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The effect of the most common gait perturbations on the compensatory limb's ankle, knee, and hip moments during the first stepping response

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ABSTRACT

Background: Trips and slips, the two most common gait perturbations, often cause falls. Multiple studies have focused mainly on the kinematics of multiple body segments in response to an unexpected trip or slip induced by mechanical obstacles, cables, treadmills, and slippery agents or contaminants on a floor. Few studies have examined the joint moments of the compensatory limb following an unexpected trip on an obstacle.
Research question: This proof-of-concept study sought to assess the ankle, knee, and hip moments of the compensatory limb during normal walking and the first stepping response following the two most common gait perturbations.
Methods: Eighteen healthy young adults completed 4 trials (2 trials with a random trip perturbation and 2 trials with a random slip perturbation) while walking on a split-belt treadmill. In each trial, the motorized treadmill induced either an unexpected trip or slip perturbation to the left foot between the 31 st and 40th step randomly. A motion capture system recorded the positions of body segments, the joint moments (i.e., ankle, knee, and hip moments) of the compensatory limb were quantified, and the maximum joint moments were assessed during normal walking and the first stepping response.
Results: Compensatory limb's ankle plantarflexion, knee flexion, hip flexion, and hip extension moments were significantly higher for a slip perturbation than for a trip perturbation during the first stepping response. Compensatory limb's knee flexion, hip flexion, and hip extension moments were also significantly higher during the first stepping response to a slip perturbation compared to normal walking.
Significance: This proof-of-concept study is the first to investigate the ankle, knee, and hip moments of the compensatory limb during the first stepping response following unexpected gait perturbations induced by a split-belt treadmill. The findings are expected to improve the gait perturbation paradigms developed for training balance-impaired individuals.

1. Introduction

Unexpected gait perturbations (i.e., trips and slips) while walking are the main causes of falls in young and older adults [1,2], because they contribute to loss of balance and gait stability. Fall-related injuries (e.g., fracture) and consequences (e.g., fear of falling) increase the cost of healthcare [3] and lead to losses in disability-adjusted life years, respectively [4].

Multiple studies have focused mainly on kinematic analyses of various body segments responding to the unexpected trips and slips induced by mechanical obstacles, cables, treadmills, and slippery agents or contaminants on the floor (see [5–7] for review). For example, a trip perturbation causes increased maximum knee, hip, and trunk flexion angles during the first stepping response [8], whereas a slip perturbation causes loss of balance in a backward direction during the same stepping response [9,10]. Although successful recovery depends on coordinating the movements of multiple body segments, the muscles in the lower limbs play a major role in reacting to, and preventing, a fall after the loss of balance and gait stability [11]. In addition, the muscle strength of the lower limbs is one of the crucial predictors of falls by older adults [12]. Previous studies have demonstrated that older adults with weakened muscles of the lower limbs fell more often than young

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adults as well as their elderly counterparts, because both young adults and their elderly counterparts had relatively stronger lower limbs [13,14].

Considering that the muscles of the lower limbs contribute to moment generation of the lower limb joints (e.g., hip, knee, and foot), several studies have assessed the kinetic (i.e., moment) responses of a slipped limb [15,16]. One study found that a slipped limb's plantarflexion, knee flexion, and hip extension moments increased enough to actually stop the rapid acceleration of a slipped foot [16], while another study found that the high knee flexion and extension moments of a slipped limb helped to decrease the velocity of a slipped foot [15]. Although stepping with the compensatory limb (i.e., contralateral side of the perturbed limb) is the normal response to an unexpected gait perturbation [17,18], only a few studies have examined the joint moments of the compensatory limb following an unexpected trip on an obstacle [19,20]. To our knowledge, no published studies have investigated the effects of the two most common gait perturbations induced by a split-belt treadmill on the joint moments of the compensatory limb during the first stepping response.

The purpose of this proof-of-concept study is to assess the ankle, knee, and hip moments of the compensatory limb during normal walking and the first stepping response following the two most common gait perturbations. A new fall-inducing system with a programmable split-belt treadmill that we developed provided the trip and the slip perturbation [21,22]. The observed kinematic changes due to treadmill-induced trips and slips were consistent with those induced by methods incorporating mechanical obstacles or slippery contaminants; in particular, a trip and slip perturbation induced by our fall-inducing system caused the body to move forward and backward, respectively.

