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The influence of increased passive stiffness of the trunk and hips on balance control during reactive stepping

Steven P. Pretty, Daniel P. Armstrong, Tyler B. Weaver, Andrew C. Laing*

Injury Biomechanics and Aging Laboratory, Department of Kinesiology, University of Waterloo, 200 University Ave West, Waterloo, Ontario, N2L 3G1, Canada

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ABSTRACT

Background: Age-related changes, which include increased trunk and hip stiffness, negatively influence postural balance. While previous studies suggest no net-effect of trunk and hip stiffness on initial trip-recovery responses, no study to date has examined potential effects during the dynamic restabilisation phase following foot contact.
Research question: Does increased trunk and hip stiffness, in isolation from other ageing effects, negatively influence balance during the restabilisation phase of reactive stepping.
Methods: Balance perturbations were applied using a tether-release paradigm, which required participants to react with a single-forward step. Sixteen young adults completed two blocks of testing: a baseline and an increased stiffness (corset) condition. Whole-body kinematics were utilized to estimate spatial step parameters, center of mass (COM), COM incongruity (peak - final position) and time to restabilisation, in anteroposterior (AP) and mediolateral (ML) directions.
Results: In the corset condition, peak COM displacement was increased in both directions ($p < 0.024$), which drove reductions in minimum margins of stability ($p < 0.032$) as step width and length were unchanged ($p > 0.233$). Increased passive stiffness also increased the magnitude and variability of peak shear ground reaction force, COM incongruity, and time to restabilisation in the ML (but not AP) direction ($p < 0.027$).
Significance: In contrast to previous literature, increased stiffness resulted in greater peak COM displacement in both directions. Our results suggest increased trunk and hip stiffness have detrimental effects on dynamic stability following a reactive step, particularly in the ML direction. Observed increases in magnitude and variability of COM incongruity suggest the likelihood of a sufficiently large loss of ML stability - requiring additional steps - was increased by stiffening of the hips and trunk. The current findings suggest interventions aiming to mobilize the trunk and hips, in conjunction with strengthening, could improve balance and reduce the risk of falls.

1. Introduction

Falls are a major health problem among older adults, resulting in injuries that lead to increased morbidity and mortality [1–3]. The increased fall incidence observed with ageing is, at least in part, due to balance detriments including reduced movement speed [4,5], decreased strength [6,7] and chronic musculoskeletal pain [8].

As the trunk constitutes a large proportion of body weight and has a large influence on center of mass (COM) location, age-related changes in control of trunk movements may be an important factor in fall risk [9]. Increased trunk stiffness has been suggested to be a key biomechanical change with age [10] and disease state [11] that interferes with compensatory trunk movements and results in abnormal motion following a perturbation [10]. Difficulty in controlling trunk stability and reduced trunk flexibility is associated with an increased risk of falls during gait [12,13] as well as following large balance disturbances [14].

Contributions of motion at the hips and trunk to balance recovery have been studied using healthy young adults whose hips and lower trunk were artificially stiffened using a rigid corset [15,16]. This experimental paradigm provides an empirical basis for investigating the isolated role of reduced joint motion/ increased stiffness (associated with aging/pathology) on the underlying mechanisms and/or success of balance recovery attempts. Utilizing a pitch-roll platform, Grünenberg and colleagues [15] demonstrated that artificially increasing trunk and hip stiffness with thoracolumbosacral orthosis in young adults resulted in a reversal of medial-lateral (ML) trunk motion characterized by a destabilizing trunk motion in the direction of the impending fall. This

* Corresponding author at: Department of Kinesiology, Faculty of Applied Health Sciences, University of Waterloo, 200 University Avenue West, Waterloo, ON, N2L 3G1, Canada.
E-mail address: actlaing@uwaterloo.ca (A.C. Laing).

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pattern has previously been observed among older adults and attributed to increased roll (ML) stiffness [10], suggesting the young adults in this study could not modify movement strategy to sufficiently account for changes in link flexibility with the corset.

