Title

Proactive modifications to walking stability under the threat of large, anterior or posterior perturbations

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Abstract

Biomechanically, fall likelihood after a walking perturbation may be influenced by: (1) the pre-perturbation state of stability (i.e., “initial conditions”) and (2) how well someone responds to a perturbation (i.e., “recovery skill”). Anteroposterior walking stability must be modifiable—ideally while preserving gait speed—to be a target for fall-prevention interventions. We investigated if neurotypical adults could proactively modulate the pre-perturbation walking state of stability represented by anteroposterior stability margins. Eleven neurotypical adults walked on a treadmill at three speeds with and without anterior or posterior perturbations. We measured the margin of stability anteriorly at mid-swing and posteriorly at foot strike for pre-perturbation left and right steps. A repeated-measures factorial ANOVA evaluated main effects and interactions of walking speed, perturbation type, and limb. With posterior perturbation threats, posterior margin of stability increased at foot strike ($p < 0.01$) compared to that with no perturbations. With anterior perturbation threats, anterior margin of stability decreased at mid-swing during stance on the dominant limb compared to the dominant limb with no perturbations ($p < 0.01$). With any perturbation threat, step lengths shortened ($p < 0.01$) and step rates increased ($p < 0.01$). At slow speeds with posterior perturbation threats, double-support time decreased ($p = 0.04$). Proactive modifications to stability margins are indeed possible in a neurotypical population within a given walking speed. Consequently, anteroposterior stability may be a feasible target for fall-prevention interventions by targeting decreased step lengths or increased step rates. Beneficial modifications appear to be dependent upon measure direction and gait phase.

1. Introduction

Walking is the most common activity concurrent with falls across young, middle-aged, and older adult populations, persons with chronic stroke, and persons with Parkinson's disease (Ashburn et al., 2008; Berg et al., 1997; Geerse et al., 2019; Harris et al., 2005; Simpson et al., 2011; Talbot et al., 2005; Tinetti et al., 1988; Van Ooijen et al., 2016). Biomechanically, the ability to prevent a fall after a walking perturbation may be influenced by: (1) the pre-perturbation state of stability ("initial conditions") and (2) the response to a perturbation ("recovery skill"). Therefore, a stable gait is one in which a relatively large perturbation is needed to initiate a loss of balance, potentially because the initial conditions before a perturbation are modified. Given this proposed relationship of gait stability to fall risk, gait stability is a relevant target for interventions to reduce falls in at-risk populations.

Young adults typically respond to a potential loss of balance by adopting a more 'cautious gait' where step lengths decrease (Cham and Redfern, 2002; Hak et al., 2013, 2012; Madehkhaksar et al., 2018; Yang et al., 2016) and step rates increase (Hak et al., 2012; Madehkhaksar et al., 2018; Major et al., 2018; Yang et al., 2016) and lower-extremity muscle recruitment and joint kinematics are altered (Heiden et al., 2006). For an unanticipated potential loss of balance, recovery skills such as rapid, coordinated lower-extremity muscle activations, arm elevation, and coordination between the lower- and upper-extremities are used to regain stability (Eng et al., 1994; Marigold et al., 2003).

The margin of stability (MoS) (Hof et al., 2005) is one method of measuring gait stability during walking that relates the velocity and position of the center of mass to the edge of the base of support. Previously, we observed no differences in the anterior MoS during walking (“initial conditions”) between children with and without cerebral palsy (Tracy et al., 2019), despite those children with cerebral palsy exhibiting an impaired balance reaction (“recovery skill”) (Crenshaw et al., 2020b). The lack of altered anterior MoS could be due to little threat of a perturbation or an inability to modify anterior MoS. Anteroposterior walking stability must be modifiable—ideally while preserving gait speed—to be a target for fall-prevention interventions. One method to encourage such modifications is to introduce a perturbation threat (Johnson et al., 2019a, 2019b; Nestico et al., 2021; Shaw et al., 2012).

This study investigated whether neurotypical adults could proactively modify anteroposterior MoS when threatened with perturbations and gait parameters as a potential strategy for modifying gait stability. We controlled walking speed, as velocity directly affects the MoS and maintaining or improving gait speed is an important target for rehabilitation. We hypothesized that anteroposterior MoS would be modifiable. We predicted that neurotypical participants would increase anterior and posterior MoS when threatened with perturbations in those directions.

2. Methods

2.1. Participants

A convenience sample of 14 young adults (Table 1) participated in this IRB-approved study after providing informed consent. Participants had no self-reported neurological or musculoskeletal impairments or injuries at the time of the study and had no fractures or surgeries in the previous 18 months. The self-reported, preferred kicking limb determined limb dominance.

