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Reactive gait and postural adjustments following the first exposures to (un)expected stepdown

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ABSTRACT

This study evaluated the reactive biomechanical strategies associated with both upper- and lower-body (lead and trail limbs) following the first exposures to (un)expected stepdown at comfortable (1.22 ± 0.08 m/s) and fast (1.71 ± 0.11 m/s) walking velocities. Eleven healthy adults completed 34 trials per walking velocity over an 8-m, custom-built track with two forceplates embedded in its center. For the expected stepdown, the track was lowered by 0-, -10- and -20-cm from the site of the second forceplate, whereas the unexpected stepdown was created by camouflaging the second forceplate (-10-cm). Two-way repeated-measurement ANOVAs detected no velocity-related effects of stepdown on kinematic and kinetic parameters during lead limb stance-phase, and on the trail limb stepping kinematics. However, analyses of significant interactions revealed greater peak flexion angles across the trunk and the trail limb joints (hip, knee and ankle) in unexpected versus expected stepdown conditions at a faster walking velocity. The -10-cm unexpected stepdown (main effect) had a greater influence on locomotor behavior compared to expected conditions due mainly to the absence of predictive adjustments, reflected by a significant decrease in peak knee flexion, contact time and vertical impulse during stance-phase. Walking faster (main effect) was associated with an increase in hip peak flexion and net anteroposterior impulse, and a decrease in contact time and vertical impulse during stepdown. The trail limb, in response, swung forward faster, generating a larger and faster recovery step. However, such reactive stepping following unexpected stepdown was yet a sparse compensation for an unstable body configuration, assessed by significantly smaller step width and anteroposterior margin-of-stability at foot-contact in the first-recovery-step compared with expected conditions. These findings depict the impact of the expectedness of stepdown onset on modulation of global dynamic postural control for a successful accommodation of (un)expected surface elevation changes in young, healthy adults.

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1. Introduction

A brief reference to the literature reveals the existence of diversity across (a) gait disorders (Lim et al., 2007; Mahlknecht et al., 2013; Snijders et al., 2007); and (b) studies suggesting task-related (AminiAghdam and Blickhan, 2018; AminiAghdam et al., 2017; AminiAghdam and Rode, 2017; Dijkstra et al., 2015; Patel et al. 2018; Winter et al., 1990), method-specific (Haeufle et al., 2018; Mansfield and Maki, 2009; Verniba, 2017) and age-related (Arampatzis et al., 2008; Graham et al., 2014; Kluff et al. 2018;

McIlroy and Maki, 1996; Shulman et al., 2019) walking control under perturbation-negotiation paradigm. Moreover, the inherently unstable human bipedal locomotion (Winter, 1995) can be further challenged by exposure to a wide variety of possible external and internal balance-threatening perturbations. For instance, a regular daily ambulation entails the accommodation of various terrain attributes and hazards, such as trip, slip, compliance and surface elevation changes. All these evidence warrant further research to investigate both stability and adaptability of gait in response to different perturbation paradigms that vary in terms of objective intensity, direction, frequency and combination (Bohm et al., 2015; McCrum et al., 2017). A specific experimental analysis of surface elevation changes in form of curb descent, varying in elevation and visibility, as a function of walking velocity would be of great

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interest for improving understanding of underlying strategies of walking control.

Walking control requires both or either online predictive (feedforward-driven) and reactive (feedback-driven) adaptations in gait pattern to environmental changes (Patla, 2003; Pavol et al., 2004). In case of unexpected mechanical perturbations to the lead limb, the control of walking demands not only reactive kinematic and kinetic adjustments in the lead limb, but also swift reactive stepping by the trail limb (AminiAghdam et al., 2019; van Dieen et al., 2007). Dynamic balance recovery by stepping reflects the capability of rapid muscle force generation and coordination of the recovery behavior in response to unexpected perturbations (Mackey and Robinovitch, 2005). Thus, this capability seems to be a more realistic indicator of the ability to actively recover from almost-falling situations than other assessments like postural sway analysis or timed-up-and-go test. In response to a risk of anterior loss of balance, the ability to lengthen the single recovery step rapidly while restraining the upper-body motion is paramount to retain walking control (Arampatzis et al., 2008; Carty et al., 2011). An inefficiency in producing such compensatory strategy has been documented as a key identifier of future risk falls in elderly (Carty et al., 2014).

The stepdown on the lower levels with varying elevations may frequently occur in daily life at various walking velocities. Moreover, changes in ground levels are not always predictable due, e.g., to a camouflage or inattention. Under faster walking velocities, recovery from an (un)expected stepdown can be more demanding due mainly to a higher center-of-mass (CoM) horizontal velocity (Süptitz et al., 2012), leading possibly to more unstable body configurations (described by a smaller margin-of-stability (MoS)) (Hof et al., 2005; Süptitz et al., 2012). In our recent study using a detailed analysis of MoS concept (AminiAghdam et al., 2019), we showed a smaller MoS values when walking faster across or recovering from an unexpected 10-cm curb. This was due to a limited potential to increase the base-of-support to compensate for an increased CoM velocity when walking faster. In unexpected stepdown (UX_{ST}) versus expected stepdown (X_{ST}) when walking faster, the balance loss should be precluded by a rapid exploitation of the key components of balance recovery, i.e., counter-rotation movements or compensatory stepping strategies (Hof, 2007). Therefore, an investigation of gait behavior as a function of (un)expected surface elevation changes at different walking speeds can shed light onto the capacity of human locomotor adaptability, with implications for improving stepping responses in fall-prone populations. This study aims to expand on our previous work (AminiAghdam et al., 2019) by elucidating important biomechanical strategies associated with both upper- and lower-body (lead and trail limbs) during first exposures to X_{ST} (–10- and –20-cm) and UX_{ST} (–10-cm) at a fast walking velocity. Given unexpected mechanical perturbations to gait pattern may restrain the feed-forward locomotor adaptations, we hypothesized greater kinematic and kinetic changes in both upper-body and the lead limb in UX_{ST} versus X_{ST} during fast walking, caused by un- or less-expected onset of ground contact. Subsequently, the ability to retain dynamic postural stability would necessitate a prompter application of reactive responses by the trail limb in the first-recovery-step.

