Journal of Biomechanics 157 (2023) 111731

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech

The influence of step width on balance control and response strategies during perturbed walking in healthy young adults

Lindsey K. Molina, Gabriella H. Small, Richard R. Neptune *

Walker Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA

ARTICLE INFO

Keywords:
Balance
Biomechanics
Gait
Perturbation
Step width

ABSTRACT

Individuals with neuromuscular deficits often walk with wider step widths compared to healthy adults. Wider steps have been linked to a higher destabilizing frontal-plane external moment and greater range of frontal-plane whole-body angular momentum (H_R), which is an indicator of decreased balance control. The purpose of this study was to experimentally determine 1) how step width alters balance control during steady-state walking, and 2) if step width changes the balance response strategies following mediolateral surface perturbations in healthy adults.

Fifteen healthy young adults (7 male, age: 25 ± 4 years) walked on an instrumented treadmill at narrow, self-selected, wide and extra-wide step widths. During perturbed trials, the treadmill provided random mediolateral surface translations to each foot midway through single-leg-stance. Muscle electromyography, biomechanical measures (H_R, frontal-plane external moment and joint moments) and deviations (differences in these measures between the perturbed and unperturbed walking trials) were compared across step widths.

During steady state walking, wider steps were associated with decreased balance control. Increasing step widths were also associated with increased gluteus medius activity and reduced hip abduction and ankle inversion moments, which suggests healthy subjects rely more on a lateral ankle strategy to maintain balance at increasing step widths. There was no change in the plantarflexion moment. During perturbed walking, lateral, but not medial, surface translations adversely affected balance control. Further, wider steps did not change the balance response strategies following the perturbations, which suggests healthy individuals have the capacity to respond similarly to the perturbations at different step widths.

1. Introduction

Individuals with neuromuscular deficits are at an increased risk of falling compared to healthy young adults (Campbell et al., 1981; Forster and Young, 1995; Miller et al., 2001). Falls can be detrimental for these clinical populations, leading to injuries and a fear of falling (Painter et al., 2012; Rubenstein, 2006). A better understanding of the mechanisms used to maintain balance during steady-state walking and restore balance following perturbations could help guide rehabilitation aimed at reducing fall rates among clinical populations.

Individuals with neuromuscular deficits often walk with wider step widths (Hof et al., 2007; Roerdink et al., 2007) and increased step width variability (McAndrew Young and Dingwell, 2012; Owings and Grabiner, 2004) compared to healthy individuals. Walking with wider steps increases the mediolateral moment arm between the body center of mass (COM) and center of pressure (COP). While a larger moment arm may lead to greater stability during static tasks, a larger mediolateral moment arm has been linked to a higher destabilizing frontal-plane external moment and greater range of frontal-plane whole-body angular momentum (H_R) (Vistamehr et al., 2016), which are indicators of poor balance during dynamic tasks such as walking (Neptune and Vistamehr, 2019; Nott et al., 2014). Wider step widths are also linked to increases in hip abduction moment (Vistamehr and Neptune, 2021) and stance phase gluteus medius activity (Kubinski et al., 2015). These greater demands placed on the hip abductor muscles when walking with wider steps could be disadvantageous for clinical populations with reduced abductor muscle strength (e.g., Andrews et al., 1996; Larsson et al., 2019; Neckel et al., 2006). In addition, wider step widths could lead to adaptations at the ankle, as modeling and simulation studies have identified the plantarflexors as primary contributors to

* Corresponding author at: Walker Department of Mechanical Engineering, The University of Texas at Austin, 204 E. Dean Keeton Street, Stop C2200, Austin, TX 78712-1591, USA.

E-mail address: rneptune@mail.utexas.edu (R.R. Neptune).

https://doi.org/10.1016/j.jbiomech.2023.111731
Accepted 14 July 2023
Available online 17 July 2023

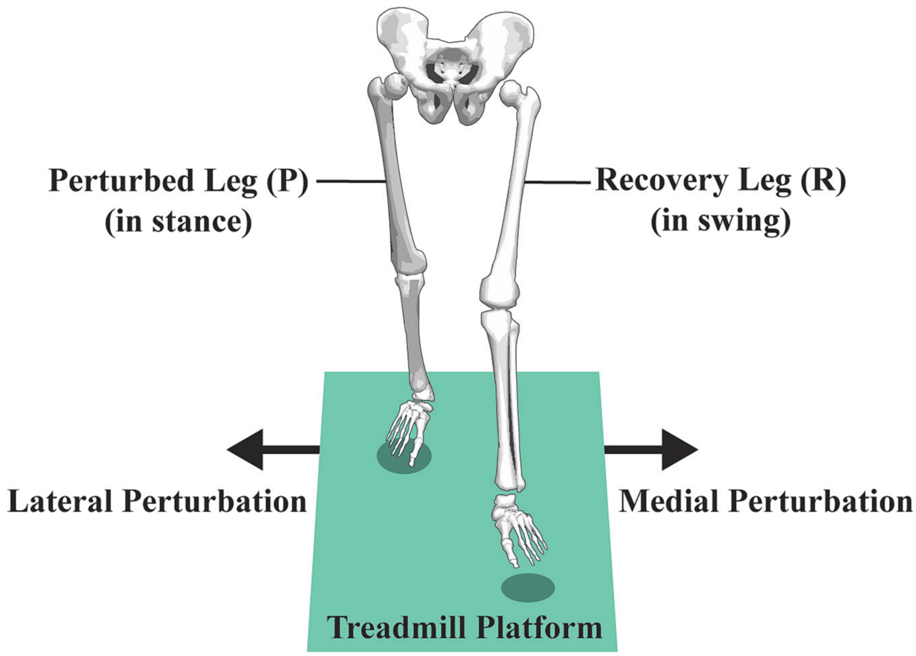

Fig. 1. Schematic of stance leg (perturbed leg) and contralateral swing leg (recovery leg) during mediolateral surface translation perturbations.

Table 1
Desired and actual % change from the self-selected (SS) step width condition.

	Desired % change from the SS condition (%)	Actual % change from the SS condition (%)
Narrow	−25	−12 ± 7
Wide	50	35 ± 15
Extra Wide	100	74 ± 20

mediolateral balance control (e.g., Neptune and McGowan, 2016; Pandy et al., 2010). Further understanding of how step width affects balance control could provide insight into why those who walk with wider steps are more susceptible to falling (Campbell et al., 1981; Forster and Young, 1995; Miller et al., 2001).

Several strategies are used to maintain balance during perturbed and unperturbed walking (Leeuwen et al., 2022; Reimann et al., 2018). A foot placement strategy can keep the COM within the base of support (e.g., Reimann et al., 2018; Segal and Klute, 2014). The stance leg can be used to modify various aspects of the COP and ground reaction force (GRF) to stabilize gait: a lateral ankle strategy shifts the COP mediolaterally to restore the COM location relative to the COP (e.g., Hof et al., 2010), a lateral hip strategy moves the trunk to reposition the COM relative to the COP (e.g., Winter, 1995) and an ankle push-off strategy accelerates the COM towards the middle of the base of support (e.g., Kim and Collins, 2013).

The purpose of this study was to experimentally determine: 1) how step width alters balance control during steady-state walking, and 2) if altered step width changes the balance response strategies following mediolateral surface perturbations in healthy young adults. During steady-state walking, we hypothesized that an increased step width will increase H_R and systematically reduce the ankle inversion moment and increase the hip abduction and ankle plantarflexion moments during stance. Further, following a perturbation, we hypothesize that walking with increasing wider steps will require the recruitment of additional strategies to restore balance. Specifically, we expected that following medial/lateral surface translation perturbations, walking with a wider step width will: 1) lead to a greater decrease/increase in deviations (i.e., difference between the perturbed and unperturbed walking trials) of H_R and 2) require a greater increase/decrease in deviations of hip abduction activity and hip abduction moment (i.e., hip strategy).

2. Methods

2.1. Data collection

Fifteen healthy young adults (7 male, age: 25 ± 4 years) (Table A1), all below the age of 35, were recruited from the local community. All participants provided informed written consent approved by The University of Texas at Austin Institutional Review Board and were free from any musculoskeletal or neuromuscular injuries. Kinematic, kinetic and electromyography (EMG) data were collected. Three-dimensional kinematic data were collected at 120 Hz with a 10-camera motion capture system (Vicon, Oxford, UK) using a full-body marker set with 56 reflective markers. Three-dimensional GRF data were collected at 960 Hz from an instrumented treadmill (Motek, Amsterdam, NL). EMG data were collected (Motion Labs Systems, Inc., Baton Rouge, USA) at 2160 Hz using bilateral electrodes placed on the medial gastrocnemius, soleus, tibialis anterior and gluteus medius.

Each subject's self-selected (SS) walking speed was determined from three averaged 10 m overground walking trials. Each subject's SS step width was found by averaging their step width, defined as the

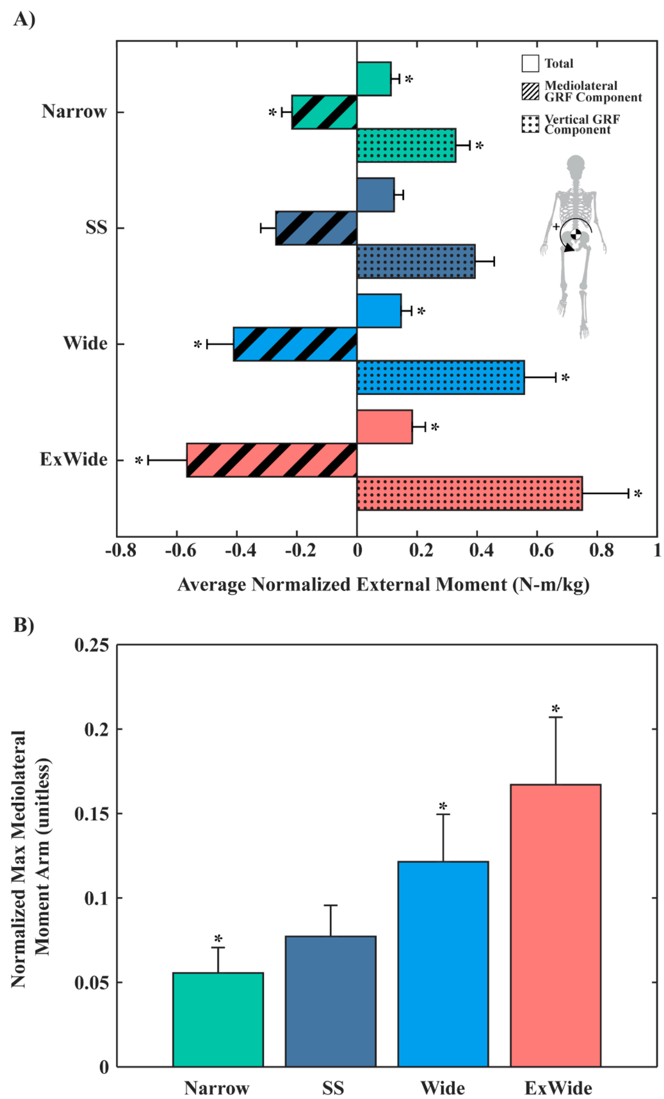

Fig. 2. **A)** The frontal-plane external moment and the vertical and mediolateral ground reaction force (GRF) components of frontal-plane external moment (± standard deviation), averaged across the gait cycle and between both legs and normalized by subject mass, and **B)** the maximum (max) mediolateral moment arm (± standard deviation) for unperturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths, normalized by leg length. A positive external moment acts to rotate the body towards the swing leg. "*" indicates a significant difference in the outcome measure from SS step widths.

mediolateral distance between heel markers at consecutive heel strikes, from 20 consecutive steps while walking on the treadmill at their SS speed. Each subject performed eight 30-second walking trials on the instrumented treadmill, consisting of steady-state walking at narrow (25% narrower), SS, wide (50% wider) and extra-wide (100% wider) step widths. A custom D-flow (Motek, Amsterdam, NL) script was developed to determine heel strike timing (Zeni et al., 2008), which was used to project foot placement targets on the treadmill at the desired step width during each trial. During four of the eight trials, the treadmill provided medial and lateral 2.5 cm surface translations lasting 0.25 s (i.e., the surface moved the stance surface medially or laterally) to each stance foot midway through single-leg-stance. The perturbation timing occurred at the point of maximum rate of change of frontal-plane whole-body angular momentum (*H)* during stance (~30% of the gait cycle). All trial conditions were randomized.

