Influence of Handrail Height and Fall Direction on Center of Mass Control and the Physical Demands of Reach-to-Grasp Balance Recovery Reactions

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Influence of handrail height and falling direction on center-of-mass control and the physical demands of reach-to-grasp balance recovery reactions

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Abstract

The ability to maintain and recover center of mass (COM) and trunk control after a destabilization is critical for avoiding falls and fall-related injuries. Handrails can significantly enhance a person's ability to recover from large destabilizations, by enabling the person to grasp and apply high forces to the rail to stabilize their COM. However, the influence of handrail height and falling direction on COM control and the demands of grasping are unknown. We investigated the effect of handrail height (34, 38, 42 inches) and fall direction (forward, backward) on COM and trunk control, and the corresponding physical demands of reach-to-grasp balance reactions. Thirteen young adults were destabilized with platform perturbations, and reached to grasp a nearby handrail to recover balance without stepping. COM kinematics and applied handrail forces were collected. COM control was evaluated in terms of: (1) COM range and peak displacement, velocity and momentum in all Cartesian axes; and (2) trunk angular displacement, velocity and momentum in the roll and pitch axes. The physical demands of grasping were estimated via resultant handrail impulse. Compared to forward-directed falling, backward-directed falling was generally associated with greater peak COM and trunk angular displacement, velocity and momentum, along with greater handrail impulse. Higher handrails generally resulted in reduced peak COM and trunk angular displacement, velocity and momentum, as well as reduced handrail impulse. These results suggest that higher handrails (within the range of heights tested) may provide a stability advantage within the range of handrail heights tested, with better COM control achieved with lower physical demands of grasping.

1 Introduction

Many falls result from activities that challenge control of a person's center of mass (COM) with respect to their base of support (BOS), such as walking, incorrect weight shifting, tripping, stumbling, or bending [1, 2]. The position and velocity of the COM with respect to the BOS can influence fall risk during slipping [3]. Accordingly, the ability to maintain and recover COM and trunk control from destabilizations is critical for avoiding falls and fall-related injuries.

Individuals employ many strategies to control their COM and trunk following balance disturbances. However, the effectiveness of these strategies is heavily context-dependent. For small perturbations, "fixed-support" strategies, such as quickly contracting muscles in the trunk and lower limbs [4, 5], can provide stabilizing torques to counteract the rotational forces acting on the COM. Conversely, "change-in-support" strategies (e.g. stepping; reaching to grasp nearby handholds) are often required to recover from large destabilizations [6]. In situations where stepping reactions may not be reliable (e.g. on stairs or icy walkways), grasping reactions are important for balance recovery. It follows that handrails can significantly enhance ability to recover from balance loss [7], provided that their design enables users to quickly and accurately reach to grasp the rail, and then apply sufficient grasping forces to stabilize their COM [6].

While the value of grasping reactions for balance recovery is well-established, our understanding of how handrail height affects both COM control, and the physical demands of achieving this control during reach-to-grasp reactions, is limited. When considering the inverted pendulum model of balance control, individuals may be able to stabilize their COM while applying lower forces to the handrail when the handrail is high, due to the stabilizing moment advantage gained from higher handrails [8]. Unknown to date is how handrail height impacts COM control following forward and backward balance loss when a reach-to-grasp reaction is executed.

This study investigates the effect of handrail height and fall direction on COM control and the corresponding physical demands of reach-to-grasp balance reactions, following forward and backward platform perturbations. We hypothesized that handrail height and fall direction would affect key COM control measures, along with the physical demands of reach-to-grasp reactions in terms of the impulse applied to the rail in response to balance loss.

2 Materials and Methods

2.1 Participants

A secondary analysis of a previously-collected dataset was performed [9]. Fifty healthy young adults participated in the larger study, of which 13 participants met our inclusion criteria for analysis in this study (nine males; 18 to 28 years old). To be included in this analysis, participants wore a full-body motion capture marker set and completed all six testing conditions at perturbation magnitudes of $3.5m/s^2$ without falling or stepping (see Sections 2.3 and 2.4 for protocol details). For this study, participant heights ranged from 159 to 193cm (mean height: 177$\pm$10cm); weights ranged from 58.1 to 95.3kg (mean weight: 75.9$\pm$11.4kg). All participants reported being free of neurological, vestibular and musculoskeletal disorders. The institutional and university's Research Ethics Boards approved this work. All participants provided informed consent.

2.2 Experimental setup

Data were collected using a 5m x 5m laboratory, secured to a robotic platform that can deliver balance perturbations (Figure 1a). An overhead safety harness protected participants from contact with the floor; slack in the harness line allowed participants to move naturally after balance loss. Participants wore knee guards to minimize possible impact with the floor, and a guard on the right elbow to minimize impact from potential contact with the handrail. Participants recovered balance using a height-adjustable, horizontal handrail (outer diameter: 3.8cm) (Figure 1b). Fourteen passive motion capture cameras collected kinematic data (Motion Analysis Inc, Santa Clara, CA). Two load cells (one at either end of the handrail) collected handrail loading data (AMTI MC3A-1000; Advanced Medical Technology, Inc, Watertown, MA).

