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The effect of vibrotactile cuing on recovery strategies from a treadmill-induced trip

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***Abstract*—Effective fall prevention technologies need to detect and transmit the key information that will alert an individual in advance about a potential fall. This study investigated advanced vibrotactile cuing that may facilitate trip recovery for balance-impaired individuals who are prone to falling. A split-belt treadmill that simulated unpredictable trip perturbations was developed to compare balance recovery without and with cuing. Kinetic and kinematic measures from force plates and full body motion capture system were used to characterize the recovery responses. Experiment I evaluated recovery adaptation resulting from repeated trip exposure without vibrotactile cuing. Experiment II investigated the effects of vibrotactile cuing as a function of cuing location (upper arm, trunk, lower leg) and lead time prior to a trip (250, 500 ms). Experiment I showed that trip recovery improved progressively from the fourth to the eighth trial. Experiment II showed that trip recovery was almost the same as the eighth trial in Experiment I, regardless of the location of the cuing stimulus and lead time. The results suggest that a combination of vibrotactile cuing and hazard detection technology could reduce the risk of trips and falls**

***Index Terms*—Vibrotactile cuing, induced trip, fall recovery, recovery kinetics and kinematics, fall prevention**

I. Introduction

FALLS are a significant hazard, particularly for the elderly. More than one third of individuals older than 65 years fall at least once per year [1-4]. Aging per se increases the risk of falling [5, 6]; 6% and 11% of aging-related falls result in fractures [7] and serious injury [8], respectively. Trips and slips account for 59% of falls in community-dwelling adults [9] and result in 57% of the fall injuries [3]. Medical costs for non-fatal and fatal falls in older adults are approximately 19 billion and 200 million dollars, respectively, per year [10]. Falls affect quality of life (sense of independence) by increasing the anxiety and fear of falling and reducing confidence in performing daily tasks [11-13].

A number of studies have examined the effects of therapeutic programs (e.g., resistance, endurance, balance, gait training) to improve posture, which in turn may reduce rates of falling (e.g., [14-18]). However, some therapeutic programs that involve various physical exercises show no reduction in falls [19-21], whereas others fail to adequately address the unexpected balance perturbations (e.g., trips and slips) that are the major causes of falls during walking [9]. Thus, there is growing interest in developing technologies that can detect physical hazards (e.g., obstacles, uneven surfaces) in the walking path.

In tandem with advances in actuator and sensor technologies, attempts have been made to develop insole-based systems equipped with miniature actuators (e.g., vibration motors) and shoe-based systems equipped with miniature sensors (e.g., ultrasonic and/or infrared sensors). Multiple studies have demonstrated that the insole-based systems improve balance and gait performance (e.g., reduce the range of postural sway, variability of walking, and chance of falling) via imperceptible vibratory noise (a.k.a. stochastic resonance) when applied to the soles in elderly people [22-25] and individuals with diabetic neuropathy [26]. Shoe-based systems also enable detection of ground-level obstacles [27, 28] and differentiate the ground's physical characteristics (e.g., deformable vs non-deformable) [29]. Potentially more important than detecting obstacles and ground conditions, Zhang et al. [30] were the first to propose a shoe-based alarm system using vibrotactile stimulation applied to the upper arm that signals the wearer about upcoming hazards. Indeed, vibrotactile information delivered as alert cuing may be advantageous because it does not interfere with visual or auditory modalities and it also encodes extrinsic feedback about the environment [31-34]. Shoe-based alarm systems utilizing vibrotactile cuing, however, need to address two important questions. Can vibrotactile cuing giving advanced warning information facilitate successful recovery from the hazards likely to cause a fall? What body locations and lead times (i.e., warning interval) are optimal for applying vibrotactile cuing?

Motivated by the questions, we developed a system that simulates the unpredictable trips associated with force perturbation acting at foot level. Prior studies have applied the technique to understand the benefits of adaptation to repeated exposure to tripping in order to improve fall recovery strategies [35-42]. We chose a programmable split-belt

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treadmill because it does not require external mechanical obstacles [35-40] or cables/pulleys [41, 42]. Our two experiments compared trip recovery performance resulting from the recognized adaptation by repetition method (e.g., [35-42]) and from vibrotactile cuing by quantitative assessment of whole-body kinetic and kinematic performance. We tested three locations of application and two lead times to determine the best cuing parameter. In this paper we describe the software design for unpredictable trips and advanced vibrotactile cuing based on a gait phase detection algorithm, quantitatively assess whole-body recovery performance based on established adaptation or cuing from simulated random trip perturbations, explore an optimal cuing location and lead time, and summarize the advantages of cuing over adaptation. Preliminary reports pertaining to this study have been published previously in abstract form [43].

