The Body's Compensatory Responses to Unpredictable Trip and Slip Perturbations Induced by a Programmable Split-Belt Treadmill

Beom-Chan Lee, *Member, IEEE*, Chul-Soo Kim, and Kap-Ho Seo

***Abstract*— This paper investigates the influence of two types of gait perturbation (i.e., trip and slip) induced by a programmable split-belt treadmill on the body's compensatory responses. Our fall-inducing technology equipped with a commercially available programmable split-belt treadmill provides unpredictable trip and slip perturbations during walking. Two force plates beneath the split-belt treadmill and a motion capture system quantify the body's kinetic and kinematic behaviors, and a wireless surface electromyography (EMG) system evaluates the lower limb muscle activity. Twenty healthy young adults participated. The perturbations (i.e., trip and slip) were applied randomly to the participant's left foot between the 31st and 40th steps. The kinetic and kinematic behaviors and lower limb muscle activity were assessed during the standing, walking, and recovery periods. Compared with trip perturbations, stepping responses to slip perturbations were quicker and trunk, shoulder, and whole body center of mass movements after slip perturbations were higher; the EMG results showed that tibialis anterior, gastrocnemius, rectus femoris, and biceps femoris activities were also higher. The two common types of gait perturbation (i.e., trip and slip) induced by a commercially available programmable split-belt treadmill influenced the body's compensatory responses.**

***Index Terms*— Gait perturbation, programmable split-belt treadmill, compensatory responses, kinetics and kinematics, electromyography.**

I. Introduction

FALLS and the resulting fractures and injuries [1], [2] are a significant health concern, particularly for older people and people with balance disorders [3]. Unexpected gait perturbations (i.e., trips and slips, the two most common gait perturbations) while walking account for approximately 60% of unintentional falls by young and older adults [1], [3]. It has been shown that compensatory motor learning and adaptation principles aided by fall-inducing technologies can facilitate the body's responses to unexpected perturbations and falls (see [4] for review). A recent review has indicated that reactive compensatory stepping (stepping is the normal response to an unexpected gait perturbation [5], [6]) by applying gait perturbations increases the task specificity [4] compared to the strength and balance exercises that are generally prescribed.

To date, many fall-inducing technologies incorporate external mechanisms including obstacles (e.g., [7], [8]), cables (e.g., [9], [10]), and slippery agents and contaminants (e.g., [11], [12]) on the floor, or low-friction movable platforms (e.g., [13], [14]) to induce unexpected gait perturbations. For example, mechanical obstacles and cables induce a trip perturbation that causes forward body movements, whereas a low-friction movable platform and slippery agents and contaminants induce a slip perturbation that causes backward body movements. Multiple studies have focused on understanding the biomechanical and physiological mechanisms and motor adaptations after multiple exposures to induced trips and slips [7]–[14].

External mechanisms, however, add to the cost and operational complexity of fall-inducing technologies. Therefore, split-belt treadmills that provide gait perturbations are being developed. For example, Mueller *et al.* [15] have developed a custom split-belt treadmill system that induces stumbling or slipping by accelerating one belt posteriorly (stumbling) and anteriorly (slipping); the system assesses the trunk's neuromuscular responses after induced stumbling and slipping. Sessoms *et al.* [16] have demonstrated the beneficial effects of perturbation-based gait training on improvements in trunk dynamics during the initial compensatory step after multiple exposure to trip-like probations induced by a commercially available programmable split-belt treadmill (GRAIL, Motekforce Link B.V., Amsterdam, NL). Recently, we developed and assessed a fall-inducing technology equipped with a commercially available programmable split-belt treadmill (Bertec Corporation, Columbus, OH, USA) that induced unpredictable trip perturbations [17], [18]. The results of our previous studies demonstrated that the kinematic

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B.-C. Lee is with the Department of Health and Human Performance, University of Houston, Houston, TX 77204 USA, and also with the Michael E. DeBakey Veterans Affairs Medical Center, Houston, TX 77030 USA (e-mail: blee24@central.uh.edu).

C.-S. Kim is with the E-Mobility R&D Center, Korea Automotive Technology Institute, Cheonan-si 31214, South Korea (e-mail: cskim@katech.re.kr).

K.-H. Seo is with the System Control Research Center, Korea Institute of Robot and Convergence, Pohang 37666, South Korea (e-mail: neoworld@kiro.re.kr).

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changes following the unpredictable trip perturbations are consistent with the changes induced by mechanical obstacles or cables/pulleys [17], [18]. We also found that the initial compensatory stepping response and kinematic responses significantly improved after repeated exposure to unpredictable trip perturbations [17], [18]. Most recently, we have equipped our fall-inducing technology with the ability to provide trip and slip perturbations selectively [19].

Recognizing that the effects and the relationship of the two types of gait perturbation (i.e., trip and slip) induced by our fall-inducing technology on the body's compensatory responses have not been studied in depth, the main objective of this study is to quantitatively assess the kinetic and kinematic behaviors and lower limb muscle activity during quiet standing, steady state walking, and recovering following the unpredictable trips and slips induced by our fall-inducing technology in healthy young adults. In particular, for kinetic behaviors, ground reaction ground reaction forces (GRFs) and first step response times were analyzed. For kinematic behaviors, trunk, shoulder, and whole body center of mass (COM) dynamics were analyzed. For both behaviors, lower limb muscle activities of the compensatory limb (i.e., non-perturbed limb) were also analyzed. Our eventual goal is to improve next-generation perturbation-based gait training in clinical settings with a commercially available programmable split-belt treadmill.