2.3 Perturbation design

Sudden platform translations disrupted balance: backward platform movements simulated forward-directed falling (Figure 1d); forward platform movement simulated backward-directed falling (Figure 1e). Perturbations consisted of square-wave acceleration profiles with a 300ms acceleration pulse, followed immediately by an equal and opposite deceleration pulse (acceleration magnitude=$3.5m/s^2$; peak velocity=1.1m/s; displacement=0.32m).

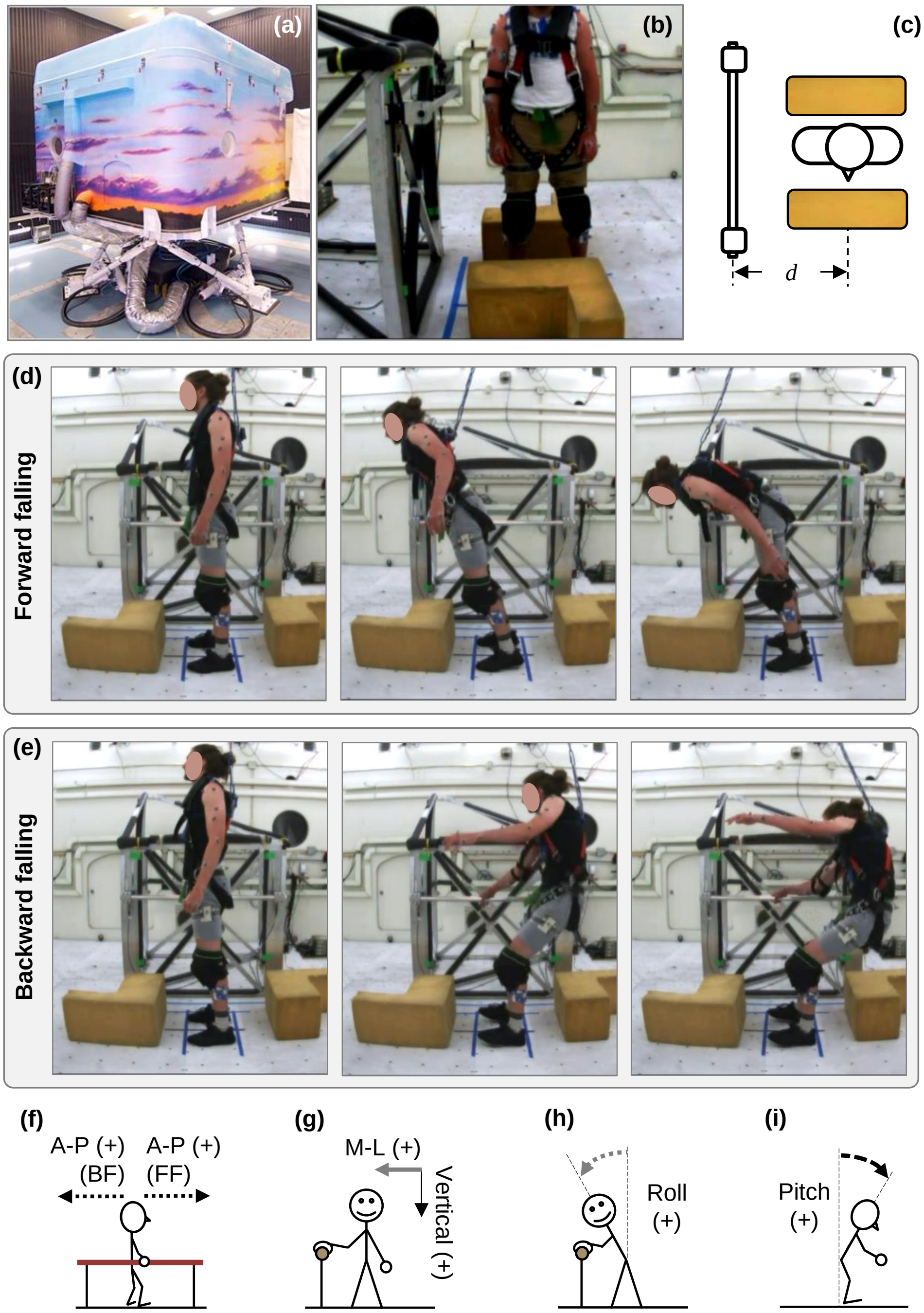

Figure 1: Testing environment and axis conventions. (a) The Challenging Environments Assessment Laboratory. **(b)** Inside the lab: a participant stands beside the handrail while wearing a safety harness. Foam blocks were used to discourage participants from stepping. **(c)** Participants stood beside the rail during testing, with their centerlines a distance of *d*=58% of their arm length away from the rail. **(d)** Sample screen captures of forward falling (induced via backward platform translations) with the LOW handrail. **(e)** Sample screen captures of backward falling (induced via forward platform translations). **(f)** COM conventions for the antero-posterior (A-P) axis for forward falling (FF) and backward falling (BF): A-P variables were calculated in the falling direction. **(g)** COM conventions for the medial-lateral (M-L) and vertical axes: M-L variables were calculated based on movement toward the handrail. **(h)** Trunk roll angular convention. **(i)** Trunk pitch angular convention.