2. Methods

2.1. Participants

The data used in this study are from the same experiment (AminiAghdam et al., 2019; Vilemeyer et al., 2019). However, due to missing kinematic data for one participant during negotiating –20-cm stepdown, data analysis was only available for ele-

ven healthy volunteers (3 female, 8 male; mean ± standard deviation (SD); age = 25.5 ± 4.7 years, mass = 68.8 ± 8.0 kg, height = 179.1 ± 9.0 cm). The experimental protocol was approved by the local Ethics Committee of Friedrich-Schiller-University Jena (3532-08/12) and conducted according to the Declaration of Helsinki.

2.2. Experimental design and protocol

Kinematic data was collected using eight infra-red Qualisys motion capture cameras (240 Hz; Qualisys, Sweden). GRFs during walking were measured using two consecutive force platforms (960 Hz; Kistler, Switzerland) embedded in the middle portion of an 8 m-long, custom-built track (Fig. 1). An eleven-body segment model was defined using eighteen 19-mm retro-reflective markers (AminiAghdam et al., 2019). To analyze the X_{ST} during ongoing walking, the track was visibly lowered from the site of the second forceplate by 0-cm (X0_{ST}; unobstructed track), –10-cm (X10_{ST}) and –20-cm (X20_{ST}) (Fig. 1). The UX_{ST} was simulated by camouflaging a lowered second forceplate (–10-cm) using a wooden solid (0-cm camouflaged stepdown) or a hollow (–10-cm camouflaged stepdown; UX10_{ST}) forceplate-size box (covered by a non-transparent thin paper). The elevation of the first forceplate was set in flush with the surface of the first-half of the track across all (un)expected walking conditions. Prior to data collection, each participant underwent sufficient practice trials on unobstructed track, ensuring that the first forceplate was struck with left foot while walking at two velocities: comfortable (1.22 ± 0.08 m/s) and fast (1.71 ± 0.11 m/s). Gait velocity was calculated as the average horizontal velocity of the CoM during a stride taken by the left foot over the step. Participants, first, completed eight successful trials per X_{ST}. Visibility and velocity were block randomized. Afterwards, they had to accomplish ten successful trials (6 × 0-cm camouflaged step and 4 × UX10_{ST}) with the camouflage occurrence being randomized. Trials were considered successful when each single

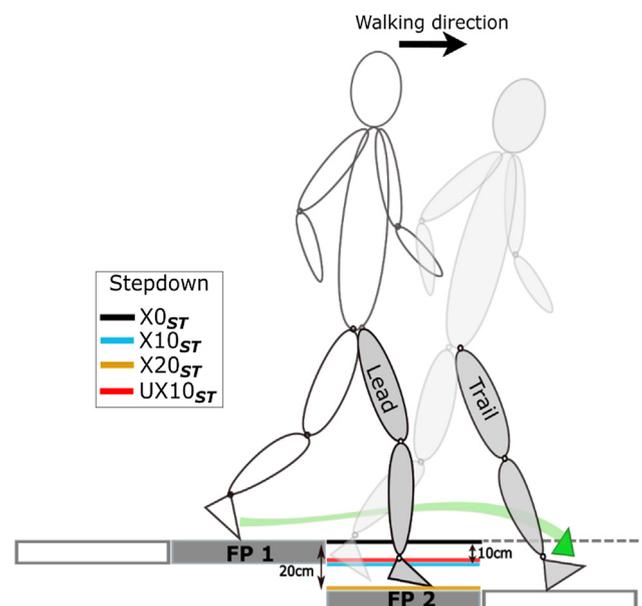


Fig. 1. Schematic illustration of stepdown during ongoing walking. Un(expected) stepdown was created by lowering the second, elevation-adjustable, forceplate (FP2) (in)visibly. Shown is the illustration of the expected stepdown of –20-cm by the lead limb and subsequently the first-recovery-step by the trail limb. X0_{ST}, X10_{ST} and X20_{ST}, expected stepdown of 0-, –10- and –20-cm, respectively; UX10_{ST}, –10-cm unexpected stepdown.

force plate was fully struck by one foot without losing any reflective markers.

2.3. Parameters of interest

For the stepdown at lower levels (lead limb), the ensemble average of the first trial of the trunk, hip, knee and ankle peak flexion angles in sagittal plane; contact time; vertical impulse (VIMP); net anteroposterior impulse (HIMP) were determined. Trunk angle was defined by the angle sustained by the line connecting the midpoint between the L5–S1 junction and the seventh cervical spinous process with respect to the vertical. VIMP and HIMP were computed by integrating the vertical GRF and net anteroposterior GRF, respectively, over the stance time, and were normalized to the product of body-weight and $\sqrt{\text{leglength}/\text{gravitational acceleration}}$ (Hof et al., 2005). For the recovery step (trail limb), the ensemble average of the first trial of the hip, knee and ankle peak flexion angles in sagittal plane; spatiotemporal gait parameters (normalized by the dimensionless normalization method (Hof, 1996)), including step length, step time, step width, swing time and swing velocity were determined during the time course between touchdown and toe-off of the lead foot at lower levels. Furthermore, we determined the anteroposterior MoS at foot-contact in the first-recovery-step as the difference between the anterior boundary of the base-of-support and the extrapolated CoM (Hof et al., 2005). A detailed description of MoS calculation is outlined in Appendix. The whole-body CoM was calculated using body segment inertial parameters (Plagenhoef et al., 1983).

