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Biomechanical response to mediolateral foot-placement perturbations during walking

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ABSTRACT

Dynamic balance in the frontal plane requires active control, which is accomplished largely through control of mediolateral foot placement. Individuals without mobility impairments have the ability to compensate for variability in foot-placement to maintain their balance; however, it is unknown how individuals respond to unexpected mediolateral perturbations to their foot placement that alter their balance control. The purpose of this study was to identify the biomechanical responses of individuals without mobility impairments to medial and lateral foot-placement perturbations during walking. Three-dimensional body segment kinematic and ground reaction force data were collected from 15 participants at 1.0 m/s and their self-selected speed on an instrumented treadmill. Dynamic balance was assessed by analyzing whole-body angular momentum in the frontal plane. We hypothesized that participants would respond to the perturbations with a combination of a lateral ankle strategy, hip adduction strategy and/or ankle push-off strategy to restore their balance. Overall, the medial perturbations adversely affected dynamic balance while lateral perturbations had little effect. Individuals responded to medial (lateral) perturbations with an increased (decreased) ankle inversion moment, which correlated to lateral (medial) shifts in their foot center of pressure. In addition, individuals responded to medial (lateral) perturbations with a decreased (slightly decreased) hip abduction moment. Contrary to our hypothesis, we did not observe an ankle push-off moment response but rather, a small response in the opposite direction. These results highlight the response of individuals without mobility impairments to unexpected foot-placement perturbations and provide a basis of comparison for those with impaired balance control.

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1. Introduction

The ability to control and maintain dynamic balance is critical to participating in activities of daily living. A fear of falling is linked to reduced activity levels, depression and anxiety among older adults (Painter et al., 2012). Previous work has shown that dynamic balance is maintained passively in the sagittal plane, but active control is required to maintain balance in the frontal plane, largely through control of foot placement (e.g., Bauby and Kuo, 2000; MacKinnon and Winter, 1993). A number of studies have focused on the role of the swing leg gluteus medius in foot placement (e.g., Dean and Kautz, 2015; Rankin et al., 2014). However, a recent simulation study demonstrated that both swing and stance leg

muscles are utilized to control foot placement (Roelker et al., 2019), highlighting the complex muscle coordination needed from both legs to control balance.

Foot placement has an important role in regulating whole-body angular momentum (H), which is an effective measure used to assess dynamic balance (Neptune and Vistamehr, 2019). In the frontal plane, foot placement affects both the mediolateral (ML) and vertical moment arms through which the ground reaction forces (GRFs) produce an external moment about the center of mass, which is equal to the time rate of change of H . Frontal-plane H is tightly regulated during unimpaired walking (Herr and Popovic, 2008), but has a higher range (Nott et al., 2014; Vistamehr et al., 2016) and is less tightly regulated during single-leg stance among individuals with impaired balance control (Nott et al., 2014). Unlike clinical balance measures, H provides insights into the underlying biomechanical factors influencing dynamic balance, including foot-placement, GRFs and body segment motion (Neptune and Vistamehr, 2019).

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Variations in ML foot placement can lead to a loss of balance, but individuals without mobility impairments have the ability to compensate for altered foot-placement to maintain their balance. A combination of strategies can be used depending on the timing and severity of the balance perturbation, including stance leg lateral ankle, hip and ankle push-off strategies (Reimann et al., 2018a). The lateral ankle strategy acts to shift the center of pressure (COP) location quickly to correct small errors in foot placement (Hof et al., 2007). Shifting the COP provided the fastest response following a visual perturbation (Reimann et al., 2018b), while foot placement could only be altered on the subsequent step. Using a passive prosthesis, individuals with lower-limb amputations cannot generate active ankle moments to shift their COP shift in response to foot-placement perturbations. As a result, amputees often use a hip strategy during single-leg stance to modulate their GRFs to maintain balance after a foot placement perturbation (Miller et al., 2018; Segal et al., 2015). Similarly, individuals use a hip strategy to maintain balance when they are prevented from using a COP shift (Otten, 1999). Others have shown during steady-state walking, even when a lateral ankle strategy is possible, the hip strategy still plays a dominant role in frontal-plane balance control (MacKinnon and Winter, 1993; Winter, 1995).

In addition to lateral ankle and hip strategies, an ankle push-off strategy can also be used to control balance (Reimann et al., 2018b). The ankle plantarflexors are important contributors to controlling frontal-plane balance during steady-state walking through their contributions to the GRFs (Neptune and McGowan, 2016). Impaired plantarflexor coordination is a predictor of poor balance control among individuals post-stroke (Brough et al., 2019). In response to visual mediolateral perturbations, an ankle push-off strategy has been observed to help restore balance (Reimann et al., 2018b). However, it is unclear if such a strategy would be used following foot-placement perturbations because of the altered interactions between foot placement and GRF generation by the plantarflexors. For example, the lateral moment arm of the vertical GRF would be decreased with a medial foot placement perturbation, thus reducing its potential to help control frontal plane H .