2.2. Protocol

All walking tasks were completed on a computer-controlled treadmill (ActiveStep®, Simbex). A safety harness attached to an overhead rail was adjusted to only arrest falls before the knees or hands touched the treadmill. If such falls occurred, the session was paused until the participant returned to a standing position on the treadmill, ready to continue. Participants completed a five-minute walking warmup at 0.8 statures per second (stats·s^{-1}), a preferred walking speed estimate (Arch and Stanhope, 2015; Bohannon, 1997).

The protocol included combinations of three walking speeds and three perturbation types, totaling nine three-minute trials. Each participant walked at 0.6 (slow), 0.8 (preferred), and 1.0 (fast) stats·s^{-1} to create comparable walking conditions between participants (Hof, 2018, 1996). One trial with each perturbation type (i.e., anterior, none, posterior) was completed within each speed. Participants were informed of the speed and perturbation combination before the trial, but they were not aware of the perturbation timing. Participants completed all nine combinations in random orders, separated by two-minute minimum rest periods. After several participants completed the protocol, we questioned whether participants perceived the perturbation difficulty to be easier with practice. To ascertain this perception, a subset of seven participants self-reported their perceived change in recovery difficulty after each trial using a five-point Likert scale.

The treadmill delivered perturbations relative to foot strike, determined with ActiveStep® software, every 12 ± 2 steps. Anterior perturbations refer to simulated trips requiring forward recovery steps, and posterior perturbations refer to simulated slips requiring backward recovery steps. We selected large perturbations with the goal of encouraging proactive modifications to stability. The perturbations, however, were not large enough to make successful recovery unfeasible, demotivating proactive modifications. Anterior perturbations were delivered 0.20 s after foot strike with the goal of perturbing mid-swing. The treadmill then resumed the predetermined belt speed. Posterior perturbations occurred immediately after foot strike. After the perturbation and a 0.21 s delay, the treadmill resumed the predetermined belt speed. Both perturbation types were similar to standing perturbations previously applied to young adults (Crenshaw et al., 2012; Crenshaw and Grabiner, 2014). Figures, characteristics, and videos of perturbations are provided as supplementary material.

2.3. Analysis

All movement was recorded from 12 cameras (Qualisys, 120 Hz) with a modified Helen-Hayes marker set creating a 13-segment whole-body model (Visual 3D, C-Motion, Inc., v2021). We low-pass filtered marker data (4^{th} order Butterworth, 6 Hz cutoff) and determined the whole-body center of mass in Visual 3D (Dempster, 1955; Hanavan, 1964). Anteroposterior stability was quantified using the MoS (Hof et al., 2005) at mid-swing and foot strike for each analyzed step using a custom script (Visual Basic for Applications, Microsoft, v2016), and gait events were determined using a coordinate-based treadmill algorithm (Zeni et al., 2008). Mid-swing was identified as the first frame where the swing-limb toe passed anterior to the stance-limb toe, a point where a trip or stumble is likely to occur (Schulz, 2011). The MoS was adapted to account for the velocity of the treadmill belt (Crenshaw et al., 2012) (Figure 1, Equation 1) and then scaled to the participant's height (Hof, 1996; Tracy et al., 2019). The final right and left steps before each perturbation were evaluated for each trial. For the no-perturbation conditions, sequential right and left steps were evaluated every 10-15 seconds. We also calculated step length and width (anteroposterior and mediolateral distances between heels at foot strike), step rate (gait speed divided by step length), and percent time in double support (percentage of stance in double support) to identify potential strategies for modifying gait stability.

The extrapolated center of mass ($xCoM$) represents the position of the whole-body center of mass (CoM) plus the velocity of the CoM (v_{CoM}) scaled. The anteroposterior position of the $xCoM$ during treadmill walking was calculated as

$$xCoM = CoM + \frac{v_{CoM} - v_{belt}}{\sqrt{\frac{g}{l}}}, \qquad \text{Equation 1}$$

where v_{belt} represents the velocity of the treadmill belt (negated because the belt direction is opposite the direction of walking), g the gravity acceleration, and l the pendulum length comprised of the instantaneous distance between the CoM and the ankle joint center of the stance limb. At foot strike, when a posterior perturbation could occur, we measured the posterior MoS (MoS_{FS}) as the distance between the $xCoM$ (Equation 1) and the stepping limb's heel (Figure 2D). A positive MoS_{FS} value indicated a stable position relative to a slip (i.e., a posterior loss of balance) where the $xCoM$ was located anterior to the stepping limb heel. At mid-swing, when an anterior perturbation could occur, we measured the anterior MoS (MoS_{MS}) as the distance between the $xCoM$ and the stance limb's toe (Figure 3D). A negative MoS_{MS} value indicated that the $xCoM$ was located anterior to the stance limb toe.