II. Methods

A. Apparatus

An experimental apparatus consists of a 12-camera motion capture system (VICON, Vicon Motion Systems Ltd., Oxford, UK), an electromyography (EMG) system (Trigno ™IM, Delsys Inc., Natick, MA, USA), and a fall-inducing system with custom software containing a gait phase detection algorithm [17]–[19]. Our fall-inducing technology includes a commercially available programmable split-belt treadmill that has two force plates underneath (Bertec Corporation, Columbus, OH, USA). The custom software controls the treadmill's two belts based on the gait phase detection algorithm (detailed information about the algorithm is available in [17]–[19]); stopping one belt triggers a trip perturbation, and accelerating one belt in the anterior direction triggers a slip perturbation at foot level at approximately 10% of the gait cycle corresponding to the initial double-limb support. The stopped or accelerated belt returns to a pre-perturbation speed after the first heel strike of the non-perturbed foot, because stepping is the normal response [5], [6]. Returning to a pre-perturbation speed also allows the treadmill user to recover from gait perturbations by continuing to walk.

The motion capture system uses 35 reflective passive markers, typically used in gait studies, to measure the body kinematics [20]. The Nexus 1.8 software continuously samples the positions of the markers at a rate of 100 Hz, and records the EMG signals from 10 wireless surface EMG sensors and the ground reaction forces (GRFs) at a rate of 1 KHz. The custom software synchronizes the Nexus 1.8 software to start and stop recording. The custom software's other functions include: 1) specifying the step, foot, and velocity/acceleration associated with the perturbation; 2) specifying the type of perturbation; 3) specifying the forward or backward direction of the slip perturbation; 4) identifying a recovery time based on cross-correlation analysis of vertical GRFs before and after the perturbation [17]–[19]; 5) recording the onset time and gait events (heel strike and toe-off) of the perturbation; and 6) recording GRF signals from the two force plates.

B. Participants

Twenty healthy young adults (10 females and 10 males; age: 23.3 ± 3.3 yrs; stature: 173.2 ± 7.6 cm; weight: 67.6 ± 12.2 kg) participated. They were naïve to the purpose of the study.

Exclusion criteria included neurological disorders (e.g., myelopathy, stroke, etc.); peripheral sensory diseases (vestibular disorders, peripheral neuropathy, etc.); musculoskeletal dysfunctions; use of a walking aid; pregnancy; left-footedness; and body mass index (BMI) greater than 30 kg/m^2 (BMI over 30 may affect gaits (e.g., [21])).

The University of Houston Institutional Review Boards approved the study protocol, which is in accordance with the Helsinki Declaration. Each participant reviewed and signed the informed consent prior to the study.

C. Experimental Protocol

Thirty-five reflective passive markers were placed on the head, neck, trunk, shoulders, arms, upper and lower legs, and feet, ten surface EMG sensors were placed on the bilateral tibialis anterior, gastrocnemius (medialis and lateralis), rectus femoris, and biceps femoris, and a safety harness was worn. EMG placement for the trunk was excluded from this study because the safety harness causes movement artifacts during measurements of the trunk's muscle activity. Each participant chose a comfortable walking speed (0.9 ± 0.2 m/s) by adjusting the treadmill's speed.

Practice trials were not allowed and participants were given no instruction about when to expect a perturbation and how to respond. Since our previous studies found learning effects on the initial compensatory stepping response and trunk dynamics after four trials of trip perturbations [17], [18], in this study all participants completed 2 trials with a trip perturbation and 2 trials with a slip perturbation. The trial order was randomized to exclude potential learning effects [17], [18]. Numerous studies have demonstrated that the central nervous system is involved in the planning and execution of motor responses for maintaining stable balance during standing, walking, and compensatory reactions (see [22] for review). Therefore, each trial consisted of three consecutive periods of standing (15 s quiet standing), walking (steady state walking, at the participant's self-selected speed, corresponding to approximately 31 to 40 gait cycles), and recovery (compensatory responses, after perturbations, corresponding to approximately 4 to 6 gait cycles). While walking, the participant stared at an "X" mark approximately 4.5 m ahead and at eye level, which helped to reduce head movements and medial-lateral walking variations [17]–[19]. To be consistent with our previous studies [17], [18] and to simplify the experimental protocols, our fall-inducing technology applied the trip or slip perturbation randomly to the participant's non-dominant foot (i.e., left foot) between the 31st and 40th steps. Stopping the left belt induced a trip perturbation and accelerating it in the forward direction induced a slip perturbation as shown in Fig. 1. The left belt was controlled at a rate of 10 m/s^2 for the trip perturations and 20 m/s^2 for the slip perturbations. After the first heel strike of the right foot, the stopped or accelerated left belt returned to the participant's self-selected pre-perturbation speed at approximately 100 ms. Each trial ended when the participant returned to normal walking during the walking period [17]–[19]; thus each trial lasted approximately 60 s (15 s standing plus ~40 s normal walking plus ~5 s recovery walking after the gait perturbation). Participants could relax during the 20 s rest period between trials.