2. Methods

2.1. Participants

According to the kinematic results of our previous studies [21,22], the power analysis indicated a minimum of 16 participants, with an effect size (f) = 0.67, power (1-β) = 0.80, and α = 0.05. Thus, we recruited eighteen young adults (9 males and 9 females; age: 23.06 ± 3.42 yrs; stature: 172.96 ± 7.96 cm; weight 66.87 ± 12.36 kg). All participants had: 1) no major operation negatively impacting balance in the last 6 months, 2) less than 30 kg/m^2 in BMI (BMI < 30 kg/m^2 affects balance and gait performance [23]), 3) a cognitive score 26 or above as determined by the Montreal Cognitive Assessment (representing normal cognitive ability), and 4) no left-footedness. Exclusion criteria included neurological disorders (e.g., stroke and Parkinson's disease), musculoskeletal dysfunction, and peripheral sensory disease (e.g., diabetes and vestibular disorders).

The study protocol was approved by the University of Houston Institutional Review Boards, which is in line with the Helsinki Declaration. Each participant reviewed and signed the informed consent form before the experimental procedures.

2.2. Experimental procedures

Fig. 1 shows the experimental apparatus, which included a motion capture system (VICON, Vicon Motion Systems Ltd., Oxford, UK) and a fall-inducing system with a programmable split-belt treadmill equipped with a force plate under each belt (Bertec Corporation, Columbus, OH, USA) and custom software (i.e., fall-inducing software). Each participant was instrumented with 35 reflective passive markers placed on the body landmarks bilaterally (i.e., head, trunk, arms, legs, and feet) and wore a safety harness, also shown in Fig. 1. A load cell (LC101-250; Omega Engineering Inc., CT, USA) was connected between the ceiling and the safety harness to measure the loading force (LF) exerted by the body weight due to gait perturbations.

The fall-inducing software, developed using Microsoft Visual C++, sampled the force signals (i.e., LF and ground reaction force (GRF) from the two force plates) and ran the gait phase detection algorithm at a rate of 100 Hz. The force signals were low pass filtered with a 10 Hz cut-off frequency to remove unwanted noises. The gait phase detection algorithm detected heel strike, toe off, and gait cycle using the filtered GRF signals. Fall-inducing software provided the trip or slip perturbation at the loading response phase corresponding to the initial double-limb support (i.e., approximately 10% of the gait cycle) by controlling one of the treadmill's belts (stopping one belt triggered a trip perturbation and accelerating one belt in the anterior direction triggered a slip perturbation [22]). Stopping or accelerating one belt occurred at a rate of 25 m/s^2 (a maximum acceleration indicated by the manufacturer). The belt stopped within 200 ms for a trip perturbation, and its speed reached 2.5 m/s within 200 ms for a slip perturbation. A fall incident was defined when the maximum LF exceeded 30% of the participant's body weight [24]. The motion capture system was synchronized with fall-inducing software and recorded the positions of the markers at a rate of 100 Hz.

Before starting an experimental trial, each participant chose a comfortable walking speed (0.86 ± 0.15 m/s) by adjusting the treadmill's speed. All participants completed 4 trials: 2 trials with a trip perturbation and 2 trials with a slip perturbation. The order of the trials was randomized to rule out the potential for learning effects [21,22]. During each trial, the participants walked on the treadmill at their chosen walking speed and fixed their gaze on an "X" mark on the wall ahead at approximately 4.5 m and eye level. For all four trials, the fall-inducing system provided a trip or slip perturbation to the left foot between the 31 st and 40th step randomly. The stopped or accelerated left belt returned to a pre-perturbation speed after the first heel strike of the right foot (i.e., first stepping response). Each trial ended at 10 steps after the perturbation. Each trial lasted for approximately 50 s (˜ 40 s walking before the gait perturbation plus ˜ 10 s walking after the gait perturbation). Consecutive trials were separated by approximately 30 s to provide a rest period. No participants performed practice trials before the experimental trials, received information about the onset of perturbations, and how to respond. During all experimental trials, there were no system malfunctions or marker drops.

2.3. Data processing

Using Nexus 1.8 software (Vicon, Centennial, CO, USA), recorded marker positions were low pass filtered by a sixth-order Butterworth filter with a 6 Hz cut-off frequency. Using a Plug-in-Gait model (Vicon, Centennial, CO, USA), joint moments (i.e., ankle, knee, and hip moments) of the lower limb were computed based on the filtered positions of the markers. Fig. 2 illustrates the definition of the joint moments for the compensatory limb and representative profiles of ankle, knee, and hip moments for the compensatory limb.