II. Methods

A. Trip Simulation Apparatus

Fig. 1 shows the system's components: a split-belt treadmill (Bertec Corporation, Columbus, OH, USA), two load cells (LC101-250; Omega Engineering Inc., Stamford, CT, USA), and a customized control unit with three stimulators - C2 tactors (Engineering Acoustics Inc., Casselberry, FL, USA). The motorized split-belt treadmill equips with two force plates located underneath each belt as shown in Fig. 1(a). A trip was simulated by stopping the belt at the left foot loading phase,

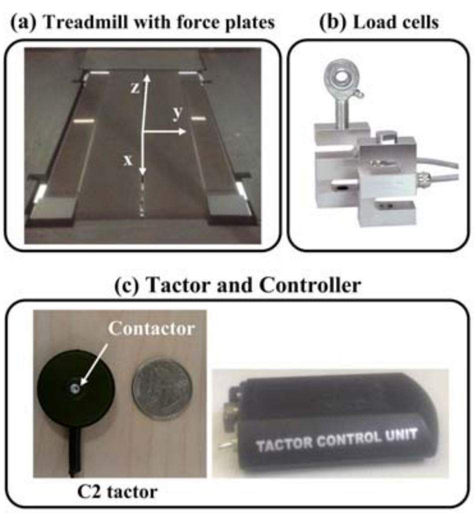

Fig. 1. Hardware components.

i.e., abruptly changing the horizontal forces at the foot [44-46]. Each force plate measures the ground reaction force (GRF) along the vertical z axis exerted on each foot when the foot is in contact with the force plate. Two load cells [Fig. 1(b)] connected between a support frame and a safety harness measure the loading force (LF) exerted by the body weight. Each load cell has a capacity of 1.11 KN with a sensitivity greater than 0.3N.

The moving contactor (0.8 cm in diameter) of the C2 tactor [Fig. 1(c)] is lightly preloaded against the skin to provide stable contact when delivering vibrotactile cues. The moving

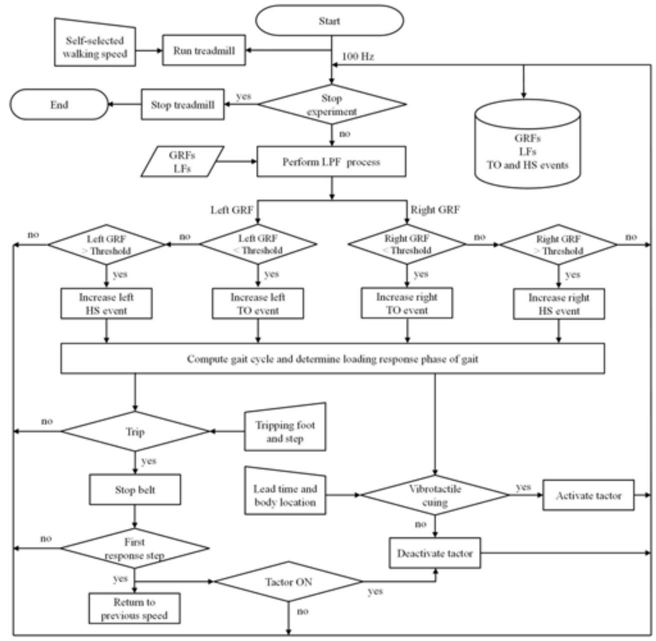

Fig. 2. Software architecture flow chart.

contactor oscillates perpendicularly to the skin, and the surrounding skin area is shielded with a passive housing (3 cm in diameter). The tactor is driven by a 250 Hz sinusoidal signal generated by a customized control unit, and the peak-to-peak displacement amplitude of the vibration is approximately 200 μm at the selected frequency [47].

We developed custom software (Microsoft visual C++) run a real-time gait phase detection algorithm, generate random trip perturbations by controlling each belt of the treadmill, generate and control vibrotactile cuing, and record the force data. Fig. 2 shows the software's flow chart. GRF and LF signals were sampled at a rate of 100 Hz and low-pass filtered by a second-order Butterworth filter with a 10 Hz cut-off frequency. Then the gait recognition algorithm detected heel strike (HS) and toe off (TO) events based on a pre-specified threshold (T) of 2% of vertical forces normalized to the body weight of each participant [48]. A HS event was registered when the GRF increased to greater than T, and a TO event was registered when the GRF decreased to less than T. Consecutive HS events were used to compute the duration of each gait cycle. Then the algorithm computed the loading response phase of gait occurring at approximately 10% of the gait cycle constituting the period of initial double-limb support [48]. As mentioned, a trip was simulated by stopping the belt at the left foot loading phase. During the perturbation, the belt was decelerated uniformly at a rate of 10 m/s^2 and completely stopped within 100 ms. The stopped belt returned to the pre-trip speed within 100 ms after the first heel strike of the non-trip foot (the first response step). This procedure was based on the commonly observed stepping behavior in balance recovery from unexpected perturbations [44, 49, 50]. When the tests were run with cuing information, the vibrotactile cue was generated by activating the tactor 250 or 500 ms before the trip perturbation and deactivating it when the HS of the first response step occurred.

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B. Participants

Twenty healthy young adults (8 females and 12 males; age: 25.4 ± 3.5 yrs; stature: 168.7 ± 8.6 cm; weight: 65.9 ± 11.0 kg) participated. All participants were randomly assigned to one of two groups of ten (4 females, 6 males). Exclusion criteria included any self-reported neurological disorder (e.g., myelopathy, stroke, Parkinson's disease), musculoskeletal dysfunction, peripheral sensory disease (e.g., peripheral neuropathy, Type 2 diabetes, vestibular disorder, etc.), use of any walking aid, pregnancy, left-footedness as determined by which foot was used to kick a soft rubber ball slowly rolled towards the participant, or a body mass index greater than 30 kg/m^2 (Previous studies have shown that a BMI over 30 may influence balance and gait stability [51-53]; therefore we aimed at recruiting participants with a BMI <30 to avoid possible interaction effects.). The University of Houston Institutional Review Boards approved the experimental protocol, which is in accordance with the Helsinki Declaration. Prior informed consent was obtained from each participant.