2.2. Data analysis

Kinematic and GRF data were filtered at 6 Hz and 15 Hz, respectively, using a 4th-order Butterworth low-pass filter. GRF data were recorded during platform movements without a participant, and then resulting inertial artifacts were subtracted from the raw GRF signals during the perturbation trials. Steps on the contralateral force plate (cross-over steps) were identified and removed from the kinetic analyses. A 13-segment inverse dynamics model was created for each subject using Visual3D (C-Motion, Germantown, USA) to quantify dynamic balance and calculate ankle and hip moments. Dynamic balance was quantified using *H* (Neptune and Vistamehr, 2019) and normalized by subject mass, height and walking speed. H_R was defined as the peak-to-peak range of normalized *H* in the frontal-plane. The frontal-plane external moment was calculated as:

$$\dot{\vec{H}} = \dot{\vec{H}}_{vertical} + \dot{\vec{H}}_{ML} = \vec{F}_{GRFvertical} * \vec{z} + \vec{F}_{GRFML} * \vec{y} \quad (1)$$

where $\dot{\vec{H}}_{vertical}$ and $\dot{\vec{H}}_{ML}$ are separate contributions from the vertical and mediolateral GRF to the frontal-plane external moment, $\vec{F}_{GRFvertical}$ and $\vec{F}_{GRFML}$ are the vertical and mediolateral GRFs, and $\vec{z}$ and $\vec{y}$ are the mediolateral and vertical moment arms from the body COM to the COP. Frontal-plane external moments were averaged across the gait cycle and between both legs. Joint moments and frontal-plane external moment were normalized by subject mass. Mediolateral moment arms were normalized by leg length. EMG signals were high-pass filtered at 40 Hz with a zero-lag 4th-order Butterworth filter, demeaned, rectified and low-pass filtered at 4 Hz with a zero-lag 4th-order Butterworth filter. For each subject, EMG was normalized by the peak EMG value observed in the unperturbed SS step width walking condition trial.

For each trial condition (perturbation type, step width and leg), to capture representative responses, data (H_R, ankle inversion moment, ankle plantarflexion moment, hip abduction moment and EMG) three standard deviations from their means were removed as outliers. For unperturbed trials, outcome measures were averaged across all gait cycles and compared across step widths to determine how step width alters balance control during steady-state walking. To determine differences in response strategy across step widths, deviation of H_R was calculated as the difference in H_R between perturbed and unperturbed steps and compared across step widths. Similarly, deviations of joint moments were calculated as the difference in joint moments between the perturbed and unperturbed trails across the gait cycle. For clarity, we will refer to the leg in stance during the perturbation as the *perturbed leg* and the contralateral leg on the subsequent step as the *recovery leg* (Fig. 1). Balance response strategies were defined using deviations of H_R and deviations of joint moments.

2.3. Statistics

2.3.1. Balance control during steady-state walking

For unperturbed trials, linear mixed effects models determined differences in frontal-plane external moment and H_R with subject as the random effect and step width as the fixed effect. If significant differences were found, follow-up paired t-tests with a Bonferroni post-hoc correction were used to determine differences between the SS step width and all other step widths. Differences in joint moment measures (ankle inversion moment, ankle plantarflexion moment and hip abduction moment) and EMG were evaluated using a one-way repeated-measures Statistical Parametric Mapping (SPM) Analysis of Variance (ANOVA)

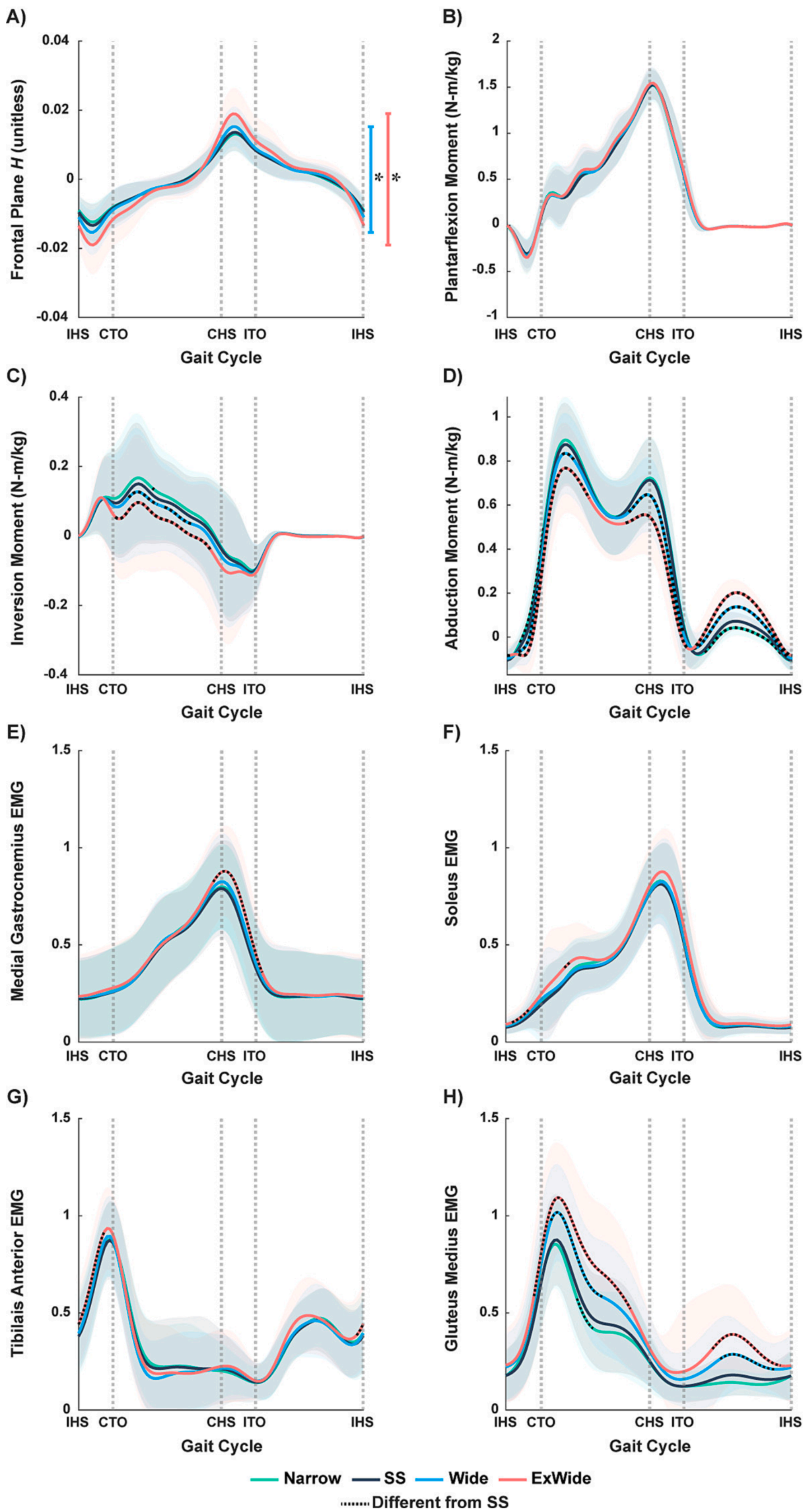

Fig. 3. **A)** Frontal-plane whole-body angular momentum (*H*) and **B)** ankle plantarflexion moment, **C)** ankle inversion moment, **D)** hip abduction moment, **E)** medial gastrocnemius electromyography (EMG), **F)** soleus EMG, **G)** tibialis anterior EMG and **H)** gluteus medius EMG (± standard deviation) averaged between both legs for unperturbed trials during walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Vertical grey dashed lines indicate ipsilateral leg heel strike (IHS) and toe off (ITO) and contralateral leg heel strike (CHS) and toe off (CTO). Black dotted lines or "*" indicate a significant difference in the outcome measure from SS step widths.

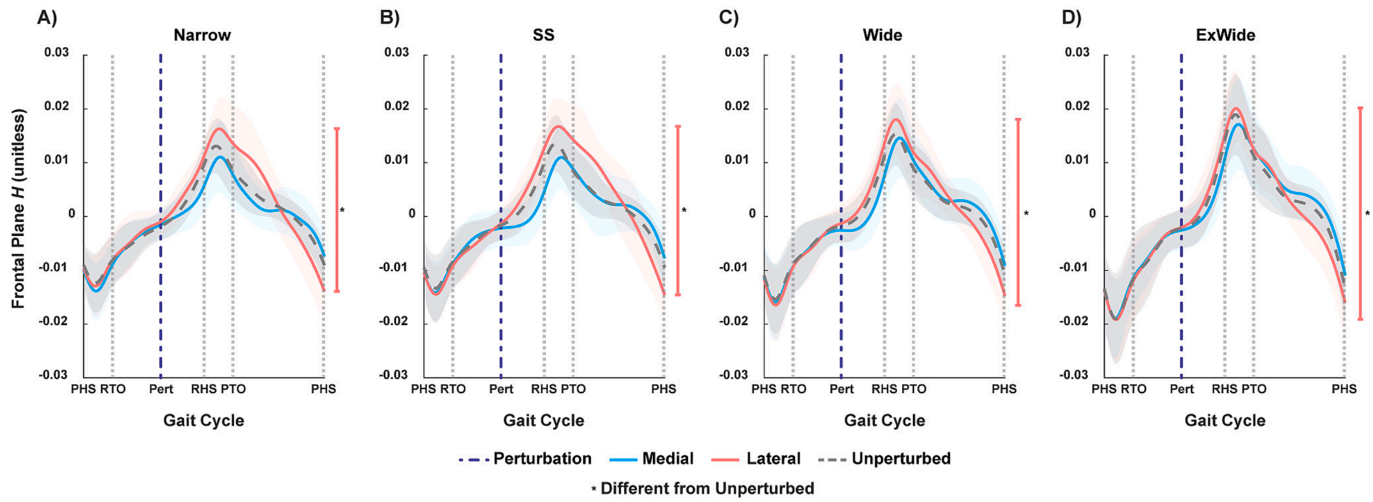

Fig. 4. Frontal-plane whole-body angular momentum (H) (± standard deviation) during perturbed and unperturbed walking at **A)** narrow, **B)** self-selected (SS), **C)** wide and **D)** extra-wide (ExWide) step widths. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). "*" indicates a significant difference in H_R from unperturbed trials.

