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Effects of Self-Selected Handrail Use on Foot Placement and Centre-of-Mass Control During Unperturbed Stair Descent in Older Adults

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Abstract

Introduction: Stair descent is a major cause of falls in older adults, accounting for significant morbidity, mortality, and healthcare costs. Handrails are widely recommended for enhancing stair safety, mainly because they can be used to arrest a fall if stability is perturbed. However, little is known about the biomechanical impact of handrails when used with the intention to avoid postural instabilities, before the need arises to arrest a potential fall. This study explores whether using the handrails in a self-preferred way during stair descent affects biomechanical parameters associated with fall risk in older adults. Method: Fourteen older adults (age: 72.6 ± 6.2 years) descended a seven-step staircase in two conditions: self-selected handrail use and no handrail use. Motion capture data were used to determine the proportion of foot length in contact with the step (PFLCS), as well as peak Centre of Mass (CoM) acceleration and velocity during the landing and lowering phases and variability. Paired- samples *t*-tests or Wilcoxon signed-rank tests were applied for statistical comparisons, as appropriate. Results: PFLCS or CoM parameters during landing or lowering. showed no significant differences. Handrail use was associated with a significant increase in CoM peak acceleration variability during the lowering phase (p = 0.012, Cohen's d = 0.87). Conclusion: The similarity in PFLCS between conditions indicates that, in the present study, allowing participants to use the handrail in a self-selected manner did not produce a measurable improvement in foot placement on the step. Greater variability in CoM peak acceleration during the lowering phase suggests less consistent movement patterns when using the handrails. This increased variability may be linked with perceived assurance that a fall could be arrested if a destabilising event occurred, although the mechanism underlying this finding remains unclear. Given the laboratory conditions, the task may not have been sufficiently challenging for healthy older adults with low concern about falling; therefore, using the handrails in their preferred way did not have a measurable effect in terms of safer stepping behaviours. Future research should assess handrail use under more demanding, real-world conditions.

Keywords: handrail use; stair descent; older adults; foot placement; centre of mass; postural control; movement variability; stair safety; biomechanics; fall prevention

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1. Introduction

Falls represent a major cause of injury and mortality among older adults and carry considerable social and economic costs for individuals, families, and healthcare systems [1,2]. Notably, falls on stairs are estimated to account for around 20% of all falls and represent the leading cause of accidental death among older adults [3,4]. Stair descent presents greater hazards than stair ascent [4–6], as it requires substantial demands on lower-limb strength [7], coordination [8], balance control [9], and visual feedback [10,11]. Furthermore, falls during stair descent often lead to more severe injuries, as individuals have limited opportunities to arrest their fall before descending the subsequent steps of the staircase. Research suggests that stair falls during descent result mainly from over-stepping, where the leading foot placed one step below rolls off the step edge, potentially resulting in a forward stumble or a backwards slip [9,12–14]. Alternatively, under-stepping can arise from the heel of the leading foot striking the riser, causing a forward stumble. In addition, increased Centre of Mass (CoM) acceleration and velocity can result in loss of balance and a consequent forward fall [15,16]. Inconsistent or poorly timed CoM control can also make foot placement less precise, can increase impact forces upon landing, and can reduce the ability to recover from small perturbations, each increasing fall risk [15,17,18].

Given the potential hazards associated with stair descent in older adults, identifying strategies that may improve safety is important. The use of handrails is one strategy, as they are an environmental feature specifically intended to support safer stair negotiation [19–23]. By providing mechanical support and redistributing some of the load, handrails can reduce the demands placed on the lower limbs [20,23]. Moreover, handrails may also provide additional guidance when visibility is limited, such as under poor lighting conditions, while enhancing perceived stability and providing reassurance during stair negotiation, particularly among individuals who are concerned about falling [20,21,23,24]. More directly, handrails have been shown to be instrumental for arresting a fall by relying on them following sudden perturbations of postural stability [22].

To date, relatively few studies have examined how handrail use influences stepping behaviour and dynamic control during unperturbed stair descent [19], for instance, reported that handrail use reduced CoM–CoP separation, suggesting improved postural control during descent. This support may allow individuals to transfer part of the stabilising demand from the lower limbs to the handrail [20]. similarly observed that older adults with a concern about falling ($n = 3$) demonstrated lower centre-of-pressure velocity (VCOP) during stair descent when using a handrail, suggesting more controlled postural adjustments. VCOP reflects the neuromuscular response involved in shifting the body's CoM and can be used as a stability indicator [20]. found no significant differences in either cadence or VCOP between self-selected handrail use and stair descent without handrails.

