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Lower-limb joint quasi-stiffness in the frontal and sagittal planes during walking at different step widths

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A R T I C L E I N F O

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A B S T R A C T

Quasi-stiffness describes the intersegmental joint moment–angle relationship throughout the progression of a task. Previous work has explored sagittal-plane ankle quasi-stiffness and its application for the development of powered lower-limb assistive devices. However, frontal-plane quasi-stiffness remains largely unexplored but has important implications for the development of exoskeletons since clinical populations often walk with wider steps and rely on frontal-plane balance recovery strategies at the hip and ankle. This study aimed to characterize frontal-plane hip and ankle quasi-stiffness during walking and determine how step width affects quasi-stiffness in both the frontal and sagittal planes. Kinematic and kinetic data were collected and quasi-stiffness values computed for healthy young adults ($n = 15$) during treadmill walking across a range of step widths. We identified specific subphases of the gait cycle that exhibit linear and quadratic frontal-plane quasi-stiffness approximations for the hip and ankle, respectively. In addition, we found that at wider step widths, sagittal-plane ankle quasi-stiffness increased during early stance (~12–35% gait cycle), sagittal-plane hip quasi-stiffness decreased in late stance (~40–55% gait cycle) and frontal-plane hip quasi-stiffness decreased during terminal stance (~48–65% gait cycle). These results provide a framework for further exploration of frontal-plane quasi-stiffness, lend insight into how quasi-stiffness may relate to balance control at various step widths, and motivate the development of stiffness-modulating assistive devices to improve balance related outcomes.

1. Introduction

Dynamic joint stiffness, also referred to as *quasi-stiffness* (Davis and DeLuca, 1996), quantifies the simultaneous changes in intersegmental joint angle and moment throughout the progression of a task. During human movement, quasi-stiffness differs from passive joint stiffness in that quasi-stiffness includes the effect of active muscle force generation, which can generate positive or negative work, in addition to passive soft tissue energy storage and return (Rouse et al., 2013).

Sagittal-plane ankle quasi-stiffness has been used to develop control strategies for powered ankle prostheses, exoskeletons and orthotic devices to provide assistance during various phases of the gait cycle (e.g., Au and Herr, 2008; Caputo and Collins, 2013; Rouse et al., 2014). These devices aim to improve various aspects of walking performance such as walking speed as well as reduce the metabolic cost of walking and/or compensations needed from the unaffected limb (e.g., Hedrick et al., 2019; Herr and Grabowski, 2011). However, improving balance control and decreasing fall risk are rarely considered when developing quasi-stiffness-based controllers. Although current assistive devices primarily focus on emulating sagittal-plane dynamics, frontal-plane quasi-stiffness remains largely unexplored but has important implications given that maintaining frontal-plane balance requires more active control than the sagittal plane (Bauby and Kuo, 2000).

Clinical populations, such as those with lower-limb prostheses or neuromuscular impairments, have poorer frontal-plane balance control compared to healthy adults (e.g., Nolasco et al., 2021; Pickle et al., 2014; Silverman and Neptune, 2011) and are at an increased risk of falling. These populations also often walk with wider step widths (Hof et al., 2007; Kurz et al., 2012; Roerdink et al., 2007), which are associated with poorer balance control in healthy young (Molina et al., 2023) and older (Vistamehr and Neptune, 2021) adults. At wider step widths, a larger moment arm between the body center-of-mass (COM) and center-of-pressure (COP) leads to a greater destabilizing moment due to gravity (MacKinnon and Winter, 1993). Thus, walking with wider steps likely requires greater neuromuscular control to maintain balance and prevent a fall.

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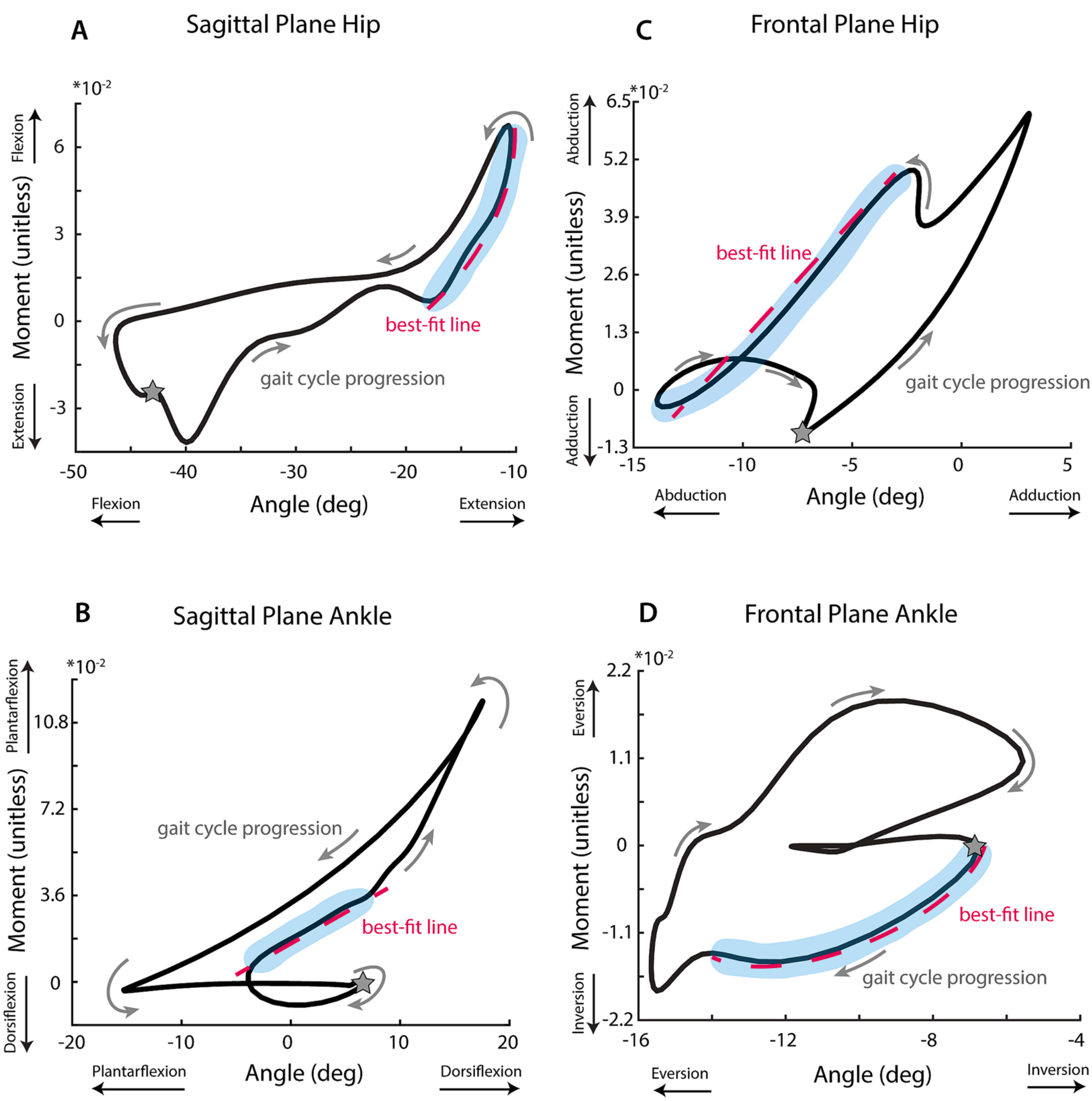

