Effect of Handrail Height and Age on Trunk and Shoulder Kinematics Following Perturbation-Evoked Grasping Reactions During Gait

Vicki Komisar, University of British Columbia, Kelowna, Canada, Toronto Rehabilitation Institute – University Health Network, ON, Canada, and University of Toronto, ON, Canada and **Alison C. Novak**, Toronto Rehabilitation Institute – University Health Network, ON, Canada, and University of Toronto, ON, Canada

Objective: To characterize the effect of handrail height and age on trunk and shoulder kinematics, and concomitant handrail forces, on balance recovery reactions during gait.

Background: Falls are the leading cause of unintentional injury in adults in North America. Handrails can significantly enhance balance recovery and help individuals to avoid falls, provided that their design allows users across the lifespan to reach and grasp the rail after balance loss, and control their trunk by applying hand-contact forces to the rail. However, the effect of handrail height and age on trunk and shoulder kinematics when recovering from perturbations during gait is unknown.

Method: Fourteen younger and 13 older adults experienced balance loss (sudden platform translations) while walking beside a height-adjustable handrail. Handrail height was varied from 30 to 44 inches (76 to 112 cm). Trunk and shoulder kinematics were measured via 3D motion capture; applied handrail forces were collected from load cells mounted to the rail.

Results: As handrail height increased (up to 42 inches/107 cm), peak trunk angular displacement and velocity generally decreased, while shoulder elevation angles during reaching and peak handrail forces did not differ significantly between 36 and 42 inches (91 and 107 cm). Age was associated with reduced peak trunk angular displacements, but did not affect applied handrail forces.

Conclusion: Higher handrails (up to 42 inches) may be advantageous for trunk control when recovering from destabilizations during gait.

Application: Our results can inform building codes, workplace safety standards, and accessibility standards, for safer handrail design.

Keywords: slips and falls, biomechanics, gait and posture, built environment design, design for older adults

Address correspondence to Vicki Komisar, School of Engineering, The University of British Columbia, 1137 Alumni Ave, Kelowna, BC V1V 1V7, Canada; e-mail: vicki.komisar@ubc.ca

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INTRODUCTION

Approximately 30%–40% of older adults experience falls each year (Hausdorff et al., 2001; Tinetti et al., 1988), and approximately one-half to three-quarters of these falls occur while walking (Robinovitch et al., 2013; Seniors' Falls in Canada: Second Report, 2014). While mechanisms such as rapid stepping or stiffening the lower limb muscles can help with balance recovery, their effectiveness may be limited in locations such as slippery ramps or on stairs, or following large destabilizations (Horak & Nashner, 1986; Maki & McIlroy, 1997). In these contexts, handrails play an important role in balance recovery, by allowing individuals to apply stabilizing forces and torques to the rail with their upper limbs and control the movement of their torso (Komisar, McIlroy, et al., 2019; Maki & McIlroy, 1997; Maki et al., 1998).

Despite the often-compulsory presence of handrails for preventing falls along high-risk walkways (e.g., stairs, slopes, bridges, and corridors of care facilities), knowledge of how their installation height affects trunk control during balance recovery in older adults is limited. Greater trunk angular displacements after balance loss have been associated with increased laboratory falls risk in older adults (Grabiner et al., 2008). In studies involving upright-stance platform perturbations in young adults, higher handrails (among 34, 38, and 42 inches) resulted in reduced peak center-of-mass and trunk displacement and velocity, and improved ability to withstand high-magnitude destabilizations without stepping (Komisar et al., 2018; Komisar, Nirmalanathan, et al., 2019).

However, this work explored young adults only, who were not permitted to step before or after balance loss. The extent to which higher handrails affect trunk control in older adults, who have been shown to experience greater trunk displacements during balance recovery with stepping (Carty et al., 2011; Grabiner et al., 2008), is unknown.

The influence of handrail height on concomitant applied stabilizing handrail forces during balance recovery must also be considered. Maki et al. (1984) reported that handrail height significantly affected the maximum voluntary forces that both young and older adults could apply to a handrail while standing still, which increased when applied anteriorly or posteriorly, and decreased when applied downward. Furthermore, young adults applied higher maximum voluntary forces to the handrail than older adults (Maki et al., 1984). However, it has not yet been established whether these effects of handrail height or age on applied handrail forces translate to balance recovery contexts during gait, where the brief time scales over which falls occur—often in under 1500 ms (Choi et al., 2015; Hsiao & Robinovitch, 1997)—limit the time available for individuals to select their grip location after balance loss.

Finally, shoulder kinematics of the grasping arm during balance recovery should be characterized with respect to handrail height. To reach higher handrails, individuals may require greater elevation of the upper arm with respect to the thorax, similar to the greater upper arm elevation angles previously reported during voluntary reaching tasks for targets with different heights (Gates & Dingwell, 2011). Reaching for higher handrails may be challenging when recovering from gait destabilizations where the ongoing movement of the trunk must be considered. These grasping postures may be difficult for healthy older adults with age-related declines in shoulder flexibility (Barnes et al., 2001), and/or individuals with upper-limb arthritis, hemiplegia due to stroke, or other conditions that limit joint range of motion (Frankle et al., 2005; Mouawad et al., 2011). However, the effect of handrail height on upper limb posture during balance recovery is unknown.