Across these studies, handrail use was standardised, with participants instructed to glide their non dominant hand along the rail without actively gripping or pulling [19,20,22]. Although this standardised approach improves consistency when measuring task performance, it may reduce the extent to which these findings reflect how handrails are typically used in real-world settings. This method could constrain natural behaviour and may not reflect the variability in how individuals typically interact with handrails (e.g., gripping, using one or both hands, or intermittent contact) in everyday settings. Thus, it remains unclear whether using the handrails in one's preferred way improves biomechanical risk factors of stair safety in older adults. Therefore, the purpose of the present study was to explore whether handrails modify certain stair fall risk factors during descent when participants are allowed to use them in their preferred way to better reflect real world handrail use. We hypothesised that handrail use would improve foot positioning on the step, measured as the proportion of foot length in contact with the stair (PFLCS),

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designated as the primary outcome. CoM control parameters were treated as secondary, exploratory outcomes to further characterise dynamic stability during handrail use.

2. Methods

2.1. Participants

Fourteen older adults (age: 72.6 ± 6.2 years; stature: 163.3 ± 7.1 cm; body mass: 69.3 ± 12.6 kg) participated in the study. Nine participants were female (5 males). All participants lived independently, in the UK, and could walk without aids apart from the handrail to negotiate stairs. Participants were recruited from a previous study [25]. The study obtained ethics approval from the Liverpool John Moores University (LJMU) Ethics Committee (Ref No: 23/SPS/017), and consent was acquired prior data collection.

2.2. Procedure

All participants were attached to the overhead belay safety system through a five-point safety harness that they wore. During each trial, a trained member of the research team operated the belay system while ensuring that the rope remained free of tension. Measurements were collected using a custom-built staircase equipped with three individual force platforms (1080 Hz, 9260AA, Kistler AG, Winterthur, Switzerland) embedded within the lower three steps. Kinematics were captured using a 26-infrared-camera system (120 Hz, Vicon, Oxford Metrics, Oxford, UK). The staircase dimensions were designed in accordance with UK building regulations [26–28], with the rise set at 20 cm and the going at 25 cm. Handrails were installed on both sides of the staircase at a height of 900 mm and remained in place throughout testing (Figure 1A). Participants descended the staircase at a self-selected pace using a step-over-step pattern, alternating the leading limb on successive steps. To allow for familiarisation, participants performed up to five practice trials with no specific handrail use. Afterwards, participants completed two trial conditions in a fixed and instructed order: (1) five trials using the handrail(s) in their preferred way (for example, using one handrail or both handrails, grabbing, gliding, etc.), and (2) five trials without using a handrail. Participants' preferred handrail use during trials was reported, and their habitual handrail use at home was observed and reported from a previous study [25]. The fixed order was selected to prioritise participant safety and comfort while descending an unfamiliar experimental staircase, allowing participants to initially complete the task with access to their preferred source of support before undertaking the no-handrail condition.

2.3. Biomechanical Measurements

A full-body six-degree-of-freedom model comprising 15 segments (head, thorax, pelvis, upper arms, lower arms, hands, thighs, shanks, and feet) was used to obtain whole-body kinematics from 74 reflective markers (diameter of 14 mm). Whole-body marker coordinates were established from a static anatomical calibration trial. A digitising wand (C-Motion, Germantown, MD, USA) was used during a separate trial to define virtual landmarks at the toe and heel tips of participants' footwear [29]. Landmarks for the toe and heel tips were at the anterior–inferior (toe tip) and posterior–inferior (heel tip) point of each shoe. A triangular cluster of markers placed over the forefoot portion of the shoe was used to track the virtual landmarks during movement trials. The toe-tip and heel-tip landmarks were positioned at the anterior–inferior and posterior–inferior points of each shoe, respectively. During the movement trials, these virtual landmarks were tracked using a triangular cluster of markers positioned over the forefoot of the shoe.