While negative effects on feet-in-place postural balance have been observed [10,15], the role of trunk and hip stiffness on larger balance disturbances, such as trips during gait, is less clear. Trips are a leading cause of falls [17] and attempts to recover balance via reactive stepping have been noted in 42% of falls observed in long-term care [18]. Theoretically, increased stiffness may attenuate any perturbation to trunk posture during a trip, providing torques to resist the angular momentum of the trunk and reducing gravitational moment arms acting on the COM. However, increased trunk stiffness may additionally hamper compensatory trunk motions, complicating the net effect on balance recovery. This framework is supported by the findings of van der Burg and colleagues [16], whom concluded no net effect of trunk stiffness on reactive stepping balance. During simulated trips with young adults and thoracolumbosacral orthosis, increased stiffness was found to decrease trunk acceleration following a perturbation, but later in the positioning phase (immediately before stepping foot contact), no significant differences in trunk posture were found [16]. This analysis however focused on only anterior-posterior trunk motion and did not account for the potential effects of trunk stiffness following foot contact during a reactive step, an important phase in regulating COM position and arresting the motion of the body.

Failure to effectively control the COM following a reactive step results in the need for subsequent steps and is associated with increased fall risk compared to single steppers [19]. Older adults have been found to require additional steps as a result of lateral instability, primarily in the direction of the unsupported leg during a reactive step [20,21]. Focusing on the initial step, in older adults a greater ML ground reaction force (GRF) is required to recover balance, compared to young adults [22]. After initial foot contact, there is a restabilisation phase in which an individual must control their COM through reactive control of applied forces and postural adjustments to regain stability [23,24]. Utilizing a tether release paradigm, Singer and colleagues [24] observed that older adults have greater and more variable ML incongruity (overshoot toward the unsupported side) during this restabilisation phase, as well as took 50% longer to regain stability compared to young adults. These findings suggest a function of dynamic stability dyscontrol [23,24] but it is unclear how trunk and hip stiffness in isolation from other ageing effects contribute to ML instability following a reactive step, and if the same effects are present in the AP direction. A comprehensive understanding of balance recovery following a trip should consider stability in both the direction of the perturbation (AP), as well as laterally (ML).

Thus, the purpose of this study was to investigate the role of increased passive hip and trunk stiffness on the restabilisation phase of a reactive step. It was hypothesized that during the dynamic reactive stepping response, increased passive stiffness would illicit no differences in peak shear GRF, COM displacement, and minimum margin of stability in the 1) AP and 2) ML directions. However, it was further hypothesized that increased passive stiffness would negatively influence the restabilisation phase of a reactive step in both the 3) AP and 4) ML directions as inferred through increased time to restabilisation and COM incongruity (overshoot). A secondary analysis of the trial to trial variability of each outcome variable was also completed. Greater variability could suggest increased probability of a failed single step recovery, requiring additional steps, even in the absence of differences in mean magnitudes. These findings may provide insights into the mechanisms underlying dynamic stability control observed in older adults, and in the longer term, inform clinical interventions focused on reactive step training.

2. Materials and methods

2.1. Participants

Sixteen healthy young adults (8 males) participated in this study (mean (SD) age = 22.5 (2.5) years; height = 1.70 (1.10) m; mass = 68.4 (9.7) kg). Exclusion criteria included any anatomical or neurological impairments with the potential to influence balance. All participants provided written informed consent. This study was approved by the Office of Research Ethics at the University of Waterloo.

2.2. Instrumentation

A 3-dimensional motion capture system (Optotrak Certus, NDI, Waterloo, ON, Canada), a force platform (OR6-7, Advanced Mechanical Technology Incorporated, Watertown, MA, USA), and two load cells (MLP-300-CO, Transducer Techniques, Temecula, CA, USA) were used to acquire whole body kinematics (100 Hz), step kinetics (2000 Hz), and tether force data (2000 Hz) respectively. An 11-segment kinematic model was generated utilizing 4-marker rigid body clusters placed on each foot, and thigh, as well as individual markers placed on the acromion, lateral epicondyle and ulnar styloid bilaterally. Participants were anchored to a rigid steel frame, via adjustable cables attached to a safety harness above the left and right ilium (Fig. 1). Each cable was placed in series with a load cell to monitor tether force symmetry and magnitude.