This study characterizes the influence of handrail height and age on trunk and shoulder kinematics following perturbation-evoked reach-to-grasp reactions during gait. We also evaluated the peak handrail forces during balance recovery as an explanatory variable, to gain insights into the concomitant demands on the handrail during grasping. While handrails are more common for balance recovery on stairs, our protocol involved level ground perturbations for the safety of older adults who participated. We hypothesized that: (1) trunk control would improve with increased handrail height; (2) shoulder elevation angles would similarly increase with handrail height; and (3) older adults would demonstrate poorer stability than younger adults, indicated by higher trunk angular displacements and velocities following balance perturbations. This study was a secondary analysis of a larger dataset, from which we previously analyzed a different subset of the data and found no association between handrail height and the time to contact the handrail after balance loss (Komisar, Maki, et al., 2019). The present study analyzes complimentary research questions related to torso control and shoulder kinematics, and their association with handrail height.

METHODS

Participants

We collected and analyzed data from 14 young adults and 13 older adults in this study (see demographics in Table 1; all participants were right-handed). This analysis was part of a larger study (Komisar, Maki, et al., 2019). This study complied with the Declaration of Helsinki and was approved by institutional research ethics boards, and all participants provided written informed consent.

Setup

The testing environment consisted of a 5 m × 5 m laboratory mounted to a robotic platform (Figure 1). The platform can translate quickly to perturb participant balance. Participants walked along a 4.8 m × 2.4 m walkway, which consisted of four force plates and four wooden tiles with the same dimensions (1.2 m × 1.2 m; see layout in Figure 1c). To maintain consistency in the walkway surface, the force plates and walkway tiles

TABLE 1: Participant Demographics

		Age	Height	Weight
Young adults (*n* = 14)	Mean (*SD*)	24.4 (3.3) years	170.3 (9.9) cm	68.4 (15.8) kg
	Range	19 to 33 years	157.5 to 186.0 cm	50.8 to 107.0 kg
Older adults (*n* = 13)	Mean (*SD*)	67.8 (4.9) years	168.1 (6.8) cm	72.4 (16.6) kg
	Range	60 to 77 years	155 to 178.5 cm	46.3 to 100.2 kg