Fig. 1. Representative moment–angle loops for the hip and ankle in the sagittal (A-B) and frontal (C-D) planes. Star indicates the start of the gait cycle (heel-strike).

Commonly used strategies for maintaining balance during walking include altered foot placement, lateral ankle, ankle push-off and hip strategies (Reimann et al., 2018). Consistent with these strategies, the ankle plantarflexors and hip abductors are primary contributors to mediolateral COM acceleration (Pandy et al., 2010) and frontal-plane whole-body angular momentum (Neptune and McGowan, 2016), which is a mechanics-based measure of dynamic balance that is correlated with clinical balance measures (Nott et al., 2014; Vistamehr et al., 2016). Previous studies investigating balance control found that healthy young adults rely more on a lateral ankle strategy than a hip strategy to maintain balance even at wider step widths (Molina et al., 2023) whereas older adults rely more on a hip strategy (Vistamehr and Neptune, 2021). These studies evaluated joint moments to determine which balance control strategies were used to counteract the increasing destabilizing moment about the body COM at wider step widths. However, how quasi-stiffness at the hip and ankle joints change with step width to maintain balance is more complex as it is influenced by both the joint moment and angle.

The objective of this study was to characterize frontal-plane hip and ankle quasi-stiffness in healthy young adults as well as determine how quasi-stiffness in both the sagittal and frontal planes varies across a range of step widths. We hypothesized that when healthy young adults walk at wider step widths, quasi-stiffness would be primarily modulated at the ankle joint to maintain balance. We expected that ankle quasi-stiffness in both the sagittal and frontal planes would increase at wider step widths and that hip quasi-stiffness would not change in either plane at wider step widths. Understanding how hip and ankle quasi-stiffness relate to specific balance control strategies would provide important insight into the concurrent joint moment and angle responses as well as provide the foundation for future studies seeking to identify differences between quasi-stiffness in healthy individuals and clinical populations. In addition, the characterization of quasi-stiffness in response to altered step widths would help inform the development of biomimetic controllers for assistive devices to improve balance control and decrease fall risk.

Table 1
Subphases of the moment–angle loops evaluated and corresponding gait events used to determine the start and end of each subphase. Gait events used were developed based on methods from [1] (Shamaei et al., 2013b), [2] (Crenna and Frigo, 2011) and [3] (Shamaei et al., 2013a).

Joint	Plane	Phase	Approximate % Gait Cycle	Start of Phase Event	End of Phase Event
Ankle	Sagittal	Dorsi-Flexion	12-35%	First max plantarflexion angle[1]	Avg: Local min VGRF ~30% gait cycle[1] Zero AP GRF[1]
		Dual-Flexion	35-51%	Avg: Local min VGRF ~30% gait cycle[1] Zero AP GRF[1]	Avg: Max dorsiflexion moment[2] Max dorsiflexion angle[1]
		Rising	12-51%	First max plantarflexion angle[1]	Avg: Max dorsiflexion moment[2] Max dorsiflexion angle[1]
		Falling	51-65%	Avg: Max dorsiflexion moment[2] Max dorsiflexion angle[1]	Toe-off[1]
	Frontal	Early Eversion	1-18%	Heel-Strike	First peak ML GRF
Hip	Sagittal	Extension	40-55%	Local discontinuity in second derivative of flexion moment with respect to angle ~40% gait cycle[3]	Max flexion moment[3]
		Flexion	55-65%	Max flexion moment[3]	Toe-off[3]
	Frontal	Adduction	3-21%	First min abduction moment	First max abduction moment
		Transition	21-48%	First max abduction moment	Local min ML GRF ~35% gait cycle
		Abduction	48-65%	Second max abduction moment	Toe-off

2. Methods

2.1. Data collection

Kinematic and kinetic data were previously collected (Molina et al., 2023) from fifteen healthy young adults (8 female; age: 25 ± 4 years; height: 169 ± 13 cm; mass: 69 ± 12 kg). All participants were free from musculoskeletal or neuromuscular impairments and provided written informed consent approved by The University of Texas at Austin Institutional Review Board. Three-dimensional full-body kinematic data were collected at 120 Hz using a 10-camera motion capture system (Vicon, Oxford, UK) and a set of 56 retroreflective markers. Three-dimensional ground reaction force (GRF) data were collected at 960 Hz using a split-belt instrumented treadmill (Motek, Amsterdam, NL).