2.4 Protocol

Individual height, weight and right arm length (acromion to fingertip) were measured. Reflective motion capture markers were used to track whole-body COM (see Section 2.5 for details on COM estimation calculations). Rigid marker clusters were secured to the pelvis (cluster at the sacrum), upper body (cluster near the thoracic level of T12), and bilaterally mid-thigh and mid-shank. The distal and proximal ends of these segments with respect to tracking markers were identified in a neutral, stationary pose. Upper limb movement was approximated with tracking markers on the hands (base of the second and fifth metacarpals), wrists (ulnar and radial joints), elbows (medial and lateral epicondyles) and shoulders (acromion; front and back of the glenohumeral joint). Participants wore standardized athletic shoes during testing.

To begin testing, participants stood erect beside the handrail with their arms relaxed at their sides and feet approximately shoulder-width apart. Their body's center-line was 58% of their right arm length away from the handrail (Figure 1c), which approximates the distance between the elbow and middle fingertip (adapted from [10]). Foam blocks in front of and behind participants' feet discouraged compensatory stepping.

Upon experiencing a perturbation, participants were instructed to reach to grasp the handrail as quickly as possible to re-stabilize, and to avoid stepping or falling into the harness. All participants received at least four lower-magnitude perturbations for each handrail height, which allowed participants to gain familiarity with the protocol for each handrail height. These perturbations were initially delivered at 2.5 m/s^2 in each direction (forward and backward) and increased in 0.5 m/s^2 intervals before the trials analyzed in this study were reached (at 3.5 m/s^2). If the participant took a step or fell during a familiarization trial, the trial was repeated. This resulted in a minimum of four lower-magnitude perturbations for each handrail height before data was included for analyses. To minimize pre-planning of movements, perturbation timing and falling direction were randomized. Participants counted backward from a randomly-selected start number by an integer between two and nine to distract attention.

For each fall direction, three handrail heights were tested: LOW (86.5cm/34 in); MED (96.5cm/38 in); and HIGH (106.5cm/42 in). 34 inches and 38 inches approximate the lower and upper boundaries of the International Building Code handrail height requirements on stairs and ramps [11], while 42 inches approximates the maximum height of handrails built into stairway landings in the Ontario

Building Code [12]. This resulted in six testing conditions, with one trial per testing condition for each participant. The testing order of rail heights and fall directions was randomized.

2.5 Data processing

Motion capture and handrail force data were sampled at 250Hz and 1000Hz respectively, and synchronized offline [13]. Inertial artifacts in the handrail force signals due to platform motion were removed by subtracting force recordings collected without a participant contacting the handrail. COM and trunk angular kinematics were estimated with a twelve-segment, link-segment model (Visual 3D; C-Motion Inc, Germantown, MD). COM kinematics were calculated from a weighted average of trunk, pelvis, upper- and lower-limb segments, with individual segment COMs approximated with existing anthropometric models [14, 15].

All kinetic and kinematic signals were filtered with zero-lag, low-pass Butterworth filters with the following orders and cut-off frequencies: 1) load cell signals: 2^{nd}-order/20Hz; 2) COM kinematics: 2^{nd}-order/6Hz; 3) trunk angular kinematics: 4^{th}-order/6Hz. Power analyses revealed that 99% of the signal power was under 6Hz for all of our analyzed COM and trunk angular position signals. Visual inspection of all filtered kinematic signals further confirmed that the overall signal shape was preserved, particularly where peak position and velocity values were extracted. Force and COM data filters were applied in MATLAB (The Mathworks, Inc, Natick, MA); trunk angular data filters were applied in Visual 3-D. COM position data were differentiated to calculate velocity, and multiplied by individual mass to calculate momentum.

2.6 Data analysis

To evaluate balance control, COM and trunk angular kinematics were analyzed (Figure 1d/e).

Peak COM displacement, velocity and momentum were calculated 1) along the A-P axis, in the fall direction (Figure 1f); 2) along the M-L axis, toward the handrail (Figure 1g); and 3) along the vertical axis, downward (Figure 1g). COM positional range was calculated along the anterior-posterior (A-P), medial-lateral (M-L) and vertical axes. In contrast with displacement, the COM range describes the difference between the highest and lowest COM position recorded during a trial; the measurement of range accounts for when participants moved in both directions along the same axis. Similar to COM kinematics, the trunk kinematic metrics included determination of both peaks and ranges: 1) Trunk roll angle range, and peak trunk roll angular velocity and momentum (toward the rail); and 2)

Peak forward trunk pitch angular displacement, velocity and momentum. Trunk roll and pitch angles were defined with respect to the vertical axis (Figure 1h,i), in the coronal (roll), and sagittal (pitch) planes. All kinematic metrics were calculated after perturbation onset (platform acceleration $\geq$ 0.1m/s^2 [13]). The exception was displacement metrics, which were calculated with respect to participants' standing position immediately before perturbation onset. Note that the coordinate system translated with the platform.

The resultant handrail impulse was calculated as a proxy measure of the physical demands of reactive grasping. Handrail impulse was defined as the time integral of the normalized resultant handrail force curve, calculated over 1) 500ms, 2) 1000ms, 3) 1500ms, and 4) 2000ms after initial handrail contact (where initial contact is defined by handrail force along any axis > 25N). Handrail force data were normalized to a percentage of the participant's body weight (%BW) to facilitate comparisons between participants, resulting in units of %BW*s for impulse. Impulse was calculated up to 2000ms after initial handrail contact, to capture key elements of balance recovery with the handrail, including slowing of COM velocity and restoration of COM position for at least 1s after a participant's highest COM velocity and displacement were observed.

