BALANCE AND AGEING

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BALANCE AND AGEING

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Associations Between Factors Across Life and One-Legged Balance Performance in Mid and Later Life: Evidence From a British Birth Cohort Study

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Introduction: Despite its associations with falls, disability, and mortality, balance is an under-recognized and frequently overlooked aspect of aging. Studies investigating associations between factors across life and balance are limited. Understanding the factors related to balance performance could help identify protective factors and appropriate interventions across the life course. This study aimed to: (i) identify socioeconomic, anthropometric, behavioral, health, and cognitive factors that are associated with one-legged balance performance; and (ii) explore how these associations change with age.

Methods: Data came from 3,111 members of the MRC National Survey of Health and Development, a British birth cohort study. Multilevel models examined how one-legged standing balance times (assessed at ages 53, 60–64, and 69) were associated with 15 factors across life: sex, maternal education (4 years), paternal occupation (4 years), own education (26 years), own occupation (53 years), and contemporaneous measures (53, 60–64, 69 years) of height, BMI, physical activity, smoking, diabetes, respiratory symptoms, cardiovascular events, knee pain, depression and verbal memory. Age and sex interactions with each variable were assessed.

Results: Men had 18.8% (95%CI: 13.6, 23.9) longer balance times than women at age 53, although this difference decreased with age (11.8% at age 60–64 and 7.6% at age 69). Disadvantaged socioeconomic position in childhood and adulthood, low educational attainment, less healthy behaviors, poor health status, lower cognition, higher body mass index (BMI), and shorter height were associated with poorer balance at all three ages. For example, at age 53, those from the lowest paternal occupational classes had 29.6% (22.2, 38.8) worse balance than those from the highest classes. Associations of balance with socioeconomic indicators, cognition and physical activity became smaller with age, while associations with knee pain and depression became larger. There were no sex differences in these associations. In a combined model, the majority of factors remained associated with balance.

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Discussion: This study identified numerous risk factors across life that are associated with one-legged balance performance and highlighted diverse patterns of association with age, suggesting that there are opportunities to intervene in early, mid and later life. A multifactorial approach to intervention, at both societal and individual levels, may have more benefit than focusing on a single risk factor.

Keywords: balance, aging, life course, risk factor, epidemiology

INTRODUCTION

From getting out of bed in the morning to sitting, standing and walking throughout the day, balance is a crucial component of everyday life (Muir et al., 2010). Poor balance is linked with several adverse health outcomes, perhaps most notably increased falls risk (Ganz et al., 2007), but also with increased risk of disability, fractures, hospitalization, and premature mortality (Cooper et al., 2011b, 2014; Nofuji et al., 2016; Keevil et al., 2018). Despite the growing awareness of the importance of balance in aging—as reflected in recent physical activity guidelines (Centre for Ageing Better, 2018; US Department of Health Human Services., 2018; Department of Health Social Care., 2019)—the life course epidemiology of balance performance has been under-investigated compared with other measures of physical capability such as grip strength and chair rise performance.

In the few studies that have examined factors across life in relation to balance performance, several associations have been found. Across a range of ages, males tend to have better balance performance than females (Wolfson et al., 1994; Schultz et al., 1997; Cooper et al., 2011a; Kim et al., 2012). Low socioeconomic position (SEP) has been found to have a negative cumulative association with balance performance, with an additive effect of low SEP in childhood and adulthood on risk for poor balance in later life (Birnie et al., 2011b; Strand et al., 2011a). Smoking history (Strand et al., 2011b), low cognitive ability in both childhood and adulthood (Kuh et al., 2009a; Blodgett et al., 2020), higher levels of depression (Nitz et al., 2005), and low levels of physical activity (Cooper et al., 2011c, 2015; Chang et al., 2013), have also been shown to be associated with poor balance.

These previous studies have primarily examined associations between a single risk factor and balance ability at one time point. With the exception of our recent study of the association between childhood cognitive ability and balance performance (Blodgett et al., 2020), to our knowledge, no study has examined whether associations change with age. This is a limitation, given that balance is a complex process that relies on sensory input including visual cues, proprioception, vestibular processes as well as muscular strength and cognitive processing (Merla and Spaulding, 1997), and so may be affected by age-related changes, such as increased levels of morbidity and decline in cognitive functioning. In addition, few studies have investigated sex differences in the associations between risk factors and balance ability. This is despite the fact that investigating sex differences in the relationships between different risk factors and balance may help elucidate why men have better average balance performance than women, as the reasons for this are still not fully understood (Maki et al., 1990; Wolfson et al., 1994; Hageman et al., 1995; Schultz et al., 1997; Bryant et al., 2005).