The purpose of this study was to identify the biomechanical responses of individuals without mobility impairments to medial and lateral foot-placement perturbations during walking. The perturbations were generated using a custom pneumatic device which moved the foot medially or laterally just prior to heel strike. We hypothesized that on the perturbed leg, individuals would compensate for medial (lateral) foot-placement perturbations with (1) a lateral (medial) COP shift, (2) a decreased (increased) hip abduction moment impulse, and/or (3) an increased (decreased)

ankle plantarflexion moment impulse. We hypothesized that these responses would occur during single-leg stance. Characterizing responses to foot placement perturbations in individuals without mobility impairments can provide a basis of comparison for those with neurological deficits and impaired balance control.

2. Methods

2.1. Data collection

Fifteen young adults without mobility impairments gave informed consent to an IRB-approved protocol (Table 1). To determine their self-selected over-ground walking speed, participants performed 3 trials of a 10-meter walk test at their “comfortable, typical walking speed.” Kinematic data were collected at 120 Hz using a 10-camera motion capture system (Vicon, Oxford, UK) and a full-body set of 65 reflective markers. Kinetic data were collected at 960 Hz from a split-belt instrumented treadmill (Motek, Amsterdam, Netherlands). Participants performed ten 30–45s walking trials at a standard speed of 1.0 m/s and ten trials at their self-selected over-ground walking speed (20 trials total). Ten trials (5 at each speed) included foot-placement perturbations. During the perturbation trials, two medial and two lateral perturbations were applied to random steps, resulting in 20 medial perturbations and 20 lateral perturbations for each subject. All trial conditions (speeds and perturbations) were randomized.

2.2. Perturbations

Foot-placement perturbations were performed using a custom pneumatic device (Segal and Klute, 2014). A compressed air tank was connected to the ankle via flexible tubes (Fig. 1). An inertial measurement unit and microprocessor (Sparkfun, Niwot, CO) were used to identify gait events. Based on the average cadence of 10 unperturbed steps, the microprocessor triggered solenoid valves (ASCO) to release compressed air 140 ms prior to the expected timing of the perturbed heel strike. Air exited through elbow joints for 180 ms, or until after heel strike occurred, producing a medial or lateral force of ~15 N on the ankle to perturb foot-placement. The original system was modified with an additional valve so the participants were unaware of the direction or timing of the perturbations.

2.3. Data analysis

Marker and force plate signals were low-pass filtered at 6 Hz and 15 Hz, respectively. A 13-segment inverse dynamics model

Table 1
Participant demographics and self-selected walking speeds.

Subject	Height (cm)	Mass (kg)	Age (years)	Gender	Self-Selected Speed (m/s)
1	161.0	58.8	24	F	1.28
2	166.0	62.4	25	F	1.49
3	166.0	56.9	25	F	1.48
4	180.0	75.3	27	M	1.38
5	171.5	60.8	28	F	1.50
6	188.0	83.0	27	M	1.46
7	179.5	68.7	28	F	1.20
8	191.0	67.7	26	M	1.14
9	184.0	71.5	26	M	1.21
10	183.0	80.4	26	M	1.31
11	157.0	46.9	20	F	1.24
12	189.0	78.4	32	M	1.19
13	161.5	51.9	19	F	1.32
14	178.5	64.1	21	F	1.38
15	174.0	72.8	22	F	1.40
Average (SD)	175.3 (11.0)	66.6 (10.6)	25.1 (3.4)		1.33 (0.12)

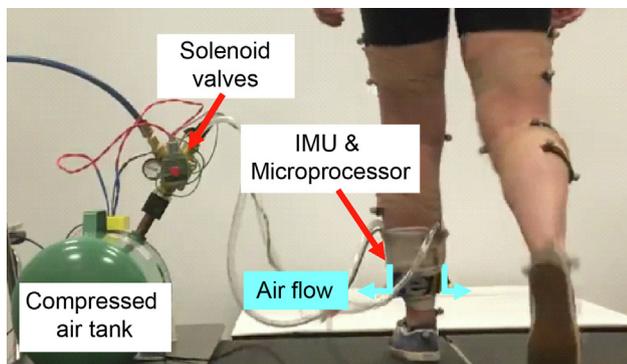


Fig. 1. The perturbation device consisted of a compressed air tank with solenoid valves, flexible hoses and elbow joints that released air to produce a medial or lateral force on the ankle.

was created for each subject (Visual 3D, C-Motion, Germantown, MD). Net internal joint moments generated by muscles were calculated using inverse dynamics. H was calculated by summing the angular momentum of each body segment about the whole-body center of mass. GRF impulses were calculated by integrating the GRF signals over stance and normalizing by body weight. Moment impulses were calculated by integrating the joint moments over stance and within four regions of stance (first double support, early ipsilateral single-leg stance, late ipsilateral single-leg stance, second double support). Joint moments were normalized by subject mass and H was normalized by subject mass, walking speed and leg length. Perturbation distance was defined as the peak divergence of the heel marker from the average unperturbed heel trajectory in the mediolateral direction following heel strike. COP excursion was calculated as the maximum lateral difference between the heel marker and COP location over stance in the lab coordinate frame. Crossover steps were identified and removed from kinetic analyses. Kinetic and kinematic measures exceeding three standard deviations of each subject's average trajectory during that condition were considered outliers and excluded from further analysis.