Since our previous studies demonstrated that whole-body movements after unexpected gait perturbations predominated in the sagittal plane [21,22], flexion and extension moments for the compensatory limb's ankle, knee, and hip were analyzed during normal walking and the first stepping response. Joint moments of the compensatory limb were computed during the first stepping response corresponding to an instance from heel strike to toe-off after the perturbation began. For normal walking, joint moments of the compensatory limb were computed for the 5 steps before either a trip or slip perturbation, and then averaged for the 5 steps. A flexion movement for each joint was expressed as a positive value, and an extension movement for each joint was expressed as a negative value, as shown in Fig. 2. The maximum dorsiflexion, plantarflexion, knee flexion, knee extension, hip flexion, and hip extension moments of the compensatory limb were used as the six outcome measures for the two gait perturbations (trip and slip) and conditions (normal walking and perturbation).

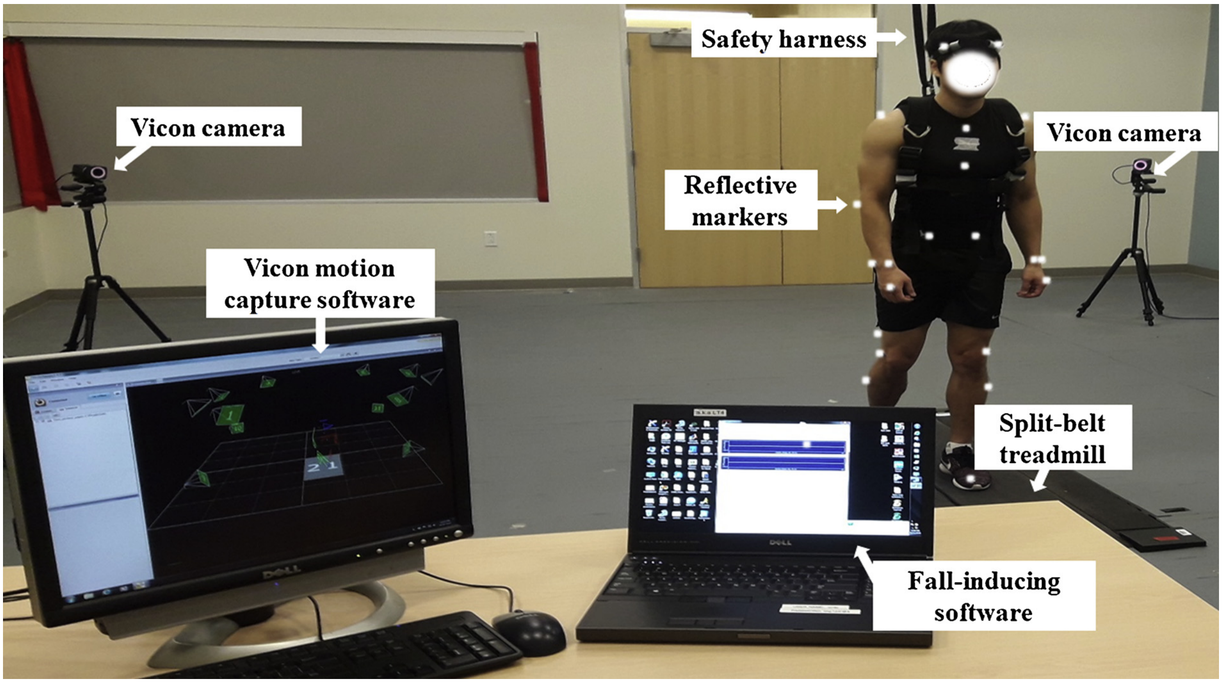

Fig. 1. Experimental apparatus.

2.4. Statistical analysis

Statistical analysis was conducted by SPSS (IBM Corp., Armonk, NY, USA) for the six outcome measures. Levene's test of equality of error variances indicated the normal distributions of the outcome measures. No trial repetition effect (i.e., 2 trials for the trip or slip perturbation) was confirmed by a repeated measures analysis of variance (RMANOVA). Each outcome measure for the 2 trials was averaged for each participant as a function of the two gait perturbations.

Two-way analysis of variance (ANOVA) determined the main effects of the two types of gait perturbation (trip and slip) and conditions (normal walking and perturbation), and the interaction effect (type of perturbation × condition). The main effects and the interaction effects were tested using an F test. Post hoc analysis was conducted with a Sidak's method to identify the factors which influenced the main and interaction effects. The significant level was defined at the $p < 0.05$ level.