Mean MoS values and gait parameters were calculated for each limb in each condition. To evaluate the main effects and interactions of the reference limb (dominant, non-dominant), walking speed (0.6, 0.8, 1.0 stats·s^{-1}), and perturbation type (anterior, none, posterior), a repeated-measures factorial ANOVA was conducted (SPSS, IBM, v28) for anterior and posterior MoS and for gait parameters. Pairwise comparison post-hoc analyses were made with Sidak adjustments for multiple comparisons. Significance was set at $p < 0.05$ and effect sizes were reported using partial eta squared (η^2) values. Conservatively assuming independence between conditions, 12 participants provided 80% power to detect a medium-to-large main effect or interaction ($\eta^2 = 0.10$) as significant (Cohen, 1988).

3. Results

3.1. Participants

Three participants (1F/2M) were excluded from the analysis due to incomplete protocols (Table 1; one treadmill mechanical error, one partial-file corruption, and one elective end to participation due to a heightened level of excitement/nervousness—an anticipated risk (Crenshaw et al., 2020a)). Of these remaining 11 participants, all reported right-limb dominance.

3.2. Posterior Margin of Stability at Foot Strike

The MoS_{FS} represents stability relative to a backward loss of balance; therefore, the primary comparisons are between trials with and without posterior perturbations. The main effects of perturbation type ($p < 0.01$, $\eta^2 = 0.77$) and walking speed ($p < 0.01$, $\eta^2 = 0.99$) were significant (Figure 2A-C). Post-hoc comparisons showed that, compared to unperturbed trials, MoS_{FS} was more positive during trials with posterior perturbations ($p < 0.01$, mean difference (standard error) 1.70 (0.26) %height, Figure 2E).

3.3. Anterior Margin of Stability at Mid-Swing

The MoS_{MS} represents stability relative to a forward loss of balance; therefore, the primary comparisons are between trials with and without anterior perturbations. A two-way interaction of perturbation type and limb ($p < 0.01$; $\eta^2 = 0.40$) was significant (Figure 3A-C). Post-hoc comparisons showed that, compared to unperturbed trials, MoS_{MS} was more negative during dominant-limb stance during trials with anterior perturbations ($p < 0.01$, mean difference (standard error) 0.63 (0.15) %height, Figure 3E). Post-hoc comparisons between stance limbs showed no difference in MoS_{MS} between stance limbs without perturbations ($p = 0.31$), but a more negative MoS_{MS} for stance on the dominant limb during trials with anterior perturbations ($p = 0.03$, 0.48 (0.18) %height, Figure 3E).

A two-way interaction of perturbation type and walking speed ($p < 0.01$; $\eta^2 = 0.29$) was also significant. Post-hoc comparisons showed no differences in MoS_{MS} between perturbation types for the fast and estimated preferred walking speeds ($p > 0.05$), while MoS_{MS} was more negative during trials with anterior perturbations compared to unperturbed trials within the slow walking speed ($p = 0.03$, 0.58 (0.18) %height, Figure 3F).

3.4. Gait Parameters

In response to the threat of perturbations, we observed changes in step length and step rate, but not step width. There were significant main effects for step length and step rate of perturbation type ($p < 0.01$, $\eta^2 = 0.68$ and $\eta^2 = 0.67$, respectively) and walking speed ($p < 0.01$, $\eta^2 = 0.99$ for both). During trials with anterior perturbations, steps were shorter ($p < 0.01$, mean difference (standard error) 0.02 (0.004) m) and step rates higher ($p < 0.01$, 0.05 (0.010) steps·s^{-1}) compared to trials with no perturbations (Figure 4A and 4C). During trials with posterior perturbations, steps were shorter ($p < 0.01$, 0.02 (0.004) m, Figure 4A) and step rates were higher ($p < 0.01$, 0.06 (0.012) steps·s^{-1}) compared to trials with no perturbations (Figure 4A and 4C). Across walking speeds, walking faster increased step length (0.6 to 0.8 stats·s^{-1}: $p < 0.01$, 0.11 (0.004) m; 0.8 to 1.0 stats·s^{-1}: $p < 0.01$, 0.10 (0.004) m, Figure 4B) and step rate (0.6 to 0.8 stats·s^{-1}: $p < 0.01$, 0.22 (0.008) steps·s^{-1}; 0.8 to 1.0 stats·s^{-1}: $p <0.01$, 0.19 (0.008) steps·s^{-1}, Figure 4D). For step width, there were no significant main effects of perturbation type ($p = 0.11$, $\eta^2 = 0.20$, Figure 4E) or walking speed ($p = 0.31$, $\eta^2 = 0.11$, Figure 4F).

