t_0 were calculated as the time phase of interest, similar to a previous study [40]. This EMG phase was shifted 50 ms early compared with the COP time phase because of electromechanical delay [11], [41]. Typical patterns of EMG and COP during the postural control test in response to external perturbation are shown in Fig. 4. The integral of EMG signal (IEMG) was calculated. The IEMG of each muscle was normalized to compare IEMG between participants and groups. The maximum value of IEMG for each participant was chosen from a given muscle across all one-minute stance task in the

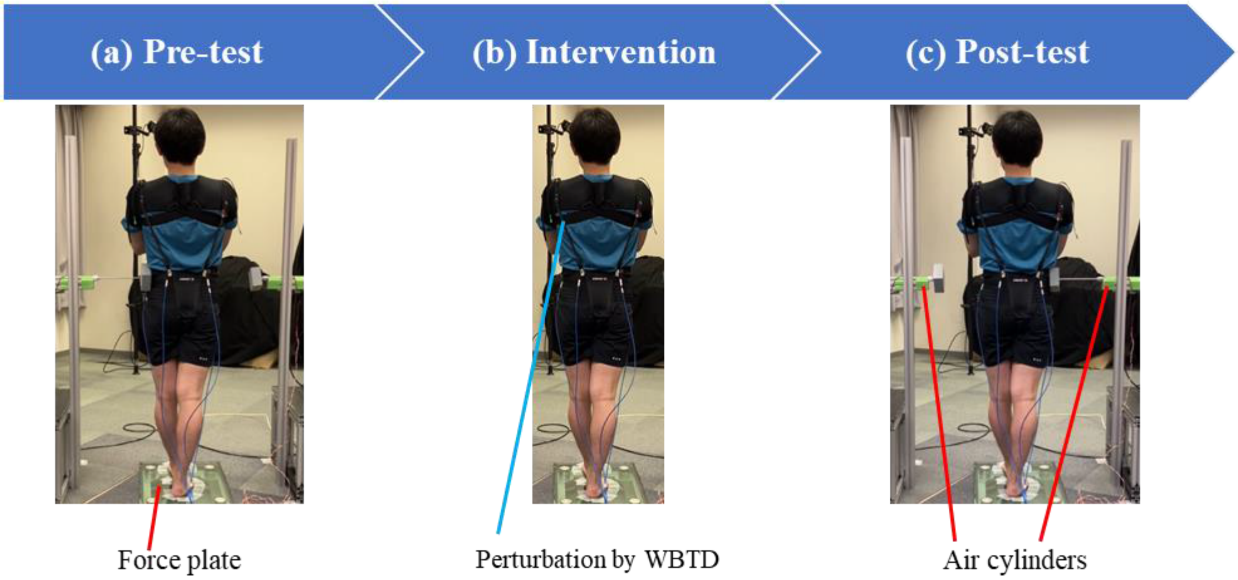

Fig. 3. Experimental procedure of this study. This experiment was conduct in following order: pre-test (a), intervention (b), and post-test (c). In (a) and (c) sessions, participants in both groups underwent unexpected lateral perturbation postural control test by air cylinders during tandem stance. In (b), the intervention session consisted of 16 tandem stance training trials. The Participants in the WBTD group were perturbed by a WBTD in intervention session, while the Sham group were not perturbed.