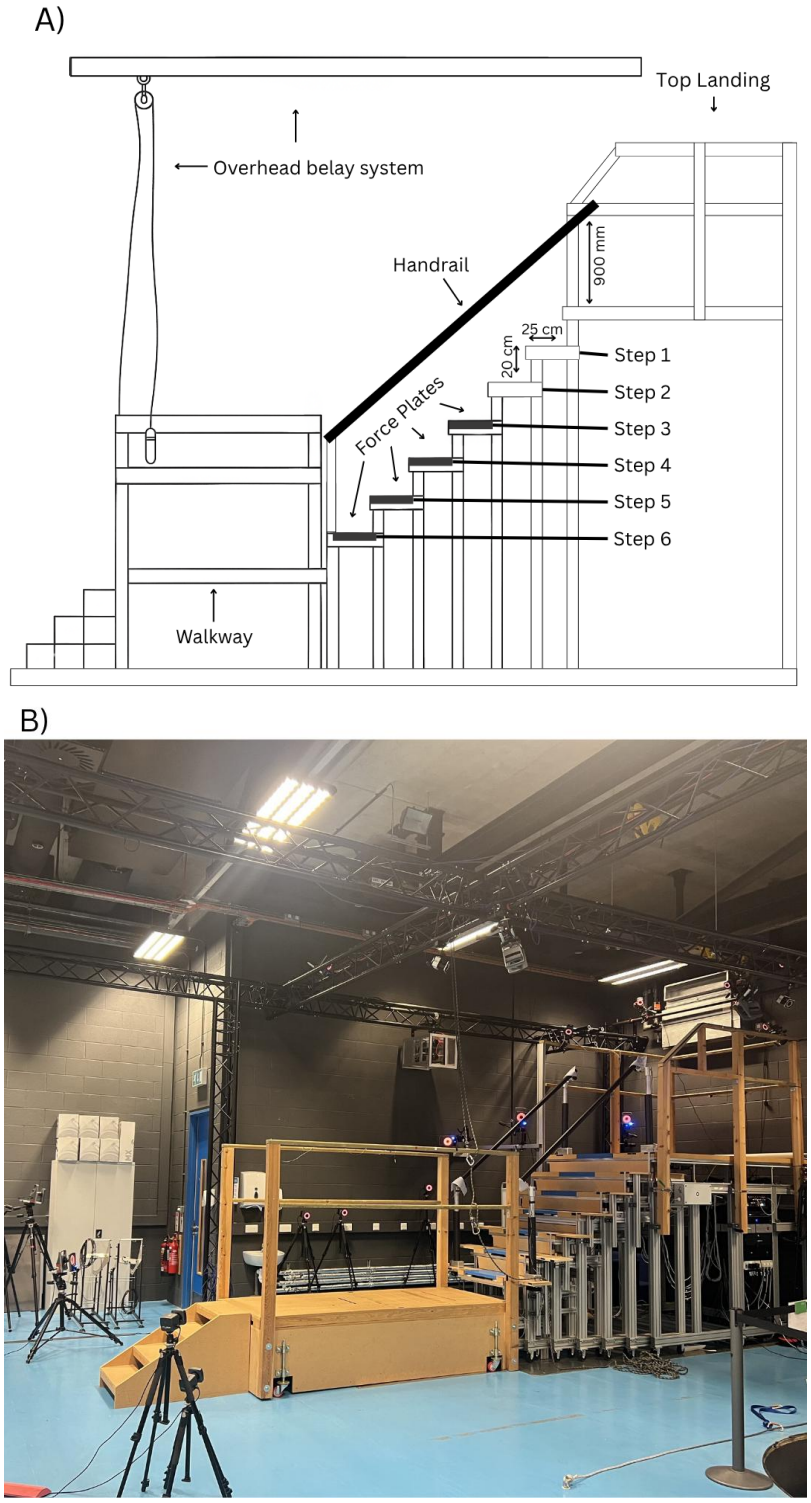


Figure 1. A schematic representation (**A**) and image (**B**) of the custom-built instrumented seven-step staircase.