C. Experimental Protocol

Experiment I investigated adaptation from learning when the participant was repeatedly exposed to trip perturbation without vibrotactile cuing. The objectives were to explore the time course of adaptation and to provide the data that characterized the adapted response to multiple trip exposures without cuing. Experiment II investigated the influence of vibrotactile cuing on recovery performance from trip perturbation as a function of the stimulus location of application on the body and the lead time. The participants performing experiments I and II were termed the "adaptation" and "cuing" groups, respectively.

To evaluate trip kinematics and the subsequent recovery attempt, body kinematics were recorded with a 12-camera motion capture system (Vicon, Centennial, CO, USA) at a rate of 100 Hz synchronized with the custom software. All participants were instrumented with reflecting markers and tactors. The safety harness, which was adjusted to fit the participant, was attached by a pair of dynamic ropes to an overhead frame [Fig. 3(a)]. The ropes, which connected to the load cells, were adjusted so that the participant would not come into knee contact with the treadmill when tripping. Tactors were attached with Velcro to an elastic belt placed on the skin over the left upper arm (lateral head of triceps brachii), left trunk (external oblique area corresponding approximately to the L4/L5 level), and left lower leg (fibularis longus area) muscles. To obtain whole body kinematics, 24 reflective passive markers were placed on the head (frontal and occipital bones), neck (C7), shoulders (acromion), arms (lateral epicondyle of the ulnar and ulnar styloid process), trunk (manubrium, S1 vertebrae level on anterior superior iliac spine, and S1 level on erector spinae), upper legs (great trochanter), knees (lateral epicondyle of femur), ankles (lateral malleolus), and feet (great toe and heel bone), as shown in Fig. 3(b).

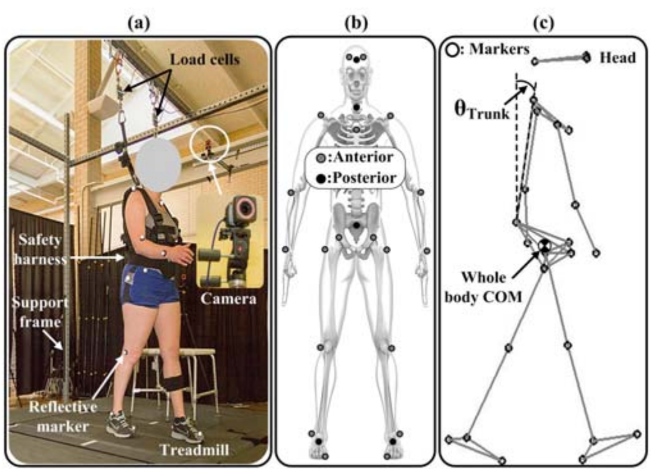

Fig. 3. An experimental platform and kinematic measurement. (a) Instrumentation. (b) Digital image of the 24 passive markers placed on the body landmarks. (c) A representative kinematic skeleton obtained with the Vicon system.

In experiment I, the adaptation group performed 8 trials without vibrotactile cuing. In experiment II, the cuing group performed 1 control trial without cuing, followed by 6 trials with vibrotactile cuing (3 cuing locations x 2 lead times), and finally 1 trial without cuing (post-control trial), for a total of 8 trials. During trials involving vibrotactile cuing, the location of the cuing stimulus and lead time were randomized. Vibrotactile cuing was provided to one of the locations indicated above with a lead time of either 250 or 500 ms prior to a trip. The lead time was determined to provide vibrotactile cuing during the swing phase (40% of the gait cycle) by assuming that the reaction time to touch stimuli was at least 155 ms [54], and that the mean duration of the self-paced gait cycle period for 20-35 year-old adults was approximately 1.06 s [55].

All participants walked on the split-belt treadmill at a self-selected walking speed (the self-selected speed was determined by adjusting the treadmill's speed prior to repeated exposure to the trip perturbation) while fixing their gaze on an "X" mark placed approximately 4.5 m ahead at eye level. The first 10 steps (pre-trip) in each trial were used to obtain a steady state cycle, prevent trip anticipation, and compute gait cycle parameters and average speed. Then a trip was randomly applied to the left foot between the tenth and twentieth steps. The trial terminated 10 steps after the trip step (post-trip). Each one-trip trial ranged from 21 to 30 total steps. The participants were given no information regarding the onset of the trip, body location and lead time for cuing, and no instruction regarding how they should respond to a trip. The duration of each trial was less than 1 min. Consecutive trials were separated by a 20 s rest period during which the treadmill was stopped. During rest, participants were instructed to relax by bending the torso and shaking their upper and lower extremities.

The measured self-selected walking speed was 0.99 ± 0.04 m/s (adaptation group) and 0.98 ± 0.06 m/s (cuing group). The self-paced gait cycle periods were 1.14 ± 0.07 s and 1.11 ± 0.08 s for the adaptation and cuing groups, respectively.

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D. *Data Analysis*

MATLAB (The MathWorks, Natick, MA) was used to post-process recorded signals from the force plates, load cells, and motion capture system. Force signals from two load cells and markers trajectories were low-pass filtered (2nd order Butterworth with zero lag and 10 Hz cut-off frequency).