3. Results

All participants resumed walking normally (i.e., no falls) within three or four steps after gait perturbations. Table 1 summarizes the

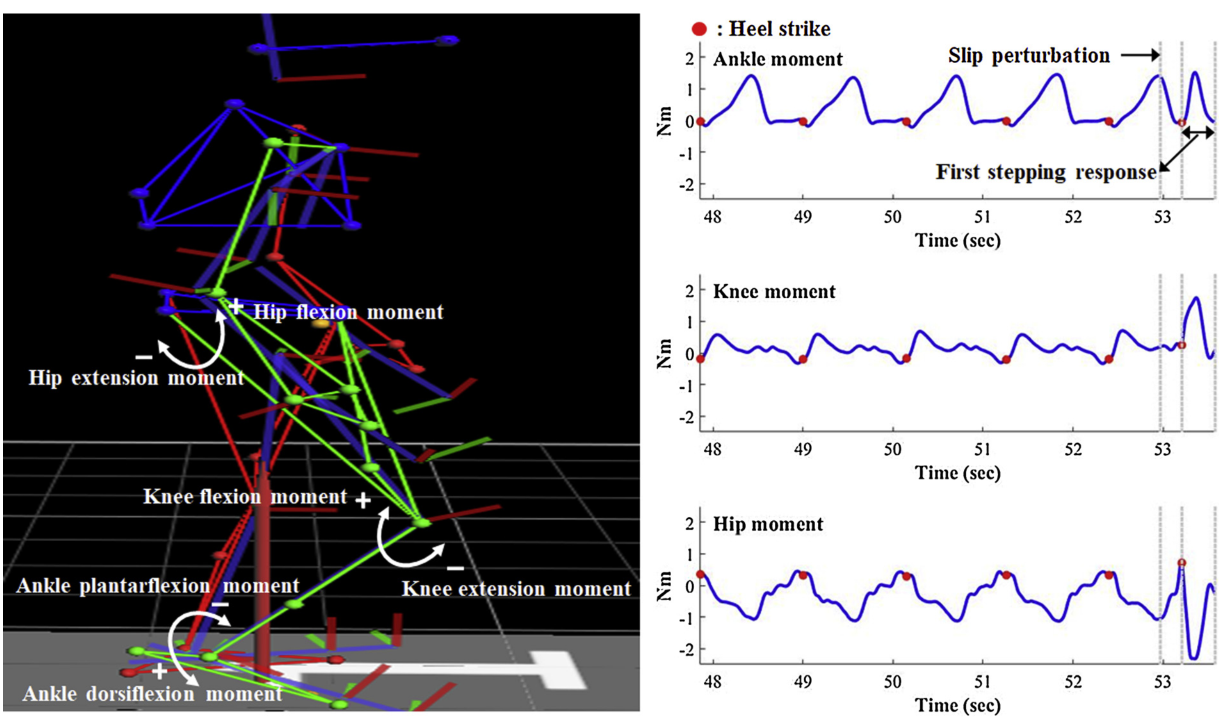

Fig. 2. Description of joint moments and representative profiles of ankle, knee, and hip moments for the compensatory limb.

Table 1
Statistical analysis results of all outcome measures for the type of perturbation (P), condition (C), and interaction (P × C) (*$p < 0.05$, **$p < 0.01$, and ***$p < 0.001$).

Dependent variable	Effects	DF	F value	p value
Dorsiflexion moment	P	1, 68	0.045	0.832
	C	1, 68	5.957	0.017*
	P × C	1, 68	0.002	0.966
Plantarflexion moment	P	1, 68	6.538	0.013*
	C	1, 68	20.259	0.000***
	P × C	1, 68	6.538	0.013*
Knee flexion moment	P	1, 68	9.062	0.004**
	C	1, 68	23.741	0.000***
	P × C	1, 68	5.198	0.026*
Knee extension moment	P	1, 68	0.046	0.830
	C	1, 68	1.983	0.164
	P × C	1, 68	0.103	0.750
Hip flexion moment	P	1, 68	10.283	0.002**
	C	1, 68	6.628	0.012*
	P × C	1, 68	11.024	0.001**
Hip extension moment	P	1, 68	23.034	0.000***
	C	1, 68	32.640	0.000***
	P × C	1, 68	22.434	0.000***

statistical analyses of the six outcome measures as a function of the type of perturbation and condition. Figs. 3–5 show the results of the six outcome measures as a function of the type of perturbation and condition, including the statistical significance from the post hoc analyses.