D. Data and Statistical Analyses

The Nexus 1.8 software ran the full body Plug-in-Gait model to process the position data of the 35 reflective passive markers, and MATLAB (The MathWorks, Natick, MA, USA) processed the GRFs and the EMG signals. The body kinematics and the GRFs were low pass filtered with a second-order Butterworth filter (cut-off frequency of 10 Hz) [23]. Since compensatory responses after gait perturbations can be quantified by kinetic and kinematic variables controlling the trunk, arms, whole body COM, and feet (see [4] for review), response step time, trunk range of motion (ROM), shoulder ROM, COM ROM, maximum trunk velocity, maximum shoulder velocity, and maximum COM velocity were used as the seven kinetic and kinematic outcome measures. Based on the GRFs, the response step time (i.e., first stepping time) denoted the time from the onset of perturbation to the first heel strike of the non-trip foot (i.e., right foot) [17]–[19]. Trunk and shoulder ROM denoted the range of motion in degrees between the flexion and extension maxima with respect to the vertical direction, and COM ROM denoted the range of motion in cm between its maximum elevation and depression along the vertical direction. Maximum trunk, shoulder, and COM velocity denoted the maximum trunk flexion, shoulder flexion, and COM elevation velocity, respectively. The computed trunk ROM, shoulder ROM, maximum trunk velocity, and maximum shoulder velocity corresponded to components in the sagittal plane. The computed COM ROM and maximum COM velocity corresponded to components in the transverse plane. In general, whole body movements predominated in both planes.

The EMG signals from the 10 sensors were also band pass filtered with a fifth-order Butterworth filter (high cut-off frequency of 300 Hz and low cut-off frequency of 20 Hz) [23]. For each sensor's EMG signals, a baseline correction (i.e., normalization) was performed by subtracting an average of EMG signals corresponding to the first 1 s of the standing period. The root-mean-square (RMS) value of the normalized EMG signals was computed in μV. Since the trip and slip perturbations were applied to the left foot, the four RMS values corresponding to the right tibialis anterior, gastrocnemius, rectus femoris, and bicep femoris, were used as the four EMG outcome measures.

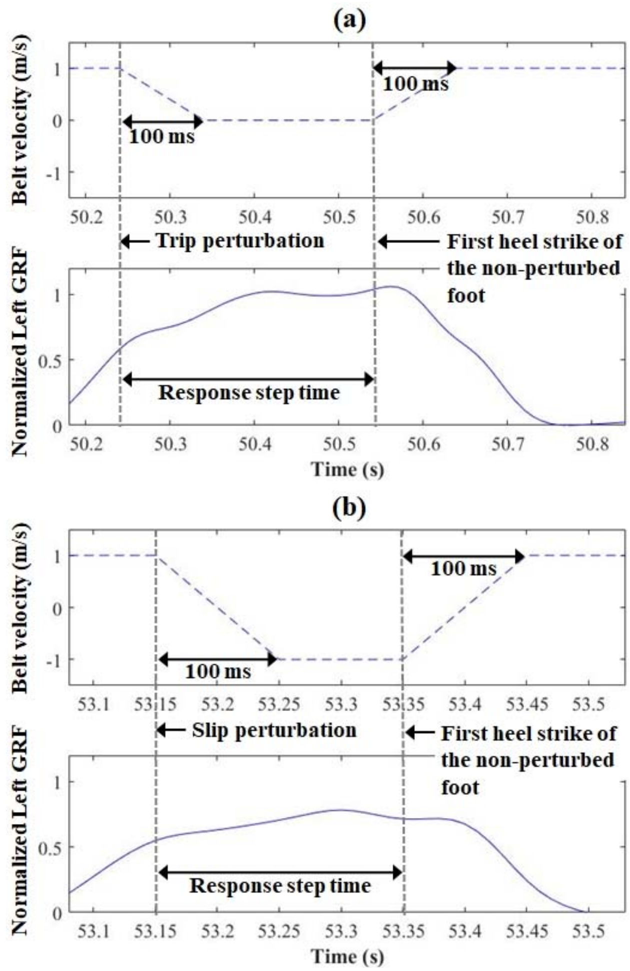

Fig. 1. Characteristics of the trip and slip perturbations induced by the split-belt treadmill and associated GRFs of the left foot from one trial of a representative participant who selected 1.0 m/s as a comfortable walking speed. (a) Trip perturbation. (b) Slip perturbation.

The six kinematic and four EMG outcome measures (i.e., trunk range of motion (ROM), shoulder ROM, COM ROM, maximum trunk velocity, maximum shoulder velocity, maximum COM velocity, and the four RMS values of each EMG location (i.e., right tibialis anterior, gastrocnemius, rectus femoris, and bicep femoris)) were computed for three periods of each trial. The kinetic measure, response step time, was computed only for the recovery period, and the average step time of the right foot was computed only for the walking period to compare the response and normal right step times.