2.4. Data processing and statistics

Kinetic and kinematic data were analyzed using custom written Matlab code (R2017a; Mathworks). Qualisys system was triggered using the Kistler hardware (external trigger) and software (BioWare). Marker trajectories data were filtered using a bidirectional, fourth-order, low-pass Butterworth-Filter with a cutoff frequency of 12 Hz. The instants of foot touchdown and toe-off were determined using a vertical GRF threshold of 0.02 body-weight. For the recovery step, there was no GRF available and, therefore, the instant of foot touchdown was determined by the velocity profile of the foot markers (O'Connor et al., 2007).

Data from the first exposures to each walking task were separately analyzed for lead and trail limbs. In case of MoS calculation, a missing value for the first trial, due to loss of marker data, was replaced by the ensemble average of the existing trials per condition per participant. For normally distributed data sets, repeated-measurement ANOVAs were implemented with SPSS 23.0 (SPSS, Inc., Chicago, IL, USA) using two within-subjects factors: (1) velocity (comfortable and fast), and (2) stepdown (X0_{ST}, X10_{ST}, X20_{ST}, and UX10_{ST}). A significant interaction was evaluated for each dependent variable of interest. In case of a nonsignificant interaction, bilateral differences were evaluated for data pooled from all four stepdown conditions, and between-stepdown differences were evaluated for data pooled from two walking velocities. For a significant interaction, post-hoc comparisons were performed to evaluate between-stepdown differences across each walking velocity, and bilateral differences for each stepdown condition. For post-hoc comparisons, a Bonferroni adjustment was made to reduce experiment-wise type I error. The significance level was $\alpha = 0.05$. Effect sizes (EFs) were reported using partial eta-squared. According to (Portney and Watkins, 2009), values of 0.1, 0.25 and 0.4 represent small, medium and large EFs, respectively for ANOVA.

3. Results

3.1. Kinematic and kinetic adjustments pertaining to the lead limb

ANOVA revealed no Velocity \times Stepdown interaction on peak flexion angles across lower-limb joints, VIMP, HIMP and Contact_T (Table 1), indicating that the kinematics and kinetics changes induced by (un)expected stepdown were not walking velocity related. Post-hoc analysis of significant main effects of Velocity revealed an increase in Hip_{PF} and HIMP, and a decrease in VIMP and Contact_T in fast walking (Fig. 5A–F, Table 1). For between-stepdown comparisons, there were a significant decrease in Hip_{PF}, Knee_{PF} and Contact_T in X10_{ST}, X20_{ST} and UX10_{ST} compared with X0_{ST} (Figs. 3 and 5). Further comparisons revealed a significantly decreased HIMP and Contact_T in X20_{ST} versus X10_{ST} (Figs. 3 and 5). Additionally, there were a significantly reduced Knee_{PF} and VIMP in UX10_{ST} compared with all X_{ST} conditions (Figs. 2 and 5). A significant interaction on Trunk_{PF} revealed a drastic increase in the trunk pitch angle while negotiating UX10_{ST} compared with all X_{ST} conditions, and a significant bilateral increase in fast walking (Fig. 6A).

3.2. Reactive recovery strategies adopted by the swing (trail) limb

ANOVA revealed significant Velocity \times Stepdown interactions on peak flexion angles across lower-limb joints (hip, knee and ankle; Table 1), indicating that reactive adaptability in joint kinematics of the swing limb are walking velocity related. Post-hoc comparisons revealed a significant increase of Hip_{PF} in UX10_{ST} versus all X_{ST} conditions at both comfortable and fast walking velocities (Fig. 4A and 6B). Furthermore, there was a significant increase in Hip_{PF} at a fast walking velocity when negotiating UX10_{ST} (Fig. 6B). Knee_{PF} significantly increased in X10_{ST} and X20_{ST} at both comfortable and fast walking velocities compared with X0_{ST} (Fig. 4B and 6C). Under UX10_{ST} condition, Knee_{PF} demonstrated a greater flexion angle compared with all X_{ST} conditions at both walking velocities (Fig. 4B and 6C). At both walking velocities, Ankle_{PF} exhibited a significant increase in UX10_{ST} versus all X_{ST} conditions, whereas no significant changes were observed between visible stepdown conditions (Fig. 4C and 6D). Analyses considering main effects revealed a significant increase in Step_L and Swing_V, and a significant decrease in Step_T, Swing_T and MoS at a fast versus a comfortable walking velocity (Fig. 5G–L). For between-stepdown comparisons, there were a significant increase in Swing_V (Fig. 5L), and a significant decrease in Step_T in X10_{ST} versus X0_{ST} (Fig. 5I). In recovery from X20_{ST}, there were a significant increase in Swing_V (Fig. 5L) compared with X0_{ST}, and a significant decrease in Step_T (Fig. 5I) and Swing_T (Fig. 5K) compared with X0_{ST}. In UX10_{ST} versus visible conditions, there were a significant increase in Step_L and Swing_V, and a significant decrease in MoS, Step_T and Step_W compared with X0_{ST}; a significant increase in Step_L, and a significant decrease in MoS, Step_T and Step_W compared with X10_{ST}; a significant increase in Step_L, and a significant decrease in MoS and Step_W compared with X20_{ST} (Fig. 5G–L).