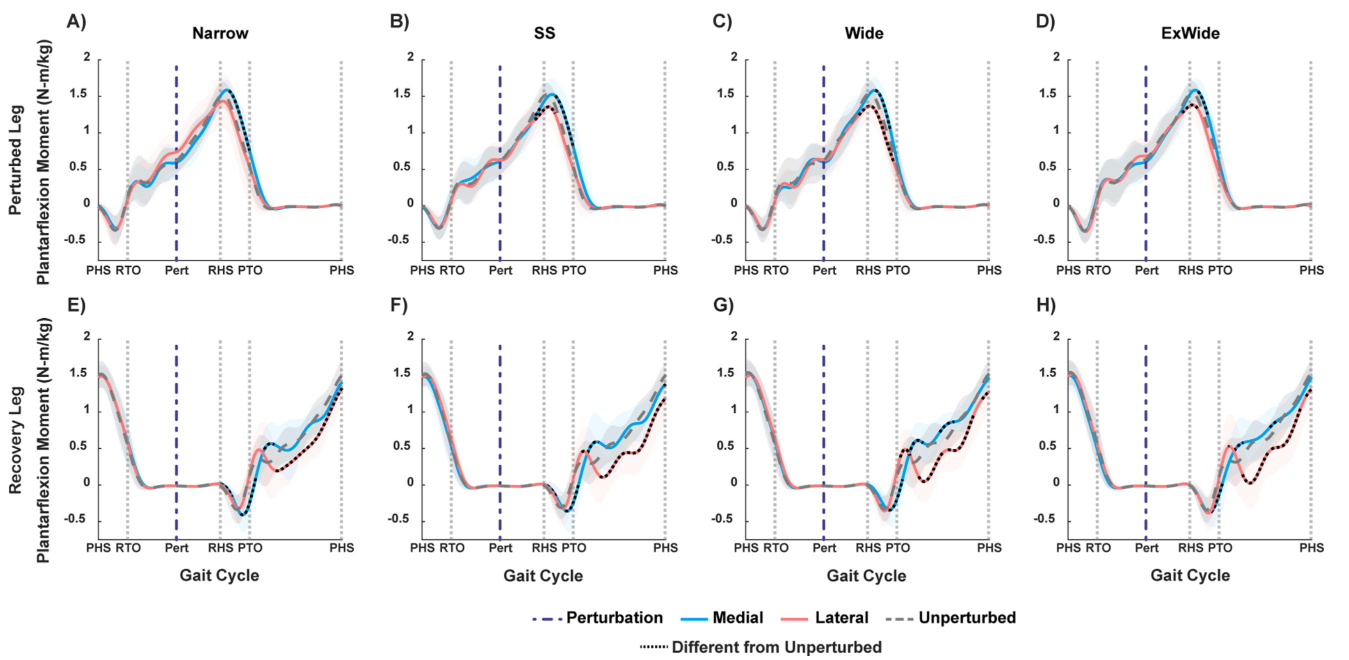

Fig. 5. Ankle plantarflexion moment (± standard deviation) during perturbed and unperturbed walking at narrow **(A and E)**, self-selected (SS) **(B and F)**, wide **(C and G)** and extra-wide (ExWide) **(D and H)** step widths for the perturbed and recovery leg. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). Black dotted lines indicate a significant difference in ankle plantarflexion moment from unperturbed trials.

(Pataky, 2012) across the gait cycle. Significance was defined as $p \leq 0.05$. If the SPM ANOVA found statistical differences at any point in the gait cycle, follow-up SPM paired t-tests with a Bonferroni post-hoc correction were used to determine differences between the SS step width and all other step widths.

2.3.2. Balance response strategies to mediolateral perturbations

SPM paired t-tests were used to determine differences in joint moment outcome measures and EMG across the stance phase of the gait cycle between perturbed and unperturbed trials. Linear mixed effects models evaluated differences in H_R with subject as the random effect and perturbation condition (perturbed or unperturbed) as the fixed effect.

2.3.3. Balance response strategies compared across step widths

For perturbed trials, linear mixed effects models determined differences between deviation of H_R with subject as the random effect and step width as the fixed effect. If significant differences were found, follow-up paired t-tests with a Bonferroni post-hoc correction were used to determine differences between the SS step width and all other step widths. Differences in deviation of joint moment measures were

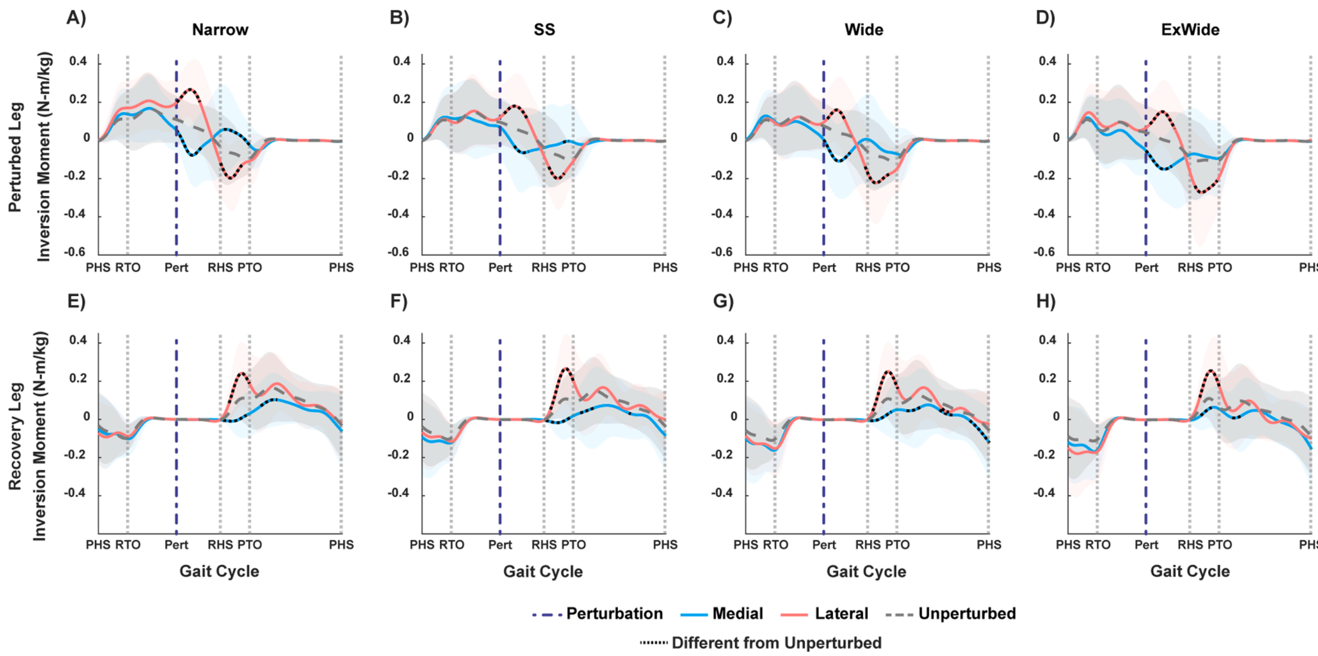

Fig. 6. Ankle inversion moment (± standard deviation) during perturbed and unperturbed walking at narrow **(A and E)**, self-selected (SS) **(B and F)**, wide **(C and G)** and extra-wide (ExWide) **(D and H)** step widths for the perturbed and recovery leg. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). Black dotted lines indicate a significant difference in ankle inversion moment from unperturbed trials.

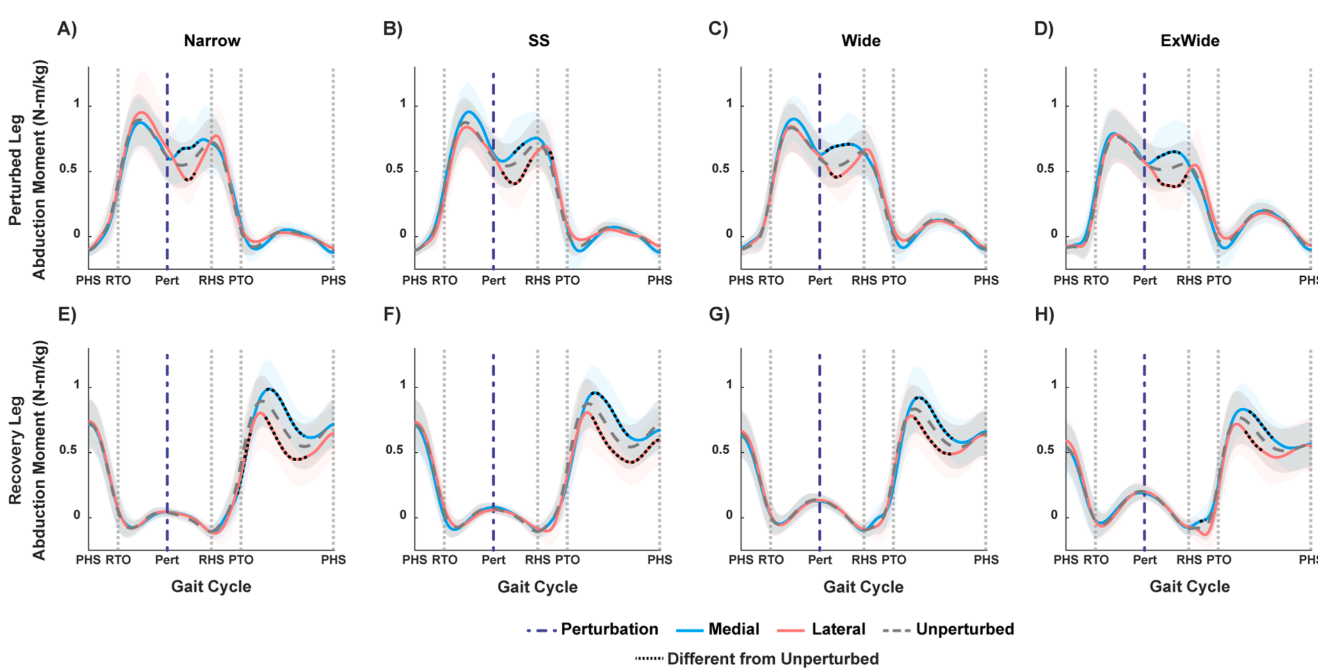

Fig. 7. Hip abduction moment (± standard deviation) during perturbed and unperturbed walking at narrow **(A and E)**, self-selected (SS) **(B and F)**, wide **(C and G)** and extra-wide (ExWide) **(D and H)** step widths for the perturbed and recovery leg. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). Black dotted lines indicate a significant difference in hip abduction moment from unperturbed trials.