Differences between perturbed and unperturbed steps were evaluated using linear mixed effects models. Separate models were created for medial and lateral perturbations throughout stance and each region of stance. Gait cycles began at the perturbed-side heel strike immediately following the perturbation. After confirming there were no significant differences in outcome measures, self-

selected and standardized walking speeds were pooled for statistical analyses. Fixed effects were perturbation condition (perturbed or unperturbed) and random effects were study subjects. Outcome measures included frontal-plane range of H (H_R), ankle inversion moment impulse, ankle plantarflexion moment impulse, hip abduction moment impulse, COP excursion and ground reaction force moment impulses. A linear mixed effects model was also created to test for correlation between ankle inversion moment impulse and COP excursion. For this model, fixed effects were inversion moment and random effects were study subjects. Statistical analyses were performed in MATLAB (Mathworks, Natick, MA).

3. Results

3.1. Perturbation effect

Relative to the average unperturbed heel trajectory, perturbations caused an average of 3.6 ± 2.8 cm more medial and 3.6 ± 2.2 cm more lateral foot-placement compared to average unperturbed foot placement (Fig. 2).

3.2. Dynamic balance

H_R was 0.026 higher after medial perturbations ($p < .001$) and 0.002 lower after lateral perturbations ($p = .002$) compared to an unperturbed value of 0.048 (non-dimensional units; Fig. 3). See Appendix Table A1 for results of all dependent measures.

3.3. Ground reaction forces

Over stance, the ML GRF impulse decreased by an average of 0.58% BW-s after medial perturbations and increased by 0.92% BW-s after lateral perturbations compared to 3.30% BW-s for unperturbed ML GRF impulse ($p < .001$ for both), while the vertical GRF impulse was not significantly different for any conditions (Fig. 4).

3.4. Lateral ankle strategy

Over stance, the total perturbed ankle inversion moment impulse increased by an average of 0.039 N-m-s/BW after medial perturbations and decreased by 0.032 N-m-s/BW after lateral perturbations compared to -0.001 N-m-s/BW for unperturbed moment impulse. Inversion moment impulses for medially and lat-

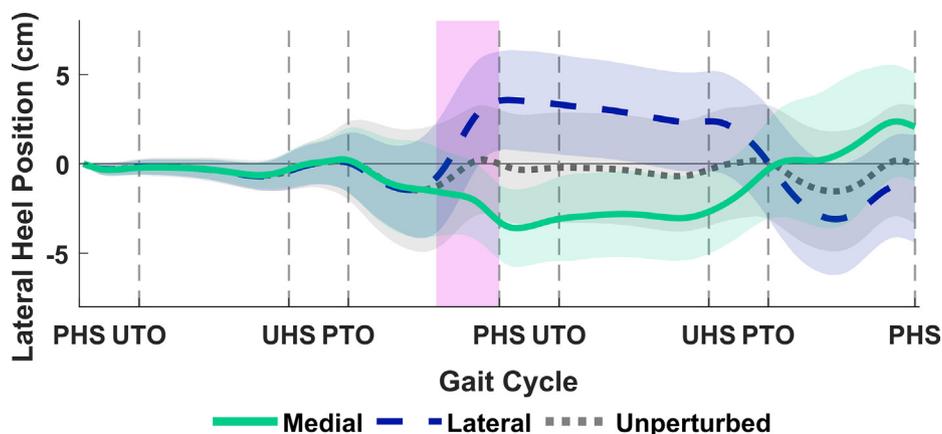


Fig. 2. Average perturbed-side (left leg) lateral (+) and medial (-) heel marker position relative to unperturbed heel trajectory ± 1 SD before and after medially perturbed, laterally perturbed and unperturbed steps. The vertical shaded region indicates perturbation duration. Vertical dashed lines indicate perturbed-side heel strike (PHS) and toe-off (PTO) and unperturbed-side heel-strike (UHS) and toe-off (UTO).