4. Discussion

In this study, we investigated the reactive biomechanical strategies associated with both upper- and lower-body (lead and trail limbs) following the first exposures to X_{ST} and UX_{ST} as a function of walking velocity. UX_{ST} versus X_{ST} conditions was characterized by accentuated changes across the majority of analyzed kinematic and kinetic variables, due possibly to the absence of preparatory or predictive locomotor adjustments. This yielded a more unstable

Table 1
Kinematic, kinetic and spatiotemporal gait parameters (mean \pm S.D.) during (un)expected stepdown.

| | Velocity | Stepdown (mean \pm SD) | | | | Main effect | | | | Interaction | |
|-------------------------|----------|--------------------------|------------------------------|--------------------------------|---|-------------|------|----------|------|-------------|------|
| | | X0 _{ST} | X10 _{ST} | X20 _{ST} | UX10 _{ST} | Velocity | | Stepdown | | P | F |
| | | P | F | DF | ES | P | F | DF | ES | P | F |
| Trunk _{PF} (°) | C | 7.04 \pm 3.42 | 6.99 \pm 3.34 | 6.97 \pm 4.05 | 10.8 \pm 5.28 | 0.001 | 52.3 | 0.001 | 25.8 | 0.001 | 15.1 |
| | F | 7.22 \pm 3.26 | 7.66 \pm 4.56 | 8.16 \pm 5.55 | 18.2 \pm 4.74^{a,b,c} | 1 | 0.84 | 3 | 0.72 | 1.29 | 0.61 |
| <i>Lead limb</i> | | | | | | | | | | | |
| Hip _{PF} (°) | C | 15.9 \pm 4.09 | 9.14 \pm 5.73 | 8.35 \pm 7.07 | 4.78 \pm 9.56 | 0.001 | 35.9 | 0.002 | 10.9 | 0.42 | 0.85 |
| | F | 18.6 \pm 6.17 | 13.8 \pm 6.39 | 10.4 \pm 9.22 | 9.71 \pm 7.71 | 1 | 0.78 | 1.63 | 0.52 | 1.71 | 0.07 |
| Knee _{PF} (°) | C | 44.2 \pm 4.82 | 42.1 \pm 6.25 | 40.6 \pm 5.62 | 35.1 \pm 4.87 | 0.41 | 0.72 | 0.001 | 19.9 | 0.84 | 0.27 |
| | F | 43.8 \pm 4.75 | 40.6 \pm 5.01 | 40.2 \pm 3.71 | 35.4 \pm 10.2 | 1 | 0.06 | 1.75 | 0.66 | 3 | 0.27 |
| Ankle _{PF} (°) | C | 10.1 \pm 4.93 | 7.02 \pm 7.31 | 8.25 \pm 7.11 | 7.26 \pm 7.35 | 0.96 | 0.02 | 0.41 | 0.79 | 0.41 | 0.78 |
| | F | 9.47 \pm 4.62 | 9.33 \pm 5.01 | 12.8 \pm 5.01 | 10.6 \pm 8.31 | 1 | 0.00 | 1.08 | 0.07 | 1.04 | 0.07 |
| VIMP | C | 2.12 \pm 0.13 | 2.07 \pm 0.12 | 2.11 \pm 0.14 | 1.86 \pm 0.20 | 0.001 | 325 | 0.001 | 22.4 | 0.16 | 1.85 |
| | F | 1.75 \pm 0.08 | 1.83 \pm 0.09 | 1.82 \pm 0.10 | 1.47 \pm 0.16 | 1 | 0.97 | 3 | 0.69 | 3 | 0.15 |
| HIMP | C | -0.01 \pm 0.02 | -0.01 \pm 0.02 | 0.01 \pm 0.03 | -0.01 \pm 0.02 | 0.001 | 21.7 | 0.08 | 2.43 | 0.18 | 1.73 |
| | F | -0.03 \pm 0.02 | -0.05 \pm 0.02 | -0.02 \pm 0.03 | -0.02 \pm 0.04 | 1 | 0.68 | 3 | 0.19 | 3 | 0.14 |
| Contact _T | C | 2.56 \pm 0.10 | 2.35 \pm 0.12 | 2.24 \pm 0.11 | 2.01 \pm 0.18 | 0.001 | 377 | 0.001 | 111 | 0.34 | 1.03 |
| | F | 2.04 \pm 0.08 | 1.94 \pm 0.09 | 1.82 \pm 0.07 | 1.53 \pm 0.21 | 1 | 0.97 | 3 | 0.91 | 1.26 | 0.09 |
| <i>Trail limb</i> | | | | | | | | | | | |
| Hip _{PF} (°) | C | 22.1 \pm 3.32 | 22.5 \pm 4.79 | 24.5 \pm 4.95 | 44.3 \pm 6.28 ^{a,b,c} | 0.001 | 68.1 | 0.001 | 131 | 0.001 | 21.1 |
| | F | 23.2 \pm 3.11 | 25.8 \pm 5.01 | 28.2 \pm 5.37 | 59.9 \pm 6.63^{a,b,c} | 1 | 0.87 | 1.19 | 0.92 | 1.43 | 0.67 |
| Knee _{PF} (°) | C | 68.5 \pm 4.95 | 78.6 \pm 3.44 ^a | 92.6 \pm 4.37 ^{a,b} | 100 \pm 8.97 ^{a,b,c} | 0.001 | 110 | 0.001/ | 210 | 0.001 | 16.9 |
| | F | 68.3 \pm 4.24 | 80.3 \pm 5.31 ^a | 93.1 \pm 4.38 ^{a,b} | 110 \pm 8.34^{a,b,c} | 1 | 0.91 | 1.36 | 0.95 | 3 | 0.62 |
| Ankle _{PF} (°) | C | 0.91 \pm 6.61 | 4.01 \pm 6.33 | 5.42 \pm 7.97 | 14.6 \pm 0.13 ^{a,b,c} | 0.01 | 11.5 | 0.001 | 27.1 | 0.01 | 4.40 |
| | F | 3.17 \pm 5.83 | 4.75 \pm 5.82 | 7.89 \pm 7.23 | 14.6 \pm 0.13 ^{a,b} | 1 | 0.53 | 3 | 0.73 | 3 | 0.31 |
| Step _L | C | 0.83 \pm 0.03 | 0.85 \pm 0.03 | 0.85 \pm 0.07 | 0.92 \pm 0.04 | 0.001 | 60 | 0.001 | 14.2 | 0.96 | 0.09 |
| | F | 0.92 \pm 0.05 | 0.95 \pm 0.07 | 0.94 \pm 0.07 | 1.01 \pm 0.07 | 1 | 0.85 | 3 | 0.58 | 3 | 0.01 |
| Step _T | C | 2.05 \pm 0.13 | 1.87 \pm 0.10 | 1.78 \pm 0.13 | 1.63 \pm 0.16 | 0.001 | 350 | 0.001 | 36.3 | 0.13 | 2.01 |
| | F | 1.69 \pm 0.07 | 1.62 \pm 0.08 | 1.50 \pm 0.07 | 1.39 \pm 0.10 | 1 | 0.97 | 3 | 0.78 | 3 | 0.16 |
| Step _W | C | 0.17 \pm 0.03 | 0.17 \pm 0.04 | 0.18 \pm 0.05 | 0.14 \pm 0.04 | 0.29 | 1.21 | 0.001 | 10.9 | 0.17 | 1.77 |
| | F | 0.18 \pm 0.05 | 0.16 \pm 0.02 | 0.17 \pm 0.05 | 0.11 \pm 0.03 | 1 | 0.03 | 3 | 0.52 | 3 | 0.15 |
| Swing _T | C | 1.59 \pm 0.11 | 1.48 \pm 0.11 | 1.45 \pm 0.11 | 1.48 \pm 0.12 | 0.001 | 103 | 0.01 | 4.70 | 0.07 | 2.52 |
| | F | 1.36 \pm 0.06 | 1.33 \pm 0.05 | 1.27 \pm 0.07 | 1.35 \pm 0.10 | 1 | 0.11 | 3 | 0.32 | 3 | 0.21 |
| Swing _V | C | 0.83 \pm 0.07 | 0.90 \pm 0.08 | 0.90 \pm 0.09 | 0.96 \pm 0.10 | 0.001 | 361 | 0.001 | 7.59 | 0.14 | 1.92 |
| | F | 1.09 \pm 0.08 | 1.15 \pm 0.09 | 1.19 \pm 0.07 | 1.16 \pm 0.07 | 1 | 0.97 | 3 | 0.43 | 3 | 0.16 |
| MoS (cm)* | C | 12.8 \pm 3.15 | 10.5 \pm 2.82 | 11.8 \pm 4.65 | 6.99 \pm 5.54 | 0.001 | 454 | 0.001 | 13.3 | 0.25 | 1.48 |
| | F | 1.43 \pm 4.16 | -0.04 \pm 4.82 | -2.47 \pm 4.64 | -7.73 \pm 6.04 | 1 | 0.97 | 3 | 0.57 | 1.51 | 0.12 |