2.3. Protocol

Balance perturbations were applied using a tether-release paradigm [24,25], which required participants to react with a single forward step onto a force platform and remain in a forward stance configuration for 10 s [24]. Participants were instructed to adopt an initial foot position (standardized to 50% hip width) and forward lean equivalent to 15% of body weight with arms folded across the chest (Fig. 1). This lean magnitude was selected based on pilot work in which participants were unable to maintain balance with feet in place responses and were able to successfully recover balance with a single step. To ensure consistency, tether force was monitored in real-time throughout the interval prior to release. Cable release occurred at unpredictable time intervals following adoption of the forward lean, via release of an electromagnet (model DCA-400 T-24C, AEC Magnetics, Cincinnati, OH, USA).

Two blocks consisting of 10 trials (5 releases and 5 catch trials) were completed. In one block, passive stiffness of the trunk and hip was increased using a rigid, plastic corset, which crossed the hips and extended upwards to the xiphoid process (Fig. 1). Maximal forward flexion bending tests were completed prior to each block of releases to assess the stiffening effect of the corset. The corset reduced maximal forward flexion by an average of 42.6 degrees (mean (SD) = 91.4 (15.1) vs. 48.8 (13.5); $p < 0.001$) and the means of both conditions were within 7 degrees of those reported by van der Burg et al. [16]. The influence of the corset on AP and ML range of motion at the trunk and hips was additionally assessed and is presented in Appendix A Supplementary data. The order of block (baseline vs. increased passive stiffness), and trial (release vs. catch) were randomized.

2.4. Data analysis

All data processing was performed using customized software routines (MATLAB version 7.10, Mathworks, Natick, MA, USA). Gaps in kinematic data (< 200 ms) were interpolated using a cubic spline [26]. All data was low-pass filtered with 2nd order, dual pass, Butterworth filters with effective cut-off frequencies of 6 Hz [27], 50 Hz [24], and 3 Hz [25], for kinematic, force platform, and tether data respectively. Peak GRF produced by the stepping leg in the ML ($Shear_{ML}$) and AP

Fig. 1. a) Experimental tether-release setup. The corset crossed the hips anteriorly and posteriorly and was secured to the thigh using compression straps; **b)** Corset anterior view with compression straps removed; **c)** Corset posterior view with compression straps removed.

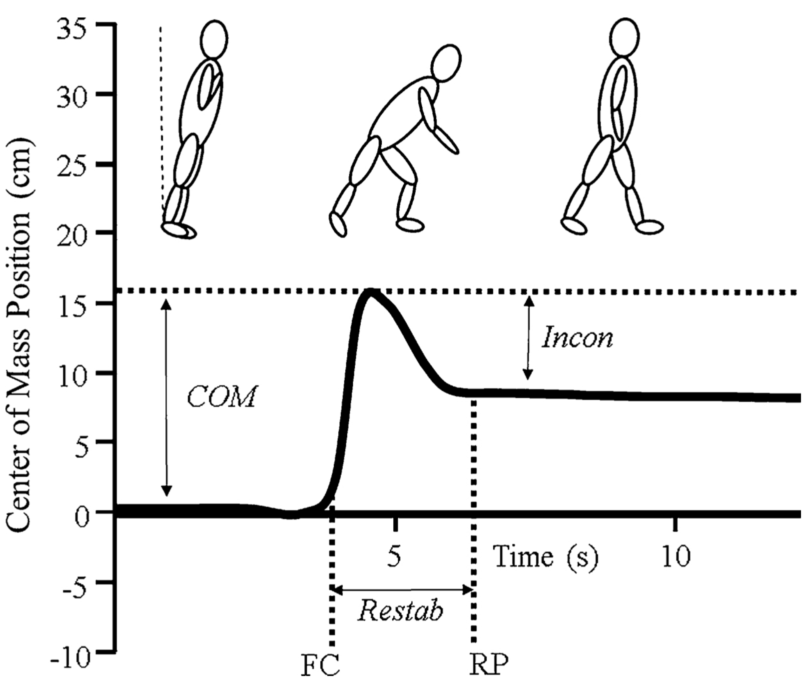

Fig. 2. Time-varying center of mass traces were analysed in both AP and ML directions. *COM* = peak center of mass displacement referenced to pre-release. *Restab* = duration from foot contact (FC) to a restabilsation point (RP) – defined by the center of mass velocity waveform. *Incon* = peak center of mass displacement referenced to post-restabilisation position.