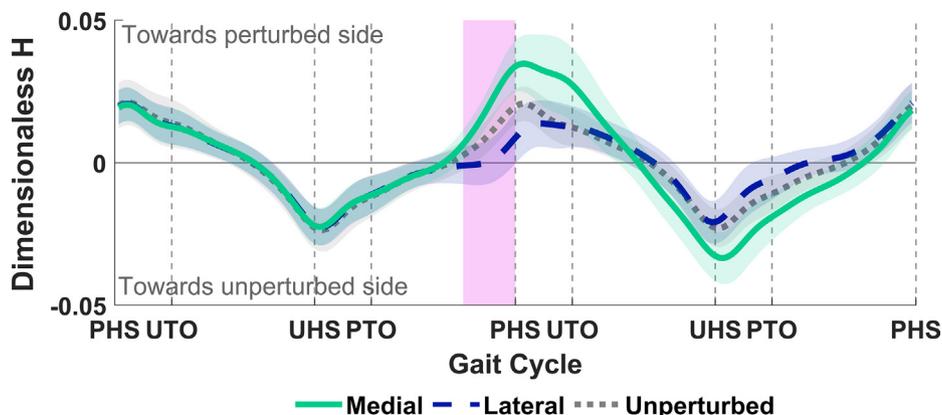


Fig. 3. Average frontal-plane $H \pm 1$ SD during the gait cycle before and after medially perturbed, laterally perturbed and unperturbed steps. The vertical shaded region indicates perturbation duration. Vertical dashed lines indicate perturbed-side heel strike (PHS) and toe-off (PTO) and unperturbed-side heel-strike (UHS) and toe-off (UTO).

erally perturbed steps were significantly different for all four regions of stance compared to unperturbed walking (all $p < .01$) (Fig. 5A). COP excursion was 1.9 cm more lateral after medial perturbations and 2.3 cm more medial after lateral perturbations compared to 0.6 cm for unperturbed steps ($p < .001$ for both) (Fig. 5B). The increases in ankle inversion moment were correlated to greater lateral COP excursion ($p < .001$, $R^2 = 0.79$) (Fig. A1).

3.5. Hip strategy

Over stance, the perturbed hip abduction moment impulse decreased by 0.07 N-m-s/kg after medial perturbations and decreased by 0.04 N-m-s/BW after lateral perturbations compared to unperturbed walking at 0.43 N-m-s/BW ($p < .001$ for both). Compared to unperturbed gait regions, the hip abduction moment impulse decreased during single-leg stance and second double support after medial perturbations ($p < .001$ for all) and decreased slightly in all regions after lateral perturbations ($p < 0.05$ for all) (Fig. 6A).

3.6. Ankle push-off strategy

The ankle plantarflexion moment impulse decreased by an average of 0.035 N-m-s/BW after medial perturbations ($p < .001$) and decreased slightly by 0.016 N-m-s/BW after lateral perturbations ($p = .05$) during stance compared to unperturbed walking at 0.440 N-m-s/BW (Fig. 6B). After lateral perturbations, plantarflexion moment impulses were slightly higher after lateral perturbations during the first two regions of stance and slightly lower for the second two ($p < 0.05$ for all) and lower after medial perturbations during second double support ($p < .001$) (Fig. 6B).

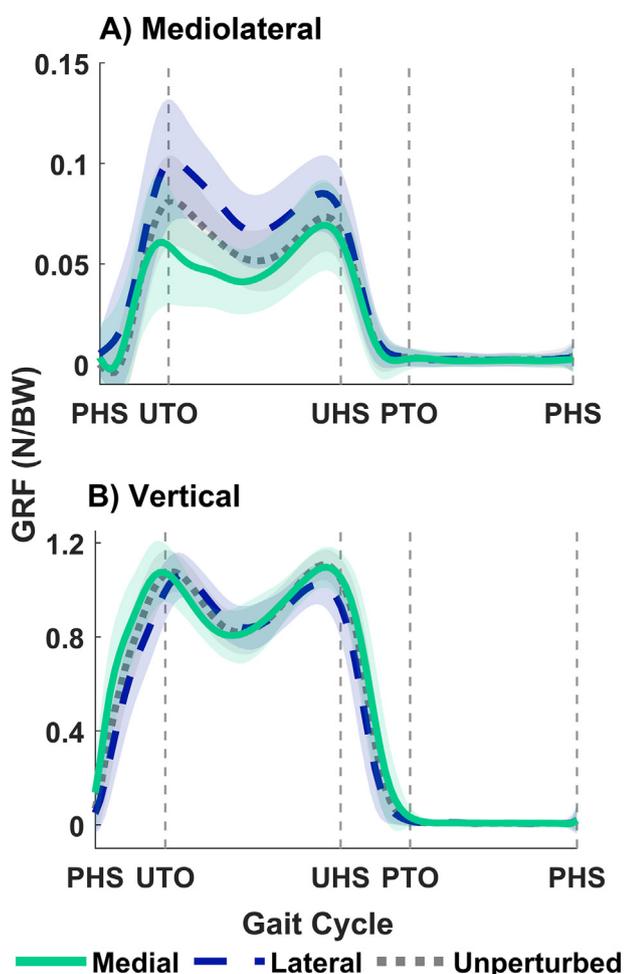


Fig. 4. Average perturbed-side (A) medial and (B) vertical GRFs ± 1 SD during the gait cycle after medially perturbed, laterally perturbed and unperturbed steps. Vertical dashed lines indicate perturbed-side heel strike (PHS) and toe-off (PTO) and unperturbed-side heel-strike (UHS) and toe-off (UTO).