All outcome measures were analyzed with SPSS (IBM Corp., Armonk, NY, USA). The Shapiro-Wilk test was used to confirm the normal distributions of the outcome measures. To assess the effect of the trial repetition (i.e., two trials for the trip or slip perturbation), initially a repeated measures analysis of variance (RMANOVA) was conducted for all outcome measures. Since the RMANOVA showed no significant effect of the trial repetition, each outcome measure for the two trials of each perturbation type was averaged as a function of the three periods for each participant. A one-way analysis of variance (ANOVA) was conducted for the response step time to assess the main effect of the two perturbations. A two-

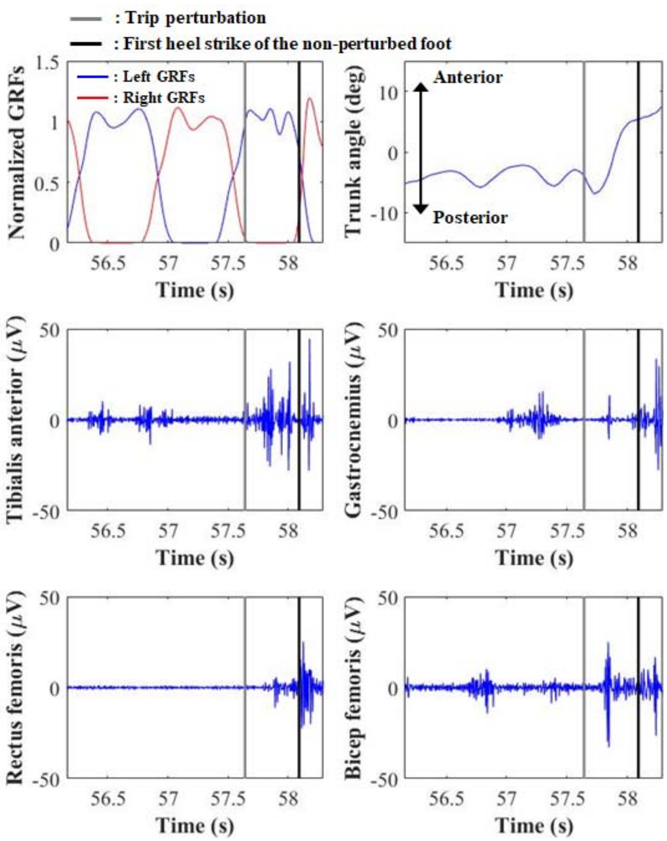

Fig. 2. Representative GRFs, trunk angles, and EMG signals for the right tibialis anterior, gastrocnemius, rectus femoris, and bicep femoris before and after a trip perturbation in one trial of a participant.

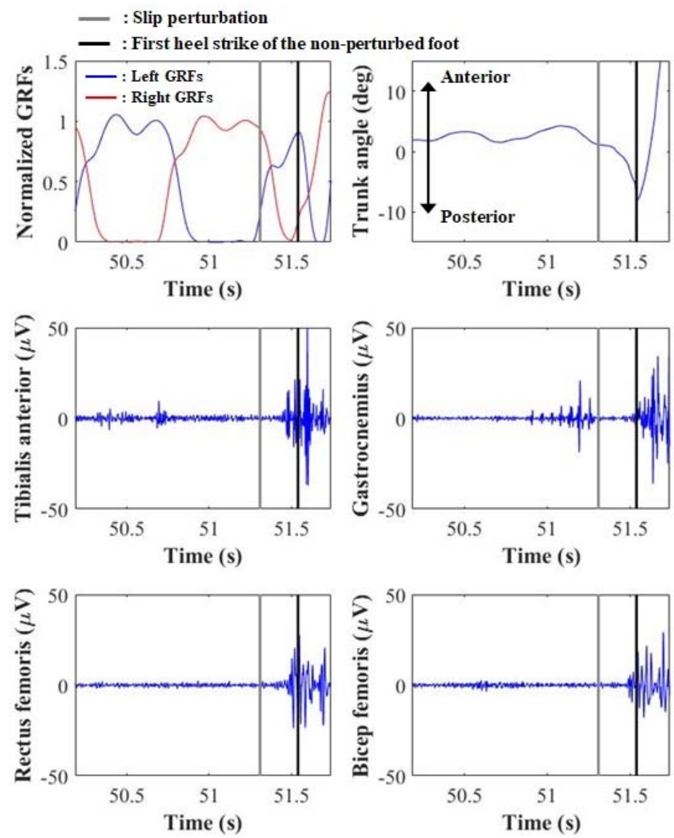

Fig. 3. Representative GRFs, trunk angles, and EMG signals for the right tibialis anterior, gastrocnemius, rectus femoris, and bicep femoris before and after a slip perturbation in one trial of a participant.

way ANOVA was conducted for the six kinematic and four EMG outcome measures to assess the main effect of the two types of gait perturbation (i.e., trip and slip), the three periods (standing, walking, and recovery), and their interactions (type of perturbation $\times$ period). An F test confirmed the hypothesis for the main effects of the independent variables and their interactions. Last, the Šídák method tested for the influence of any factors on the main effects and their interactions. Significance was defined at the $p < 0.05$ level.

III. Results

Figs. 2 and 3 show representative GRF profiles for both feet, trunk angles, and EMG signals for the right tibialis anterior, gastrocnemius, rectus femoris, and bicep femoris before and after the trip and slip perturbation in one trial of a participant. The trip and slip perturbation induced by our fall-inducing technology equipped with a commercially available programmable split-belt treadmill caused forward and backward trunk movements, respectively. Muscle activity of the compensatory limb (i.e., non-perturbed limb) also increased after the trip and slip perturbation.