($Shear_{AP}$) directions were extracted. Whole-body COM was calculated using the filtered kinematic data and the anthropometric tables of de Leva [28]. Maximum COM displacements in the ML (COM_{ML}) and AP (COM_{AP}) planes were extracted and referenced to the starting COM position (mean of 2 s pre-tether release) (Fig. 2). Tether release was defined when tether force dropped and remained below 2 SD of the initial 1 s mean. Restabilisation in each direction was defined as the point when the COM velocity waveform entered and remained within 2

SD of the mean COM velocity extracted during a forward-stance quiet standing trial (configuration obtained following a forward volitional step) [23,24]. Time to restabilisation was calculated from foot contact (> 10 N vertical GRF) to restabilisation of the ML (*Restab*$_{ML}$) and AP (*Restab*$_{AP}$) COM [24]. COM incongruity (the difference between the maximum COM position and mean position 2 s post-restabilisation) was extracted in the ML (*Incon*$_{ML}$) and AP (*Incon*$_{AP}$) directions (Fig. 2) [23,24]. Minimum margin of stability in the ML (*MOS*$_{ML}$) and AP (*MOS*$_{AP}$) directions after foot contact were calculated as the minimum ML distance from the COM to the lateral border of the step foot (5th metatarsal head), and minimum AP distance from the COM to the anterior border of the tip of the stepping foot big toe, respectively. Trial-to-trial variability of all measures was assessed by the standard deviation within each condition.

2.5. Statistical analysis

Separate one-way repeated measures analysis of variance (ANOVA) were performed to assess the influence of increased passive stiffness (corset) on magnitude and variability of: 1) *Shear*; 2) *COM*; 3) *MOS;* 4) *Restab*; and 5) *Incon* in both directions. All statistical analyses were performed with a software package (SPSS Version 20, SPSS Inc., Chicago, IL, USA) using an α of 0.05.

3. Results

3.1. Dynamic reactive stepping

Increased passive stiffness of the trunk and hips influenced reactive stepping AP kinematics, but no differences were observed in *Shear*$_{AP}$ magnitude ($p = 0.578$) nor variability ($p = 0.329$) (Fig. 3, Tables 1 and 2). *COM*$_{AP}$ was 14% greater in the corset (318.4 mm) compared to baseline (279.0 mm) condition ($p = 0.024$). The observed increases in *COM*$_{AP}$ contributed to a 45% reduced *MOS*$_{AP}$ in the corset (56.6 mm) compared to baseline (102.7 mm) condition ($p = 0.032$), with no significant difference in step length ($p = 0.233$). No differences in variability of *COM*$_{AP}$ ($p = 0.188$) nor *MOS*$_{AP}$ ($p = 0.288$) were observed across conditions.

Artificial stiffening of the trunk and hips influenced reactive stepping kinetics and kinematics in the ML direction (Fig. 3). *Shear*$_{ML}$ was 29% greater in the corset (78.4 N) compared to baseline (61.0 N) condition ($p = 0.016$), as well as 68% more variable ($p = 0.016$). COM_{ML} was 41% greater in the corset (77.6 mm) compared to baseline (54.9 mm) condition ($p = 0.002$), as well as 50% more variable ($p = 0.027$). The observed increases in *COM*$_{ML}$ contributed to a 22% reduced MOS_{ML} in the corset (150.5 mm) compared to baseline (195.6 mm) condition ($p = 0.010$), with no difference in step width ($p = 0.604$). No differences in MOS_{ML} variability were observed across conditions ($p = 0.544$).

Table 2
Statistical summary of ANOVA results for the magnitude and variability of each dependent variable.

Phase		Magnitude		Variability	
		F	(p)	F	(p)
Dynamic Reactive Stepping	**Hypothesis 1 (AP)**				
	Shear$_{AP}$	0.32	(0.578)	1.02	(0.329)
	COM$_{AP}$	**6.26**	**(0.024)**	1.90	(0.188)
	MOS$_{AP}$	**5.59**	**(0.032)**	1.21	(0.288)
	Hypothesis 2 (ML)				
	Shear$_{ML}$	**7.31**	**(0.016)**	**7.29**	**(0.016)**
	COM$_{ML}$	**13.91**	**(0.002)**	**6.04**	**(0.027)**
	MOS$_{ML}$	**8.78**	**(0.010)**	0.39	(0.544)
Restabilisation	**Hypothesis 3 (AP)**				
	Restab$_{AP}$	2.02	(0.176)	1.91	(0.188)
	Incon$_{AP}$	2.13	(0.165)	2.06	(0.172)
	Hypothesis 4 (ML)				
	Restab$_{ML}$	**28.18**	**(< 0.001)**	**11.04**	**(0.005)**
	Incon$_{ML}$	**5.99**	**(0.027)**	**6.26**	**(0.024)**

Bold font indicates statistical significance. *Shear* = stepping foot shear force; *COM* = peak center of mass displacement; *MOS* = minimum margin of stability; *Restab* = time to restabilisation; *Incon* = center of mass incongruity.