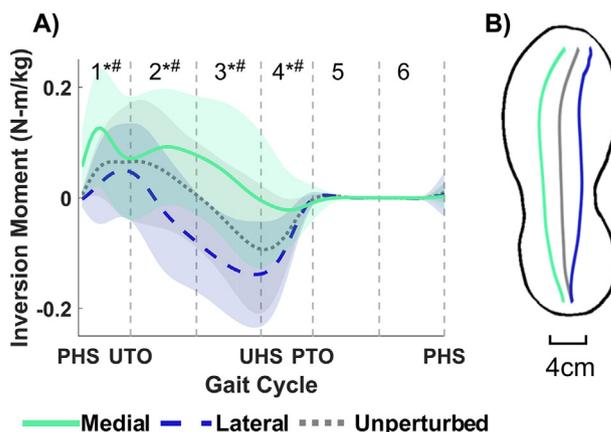


Fig. 5. (A) Average stance leg ankle inversion (+) and eversion (-) moment ± 1 SD for medially perturbed, laterally perturbed and unperturbed steps. Vertical lines indicate the regions of the gait cycle. ** denotes significance for medial perturbations and # denotes significance for lateral perturbations for moment impulse within each gait phase. Vertical dashed lines indicate perturbed-side heel strike (PHS) and toe-off (PTO) and unperturbed-side heel-strike (UHS) and toe-off (UTO). (B) Average COP excursion relative to the heel marker for each perturbation condition.

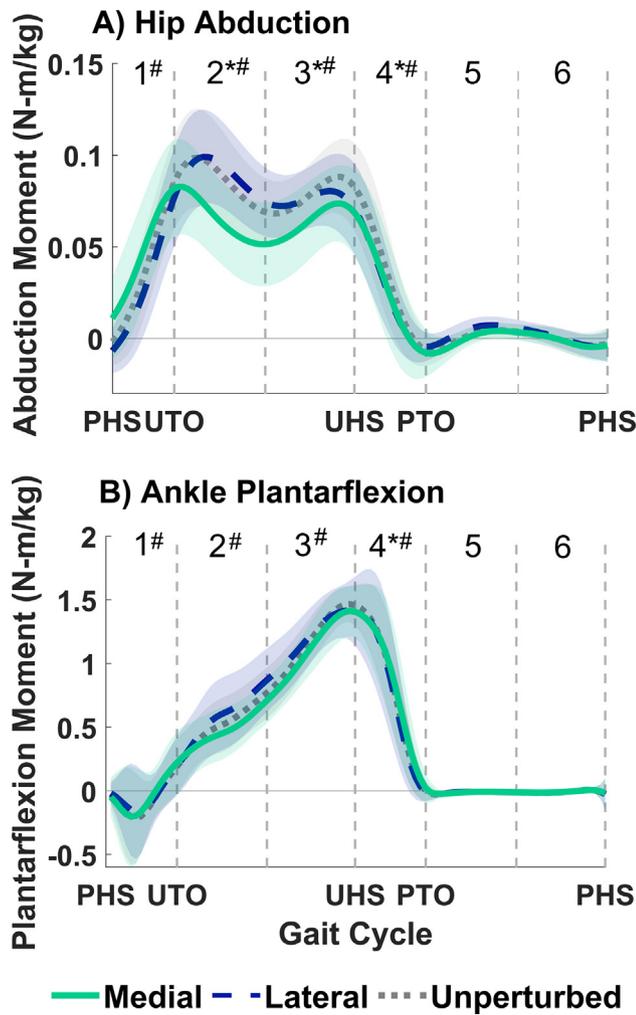


Fig. 6. (A) Average hip abduction (+) and adduction (-) moment and (B) average ankle plantarflexion (+) and dorsiflexion (-) moment \pm 1 SD for medially perturbed, laterally perturbed and unperturbed steps. Vertical lines indicate regions of the gait cycle. "*" denotes significance for medial perturbations and "#" denotes significance for lateral perturbations for moment impulse within each gait phase. Vertical dashed lines indicate perturbed-side heel strike (PHS) and toe-off (PTO) and unperturbed-side heel-strike (UHS) and toe-off (UTO).

4. Discussion

The purpose of this study was to identify the biomechanical responses of individuals without mobility impairments to medial and lateral foot-placement perturbations during walking. In agreement with our hypothesis, participants compensated for medial (lateral) perturbations with a lateral (medial) COP shift and decreased hip abduction moment impulse after medial perturbations. However, contrary to our hypothesis the ankle plantarflexion moment impulse decreased slightly after medial perturbations and hip abduction moment impulse decreased slightly after lateral perturbations. While balance responses primarily occurred during single-leg stance as hypothesized, other small changes in ankle and hip responses occurred during first and second double support.