A. Kinetics

Following an induced perturbation, the participants returned to their normal walking pace between 3 and 6 steps (trip: 3.2 ± 0.5 steps and slip: 4.3 ± 0.6 steps) corresponding to 3.36 ± 0.5 s for the trip perturbation and 4.7 ± 0.8 s for the slip perturbation. There were no falls at any time during the study. The normal GRF profiles during the walking period and the superimpositions of perturbed GRF profiles for one gait cycle (the GRFs were normalized to the body weight of each participant for illustrative purposes) are shown in Fig. 4. There were more variations in the perturbed gait cycle and GRF profiles regardless of the type of perturbation in

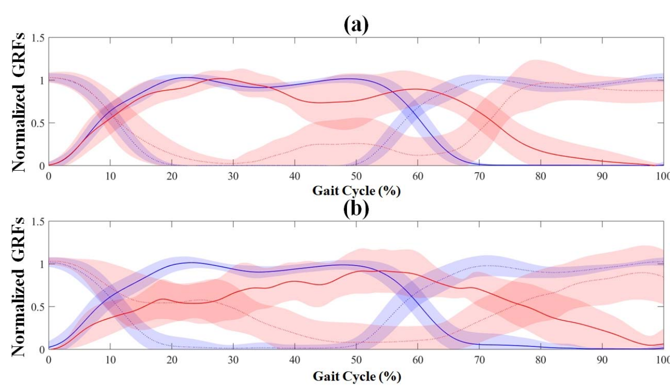

Fig. 4. Normal and perturbed GRF profiles as a function of the type of perturbation across all participants. (a) Average normal GRF profiles during the walking period and perturbed GRF profiles for the trip perturbation. (b) Average normal GRF profiles during the walking period and perturbed GRF profiles for the slip perturbation. Blue and red lines indicate normal GRF profiles and perturbed GRF, respectively. Solid and dashed lines indicate GRF profiles for the left and right foot, respectively. Shaded areas represent the standard deviation of the corresponding average.

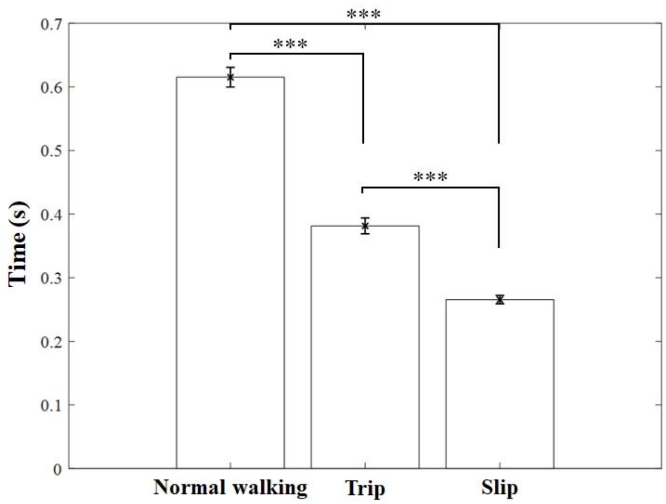

Fig. 5. Average step response time across all participants. Error bars indicate standard error of the corresponding mean (** $p < 0.01$ and *** $p < 0.0001$).

TABLE I

STATISTICAL ANALYSIS RESULTS OF THE SIX KINEMATIC OUTCOME MEASURES FOR THE TYPE OF PERTURBATION (T), PERIOD (P), AND INTERACTION (T × P). * $p < 0.05$, ** $p < 0.01$, AND *** $p < 0.0001$

Variable	Effects	DF	F value	Pr > F
Trunk ROM	T	1, 114	4.632	0.033*
	P	2, 114	170.819	< 0.0001***
	T × P	2, 114	3.678	0.028*
Maximum trunk velocity	T	1, 114	10.589	0.001**
	P	2, 114	173.713	< 0.0001***
	T × P	2, 114	11.400	< 0.0001***
Shoulder ROM	T	1, 114	4.196	0.043*
	P	2, 114	100.860	< 0.0001***
	T × P	2, 114	3.237	0.043*
Maximum shoulder velocity	T	1, 114	8.838	0.004**
	P	2, 114	110.677	< 0.0001***
	T × P	2, 114	8.264	< 0.0001***
COM ROM	T	1, 114	38.296	< 0.0001***
	P	2, 114	340.050	< 0.0001***
	T × P	2, 114	35.695	< 0.0001***
Maximum COM velocity	T	1, 114	19.233	< 0.0001***
	P	2, 114	252.722	< 0.0001***
	T × P	2, 114	20.903	< 0.0001***

the superimposition. The perturbed gait cycle and GRF profiles were more variable for the slip perturbation than for the trip perturbation.

One-way ANOVA applied to the response step times showed a significant main effect of the type of perturbation [F (2, 60) = 220.258, $p < 0.0001$]. The average step response times as a function of the type of perturbation, including the statistical significance from post hoc multiple comparisons, are shown in Fig. 5. Following both perturbations, the response step times were significantly quicker than the right step times during normal walking. Post hoc analysis also showed that a slip perturbation and a trip perturbation produced the quickest and second quickest response step times, respectively.