The last two columns outline the p-values (P), F-value (F), degrees of freedom (DF) and effect size (ES, partial eta-squared) pertaining to the main effects of Velocity and Stepdown as well as their interaction, respectively. In case of interaction effect, significant differences from X0_{ST}, X10_{ST} and X20_{ST} across each walking velocity are indicated by 'a', 'b' and 'c', respectively ($p < 0.05$). Accordingly, bold values indicate significant bilateral differences for each stepdown condition ($p < 0.05$). X0_{ST}, X10_{ST} and X20_{ST}, expected stepdown of 0-, -10- and -20-cm, respectively; UX10_{ST}, -10-cm unexpected stepdown; MoS, margin-of-stability; VIMP, vertical impulse; HIMP, net anteroposterior impulse; 'C', comfortable; 'F', fast; Subscripts: 'PF', peak flexion; 'T', time; 'L', length; 'W', width; 'V', velocity. * due to loss of marker data, a missing value for the first trial (11%; 8 out of 88: 8 conditions \times 11 participants) was replaced by the ensemble average of the existing trials per condition per participant.

body configuration (i.e., inadequate trunk stabilization) during stance-phase at the lower level, leading to a reduced walking control in the first-recovery-step, particularly at a faster walking velocity.

The ability to both arrest and reverse trunk motion, and to simultaneously execute a timely, proper recovery step in exposure to forward balance perturbations are two key strategies for successful reactive balance recovery (AminiAghdam and Blickhan, 2018; Arampatzis et al., 2008). The performance of such strategies is largely determined by the ability of positioning the CoM more posterior relative to the leading limb at foot-contact (BRODIE et al., 2018). In our study, participants seemed to manage to do so in both X10_{ST} and X20_{ST} by controlling the trunk motion, as reflected in a less-decreased MoS at foot-contact in the first-recovery-step compared with X0_{ST}. Following UX10_{ST}, the trunk pitch angle progressively increased, with a large EF, toward the end of the stance-phase, particularly when walking faster compared with both X10_{ST} and X20_{ST} (Fig. 2A). In the study conducted by (van Dieen et al., 2007), they reported the mean increase of $\sim 7^\circ$ in peak trunk pitch angle following UX10_{ST}. In comparison, the magnitude of peak trunk pitch angle in our study was 10.8 ± 5.28 and 18.2 ± 4.74 during comfortable and fast walking velocities, respectively. The greater values in our study can be attributed to

a) the experimental design, by which natural movements of participants were not restrained by e.g. wearing safety harness, thus allowing for a greater upper-body degree of freedom; and b) the approach of analyzing the first exposures to stepdown conditions.