Using a British birth cohort study, previously used to study factors associated with balance at a single age (Kuh et al., 2006, 2009a; Birnie et al., 2011a; Cooper et al., 2011a,c, 2015; Strand et al., 2011a,b; Mulla et al., 2013; Murray et al., 2013; Blodgett et al., 2020), we aimed to investigate associations of socioeconomic, behavioral, health and cognitive risk factors across life with one-legged balance performance over 16 years and assess if these associations change with age or sex. We hypothesized that positive factors such as high SEP, low BMI, participation in healthy behaviors, absence of poor physical and mental health as well as higher adult cognitive ability would be associated with better balance performance. As physical and mental comorbidities become more common with age, we hypothesized that the associations of health status with balance performance would get stronger with age. Conversely, as health status becomes more important, the relative contributions of SEP were hypothesized to decrease.

METHODS

Sample

The MRC National Survey of Health and Development (NSHD) is an ongoing study of 5,362 individuals born in England, Scotland, or Wales within 1 week in March 1946. Since 1946, study members have been followed up to 24 times in infancy, across childhood, adolescence, and adulthood, most recently at ages 53 (n = 2,988), 60–64 (n = 2,229), and 69 (n = 2,149) using a combination of questionnaires, interviews, and clinical examinations (Kuh et al., 2016). Details of loss to follow-up (e.g., death, emigration, refusal, incapacity) in this sample have been previously described (Blodgett et al., 2020). Ethical approval for the most recent data collection wave (2015) was obtained from Queen Square Research Ethics Committee (14/ LO/1073) and Scotland A Research Ethics Committee (14/SS/1009).

Assessment of Balance Ability

One-legged balance performance was assessed by trained nurses during clinical assessments at ages 53, 60–64, and 69 using standardized protocols. Study members were asked to fold their arms and stand on their preferred leg with their eyes closed for as long as possible up to a maximum of 30 s. If individuals were unable to perform the test, the reason was recorded. In these analyses, individuals who could not perform the test due to health reasons and those who attempted but could not maintain the balance position were given a score of zero. The final analytical sample consisted of individuals with a balance time at one or more ages ($n = 3{,}111$). The one-legged balance test is considered to be a reliable and valid measure of static balance and has been shown to have high inter-rater and test-retest reliability (Giorgetti et al., 1998; Bohannon, 2006; Springer et al., 2007; Michikawa et al., 2009; Choi et al., 2014; Ortega-Pérez de Villar et al., 2018). Many studies have consistently demonstrated associations between poor one-legged balance performance and higher risk of falls, disability, poor gait speed, frailty and premature mortality (Drusini et al., 2002; Michikawa et al., 2009; Cooper et al., 2010, 2014; Delbaere et al., 2010; Oliveira et al., 2018).

Assessment of Risk Factors

We selected a set of risk factors *a priori* that had previously been shown to be associated with balance or other measures of physical capability at a single time point in NSHD and other studies (Kuh et al., 2006, 2009a; Birnie et al., 2011b; Cooper et al., 2011a,c, 2015; Strand et al., 2011a,b; Welmer et al., 2012; D'Andréa Greve et al., 2013; Amemiya et al., 2019; Thomas et al., 2019).

Socioeconomic Indicators

Paternal occupational class (at age 4) and *own occupational class* (reported at age 53 years) were grouped into three categories as distinguished by the Registrar General's Social Classification (Galobardes et al., 2006): (1) I Professional and II Intermediate; (2) III Skilled (non-manual) and III Skilled (manual); and (3) IV Partly skilled and V Unskilled manual. *Maternal education* was classified into four categories: (1) Primary only; (2) Primary and further education; (3) Secondary only; (4) Secondary and further education. Participants reported their highest level of *educational attainment* by age 26, which was categorized as degree or higher, advanced secondary qualifications generally attained at 18 years (GCE A level or Burnham B), ordinary secondary qualifications generally attained at 16 years, (e.g., GCE O level or Burnham C), below ordinary secondary qualifications, or none.

Anthropometric Indicators (Ages 53, 60–64, 69)

Height (m) and *BMI (kg/m^2)*, derived from height and weight measurements ascertained by nurses using standardized protocols, were used (Braddon et al., 1986).