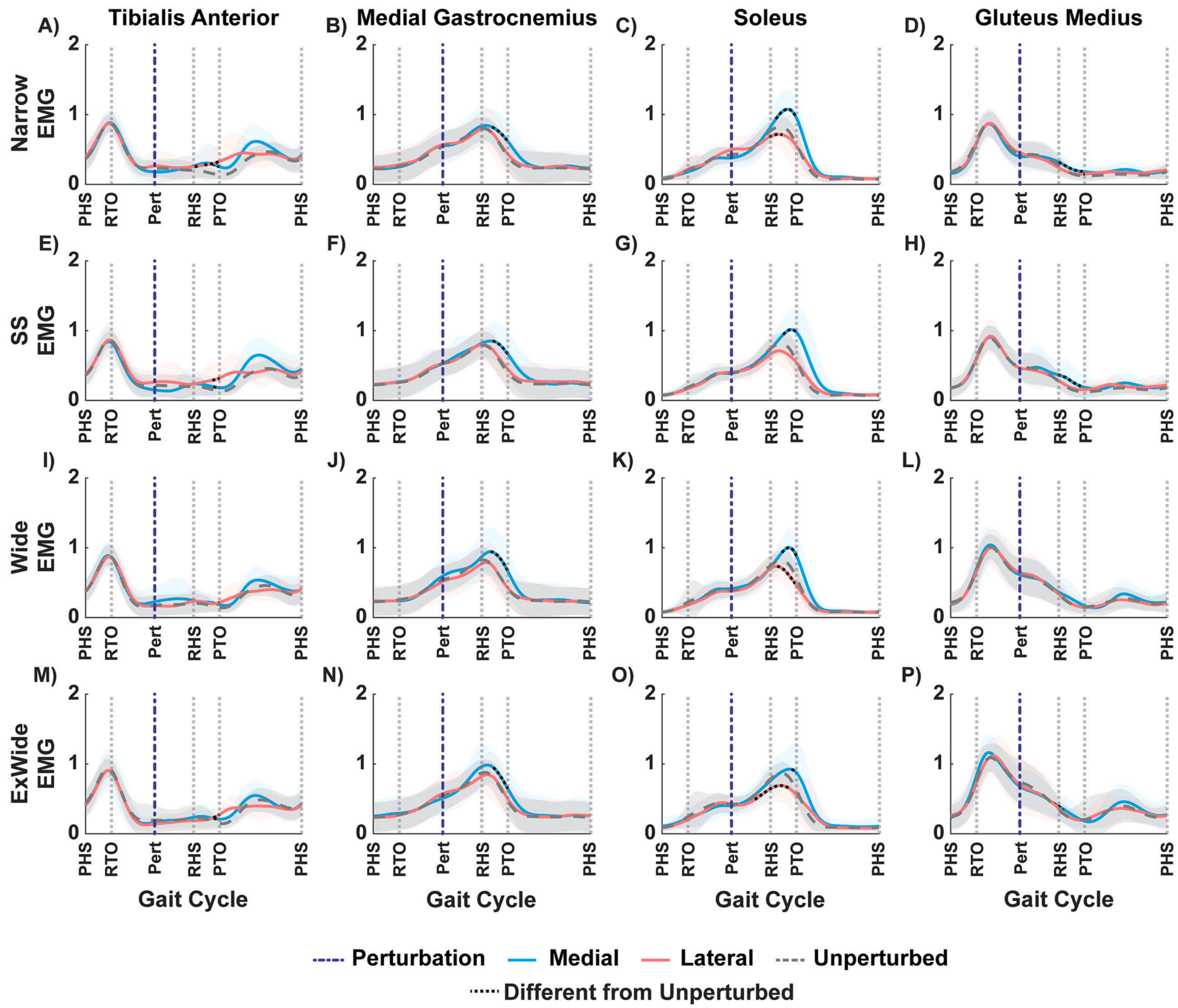

Fig. 8. Electromyography (EMG) signals (± standard deviation) during perturbed and unperturbed walking at narrow **(A-D)**, self-selected (SS) **(E-H)**, wide **(I-L)** and extra-wide (ExWide) **(M−P)** step widths on the *perturbed leg* for the tibialis anterior **(A, E, I and M)**, medial gastrocnemius **(B, F, J and N)**, soleus **(C, G, K and O)** and gluteus medius **(D, H, L and P)**. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). Black dotted lines indicate a significant difference in EMG activity from unperturbed trials.

evaluated using a one-way repeated-measures SPM ANOVA during the stance phase of the gait cycle for both legs. If the SPM ANOVA found statistical differences at any point in the gait cycle, follow-up SPM paired t-tests with a Bonferroni post-hoc correction were used to determine differences between the SS step width and all other step widths. All statistical analyses were performed using MATLAB (Mathworks, Natick, MA).

3. Results

Actual step widths were narrower than targeted step widths (Table 1).

3.1. Balance control during steady-state walking

During unperturbed trials, average frontal-plane external moment increased with step width due to increases in the mediolateral moment arm, resulting in an increased vertical GRF component, which acts to rotate the body towards the swing leg, while the mediolateral GRF component, which acts to rotate the body towards the stance leg, became more negative (Fig. 2, Table A2). H_R increased for wide and extra-wide step widths, ankle inversion moment decreased as step width increased during single-leg-stance, and hip abduction moment decreased/increased as step width increased during stance/swing (Fig. 3, Table A2 & A3). Plantarflexion moment did not change across step widths. For extra-wide step widths, medial gastrocnemius activity increased during second double support, soleus activity increased slightly during some periods of stance and tibialis anterior activity increased during first double support. As step width increased, gluteus medius activity also increased.

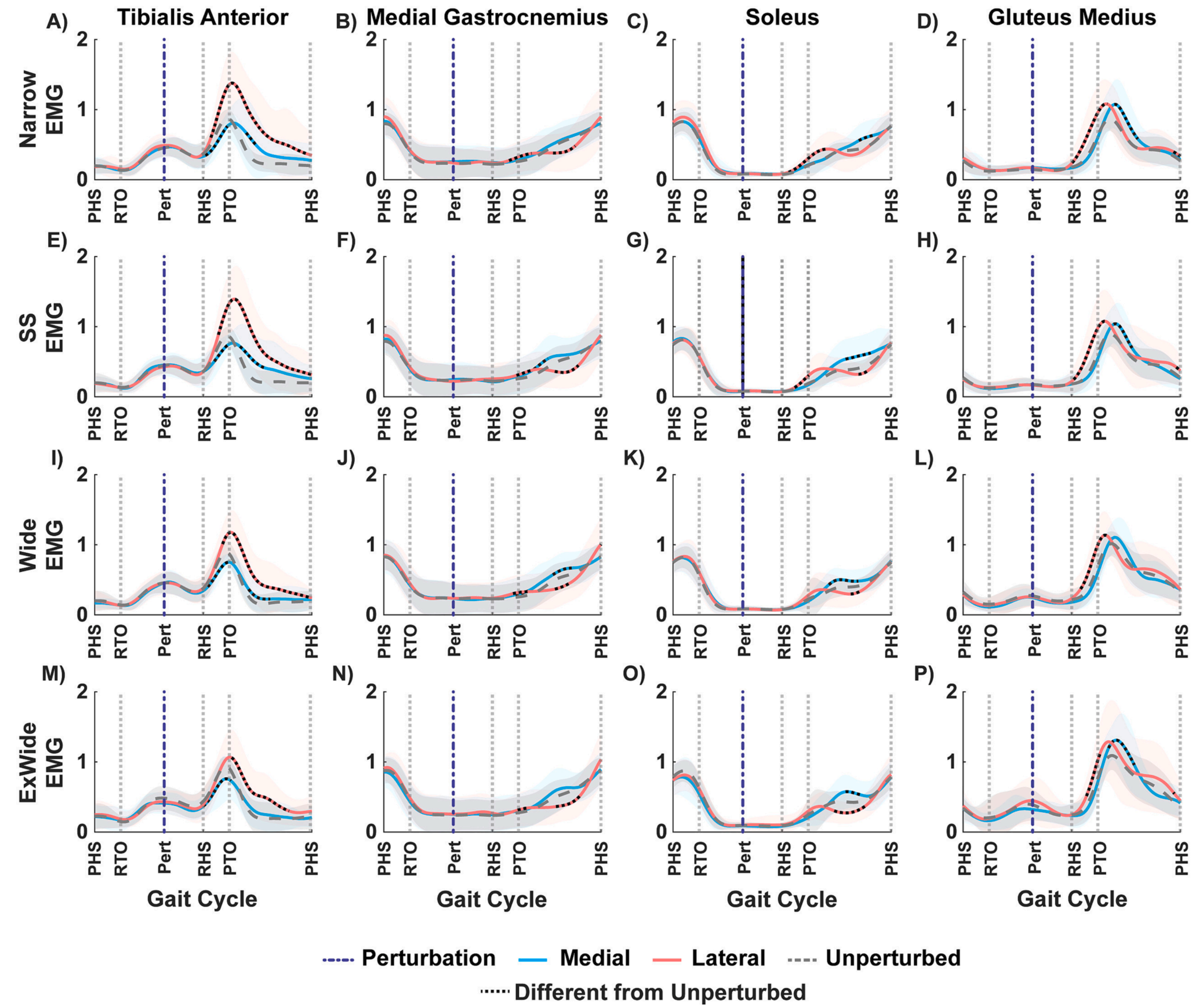

Fig. 9. Electromyography (EMG) signals (± standard deviation) during perturbed and unperturbed walking at narrow **(A-D)**, self-selected (SS) **(E-H)**, wide **(I-L)** and extra-wide (ExWide) **(M−P)** step widths on the *recovery leg* for the tibialis anterior **(A, E, I and M)**, medial gastrocnemius **(B, F, J and N)**, soleus **(C, G, K and O)** and gluteus medius **(D, H, L and P)**. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). Black dotted lines indicate a significant difference in EMG activity from unperturbed trials.

3.2. Balance response strategies to mediolateral perturbations

Perturbations occurred at 32.2 ± 4.4% of the gait cycle while the perturbed leg was in single-leg-stance and the rate of change of H was at its maximum (Fig. 4). Following the perturbation, H_R increased for lateral perturbations and remained unchanged during medial perturbations across all step widths (Fig. 4, Table A4).

Changes in joint moments were consistent across all step widths. *Following medial perturbations,* plantarflexion moment increased during double support on the perturbed leg and single-leg-stance on the recovery leg (Fig. 5, Table A5). On the perturbed leg, ankle inversion decreased during single-leg-stance and increased during double support. During the subsequent recovery step, ankle inversion moment decreased during stance for medial perturbations (Fig. 6, Table A5). Hip abduction moment increased during single-leg-stance on the perturbed and recovery legs (Fig. 7, Table A5). *Following lateral perturbations,* plantarflexion moment decreased during double support on the perturbed leg and single-leg-stance on the recovery leg (Fig. 5, Table A5). On the perturbed leg, ankle inversion moment increased during single-leg-stance and decreased during double support (Fig. 6, Table A5). During the subsequent recovery step, ankle inversion moment increased during double support. Hip abduction moment decreased during single-leg-stance on the perturbed and recovery legs (Fig. 7, Table A5).

Following medial perturbations, the perturbed leg medial gastrocnemius and soleus activity increased during double support across all step widths (Fig. 8, Table A6). During the subsequent recovery step, recovery leg tibialis anterior activity decreased during double support and medial gastrocnemius, soleus and gluteus medius activity increased during single-leg-stance across all step widths (Fig. 9, Table A6). *Following lateral perturbations*, the perturbed leg soleus and gluteus medius activity increased during double support across all step widths (Fig. 8, Table A6). During the subsequent recovery step, recovery leg tibialis anterior activity increased during stance, medial gastrocnemius and soleus activity decreased during single-leg-stance and gluteus medius activity increased

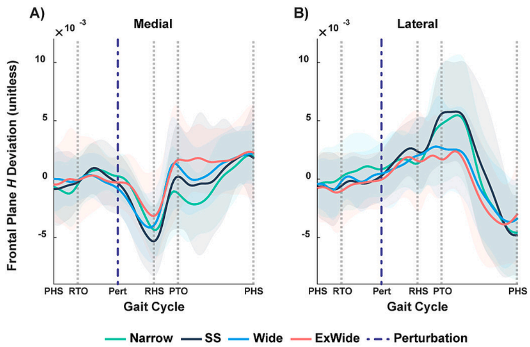

Fig. A1. Frontal-plane whole-body angular momentum (*H*) deviation (± standard deviation) after medial **(A)** and lateral **(B)** perturbations walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). No significant differences were found in deviation of the range of *H* between SS and all other step widths.

during double support across all step widths (Fig. 9, Table A6).