Participants completed four 30-second steady-state walking trials at the following step widths: Narrow (25% narrower), Self-Selected (SS), Wide (50% wider) and Extra-Wide (100% wider). Pilot testing determined a 25% decrease in step width was the narrowest width possible to avoid a substantial number of crossover steps (i.e., foot lands on both treadmill belts), while the wider step width magnitudes were chosen to maximize the opportunity for seeing an effect of step width. Each trial was performed at the participant's SS walking speed, which was determined using the average of three 10-meter overground walking trials. Each participant's SS step width was determined by averaging the mediolateral distance between markers placed on the calcanei during heel-strikes of 20 consecutive steps at their SS speed. A custom D-flow (Motek, Amsterdam, NL) script was developed to project continuous foot placement targets on the treadmill surface at each prescribed step width.

2.2. Data processing and analysis

Kinematic and GRF data were low-pass filtered at 8 Hz and 15 Hz, respectively, using a fourth-order double-pass Butterworth filter. A 13-segment inverse dynamics model was created for each participant in Visual3D (C-Motion, Germantown, USA) and used to compute lower-limb joint angles and moments. Individual strides were determined and crossover steps were removed using custom MATLAB (MathWorks, Natick, USA) scripts.

Table 2
RMSE values for linear and quadratic least-squares regressions of each subphase. Best fit type (linear or quadratic) used for calculating quasi-stiffness during each subphase was chosen as linear for subphases in which there was less than 10% difference in RMSE between fit types to prevent overfitting the data. For subphases in which there was greater than 10% difference in RMSE, the fit with the lower RMSE was chosen.

Joint	Plane	Phase	Linear RMSE	Quadratic RMSE	% Difference in RMSE	Fit Type Used
Ankle	Sagittal	Dorsi-Flexion	1.70E-04	1.57E-04	8.0	Linear
		Dual-Flexion	1.78E-04	1.74E-04	2.3	Linear
		Rising	2.15E-04	1.89E-04	12.9	Quadratic
		Falling	1.76E-04	1.53E-04	14.0	Quadratic
	Frontal	Early Eversion	2.58E-05	2.33E-05	10.2	Quadratic
Hip	Sagittal	Extension	1.20E-04	1.05E-04	13.3	Quadratic
		Flexion	8.10E-05	7.32E-05	10.1	Quadratic
	Frontal	Adduction	1.76E-04	1.69E-04	4.1	Linear
		Transition	6.56E-05	6.41E-05	2.3	Linear
		Abduction	9.05E-05	8.58E-05	5.3	Linear

Sagittal and frontal-plane joint moments were normalized by each participant's body height and weight. Moment-angle loops (Fig. 1) in the sagittal plane were divided into subphases of stance similar to previous studies (Crenna and Frigo, 2011; Nigro and Arch, 2022; Shamaei et al., 2013a, 2013b, 2013c). Similarly, we defined subphases for the frontal-plane moment–angle loops using characteristic frontal-plane hip and ankle angle, moment and GRF events outlined in Table 1.

Each subphase of the moment–angle loops was fit with either a linear or quadratic polynomial least-squares regression following the methods in (Nigro et al., 2021) to determine the best fit. Briefly, we initially computed both linear and quadratic polynomial least-squares regressions for each subphase, then compared RMSE values between the linear and quadratic fits on a group level. If there was less than 10% difference in RMSE between the fit types, we used a linear fit to prevent overfitting the data. If there was greater than 10% difference in RMSE, we used the fit with the lower RMSE, indicating a better fit for that subphase.

Quasi-stiffness (K_{QS}) approximates the relationship between joint angle (θ) and corresponding moment (M) for a given phase as:

$$M = K_{QS} * \theta \tag{1}$$

For linear-fit conditions (Eq. (2)), K_{QS} remains constant throughout the duration of the phase and is equal to the linear fit coefficient (A).

$$M = A * \theta \tag{2}$$

where,

$$K_{QS} = A \tag{3}$$

However, for quadratic-fit conditions (Eq. (4)), K_{QS} changes throughout the phase as it is angle-dependent. As such, the median quasi-stiffness throughout the duration of the phase was taken as:

$$M = A\theta^2 + B\theta = (A\theta + B) * \theta \tag{4}$$

where,

$$K_{QS} = median(A\theta + B) \tag{5}$$

2.3. Statistical analyses

For each subphase, a linear mixed-effects model determined the relationship between quasi-stiffness (K_{QS}) and step width. K_{QS} was the dependent variable, and the fixed factors included the limb (left or right) as well as step width condition (Narrow, SS, Wide or Extra-Wide). Step width conditions were included as ordinal categorical variables to account for the relationship between step width sizes. The random factors included participant, the interaction between limb and participant as well as a random intercept. Significance was defined as $\alpha = 0.05$. In the case of a significant effect of step width, Bonferroni-adjusted follow-up paired t-tests were performed to compare K_{QS} between the SS step width versus each of the other step width conditions as well as the most extreme case of Narrow versus Extra-Wide. Although the scope of this work focused on hip and ankle quasi-stiffness as they relate most closely to balance control strategies, we similarly evaluated quasi-stiffness responses of the knee across step widths for completeness (see Appendix A for the knee analyses). Since previous studies have noted that healthy individuals exhibit between-limb differences in lower-limb joint angles (Forczek and Staszkiewicz, 2012) and moments (Lambach et al., 2014) during walking, data from both limbs were included in our linear mixed-effects model. In the case of a significant limb effect, follow-up paired t-tests were performed to compare K_{QS}, pooled across step widths, between the right and left limb.