There was also a modification in the percent of time spent in double support with a significant two-way interaction of condition and speed ($p < 0.01$, $\eta^2 = 0.31$). During trials with posterior perturbations, a shorter percent of time was spent in double support within the slow walking speed (post-hoc $p = 0.04$, mean difference (standard error) 0.99 (0.34) %gait cycle, Figure 4G) compared to unperturbed trials.

3.5. Additional Results

The seven-participant subset reported decreased difficulty in perturbation recovery from the beginning to the end of the trial (range -0.14 to -1.00 points, Figure 5). Individual participant data and complete ANOVA results, including means and standard deviations for MoS measures and gait parameters, are included as supplementary material along with figures of gait parameters and number of steps analyzed per condition results.

4. Discussion

The purpose of this study was to investigate whether neurotypical adults could proactively modify anteroposterior MoS when threatened with perturbations. At foot strike, when a posterior perturbation such as a slip could occur, participants significantly increased MoS_{FS} when threatened with posterior perturbations (Figure 2E). At mid-swing, when an anterior perturbation could occur, participants did not increase MoS_{MS} when threatened with anterior perturbations (Figure 3E). Proactive modifications to posterior MoS occurred, and modifications to stability were dependent on the phase of gait and direction of the postural threat.

Fall-prevention interventions may be able to improve posterior MoS within a given speed by promoting shorter, more frequent steps. A secondary analysis of trials with and without posterior perturbations indeed showed that, within each condition and limb, MoS_{FS} was meaningfully correlated to step lengths (r = -0.75 to -0.50) and step rates (r = 0.48 to 0.75). In anticipation of slipping while walking or when responding to mediolateral treadmill perturbations, the adoption of a more 'cautious gait' is selected where step lengths decrease (Cham and Redfern, 2002; Hak et al., 2013, 2012; Madehkhaksar et al., 2018; Yang et al., 2016) and step rates increase (Hak et al., 2012; Madehkhaksar et al., 2018; Major et al., 2018; Yang et al., 2016). In comparable samples of young adults walking without perturbations and with the awareness that a posterior perturbation may occur, Yang and colleagues also observed improved posterior stability with awareness (Yang et al., 2016), while Eichenlaub and colleagues did not observe a change in posterior stability while anticipating a perturbation (Eichenlaub et al., 2023). An increase in step width is also common (Ahuja and Franz, 2022; Hak et al., 2013, 2012; Major et al., 2018; Wu et al., 2015), but we observed no change. Possibly, the lack of a direct threat to lateral stability did not require the same kinematic adaptation to protect against a lateral loss of balance. Mathematically, anteroposterior MoS can be altered by changing the COM velocity associated with gait speed (Figures 2F and 3F). Previous studies have shown that faster gait speeds increase posterior stability (Espy et al., 2010), but also increase the required coefficient of friction between the foot and floor (Kim et al., 2005). We have demonstrated that MoS_{FS} can be modified with no change to walking speed—an outcome that is often targeted by gait rehabilitation. Decreasing step length to improve posterior stability may be challenging for populations who already walk with shorter strides (e.g., persons with Parkinson's disease) (Morris et al., 1996). A meaningful argument against modifying step length away from self-selected parameters is the likelihood that this change will decrease walking economy and increase the mechanical work done at the joints (Ahuja and Franz, 2022; Gordon et al., 2009; Hreljac and Martin, 1993). Stability, economy, and joint work are all important considerations for gait interventions, and the risks and benefits of which take priority are likely individually specific.

The decrease in the percent of time spent in double support (Figure 4G) was unexpected as double-support allows for effective center-of-pressure modulation in response to perturbations (van Mierlo et al., 2021). This decrease may exemplify the neurotypical participants relying on their balance reaction capabilities in addition to improving their initial stability conditions as posterior stability was increased at foot strike (Figure 2E). By decreasing the percent of time spent in double support, resulting in an increase in the percent of time spent in single support, the participant would be in a position to respond more quickly to a perturbation by placing the swing limb after a posterior perturbation rather than repositioning a previously placed step. This strategy could be particularly useful for posterior perturbations as the direction of the recovery step is opposite the walking direction. The need to maintain a faster walking speed in other conditions may not allow for this modification, and individuals with decreased recovery skill may not be able to utilize this modification.

Modifications to MoS were dependent on the gait phase and the measurement direction. Unlike MoS_{FS} which can be modified by step placement, modifying MoS_{MS} may require center of mass velocity control. This control deviates from the inverted pendulum trajectory and increases energetic cost (Kuo, 2007). As walking speed must be maintained on a treadmill, this increased cost may be too much to be a desirable proactive modification, especially in a young, neurotypical population where recovery skill is high. Eichenlaub et al. posit that a young population may be able to rely on rapid reactive adjustments limiting this population's need to induce proactive adjustments (Eichenlaub et al., 2023). Alternatively, modifications may have occurred to improve the anterior MoS at points of the gait cycle other than mid-swing. Decreasing trip-related fall risk at mid-swing may require focusing on other factors such as toe clearance. The lack of beneficial adaptations to MoS_{MS} were not due to ineffective threats to anterior stability as one participant withdrew from participating and four participants fell into the harness during one trial each.