The contact time was substantially decreased as a result of both increasing walking velocity and stepping down, confirmed by large ESs. However, the magnitude of reduction following UX10_{ST} was significantly greater than following both X10_{ST} and X20_{ST} (Fig. 5F, Table 1). A shorter contact time during UX10_{ST} yielded a smaller VIMP. Given impulse due to vertical GRF directly promotes the CoM support (Seeley et al., 2008), a reduction in VIMP would imply an ineffective contribution from lead limb to the control of upper-body. Net HIMP appeared to be negative across all walking conditions, suggesting the braking impulse applied to decelerate the CoM was greater than propulsive impulse. Unlike VIMP (ES of 0.69), HIMP (ES of 0.19) was not influenced by UX10_{ST} compared with X_{ST} conditions. These adjustments, alongside a greater trunk flexion during the stance-phase of UX10_{ST} in fast walking indicate that impulses due to vertical GRF are insufficient to stabilize the upper-body, reflected in a reduced walking control in the first-recovery-step (negative MoS). For older adults, a greater negative net HIMP in the first-recovery-step following sudden induced forward falls has been found responsible for controlling the CoM

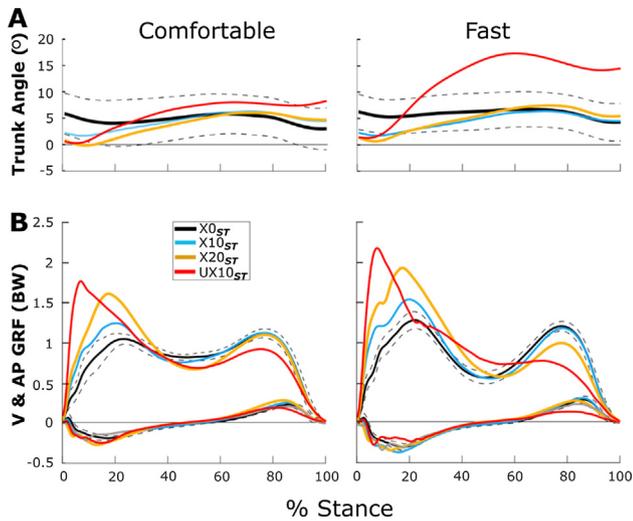


Fig. 2. Trunk and ground reaction forces (GRF) during (un)expected stepdown. Shown are ensemble-averaged (A) trunk angle trajectory and (B) vertical and anteroposterior GRFs (normalized to participant body weight (BW)) following the first exposures to (un)expected stepdown at comfortable (left) and fast (right) walking velocities during the stance-phase (normalized to 100%) of the lead limb ($N = 11$). Dashed lines represent S.D. for the level, unobstructed contact (X_{0ST}), X_{0ST} , X_{10ST} and X_{20ST} , expected stepdown of 0-, -10- and -20-cm, respectively; UX_{10ST} , -10-cm unexpected stepdown.

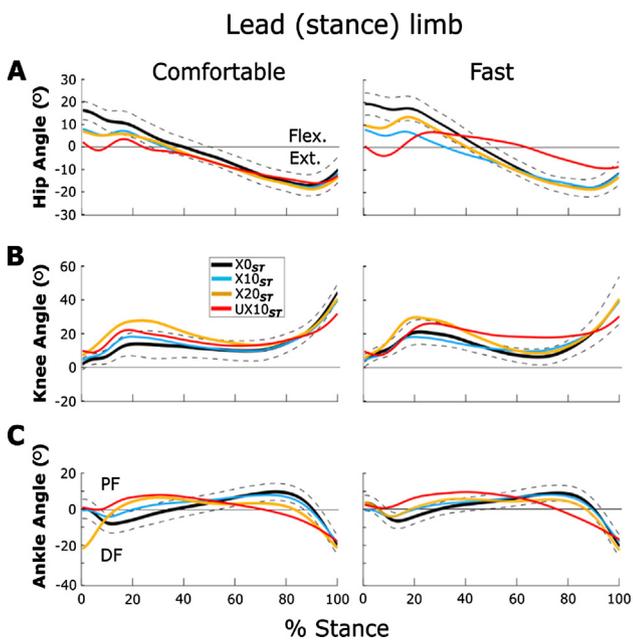


Fig. 3. The lead limb's joint angle trajectories during (un)expected stepdown. Shown are ensemble-averaged (A) hip, (B) knee and (C) ankle joint trajectories following the first exposures to (un)expected stepdown on lower levels at comfortable (left) and fast (right) walking velocities during the stance-phase (normalized to 100%) of the lead limb ($N = 11$). Dashed lines represent S.D. for the level, unobstructed contact (X_{0ST}). Flex., flexion; Ext., extension; PF, plantarflexion; DF, dorsiflexion; X_{0ST} , X_{10ST} and X_{20ST} , expected stepdown of 0-, -10- and -20-cm, respectively; UX_{10ST} , -10-cm unexpected stepdown.

anteroposterior velocity, and thus leading to positive MoS values (Arampatzis et al., 2008). On the other hand, walking faster was characterized by a significantly smaller VIMP due mainly to a shorter contact time, as indicated by a large ES of 0.97 for both variables. Moreover, a larger HIMP (ES of 0.68) in a faster walking reflects modulation of the braking impulse more than the propulsive impulse. This latter finding is in line with a previous study