3.3. Balance response strategies compared across step widths

There were no differences between the SS and other step widths for H_R (Fig. A1, Table A7), ankle plantarflexion moment (Fig. A2, Table A8), ankle inversion moment (Fig. A3, Table A8) or hip abduction moment (Fig. A4, Table A8) deviations on either leg.

4. Discussion

The purpose of this study was to determine: 1) how step width alters balance control during steady-state walking, and 2) if step width changes the balance response strategies following mediolateral surface perturbations in healthy young adults. As expected, walking with wider steps led to an increase in H_R and a decrease in ankle inversion moment (Fig. 3). Contrary to our hypotheses, increasing step width did not affect the plantarflexion moment or the balance response strategy used and led to a decrease in hip abduction moment (Fig. 3, Tables A1–A4).

4.1. Balance control during steady-state walking

Consistent with previous studies (McAndrew Young and Dingwell, 2012; Vistamehr et al., 2016; Vistamehr and Neptune, 2021), walking with wide and extra-wide step widths decreased balance control, defined as an increase in H_R (Fig. 3). In contrast, walking with narrow steps did not affect balance control compared to SS steps. These results may partially explain why those that walk with wider steps are more susceptible to falling (Campbell et al., 1981; Forster and Young, 1995; Miller et al., 2001).

In agreement with our hypothesis, ankle inversion moment systematically decreased with increasing step widths (Fig. 3). Walking with a wider step creates a larger mediolateral COM displacement (Orendurff

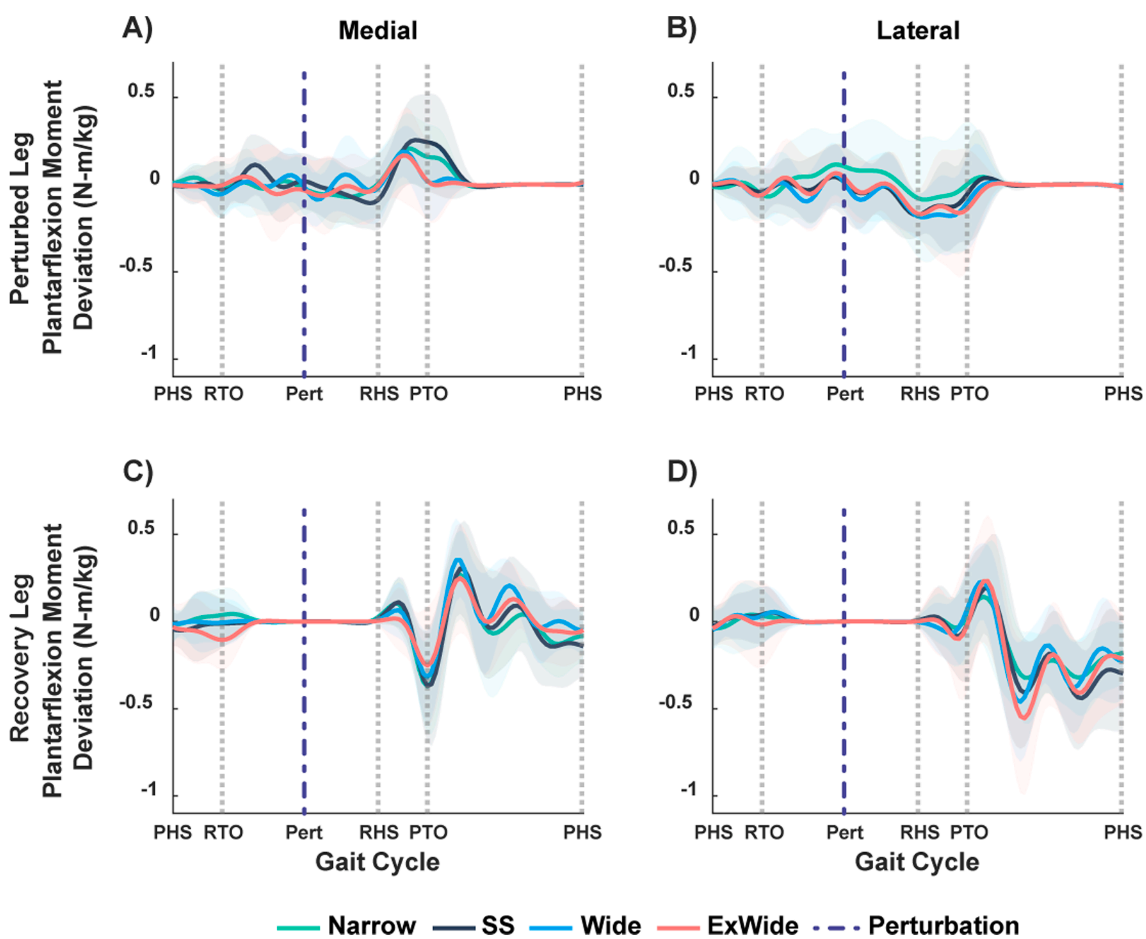

Fig. A2. Ankle plantarflexion moment deviation (± standard deviation) after medial **(A, C)** and lateral **(B, D)** perturbations walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths for the perturbed leg **(A, B)** and recovery leg **(C, D)**. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). No significant differences were found between SS step widths and all other step widths.

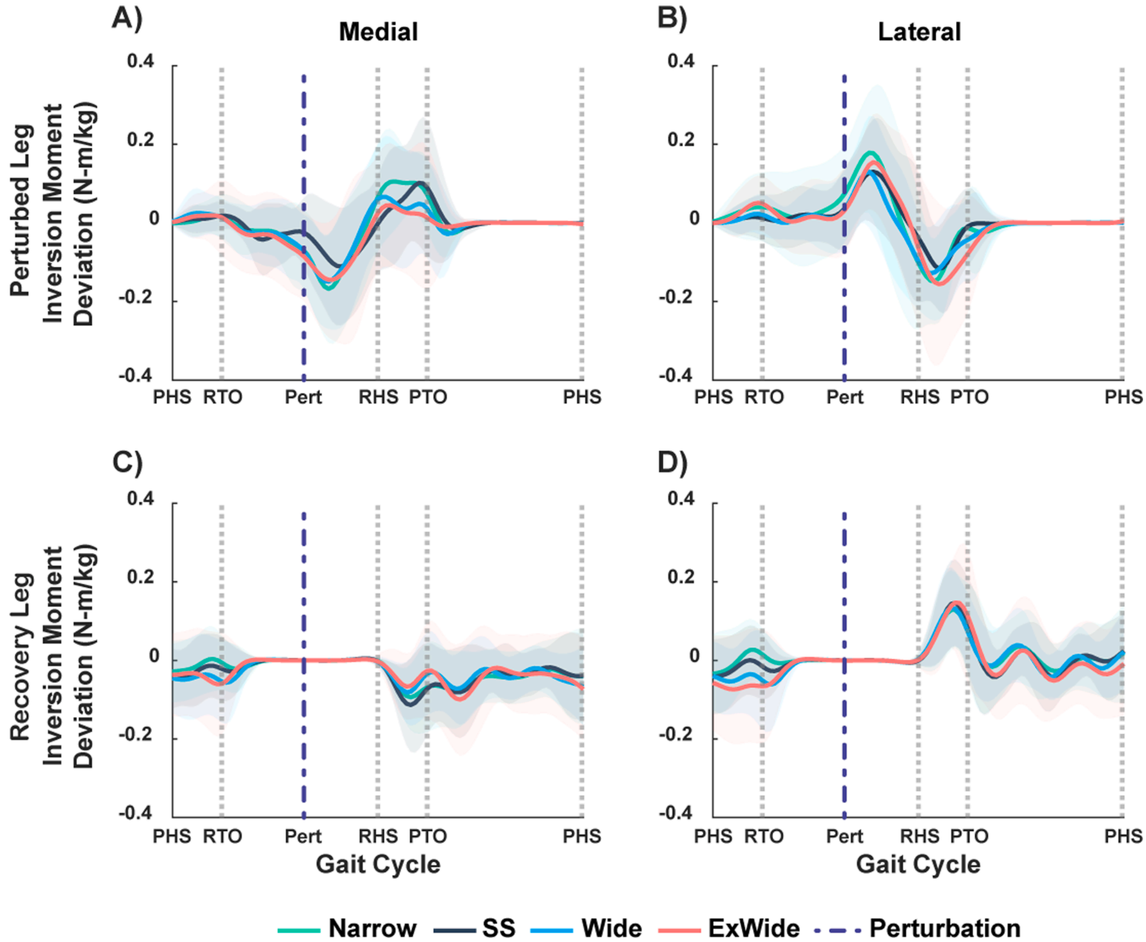

Fig. A3. Ankle inversion moment deviation (± standard deviation) after medial **(A, C)** and lateral **(B, D)** perturbations walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths for the perturbed leg **(A, B)** and recovery leg **(C, D)**. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). No significant differences were found between SS step widths and all other step widths.

et al., 2004), which requires a medial COP shift and an increase in ankle eversion moment (Hof et al., 2010). In addition, at wider steps, we saw a systematic increase in gluteus medius activity (Fig. 3), consistent with others (Kubinski et al., 2015; Rankin et al., 2014), and an increase in the destabilizing external moment that acts to rotate the body towards the swing leg (Fig. 2). Surprisingly, we saw a resulting decrease in hip abduction moment during stance despite the increase in gluteus medius activity. Wider steps may shorten the gluteus medius muscle fibers from their optimal length, which can explain why hip abductor moment decreased despite an increase in gluteus medius activity (Kubinski et al., 2015). In contrast, previous work found that compared to younger adults, older adults increase their hip abduction moment during walking to counteract the higher destabilizing external moment as a result of the increased mediolateral moment arm (Vistamehr and Neptune, 2021). Compared to our results, that study reported smaller moment arms in healthy older adults (0.09 m/leg length) (Vistamehr and Neptune, 2021) compared to our moment arms at wide (0.12 m/leg length) and extra wide (0.17 m/leg length) step widths. Therefore, 0.09 m/leg length may have been below the threshold to negatively influence muscle fiber length. Future modeling and simulation studies should analyze individual muscle states during walking at different step widths to further understand this decrease in hip abduction moment.

Another surprising result was that plantarflexion moment did not increase along with increasing H_R at wider steps (Fig. 3). Previous studies of steady-state walking have shown the plantarflexors are primary contributors to the vertical GRF (Pandy et al., 2010) and frontal-plane external moment that acts to rotate the body towards the swing leg (Neptune and McGowan, 2016). In addition, previous work has shown the ankle plantarflexors are important contributors to body forward propulsion (Neptune et al., 2001). Because we kept treadmill walking speed the same across all step widths, the potential for the plantarflexors to respond to a change in step width may have been limited. Future modeling and simulation studies should investigate changes in individual muscle contributions to frontal-plane external moment at increasing step widths to further understand the lack of increase in plantarflexion moment.