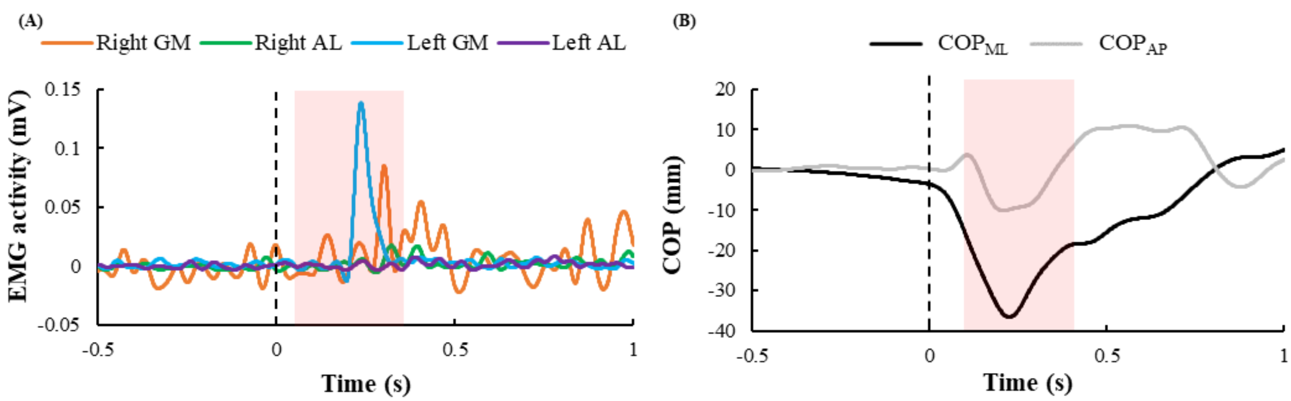

Fig. 4. Typical pattern of EMG and COP during the postural control test against external perturbation. Time 0 means the timing of the perturbation by the air cylinder. The red area indicates the CPA phase in EMG and COP. (A): EMG of both GM and AM, bilaterally, (B): COP_{ML} and COP_{AP} displacement.

pre-test and post-test sessions. After that, all IEMG values of each muscle of each participant were divided by this maximal value of IEMG [42]. Therefore, the range of normalized IEMG was limited from 0 to 1. In addition, the co-contraction index (CCI) of right GM and right AL (Right CCI), left GM and left AL (Left CCI), and right GM and left GM (GM CCI) was calculated to estimate hip joint or pelvis co-contraction in the frontal plane. For each participant, the maximal value of EMG amplitude for a given muscle activity across all pre-test and post-test sessions was determined. After that, all EMG data of the four muscles of each participant were divided by this maximal EMG amplitude [43]. These CCI were calculated by using following calculation [44]:

$$CCI = \frac{1}{N}\sum_{i=1}^{N}\left(\frac{EMG_{low_i}}{EMG_{high_i}}\right)\left(EMG_{low_i} + EMG_{high_i}\right) \quad (1)$$

where N is total number of datapoints for the EMG time phase of interest, EMG_{low} is the normalized lower EMG value of pair muscles EMG at ith data point. EMG_{high} is normalized higher EMG value of pair muscles EMG at ith data point. These CCI can range from 0 to 2. This CCI calculation takes similarity and magnitude of the EMG of muscle pair into account, therefore, it can be preferable for estimating joint stiffness [17]. If this CCI value in the post-test decreases compared with that of the pre-test, it is indicative of improvement in postural control adaptability against perturbation.

COP data from a force plate were recorded with a sampling frequency of 1000 Hz. Mediolateral and anteroposterior direction of COP (COP_{ML} and COP_{AP}) data were filtered with a 10 Hz cut-off frequency fourth-order low-pass Butterworth filter. Baseline values of these COP metrics were calculated using mean value from -500 to -350 ms (before balance perturbation, t_0). Then, these mean values were subtracted from COP_{ML} and COP_{AP} data, respectively. Peak displacement (D-COP), root mean square (RMS), and peak velocity (V-COP) of COP_{ML} and COP_{AP} were calculated using data from 100 to 400 ms after t_0, as in a previous study [40].