Behavioral Risk Factors (Ages 53, 60–64, 69)

Individuals self-reported their *leisure time physical activity participation* (never, 1–4 times/month, 5+ times/month) and their *smoking status* (never, past smoker, current smoker) (Kuh et al., 2009b; Strand et al., 2011b). Current and past smokers were defined as those who smoked at least one cigarette a day for 12 months or more.

Health Status (Ages 53, 60–64, 69)

Current health conditions (yes/no for each) were ascertained using a series of self-reported questions on *history of diabetes, cardiovascular events, respiratory symptoms, and knee pain* (Kuh et al., 2005; Cooper et al., 2014). *Symptoms of depression and anxiety* were assessed using the 28-item self-reported General Health Questionnaire; each item was scored from 1 to 4 and summed together (range: 0–84) (Goldberg and Hillier, 1979).

Cognitive Ability (Ages 53, 60–64, 69)

Verbal memory was assessed using a 15-item word list. Each word was presented for 2 s before individuals were instructed to write down as many words as they could remember. This was repeated over three identical trials and the number of words recalled during each trial were summed (range: 0–45). To minimize any practice effects, two word lists were rotated between follow-up assessments (Davis et al., 2017).

Statistical Analyses

Sex differences in each risk factor were assessed using t-tests or chi-square tests, as appropriate, and described by the mean ($\pm$SD) or proportion (n). Separate multilevel models were used to examine the associations between each risk factor (independent variable) and log transformed balance time (dependent variable) in the maximal available sample size. Cross-sectional associations were assessed for time-varying covariates (e.g., anthropometric, behavioral, health, cognitive factors), whereas SEP measures were based on reports from one age. Balance times at each age (level 1) were nested within individuals (level 2) and both the intercept and slope were modeled as random effects. As the sample was age-homogenous, age was employed as a linear time metric and was centered at age 53 (intercept); age 63 was utilized as the time integer for age 60–64 (Kuh et al., 2019; Blodgett et al., 2020). Balance times were log-transformed due to the skewed distribution of balance. Non-linearity of the association between each risk factor and balance was assessed using likelihood ratio tests.

A variable-by-sex interaction term was estimated, with subsequent models stratified by sex if there was evidence of an interaction. An interaction between age and each risk factor was added to the model to test whether the association between each risk factor and balance changes with age. Age interactions were considered if $p < 0.05$; an alpha of 0.05 was used for both age and sex interactions in order to parsimoniously build each model. Finally, all risk factors and significant interaction terms were included in a combined model. To account for the non-random events of mortality and attrition (not due to death), the model was adjusted for separate binary indicators of both death and attrition (not due to death) between ages 53 and 69. This approach minimizes the correlation between non-random loss to follow up and poorer performance on the balance test, thus reducing bias in the other estimates (Botoseneanu and Liang, 2012; Botoseneanu et al., 2013). All estimates are presented as sympercents (i.e., as % change) to aid interpretation (Cole and Kryakin, 2000). Stata 14 was used for all statistical analyses.

RESULTS

Characteristics of the sample are described in **Table 1**. Men were taller than women, had higher adult SEP, higher educational attainment, lower verbal memory, and were more likely to have a history of smoking. Men also had a higher prevalence of diabetes and CVD events, although women reported higher prevalence of knee pain and symptoms of anxiety and depression.

TABLE 1 | Characteristics of analytical sample (n = 3,111), MRC National Survey of Health and Development.