3. Results

3.1. Quasi-stiffness calculation

Hip and ankle quasi-stiffness values were derived for each subphase using either linear or quadratic approximations of moment–angle relationships. The fit type used for each subphase and corresponding RMSE are listed in Table 2.

3.2. Sagittal-plane quasi-stiffness

3.2.1. Hip

During the Extension phase, quasi-stiffness decreased when walking at the Extra-Wide step widths compared to SS ($p = 0.006$) and Narrow ($p < 0.001$) widths (Fig. 2A). During the Flexion phase, quasi-stiffness did not change with step width ($p > 0.096$) (Fig. 2A).

3.2.2. Ankle

During the Dorsi-Flexion phase, quasi-stiffness increased when walking at the Extra-Wide step widths compared to SS ($p < 0.001$) and Narrow ($p < 0.001$) widths (Fig. 2B). During the Dual-Flexion phase, there were no changes across step widths ($p > 0.362$) (Fig. 2B). For the entire Rising phase, which spans both the Dorsi-Flexion phase and the Dual-Flexion phase, there was a trend of quasi-stiffness increasing with increasing step width (Fig. 2B). However, this difference did not reach significance with the Bonferroni adjustment applied ($p = 0.022$). During the Falling phase, there were no changes in quasi-stiffness with step

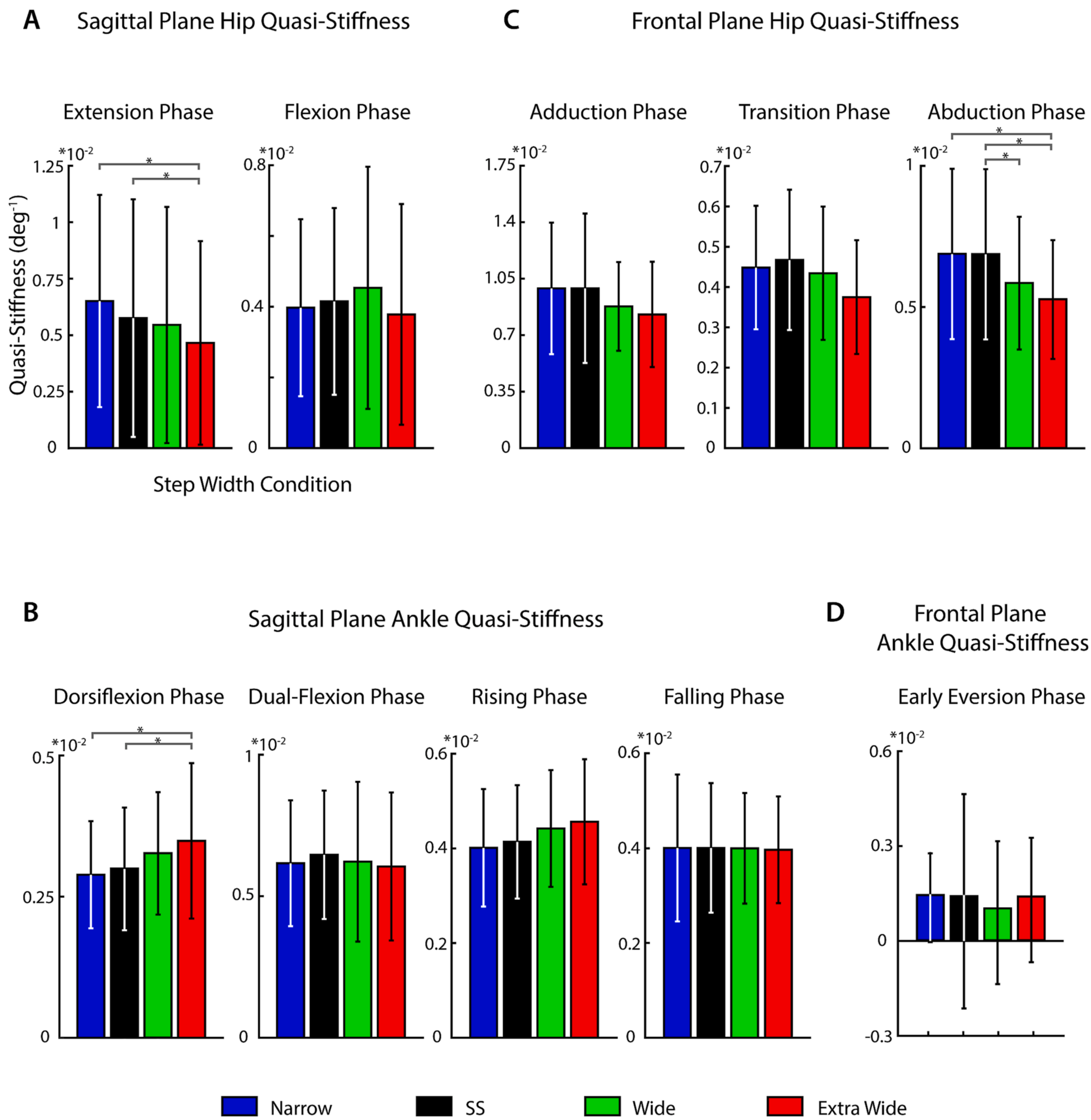

Fig. 2. Quasi-stiffness values across step widths: A) Sagittal-plane hip quasi-stiffness decreased when walking at wider step widths during the Extension phase but does not change during the Flexion phase. B) Sagittal-plane ankle quasi-stiffness increased when walking at wider step widths only during the Dorsi-Flexion phase. C) Frontal-plane hip quasi-stiffness decreased when walking at wider step widths only during the Abduction phase. D) Frontal-plane ankle quasi-stiffness did not change across step widths. "*" indicates significant difference between step width conditions.

width ($p > 0.227$) (Fig. 2B).