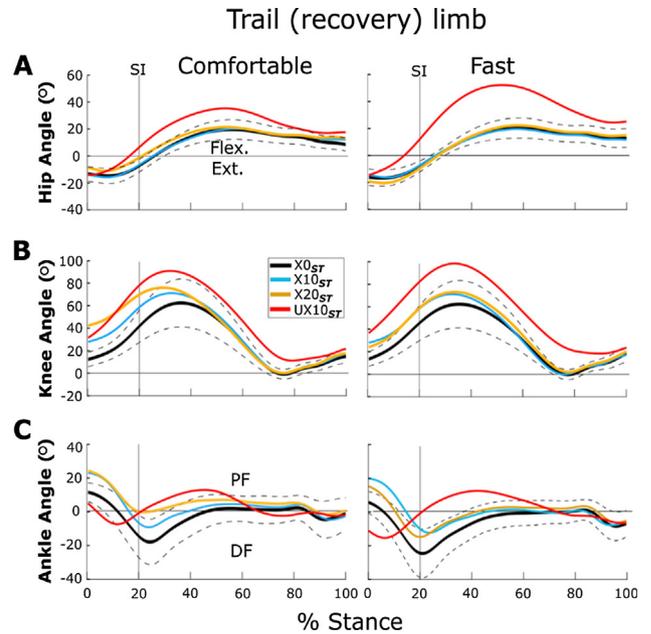


Fig. 4. The swing limb's joint angle trajectories during (un)expected stepdown. Shown are ensemble-averaged (A) hip, (B) knee and (C) ankle joint trajectories of the trail limb ($N = 11$) following the first exposures to (un)expected stepdown at comfortable (left) and fast (right) walking velocities (normalized to 100%). Dashed lines represent S.D. for the level, unobstructed contact (X_{0ST}). Vertical solid lines mark the initiation of the swing-phase (SI). Flex., flexion; Ext., extension; PF, plantarflexion; DF, dorsiflexion; X_{0ST} , X_{10ST} and X_{20ST} , expected stepdown of 0-, -10- and -20-cm, respectively; UX_{10ST} , -10-cm unexpected stepdown.

(Peterson et al., 2011) suggesting that the braking versus propulsive impulse has a greater relationship with speed. Furthermore, the vertical GRF pattern was seemingly altered in UX_{10ST} at both walking velocities, illustrating a more accentuated peak earlier during the stance-phase compared with X_{ST} conditions (Fig. 2B).

In UX_{ST} versus X_{ST} , the trail limb propelled forward with an accentuated sagittal lower-limb joint flexion angles at both walking velocities. An analysis of ESs revealed a greater velocity-related impact of stepdown on joint peak flexion angles of hip and knee than on that of ankle (Table 1). Such kinematic behavior contributed to a longer step length, but a narrower step width in the first-recovery-step which is indicative of a smaller mediolateral margin for the extrapolated CoM and thus a compromised stability in the frontal plane. This is attributable to insufficient sensory information induced by the camouflage of stepdown, which is key for the control of lateral foot placement (Bauby and Kuo, 2000). Although the swing velocity and step time were shorter in X_{ST} and UX_{ST} compared with unobstructed walking, UX_{10ST} did not represent significant changes compared with X_{10ST} and X_{20ST} . While UX_{10ST} caused more pronounced alterations than X_{20ST} in postural control and dynamics of the lead limb, but a larger step length by the trail limb following UX_{10ST} implies the capability of generating a rapid reactive recovery step in young, healthy adults. Such reactive motor response for controlling the forward rotation of the upper-body in young adults has already been shown (van Dieen et al., 2007). Despite achieving the largest step length by the trail limb following UX_{10ST} , indicating attempts to provide greater margins for the compromised stability, the MoS appeared negative. MoS, as a metric estimate of stability, represents the effectiveness of both step and trunk kinematics in compensatory post-perturbation recovery steps to restore balance (Crenshaw et al., 2012). The magnitude of decrease in MoS following X_{20ST} did not significantly differ from that of both X_{0ST} and X_{10ST} in spite of a smaller HIMP compared with all other