		Men (n = 1,550)	Women (n = 1,561)	Tests of sex differences (p-value)
ONE-LEGGED BALANCE TIME (s), MEDIAN (Q1, Q3), *n*				
Age 53		5 (3, 10), n = 1421	4 (3, 7), n = 1,476	<0.001
Age 60-64		3.57 (2.35, 5.53), n =1,055	3.16 (2.16, 4.72), n = 1,148	<0.001
Age 69		2.94 (1.84, 4.78), n = 1,037	2.72 (1.69, 4.15), n = 1,079	<0.005
SOCIOECONOMIC INDICATORS, *n* (%)				
Paternal occupational class				
I Professional/II Intermediate		407 (27.6)	383 (26.0)	0.56
III Skilled (non-manual or manual)		692 (46.9)	716 (48.7)	
IV Partly skilled/V Unskilled		377 (25.5)	372 (25.3)	
Maternal education				
Secondary and further education		162 (11.74)	169 (12.2)	0.49
Secondary only		167 (12.1)	153 (11.0)	
Primary and further education		213 (15.4)	193 (13.90)	
Primary only		838 (60.7)	873 (62.9)	
Highest household occupational class				
I Professional/II Intermediate		788 (51.6)	559 (36.1)	<0.001
III Skilled (non-manual or manual)		578 (37.8)	659 (42.6)	
IV Partly skilled/V Unskilled		162 (10.6)	329 (21.3)	
Educational attainment at age 26				
Degree or higher		212 (14.5)	81 (5.5)	<0.001
GCE A level or Burnham B		408 (27.9)	343 (23.3)	
GCE O level or Burnham C		211 (14.4)	377 (25.6)	
Sub GCE		92 (6.3)	134 (9.1)	
None attempted		540 (36.9)	537 (36.5)	
ANTHROPOMETRY, MEAN (SD)				
Height (m)				
Age 53		1.75 (0.07), n = 1,436	1.62 (0.06), n = 1,498	<0.001
Age 60–64		1.75 (0.09), n=1062	1.62 (0.06), n=1159	<0.001
Age 69		1.73 (0.09), n = 1,023	1.61 (0.06), n = 1,077	<0.001
BMI (kg/m^2)				
Age 53		27.4 (4.0), n = 1,435	27.4 (5.5), n = 1,486	0.89
Age 60–64		27.9 (4.1), n = 1,061	27.9 (5.5), n = 1,158	0.92
Age 69		28.2 (4.6), n = 1,040	28.2 (5.7), n = 1,081	0.91
BEHAVIORAL RISK FACTORS, *n* (%)				
Leisure time physical activity				
Age 53	None	693 (47.9)	761 (50.4)	0.18
	1–4 times/month	270 (18.7)	245 (16.2)	
	5+ times/month	485 (33.5)	503 (33.3)	
Age 60–64	None	681 (65.2)	716 (62.9)	0.52
	1–4 times/month	137 (13.1)	162 (14.2)	
	5+ times/month	227 (21.7)	261 (22.9)	
Age 69	None	711 (59.9)	777 (61.33)	0.08
	1–4 times/month	135 (11.4)	170 (13.42)	
	5+ times/month	341 (28.7)	320 (25.3)	
Smoking status				
Age 53	Current	343 (23.6)	339 (22.5)	<0.001
	Previous smoker	737 (50.8)	671 (44.5)	
	Never smoker	371 (25.6)	499 (33.1)	
Age 60–64	Current	137 (12.3)	142 (11.8)	<0.001

(Continued)

TABLE 1 | Continued

		Men (n = 1,550)	Women (n = 1,561)	Tests of sex differences (p-value)
	Previous smoker	663 (59.5)	629 (52.1)	
	Never smoker	314 (28.2)	436 (36.1)	
Age 69	Current	123 (10.3)	111 (8.8)	<0.001
	Previous smoker	756 (63.5)	723 (57.2)	
	Never smoker	311 (26.1)	430 (34.0)	
HEALTH STATUS, *n* (%)				
Diabetes	Age53	57 (3.1)	43 (2.4)	0.18
	Age60–64	129 (10.1)	99 (7.2)	<0.01
	Age69	175 (13.7)	136 (10.0)	<0.005
CVD events	Age53	85 (5.8)	48 (3.2)	<0.01
	Age60–64	131 (11.5)	62 (5.1)	<0.001
	Age69	193 (17.6)	114 (10.0)	<0.001
Respiratory symptoms	Age 53	292 (19.9)	276 (18.2)	0.22
	Age 60–64	233 (20.1)	224 (18.2)	0.15
	Age 69	264 (24.5)	266 (22.4)	0.23
Knee pain	Age 53	226 (15.5)	310 (20.6)	<0.001
	Age 60–64	216 (20.3)	288 (24.6)	0.01
	Age 69	190 (18.1)	241 (22.1)	0.02
Depression/anxiety	Age 53	15.2 (7.3), n = 1,051	17.8 (8.9), n = 1,137	<0.001
	Age 60–64	15.7 (8.6), n = 1,407	18.9 (10.3), n = 1,470	<0.001
	Age 69	14.1 (7.5), n = 1025	16.2 (8.2), n = 1068	<0.001
VERBAL MEMORY SCORES, MEAN (SD)				
	Age 53	23.0 (6.2), n = 1,397	24.9 (6.2), n = 1473	<0.001
	Age 60–64	23.0 (5.9), n = 1,023	25.4 (6.1), n = 1,127	<0.001
	Age 69	21.2 (6.0), n = 1,005	23.1 (6.0), n = 1057	<0.001