4.1. Perturbation effect on dynamic balance

Frontal-plane H_R increased (decreased) after medial (lateral) perturbations, which was consistent with previous work (Miller et al., 2018). The time rate of change of H (\dot{H}) is equal to the sum of the external moments acting about the body's center of mass

(COM). During gait, these external moments are generated by the GRFs acting at a perpendicular distance away from the COM (i.e., the moment arm) and can be modulated by adjusting ML and vertical GRF magnitude and moment arms (Fig. 7). During steady-state walking, the external moments produced by each leg counteract the H produced by the opposite leg and rotate the body towards the contralateral side, keeping the integral of H over the gait cycle close to zero to prevent a body lean. Thus, a change in H caused by a foot placement perturbation requires altered balance control to compensate for increases or decreases to net H .

Higher H_R levels are indicative of balance deficits (Vistamehr et al., 2016), thus the increase in H_R after medial perturbations suggests a disruption to dynamic balance. Compared to unperturbed walking, more medial foot placement reduces the ML distance between the COP and COM, resulting in a shorter ML moment arm. The shortened ML moment arm may cause an increase in H_R by reducing the potential of the vertical GRF to counteract the momentum generated by the unperturbed leg. In response, participants used a lateral ankle strategy to lengthen the ML moment arm and a hip abduction strategy to reduce the external moment towards the perturbed side generated by the ML GRF.

In contrast, H_R decreased after the lateral foot placement perturbations. Lower values of H_R are associated with better clinical balance scores (Vistamehr et al., 2016) and the wider base of support created by the lateral perturbation was unlikely to challenge balance. However, the change in H still required a response to keep net H over the gait cycle close to zero. Previous work suggested that a decrease in H_R after lateral foot-placement perturbations was caused by an increase in perturbed leg ML GRF (Miller et al., 2018). While we also observed an increased perturbed-side ML GRF after lateral perturbations (Fig. 4A), the decrease in H occurred before heel strike and H did not continue to decrease during stance

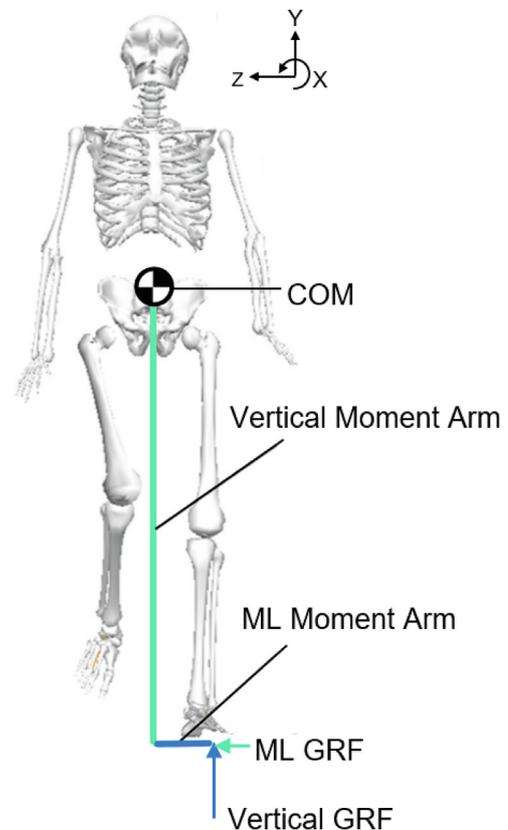


Fig. 7. Vertical and mediolateral (ML) forces and moment arms contributing to frontal-plane H about the center of mass (COM). Adapted from Miller et al. (2018).

(Fig. 3), indicating that the perturbation itself caused H to decrease rather than the increased ML GRF. Lateral perturbations also caused wider steps, which are associated with higher external moments and thus higher H_R (Nott et al., 2014) and are used by individuals at a greater risk of falling (e.g., Dean et al., 2007; Frame et al., 2020). Thus, a lateral perturbation would likely cause H_R to increase during the subsequent step unless active control occurred. However, participants were able to maintain low levels of H_R to control their balance in part by using a lateral ankle strategy.

4.2. Lateral ankle strategy

Studies have shown individuals may shift the location of the COP to compensate for a step that is too medial or lateral (Hof et al., 2010, 2007; Segal et al., 2015) in order to correct the ML moment arm of the vertical GRF. This COP shift can be accomplished via an ankle inversion or eversion moment (Segal et al., 2015). Shifting the COP medially (laterally) shortens (lengthens) the ML moment arm, thus decreasing (increasing) the external moment produced by the perturbed-side vertical GRF. As expected, subjects responded to medial (lateral) perturbations with an increased (decreased) ankle inversion moment impulse over stance and a correlated COP shift opposite the perturbation direction. Small but significant ankle inversion moment responses occurred immediately after heel strike during double support. This early reaction despite the neural and electromechanical delays in muscular response to balance disruptions (Pijnappels et al., 2005) indicates that a reflex response occurred or that participants initiated an active response during the perturbation.