TABLE I
STATISTICS OF ANALYSIS OF VARIANCE FOR SPLIT-PLOT FACTORIAL DESIGN FOR IEMG, CCI, AND COP PARAMETERS

		Perturb direction	Effect of group	Effect of time	Interaction
IEMG	Right GM	Left-to-right	$F = 0.276, p = 0.610, \eta_p^2 = 0.022$	$F = 0.034, p = 0.857, \eta_p^2 = 0.003$	$F = 0.799, p = 0.389, \eta_p^2 = 0.062$
		Right-to-left	$F = 0.060, p = 0.811, \eta_p^2 = 0.005$	$F = 0.480, p = 0.501, \eta_p^2 = 0.039$	$F = 0.020, p = 0.889, \eta_p^2 = 0.002$
	Right AL	Left-to-right	$F = 0.315, p = 0.585, \eta_p^2 = 0.026$	$F = 0.450, p = 0.515, \eta_p^2 = 0.036$	$F = 0.195, p = 0.667, \eta_p^2 = 0.016$
		Right-to-left	$F = 0.772, p = 0.397, \eta_p^2 = 0.061$	$F = 0.081, p = 0.780, \eta_p^2 = 0.007$	$F = 1.201, p = 0.294, \eta_p^2 = 0.091$
	Left GM	Left-to-right	$F = 0.097, p = 0.761, \eta_p^2 = 0.008$	$\boldsymbol{F = 4.761, p = 0.049, \eta_p^2 = 0.284}$	$F = 2.744, p = 0.124, \eta_p^2 = 0.186$
		Right-to-left	$F = 0.488, p = 0.498, \eta_p^2 = 0.039$	$F = 0.245, p = 0.630, \eta_p^2 = 0.020$	$F = 3.303, p = 0.094, \eta_p^2 = 0.216$
	Left AL	Left-to-right	$F = 1.848, p = 0.199, \eta_p^2 = 0.134$	$F = 0.089, p = 0.770, \eta_p^2 = 0.007$	$F = 0.471, p = 0.506, \eta_p^2 = 0.038$
		Right-to-left	$F = 1.668, p = 0.221, \eta_p^2 = 0.122$	$F = 0.856, p = 0.373, \eta_p^2 = 0.067$	$F = 0.570, p = 0.465, \eta_p^2 = 0.045$
CCI	Right CCI	Left-to-right	$F = 1.108, p = 0.313, \eta_p^2 = 0.085$	$F = 0.149, p = 0.707, \eta_p^2 = 0.012$	$F = 2.093, p = 0.174, \eta_p^2 = 0.149$
		Right-to-left	$F = 1.733, p = 0.213, \eta_p^2 = 0.126$	$F = 1.504, p = 0.244, \eta_p^2 = 0.111$	$F = 0.802, p = 0.388, \eta_p^2 = 0.063$
	Left CCI	Left-to-right	$F = 0.001, p = 0.971, \eta_p^2 < 0.001$	$F = 3.982, p = 0.069, \eta_p^2 = 0.249$	$F = 0.025, p = 0.877, \eta_p^2 = 0.002$
		Right-to-left	$F = 0.118, p = 0.738, \eta_p^2 = 0.010$	$F = 0.008, p = 0.932, \eta_p^2 < 0.001$	$F = 0.685, p = 0.424, \eta_p^2 = 0.054$
	GM CCI	Left-to-right	$F = 0.336, p = 0.573, \eta_p^2 = 0.027$	$\boldsymbol{F = 6.331, p = 0.027, \eta_p^2 = 0.345}$	$F = 0.326, p = 0.579, \eta_p^2 = 0.026$
		Right-to-left	$F = 1.081, p = 0.319, \eta_p^2 = 0.083$	$F = 0.174, p = 0.684, \eta_p^2 = 0.014$	$F = 0.365, p = 0.557, \eta_p^2 = 0.030$
COP_{ML}	D-COP_{ML}	Left-to-right	$F = 0.373, p = 0.553, \eta_p^2 = 0.030$	$F = 0.001, p = 0.970, \eta_p^2 < 0.001$	$F = 1.568, p = 0.234, \eta_p^2 = 0.116$
		Right-to-left	$F = 0.120, p = 0.735, \eta_p^2 = 0.001$	$F = 1.271, p = 0.282, \eta_p^2 = 0.096$	$F = 1.155, p = 0.304, \eta_p^2 = 0.088$
	RMS_{ML}	Left-to-right	$F = 0.697, p = 0.420, \eta_p^2 = 0.055$	$F = 0.005, p = 0.946, \eta_p^2 < 0.001$	$F = 0.750, p = 0.404, \eta_p^2 = 0.059$
		Right-to-left	$F = 0.006, p = 0.939, \eta_p^2 < 0.001$	$F = 0.901, p = 0.361, \eta_p^2 = 0.070$	$F = 1.934, p = 0.190, \eta_p^2 = 0.139$
	V-COP_{ML}	Left-to-right	$F = 1.940, p = 0.189, \eta_p^2 = 0.139$	$\boldsymbol{F = 6.484, p = 0.026, \eta_p^2 = 0.351}$	$F = 2.368, p = 0.150, \eta_p^2 = 0.165$
		Right-to-left	$F = 0.372, p = 0.553, \eta_p^2 = 0.030$	$F = 2.852, p = 0.117, \eta_p^2 = 0.192$	$F = 0.562, p = 0.468, \eta_p^2 = 0.045$
COP_{AP}	D-COP_{AP}	Left-to-right	$F = 0.544, p = 0.475, \eta_p^2 = 0.043$	$F = 0.495, p = 0.495, \eta_p^2 = 0.040$	$F = 1.642, p = 0.224, \eta_p^2 = 0.120$
		Right-to-left	$F = 0.115, p = 0.741, \eta_p^2 = 0.001$	$F = 0.144, p = 0.711, \eta_p^2 = 0.012$	$F = 0.845, p = 0.376, \eta_p^2 = 0.066$
	RMS_{AP}	Left-to-right	$F = 0.013, p = 0.912, \eta_p^2 = 0.001$	$F = 0.057, p = 0.851, \eta_p^2 = 0.005$	$F = 0.003, p = 0.956, \eta_p^2 < 0.001$
		Right-to-left	$F = 1.361, p = 0.266, \eta_p^2 = 0.102$	$F = 0.472, p = 0.505, \eta_p^2 = 0.038$	$F < 0.001, p = 0.979, \eta_p^2 < 0.001$
	V-COP_{AP}	Left-to-right	$F < 0.001, p = 0.988, \eta_p^2 < 0.001$	$F = 0.573, p = 0.464, \eta_p^2 = 0.046$	$F = 1.613, p = 0.228, \eta_p^2 = 0.119$
		Right-to-left	$F = 4.062, p = 0.067, \eta_p^2 = 0.253$	$F = 0.165, p = 0.692, \eta_p^2 = 0.014$	$F = 0.431, p = 0.524, \eta_p^2 = 0.035$