3.1. Ankle moments

Two-way ANOVA applied to the maximum dorsiflexion moment showed significant main effects of the condition [F (1, 68) = 5.957, p = 0.017]. However, the main effects of the type of perturbation and type of perturbation × condition interaction were insignificant. The two-way ANOVA applied to the maximum plantarflexion moment showed significant main effects of the type of perturbation [F (1, 68) = 6.538, p = 0.013], condition [F (1, 68) = 20.259, p = 0.000], and type of perturbation × condition interaction [F (1, 68) = 6.538, p = 0.013].

Post hoc analysis showed a significant decrease in maximum plantarflexion moment during the first stepping response compared to normal walking for the trip perturbation, as shown in Fig. 3(B). Maximum plantarflexion moment was significantly higher for the slip perturbation than for the trip perturbation during the first stepping response. Other pairwise comparisons for maximum dorsiflexion moments and maximum plantarflexion moments were insignificant.

3.2. Knee moments

Two-way ANOVA showed significant main effects of the type of perturbation [F (1, 68) = 9.062, p = 0.004], condition [F (1, 68) = 23.741, p = 0.000], and type of perturbation × condition interaction [F (1, 68) = 5.198, p = 0.026] for the maximum knee flexion moment. There were no significant main effects of the type of

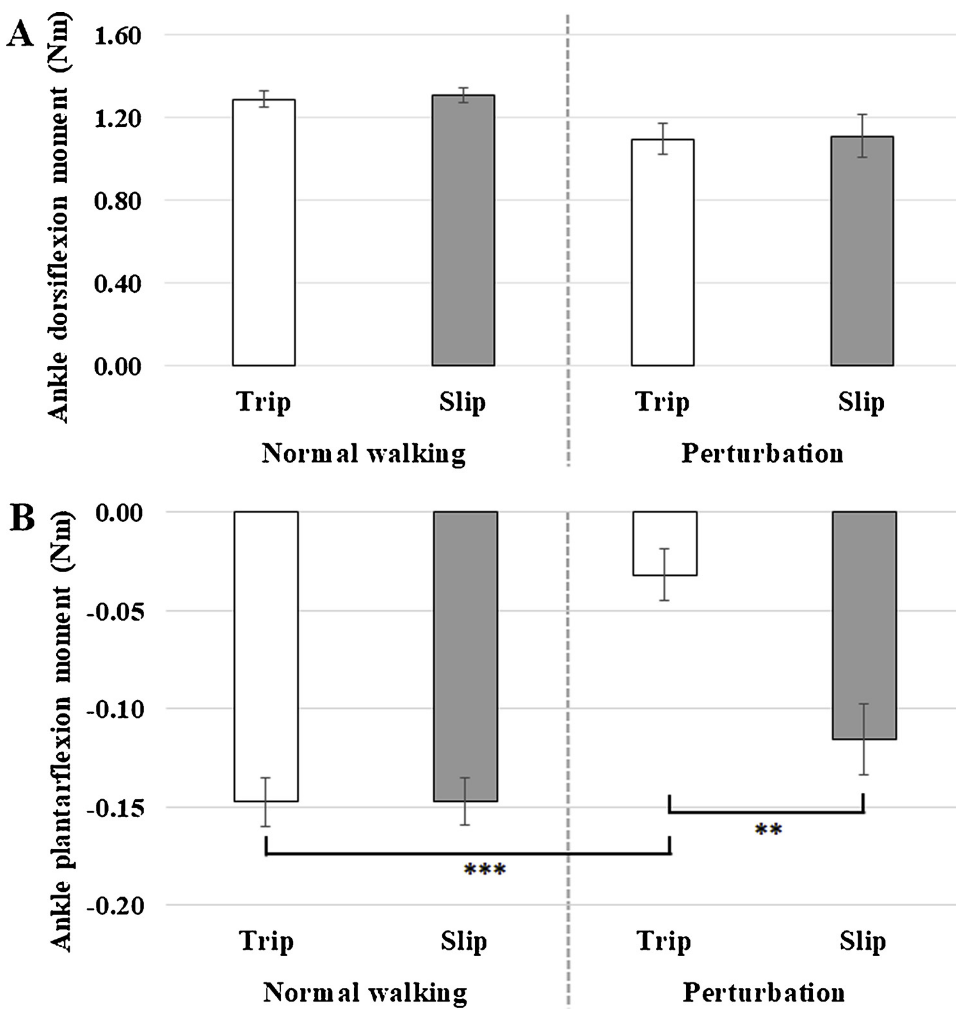

Fig. 3. Maximum ankle moments across all participants. Ankle dorsiflexion moment (A) and Ankle plantarflexion moment (B). White and light gray bars indicate a trip and a slip perturbation, respectively. Error bars indicate standard error of the corresponding average (**$p < 0.01$ and ***$p < 0.001$).