4.3. Hip strategy

COP shifts produced by the lateral ankle strategy are limited by the surface area of the foot. Thus, a hip strategy was also necessary after medial perturbations to restore pre-perturbation gait patterns. Lower hip abduction moments correlated with higher frontal plane H_R in previous work (Silverman et al., 2012), which suggested that increasing hip abduction moment might be an effective strategy to reduce H_R . In contrast, these participants decreased their hip abduction moments during single-leg stance.

This reduction in abduction moment counteracts the increased H_R following medial perturbations. Hip abductor muscles produce medially directed GRFs that rotate the body towards ipsilateral side in early and late stance (Neptune and McGowan, 2016). When the shortened ML moment arm reduces the H generated by the vertical GRF towards the contralateral side, decreasing the ML GRFs would counteract that change by decreasing H in the opposite direction (Fig. 7). Because the hip abductors are the primary contributors to the ML GRF during single-leg stance (John et al., 2012), reducing the hip abduction moment could accomplish this decrease in ML GRFs. Indeed, we observed corresponding decreases in hip abduction moments and ML GRFs after medial perturbations (Fig. 4A).

Lateral perturbations also produced a slight decrease in hip abduction moment throughout double support and single-leg stance (Fig. 6A). However, the ML GRF increased despite the decrease in hip abduction moment (Fig. 4), possibly due to the perturbation force or an unmeasured response. The small change opposite the direction hypothesized suggests that the lateral perturbation did not challenge balance to the extent of the medial perturbation. Previous work reported an increase in positive and decrease in negative hip abduction work after lateral perturbations and the opposite for medial perturbations (Miller et al., 2018), suggesting that a measure of work may identify some responses that were not clear in the moment impulse.

4.4. Ankle push-off strategy

Modifying the vertical GRF could be an additional strategy to modulate H . The ankle plantarflexors are primary contributors to vertical GRFs in late stance (Neptune et al., 2001; Anderson and Pandy, 2003) and frontal plane H (Neptune and McGowan, 2016). Thus, we hypothesized that ankle plantarflexor moments would be used to adjust vertical GRFs and compensate for altered ML moment arms after perturbations. To overcome the shorter (longer) moment arm after medial (lateral) perturbations, we expected to see increased (decreased) ankle plantarflexion moments to increase (decrease) the vertical GRF. However, the opposite occurred after medial perturbations and minimal decreases were observed after lateral perturbations. Because there were no changes to the vertical GRFs and changes in the ankle plantarflexion moment were small (Fig. 6B), we suspect that they occurred as a byproduct of the lateral ankle strategy: i.e., the ankle everter and inverter muscles used to accomplish the COP shifts also have plantarflexion moment arms, and vice versa for ankle plantarflexor muscles (Lee and Piazza, 2008). Thus, we do not believe that an ankle plantarflexor strategy was intentionally used to respond to these perturbations. Unlike the ankle push-off strategy used to recover from frontal plane visual perturbations (Reimann et al., 2018b) and trips in the anterior direction (Pijnappels et al., 2005), an ankle push-off strategy may not be an efficient way to recover from a foot placement perturbation due to the relatively small moment arm and high force of the vertical GRF. Moreover, changing ankle push-off moments could interfere with anterior-posterior balance control. Future modeling work will investigate individual muscle contributions to the observed perturbation responses.

A potential limitation to this study was that the perturbation was intended to produce an imposed error in foot placement, which would subsequently require a balance response. However, based on the measured H values, the perturbation itself affected dynamic balance prior to foot placement. Moreover, because EMG data were not collected and causal relationships were not analyzed, we could not determine whether some changes to ground reaction forces and H were caused by the perturbation itself or by an active balance response. In future studies, modeling and simulation techniques should be used to determine individual muscle contributions to these biomechanical responses. Another limitation was that a learning effect could occur throughout data collection. In a post-hoc test, linear mixed effects models were created for medial and lateral perturbations to evaluate the relationship between H_R and step sequence. A group effect was found for medial perturbations ($p=.04$) and separate linear regressions were created for each subject, with four of fifteen participants demonstrating learning effects ($p < .05$). There were no group or individual learning effects for lateral perturbations. Because the effect was small and the majority of the subjects had no learning effect, this likely had little impact on our results and conclusions. Finally, medial perturbations were more likely to cause a crossover step. However, due to the large number of steps that were analyzed (225 medial and 279 lateral), we do not believe the removal of crossover steps affected our primary conclusions.