Statistical analyses were performed using 2x3 repeated measures ANOVAs (SAS Enterprise Guide version 9.1, Cary, NC), with fall direction (forward, backward) and handrail height (LOW, MED, HIGH) comprising the within-subject factors. Data were rank-transformed to meet ANOVA normality assumptions [16]. Post-hoc pairwise comparisons with Tukey adjustments to account for multiple comparisons were performed following identification of significant main effects. Where interaction effects were identified, only pairwise comparisons of handrail height within each fall direction were considered (i.e., forward-LOW was not compared to backward-HIGH). Sphericity was confirmed with Mauchly's test in SPSS (IBM, Armonk, NY). Significance levels were $p<0.05$ for all analyses.

3 Results

3.1 Balance recovery strategy – forward-directed versus backward-directed falling

Balance recovery strategies from one participant, during forward-directed and backward-directed falling, are shown in Figure 1d and Figure 1e. Characteristic COM kinematic, trunk angular kinematic and resultant handrail force profiles are shown below in Figure 2.

Forward-directed falling was generally characterized by the trunk pitching forward, with little downward displacement of the hips (Figure 1d; Figure 2c/d). Conversely, backward-directed falling often involved dropping the hips, while the trunk remained relatively upright (Figure 1e). Backward-directed falling generally demonstrated greater COM displacement and velocity magnitudes (Figure 2a/b) and handrail forces (Figure 2e) compared to forward-directed falling. On average, participants contacted the handrail more quickly during forward falling (338$\pm$41ms) than backward falling (351$\pm$32ms) ($F(1,12)=8.17$;$p=0.014$). Handrail contact time did not vary significantly with rail height ($F(2,24)=1.24$; $p=0.306$).