After trials with perturbations, a seven-participant subset reported a trend that perturbation recovery became easier (range -0.14 to -1.00 points on a five-point scale, Figure 5) within the duration of the trial. This trend either suggests that we detected proactive modifications despite a learning effect on recovery ability, or that the modifications increased stability easing recovery. With a novel walking challenge, participants may adapt their perturbation recovery strategy over time, and improving *post*-perturbation recovery may reduce the participant's *pre*-perturbation modifications limiting the observed effects from our analyses.

Moving forward, studies of other at-risk populations will provide insight into the influence of age, impairments, and walking confidence on proactive modifications to walking stability, especially for those less able to rely on their reactive capabilities. As this study only evaluated neurotypical adults, the extent to which clinical populations can modify MoS and the ability to transfer these modifications to unperturbed walking are still unknown. In addition, further study is needed to evaluate the extent to which these proactive modifications benefit the risk of falling from a given perturbation magnitude. With an improved understanding of proactive walking stability control, rehabilitation protocols can specifically target increased walking stability for all environments.

In conclusion, these results suggest that posterior MoS_{FS} is modifiable. Consequently, walking stability may be a feasible target for fall-prevention interventions by decreasing step length and increasing step rate while maintaining walking speed. However, potentially detrimental effects on walking economy should be considered. These proactive modifications to posterior MoS were implemented despite the capacity for neurotypical participants to rely on their ability to recover from perturbations. These results also provide a framework with which to interpret results from at-risk populations.

Conflict of Interest Statement

The authors declare no competing financial or non-financial interests which could or appear to influence the interpretation or presentation of this work.

CRediT Statement

James Tracy: Conceptualization, Methodology, Software, Formal Analysis, Investigation, Data Curation, Writing - Original Draft, Visualization, Funding Acquisition. **Jocelyn Hafer:** Conceptualization, Methodology, Writing - Review & Editing. **Hendrik Reimann:** Conceptualization, Methodology, Writing - Review & Editing. **Thomas Buckley:** Conceptualization, Methodology, Writing - Review & Editing. **Jessica Allen:** Conceptualization, Methodology, Writing - Review & Editing. **Jeremy Crenshaw:** Conceptualization, Methodology, Formal Analysis, Writing - Review & Editing, Supervision, Project Administration.

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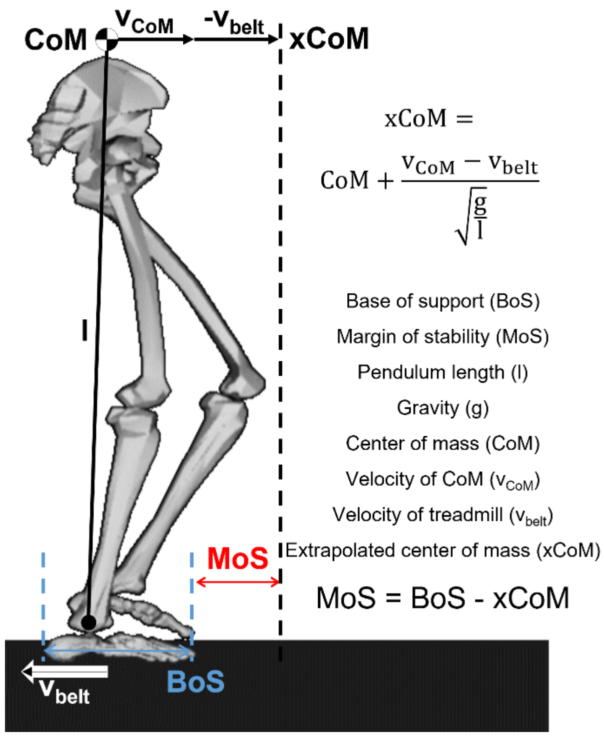

Figure 1: Illustration of the anteroposterior margin of stability at mid-swing accounting for treadmill belt velocity. Figure modified from previous publication [10]. The margin of stability represents the distance between the base of support and the extrapolated center of mass (center of mass position + scaled center of mass velocity + treadmill belt velocity). Positive values represent a state of stability (i.e. the extrapolated center of mass is within the base of support or advantageously placed away from the edge of the base of support), and a perturbation is needed to initiate a fall in that direction. Negative values represent a state of instability (i.e. the extrapolated center of mass is outside the base of support), and a compensatory action such as taking a step, applying an external force, or counter-rotating segments about the center of mass is needed to prevent a fall.