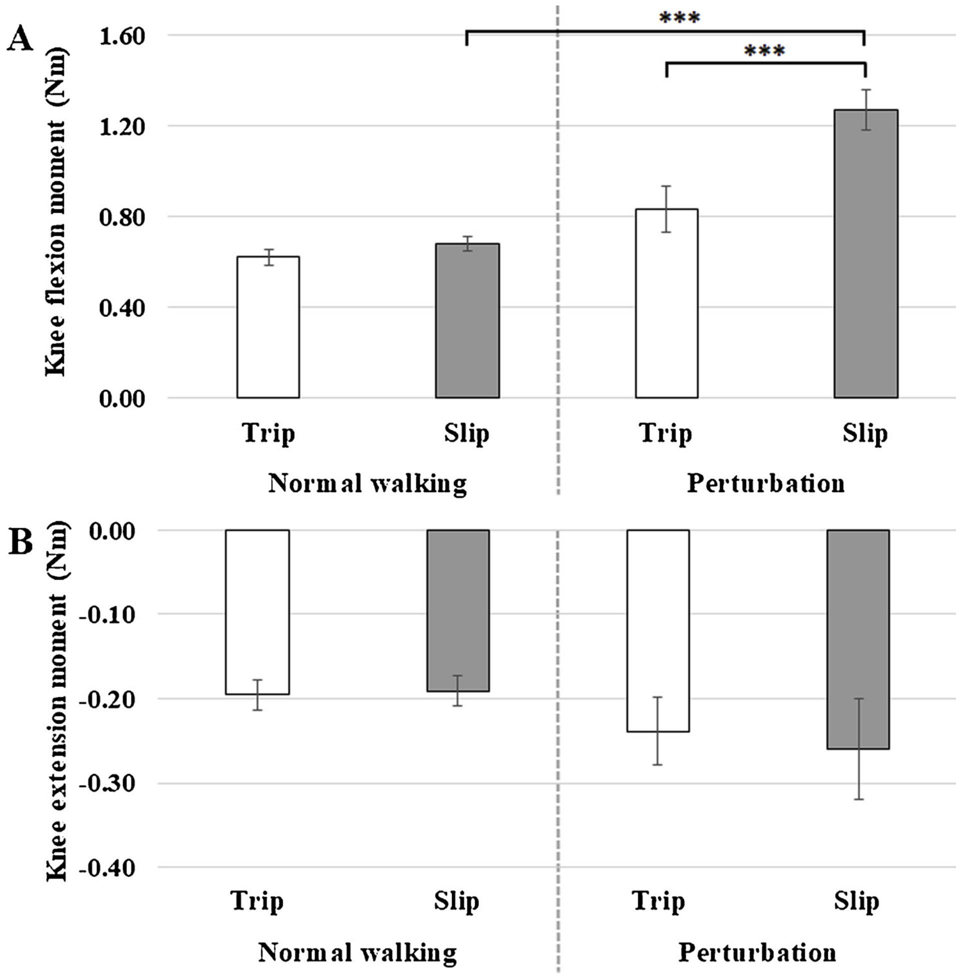

Fig. 4. Maximum knee moments across all participants. Knee flexion moment (A) and Knee extension moment (B). White and light gray bars indicate a trip and a slip perturbation, respectively. Error bars indicate standard error of the corresponding average (***$p < 0.001$).

perturbation, condition, and type of perturbation × condition interaction for maximum knee extension moments.

Post hoc analysis showed that the maximum knee flexion moment for the slip perturbation was significantly higher during the first stepping response compared to normal walking, as shown in Fig. 4(A). Maximum knee flexion moments were significantly higher for slip perturbations than for trip perturbations. Other pairwise comparisons for maximum knee flexion and maximum knee extension moments were insignificant.