3.2. Restabilisation phase

The corset had no significant effect on restabilisation phase metrics in the AP direction (Fig. 4). Although *Incon*$_{AP}$ was on average 35% greater ($p = 0.165$) and 28% more variable ($p = 0.172$) in the corset condition compared to baseline, these differences were not statistically significant. Similarly, neither *Restab*$_{AP}$ magnitude ($p = 0.176$) nor variability ($p = 0.188$) were significantly different across conditions.

In contrast, the corset influenced ML restabilisation phase metrics following reactive stepping (Fig. 4, Tables 1 and 2). *Incon*$_{ML}$ and *Restab*$_{ML}$ were 41% and 71% greater in the corset (25.0 mm; 3.41 s) compared to baseline (17.7 mm; 2.00 s) condition, respectively (*Incon*$_{ML}$: $p = 0.027$; *Restab*$_{ML}$: $p < 0.001$). *Incon*$_{ML}$ and *Restab*$_{ML}$ were also 46% and 80% more variable in the corset (9.70 mm; 1.26 s) compared to baseline (6.65 mm; 0.70 s) condition, respectively (*Incon*$_{ML}$: $p = 0.024$; *Restab*$_{ML}$: $p = 0.005$).

4. Discussion

The primary goal of this study was to investigate the role of increased passive stiffness of the hip and trunk, in isolation from other

Table 1
Descriptive statistics mean (SD) across corset condition.

Phase		Magnitude			Variability		
		Baseline	Corset	Change (% Baseline)	Baseline	Corset	Change (% Baseline)
Dynamic Reactive Stepping	**Hypothesis 1 (AP)**						
	Shear$_{AP}$ (N)	194.6 (83.6)	189.2 (76.0)	−2.8	39.3 (22.0)	46.1 (29.0)	17.4
	COM$_{AP}$ (mm)	279.0 (39.9)	318.4 (53.8)	14.1	35.3 (12.2)	43.1 (26.7)	22.2
	MOS$_{AP}$ (mm)	102.7 (84.8)	56.6 (30.8)	−44.9	40.3 (22.9)	32.2 (18.7)	−20.1
	Hypothesis 2 (ML)						
	Shear$_{ML}$ (N)	60.9 (20.8)	78.4 (20.3)	28.7	15.0 (7.2)	25.2 (12.9)	68.0
	COM$_{ML}$ (mm)	54.9 (21.6)	77.6 (28.1)	41.4	16.2 (9.7)	24.3 (6.9)	49.8
	MOS$_{ML}$ (mm)	192.6 (65.3)	150.5 (78.6)	−21.9	32.9 (30.3)	41.4 (39.5)	26.0
Restabilisation	**Hypothesis 3 (AP)**						
	Restab$_{AP}$ (s)	1.59 (0.41)	1.91 (0.72)	20.1	0.53 (0.35)	0.73 (0.38)	37.7
	Incon$_{AP}$ (mm)	42.2 (31.0)	56.8 (32.1)	34.6	18.1 (11.9)	23.1 (10.5)	27.8
	Hypothesis 4 (ML)						
	Restab$_{ML}$ (s)	2.00 (0.42)	3.41 (1.07)	70.5	0.70 (0.36)	1.26 (0.58)	80.0
	Incon$_{ML}$ (mm)	17.7 (6.2)	25.0 (10.0)	41.1	6.7 (2.9)	9.7 (4.4)	45.9