Normalized integral of electromyography (IEMG) of Gluteus Medius (GM) and Adductor Longus (AL), normalized co-contraction index (CCI), and center of pressure (COP) parameters are shown in this table. The main effects of group (WBTD/Sham), time factors (pre-test/post-test), and interaction are shown. Bold indicates statistically significant difference (p < 0.05).

These COP parameters were used as a representative index for evaluating postural control. Peak D-COP_{ML} and V-COP_{ML} were quantified as the maximum value in the same external force direction by the air cylinder in this time phase, while peak D-COP_{AP} and V-COP_{AP} were quantified as the absolute maximum value. Since the unexpected perturbation during pre-test and post-test were bi-directional, these parameters were averaged separately for the left-to-right and right-to-left direction of perturbation.

E. Statistical Analysis

To compare these IEMG, CCI, and COP parameters in two perturbation direction conditions, the main effects of group (WBTD/Sham) and time (pre-test/post-test) were evaluated by analyzing the variance for a split-plot factorial design. Paired t-test and unpaired t-test were used as a post-hoc test to compare the time and group factor, respectively. These statistical analyses were performed using R (version 4.2.1; CRAN, freeware). The partial eta-squared (η^2) was calculated to investigate effect size for the variance for a split-plot factorial design, and η^2 was defined as small ($<$0.01), medium (0.01 – 0.06), and large ($>$0.06). The Effect size of post-hoc test was also calculated using r value, and r was defined as small ($<$0.1), medium (0.1 – 0.5), and large ($>$0.5). Statistical significance was set at $p < 0.05$.