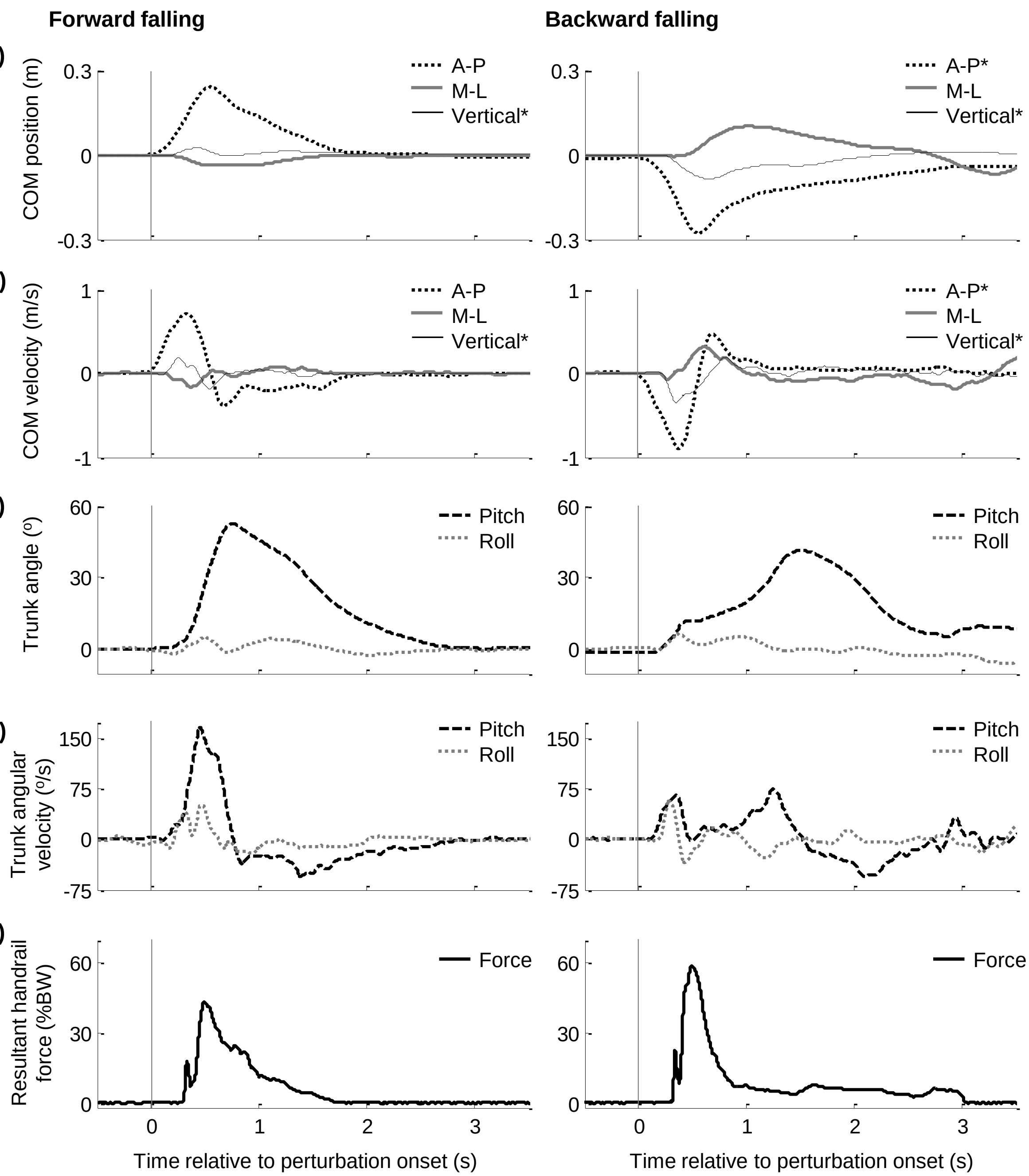

Figure 2: Characteristic COM kinematics, trunk angular kinematics and resultant handrail force profiles during forward and backward falling. All traces are from one participant, when the handrail was LOW. **(a)** COM position in the anterior-posterior (A-P), medial-lateral (M-L) and vertical axes. Note that the sign convention for the A-P axis during backward falling was not flipped, to more clearly indicate backward COM movement. **(b)** COM velocity in all three Cartesian axes. **(c)** Trunk pitch and roll angles. **(d)** Trunk pitch and roll angular velocity. **(e)** Resultant handrail force. Momentum traces were excluded as the shape of these traces was the same as that of velocity. The asterisk (*) for A-P COM traces during backward falling, and all vertical COM traces, indicates that the axis conventions are opposite to those shown in Figures 1f/g. The sign convention for the A-P axis during backward falling was not flipped to more clearly indicate backward COM movement, while the vertical axes were not flipped to highlight initial upward COM movement during forward falling, versus downward movement during backward falling.

3.2 Effect of fall direction and handrail height on the physical demands of reactive grasping – handrail impulse

Handrail impulse (Figure 3) during backward-directed falling significantly exceeded that of forward-directed falling (all $F(1,12)$'s$\geq$16.51; all p's$\leq$0.002), with the greatest increase to impulse occurring in the first 500ms after handrail contact. Further, impulse decreased significantly as handrail height increased (all $F(2,24)$'s$\geq$3.49; all p's$\leq$0.047). However, only the increase from LOW to HIGH was significant (500ms: $p_{L\text{-}H}<0.001$; 1000ms: $p_{L\text{-}H}=0.014$; 1500ms: $p_{L\text{-}H}=0.039$; all other pairwise p's (LOW-MED, and MED-HIGH) $\geq$ 0.083). The increase from LOW to HIGH was not significant at 2000ms, following Tukey corrections ($p_{L\text{-}H}=0.072$).

Significant interaction effects between fall direction and handrail height were not observed for impulse (all interaction $F(2,24)$'s$\leq$0.98; all interaction p's $\geq$0.115).

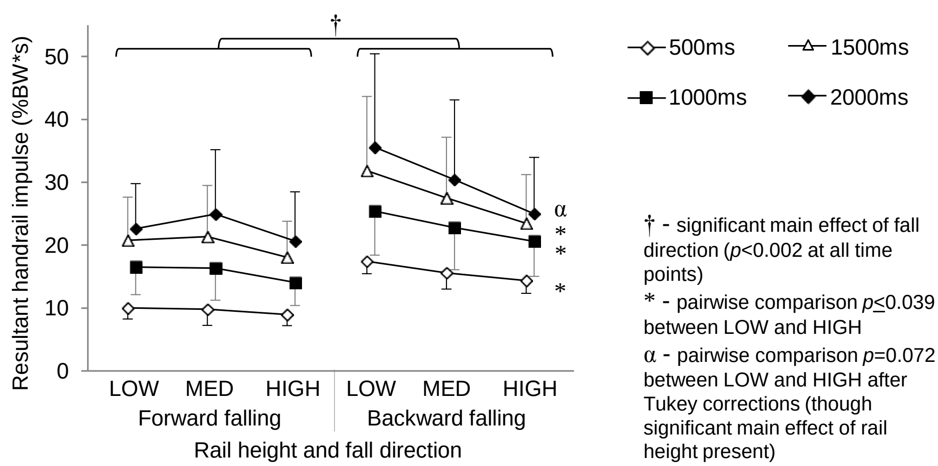

Figure 3: Resultant handrail impulse for each condition, measured 500ms, 1000ms, 1500ms and 2000ms after initial handrail contact. Mean values are plotted; error bars represent standard deviation.

3.3 Effect of fall direction and handrail height on COM and trunk control

3.3.1 COM kinematics

Compared to forward-directed falling, backward-directed falling resulted in significantly worse COM control (i.e., higher COM range, displacement, peak velocity and peak momentum) in all axes, with the exception of COM range in the A-P axis (A-P COM range: fall direction $F(1,12)=1.23$; $p=0.082$; all other COM metrics: fall direction $F(1,12)$'s$\geq$15.38; p's$\leq$0.002) (Figure 4).