4.2. Balance response strategies to mediolateral perturbations

Across all step widths, H_R increased following lateral perturbations and surprisingly did not change following medial perturbations, suggesting medial perturbations were less challenging to balance control (Fig. 4). The shear force on the bottom of the foot from the translating treadmill caused a change in joint moments, distinct from the subject response to the perturbation. Therefore, we used EMG to distinguish between these changes. The perturbed EMG signals did not deviate from unperturbed walking until the recovery leg heel strike after the perturbation. Therefore, changes in joint moment that occurred before double

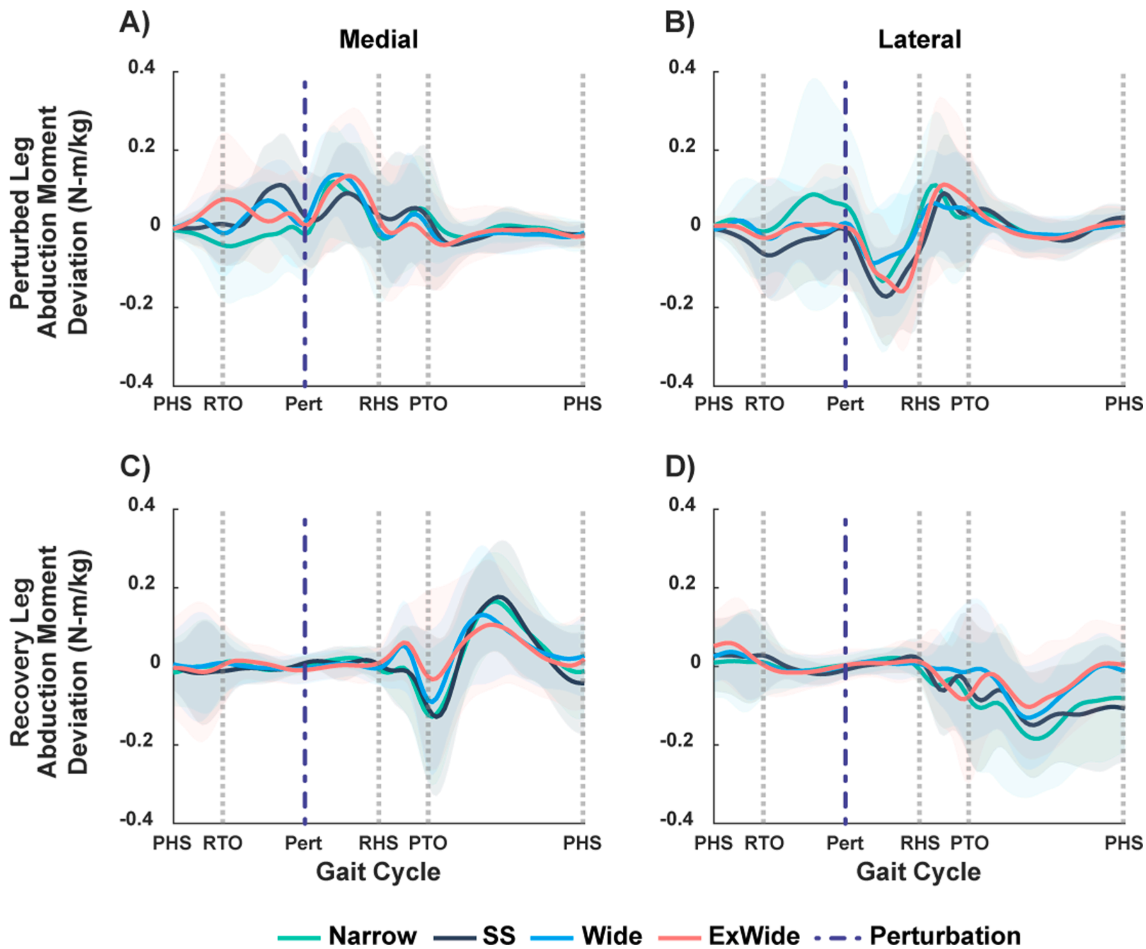

Fig. A4. Hip abduction moment deviation (± standard deviation) after medial **(A, C)** and lateral **(B, D)** perturbations walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths for the perturbed leg **(A, B)** and recovery leg **(C, D)**. Vertical grey dashed lines indicate perturbed side heel strike (PHS) and toe off (PTO) and recovery side heel strike (RHS) and toe off (RTO). No significant differences were found between SS step widths and all other step widths.

Table A1
Demographic information for all participants.

Subject	Height (cm)	Mass (kg)	Speed (m/s)	Sex	Age (years)
1	155.0	59.8	1.34	F	27
2	158.5	68.0	1.53	F	27
3	158.5	64.4	1.17	F	24
4	152.5	58.0	1.3	F	23
5	168.5	73.2	1.24	M	20
6	155.5	52.0	1.51	F	24
7	161.5	50.6	0.89	F	22
8	181.0	76.6	1.34	M	23
9	165.5	66.6	1.38	F	22
10	197.5	75.0	0.98	M	35
11	173.0	85.8	1.33	M	27
12	182.0	90.4	1.37	M	28
13	170.5	56.2	1.39	F	22
14	178.0	78.0	1.31	M	21
15	180.0	80.6	1.01	M	34
Average	169 ± 13	69 ± 12	1.3 ± 0.2	(7 M/8F)	25 ± 4

support were likely not balance responses but mechanical effects of the moving treadmill.

Previous research has highlighted a number of joint-level response strategies important for balance control: a lateral ankle strategy, hip strategy and an ankle push-off strategy (Reimann et al., 2018). The deviation analyses revealed similar joint-level response strategies were used to restore balance across all step widths (Figs. A1-A4). *For lateral perturbations*, individuals used a lateral ankle strategy on the recovery leg to increase inversion moment (Fig. 6), which others have shown acts to shift the COP laterally (e.g., Hof et al., 2010). During double support, we saw a hip strategy where hip abduction moment decreased on the recovery leg (Fig. 7), which acts to shift the COM towards the recovery leg (Winter, 1995). Finally, individuals also used an ankle push-off strategy by decreasing their plantarflexion moment in single-leg-stance on the recovery leg (Fig. 5), which acts to decelerate the COM towards the middle of the base of support (e.g., Kim and Collins, 2013).

For medial perturbations, we also found individuals used similar strategies to recover, despite no change in H_R. These response strategies may have counteracted the perturbation to prevent H_R from changing. Individuals used a lateral ankle strategy by decreasing their inversion moment on the recovery leg (Fig. 6) to shift the COP medially (Hof et al., 2010). Further, a hip strategy was used on the recovery leg to increase abduction moment (Fig. 7) and shift the COM away from the recovery leg (Winter, 1995). Finally, individuals used an ankle push-off strategy by increasing their plantarflexion moment during single-leg-stance on the recovery leg (Fig. 5) to accelerate the COM towards the base of support (Kim and Collins, 2013).

4.3. *Balance response strategies compared across step widths*

Although perturbations elicited a balance response, step width did

Table A2
Frontal-plane external moment and range of whole-body angular momentum (H_R) statistics for unperturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a p-value $\leq$ 0.05.

Outcome Measure	Fixed Effect	Linear Mixed Effects p-value	Follow-Up Paired T-Test p-value
H_R	Step Width	**<0.001**	SS vs. Narrow: 0.110 SS vs. Wide: **0.002** SS vs. ExWide: **<0.001**
Average Frontal-Plane External Moment	Step Width	**<0.001**	SS vs. Narrow: **0.004** SS vs. Wide: **<0.001** SS vs. ExWide: **<0.001**
Vertical Ground Reaction Force (GRF) Component of Average Frontal-Plane External Moment	Step Width	**<0.001**	SS vs. Narrow: <**0.001** SS vs. Wide: **<0.001** SS vs. ExWide: **<0.001**
Mediolateral GRF Component of Average Frontal-Plane External Moment	Step Width	**<0.001**	SS vs. Narrow: <**0.001** SS vs. Wide: **<0.001** SS vs. ExWide: **<0.001**

Table A3
Joint moment and electromyography (EMG) statistics for unperturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a p-value $\leq$ 0.05.

Outcome Measure	ANOVA p-value	Follow-Up Paired T-Test p-value	Regions of Significance (%)
Hip Abduction Moment	**<0.001**	SS vs. Narrow: **<0.001** SS vs. Wide:**<0.001** SS vs. ExWide:**<0.001**	SS vs. Narrow: 4.5–12.9, 69.2–96.3 SS vs. Wide: 4.1–11.7, 19.4–23.4, 42.7–61.6, 65.4–98.6 SS vs. ExWide: 0–1.0, 4.0–28.8, 41.6–63.1, 66.3–100.0
Ankle Inversion Moment	**<0.001**	SS vs. Narrow:**0.016** SS vs. Wide:**<0.001** SS vs. ExWide:**<0.001**	SS vs. Narrow: 25.8–26.5 SS vs. Wide: 17.5–27.7, 30.1–39.9 SS vs. ExWide: 12.8–46.2
Ankle Plantarflexion Moment	1	–	–
Medial Gastrocnemius EMG	**0.012**	SS vs. Narrow: 1 SS vs. Wide: 1 SS vs. ExWide:**<0.001**	SS vs. ExWide: 47.3–64.5
Soleus EMG	**<0.001**	SS vs. Narrow: 1 SS vs. Wide: 1 SS vs. ExWide: **<0.001**	SS vs. ExWide: 2.2–8.0, 20.6–22.9
Tibialis Anterior EMG	**0.001**	SS vs. Narrow: 1 SS vs. Wide: 1 SS vs. ExWide:**<0.001**	SS vs. ExWide: 0–9.1, 97.7–100
Gluteus Medius EMG	**<0.001**	SS vs. Narrow: **0.009** SS vs. Wide:**<0.001** SS vs. ExWide: **<0.001**	SS vs. Narrow: 25.0–30.6 SS vs. Wide: 15.8–34.1, 75.1–94.2 SS vs. ExWide: 9.9–43.7, 72.4–96.3

Table A4
Range of whole-body angular momentum (H_R) statistics for perturbed versus unperturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a p-value $\leq$ 0.05.

Outcome Measure	Fixed Effect	Perturbation Direction	Step Width	Linear Mixed Effects p-value
H_R	Perturbation Condition	Lateral	Narrow	**<0.001**
H_R	Perturbation Condition	Medial	Narrow	0.077
H_R	Perturbation Condition	Lateral	SS	**<0.001**
H_R	Perturbation Condition	Medial	SS	0.578
H_R	Perturbation Condition	Lateral	Wide	**0.002**
H_R	Perturbation Condition	Medial	Wide	0.634
H_R	Perturbation Condition	Lateral	ExWide	**0.005**
H_R	Perturbation Condition	Medial	ExWide	0.634

not affect the response strategies used (Figs. A1-A4), suggesting that healthy young adults have the capacity to respond similarly to surface translation perturbations at a wide range of step widths. Further, postural responses to these surface translation perturbations may be more automatic in nature, and thus the response is always similar.

Other studies have found that gait variability increases at non-self-selected speeds, which suggests individuals adopt different walking strategies under non-self-selected conditions (e.g., Sekiya et al., 1997, Shih et al., 2021). Therefore, forcing participants to walk at non-self-selected step widths could affect their balance response strategies. However, our study did not find differences in the response strategy used across step widths. Since the response to a balance perturbation is critical for fall prevention, that might explain why we did not see differences between self-selected and non-self-selected conditions.

4.4. Limitations and Future work

A potential limitation was that we had subjects target step width

Table A5
Joint moment statistics for perturbed versus unperturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a *p*-value ≤ 0.05.