In summary, participants used both hip and ankle strategies to control H in response to medial and lateral foot placement perturbations. Medial perturbations caused H_R to increase, which produced a lateral ankle and hip response. Lateral perturbations caused H_R to decrease and primarily produced a lateral ankle response. Medial perturbations did not produce the expected ankle push-off response, instead producing a slight response in the opposite direction. Together, these results highlight the complex responses individuals without mobility impairments use to recover from foot placement perturbations and can provide a baseline for

Table A1Results of dependent measures ± 1 SD. *** indicates a significant difference between medially or laterally perturbed steps and unperturbed steps.

		Unperturbed	Medially Perturbed		Laterally perturbed	
H_R (Dimensionless)	Stance	0.048 \pm 0.009	0.074 \pm 0.013	* p < .001	0.045 \pm 0.010	* p < .001
COP Excursion (cm)	Stance	0.65 \pm 0.52	2.58 \pm 1.10	* p < .001	-1.62 \pm 0.77	* p < .001
ML GRF impulse (% BW-s)	Stance	3.30 \pm 0.68	2.71 \pm 0.68	* p < .001	4.21 \pm 0.83	* p < .001
Vertical GRF impulse (% BW-s)	Stance	50.75 \pm 1.89	51.32 \pm 3.72		50.52 \pm 2.94	
Ankle inversion moment impulse (N-m-s/kg $\times 10^2$)	Stance	-0.08 \pm 5.71	3.80 \pm 6.13	* p < .001	-3.30 \pm 4.81	* p < .001
	Phase 1	0.86 \pm 0.93	1.46 \pm 0.91	* p < .001	0.48 \pm 0.99	* p < .001
	Phase 2	0.94 \pm 2.14	1.78 \pm 2.31	* p < .001	-0.34 \pm 1.92	* p < .001
	Phase 3	-0.89 \pm 2.20	0.90 \pm 2.57	* p < .001	-2.11 \pm 1.66	* p < .001
	Phase 4	-1.01 \pm 1.00	-0.25 \pm 1.04	* p < .001	-1.47 \pm 0.95	* p < .001
Hip abduction moment impulse (N-m-s/kg $\times 10^2$)	Stance	43.03 \pm 10.69	36.51 \pm 8.71	* p < .001	39.25 \pm 9.00	* p < .001
	Phase 1	5.15 \pm 1.84	5.13 \pm 2.27		3.95 \pm 1.59	* p < .001
	Phase 2	18.05 \pm 4.40	15.13 \pm 3.34	* p < .001	17.26 \pm 3.77	* p < .05
	Phase 3	15.91 \pm 3.68	13.85 \pm 3.07	* p < .001	14.10 \pm 3.02	* p < .001
	Phase 4	6.11 \pm 2.38	4.38 \pm 2.91	* p < .001	5.32 \pm 2.01	* p < .01
Ankle plantarflexion moment impulse (N-m-s/kg $\times 10^2$)	Stance	43.92 \pm 4.99	40.41 \pm 4.47	* p < .001	42.27 \pm 5.72	* p < 0.05
	Phase 1	-1.54 \pm 0.61	-0.81 \pm 3.32		-0.98 \pm 0.99	* p = .001
	Phase 2	10.25 \pm 2.46	9.69 \pm 2.15		11.64 \pm 1.92	* p < .001
	Phase 3	23.58 \pm 2.92	22.37 \pm 4.80		21.20 \pm 1.66	* p < .001
	Phase 4	14.08 \pm 3.02	11.31 \pm 2.82	* p < .001	12.92 \pm 0.95	* p < .05

comparing balance recovery mechanisms for those with neurological deficits and impaired balance control.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix

See Fig. A1 and Table A1.

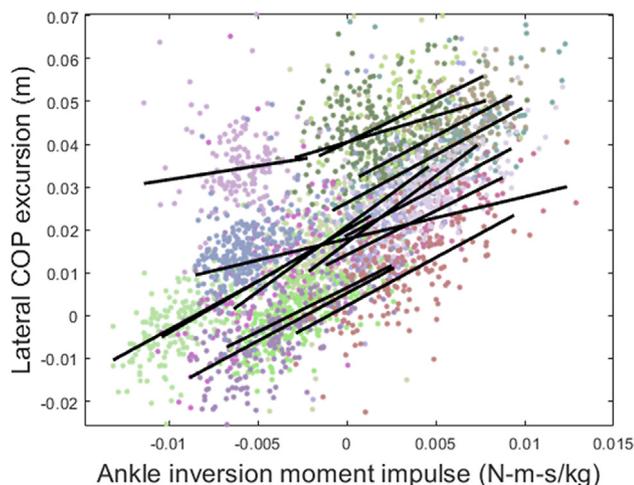


Fig. A1. Results of the linear mixed effects model analyzing the correlation between ankle inversion impulse and lateral COP excursion. Each dot represents one perturbed step and each color shows a different subject. The black lines show individual regression models for each subject.

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