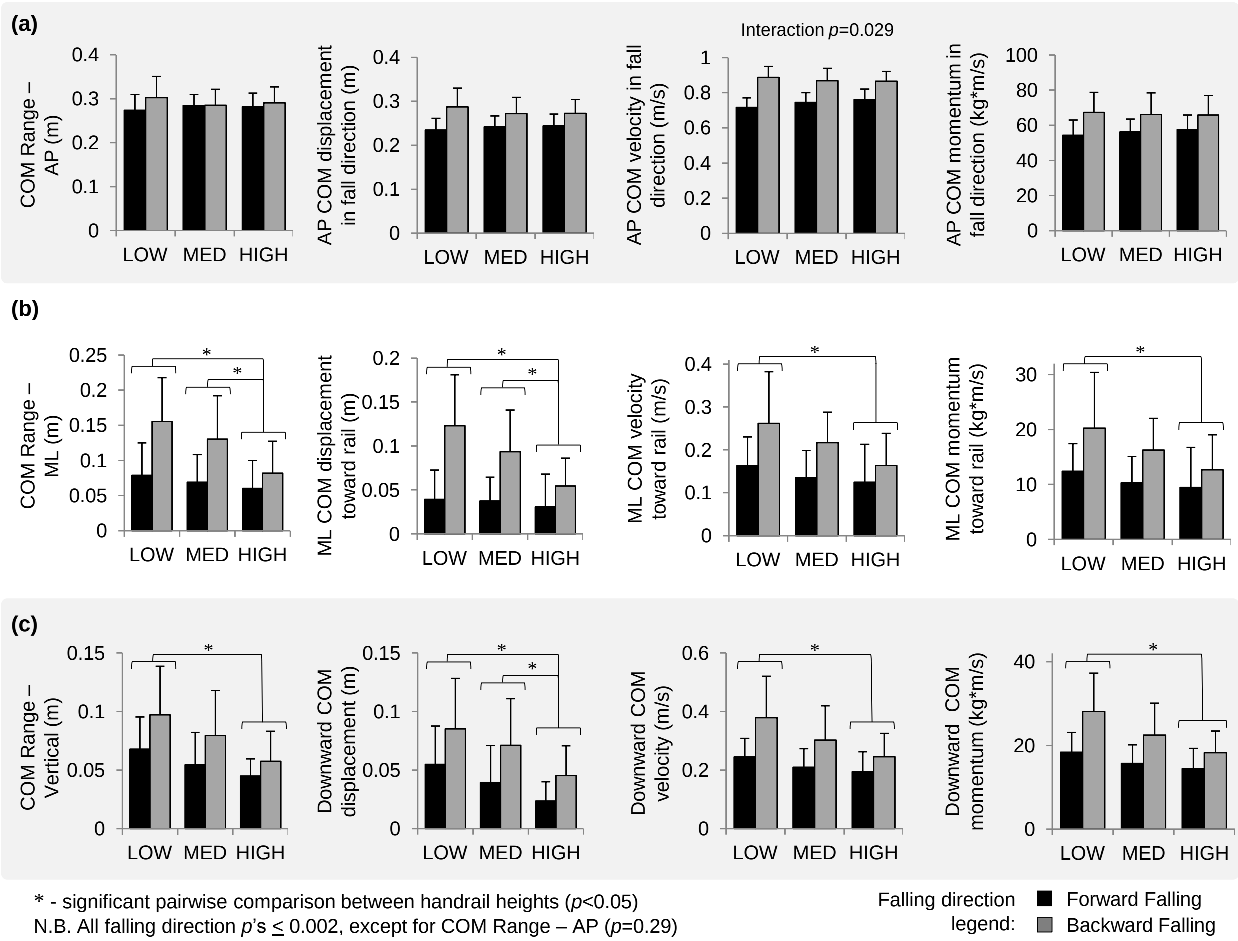

Figure 4: COM range, and peak displacement, velocity and momentum magnitudes in the **(a)** antero-posterior (A-P), **(b)** medial-lateral (M-L), and **(c)** vertical axes. Bars represent mean values; error bars represent standard deviation.

Significant main effects of handrail height on COM control were not found in the A-P axis (F(2,24)'s≤0.77; *p*'s ≥0.475). In the M-L and vertical axes, all COM control variables decreased significantly (indicating better COM control) as handrail height increased (handrail height main effect F(2,24)'s ≥5.77; *p*'s≤0.009). Pairwise comparisons revealed that the decrease in COM variables as rail height increased from LOW to HIGH was significant in both M-L and vertical axes (all LOW-HIGH *p*'s ≤ 0.007). The decrease from MED to HIGH was also significant for COM range and displacement in the ML axis, and for downward COM displacement (*p*'s≤0.029).

The only observed significant interaction effect was for peak COM velocity in the A-P axis (F(2,24)=4.13; *p*=0.029), which rose as handrail height increased during forward-directed falling, but decreased as handrail height increased for backward-directed falling. Significant interaction effects were not observed for other COM control variables ($F(2,24)$'s$\leq$2.69; p's$\geq$0.089).

3.3.2 Trunk angular kinematics

Trunk roll range was significantly higher for backward falling compared to forward falling ($F(1,12)=10.56$; $p=0.007$), while peak trunk roll velocity and momentum did not differ significantly between falling directions ($F(1,12)$'s$\leq$3.28; p's$\geq$0.095) (Figure 5a,b,c). Conversely, peak forward trunk pitch displacement, velocity and momentum were significantly higher for forward falling than for backward falling ($F(1,12)$'s$\geq$137.37; p's$<$0.001) (Figure 5 d,e,f).

All trunk angular metrics decreased significantly as handrail height increased ($F(2,24)$'s$\geq$10.88; p's$<$0.001). Pairwise comparisons revealed that all trunk angular metrics decreased significantly as handrail height increased from LOW to HIGH (p's$\leq$0.001). Trunk pitch metrics further decreased from LOW to MED (p's$\leq$0.008). For roll, all decreases from LOW to MED, and MED to HIGH, were significant, with the exception of the roll angle range LOW-to-MED decrease (roll angle range $p_{L-M}=0.124$; all other roll p_{L-M}'s and p_{M-H}'s$\leq$0.039).