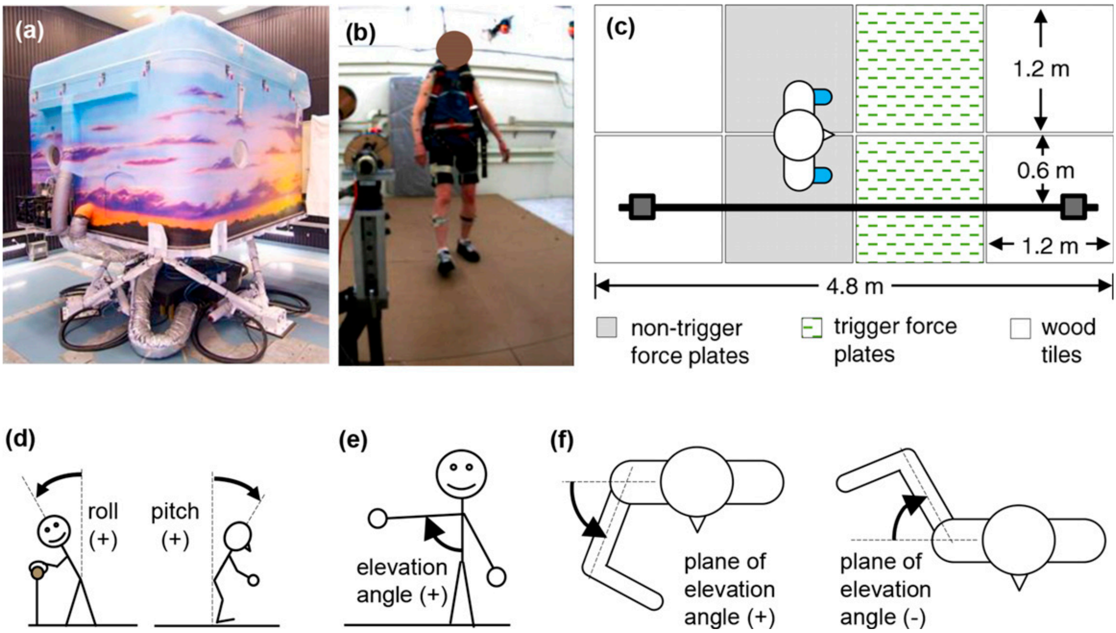

Figure 1. Laboratory setup and definitions of kinematic outcome measures. (a) External view of laboratory, mounted to a robotic platform. (b) Laboratory interior. (c) Schematic diagram of the laboratory (transverse view). Participants walked back and forth along the 4.8m-long walking surface, with the center-line of the surface approximately aligned with participants' sagittal plane. (d) Trunk roll and pitch angular definitions, comprising the roll and pitch of the trunk with respect to the vertical axis. (e) Thoracohumeral (shoulder) elevation angle definition, comprising the elevation of the upper arm with respect to the thorax. (f) Shoulder plane of elevation angle definition, comprising the angle between the humerus and the participant's coronal plane (at the level of the thorax). Anterior positioning of the humerus is characterized by a positive angle; posterior positioning of the humerus is characterized by a negative angle.

were covered with 1.2 m × 1.2 m fiberboard panels (Flakeboard, USA). A height-adjustable handrail was mounted beside the walkway, 60 cm from the center. The handrail was on the right-hand side of participants during balance disturbances. Participants wore a safety harness attached to an overhead robotic unit, which followed them as they walked to maintain consistency in harness line length. Slack in the harness line permitted trunk movement after destabilizations.

Participants wore standardized athletic shoes (Athletic Works Running Shoes) to reduce the potential for confounding due to differences in footwear. A layer of felt and 100% cotton fabric was placed on the outside of the shoe soles to reduce the traction between participants' shoes and the fiberboard walking surface, and increase the challenge of compensatory stepping during balance recovery (Komisar, Maki, et al., 2019).

Kinematic data were collected with 12 passive motion capture cameras mounted to the laboratory walls (Motion Analysis Inc., Santa Clara, USA) at 250 Hz. The coordinate system translated with the laboratory during perturbations. Handrail force data were collected with triaxial load cells mounted to each end of the handrail (AMTI MC3A-1000; Advanced Medical Technology, Inc, Watertown, MA) at 1000 Hz.

Experimental Protocol

Participants were instructed to walk back and forth along with the walking surface, at a self-selected pace. During randomly selected walks when the handrail was to the right of participants, the platform quickly translated backward relative to the participant when they stepped on and applied 100N of vertical force to either "trigger" force plate during the walk (Figure 1c), to disrupt participant balance. The perturbations were intended to represent a trip on a surface where compensatory stepping would be challenging. Participants were instructed to reach to grasp the handrail as quickly as possible upon experiencing a perturbation ("trial"). All perturbations consisted of a 300-ms square-wave acceleration pulse, followed by an equal and opposite deceleration pulse (acceleration amplitude = 3.75 m/s^2; peak velocity = 1.15 m/s; displacement = .35m; Komisar, Maki, et al., 2019).

All participants completed a verbal task while walking, leading into each perturbation. The task consisted of either (1) counting backward by 3's or 7's from a researcher-selected number (between 100 and 1000); (2) naming animals starting with every letter of the alphabet; or (3) conversing with the researcher inside the laboratory. As the purpose of the verbal task was to distract attention, participant performance in the verbal task was not tracked, and participants were permitted to choose and switch tasks during the experiment. Participants were instructed to look forward while walking and avoid directing their gaze at the researcher in the laboratory. We only induced perturbations while participants looked forward.

We tested a total of eight handrail heights per participant: 30, 32, 34, 36, 38, 40, 42, and 44 inches (76, 81, 86, 91, 97, 102, 107, and 112 cm, respectively). After completing a familiarization trial, participants completed four perturbation trials per handrail height. As pilot analysis revealed that trial number did not affect our outcomes ($p \geq .193$), the first three of these trials are analyzed in this study to reduce data processing time. We randomized the order of testing handrail heights for each participant, though all trials at a given rail height were completed back-to-back to reduce protocol duration. Participants also completed four trials per handrail height with the laboratory inclined at 8^{o} (Komisar, Maki, et al., 2019), though the slope gait trials were not analyzed in this study.

The total experimental duration (including participant setup and the sloped gait trials not included in this analysis) was approximately 2–2.5 hr per participant, depending on rest (participants were told that they could rest as often as desired, to avoid fatigue). All participants rested for approximately 10 min halfway through the perturbation trials. Every four trials, the researcher asked the participant to rate their perceived exertion based on the Borg scale (Borg, 1982), which did not surpass a rating of 13 (corresponding to "somewhat hard") for any participant during testing.

Kinematic Data Collection, Processing, and Modeling

Reflective motion capture markers were used to track trunk and upper-limb kinematics. Rigid marker clusters were secured to the pelvis (cluster at the sacrum) and upper body (cluster near the thoracic level of T12). The distal and proximal ends of these segments were identified in a neutral, stationary pose. Movement of the right upper arm was tracked via markers on the elbows (medial and lateral epicondyles) and shoulders (acromion; front, back and distal side of the glenohumeral joint).