Vicon (Vicon Nexus 2.6, Oxford Metrics, Oxford, UK) was used to label and gap fill all marker data, and Visual 3D x64 (C-Motion, Germantown, MD, USA) was used to analyse data. To filter all kinematic data. a low-pass fourth-order Butterworth filter was used with a cut-off frequency of 6 Hz. Force-plate data was used to identify gait events, with an ascending vertical ground reaction force threshold of >20 N used to identify touchdown, while a descending threshold of <20 N defined foot-off. Kinematic data were subsequently analysed to determine the following outcome measures:

1. Proportion of foot length in contact with stair (PFLCS). PFLCS was used to quantify anterior–posterior foot placement relative to the edge of the step, providing an indication of overstep/slip-induced fall risk. Lower PFLCS values indicate a smaller proportion of the foot positioned on the step. The anterior–posterior distance between the virtual toe-tip landmark and the step edge was determined at the point of peak vertical ground reaction force during foot contact. This time point was consistent across trials and represented maximal loading of the step. PFLCS was subsequently expressed as the percentage of foot length in contact with the tread using the following equation:

$$\text{PFLCS (\%)} = \frac{Foot\ Length\ (AP) - Toe\ to\ Step\ Edge\ Distance}{Foot\ Length(AP)} \times 100$$

where foot length (AP) represents the horizontal distance between the virtual heel-tip and toe-tip landmarks. PFLCS was determined for steps 2 to 4 during descent (Figure 1A) and averaged separately across the five handrail and five no-handrail trials for each participant and condition.

2. Cadence. The duration of a gait cycle was determined for each limb, and cadence was calculated from the mean duration across the left and right limbs. This value was then averaged across the three trials to obtain a single cadence measure for subsequent analysis.
3. CoM Parameters. CoM parameters were used to characterise dynamic control during stair descent, including the control of body movement during lowering and contact with each step. Greater downward CoM acceleration and velocity during lowering, together with reduced CoM acceleration at landing, have been associated with increased risk. Whole-body CoM was calculated in Visual3D as the weighted average position of the 15 body segments: head, thorax, pelvis, upper arms, lower arms, hands, thighs, shanks, and feet. CoM parameters were determined for steps 2 to 4 for each participant and averaged across the five trials in each condition (handrail and no-handrail) for subsequent analysis. The CoM parameters were split into the following phases:

   Landing Phase: The period from touch-down with the step below until the CoM's downward velocity reached zero (indicating the end of its upward acceleration) was used to define the landing phase. This phase shows weight acceptance and forward motion [18,30].

   Lowering Phase: This phase was defined as the period beginning at the completion of landing and ending when the opposite limb contacted the next step.

   3.1 CoM peak Acceleration-Land. This measure represents the peak positive acceleration of the CoM in the downward direction during the landing phase.
   3.2 CoM peak Velocity-Lower. This measure represents the peak negative velocity of the CoM in the downward direction during the lowering phase.
   3.3 CoM peak Acceleration-Lower. This measure represents the peak negative acceleration of the CoM in the downward direction during the lowering phase.
4. Within-step and between-trial variation. Variation was calculated to assess the ability to maintain a steady/safe movement pattern. A higher variation within steps and between trials may increase the risk of a fall [31]. For each participant and each experimental condition, the standard deviation (SD) of each parameter was calculated across the five trials and three analysed steps (steps 2, 3, and 4), resulting in up to 15 observations per participant per condition. These observations were pooled to provide a single SD value for each participant and condition. Thus, the resulting measure represents within participant variability across repeated steps and trials and was not treated as 15 independent observations. The participant-level SD values were

subsequently entered into the paired statistical analysis, with the 12 participants who completed both experimental conditions constituting the statistical unit of analysis.

2.4. Questionnaires and Functional Movement Tests

To expand on the findings of Reid et al. [20] that dynamic stability improved with handrail use during stair descent only for a small number of participants with concern of falling (n = 3), in the present study, the following questionnaires were completed after the staircase trials: the Activities-specific Balance Confidence Scale (ABC) to measure self-perceived balance confidence during daily activities [32], the short Falls Efficacy Scale-International (Short-FES-I) to measure concern about falling during activities of daily living [33], the Montreal Cognitive Assessment (MoCA) to measure cognitive functioning [34], the Nottingham extended Activities of Daily Living (NEADL) index to measure functional status [35] and the FRAT to assess fall risk [36].