Measure	Perturbation Direction	Leg	Step Width	T-Test *p*-value	Regions of Significance (%)
Hip Abduction Moment	Lateral	Perturbed	Narrow	**0.018**	39.3–43.2
Hip Abduction Moment	Medial	Perturbed	Narrow	**<0.001**	36.3–44.5
Hip Abduction Moment	Lateral	Perturbed	SS	**<0.001**	35.6–47.3, 55.0–56.2
Hip Abduction Moment	Medial	Perturbed	SS	**0.005**	40.3–45.4
Hip Abduction Moment	Lateral	Perturbed	Wide	**0.010**	36.4–40.8
Hip Abduction Moment	Medial	Perturbed	Wide	**<0.001**	34.6–45.6
Hip Abduction Moment	Lateral	Perturbed	ExWide	**<0.001**	37.4–49.1
Hip Abduction Moment	Medial	Perturbed	ExWide	**<0.001**	37.2–46.5
Hip Abduction Moment	Lateral	Recovery	Narrow	**<0.001**	63.5–66.2, 71.9–89.0
Hip Abduction Moment	Medial	Recovery	Narrow	**<0.001**	60.5–64.7, 72.4–87.8
Hip Abduction Moment	Lateral	Recovery	SS	**<0.001**	72.1–100
Hip Abduction Moment	Medial	Recovery	SS	**<0.001**	71.8–87.3
Hip Abduction Moment	Lateral	Recovery	Wide	**<0.001**	70.5–86.0
Hip Abduction Moment	Medial	Recovery	Wide	**<0.001**	70.8–86.1
Hip Abduction Moment	Lateral	Recovery	ExWide	**0.003**	73.0–80.4
Hip Abduction Moment	Medial	Recovery	ExWide	**<0.001**	54.5–56.9, 74.5–84.1
Ankle Inversion Moment	Lateral	Perturbed	Narrow	**<0.001**	32.2–41.6, 50.1–58.9
Ankle Inversion Moment	Medial	Perturbed	Narrow	**<0.001**	32.0–42.5, 51.5–62.0
Ankle Inversion Moment	Lateral	Perturbed	SS	**<0.001**	33.7–43.3, 52.0–58.4
Ankle Inversion Moment	Medial	Perturbed	SS	**<0.001**	37.8–43.8, 57.9–60.3
Ankle Inversion Moment	Lateral	Perturbed	Wide	**<0.001**	33.5–40.6, 49.0–58.8
Ankle Inversion Moment	Medial	Perturbed	Wide	**<0.001**	32.0–42.9
Ankle Inversion Moment	Lateral	Perturbed	ExWide	**<0.001**	33.7–43.8, 52.1–61.3
Ankle Inversion Moment	Medial	Perturbed	ExWide	**<0.001**	32.0–42.6
Ankle Inversion Moment	Lateral	Recovery	Narrow	**<0.001**	53.6–62.0
Ankle Inversion Moment	Medial	Recovery	Narrow	**<0.001**	51.2–60.8, 66.6–73.4
Ankle Inversion Moment	Lateral	Recovery	SS	**<0.001**	52.7–62.4
Ankle Inversion Moment	Medial	Recovery	SS	**<0.001**	50.7–61.2, 64.9–71.2, 84.8–85.6
Ankle Inversion Moment	Lateral	Recovery	Wide	**<0.001**	52.1–62.4, 80.6–83.4
Ankle Inversion Moment	Medial	Recovery	Wide	**<0.001**	51.1–59.8, 66.0–71.7, 79.5–84.9, 91.9–100.0
Ankle Inversion Moment	Lateral	Recovery	ExWide	**<0.001**	54.1–62.2
Ankle Inversion Moment	Medial	Recovery	ExWide	**<0.001**	54.5–59.0, 67.4–71.5
Ankle Plantarflexion Moment	Lateral	Perturbed	Narrow	1	–
Ankle Plantarflexion Moment	Medial	Perturbed	Narrow	**<0.001**	53.5–62.0
Ankle Plantarflexion Moment	Lateral	Perturbed	SS	**<0.001**	46.3–54.5
Ankle Plantarflexion Moment	Medial	Perturbed	SS	**<0.001**	46.7–48.8, 54.7–62.0
Ankle Plantarflexion Moment	Lateral	Perturbed	Wide	**<0.001**	46.4–60.7
Ankle Plantarflexion Moment	Medial	Perturbed	Wide	**<0.001**	53.1–59.4
Ankle Plantarflexion Moment	Lateral	Perturbed	ExWide	**<0.001**	46.9–53.2
Ankle Plantarflexion Moment	Medial	Perturbed	ExWide	**0.004**	53.4–57.3
Ankle Plantarflexion Moment	Lateral	Recovery	Narrow	**<0.001**	72.7–100.0
Ankle Plantarflexion Moment	Medial	Recovery	Narrow	**<0.001**	50.0–54.9, 58.3–64.5, 68.0–72.2
Ankle Plantarflexion Moment	Lateral	Recovery	SS	**<0.001**	65.5–67.2, 73.0–100.0]
Ankle Plantarflexion Moment	Medial	Recovery	SS	**<0.001**	51.2–55.3, 58.8–64.4, 67.9–72.5, 99.2–100.0
Ankle Plantarflexion Moment	Lateral	Recovery	Wide	**<0.001**	63.4–67.5, 71.0–81.1, 82.9–93.6, 97.0–100.0
Ankle Plantarflexion Moment	Medial	Recovery	Wide	**<0.001**	58.7–64.3, 66.9–72.8, 79.6–85.7
Ankle Plantarflexion Moment	Lateral	Recovery	ExWide	**<0.001**	65.8–66.3, 71.6–82.7, 83.5–95.5, 97.7–100.0
Ankle Plantarflexion Moment	Medial	Recovery	ExWide	**<0.001**	58.7–63.4, 68.2–72.0, 82.8–85.4

lines on the treadmill during perturbed walking. This could cause participants to avoid using a foot placement strategy in response to a perturbation. Similarly, for the joint moment analysis, we removed all cross-over steps. Future studies should further investigate changes in foot placement strategies at altered step widths. Further, our study investigated healthy young adults, who have a lower risk of falling compared to other clinical populations. Pathological conditions and morphological changes in muscle that lead to reduced muscle strength and rate of force development in older adults and those with various neuromuscular impairments likely affect the ability of an individual to successfully respond to balance perturbations. Thus, future studies should investigate differences in response strategies across varying step widths in clinical populations who naturally walk with increased step widths. In addition, the data we analyzed did not provide insight into the causal relationship between muscle activity and joint moments and their contributions to balance control. Future work should use modeling and simulation to gain insight into the influence of step width on muscle contributions to balance control and how these strategies change with step width. Further, this study only investigated surface translation perturbations, limiting the generalizability of our study to other perturbation types. Others have applied foot placement perturbations of similar magnitude and timing as the present study and found different balance response strategies (Brough et al., 2021). Thus, future studies should further investigate the effect of different perturbation types on balance response strategies at varying step widths to further understand the biomechanics of balance control. Finally, other studies have focused on evaluating the orbital stability of human gait (e.g., Perry and Srinivasan, 2017, Sinitksi et al., 2012). One study compared floquet multipliers across different step widths during steady-state treadmill walking and found orbital stability did not change with increasing step widths despite an increase in kinematic variability (e.g., foot placement variance) (Perry and Srinivasan, 2017). Contrary to our study, they concluded wider steps might contribute to stability control as foot placement can be less accurate without changing their orbital stability. However, our study took a different approach by evaluating balance control using a mechanics-based measure within each step and found that step width adversely influenced our measure of balance control. An interesting area of future work would be to assess under what conditions each measure can help inform the other.

Table A6
Electromyography (EMG) statistics for perturbed versus unperturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a *p*-value $\leq$ 0.05.

Muscle	Perturbation Direction	Leg	Step Width	T-Test *p*-value	Regions of Significance (%)
Tibialis Anterior	Lateral	Perturbed	Narrow	**<0.001**	50.5–51.9, 55.0–62.0
Tibialis Anterior	Medial	Perturbed	Narrow	**0.034**	58.5–62.0
Tibialis Anterior	Lateral	Perturbed	SS	**0.035**	58.7–62.0
Tibialis Anterior	Medial	Perturbed	SS	**0.037**	59.1–62.0
Tibialis Anterior	Lateral	Perturbed	Wide	**0.050**	61.7–62.0
Tibialis Anterior	Medial	Perturbed	Wide	**0.044**	60.1–62.0
Tibialis Anterior	Lateral	Perturbed	ExWide	**0.040**	59.1–62.0
Tibialis Anterior	Medial	Perturbed	ExWide	**0.042**	59.3–62.0
Gastroc	Lateral	Perturbed	Narrow	1	–
Gastroc	Medial	Perturbed	Narrow	**0.009**	54.8–62.0
Gastroc	Lateral	Perturbed	SS	1	–
Gastroc	Medial	Perturbed	SS	**0.008**	55.0–62.0
Gastroc	Lateral	Perturbed	Wide	1	–
Gastroc	Medial	Perturbed	Wide	**0.006**	54.0–62.0
Gastroc	Lateral	Perturbed	ExWide	1	–
Gastroc	Medial	Perturbed	ExWide	**0.016**	55.3–62.0
Soleus	Lateral	Perturbed	Narrow	**0.008**	49.4–55.5
Soleus	Medial	Perturbed	Narrow	**0.002**	52.5–62.0
Soleus	Lateral	Perturbed	SS	1	–
Soleus	Medial	Perturbed	SS	**0.009**	55.9–62.0
Soleus	Lateral	Perturbed	Wide	**0.004**	53.0–61.4
Soleus	Medial	Perturbed	Wide	**0.007**	55.3–62.0
Soleus	Lateral	Perturbed	ExWide	**<0.001**	42.8–58.8
Soleus	Medial	Perturbed	ExWide	**0.045**	59.7–62.0
Gluteus Medius	Lateral	Perturbed	Narrow	**0.050**	61.5–62.0
Gluteus Medius	Medial	Perturbed	Narrow	**<0.001**	50.0–62.0
Gluteus Medius	Lateral	Perturbed	SS	1	–
Gluteus Medius	Medial	Perturbed	SS	**0.002**	49.6–60.3
Gluteus Medius	Lateral	Perturbed	Wide	1	–
Gluteus Medius	Medial	Perturbed	Wide	1	–
Gluteus Medius	Lateral	Perturbed	ExWide	1	–
Gluteus Medius	Medial	Perturbed	ExWide	**0.024**	47.4–52.3
Tibialis Anterior	Lateral	Recovery	Narrow	**<0.001**	53.5–98.1
Tibialis Anterior	Medial	Recovery	Narrow	**<0.001**	51.0–61.2, 69.8–74.2
Tibialis Anterior	Lateral	Recovery	SS	**<0.001**	57.0–100.0
Tibialis Anterior	Medial	Recovery	SS	**<0.001**	55.1–62.2, 66.9–79.0
Tibialis Anterior	Lateral	Recovery	Wide	**<0.001**	58.9–99.2
Tibialis Anterior	Medial	Recovery	Wide	**<0.001**	52.1–62.4, 70.9–80.9
Tibialis Anterior	Lateral	Recovery	ExWide	**<0.001**	63.1–89.5
Tibialis Anterior	Medial	Recovery	ExWide	**<0.001**	50.0–62.8
Gastroc	Lateral	Recovery	Narrow	**<0.001**	57.6–68.1, 77.9–88.2
Gastroc	Medial	Recovery	Narrow	1	–
Gastroc	Lateral	Recovery	SS	**<0.001**	61.1–68.4, 78.7–89.7
Gastroc	Medial	Recovery	SS	**0.012**	69.1–75.7
Gastroc	Lateral	Recovery	Wide	**<0.001**	59.5–66.6, 78.8–85.8, 99.8–100.0
Gastroc	Medial	Recovery	Wide	**<0.001**	60.7–66.5, 77.9–86.5
Gastroc	Lateral	Recovery	ExWide	**<0.001**	61.7–65.9, 79.3–90.3
Gastroc	Medial	Recovery	ExWide	1	–
Soleus	Lateral	Recovery	Narrow	**<0.001**	53.0–70.2
Soleus	Medial	Recovery	Narrow	**0.006**	85.2–92.1
Soleus	Lateral	Recovery	SS	**<0.001**	55.8–62.3, 85.1–88.9
Soleus	Medial	Recovery	SS	**<0.001**	79.1–90.7
Soleus	Lateral	Recovery	Wide	**0.025**	81.5–85.5
Soleus	Medial	Recovery	Wide	**<0.001**	71.5–85.0
Soleus	Lateral	Recovery	ExWide	**<0.001**	73.9–88.7
Soleus	Medial	Recovery	ExWide	**0.005**	79.1–86.2
Gluteus Medius	Lateral	Recovery	Narrow	**<0.001**	50.2–66.5, 96.9–100
Gluteus Medius	Medial	Recovery	Narrow	**<0.001**	67.7–80.5
Gluteus Medius	Lateral	Recovery	SS	**<0.001**	50.9–65.2, 98.1–100
Gluteus Medius	Medial	Recovery	SS	**0.012**	69.3–77.8
Gluteus Medius	Lateral	Recovery	Wide	**<0.001**	54.1–66.2
Gluteus Medius	Medial	Recovery	Wide	1	–
Gluteus Medius	Lateral	Recovery	ExWide	**<0.001**	56.8–63.3, 96.9–97.3
Gluteus Medius	Medial	Recovery	ExWide	**<0.001**	67.0–78.8