Kinematic and kinetic data sources were synchronized offline (Komisar et al., 2017). Trunk and shoulder angular kinematics were estimated with a link segment model, consisting of three segments: the pelvis, the trunk, and the right upper arm (Visual 3D; C-Motion Inc, Germantown, MD). All motion capture markers were filtered with fourth-order, zero-lag, low-pass Butterworth filters prior to model estimation (MATLAB; The Mathworks, Inc, Natick, MA). Markers on the trunk and shoulders were filtered with a cut-off frequency of 10 Hz, while markers on the epicondyles were filtered at 30 Hz. Filter parameters were selected based on a combination of power analysis (Kuo et al., 2020) and visual inspection. Specifically, 99% of the signal power for all markers used in analysis was below 10 Hz, based on data from three randomly selected trials. However, the position signals from the epicondyles were sensitive to filtering around the time that participants contacted the handrail (likely due to the sudden deceleration of the forearm segment when the hand contacted the handrail), with >1 cm of position error imposed by the filter in some cases. Filtering at 30 Hz preserved the key features of these signals (based on visual inspection), thereby enabling more accurate estimation of shoulder angles.

Handrail Force Data Collection and Processing

Inertial artifacts in the handrail force signals (resulting from platform motion) were removed by subtracting force recordings collected without a participant contacting the handrail (Komisar, Nirmalanathan, et al., 2019). Handrail force signals were then filtered with a fourth-order, dual-pass, low-pass Butterworth filter with a cut-off frequency of 8 Hz, to attenuate noise due to the vibration of the handrail. The filter cut-off frequency was based on the lowest natural frequency of the handrail (approximately 10 Hz, based on Fourier analysis), which occurred when the rail was 44 inches high.

Outcome Measures

Trunk kinematics. Trunk angular kinematic data were analyzed to evaluate balance control. Metrics were defined with respect to the vertical axis in global coordinates, in the individual's coronal (roll) and sagittal (pitch) planes (Figure 1d; Novak et al., 2016). Variables of interest included peak forward trunk angular displacement and velocity, and peak angular displacement and velocity toward the handrail. Peak angular displacements were defined as the difference between peak trunk angular position along the axis of interest, and the participant's trunk angular position in the frame immediately before perturbation onset (platform acceleration $>.1\ m/s^2$; Komisar, Maki, et al., 2019).

Shoulder kinematics. To quantify the influence of handrail height on shoulder posture, we considered: (1) peak thoracohumeral ("shoulder") elevation angles (measured between perturbation onset and completion of using the handrail for balance recovery), and (2) concomitant shoulder plane-of-elevation angles (Figure 1e and f for illustrated definitions). Shoulder angles were defined based on the joint coordinate systems recommended by the International Society of Biomechanics (Wu et al., 2005). The shoulder elevation angle represents the elevation of the humerus with respect to the thorax, where 0^o was defined as when the humerus was parallel with the individual's sagittal and coronal planes (Aizawa et al., 2010). Conversely, the peak shoulder plane-of-elevation angle represents the transverse-plane rotation of the humerus relative to the participant's coronal plane (Figure 1f). The shoulder plane-of-elevation angle had a value of 0^o when the upper arm coincided with the participant's coronal plane; a value of 90^o occurs when the upper arm is anterior to the thorax and coincident with the participant's sagittal plane (Aizawa et al., 2010). Biomechanical literature describes positive and negative plane-of-elevation angles as analogous to shoulder horizontal flexion and extension, respectively (Aizawa et al., 2010).

Handrail forces. We extracted the peak resultant forces that participants applied to the handrail to quantify the contribution of the handrail to balance recovery. Handrail forces were analyzed as a percentage of participant body weight (% BW).

Gait speed. Participant gait speed leading into the perturbation was estimated to characterize possible differences in behavior due to age. Gait speed was defined as the quotient between: (a) the total anterior displacement of a marker on the participant's pelvis cluster, starting from 1.2m away from the force plate that triggered perturbations (Figure 1c), up to the frame before perturbation onset, and (b) the time required for the participant to walk that distance.

Data Analysis

We analyzed each variable of interest using a 2 × 8 mixed repeated measures analysis of variance (ANOVA; SAS Enterprise Guide version 9.1, Cary, NC). Handrail height comprised the within-subject factor; age was modeled as a between-subject factor. To control for individual height, we included individual height as a covariate. All trials were included in analysis. All metrics were rank-transformed to meet ANOVA normality assumptions (Conover & Iman, 1981). We used significance levels of $\alpha = .05$ for all analyses. Post hoc comparisons with Tukey corrections were performed following identification of significant main effects (age, handrail height) and interactions (age × handrail height), to identify differences in balance recovery strategy between handrail heights.