Recovery performance from the induced trip perturbation was quantified by several variables from the kinetic and kinematic data, as justified in prior studies [35, 37, 39, 44]. The variables focused mainly on the importance of controlling balance recovery by controlling trunk inclination and whole body center of mass (COM). Hence, eight outcome measures were defined to characterize trip recovery kinetic and kinematic responses.

Four metrics of the eight outcome measures (i.e. response step time, maximum response step force, recovery time, and peak LF) were computed using the recorded GRF profiles. Fig 4 illustrates the definitions of the first three metrics. Response step time was defined as the time from the trip to the initiation of first step with the non-trip foot (i.e., right foot), maximum response step force was defined as the maximum foot contact force during the first response step, and recovery time was defined as the time from the trip to return to baseline gait based upon the force profile. Peak LF observed in the period between the instant of the trip and the instant of the first response step was computed as a percentage of maximum LF exerted on the safety harness by the participant's body weight. The fall incident was defined to occur when the maximum LF exceeded 30% of the participant's body weight [39].

Four kinematic metrics of the eight outcome measures (maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion angular dispersion (AD), and maximum whole body COM velocity) were calculated from the recorded marker displacements. Trunk flexion angle was computed by trigonometric methods using the markers attached to C7 and L5/S1 joints, respectively [Fig. 3(c)]. The trunk flexion velocity and trunk flexion AD were defined as a time derivative and a standard deviation, respectively, of the trunk flexion angle. The twelve COM locations (head, trunk, upper arms, forearm, upper legs, lower legs, and feet) were computed based on anthropometric data. Then the whole body COM position, as shown in Fig 3(c), was calculated [56]. The whole body COM velocity was computed as the time derivative of the whole body COM position. The four kinematic metrics were computed for the period corresponding to recovery as defined by the recovery time (Fig. 4). The metrics corresponded to components in the sagittal plane where whole body movements largely predominated.

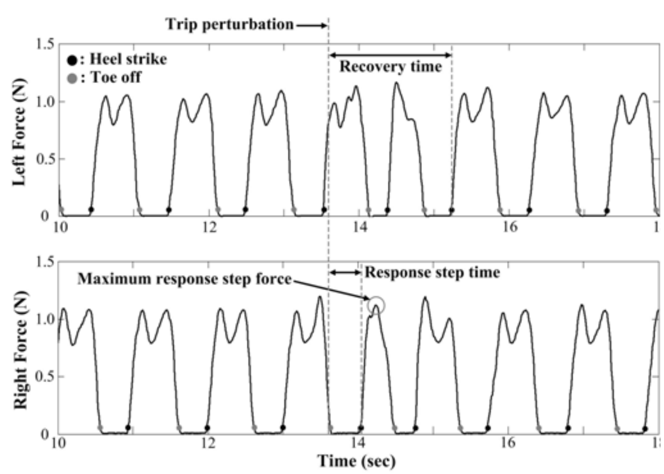

Fig. 4. Representative GRF profiles including three outcome metrics (i.e., response step time, maximum response step force, and recovery time).

The four kinematic metrics were also used to evaluate postural behaviors as a function of pre-, per-, and post-trip in each group's first trial. The first trial was considered to characterize the kinematic response to the participant's first trip. Each period was defined by the 5 steps before the instant of the trip (pre-trip), the step(s) included in the recovery time (per-trip), and the 5 steps succeeding recovery (post-trip), respectively.

All metrics were normally distributed according to Levene's test of equality of error variances. For each group's first trial, a two-way analysis of variance (ANOVA) was performed to assess the effects of the trip as a function of the group (adaptation and cuing) and period (pre-, per-, and post-trip) as well as their interactions (group x period) for maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion AD, and maximum whole body COM velocity. For the adaptation group, a one-way ANOVA was performed to determine the learning effect resulting from repeated exposure to the trip perturbation. For the cuing group, a one-way ANOVA was applied to determine the main effects of vibrotactile cuing (control, cuing, and post-control trials), and a two-way ANOVA was conducted to determine the main effects of stimulus location (upper arm, trunk, and lower leg) and lead time (250 ms and 500 ms) as well as their interactions (stimulus location x lead time). In the statistical analyses, the dependent variables corresponded to the eight outcome measures (i.e., response step time, response step force, recovery time, maximum LF, maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion AD, and maximum whole body COM velocity). The hypotheses for the main effects of all independent factors as well as their interactions were tested using an F test. To determine the factors influencing the main and interaction effects, post hoc analysis for each dependent variable was conducted using Sidak's method. Significance was defined at the $p < 0.05$ level.