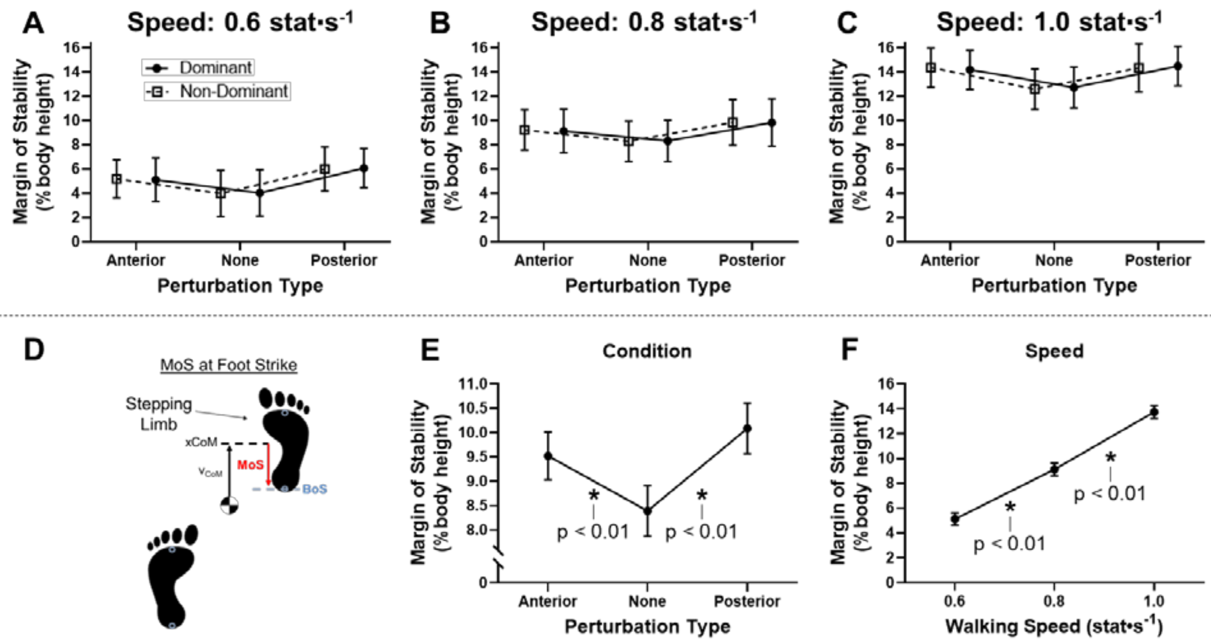

Figure 2: Panels A-C: Summary results for dominant and non-dominant limb posterior margin of stability at foot strike with and without anterior or posterior perturbations. **Panel D:** Illustration of the posterior margin of stability at foot strike. **Panel E-F:** Estimated marginal means and standard errors of posterior margin of stability across perturbation types (simulated trips, none, or simulated slips) and walking speeds (slow, estimated preferred, or fast). **Panel E:** Participants showed increased posterior stability (i.e., more positive MoS) at foot strike relative to a posterior loss of balance during trials with perturbations compared to trials without perturbations. **Panel F:** With each increase in walking speed, participants increased their stability relative to a backward loss of balance.