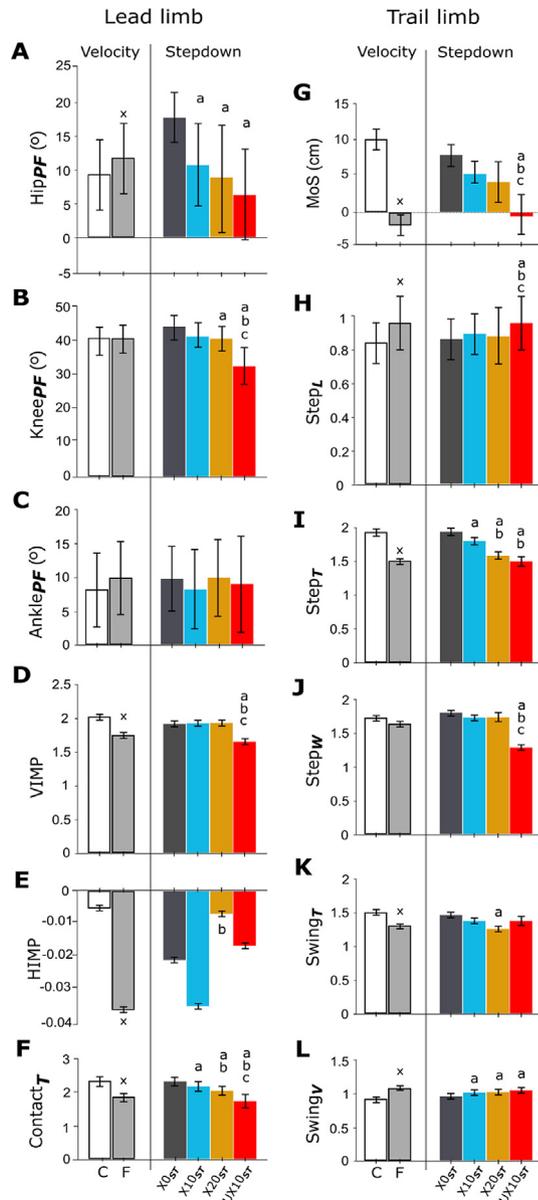


Fig. 5. Main effects of Velocity and Stepdown. Shown (mean ± S.D.) are the main effects of Velocity and Stepdown on variables (A–F, lead limb; G–L, trail limb) for whom two-way repeated-measurement ANOVAs revealed no Velocity × Stepdown interaction (N = 11). Significant bilateral differences are indicated by ‘x’. Accordingly, significant differences from X0_{ST}, X10_{ST} and X20_{ST} are indicated by ‘a’, ‘b’ and ‘c’, respectively (p < 0.05; Bonferroni post-hoc test). Error bars denote standard deviation. X0_{ST}, X10_{ST} and X20_{ST}, expected stepdown of 0-, –10- and –20-cm, respectively; UX10_{ST}, –10-cm unexpected stepdown; MoS, margin-of-stability; VIMP, vertical impulse; HIMP, net anteroposterior impulse; ‘C’, comfortable; ‘F’, fast. Subscripts: ‘PF’, peak flexion; ‘T’, time; ‘L’, length; ‘W’, width; ‘V’, velocity.

conditions. One may argue that the unexpectedness rather than the magnitude of the stepdown may impose a greater dynamic postural instability during ongoing walking, which may require multiple recovery steps to restore normal gait. Moreover, the MoS was substantially decreased when walking faster (ES of 0.97). The decrease of MoS proportionally with an increase of the CoM velocity has already been reported during unperturbed treadmill walking (McCrum et al., 2019; Süptitz et al., 2012).

Our findings concurred with previous studies (Müller et al., 2014; Shinya et al., 2009; van der Linden et al., 2007; van Dieen et al., 2007) which reported that despite the potential disturbing impacts of UX_{ST} on dynamic locomotor stability, young, healthy

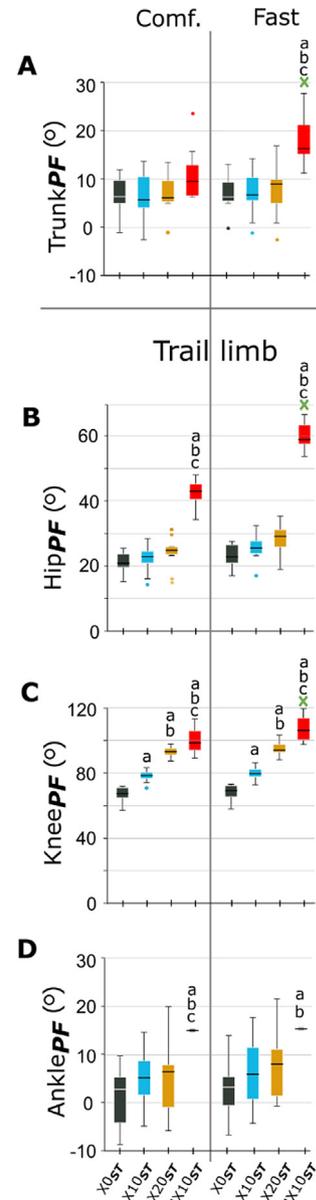


Fig. 6. Boxplots of Velocity × Stepdown interaction. Shown are peak flexion angles for the (A) trunk, and for the trail limb joint kinematics, including (B) hip, (C) knee and (D) ankle (N = 11). Significant differences from X0_{ST}, X10_{ST} and X20_{ST} at both comfortable and fast walking velocities are indicated by ‘a’, ‘b’ and ‘c’, respectively. Accordingly, a significant bilateral difference for each stepdown condition is indicated by ‘x’ (p < 0.05; Bonferroni post-hoc test). A vertical gray line separates stepdown conditions in a fast walking from that of comfortable (Comf.) walking velocity. X0_{ST}, X10_{ST} and X20_{ST}, expected stepdown of 0-, –10- and –20-cm, respectively; UX10_{ST}, –10-cm unexpected stepdown. Subscript: ‘PF’, peak flexion.

participants are able to accommodate (un)expected level changes during ongoing walking. However, the experimental setups used by other groups prompted wearing customized goggles for blocking the lower visual field and the safety harness for fall prevention (Buckley et al., 2011; Miyasike-Dasilva et al., 2019; van der Linden et al., 2007; van Dieen et al., 2007) that both may restrict natural locomotor behavior. Different from these studies, we implemented a more realistic UX_{ST} simply by camouflaging the site of the ground level change with no further experimental constraints. Furthermore, we minimized the learning effects that may stem from repetition of multiple trials by analyzing the first exposures to perturbations, attempting to improve the ecological validity of the experimental paradigms. One limitation concerns our experi-

mental setup which did not allow for further analysis of the subsequent recovery steps to determine either one or additional recovery steps are required to return to the steady-state gait following accommodating UX10₅₇. Second, owing to the inherent multiplicity in the multifactorial repeated-measurement ANOVA, our findings may be subject to higher type-one error rates and thus, possibly, less persuasive than presented.

Collectively, our study illustrates that the availability of spatial awareness or visual perception of ground level change facilitates accommodation of larger elevation differences during ongoing gait, presumably through global predictive locomotor adaptations. Although young, healthy participants demonstrated the capability of coping with an unexpected onset of stepdown at a faster walking velocity, but with a deficient performance, such environmental hazard is assumed to impose greater stability threats to fall-prone cohorts. This is due to the fact that stepping on an uneven ground has been identified as one of the most potential fall risk factors in neuropathy patients (DeMott et al., 2007) and in older adults (Li et al., 2006). As a consequence, an extension of the present experiment to these and other patient cohorts can help identify the health- and age-related challenges associated with (in)voluntary stepping, and estimate their increased falling risk under a complicated locomotor tasks (i.e., accommodation of (un)expected ground surfaces changes) at different walking speeds.

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Declaration of Competing Interest

The authors have no conflicts of interest to declare.

Appendix A. Supplementary material

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.jbiomech.2019.07.029>.

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