III. Results

A. *General Effects of Trip Perturbation*

Fig. 5 shows the trip perturbation effects in the first trial for the cuing group and in the control trial (i.e., first trial) for the cuing group. The two-way ANOVA (group, period) showed no significant difference between the groups [$F(2,54) < 0.413$, $p > 0.523$] and no group x period interaction [$F(2,54) < 1.610$, $p > 0.209$]. The main effect of period (pre-, per-, and post-trip) was significant for maximum trunk flexion angle [$F(2,54) = 55.23$, $p < 0.0001$], maximum trunk flexion velocity [$F(2,54) = 34.14$, $p < 0.0001$], trunk flexion AD [$F(2,54) = 51.09$, $p <$

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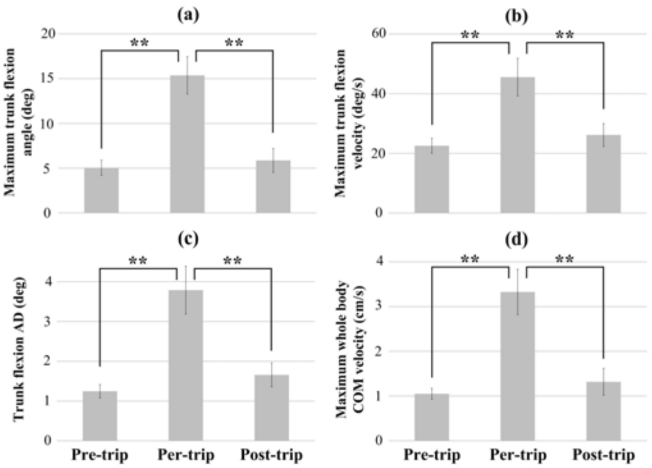

Fig. 5. Effects of the trip as a function of the period (pre-, per-, and post-trip) in the first trial for both groups (n=20). (a) Maximum trunk flexion angle. (b) Maximum trunk flexion velocity. (c) Trunk flexion AD. (d) Maximum whole body COM velocity. Error bars indicate standard error of the corresponding average (** p < 0.0001).

0.0001], and maximum whole body COM velocity [$F(2,54) = 42.50$, $p < 0.0001$]. Post hoc multiple comparisons showed that maximum trunk flexion angle ($p < 0.0001$), maximum trunk flexion velocity ($p < 0.0001$), trunk flexion AD ($p < 0.0001$), and maximum whole body COM velocity ($p < 0.0001$) were significantly greater during the trip period than the pre-trip and post-trip periods. However, these measures were not significantly different between the pre-trip and post-trip periods ($p > 0.287$), indicating that normal walking had resumed within two or three steps after the trip.

Fig. 6(a)-(b) which illustrate the perturbation of the gait cycle during a trip, superimposes GRF profiles for the first trip trial without and with cuing onto normal GRF profiles obtained during the pre-trip period (i.e., 5 steps before the instant of the trip). The superimposition shows that the gait cycle and GRF profiles were more variable for trip step(s) than for steps before the trip, and that recovery from perturbation in the first trip trial was more efficient with cuing

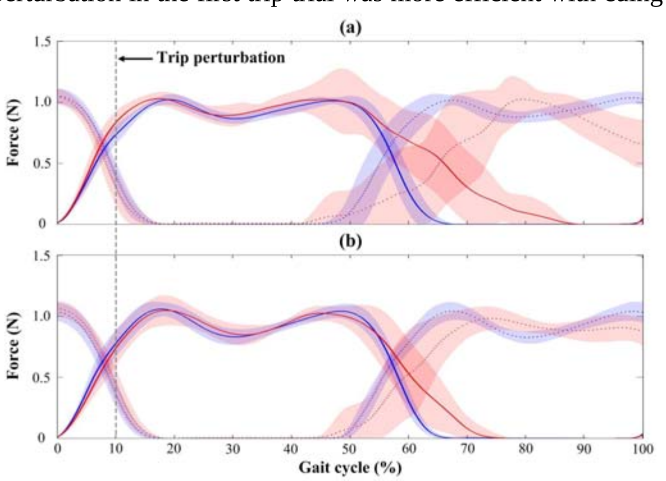

Fig. 6. Average GRF profiles for the first trip trial as a function of the group. (a) GRF profiles obtained from the adaptation group. (b) GRF profiles obtained from the cuing group. Blue and red lines indicate normal GRF profiles obtained from 5 steps before the instant of the trip (pre-trip) and perturbed GRF profiles, respectively. Solid and dashed lines correspond to GRF profiles for the left and right foot. Shaded areas represent the first standard deviation of the corresponding average.

TABLE I
STATISTICAL ANALYSIS RESULTS OF THE ADAPTATION GROUP FOR REPETITION (R). *STATISTICAL SIGNIFICANCE.

Dependent variable	Effects	DF	F Value	Pr>F
Response step time	R	7, 72	7.246	< 0.0001*
Maximum response step force	R	7, 72	8.046	< 0.0001*
Recovery time	R	7, 72	6.388	< 0.0001*
Maximum LF	R	7, 72	0.486	0.842
Maximum trunk flexion angle	R	7, 72	10.178	< 0.0001*
Maximum trunk flexion velocity	R	7, 72	6.317	< 0.0001*
Trunk flexion AD	R	7, 72	5.486	< 0.0001*
Maximum whole body COM velocity	R	7, 72	5.011	< 0.0001*

TABLE II
STATISTICAL ANALYSIS RESULTS OF THE CUING GROUP FOR VIBROTACTILE CUING (V), BODY LOCATION (L), AND LEAD TIME (T), AND THEIR INTERACTION (L X T). *STATISTICAL SIGNIFICANCE.