Sex, Age, and Balance

Women had 18.8% (95%CI: 13.6, 23.9%; **Table 2**, **Figure 1**) worse balance performance than men at age 53. The interaction between age and sex indicated that for every additional year increase in age, the sex difference in balance decreased by 0.7% (0.3, 1.2%). Thus, at ages 63 and 69, respectively, women had 11.4% (7.6, 15.2%) and 7.0% (2.1, 11.9%) lower balance times than men. Despite the sex differences in balance performance across time, there were no interactions between sex and any of the risk factors investigated.

Socioeconomic Indicators and Balance

The results of the likelihood ratio tests for deviations from linearity suggested that all four socioeconomic indicators could be modeled as continuous variables. More disadvantaged SEP for all four indicators—paternal occupation, maternal education, own education, own occupation—was associated with worse balance times (**Table 2**, **Figure 2**). For example, more disadvantaged paternal occupational class was associated with 14.8% (11.1, 18.4%; **Table 2**, **Figure 2A**) poorer balance time for each subsequent level. The associations with paternal occupational class, maternal education and own educational attainment all became smaller with age (all $p < 0.001$, **Table 2**, **Figure 2**), however there was no interaction between own occupational class and age ($p = 0.1$).

Anthropometric Indicators and Balance

Height had a quadratic association with balance such that taller individuals had better balance times than shorter individuals although this association plateaued at the tallest heights (see **Table 2**, **Figure 3**). BMI had an inverse linear association with balance, where every additional kg/m^2 was associated with 2.8% (2.5, 3.1%) poorer balance time (**Table 2**, **Figure 4**). There was no evidence of an interaction with age for either height ($p = 0.1$) or BMI ($p = 0.6$) suggesting that the association stayed constant over time (**Table 3**).

Health Behaviors and Balance

Those who participated in leisure time physical activity 1–4 times [23.9% (17.3, 30.5%)] or 5+ times [23.3% (18.0, 28.7%)] per month had better balance times than those who did not participate at age 53 (**Table 2**, **Figure 5A**). There was no difference in balance between those who participated in leisure time physical activity 1–4 times/month and those who participated 5+ times/month. There was evidence that the association got smaller with age, as shown by the age-interaction for those who participated 1–4 times/month. Individuals who had a past history of smoking or who were current smokers had worse balance ability than those who had never smoked [6.1% (3.3, 8.9%); **Table 2**,

TABLE 2 | Associations between risk factors and balance performance in multilevel models.

Risk factors[a]	*n* participants (*n* observations)	Mean difference in % balance time at age 53 (intercept)		Age (years)*risk factor interaction	
		Coefficient (%) (95% CI)	*p-value*	Coefficient (%) (95% CI)	*p-value*
1: Sex [female vs. male (ref)]	3,111 (obs = 7,216)	−18.8 (−23.9, −13.6)	<0.001	0.7 (0.3, 1.2)	<0.001
2: Paternal occupational class[b] (per 1 level change)	2,947 (obs = 6,838)	−14.8 (−18.4, −11.1)	<0.001	0.5 (0.2, 0.8)	<0.001
3: Maternal education[c] (per 1 level change)	2,768 (obs = 6,424)	−11.3 (−23.8, −8.8)	<0.001	0.4 (0.2, 0.6)	<0.001
4: Education at age 26[d] (per 1 level change)	2,935 (obs = 6,830)	−11.1 (−12.9, −9.3)	<0.001	0.3 (0.2, 0.5)	<0.001
5: Own occupational class[b] (per 1 level change)	3,075 (obs = 7,167)	−15.2 (−17.9, −12.6)	<0.001	-	-
6: Height (cm)					
linear term	3,090 (obs = 7,144)	12.5 (6.4, 18.7)	<0.001	-	-
quadratic term		−0.04 (−0.05, −0.02)	<0.001		
7: BMI (per kg/m^2)	3,083 (obs = 7,150)	−2.8 (−3.1, −2.5)	<0.001	-	-
8: Leisure time physical activity[e]	3,094 (obs = 6,960)				
1–4 times/week		23.9 (17.3, 30.5)	<0.001	−0.7 (−1.3, 0.02)	0.04
5+ times/week		23.3 (18.0, 28.7)	<0.001	−0.4 (−0.9, 0.1)	0.12
9: Smoking[f] (per 1 level change)	3,092 (obs = 6,996)	6.1 (3.3, 8.9)	<0.001	-	-
10: Diabetes[g]	3,111 (obs = 7,214)	18.0 (11.6, 24.4)	<0.001	-	-
11: CVD events[g]	3,072 (obs = 6,895)	19.1 (12.9, 25.4)	<0.001	-	-
12: Respiratory symptoms[g]	3,062 (obs = 6,634)	8.6 (4.5, 12.7)	<0.001	-	-
13: Knee pain [g]	3,108 (obs = 7,173)	10.8 (4.6, 17.0)	<0.001	−0.7 (−1.3, −0.01)	0.02
14: Symptoms of depression/anxiety[h] (per 1 SD)	3,071 (obs = 7,032)	5.3 (2.8, 7.7)	<0.001	0.3 (0.04, 0.5)	<0.02
15: Verbal memory[h] (per 1SD)	3,035 (obs = 6,979)	13.4 (11.0, 15.9)	<0.001	−0.4 (−0.6, −0.2)	<0.05