3.3. Hip moments

Two-way ANOVA showed significant main effects of the type of perturbation [F (1, 68) = 10.283, $p =$ 0.002], condition [F (1, 68) = 6.628, $p =$ 0.012], and type of perturbation × condition interaction [F (1, 68) = 11.024, $p =$ 0.001] for the maximum hip flexion moment. The same analysis applied to the maximum hip extension moment showed significant main effects of the type of perturbation [F (1, 68) = 23.034, $p =$ 0.000], condition [F (1, 68) = 32.640, $p =$ 0.000], and type of perturbation × condition interaction [F (1, 68) = 22.434, $p =$ 0.000].

Post hoc analysis showed that both maximum moments (i.e., maximum hip flexion moment and maximum hip extension moment) for slip perturbations were significantly higher during the first stepping response compared to normal walking, as shown in Fig. 5. Both maximum moments were significantly higher for slip perturbations than for trip perturbations. Other pairwise comparisons for both maximum moments were insignificant.

4. Discussion

Two main findings of this proof-of-concept study are that: 1) maximum plantarflexion, knee flexion, hip flexion, and hip extension moments of the compensatory limb were significantly higher during the first stepping response after a slip perturbation than after a trip perturbation, and 2) maximum knee flexion, hip flexion, and hip extension moments of the compensatory limb were significantly higher during the first stepping response after a slip perturbation compared to normal walking.

A previous study has demonstrated increased knee flexion and hip extension moments and decreased plantarflexion moment of the slipped limb during attempts to recover from slipping on a slippery surface [25]. Similarly, our results show increased maximum knee flexion, hip flexion, and hip extension moments of the compensatory limb during the first stepping response after a slip perturbation compared to normal walking. However, maximum knee flexion, hip flexion, and hip extension moments of the compensatory limb between the first stepping response after a trip perturbation and normal walking did not differ. In addition, maximum plantarflexion, knee flexion, hip flexion, and hip extension moments of the compensatory limb were significantly higher after a slip perturbation than after a trip perturbation. We attribute these results to reality, i.e., slip perturbations challenge gait and balance stability more than trip perturbations, because slip perturbations require more adjustments of multiple body segments than trip