III. RESULTS

Table I summarizes the results of variance for the split-plot factorial design for IEMG, CCI, and COP parameters. Normalized IEMG and normalized CCI for each group and time factor in both perturbation direction conditions are shown in Fig. 5.

In normalized IEMG parameters, there were significant differences for the main effect of time for left GM in the left-to-right perturbation condition. In the WBTD group, normalized IEMG of the left GM during the post-test significantly decreased compared to those of the pre-test in the left-to-right perturbation condition. In normalized CCI parameters, there were significant differences for the main effect of time for GM CCI in the left-to-right perturbation condition. In the WBTD group, normalized GM CCI during the post-test significantly decreased compared to those of the pre-test in the left-to-right perturbation condition. There were no statistically significant differences in the main effects of the group for IEMG and CCI parameters. In addition, there was no significant interaction effect.

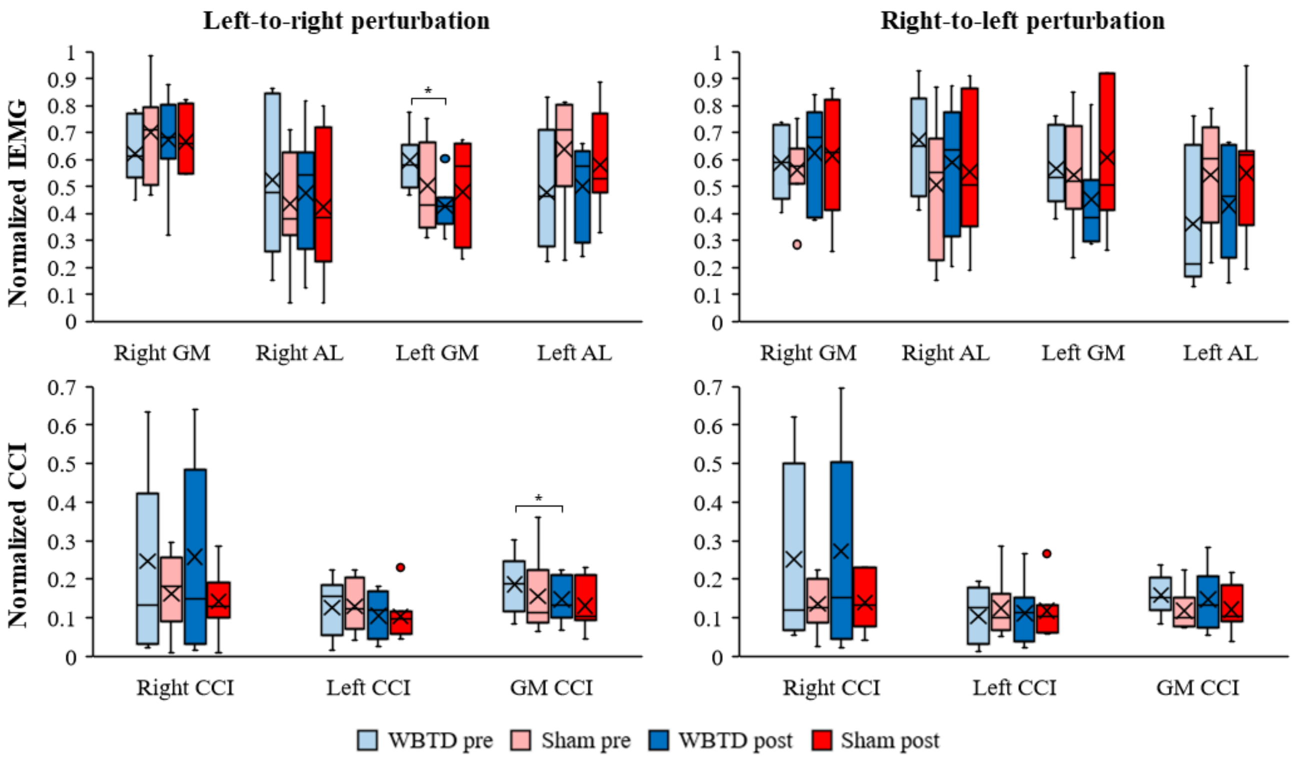

Fig. 5. Results of EMG parameters on each group and time factor in both perturbation direction conditions. The top row shows normalized integral of electromyography (IEMG) of Gluteus Medius (GM) and Adductor Longus (AL). The bottom row shows normalized co-contraction index (CCI). Boxes and inside horizontal lines of Box plots represent ranges of Q1, Q3, and median values. Upper and lower whiskers show the highest and lowest values excluding outliers. $\times$ and $\bigcirc$ indicate mean and outlier, respectively. $*$ indicates significant difference ($p < 0.05$).

The COP_{ML} and COP_{AP} for each group and time factor in both perturbation direction conditions are shown in Fig. 6. In COP parameters, there were significant differences for the main effect of time for V-COP_{ML} during the left-to-right perturbation condition. In the WBTD group, V-COP_{ML} during the post-test significantly decreased compared to those of the pre-test in the left-to-right perturbation condition. There were no statistically significant differences in the main effects of the group for these COP parameters. In addition, there was no significant interaction effect. There were no significant differences in normalized IEMG, normalized CCI, and COP parameters for the right-to-left perturbation condition.

IV. Discussion