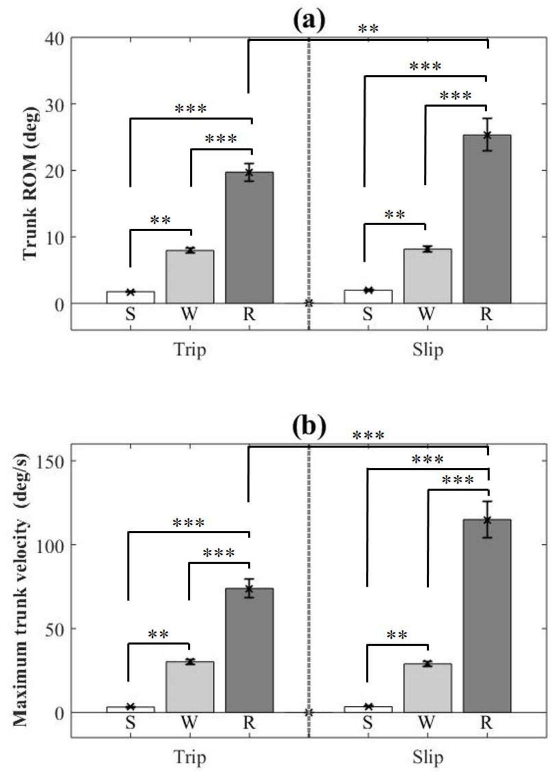

Fig. 6. Average trunk ROM and maximum trunk velocity as a function of the type of perturbation and period across all participants. (a) Trunk ROM. (b) Maximum trunk velocity. S, W, R indicate standing, walking, and recovery periods, respectively. Error bars indicate standard error of the corresponding mean (* $p < 0.05$, ** $p < 0.01$, and *** $p < 0.0001$).

B. Kinematics

The statistical analyses of the six kinematic outcome measures as a function of the type of perturbation and period are reported in Table I. Two-way ANOVA showed significant main effects of the type of perturbation, period, and type of perturbation × period interaction for the six kinematic outcome measures. The results of the six kinematic outcome measures as a function of the type of perturbation and period, including the statistical significance from post hoc multiple comparisons, are shown in Figs. 6, 7, and 8. Post hoc analysis showed that trunk ROM, maximum trunk velocity, shoulder ROM, maximum shoulder velocity, COM ROM, and maximum COM velocity significantly increased during the recovery period compared to the standing and walking periods for both perturbations. Only during the recovery period, the same analysis showed that the six kinematic outcome measures significantly increased for the slip perturbation compared to the trip perturbation.

C. Lower Limb Muscle Activity

The statistical analyses of the four RMS values as a function of the type of perturbation and period are reported in Table II. Two-way ANOVA showed significant main effects of the type of perturbation, period, and type of perturbation × period interaction for the four RMS values. The results of the four RMS values as a function of the type of perturbation and period, including the statistical significance from post hoc

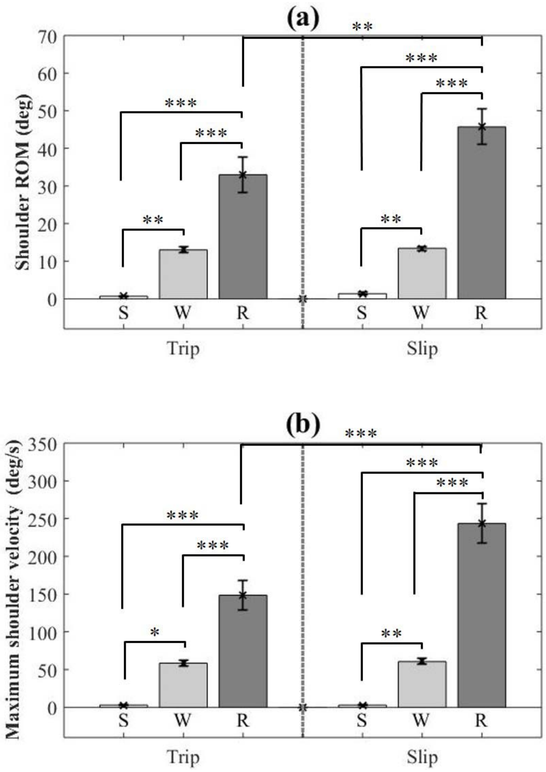

Fig. 7. Average shoulder ROM and maximum shoulder velocity as a function of the type of perturbation and period across all participants. (a) Shoulder ROM. (b) Maximum shoulder velocity. S, W, R indicate standing, walking, and recovery periods, respectively. Error bars indicate standard error of the corresponding mean (* $p < 0.05$, and *** $p < 0.0001$).

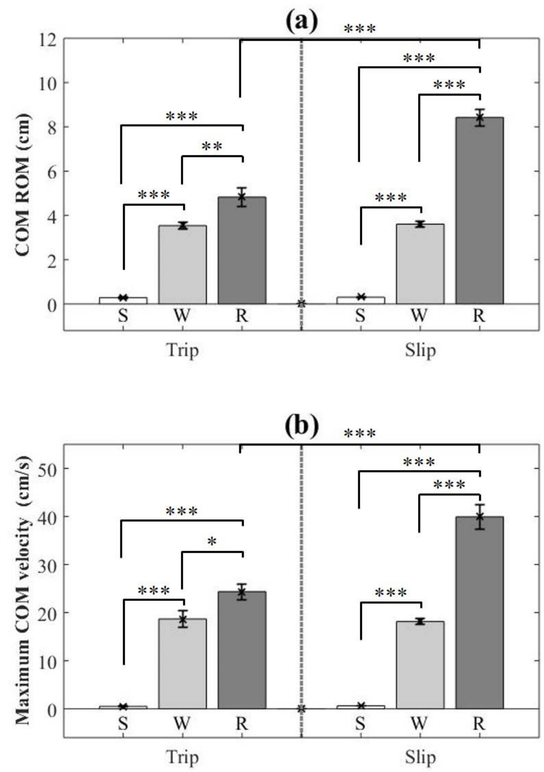

Fig. 8. Average COM ROM and maximum COM velocity as a function of the type of perturbation and period across all participants. (a) COM ROM. (b) Maximum COM velocity. S, W, R indicate standing, walking, and recovery periods, respectively. Error bars indicate standard error of the corresponding mean (* $p < 0.05$, ** $p < 0.01$, and *** $p < 0.0001$).