Afterwards, the Fullerton Functional Fitness Tests [37] were completed to measure functional movement measures: (1) chair stand to obtain functional lower body strength, (2) Arm curl to obtain functional upper body strength, (3) 2 min step test to obtain physical stamina, (4) chair sit and reach to obtain lower body flexibility, (5) back scratch test to obtain upper body flexibility and (6) timed up and go to obtain speed, agility and balance [37].

2.5. Data Analysis and Statistics

A frequency analysis was used to compare the questionnaires and functional fitness scores. A paired-samples t-test was conducted to compare the effect of handrail condition (self-selected handrails vs. no handrail usage) on the biomechanical outcome measures. For each biomechanical outcome, the average of all steps was used for the comparisons. Visual inspection of residual plots was performed to assess the normal distribution of data. Where data were not normally distributed, a Wilcoxon signed rank test was used. CoM peak velocity-lowering variability was the only measured outcome to use Wilcoxon signed rank. Effect sizes (Cohen's d) were calculated as the mean difference divided by the standard deviation and interpreted as small (0.2), medium (0.5), or large (0.8) [38].

3. Results

During the staircase trials, one participant was excluded from the analysis due to marker errors. A further participant was unable to complete the descent without using the handrail and therefore completed only the handrail trials. As a result, the data from twelve participants were included in the descriptive reporting and statistical analysis.

Handrail use during trial and habitual use at home are summarised in Table 1. Participants demonstrated varied handrail use patterns during descent trials. Handrail condition had no significant effect on most parameters (Table 2). Handrail condition had a significant effect on CoM peak acceleration-lowering variability ($t(11) = 3.027$, $p = 0.012$, Cohen's d = 0.874) (Figure 2H). The self-selected handrail condition (0.57 ± 0.20 m/s^2) was significantly greater compared to the no-handrail condition (0.45 ± 0.16 m/s^2), with a large effect size (Cohen's d = 0.874) and 95% confidence intervals ranging from 0.189 to 1.531 m/s^2.