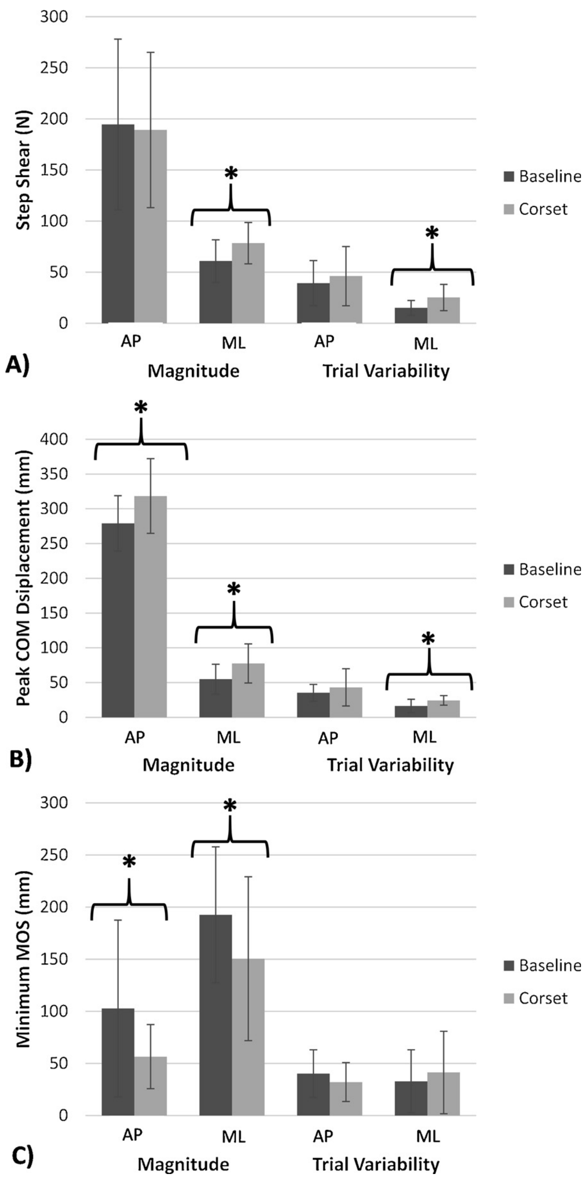

Fig. 3. Influence of increased trunk and hip stiffness on the magnitude and variability of: **a)** Step shear GRF (*Shear*); **b)** Peak COM displacement (*COM*); and **c)** Minimum margin of stability (*MOS*). Bars and error bars represent mean and standard deviation respectively. * Indicates statistical significance ($p < 0.05$).

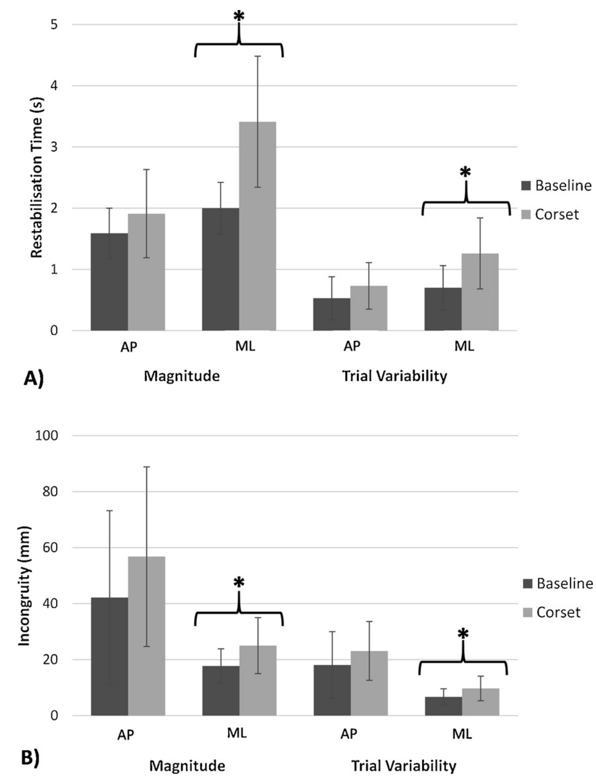

Fig. 4. Influence of increased trunk and hip stiffness on the magnitude and variability of: **a)** Restabilisation time (*Restab*); and **b)** Incongruity (*Incon*). Bars and error bars represent mean and standard deviation respectively. * Indicates statistical significance ($p < 0.05$).

ageing effects, on the restabilisation phase of a forward reactive step. In contrast to our first and second hypotheses, increased passive stiffness was associated with increased COM displacement and reduced MOS in both AP and ML directions. Contrary to our third hypothesis, the corset did not significantly affect AP restabilisation. However, in support of our fourth hypothesis, increased passive stiffness increased indicators of lateral instability during ML restabilisation. These data provide novel insights that improve our knowledge of the factors that influence stability during the restabilisation phase of a reactive step.