Improving postural control during standing with unexpected external perturbation is critical for fall prevention. The purpose of this study was to demonstrate the effect of perturbation-based balance training using a WBTD on postural control function. This study measured normalized IEMG and CCI of hip muscles, and COP parameters during unexpected lateral perturbation postural control tests. Experimental results showed that perturbation-based balance training using a WBTD could improve corrective postural control compared to a Sham group as a main effect.

In the left-to-right perturbation condition, normalized IEMG of the left GM for the WBTD group decreased compared to those of the pre-test session. Muscle activity for corrective postural control on younger adults were evaluated by inducing unexpected perturbations and measuring EMG during the CPA phase [40], [41]. Previous studies have shown that older adults increased compensatory activation in their lower limb muscles during standing postural tasks, compared with younger adults [8]. During tandem stance, instability of the COP_{ML} was higher than in normal standing [45], and both GM activation increased, contributing to stabilization of postural control. Perturbation-based balance training using a WBTD might decrease excessive muscle activity during challenging balance tasks such as the tandem standing task used in this study.

Moreover, the results also showed that normalized GM CCI during the post-test for the WBTD group significantly decreased with a large effect size compared to those of the pre-test session. Muscle co-contraction is known to increase joint stiffness by central nervous system [46], [47]. Older adults elevate the level of lower limb muscles co-contraction during gait and standing balance [48], [49]. The CCI and muscle activity also increase when younger and older individuals maintain their postural control during balance challenging tasks [50], [51]. However, assessing increased joint stiffness, as indicated by the calculated CCI has not always been successful without accurate prediction of the amount of postural sway [51]. Elevation of the CCI level during standing postural control was associated with fall risk in older adults [18]. Therefore, perturbation-based balance training using a WBTD could decrease CCI and contribute to flexible postural control adjustment. In addition, it has been reported that GM muscle activity was increased when younger individuals were perturbed from lateral direction [52]. Balance training decreased CCI during standing balance task in older adults, which could be associated with improvement of postural control