Table A7
Deviation of whole-body angular momentum (H_R) statistics for perturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a p-value ≤ 0.05.

Outcome Measure	Fixed Effect	Perturbation Direction	Linear Mixed Effects *p*-value	Follow-Up Paired T-Test *p*-value
Deviation of H_R	Step Width	Lateral	**0.035**	SS vs. Narrow: 0.9013 SS vs. Wide: 0.0667SS vs. ExWide: 0.551
Deviation of H_R	Step Width	Medial	0.463	–

Table A8
Deviation of joint moment statistics for perturbed walking at narrow, self-selected (SS), wide and extra-wide (ExWide) step widths. Bolded values denote a p-value $\leq$ 0.05.

Measure	Perturbation Direction	Leg	ANOVA *p*-value	Follow-Up Paired T-Test *p*-value
Hip Abduction Moment Deviation	Lateral	Perturbed	**0.019**	SS vs. Narrow: 1 SS vs. Wide: 1SS vs. ExWide: 1
Hip Abduction Moment Deviation	Medial	Perturbed	1	–
Hip Abduction Moment Deviation	Lateral	Recovery	1	–
Hip Abduction Moment Deviation	Medial	Recovery	1	–
Ankle Inversion Moment Deviation	Lateral	Perturbed	1	–
Ankle Inversion Moment Deviation	Medial	Perturbed	1	–
Ankle Inversion Moment Deviation	Lateral	Recovery	1	–
Ankle Inversion Moment Deviation	Medial	Recovery	1	–
Ankle Plantarflexion Moment Deviation	Lateral	Perturbed	1	–
Ankle Plantarflexion Moment Deviation	Medial	Perturbed	1	–
Ankle Plantarflexion Moment Deviation	Lateral	Recovery	1	–
Ankle Plantarflexion Deviation	Medial	Recovery	**0.020**	SS vs. Narrow: 1 SS vs. Wide: 1SS vs. ExWide: 1

5. Conclusion

These analyses suggest walking with a wider step systematically increases H_R, increases stance gluteus medius activity and reduces the hip abduction and ankle inversion moments. Further, walking with a wider step does not change the balance response strategy in healthy young adults following a mediolateral surface translation perturbation, which suggests the perturbation was not challenging enough to require recruitment of additional strategies to maintain balance. Ultimately, this study provides a baseline for comparing balance response mechanisms across step widths for individuals with neuromuscular impairments.

CRediT authorship contribution statement

Lindsey K. Molina: Conceptualization, Data curation, Writing - original draft, Visualization, Investigation, Validation, Formal analysis, Methodology, Project administration. **Gabriella H. Small:** Writing – review & editing, Methodology, Data curation, Conceptualization. **Richard R. Neptune:** Conceptualization, Writing - review & editing, Visualization, Investigation, Methodology, Supervision, Resources, Project administration.

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.

Appendix A

See Figs. A1–A4 and Table A1–A8.

References

Andrews, A.W., Thomas, M.W., Bohannon, R.W., 1996. Normative Values for Isometric Muscle Force Measurements Obtained With Hand-held Dynamometers. Phys. Ther. 76, 248–259.

Brough, L.G., Klute, G.K., Neptune, R.R., 2021. Biomechanical response to mediolateral foot-placement perturbations during walking. J. Biomech. 116, 110213.

Campbell, A.J., Reinken, J., Allan, B., Martinez, G., 1981. Falls in old age: a study of frequency and related clinical factors. Age Ageing 10, 264–270.

Forster, A., Young, J., 1995. Incidence and consequences of falls due to stroke: A systematic inquiry. BMJ 311, 83.

Hof, A.L., Bockel, R.M.V., Schoppen, T., Postema, K., 2007. Control of lateral balance in walking. Experimental findings in normal subjects and above-knee amputees. Gait Posture 25, 250–258.

Hof, A.L., Vermerris, S.M., Gjaltema, W.A., 2010. Balance responses to lateral perturbations in human treadmill walking. J. Exp. Biol. 213, 2655–2664.

Kim, M., Collins, S.H., 2013. Stabilization of a three-dimensional limit cycle walking model through step-to-step ankle control. IEEE Int. Conf. Rehabil. Robot. 13–18.

Kubinski, S.N., McQueen, C.A., Sittloh, K.A., Dean, J.C., 2015. Walking with wider steps increases stance phase gluteus medius activity. Gait Posture 41, 130–135.

Larsson, L., Degens, H., Li, M., Salviati, L., Lee, Y., Thompson, W., Kirkland, J.L., 2019. Sarcopenia: Aging-Related Loss of Muscle Mass and Function. Physiol. Rev. 99, 427–511.

Leeuwen, A.M. Van, Bruijn, S.M., Dieën, J.H. Van, 2022. Stabilization of gait, mechanisms, and opportunities for training. SportRxiv.

McAndrew Young, P.M., Dingwell, J.B., 2012. Voluntarily changing step length or step width affects dynamic stability of human walking. Gait Posture 35, 472–477.

Miller, W.C., Speechley, M., Deathe, B., 2001. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch. Phys. Med. Rehabil. 82, 1031–1037.

Neckel, N., Pelliccio, M., Nichols, D., Hidler, J., 2006. Quantification of functional weakness and abnormal synergy patterns in the lower limb of individuals with chronic stroke. J. Neuroeng. Rehabil. 3, 1–11.

Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J. Biomech. 34, 1387–1398.

Neptune, R.R., McGowan, C.P., 2016. Muscle contributions to frontal plane angular momentum during walking. J. Biomech. 49, 2975–2981.

Neptune, R.R., Vistamehr, A., 2019. Dynamic balance during human movement: Measurement and control mechanisms. J. Biomech. Eng. 141, 1–10.

Nott, C.R., Neptune, R.R., Kautz, S.A., 2014. Relationships between frontal-plane angular momentum and clinical balance measures during post-stroke hemiparetic walking. Gait Posture 39, 129–134.

Orendurff, M.S., Segal, A.D., Klute, G.K., Berge, J.S., Rohr, E.S., Kadel, N.J., 2004. The effect of walking speed on center of mass displacement. J. Rehabil. Res. Dev. 41, 829–834.

Owings, T.M., Grabiner, M.D., 2004. Step width variability, but not step length variability or step time variability, discriminates gait of healthy young and older adults during treadmill locomotion. J. Biomech. 37, 935–938.

Painter, J.A., Allison, L., Dhingra, P., Daughtery, J., Cogdill, K., Trujillo, L.G., 2012. Fear of Falling and Its Relationship With Anxiety, Depression, and Activity Engagement Among Community-Dwelling Older Adults. Am. J. Occup. Ther. 66, 169–176.

Pandy, M.G., Lin, Y.C., Kim, H.J., 2010. Muscle coordination of mediolateral balance in normal walking. J. Biomech. 43, 2055–2064.

Pataky, T.C., 2012. Computer Methods in Biomechanics and Biomedical Engineering One-dimensional statistical parametric mapping in Python. Comput. Methods Biomech. Biomed. Engin. 15, 295–301.

Perry, J.A., Srinivasan, M., 2017. Walking with wider steps changes foot placement control, increases kinematic variability and does not improve linear stability. Royal Society open science 4 (9), 160627.

Rankin, B.L., Buffo, S.K., Dean, J.C., 2014. A neuromechanical strategy for mediolateral foot placement in walking humans. J. Neurophysiol. 112, 374–383.

Reimann, H., Fettrow, T., Jeka, J.J., 2018. Strategies for the control of balance during locomotion. Kinesiol. Rev. 7, 18–25.

Roerdink, M., Lamoth, C.J.C., Kwakkel, G., Wieringen, P.C.W.V., Beek, P.J., 2007. Gait Coordination After Stroke: Treadmill Walking. Phys. Ther. 87, 1009–1022.

Rubenstein, L.Z., 2006. Falls in older people: epidemiology, risk factors and strategies for prevention. Age Ageing 35, 37–41.

Segal, A.D., Klute, G.K., 2014. Lower-limb amputee recovery response to an imposed error in mediolateral foot placement. J. Biomech. 47, 2911–2918.

Sekiya, N., Nagasaki, H., Ito, H., Furuna, T., 1997. Optimal walking in terms of variability in step length. J Orthop Sports Phys Ther. 5, 266–272.

Shih, H.S., Gordon, J., Kulig, K., 2021. Trunk control during gait: Walking with wide and narrow step widths present distinct challenges. J Biomech. 114, 110135.

Sinitksi, E.H., Terry, K., Wilken, J.M., Dingwell, J.B., 2012. Effects of perturbation magnitude on dynamic stability when walking in destabilizing environments. J. Biomech. 45 (12), 2084–2091.

Vistamehr, A., Kautz, S.A., Bowden, M.G., Neptune, R.R., 2016. Correlations between measures of dynamic balance in individuals with post-stroke hemiparesis. J. Biomech. 49, 396–400.

Vistamehr, A., Neptune, R.R., 2021. Differences in balance control between healthy younger and older adults during steady-state walking. J. Biomech. 128, 110717.

Winter, D., 1995. Human balance and posture control during standing and walking. Gait Posture 3, 193–214.

Zeni Jr., J.A., Richards, J.G., Higginson, J.S., 2008. Two simple methods for determining gait events during treadmill and overground walking using kinematic data. Gait Posture 27, 710–714.