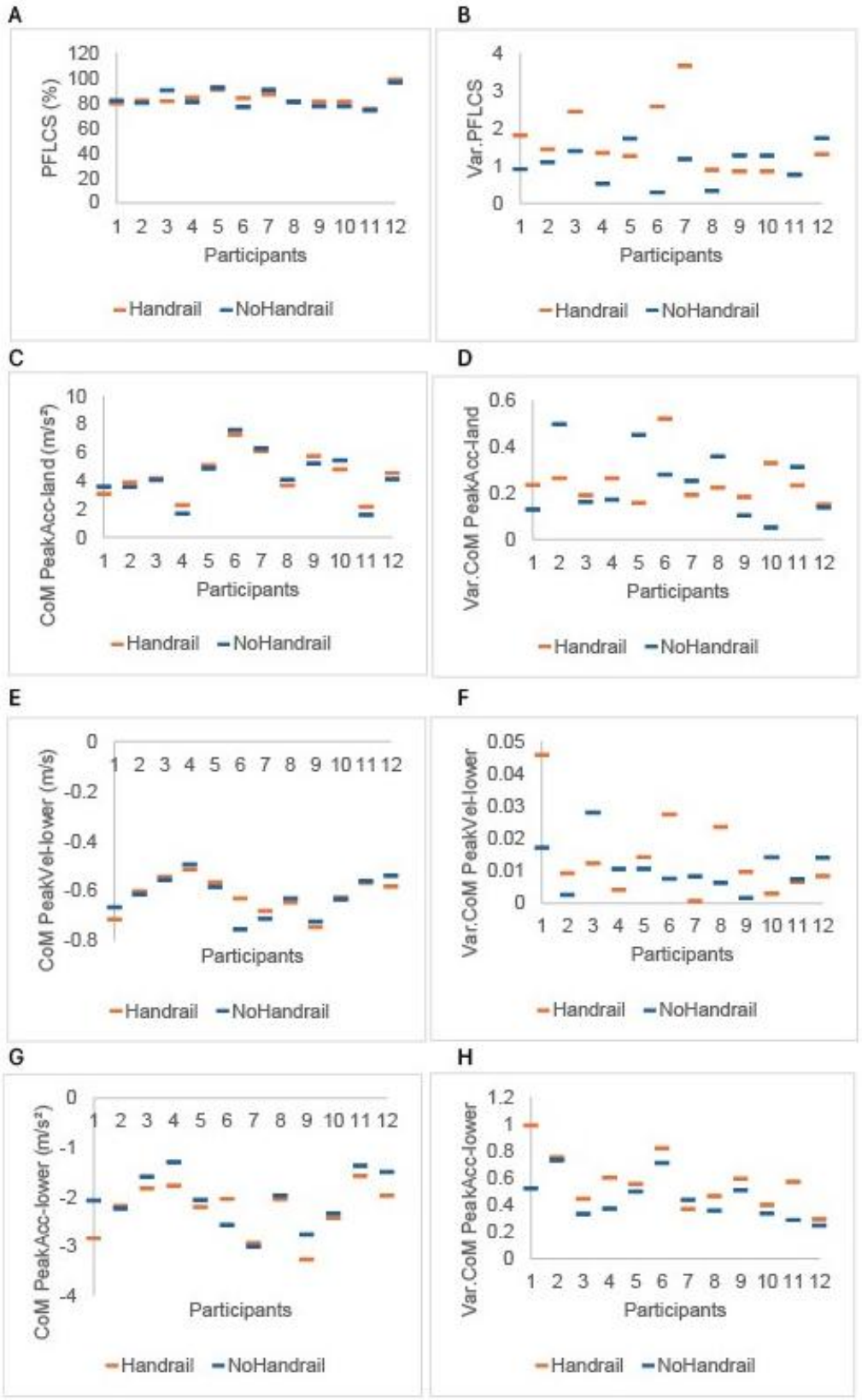


Figure 2. Scatter graphs representing biomechanical data for self-selected handrail vs. no handrail use during stair descent. (**A**) PFLCS, proportion of foot length in contact with step; (**B**) PFLCS variation; (**C**) CoM peak acceleration-land; (**D**) CoM peak acceleration-land variation; (**E**) CoM peak velocity-lower; (**F**) CoM peak velocity-lower variation; (**G**) CoM peak acceleration-lower; (**H**) CoM peak acceleration-lower variation.

Table 1. Participant characteristics: handrail use during the handrail trial condition, habitual handrail use at home, questionnaire, and functional fitness scores.

	Number of Participants (*n* = 12)
Handrail Use During the Handrail Trial Condition	
Used both handrails	10
Used only right handrail	2
Habitual Handrail Use at Home	
Use handrails at home	10
Does not use handrails at home	2
Questionnaires	
Short FES-I	8.43 (1.87)
NEADL	20.86 (1.56)
ABC (%)	93 (0.08)
MOCA	26.64 (3.13)
FRAT	0.93 (1.44)
Functional Fitness	
Chair Stand (per min)	13.29 (4.29)
Arm Curl (per min)	21.57 (10.23)
2-min Step (step/min)	93.79 (15.34)
Chair Sit and Reach (cm)	8.14 (16.24)
Back Scratch (cm)	6 (11.72)
Time Up and Go (s)	7.99 (1.92)

All questionnaires and functional fitness values are presented as mean (SD). Short FES-I, The Short Falls Efficacy Scale—International; scores <11 generally indicate low concern of falling [39]; NEADL, The Nottingham extended Activities of Daily Living; [35]; ABC, The Activities Specific Balance Scale (scores above 80% considered high confidence) [32]; MOCA, The Montreal Cognitive Assessment (scores ≥26 are typically used as a threshold for normal cognitive function) [34]; FRAT, The Falls Risk Assessment Tool (scores range from 0 to 5, with higher scores indicating greater risk) [36]; Functional Fitness, Fullerton Function Fitness Test [37].

Table 2. Mean (SD) PFLCS, CoM peak acceleration-land, CoM peak velocity-lower, CoM peak acceleration-lower, and within-step and between-trial variance for each parameter during stair descent for older adults.