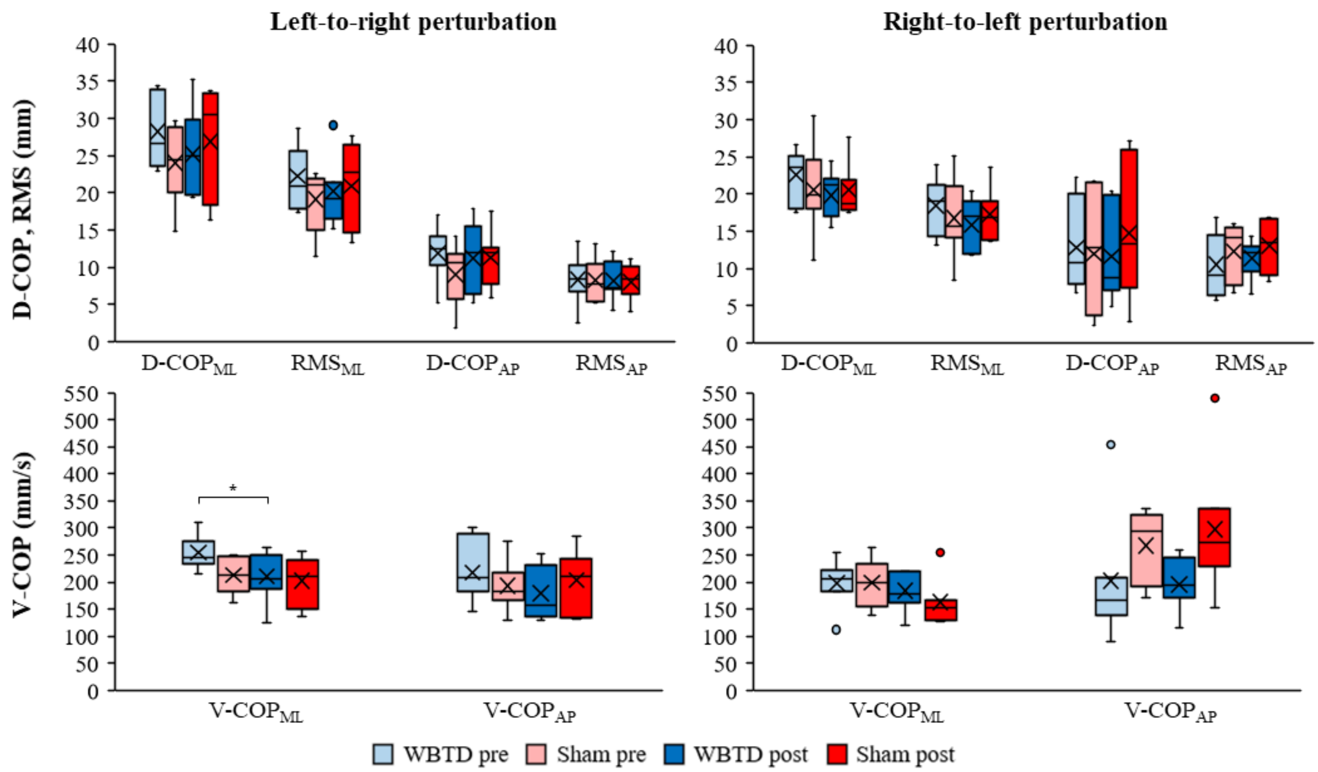

Fig. 6. Results of center of pressure (COP) parameters on each group and time factor in both perturbation direction conditions. The top row shows D-COP and RMS. The bottom row shows V-COP. Boxes and inside horizontal lines of Box plots represent ranges of Q1, Q3, and median values. Upper and lower lines of whiskers show the highest and lowest values excluding outliers. × and ○ indicate mean and outlier, respectively. * indicates significant difference ($p < 0.05$).

ability [53]. In this study, we observed decreased GM CCI during unexpected lateral perturbation postural control tests. This seems to indicate a decrease of excessive joint and inter-limb stiffness to allow flexible control by the GM on both sides. The results showing the effect of a WBTD are supported by and consistent with these previous studies.