Dependent variable	Effects	DF	F Value	Pr>F
Response step time	V	2, 77	23.564	< 0.0001*
	L	2, 54	0.101	0.904
	T	1, 54	0.890	0.350
	L x T	2, 54	0.247	0.782
Maximum response step force	V	2, 77	17.672	< 0.0001*
	L	2, 54	0.259	0.773
	T	1, 54	1.500	0.226
	L x T	2, 54	0.457	0.636
Recovery time	V	2, 77	20.202	< 0.0001*
	L	2, 54	0.389	0.680
	T	1, 54	0.009	0.927
	L x T	2, 54	0.715	0.494
Maximum LF	V	2, 77	1.192	0.309
	L	2, 54	0.044	0.957
	T	1, 54	0.000	0.992
	L x T	2, 54	0.016	0.984
Maximum trunk flexion angle	V	2, 77	104.556	< 0.0001*
	L	2, 54	0.065	0.937
	T	1, 54	0.015	0.905
	L x T	2, 54	0.059	0.943
Maximum trunk flexion velocity	V	2, 77	12.933	< 0.0001*
	L	2, 54	0.310	0.735
	T	1, 54	0.160	0.691
	L x T	2, 54	0.213	0.809
Trunk flexion AD	V	2, 77	25.040	< 0.0001*
	L	2, 54	0.513	0.601
	T	1, 54	0.283	0.597
	L x T	2, 54	0.357	0.701
Maximum whole body COM velocity	V	2, 77	16.002	< 0.0001*
	L	2, 54	0.138	0.871
	T	1, 54	0.232	0.632
	L x T	2, 54	0.309	0.735

than without cuing as indicated by the reduced variability of the right foot heel strike force profile.

With the exception of peak LFs, the other seven outcome metrics showed that repeated exposure to trip perturbation and vibrotactile cuing before a trip significantly affected recovery performance. Tables I and II summarize the results of the statistical analysis for all dependent variables. Table I presents the main effects of repetition for the adaptation group, and

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Table II presents the main effects of vibrotactile cuing, body location, and lead time and their interactions (body location x lead time) for the cuing group.

B. *Adaptation Group*

Repetition of trip trials had a significant effect on the response step time ($p < 0.0001$), maximum response step force ($p < 0.0001$), and recovery time ($p < 0.0001$), as shown in Table I and Fig. 7(a)-(c). The post hoc analysis showed that the values of the response step time, maximum response step force, and recovery time were not significantly different for trials 1-4, decreased between trials 4-8, and were significantly smaller (27.47%, 23.41%, and 23.35%, respectively) for trial 8 than for trials 1-4.

The significant effects of repetition were observed for maximum trunk flexion angle and velocity, trunk flexion AD, and maximum whole body COM velocity, as shown in Table I and Fig. 8(a)-(d). The post hoc comparisons showed significantly smaller values (64.63%, 52.10%, 60.04%, 42.37%) of the maximum trunk flexion angle and velocity, trunk flexion AD, maximum whole body COM velocity, for trial 8 than for trials 1-4.

LFs applied to the safety harness were not significantly affected by repetition of trip trials, as shown in Fig. 7(d) and Table I. The average maximum LF was 3.3% of the body weight for trials 1-8.

C. *Cuing Group*

Insignificant differences between the control and post-control trials for the response step time ($p = 0.998$), maximum response step force ($p = 0.198$), and recovery time ($p = 0.669$) suggested that no adaptation occurred with the addition of cuing despite repeated tripping. However, there were significant differences between the control trials and cued trials for response step time, maximum response step force,

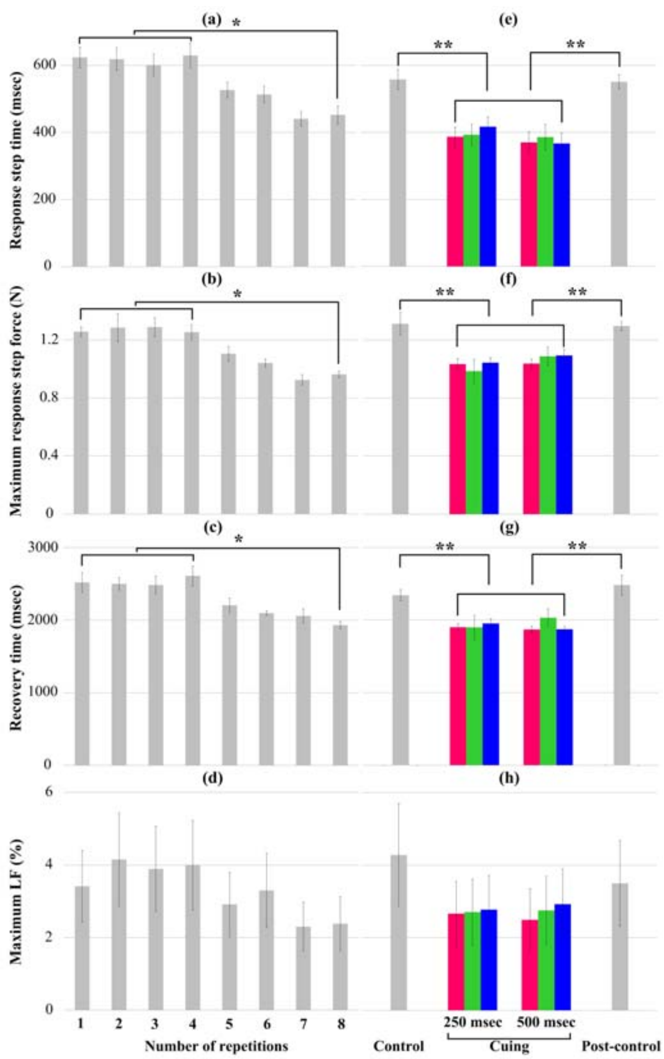

Fig. 7. Response step time, maximum response step force, recovery time, and maximum LF for the adaptive group ((a)-(d)) and cuing group ((e)-(h)). Red, green, and blue bars represent body locations for the application of vibrotactile cuing such as upper arm, trunk, and lower leg, respectively, as a function of the lead time (i.e., 250 ms and 500 ms). Gray bars correspond to the trial without vibrotactile cuing (control and post-control trials). Error bars indicate standard error of the corresponding average (** p < 0.0001 and * p < 0.05).