Significant interaction effects between fall direction and handrail height were not found (all interaction $F(2,24)$'s$\leq$1.51; all interaction p's$\geq$0.346).

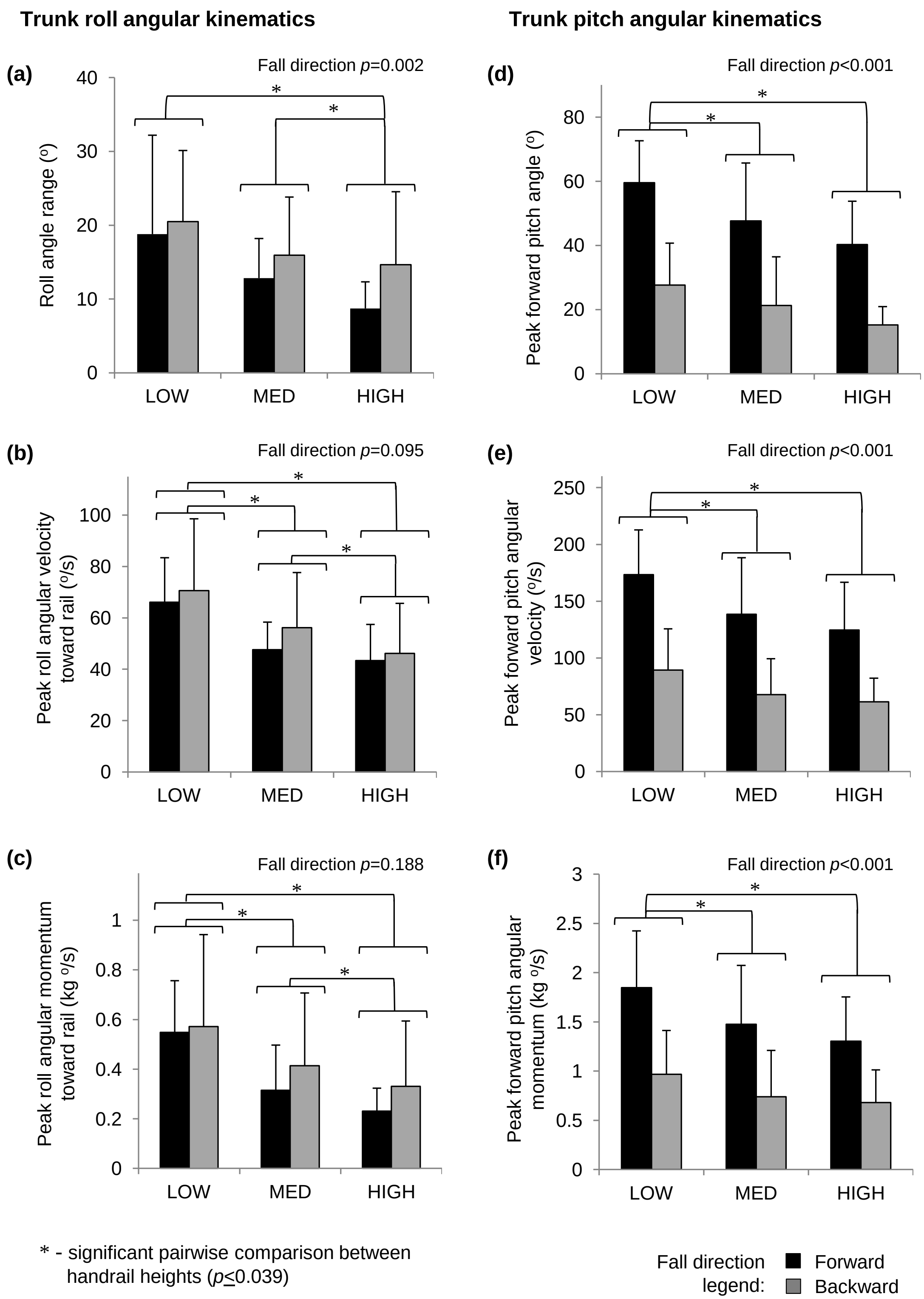

Figure 5: Trunk angular kinematics for each fall direction and handrail height. Bars represent mean values; error bars represent standard deviations. **(a)** Trunk roll angle range. **(b)** Peak roll angular velocity (toward the handrail). **(c)** Peak roll angular momentum (toward the handrail). **(d)** Peak forward pitch angle. **(e)** Peak forward pitch angular velocity. **(f)** Peak forward pitch angular momentum.

4 Discussion

To avert a fall after balance loss, one must regain control of the position and velocity of their COM and trunk. The results of this study indicate that both falling direction and handrail height significantly affected trunk and COM control following platform perturbations, along with the impulse that participants applied to the handrail to regain stability.