RESULTS

Trunk Angular Kinematics

Handrail height significantly affected trunk forward pitch displacement and velocity, and trunk roll displacement and velocity toward the rail, which generally decreased as handrail height increased (all height main effect $F(7,175) \geq 17.61$; $p < .001$; pairwise comparisons in Figure 2). Furthermore, trunk forward pitch displacement and velocity were both significantly greater in young adults than older adults (age main effect $F(1,24) \geq 16.85$; $p < .001$; Figure 2a and c). However, trunk roll displacement and velocity toward the rail did not differ significantly between age groups (age main effect $F(1,24) \leq 2.63$; $p > .118$; Figure 2b and d). Significant age × handrail height interactions were not found for any trunk angular kinematic metrics ($F(7,175) \leq 1.37$; $p > .223$).

Handrail Forces

Peak handrail forces were included as an explanatory variable for the observed differences in performance in trunk angular kinematics with respect to age and rail height. Peak resultant handrail forces showed slight, but statistically significant differences with handrail height ($F(7,175) = 2.99$; $p = .005$; Figure 3), and generally decreased as rail height increased, from 32 inches (17.9% BW) to 44 inches (15.4% BW). However, post hoc comparisons revealed that the only significant differences between individual handrail heights were for 32 inches versus 44 inches, and 34 inches versus 44 inches ($p < .05$); all other comparisons between handrail heights were not significant.

Significant main effects of age on peak handrail forces were not observed ($F(1,24) = 0.261$; $p = .614$), nor were significant age × handrail height interactions found ($F(7,175) = 1.26$; $p = .273$).

Shoulder Elevation and Plane-of-Elevation Angles

Shoulder elevation and concomitant plane-of-elevation angles varied significantly with handrail height ($F(7,175) \geq 6.27$; $p < .001$; Figure 4). As handrail height increased, shoulder elevation angles generally increased, while concomitant shoulder plane-of-elevation angles generally increased in magnitude in the negative direction (indicating shoulder extension). Significant main effects of age, or age × handrail height interactions, were not found for either shoulder elevation or concomitant plane of elevation (age main effect: $F(1,24) \leq 1.39$; $p > .250$; age × handrail height interaction: $F(7,175) \leq 1.23$; $p > .286$).

Gait Speed

Gait speed leading into perturbation onset did not vary significantly with handrail height or age (handrail height $F(7,175) = 0.681$; $p = .688$; age $F(1,24) = 0.060$; $p = .809$), nor were significant age × handrail height interactions identified ($F(7,175) = 0.565$; $p = .783$). Young adults had an average gait speed of .77 m/s ($SD = .15$ m/s), while older adults had an average gait speed of .78 m/s ($SD = .20$ m/s).

DISCUSSION

Handrails can significantly enhance a person's ability to recover from balance loss and avoid a fall, provided that their design allows the person