For COP parameters, V-COP$_{ML}$ of the post-test session for the WBTD group significantly decreased compared to those of the pre-test session (with a large effect size). Previous studies have reported that COP peak displacement and peak velocity after unexpected perturbation was found in the CPA phase. The results of this study were consistent with these previous studies on younger and older adults [35], [54]. Postural control during stance with unexpected perturbation decreases with aging, disease, and the experience of falls [14], [29]. Central nervous system disorder such as Parkinson's disease and stroke affect postural control and muscle activity during stance and gait [5], [14]. A previous study reported that older individuals with and without experience of falls had COP instability during standing task with unexpected perturbation compared to younger adults [29]. Perturbation-based balance training using a training device or medical experts' guidance could improve postural control. Perturbation-based balance training using a ride-on robotics rehabilitation instrument improved static balance function and tandem gait velocity [55]. In addition, perturbation-based balance training using waist-pull system during gait improve gait stability and response to perturbation in individuals with Parkinson's disease [56]. The COP velocity is an indicator of instability during standing balance tasks, and older individuals with experience of falls increase COP velocity during tandem stance compared to older adults without experience of falls [57]. Therefore, perturbation-based training using a WBTD might improve postural control immediately. Decreased V-COP$_{ML}$ was seemingly caused by flexible postural control to lower GM CCI, likely reducing joint stiffness and thus allowing flexible and stabilizing postural control adjustment [16].

Differing from previous studies, the results of the present study did not show significant difference under the right-to-left perturbation condition. These results may be due to the type of standing posture used. Previous studies used perturbation balance training for younger and older adults required to maintain stance posture with their feet shoulder placed at width [28], [38], [58], while the participants of this study were required to be in tandem stance with their right-side leg behind for evaluating postural control during the balance challenge task. Compared to normal standing posture, it was difficult for the participants to maintain lateral postural stability during tandem stance. Moreover, tandem stance is asymmetric foot position. Therefore, we speculate that these significant differences were only found in the left-to-right perturbation condition because of the asymmetric foot position. Also, there were no significant differences on the results of D-COP for the WBTD group. The results differed from the previous study [28]. One possible reason for no significant difference in the D-COP could be that Perturbation force is less than other methods. Previous studies use more high force than our study because they perform it in laboratory environment, namely, these studies used large perturbation system and strong harness [23], [28]. In this study, the wearable device for perturbation-based balance training has been developed which also allows users to safely practice this training at home. The generated force is carefully controlled for safety. Therefore, the components of postural control such as EMG index could detect sensitive changes, but indicators such as D-COP might not detect immediate changes. Also, p values of interaction in this study were not less than 0.05, whereas it may be worth noting that some effect sizes of the interaction were large.

There are a few limitations in this study that should be addressed. First, the sample size of this study was not very large. Research with a large number of participants may provide more discoveries due to a more detailed analysis. Second, this study did not measure other lower limb muscles such as ankle joint muscles. Although hip muscles are considered to be the primary contributors to mediolateral stability during frontal plane balance challenge tasks, ankle muscles might also contribute to mediolateral stability. Lateral perturbation during stance has been reported to increase muscle activity the ankle dorsiflexor and plantar flexor muscles as well as the hip abductor and adductor muscles [58]. Verification of muscle activity and CCI in other joint and inter-limb pairings might yield important insights into postural control and the effect of perturbation-based balance training. Lastly, this study did not include older individuals or individuals with postural control impairment including male and female. To investigate the long-term effect of perturbation-based training using a WBTD for older individuals and individuals with postural control impairment could further contribute to fall prevention and effective treatment for postural control. Despite these limitations, our findings suggest that perturbation-based balance training using a WBTD could improve postural control ability against unexpected external perturbation.

V. Conclusion

The purpose of this study was to investigate the effect of perturbation-based training using a WBTD on postural control ability against unexpected external perturbation. A WBTD was developed to allow users to perform perturbation-based balance training at their home easily. This device is lightweight, flexible, and easy to use at the user's home without requiring large instruments or medical expertise. The results showed that perturbation-based balance training using a WBTD significantly decrease postural sway. Moreover, the training also decreased co-contraction and muscle activity in hip abductors. These findings suggest that a WBTD may improve postural control ability against unexpected external perturbation. Further study should investigate the effect of similar training in pathological participants and older individuals with experience of falls.

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