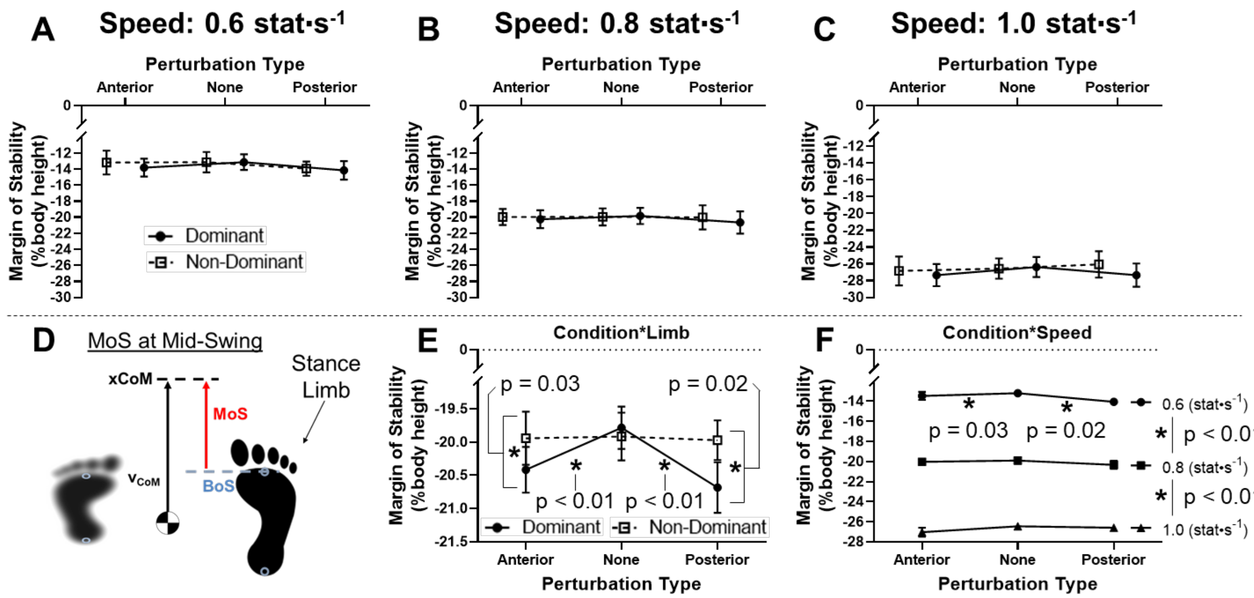

Figure 3: Panels A-C: Summary results for dominant and non-dominant limb anterior margin of stability at mid-swing with and without anterior or posterior perturbations. **Panel D:** Illustration of the anterior margin of stability at mid-swing. **Panel E-F:** Estimated marginal means and standard errors of anterior margin of stability interactions of perturbation types (simulated trips, none, or simulated slips) with stance limb (dominant or non-dominant) and perturbation types with walking speeds (slow, estimated preferred, or fast). **Panel E:** Participant's had less anterior stability (i.e., more negative MoS) during stance on the dominant limb for trials with perturbations compared to trials without perturbations. Decreased anterior stability during stance on the dominant limb compared to stance on the non-dominant limb was present within perturbation trials, but not within the no perturbation trials. The dominant limb is shown with solid circles and solid lines. The non-dominant limb is shown with open squares and dashed lines. **Panel F:** A decrease in anterior stability was observed between trials with and without perturbations within the slow walking speed condition. With each increase in walking speed, participants decreased their stability relative to a forward loss of balance.

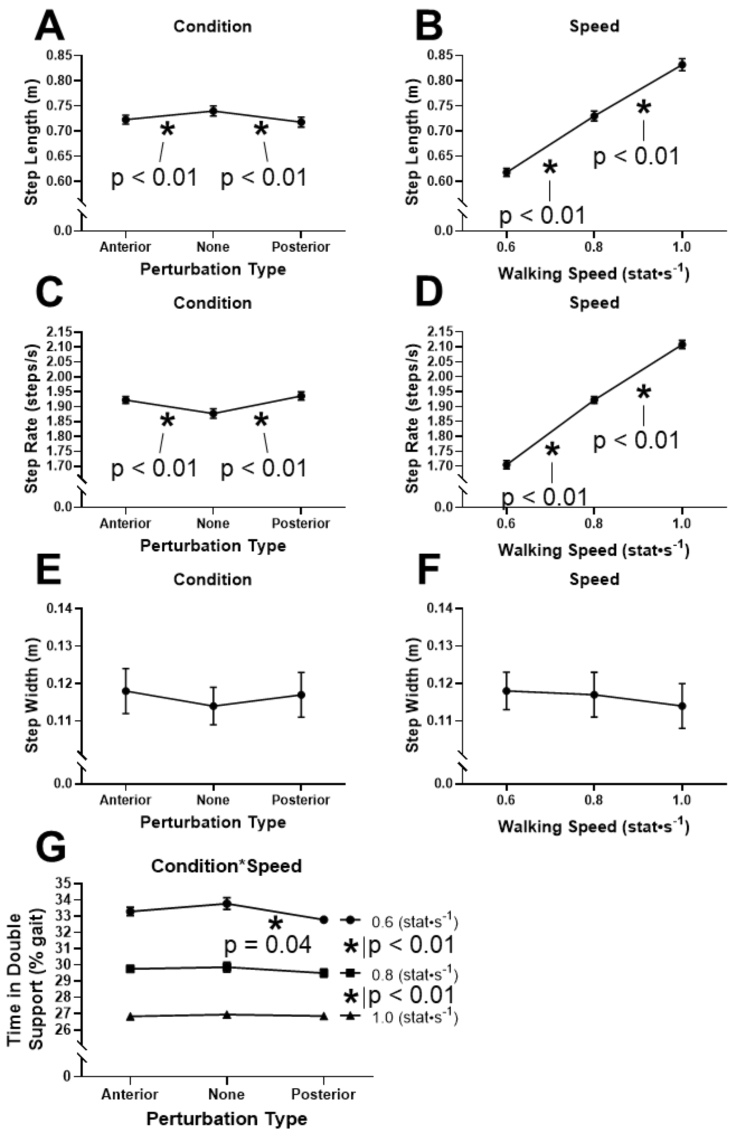

Figure 4: Gait parameters across perturbation types and walking speeds. Estimated marginal means and standard errors of gait parameters across perturbation types (simulated trips, none, or simulated slips) and walking speeds (slow, estimated preferred, or fast). **Panels A-B:** Participants decreased step length for trials with perturbations compared to trials without perturbations and increased step length with increasing walking speed. **Panels C-D:** Participants increased step rate for trials with perturbations compared to trials without perturbations and increased step rate with increasing walking speed. **Panels E-F:** Participants did not change step width when threatened with perturbations or when changing walking speeds. **Panel G:** A decrease in the percent of time spent in double support was observed when threatened with posterior perturbations at the slow walking speed. With each increase in walking speed, participants decreased their percentage of time spent in double support.