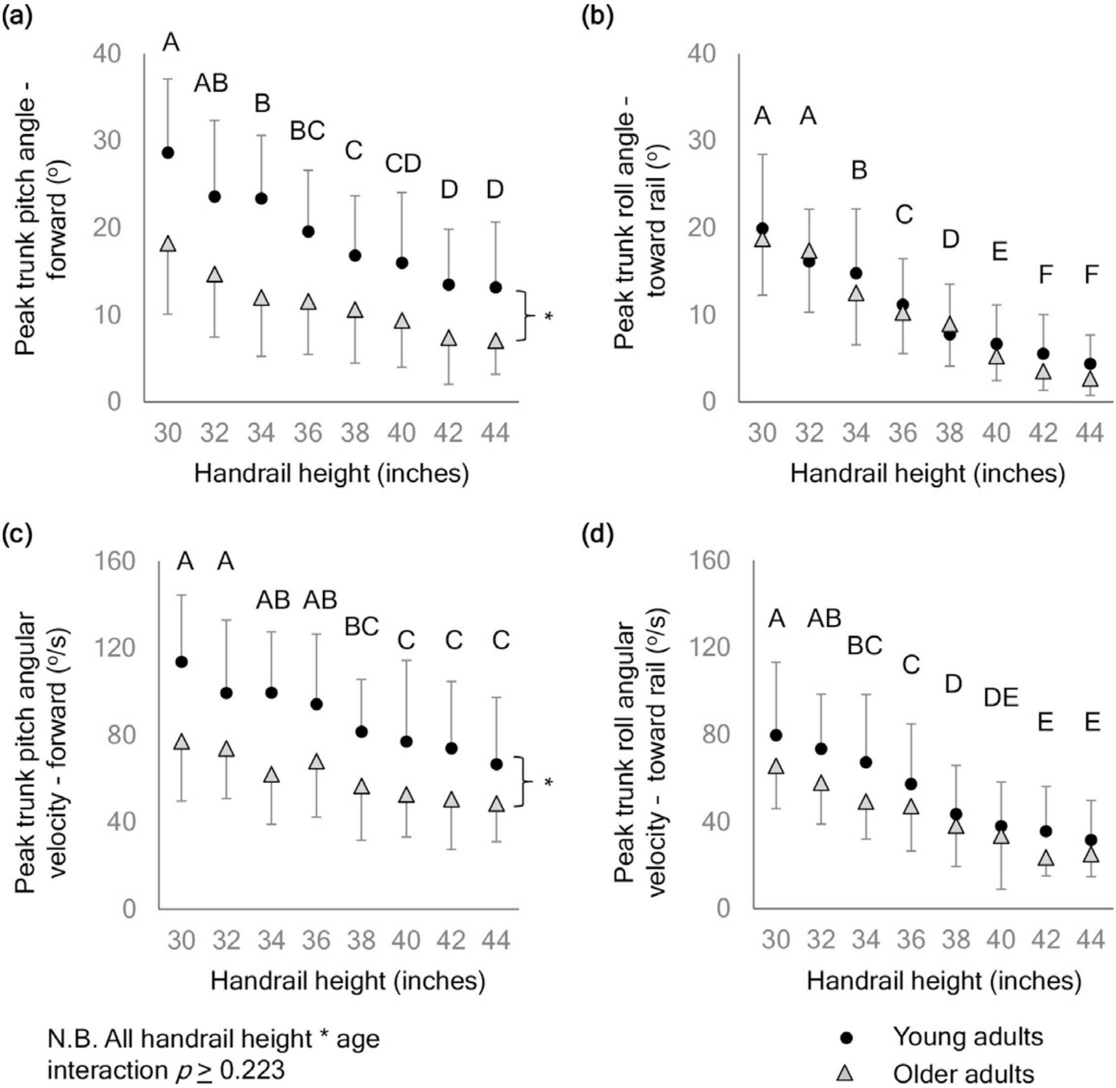

Figure 2. Trunk angular kinematics. (a) Peak forward trunk pitch displacement. (b) Peak trunk roll displacement toward the rail. (c) Peak forward trunk pitch velocity. (d) Peak trunk roll velocity toward the rail. Mean values for each population and condition are shown; error bars represent 1 standard deviation. Letters indicate post hoc pairwise comparisons: comparisons that did not differ significantly are indicated by the same letter.

to apply hand-contact forces to the rail to control the position and velocity of their trunk. This study expands our understanding of how handrail installation height affects trunk control and shoulder kinematics while reaching when recovering balance from perturbations during gait, and how these outcomes depend on age.

Consistent with handrail studies involving upright stance perturbations (Komisar et al., 2018), increasing rail height in this study mostly resulted in significant reductions to trunk roll and pitch displacements and velocities when participants recovered balance while walking. Concomitantly, peak handrail forces decreased marginally with handrail height. Together, these findings can be explained by extending the inverted pendulum model of balance to include handrail forces (Maki & Fernie, 1983), which predicts that higher handrails effectively

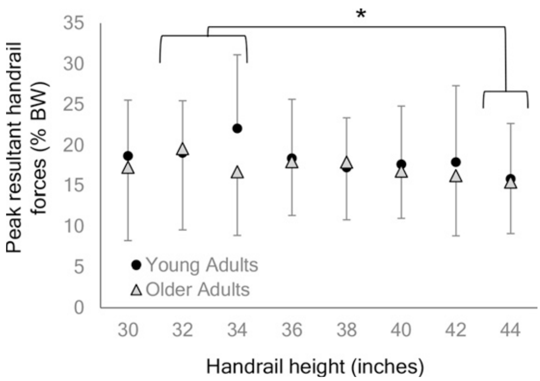

Figure 3. Peak handrail forces, normalized to %BW. Bars represent mean values for each population and condition; error bars represent 1 standard deviation. * indicates significant difference between peak handrail forces for the handrail at 32 and 34 inches versus 44 inches.

increase the lever arm between the handrail and the person's ankles or the floor, thereby enabling greater stabilizing moments with higher handrails at a given applied force. The moment advantage afforded by higher rails (within the range tested) likely contributed to controlling the rotation of the torso following balance loss.

We also found that as handrail height increased, the overall mean peak shoulder elevation angle during the reach-to-grasp reaction generally increased. Our findings are consistent with previously reported increases to shoulder elevation during volitional reaching for targets mounted at three different heights (Vandenberghe et al., 2010). However, significant differences in peak shoulder elevation angle in the present study were only observed between a limited set of handrail heights due to the high variability in shoulder elevation angles at each handrail height (consistent with the large variability in shoulder elevation reported in reach-to-grasp reactions from balance disturbances during stair descent; Gosine et al., 2019). Notably, when comparing to 34-inch-high handrails (the lower boundary of many building codes and accessibility standards in North America; 2010 ADA Standards for Accessible Design, 2010; 2015 International Building Code: Chapter 10 Means of Egress, 2016; Ontario Regulation 332/12: Building