3.3. Frontal-plane quasi-stiffness

3.3.1. Hip

During the Adduction phase, there was a trend of quasi-stiffness decreasing with increasing step width (Fig. 2C). However, this difference did not reach significance with the Bonferroni adjustment ($p = 0.021$). During the Transition phase (Fig. 2C), there was no change in quasi-stiffness across step widths ($p > 0.146$). During the Abduction phase (Fig. 2C), quasi-stiffness decreased with increased step width ($p < 0.001$). For between-limb differences in frontal-plane hip quasi-stiffness, see Appendix B.

3.3.2. Ankle

During the Early Eversion phase (Fig. 2D), quasi-stiffness did not change with step width ($p > 0.220$). Due to notable variation in the shape of the frontal-plane ankle moment–angle loops (Fig. 3), additional characteristic subphases could not be defined to further explore the effect of step width on frontal-plane ankle quasi-stiffness.

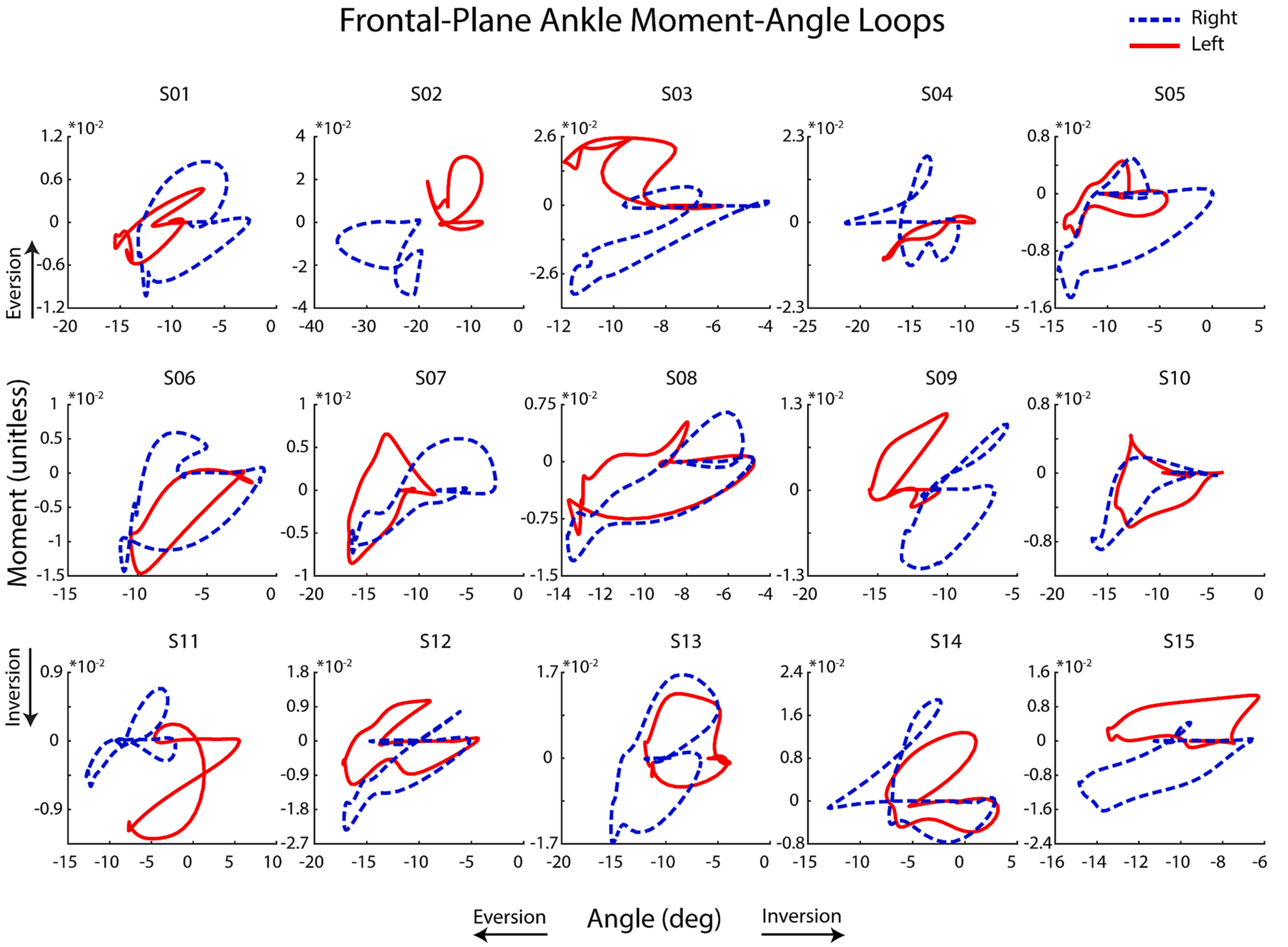

Fig. 3. Frontal-plane ankle moment–angle loops for each participant. There were notable variations in shape between participants and between the right (dashed) and left (solid) limbs of the same participant.