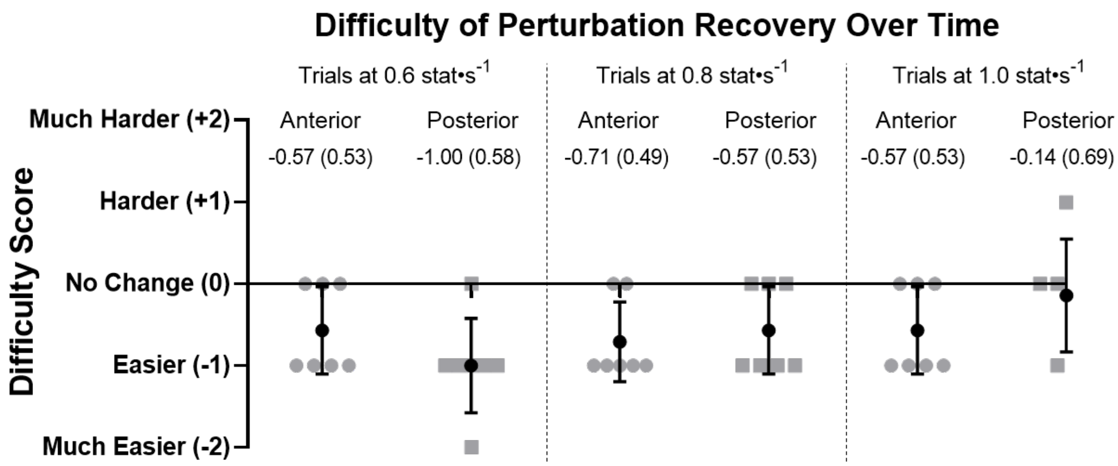

Figure 5: Perceived change in difficulty of perturbation recovery over time. Participant response mean and standard deviation represented with solid black circles and error bars. Individual responses shown with gray circles (anterior perturbation trials) and gray squares (posterior perturbation trials). Across all combinations of trials with perturbations and walking speeds, we observed a trend suggesting that the perturbations were easier to recover from over the length of the trial.

Table 1: Description of participants.

ID	Sex	Age (yrs)	Height (m)	Mass (kg)	0.6 stats·s^{-1} Condition (m/s)	0.8 stats·s^{-1} Condition (m/s)	1.0 stats·s^{-1} Condition (m/s)
*** 1**	M	26	1.87	75.0	1.12	1.50	1.87
*** 2**	M	20	1.93	74.0	1.16	1.54	1.93
*** 3**	F	21	1.77	61.0	1.06	1.42	1.77
4	M	25	1.74	62.5	1.04	1.39	1.74
5	F	27	1.74	75.5	1.04	1.39	1.74
6	M	32	1.92	90.0	1.15	1.54	1.92
† 7	F	20	1.67	57.0	1.00	1.34	1.67
† 8	M	26	1.75	58.0	1.05	1.40	1.75
9	M	21	1.90	82.5	1.14	1.52	1.90
† 10	F	27	1.68	71.5	1.01	1.34	1.68
† 11	F	30	1.75	72.0	1.05	1.40	1.75
† 12	F	30	1.71	55.5	1.03	1.37	1.71
† 13	M	36	1.79	64.5	1.07	1.43	1.79
† 14	F	24	1.65	53.5	0.99	1.32	1.65
All - Mean (SD)		26.1 (4.8)	177.4 (9.4)	68.0 (10.9)	1.07 (0.06)	1.42 (0.07)	1.78 (0.09)
Included - Mean (SD)		27.1 (4.7)	175.2 (8.7)	67.5 (11.9)	1.05 (0.05)	1.40 (0.07)	1.75 (0.09)

Note: * Indicates participants excluded from the analyses due to incomplete protocols. One participant did not complete the protocol due to a treadmill mechanical error, one participant had an incomplete data set due to a partial-file corruption, and one participant elected to end their participation due to a heightened level of excitement/nervousness—an anticipated risk [36]. † Indicates the subset of participants who completed the Likert scale question regarding perceived change in recovery difficulty.

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