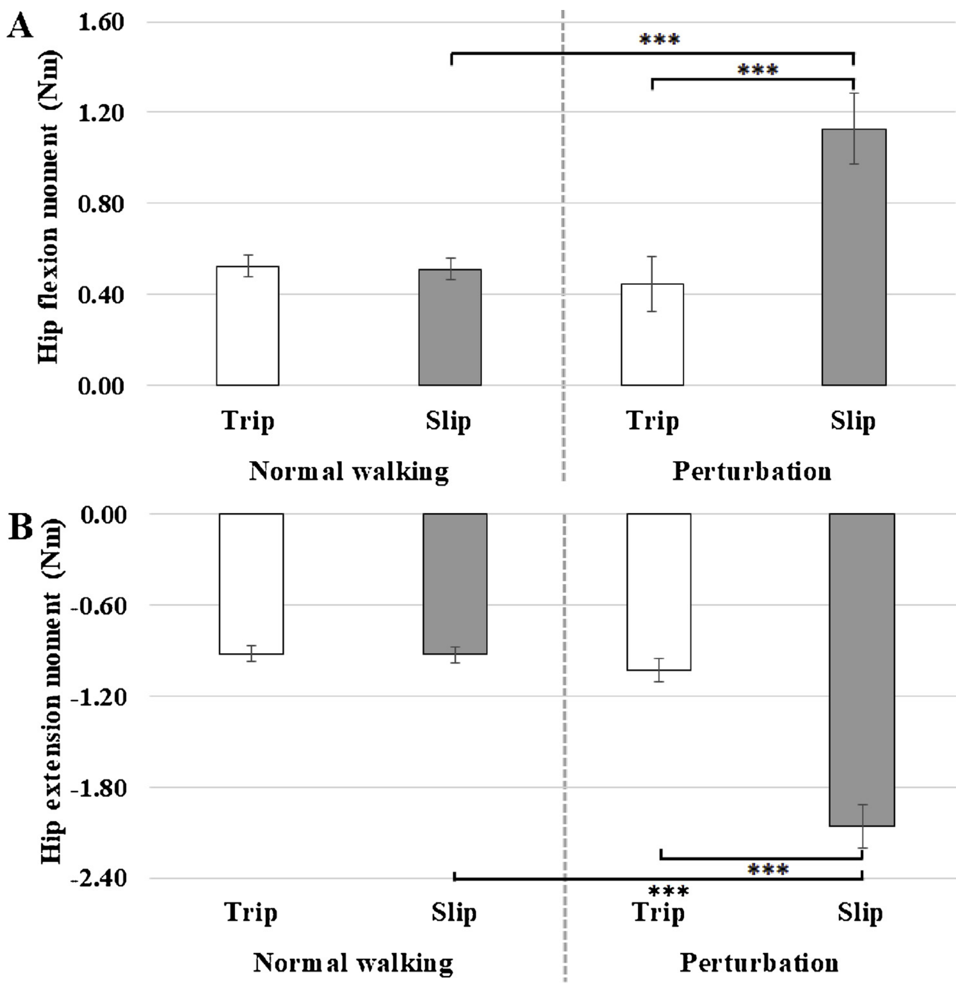

Fig. 5. Maximum hip moments across all participants. Hip flexion moment (A) and Hip extension moment (B). White and light gray bars indicate a trip and a slip perturbation, respectively. Error bars indicate standard error of the corresponding average (*** $p < 0.001$).

perturbations [26]. Our perspective is supported by two studies which found that quick stepping of the compensatory limb's foot stops or mitigates excessive trunk movements [26] and loss of balance [16].

Although we did not measure lower limb muscle activity, it is reasonable to assume that increased knee flexion, hip flexion, and hip extension moments of the compensatory limb after a slip perturbation may result from the increase in activity of the hamstring muscles (biceps femoris, semitendinosus, and semimembranosus). Our assumption is supported by a study which found that the hamstring muscles responsible for knee flexion and hip extension activate quickly to recover balance and prevent falls after a slip [27]. In addition, generating more hip extension on the support side contributes to restraining excessive backward trunk movements, increases the body's base of support after slips [27,28], and increases hip flexion, which is a common response to loss of balance caused by backward trunk movements [26]. A previous study has demonstrated that medial gastrocnemius activity of the slipped limb increase during recovery from a slip [27]. Thus, we attribute the increase in plantarflexion moments of the compensatory limb after a slip perturbation to the increase in activity of the plantarflexors (e.g., gastrocnemius, soleus, and tibialis posterior) which move the center of pressure anteriorly [29], thus contributing to produce sufficient force during push-off to recover gait and balance instability after slip perturbations.

Finding no significant change in knee flexion, hip flexion, and hip extension moments of the compensatory limb after trip perturbations, however, contradicts a previous study which demonstrated that knee flexion and hip extension moments of the compensatory limb significantly increase after trip perturbations compared to normal walking [20]. We attribute the contradiction to the method that we used to provide the unexpected trip perturbations, i.e., we controlled one belt of the split-belt treadmill, whereas the previous study utilized mechanical obstacles to induce a stumble [20]. In particular, generating more knee flexion and hip extension moments of the compensatory limb allows enough time and clearance to reposition of the tripped limb in response to the trip perturbations induced by the mechanical obstacle [20,30]. However, elevated clearance of the tripped limb is unnecessary in response to the trip perturbations induced by the treadmill.

The limitations of this study are its relatively small sample size, participant cohort (i.e., healthy young adults), and lack of measurement of the lower limb muscle activities. Despite these limitations, the findings of this study results could inform the future design of treadmill-induced gait perturbations examining the contributions of the joint moments of the compensatory limb during the first stepping response and the recovery performance following unexpected gait perturbations for different populations.

5. Conclusion

To our knowledge, this proof-of-concept study is the first to investigate the ankle, knee, and hip moments of the compensatory limb during the first stepping response after unexpected trips and slips. Our findings contribute to the published literature by characterizing the ankle, knee, and hip moments of the compensatory limb after the two most common types of gait perturbation. They will also contribute to improving the gait perturbation paradigms being developed for training susceptible individuals (e.g., older adults) to compensate for these perturbations.

Future research will focus on the effects of different perturbation types (i.e., obstacle-based trips and treadmill-based trips and slips) on the kinematics, kinetics, and lower limb muscle activities of older adults. Since gait perturbation paradigms that facilitate the body's reactive responses based on motor learning principles have been shown to reduce falls (See [5] for a review), the outcomes will improve the design of gait perturbation paradigms used for training balance-impaired people (e.g., older adults) to compensate for unexpected gait perturbations.

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Conflict of interest

The authors declare that they have no conflict of interest.

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