	Number of Participants (*n* = 12)	
	Self-Selected Handrails	**No Handrails**
PFLCS %	83.54 (1.55)	83.28 (7.23)
Var. PFLCS %	4.26 (0.30)	4.17 (1.38)
Cadence (steps/min)	22.6 (6.02)	21.33 (3.72)
Var. Cadence	2.90 (2.60)	8.34 (11.73)
CoM peakAcc-land (m/s^2)	4.55 (1.55)	4.36 (1.73)
Var. CoM peakAcc-land	0.87 (0.30)	0.74 (0.25)
CoM peakVel-lower (m/s)	−0.61 (0.07)	−0.62 (0.08)
Var. CoM peakVel-lower	0.04 (0.02)	0.04 (0.01)
CoM peakAcc-lower (m/s^2)	−2.33 (0.56)	−2.06 (0.55)
Var. CoM peakAcc-lower	0.61 (0.24) [a]	0.44 (0.16) [a]

PFLCS, Proportion of foot length in contact with step; Var., within-step and between-trial variation; CoM, Centre of Mass. [a] Refers to significant differences between handrail conditions (self-selected handrail vs. no handrail).

No significant difference in PFLCS between the handrail and no-handrail conditions ($t(11) = 0.408$, $p = 0.691$). Similarly, PFLCS variability did not differ significantly between conditions ($t(11) = 0.569$, $p = 0.581$). No significant difference was observed in CoM peak acceleration-landing between the handrail and no-handrail conditions ($t(11) = 0.410$, $p = 0.690$). Similarly, CoM peak acceleration-landing variability did not differ significantly between conditions ($t(11) = 0.866$, $p = 0.405$). No significant difference in CoM peak velocity lowering was observed between the handrail and no-handrail conditions ($t(11) = 0.199$, $p = 0.846$). For CoM peak velocity-lowering variability, a Wilcoxon signed-rank test showed no significant difference between conditions ($Z = -1.451$, $p = 0.147$). A paired-samples *t*-test showed no significant difference in CoM peak acceleration lowering between the handrail and no-handrail conditions ($t(11) = -1.945$, $p = 0.078$).

Questionnaire and Functional Fitness Characteristics

Questionnaire scores and Functional Fitness Scores are summarised in Table 1. The short FES-I score showed low concern about falling, and the ABC score indicated high functional independence, normal cognitive function, and minimal fall risk. NEADL, MOCA, and FRAT results further confirmed that the group was healthy, high-functioning, and independent.

4. Discussion

This study explored whether participants using the handrails in their preferred way during unperturbed stair descent affected foot placement (primary outcome) and CoM control (secondary, exploratory outcomes) in older adults. Despite the expected stability benefits of handrail use [19–23], handrail use did not significantly affect the primary outcome, PFLCS. This finding suggests that in healthy, high-functioning older adults with low concern about falling, using handrails in a self-selected manner may not result in a measurable safety benefit in terms of foot positioning on the step.

Among the secondary, exploratory outcomes, only variability in CoM peak acceleration during the lowering phase significantly increased during the self-selected handrail trials. This finding suggests that our participants adopted less consistent movement patterns when using the handrail, possibly due to perceived assurance that the handrail could arrest a fall should their movement be perturbed. As an exploratory finding, this result should be interpreted with caution and considered hypothesis-generating rather than confirmatory. Greater variability may reflect less consistent control of downward movement, which could influence the ability to regulate stability across steps [15,17,31,40]. However, the present study did not determine whether increased CoM variability directly influenced foot trajectory or foot placement. Previous research has suggested that greater variability may contribute to less accurate foot trajectories between steps, potentially affecting subsequent foot placement, or may reduce the ability to effectively decelerate the body when contacting the step below [15,17,18]. Unlike previous studies that reported stability benefits with handrail use, such as reduced CoP-CoM separation [19] or lower centre-of-pressure velocity (VCOP) in older adults with concern of falling [20], we did not find handrail use to enhance the stepping behaviours or stability parameters, although the mechanism underlying this finding remains unclear. Several factors may explain this discrepancy. First, our participants used handrails in their own preferred manner rather than a prescribed glide, potentially leading to inconsistent mechanical support. While this approach improves ecological validity and better reflects real-world behaviour, differences in grip type, contact duration, and the degree of force or mechanical support applied may have resulted in participants receiving different levels of assistance from the handrails. Consequently, the handrails may not have provided consistent mechanical support across participants, potentially reducing our ability to identify the effects of any specific handrail

use strategy and increasing variability in movement patterns. If the handrail was not consistently gripped, glided, or actively used for support, it may have provided limited benefit [41]. Descriptive subgroup comparisons between participants who glided their hands along the handrail (n = 10) and those who did not (n = 2) indicated similarity in biomechanical parameters, but a systematic study with larger sizes would be required to explore the role of interacting with the handrail in a specified way. Second, the sample was high-functioning, with above average balance confidence for their age, low concern of falling, and normal cognitive scores, which may have reduced reliance on handrails for foot placement and stability. Third, the stair task may not have been sufficiently challenging to reveal safety benefits. Several factors, including stair dimensions, lighting, and environmental complexity, can influence the degree of challenge for stair descent [7,9].