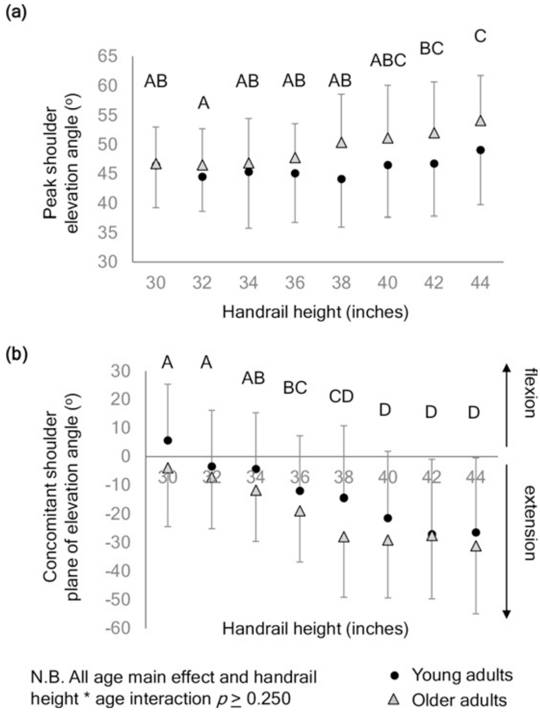

Figure 4. Peak shoulder elevation angles and concomitant plane of elevation angles during balance recovery. (a) Peak shoulder elevation angles between perturbation onset and balance recovery with the handrail. (b) Shoulder plane of elevation angles measured when peak shoulder elevation angles occurred. Mean values for each population and condition are shown; error bars represent 1 standard deviation. Letters indicate post hoc pairwise comparisons: comparisons that did not differ significantly are indicated by the same letter.

Code, 2016), only the highest-tested handrail (44 inches) resulted in significant increases to mean peak shoulder elevation angle during balance recovery; rails between 34 inches and 42 inches did not differ significantly in this metric. This suggests that the balance recovery advantages with higher handrails are not compromised by the physical demands of reaching for these rails, up to a height of 42 inches.

Surprisingly, older adults in this study generally demonstrated trunk angular displacements and peak velocities that were comparable to or less than those of younger adults during balance recovery. Furthermore, our results did not provide evidence that older adults were more reliant on the handrail for balance recovery (indicated by comparable applied handrail forces in younger and older adults), nor did gait speed differ between groups. One possible explanation for our kinematic findings is that the older participants may have experienced increased hip joint stiffness (previously reported in healthy older adults; Anderson & Madigan, 2014), which would contribute to reductions in trunk angular displacements and velocities during balance recovery. In support of this notion, Kerrigan et al. (1998) reported reduced pelvic sagittal motion in older adults during gait (independent of gait speed), while Gill et al. (2001) reported reduced trunk angular sway in older adults during gait—particularly when walking with the eyes closed and during stair gait. Our findings are also in line with age-related changes in balance control strategies during gait, including increased reliance on the hip joints for torque generation in older adults (Silder et al., 2008)—which could plausibly contribute to torso control during balance recovery.

The reductions to trunk angular displacements and velocities with increased handrail height in the present study are consistent with those observed during upright-stance perturbations in young adults (Komisar et al., 2018); however, there were differences in the magnitudes of trunk kinematics. While trunk roll kinematics were similar in both studies (averages at rail heights of 34, 38, and 42 inches tested previously within 0.9° to 2.6° for displacement, and within 1.2°/s to 3.3°/s for velocity), trunk pitch kinematics were much lower in the present study (mean displacements were 33.5% to 39.2%, and mean velocities were 57.4% to 59.3%, of those in Komisar et al. (2018)), even though the perturbation accelerations in the present study were higher in this study (3.75 m/s^2 and 3.5 m/s^2 respectively). The reduced trunk angular displacements and velocities in the present study may stem from how participants could execute stepping reactions, which slow the forward falling motion of the body (Maki & McIlroy, 1999) and supplement the balance recovery contribution of the handrail.

The peak resultant handrail forces in this study were about two-thirds of those reported for upright stance perturbations with young adults (Komisar, Nirmalanathan, et al., 2019)—approximately 18% BW in this study for younger adults, versus 27% BW for forward falling motions reported previously. This discrepancy may result from the combination of the greater perturbation magnitudes for the handrail forces reported in Komisar, Nirmalanathan, et al. (2019) (mean platform acceleration of 4.1 m/s^2 for forward falling motions), and the ability of participants in the present study to execute reactive steps—both of which would reduce the forward falling momentum of the body (Maki & McIlroy, 1999) and the concomitant forces applied to the handrail to regain balance. A further unexpected finding was the absence of a notable effect of handrail height on applied handrail forces, which differs from both research involving voluntary applied forces (Maki et al., 1984) and perturbations of upright stance without reactive stepping (Komisar et al., 2018). This result may be explained by the greater variability both in applied handrail forces and in postural responses to balance loss in the present ongoing gait study, where stepping was not prohibited in the instructions.