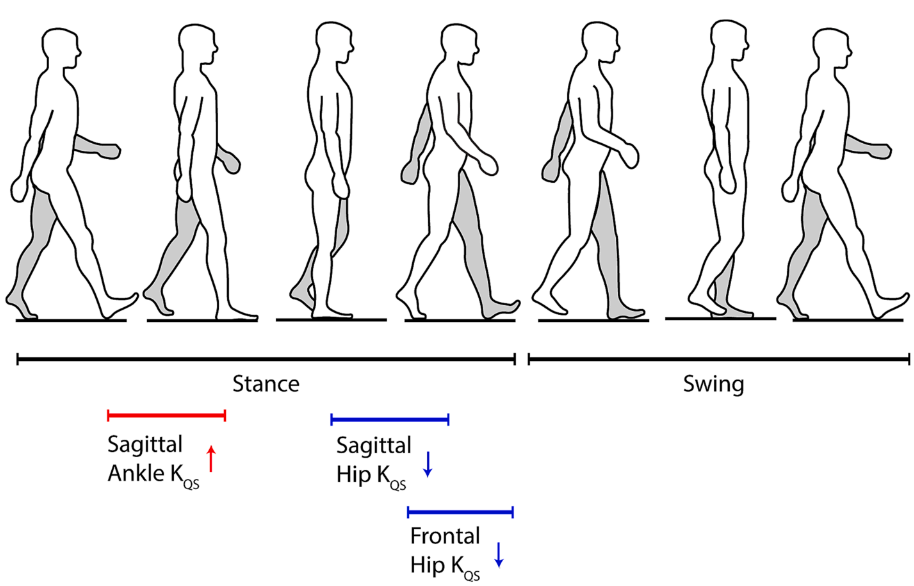

Fig. 4. Summary of changes in sagittal and frontal-plane hip and ankle quasi-stiffness values at increased step widths throughout the gait cycle.

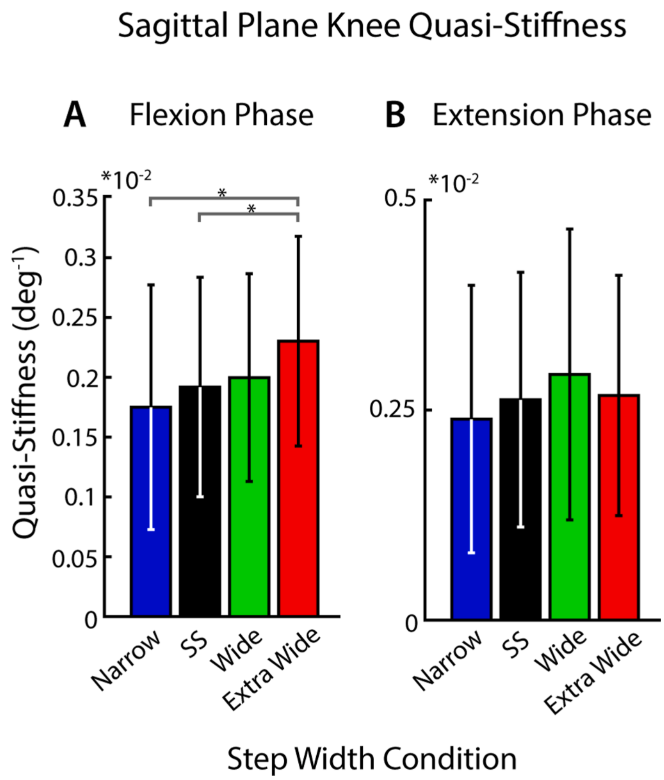

Figure A1. Quasi-stiffness values across step widths: A) Sagittal-plane knee quasi-stiffness increased when walking at wider step widths during the Flexion phase but did not change during the B) Extension phase. "*" indicates significant difference between step width conditions.

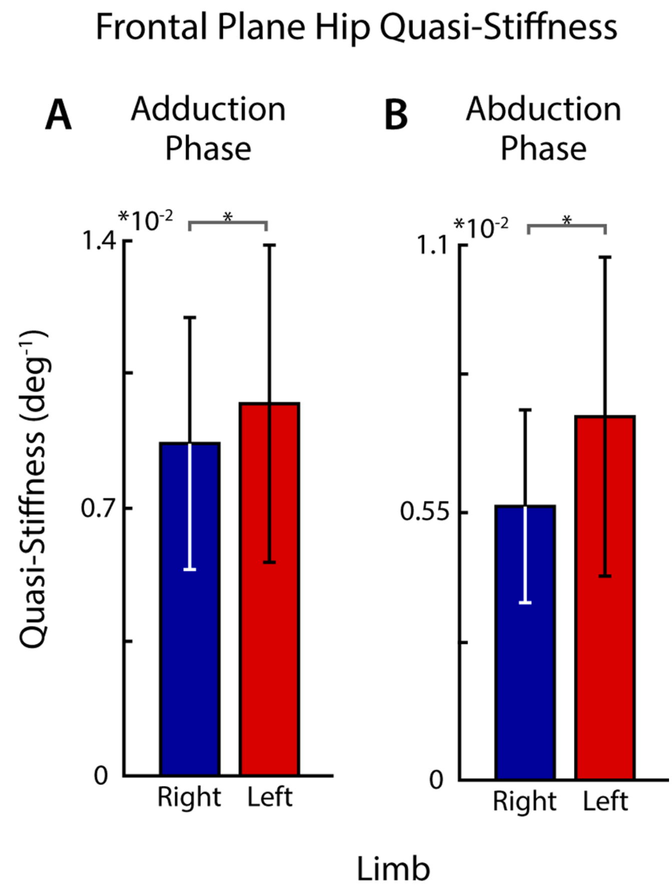

Figure B1. Frontal-plane hip quasi-stiffness. The right leg demonstrated lower quasi-stiffness compared to the left leg during the A) Adduction and B) Abduction phases. "*" indicates significant difference between step width conditions.