While no difference in foot positioning and greater variability was observed with self-selected handrail use, these findings should be considered within the broader context of handrails as a safety feature during stair negotiation. Handrails' potential benefits may be more apparent during unexpected perturbations, where handrails can assist with recovery from sudden instability [22]. Such conditions were not examined in the present study. They also provide tactile guidance in low visibility, assist those with mobility issues, and increase perceived stability, particularly in individuals with a concern of falling [20,21,23]. Therefore, external variables such as lighting, stair surface, and environmental distractions (e.g., objects on the stairs) could influence the extent to which handrails are relied upon [42–44]. Individual factors, such as reduced attention due to dual tasking [8,45], the presence of other people on the stairs [46,47], or visual aspects [29,48] could also alter handrail use in real-world settings. Simulating these external conditions in a controlled lab setting is technically challenging.

An a priori power analysis was not conducted due to the lack of sufficiently comparable data from which to derive a reliable, expected effect size; therefore, the findings should be interpreted cautiously given the lack of power analysis. It should also be noted that a limitation of the study is that the measurements were conducted on an experimental staircase within a laboratory environment using a safety harness. While the harness was necessary for participant safety, it may have influenced participants' psychological state and stair negotiation strategies compared to navigating stairs in their own homes. However, participants were given time to familiarise themselves with the staircase, and all reported feeling confident in negotiating it. Another limitation of this study is that handrail use was not standardised. Variations in how participants interacted with the handrails (e.g., grip type, force application, or contact duration) may have influenced the observed outcomes. While this variability reflects real-world behaviour, it limits the ability to isolate the effects of specific handrail use strategies. Although participants' handrail use and preferred strategies were observed, handrail forces were not measured. Consequently, the magnitude and duration of forces applied to the handrail could not be quantified, limiting our ability to determine the mechanical contribution of handrail use to the observed biomechanical responses. Future studies incorporating handrail force measurement could provide greater insight into how the magnitude and timing of handrail support influence balance and movement control during stair descent. Additionally, the participants consisted of healthy older adults with relatively high balance confidence and were not concerned with falling, as indicated by their high ABC and low FES-I scores. This limits the generalisability of the findings to older adults with poorer balance or a concern of falling [17], who might respond differently to handrail use [20]. Another limitation is that the self-selected handrail and no-handrail conditions were completed in a fixed order rather than being randomised or counterbalanced. Although the handrail condition was completed first to prioritise participant safety and comfort when negotiating an unfamiliar staircase, this design may have introduced potential order or familiarisation effects. For

example, participants may have become more accustomed to the staircase and task during the initial self-selected handrail trials, which could have influenced their performance during the subsequent no-handrail condition. Future studies should consider randomising or counterbalancing condition order where appropriate to minimise these potential effects. Lastly, we did not administer a scale or questionnaire immediately after each handrail condition to assess participants' perceived confidence or psychological state during stair descent. Such information could have provided valuable insight into how subjective feelings of safety and stability influenced stair negotiation strategies. Future research should aim to recruit more diverse participants, including older adults with greater fall risk, and incorporate post-trial confidence assessments to better understand the psychological factors influencing stair descent performance.

5. Conclusions

This study investigated whether using the handrails in their preferred way during unperturbed stair descent influences biomechanical foot placement and CoM control in older adults. We hypothesised that handrail use would improve foot positioning on the step and CoM control, but our findings did not support this. Interestingly, self-selected handrail use did not affect the primary outcome (PFLCS). This finding suggests that in healthy, high functioning older adults with low concern about falling, using handrails in a self-selected manner may not result in a measurable safety benefit in terms of foot positioning on the step. Future research should explore the interaction between handrail use, stair complexity, and individual differences in balance confidence to better understand the role of handrails in reducing stair fall risk in unperturbed conditions.

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Data Availability Statement: The data are not publicly available due to ethical restrictions relating to participant data and data sharing.

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