multiple comparisons are shown in Fig. 9. Post hoc analysis showed that the four RMS values significantly increased during the recovery period compared to the standing and walking periods for both perturbations. Only during the recovery period, the four RMS values significantly increased for the slip perturbation compared to the trip perturbation.

IV. Discussion

The results of this study demonstrated the effects of the two most common gait perturbations (i.e., trip and slip) induced by a commercially available programmable split-belt treadmill on kinetic and kinematic behaviors and muscle activity of the compensatory limb (i.e., non-perturbed limb). Notably, stepping responses to slip perturbations were quicker compared to trip perturbations. Trunk, shoulder, and whole body COM movements after slip perturbations were higher than those after trip perturbations. EMG results showed that tibialis anterior, gastrocnemius, rectus femoris, and biceps femoris activities were also higher after slip perturbations than those after trip perturbations.

Consistent with our previous findings [17]–[19], step response times (i.e., first stepping times) after a trip perturbation were significantly quicker than the normal step times during walking periods. Reactive stepping of the compensatory limb is the normal response to unexpected perturbations [5], [6], and the ability to take quick steps after unexpected perturbations is critical to recovering balance equilibrium and preventing falls [24]. Thus, we attribute the quicker stepping after the unexpected trip and slip perturbations to counteracting or preventing the loss of balance [25], [26]. Noting that step response times were significantly quicker in response to slip perturbations compared to trip perturbations, we also infer that the backward loss of balance caused by slip perturbations requires a quicker stepping response by the compensatory limb, because the margin of postural stability is smaller in a backward direction than a forward direction [27]. Our inference is supported by a previous finding that stepping time was quicker for the backward loss of balance than the forward loss of balance caused by a sudden translation of a surface platform while standing [27]. Thus, we infer that the first stepping response of the compensatory limb after a slip perturbation is largely automatic because of the relatively quick step response time. Step response times after the trip and slip perturbations (see Fig. 5) were similar to those after the forward and backward platform translation previously reported [27], which confirms that our fall-inducing technology equipped with a commercially available programmable split-belt treadmill can provide the trip and slip perturbations by inducing the forward and backward loss of balance during walking.

Successful recovery from sudden and unexpected perturbations during standing and walking depends on well- adjusted movements of the trunk, COM, and upper and lower extremities. The kinematic results showing significant increases in trunk, shoulder, and COM movements during recovery periods after the perturbations are congruent with the

TABLE II

STATISTICAL ANALYSIS RESULTS OF THE FOUR RMS VALUES AS A FUNCTION OF THE EMG LOCATION FOR THE TYPE OF PERTURBATION (T), PERIOD (P), AND INTERACTION (T × P). * $p < 0.05$, ** $p < 0.01$, AND *** $p < 0.0001$

Variable	Effects	DF	F value	Pr > F
RMS (right tibialis anterior)	T	1, 114	4.480	0.036*
	P	2, 114	208.271	< 0.0001***
	T × P	2, 114	7.709	0.001**
RMS (right gastrocnemius)	T	1, 114	9.781	0.002**
	P	2, 114	134.157	< 0.0001***
	T × P	2, 114	11.606	< 0.0001***
RMS (right rectus femoris)	T	1, 114	9.737	0.002**
	P	2, 114	165.248	< 0.0001***
	T × P	2, 114	16.353	< 0.0001***
RMS (right bicep femoris)	T	1, 114	15.337	< 0.0001***
	P	2, 114	16.353	< 0.0001***
	T × P	2, 114	16.801	< 0.0001***

body's compensatory responses to both types of perturbation during recovery periods (see [4] for review). We attribute the increased trunk, shoulder, and COM movements for the slip perturbation compared to the trip perturbation (see Figs. 6, 7, and 8) to our observation that the treadmill-induced slip perturbations challenged gait and balance stability more than the treadmill-induced trip perturbations. Since the feet provide larger limits of stability in a forward direction than in a backward direction [28], [29], it is reasonable to assume that the forward stepping response contributes to fewer compensatory adjustments by the body following trip perturbations compared to slip perturbations. The results of lower limb muscle activity also support our observation. During recovery periods, right tibialis anterior, gastrocnemius, rectus femoris, and biceps femoris activity was significantly higher after slip perturbations compared to trip perturbations. Moreover, lower limb muscle activity significantly increased during recovery periods after both perturbations.

Clearly, recovery from unexpected balance perturbations while standing involves cortical activity and cognitive processes (see [22] for review). Previously, having found that recovery from slip perturbations with our fall-inducing system required more attentional resources than standing and normal walking on a treadmill, as assessed by activity of the prefrontal cortex [30]. Thus, we infer that that more trunk, shoulder, and COM movements and higher lower limb muscle activity after slip perturbations require more involvement of the prefrontal cortex compared to trip perturbations. We expect to confirm this inference by exploring the effects of types of perturbation (i.e., trip or slip) on prefrontal cortex activity.