4.1 Backward-directed falling resulted in poorer COM control and greater physical demands of grasping, compared to forward-directed falling

Young adults have been previously observed to fall more frequently from backward destabilizations than from forward destabilizations [17], with higher wrist impact velocities [18]. In this study, backward-directed falling resulted in less-controlled COM kinematics than forward-directed falling, even though the stabilizing impulse that participants applied to the handrail during backward-directed falling consistently exceeded that of forward-directed falling. The heightened handrail impulse during backward-directed falling may be explained in part by how both the real and perceived instability of backward-directed falling may exceed that of forward-directed falling. Reduced stability during backward falling may have stemmed from the center of pressure being closer to the posterior edge of the base of support while standing, demanding a larger response during a backward-directed fall particularly in the absence of stepping. Poorer visual perception of the body relative to the floor in the falling direction may have further increased the perception of instability, and contributed to larger applied handrail forces in response to backward-directed falling [19, 20], over the course of the balance recovery response. Backward-directed destabilizations have been shown to elicit stronger responses compared to forward-directed destabilizations, including greater dorsiflexor and plantarflexor co-contraction [21] and increased likelihood of reaching for handrails [22] – even with reduced perturbation magnitudes for backward-directed falling [22]. Taken together, these factors may have led to more aggressive handrail use during backward-directed falling, compared to forward-directed falling.

4.2 As handrail height increased, COM and trunk control improved and the physical demands of grasping decreased

As handrail height increased, participants demonstrated consistent or better COM and trunk control in response to perturbations, even though handrail impulse decreased significantly as increases in height. An inverted pendulum model of balance [23] may help to explain the better COM and trunk control with increased rail height, without a concomitant increase to handrail forces. In this context, the handrail enables users to generate high stabilizing forces and moments to counteract the translational and rotational forces acting on the COM with respect to the ankles [8]. Accordingly, the higher rails evaluated in this study afforded greater stabilizing moments than did the lower handrails for a given applied handrail force, due to the increased moment arm between the user's ankles and the rail. These findings are consistent with past research, where higher maximum voluntary moment generation ability with increased handrail height has been observed in both younger and older adults [24, 25]. Building on these past studies, our findings suggest that the stabilizing moment advantage may have enabled participants to achieve better COM and trunk control with the higher rails evaluated in this study, while applying lower impulse to the handrail during reactive grasping.

While our results suggest a stability advantage with higher handrails tested in this study, our findings would not necessarily apply outside of the tested range. The mechanical advantage of a larger moment arm with increasing handrail height would be offset at some point by other factors, such as substantial reductions to volitional strength with handholds surpassing shoulder height [26], or the user eventually not being able to reach the handrail altogether. Further research is required to determine this optimal height for balance recovery across various populations.

We note that while statistically-significant effects of handrail height and falling direction on impulse and COM/trunk control were observed, the functional importance of these differences is unknown because participants did not step or fall in the included trials. The stability advantages with the higher handrails evaluated in this study may be more important to individuals with reduced trunk and upper-limb strength due to conditions – such as in persons with stroke [27], who may both (a) demonstrate worsened trunk and COM control, and (b) be unable to apply the greater handrail forces needed for balance recovery with lower rails. For example, reduced upper-limb strength and trunk flexion-extension isometric strength (while standing) have been observed in post-stroke patients [27, 28], who may experience greater challenges in restoring their trunk to upright stance with the greater pitch angular displacements observed with lower handrails in this study. Conversely, higher handrails may be problematic for users with upper-limb arthritis or other conditions that constrain range of motion. Further evaluations of handrail height on balance recovery should consider individuals with reduced strength and range of motion, including older adults with demonstrated age-related declines in handrail force generation ability [24], and speed and accuracy of reactive grasping [29].

4.3 Limitations and future work

This study enhances our understanding of how handrail height and falling direction influence balance recovery. However, several limitations should be acknowledged. First, we focused on reach-to-grasp reactions following perturbations of upright stance; ongoing gait was not studied. While both contexts are important, ongoing gait testing may reveal different results due to delays in arm movement onset and handrail contact with leg movement [30]. Reduced speed of reach-to-grasp reactions may result in increases to peak COM and trunk kinematic variables – and thus worsened control – due to the increased time after balance loss before the handrail can be used to generate stabilizing forces. However, the potentially-negative COM control implications of delayed reactive grasping during ongoing gait may be countered by being able to step in response to perturbations. Second, the high number of outcome measures in this study increases the risk of a Type I error within our main effects, although applying Tukey corrections reduces the likelihood of false positives within our post hoc comparisons. Third, this study evaluated three handrail heights only. While the tested handrail heights were selected to coincide with existing building standards (which increases the applicability of our findings to handrails in the community), performance in metrics may differ for handrails outside of our tested range. Finally, our sample was limited to healthy young adults. Further research should include other populations, including individuals with reduced trunk and upper-limb strength and range of motion, whose balance recovery reactions may be affected differently by varying handrail height.

5 Conclusions

We have characterized the influence of falling direction and handrail height on COM and trunk control and the corresponding physical demands of reactive grasping in younger adults. Backward-directed falling resulted in poorer COM control than did forward-directed falling, despite higher impulse applied to the handrail during backward-directed falling. Trunk control was generally worse during forward-directed falling. As handrail height increased, COM and trunk control improved, and impulse applied to the handrail decreased. Our findings suggest a possible stability advantage with increased handrail height in both falling directions within the range of tested handrail heights, demonstrated by participants having achieved greater COM and trunk control while applying lower impulse to the handrail during balance recovery.

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