Limitations

We acknowledge the study limitations. First, participants walked a fixed lateral distance (60 cm) away from the handrail leading into perturbation onset, and shoulder postures during balance recovery and concomitant applied handrail forces are expected to differ with other lateral distances (Maki et al., 1998). Second, this study incorporated one perturbation magnitude and direction (backward platform translations). However, perturbation design affects postural response (Brown et al., 2001), and further research is needed to understand if previously identified effects of perturbation direction on torso control strategies and applied handrail forces during balance recovery in upright stance conditions are also observed during gait (Gosine et al., 2021b; Komisar et al., 2018). Third, while we reduced the predictability of the balance perturbations by varying the timing, participants still expected balance disturbances (i.e., they were told that perturbations would occur). Individuals adapt their gait patterns when anticipating destabilizing situations (e.g., slippery floors, unlocked rollers; Cham & Redfern, 2002; Marigold & Patla, 2002), and the low gait speeds in both younger and older adults in this study (averaging about .8 m/s, compared to self-selected speeds of 1.0–1.4 m/s in the literature; Brach et al., 2001; Kaufman et al., 2016) suggests that participants may have adopted more conservative balance control strategies leading into perturbations. The absence of differences in young versus older adults in gait speed may reflect the healthy status of the participating older adults (who all completed a long perturbation protocol without perceived exertion surpassing “somewhat hard” on the Borg scale), along with both cohorts adopting a conservative balance control strategy during the study. Hence, gait speed findings should be interpreted with caution. Nevertheless, it seems unlikely that the low gait speed altered our assessment of the relative performance of different handrail heights for balance recovery.

We conducted this study on level ground. While this approach allowed us to safely test several handrail heights with older adults, the mechanics of balance loss and recovery differ on stairs (Gosine et al., 2019; Gosine et al., 2021a), and further research is required to characterize the effect of handrail height on balance recovery in that environment. We were also unable to characterize outcomes related to compensatory stepping despite its importance as a complimentary strategy to reactive grasping. Future studies should investigate the interacting effects of age, compensatory stepping, and compensatory grasping with different handrail configurations. Furthermore, participants were instructed to reach to grasp the handrail as quickly as possible (as opposed to simply responding naturally to perturbations), as young adults do not always reach to grasp handrails when reacting naturally to perturbations without an explicit instruction to grasp (Borrelli et al., 2020; Gosine et al., 2019; Gosine et al., 2021a; King et al., 2009; King et al., 2011;). While these instructions were necessary for our evaluation of handrail height, they also primed participants to look for the handrail during testing, which may have improved participants’ formation of a visuospatial map and performance when grasping. Nevertheless, it seems unlikely that the instruction to grasp affected our assessment of handrail height on balance recovery. Finally, our sample included healthy adults only, due to the challenging nature of the protocol. Other cohorts (e.g., children; frail older adults [Komisar et al., 2020]; individuals with pathologies that limit shoulder range of motion, such as stroke [Mouawad et al., 2011]) may use different handrail grasping strategies for balance recovery, and further research is needed to understand the influence of handrail height on balance recovery in these populations.

CONCLUSION

We characterized the effect of handrail height and age on trunk and shoulder kinematics following perturbation-evoked grasping reactions during level-ground walking. Increased handrail height was generally associated with reduced trunk angular displacements and velocities (up to 42 inches), indicating greater ability to remain upright. While peak shoulder elevation angles mostly increased with handrail height, the differences were nonsignificant for handrails between 34 inches and 42 inches high. Peak resultant handrail forces decreased marginally as handrail height increased. Older adults generally demonstrated comparable or reduced trunk angular displacements and velocities, potentially indicating increased hip stiffness during balance recovery. Our findings point to stability advantages with increased handrail height (up to 42 inches) in both younger and older adults, without requiring significant increases to shoulder elevation to reach the higher rails.

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KEY POINTS

- Younger and older adults experienced platform perturbations while walking beside a handrail; handrail height was varied from 30 to 44 inches (76 to 112 cm).
- As handrail height increased, trunk angular displacements and velocities generally decreased (up to 42 inches/ 107 cm) for both populations; slight increases to peak shoulder elevation angle and decreases to applied handrail forces were also observed.
- Older adults demonstrated comparable or reduced trunk angular kinematics to young adults.
- Increasing handrail height up to 42 inches (107 cm) may help with trunk stability during balance recovery.

ORCID iD

Vicki Komisar https://orcid.org/0000-0003-3504-1796

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Vicki Komisar is an assistant professor of mechanical engineering with the School of Engineering at the University of British Columbia, Canada. She completed her PhD in biomedical engineering in 2018 at the University of Toronto and Toronto Rehabilitation Institute – University Health Network, Canada, followed by a Postdoctoral Fellowship from 2018-2020 in biomedical physiology and kinesiology at Simon Fraser University, Canada.

Alison C. Novak is a scientist at Toronto Rehabilitation Institute – University Health Network, and an assistant professor with both the Department of Occupational Science and Occupational Therapy, and the Faculty of Kinesiology and Physical Education at the University of Toronto, Canada. She completed her PhD in rehabilitation sciences in 2011 at Queen’s University, Canada.

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