4. Discussion

This study aimed to characterize frontal-plane hip and ankle quasi-stiffness in healthy young adults and determine how quasi-stiffness in both the sagittal and frontal planes varies across step widths. We hypothesized that ankle quasi-stiffness would increase at wider step widths. This hypothesis was partially supported, as sagittal-plane ankle quasi-stiffness increased at wider step widths during the Dorsi-Flexion phase. Contrary to our hypothesis, frontal-plane ankle quasi-stiffness did not change with step width. Finally, we hypothesized that hip quasi-stiffness would not change with step width. However, notable decreases in hip quasi-stiffness at wider step widths occurred at sub-phases in both the sagittal (Extension phase) and frontal (Abduction phase) planes.

4.1. Characterization of frontal-plane quasi-stiffness

Quasi-stiffness was determined using linear and quadratic approximations to describe joint moment–angle relationships throughout the walking gait cycle. Frontal-plane hip quasi-stiffness was characterized with a set of linear curves during the Adduction, Transition and Abduction phases, where we observed low RMSE values indicating a strong fit (Table 2). Powered ankle prostheses and exoskeletons often utilize sagittal-plane ankle quasi-stiffness approximations from healthy populations to define their controllers (e.g., Au and Herr, 2008; Caputo and Collins, 2013). Similarly, hip exoskeleton devices with frontal-plane actuation could utilize quasi-stiffness values similar to those in the present study. Currently, few hip exoskeletons with frontal-plane actuation have been developed (Alili et al., 2023; Chiu et al., 2021). However, the quasi-stiffness profiles characterized here provide a framework for control schemes that can make future designs more approachable.

Frontal-plane ankle quasi-stiffness was characterized using a quadratic curve to describe the Early Eversion phase of the gait cycle (Table 2). Throughout the rest of the gait cycle, there were notable variations in the shape of the frontal-plane ankle moment–angle loops across participants (Fig. 3) as well as differences between limbs within participants. As such, only the Early Eversion phase of the gait cycle could be fit with a consistent curve to reasonably represent the group data. High inter- and intra-participant differences in the shape of the frontal-plane ankle moment–angle loops is likely due to the small magnitude of the moment arm between the ankle joint center and COP. Given the heterogeneity of the frontal-plane ankle moment–angle loop shapes, it would likely be difficult to establish a general control scheme to prescribe frontal-plane ankle quasi-stiffness (e.g., in a powered assistive device) that suits a range of users. Considering that lateral ankle modulation is an effective balance control strategy (Reimann et al., 2018), subject-specific stiffness profiles may be warranted to improve balance control and decrease fall risk.

4.2. Quasi-Stiffness across step widths

At wider step widths, a number of quasi-stiffness values changed: sagittal-plane ankle quasi-stiffness increased during the Dorsi-Flexion phase in early stance, sagittal-plane hip quasi-stiffness decreased during the Extension phase in late stance, and frontal-plane hip quasi-stiffness decreased during the Abduction phase in terminal stance (Fig. 4). In contrast, frontal-plane ankle quasi-stiffness remained unchanged.

During the Dorsi-Flexion phase in early stance, sagittal-plane ankle quasi-stiffness increased at wider step widths. Interestingly, previous work assessing joint moments across step widths in healthy young adults found no change in ankle plantarflexion moment throughout the gait cycle at various step widths (Molina et al., 2023). Given the relationship between joint moment and quasi-stiffness (Eq. (1)), we performed *post-hoc* statistical parametric mapping (Pataky, 2010) t-tests comparing ankle plantarflexion angle across step width conditions to determine if the observed change in quasi-stiffness during the Dorsi-Flexion phase was consistent with a change in angle. We confirmed that ankle plantarflexion moment did not change across conditions in agreement with previous work (Molina et al., 2023) but that ankle plantarflexion angle did decrease at Extra-Wide step widths ($p = 0.001$) during the Dorsi-Flexion phase. Thus, the observed increase in quasi-stiffness was more related to a reduction in ankle plantarflexion angle throughout this phase, rather than an increase in plantarflexion moment. We similarly determined that at wider step widths, knee flexion angle increased during the stance phase ($p = 0.001$) but decreased during the swing phase ($p = 0.001$). Taken together, these results reflect kinematic characteristics of stiff-knee gait (Lewek et al., 2012; van der Krogt et al., 2010). Clinical populations who commonly demonstrate stiff-knee gait (e.g., individuals post-stroke or with cerebral palsy) often walk with increased step widths (Kurz et al., 2012; Roerdink et al., 2007) and exhibit poorer balance control (Kurz et al., 2012; Nott et al., 2014) compared to healthy individuals. Likewise, simply walking at wider step widths resulted in healthy adults walking with stiff-knee gait characteristics (present study) and demonstrating poorer balance control (Molina et al., 2023). Given that comprehensive experimental protocols may be too demanding for individuals with neuromuscular impairments to perform, studying healthy individuals walking at wide step widths could be used as a surrogate to help inform targets for improving balance control in clinical populations.