Our study has four limitations: a single participant cohort (i.e., healthy young adults); a single perturbation magnitude; no measurements of upper body muscle activities; and no onset detection of lower body muscle activity. It has been well documented that aging affects the body's compensatory responses (e.g., [31]) and neurological conditions (e.g., stroke [32]) differently. Therefore, future research will focus on understanding the different magnitudes of trip and slip perturbations on the body's compensatory responses by adding more EMG sensors to record upper body muscle activities and to analyze

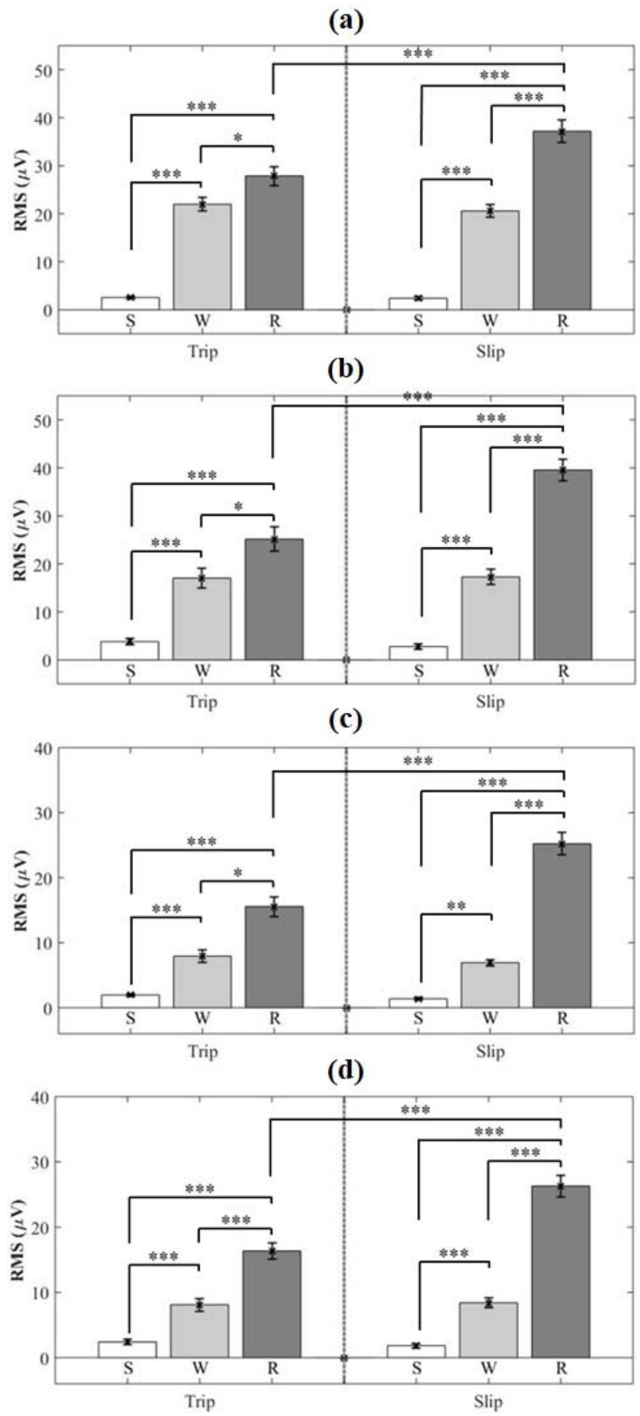

Fig. 9. Average RMS values of the four EMG locations as a function of the type of perturbation and period across all participants. (a) Right tibialis anterior. (b) Right gastrocnemius. (c) Right rectus femoris. (d) Right bicep femoris. S, W, R indicate standing, walking, and recovery periods, respectively. Error bars indicate standard error of the corresponding mean (* $p < 0.05$, ** $p < 0.01$, and *** $p < 0.0001$).

EMG onset in older adults and individuals with neurological conditions (e.g., Parkinson's disease).

V. CONCLUSION

This study quantitatively assessed the body's compensatory responses following the two types of gait perturbation induced by a commercially available programmable split-belt treadmill. Compared to trip perturbations, slip perturbations led to quicker step response times, more trunk, shoulder, and COM movements, and more lower limb muscle activity during recovery periods. The observed body's compensatory responses following treadmill-induced trip and slip perturbations were congruent with those induced by systems equipped with obstacles, cables, slippery agents and contaminants, etc.

Medical, geriatric, and physical therapies are increasingly dependent upon technologies, including instrumented treadmills, to train balance-constrained individuals. Aging, especially when combined with neurological disorders or failing eyesight, disrupts the body's "normal" responses to trips and slips, which are the most common gait perturbations [2], [33], [34]. The findings of this study suggest that a commercially available programmable split-belt treadmill can provide trip and slip perturbations without incorporating costly external mechanisms (e.g., obstacles, cables, low-friction movable platforms, etc.). Since perturbation-based gait training appears to prevent falls in different populations (see [4] for review), developing next-generation gait perturbation paradigms with the use of a commercially available programmable split-belt treadmill will improve perturbation-based gait training.

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