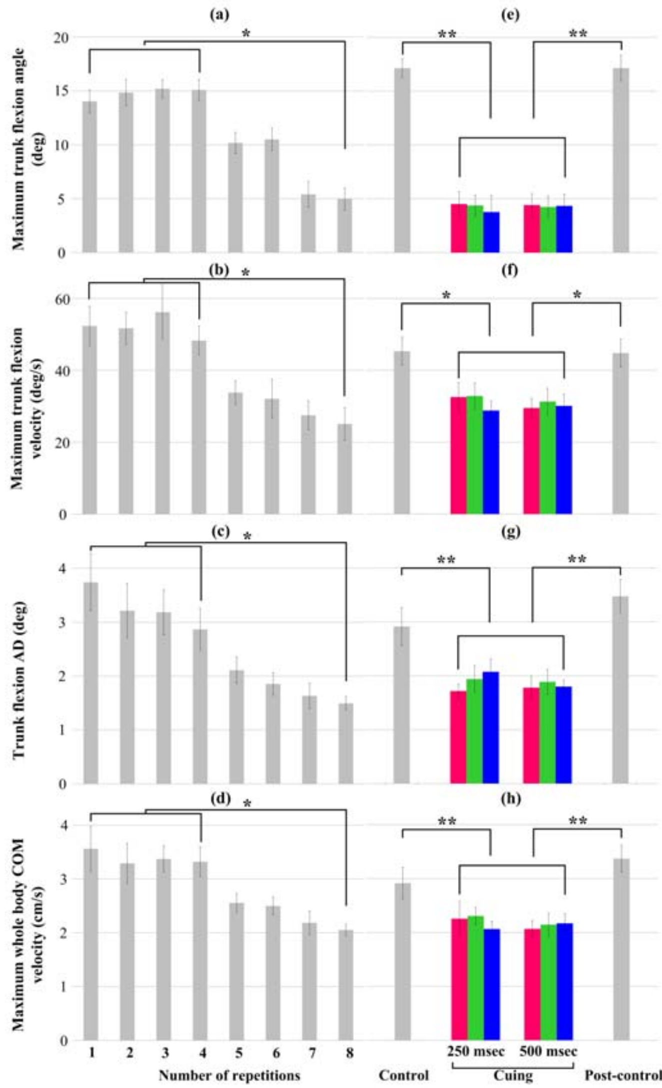

Fig. 8. Maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion AD, and maximum whole body COM velocity for the adaptation group ((a)-(d)) and cuing group ((e)-(h)). Red, green, and blue bars represent body locations for the application of vibrotactile cuing such as upper arm, trunk, and lower leg, respectively, as a function of the lead time (i.e., 250 ms and 500 ms). Gray bars correspond to the trial without vibrotactile cuing (control and post-control trials). Error bars indicate standard error of the corresponding average (** p < 0.0001 and * p < 0.05).

and recovery time as illustrated in Fig. 7(e)-(g). Response step time, maximum response step force, and recovery time decreased by 30.25%, 19.72%, and 20.38%, respectively, compared with the control trials. The post hoc analysis showed that the variables (i.e., response step time, maximum response step force, and recovery time) were not influenced by stimulus location, lead time, or by their interaction (i.e., stimulus location x lead time) in vibrotactile cuing trials as shown in Table II.

Fig. 8, right column, illustrates the average maximum trunk flexion angle and velocity, trunk flexion AD, and maximum whole body COM velocity. Significant reductions in the maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion AD, and maximum whole body COM velocity were observed with cuing. Fig. 8(e)-(h) show that average maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion AD, and maximum whole body COM velocity were significantly less (75.01%, 31.52%, 41.59%, and 31.02%, respectively) for cuing trials than for control and post-control trials. The post hoc comparisons showed that stimulus location, lead time, and their interactions (i.e., stimulus location x lead time) had no significant effects ($p > 0.9$) as shown in Table II. The post hoc analysis also showed no significant changes between control and post-control trials for the maximum whole body COM position (p = 0.918), maximum whole body COM velocity (p = 0.372), maximum trunk flexion angle (p = 1.000), maximum trunk flexion velocity (p = 0.999), and trunk flexion AD (p = 0.267).

No LFs applied to the safety harness were significantly affected by vibrotactile cuing, as shown in Fig. 7(h) and Table II. Although not significant, there was a slight decrease in the maximum LF with vibrotactile cuing. The average maximum LF was 3.90% of the body weight for trials without vibrotactile cuing and 2.72% for trials with vibrotactile cuing.