Sagittal-plane hip quasi-stiffness decreased at wider step widths during the Extension phase, which may involve energy storage during loading to then be returned during the Flexion phase (Shamaei et al., 2013a). Interestingly, quasi-stiffness during the Flexion phase did not change with step width. *Post-hoc* pairwise comparisons of Extension versus Flexion quasi-stiffness values at each step width revealed no differences at Wide or Extra-Wide step widths ($p > 0.046$) after Bonferroni adjustment. Therefore, at wider step widths (i.e., Wide and Extra Wide), the quasi-stiffness values during the Extension and Flexion phases were approximately equal, suggesting that the hip demonstrates spring-like behavior during this phase. However, at the Narrow and SS step width conditions, the Extension phase demonstrated higher quasi-stiffness values than the Flexion phase ($p < 0.0104$). In agreement with Shamaei et al., (2013a) who evaluated quasi-stiffness across walking speeds and found that the hip demonstrated spring-like behavior only in certain speed conditions, we similarly found that spring-like behavior was only present when walking at wider step widths. Since the Extension and Flexion phases have differing responses to changing step width, sagittal-plane hip behavior cannot be simply replicated using a spring with scalable stiffness values across step widths.

During the Abduction phase at terminal stance, frontal-plane hip quasi-stiffness decreased at wider step widths. Given that quasi-stiffness is in part regulated by muscle force generation across a joint (Rouse et al., 2013), changes in hip quasi-stiffness relating to changes in step width are likely accompanied by changes in muscle activity that modulate frontal and sagittal-plane hip motion. Roelker et al. (2019) found that hip flexors, extensors and abductors are primary contributors to controlling anteroposterior and mediolateral foot placement, which is an important strategy for maintaining balance control. Furthermore, Akl et al. (2023) noted that changes in sagittal-plane hip quasi-stiffness due to walking speed are associated with changes in hip flexor/extensor coactivation. The hip strategy, driven by activation of the lateral spinal muscles (MacKinnon and Winter, 1993), has been suggested to correct for errors in foot placement (MacKinnon and Winter, 1993) as well as maintain balance when mediolateral foot placement is tightly constrained (Reimann et al., 2018) such as in the current study. Shih and Kulig (2021) noted that spinal muscle coactivation decreases at wider step widths, which agrees with our finding of decreased frontal-plane hip quasi-stiffness at wider step widths. Taken together, these results suggest that changes in hip quasi-stiffness across step widths may reflect changes in the utilization of balance control strategies, as wider steps also lead to decreased balance control (Molina et al., 2023). Understanding how quasi-stiffness changes with step width is an important step in understanding balance control and mitigating fall risk in populations who walk at wider steps.

4.3. Limitations and recommendations for future work

A potential limitation of this study was that we only evaluated quasi-stiffness in healthy individuals during steady-state walking. Therefore, this work does not capture how clinical populations who walk with wider step widths may employ different balance control strategies compared to healthy individuals. Furthermore, the balance control strategies used during steady-state walking likely involve more subtle COM modulation compared to a balance response to a larger perturbation necessary to prevent a fall. Future work should explore how lower limb quasi-stiffness changes during a perturbation compared to steady-state walking, as well as how joint angles and moments each contribute to changes in quasi-stiffness across walking conditions. Finally, quasi-stiffness was only evaluated during the stance phase of the gait cycle, as sagittal and frontal-plane ankle moments approach zero during swing phase. As a result, our quasi-stiffness analysis may not fully capture the effects of other balance control strategies that occur during the swing phase. Furthermore, frontal-plane ankle quasi-stiffness could not be evaluated for the entire stance phase due to notable heterogeneity in the shape of the moment–angle loops.

5. Conclusion

This study aimed to characterize frontal-plane hip and ankle quasi-stiffness during walking and determine how hip and ankle quasi-stiffness in both the sagittal and frontal planes responds to walking at various step widths. The results identified subphases of the gait cycle that exhibit linear and quadratic frontal-plane moment–angle relationships for the hip and ankle, respectively. In addition, we found that at wider step widths, sagittal-plane ankle quasi-stiffness increased during the Dorsi-Flexion phase in early stance, sagittal-plane hip quasi-stiffness decreased during the Extension phase in late stance and frontal-plane hip quasi-stiffness decreased during the Abduction phase in terminal stance. These results provide insight into how changes in quasi-stiffness may relate to changes in balance control strategies used at various step widths as well as provide a framework for developing stiffness-modulating assistive devices for improving balance related outcomes.

CRediT authorship contribution statement

Stephanie L. Molitor: Conceptualization, Visualization, Methodology, Investigation, Formal analysis, Writing – original draft. **Richard R. Neptune:** Conceptualization, Methodology, Resources, Writing – review and editing, Visualization, Supervision, 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.

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Appendix A

Sagittal-plane knee quasi-stiffness across step widths

Sagittal-plane knee moment-angle loops were divided into subphases of stance similar to methods described by Shamaei et al. (2013c). Quasi-stiffness was calculated during knee Flexion and Extension phases using a linear fit for each phase. The effect of step width on quasi-stiffness was explored using a mixed linear effects model as described in the Methods for the main body of this work. During the Flexion phase, sagittal-plane knee quasi-stiffness increased when walking at the Extra-Wide step widths compared to the SS ($p<0.001$) and Narrow ($p<0.001$) widths (Fig. A1). During the Extension phase, sagittal-plane knee quasi-stiffness did not change across step widths ($p>0.081$) (Fig. A1).

Appendix B

Influence of limb on quasi-stiffness

A *post-hoc* comparison of quasi-stiffness values between the left and right limbs revealed an influence of limb only for frontal-plane hip quasi-stiffness. During both the Adduction ($p=0.007$) and Abduction ($p<0.001$) phases, the right leg demonstrated lower quasi-stiffness than the left leg (Fig. B1).

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