Our results provide insight into a possible factor increasing lateral instability and the need for subsequent steps following a reactive step in older adults [20]. In reactive stepping tasks, older adults exhibit greater $Shear_{ML}$ [22], $Incon_{ML}$, as well as longer $Restab_{ML}$ [24], compared to young adults. Artificial stiffening of the hips and trunk resulted in ML differences similar to that observed with ageing, suggesting young adults in this study were unable to sufficiently adjust their movement strategy following increases in link stiffness. While van der Burg et al. [16] suggested no net effect of increased stiffness on balance recovery following a trip, their work did not consider critical elements of balance recovery following foot contact. The current results suggest increased hip and trunk stiffness have detrimental effects on ML stability following foot contact (evidenced by increased COM_{ML} and $Shear_{ML}$, as well as reduced MOS_{ML}) and during restabilisation (evidenced by increased $Incon_{ML}$ and longer $Restab_{ML}$). Coupled with the increased variability we observed in these measures, the likelihood of a sufficiently large loss of ML stability - requiring additional steps - was increased by artificial stiffening of the hips and trunk. The current findings suggest interventions aimed to mobilize the trunk and hips, in conjunction with strengthening, could improve reactive balance and reduce fall risk among high stiffness populations.

Interestingly, increased passive stiffness did not significantly affect restabilisation phase metrics in the sagittal plane in which the perturbation occurred (Tables 1 and 2). Despite an average increase in COM_{AP} of **14%** (Fig. 3), no statistical differences were observed in $Incon_{AP}$ (Fig. 4). These results suggest that when wearing the corset, participants adopted a more anterior restabilisation point despite similar step length and thus available base of support across conditions. It should be noted that on average, $Incon_{AP}$ was **35%** greater in the corset condition compared to baseline, however the effect of the corset was not consistent across participants and thus variance was substantial (Fig. 4). Our results contrast those of van der Burg and colleagues [16], whom found no effect of artificial stiffness on the peak moment arm of gravity on the trunk (indicative of COM position), perhaps as their analysis was confined to the positioning phase of a simulated trip over an obstacle. Overall, these discrepancies suggest that the effects of passive trunk and hip stiffness on AP COM control may be dependent on the phase of trip recovery and/or simulation type. The robustness of the current findings should be evaluated across additional simulation types, such as translating platforms, which offer different mechanical and sensory stimuli both prior to and during perturbation [29].

There were several limitations associated with this study. First, net GRF could not be calculated as a single force platform was utilized in this study and the stance leg GRF was not measured. Future work should determine how increasing passive stiffness influences the control of the center of mass through changes in the location of the center of pressure as well as the net GRF vector [24]. Second, it is unclear if the corset appropriately simulated the increases in hip and trunk stiffness observed with aging [10], which could also be associated with active muscular co-contraction in addition to changes in passive stiffness [30]. While our use of young adults enabled investigation of the isolated effects of link stiffness, future analysis of active stiffening responses during reactive stepping, and characterization of trunk stiffness across the older adult population, would provide insights into the bio-fidelity of our approach. Third, we confined the stepping limb target area of our participants to a defined force platform. While the plate was reasonably large (46 x 51 cm), it may have inadvertently influenced potential effects of stiffness on step length/width.

In summary, this is the first study to investigate the role of increased hip and trunk stiffness, in isolation from other ageing effects, on the restabilisation phase of a reactive step. The results suggest that increased stiffness inhibits the ability of an individual to effectively control their ML COM following foot contact, increasing lateral instability. Thus, increased hip and trunk stiffness observed with aging [10] and in disease states [11], likely contributes to lateral instability and the need for subsequent steps among these populations. As lateral instability and the need for multiple steps are associated with increased fall risk [19], future work should investigate the efficacy of interventions that target increased mobilization (and potential strengthening) of the hips and trunk in these populations.

Declarations of interest

None.

Acknowledgments

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Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:https://doi.org/10.1016/j.gaitpost.2019.05.018.

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