IV. Discussion

Two methods (adaptation, cuing) to facilitate recovery from unpredictable trips while walking at a self-selected pace were compared. Facilitation of trip recovery appeared to occur immediately with vibrotactile cuing, whereas repetition-induced adaptation needed at least 8 trials to achieve similar performance. This beneficial effect of vibrotactile cuing was observed in seven outcome metrics (response step time, maximum response step force, recovery time, maximum trunk flexion angle, maximum trunk flexion velocity, trunk flexion AD, and maximum whole body COM velocity).

The method for simulating trip perturbations using the programmable split-belt treadmill was validated by statistical analyses showing significant changes in outcome measures indicative of a trip after the trip event (Figs. 5 and 6). The kinematic effects, including forward body rotation and the generation of higher net support force during the stepping phase, are consistent with those induced by trip perturbations resulting from external mechanical obstacles [35, 37, 39]. The analyses also showed that the pre-trip steps and post-recovery steps were not significantly different (Fig. 5). Hence, normal walking resumed within one-to-three steps after a trip.

Recovery time [Fig. 7(c) and 7(g)], which is implicitly subordinate to behavioral characteristics described by the selected outcome measures, could be viewed as a summary indicator of the recovery performance, since all behavioral measures concurred for both the adaptation and cuing groups. The decreases in peak response step force and loading, trunk flexion angle and velocity, trunk flexion AD (variability), and COM velocity concurred with the decrease in recovery time when comparing trials 1-4 with trial 8 in the adaptation group and comparing control trials (i.e., control and post-control trial) with trials with cuing (see Figs. 7 and 8). Hence, the behavioral changes associated with loss of balance recovery [35, 37, 39] suggest that both better control of trunk movements and most likely trunk-leg coordination resulting from compensatory reactions reduce the perturbed (per-trip) period. In other words, adaptation to repeated trip exposure and presence of cuing enable a reduction in trunk flexion, whose excess (> 45°) at the time of recovery foot contact contributes to falling [37].

The results also showed significantly less response step time when vibrotactile cuing was available prior to a trip perturbation than without vibrotactile cuing. There is speculation that vibrotactile cuing in fact facilitates a stepping strategy in response to external postural perturbation [35, 37, 39]. Similarly, Asseman et al. [57] have found that post-trip vibrotactile cuing applied about 60 ms after trip initiation (abrupt backwards translation of a support surface while standing) reduced stepping reaction times in healthy older adults, although healthy young adults and patients with vestibular deficit and with peripheral neuropathy did not benefit from vibrotactile cuing. If the limited efficacy of post-trip vibrotactile cuing could be enhanced by prior information about the motor action to be executed (a.k.a. precuing [58, 59]), the vibrotactile cuing provided prior to a trip perturbation in the present study could remove time uncertainty because it indicates the imminence of the perturbation which in turn facilitates trip recovery.

Consistent with findings in prior studies [35-42], the improved recovery observed in the adaptation group demonstrated the effectiveness of adaptive training for fall recovery strategies. The results in Figs. 7 and 8 clearly show that while the adaptive response developed progressively over trials 1-8 when learning by repetition, the adaptive response appeared in trial 1 as soon as vibrotactile cuing was provided. Notably, the immediate compensatory response did not vary with the location of application of the vibrotactile cuing or with the cuing lead time. Regardless of the location of vibrotactile cuing, 250 ms of lead time was sufficient for the central nervous system (CNS) to receive sensory data from any location on the body and modify the motor action. Hence, both the immediate efficiency and robustness of the very short lead time suggest that vibrotactile cuing can be effectively applied to any body segment immediately before a fall occurs.

V. Conclusion

This study investigated the effect of vibrotactile cuing as a function of location of the cuing stimulus and lead time on recovery strategies from a treadmill-induced trip. The experimental results demonstrated that vibrotactile cuing improves recovery from trip perturbation more efficiently than adaptation based on learning. Vibrotactile cuing may not significantly interfere with the visual, auditory, muscle proprioceptive, and vestibular information used in real-life walking and other activities. The independence and consistency of performance regardless of the location of the cuing stimulus and lead time support the use of vibrotactile cuing to inform future designs of fall prevention technologies. Moreover, positioning tactors within the vicinity of sensors may offer advantages in terms of cost, size, flexibility, and accessibility.

While the recent advances in sensor technologies allow shoe-based systems to detect ground-level obstacles [27, 28, 30] and ground conditions [29], the challenge is to accurately predict fall incidents (e.g., trips and slips) by guaranteeing low false-alarm rates. Our future research will focus on the design of a vibrotactile cuing scheme with varying intensity and frequency with respect to the proximity and direction of upcoming hazards (e.g., obstacles, groves, pits and holes, and characteristics of ground) that provides navigational guidance for hazard avoidance. To further understand how false alarms negatively change body kinetics and kinematics for individuals at a high risk of falling (e.g., older adults), our research will include a false vibrotactile cuing condition that provides vibrotactile cuing without the trip perturbation in order to investigate the effects of true and false vibrotactile cuing on older adults' recovery performance and postural behaviors. Application, especially to older adults at a high risk of falling, should have positive impacts on walking and daily activities. Our eventual goal is to design a wearable system that uses vibrotactile stimulation to warn of potential falls in advance.

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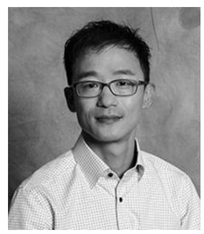

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