[a]*all models adjusted for sex as no evidence of sex interactions (see **Table 3**).*
[b]*ref: I Professional or II Intermediate.*
[c]*ref: Secondary and further education.*
[d]*ref: Degree or higher.*
[e]*ref: none in last 4 weeks.*
[f]*ref: current smoker.*
[g]*ref: individuals with no health condition.*
[h]*SD estimates at each age are provided in **Table 1**.*

Figure 5B]; there was no evidence that this association changed with age.

Current Health Status and Balance

Individuals who had a history of diabetes, a history of CVD events or current respiratory symptoms had worse balance performance; these associations remained constant with age (**Table 2**, **Figures 6A–C**). Those who reported knee pain had 10.8% (4.6, 17.0%) lower balance times at age 53; this association got larger with age [0.7% per year (0.1, 1.2)] such that those with knee pain at age 69 had 21.6% (15.4, 27.8%) poorer balance than those with no knee pain (**Table 2**, **Figure 6D**). A one standard deviation increase in depression and anxiety symptoms on the GHQ-28 questionnaire was associated with a 5.2% (2.8, 7.7%, **Table 2**, **Figure 6E**) decrease in balance times. This association also increased with age by 0.3% per year (0.04, 0.5%); by age 69, 1 SD increase in GHQ-28 score was associated with a 9.5% (7.0, 11.9%) decrease in balance time.

Cognitive Ability and Balance

One standard deviation increase in verbal memory was associated with a 13.4% (11.0, 15.9%; **Table 2**, **Figure 7**) increase in balance time at age 53. This association got smaller with age (0.5% per year, (0.3, 0.7%), but remained associated with balance at age 69 [5.1%, (2.6, 7.5%)].

Combined Model of All Covariates and Their Association With Balance

Table 4 provides the estimates for the combined model of all covariates and the relevant age interaction terms. Notably, being female, having higher BMI, lower maternal education, lower educational attainment, lower own occupational class, not participating in leisure time physical activity, reporting a history of CVD events, higher levels of anxiety and depression and lower verbal memory remained associated with lower balance time. Nearly all age interaction terms weakened and were no longer statistically significant (at the 5% level) in this model, although there remained evidence that the associations with sex and verbal memory decreased with age.

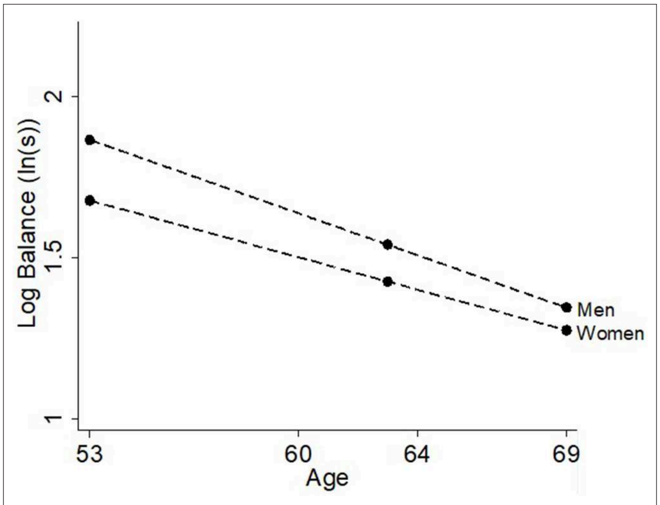

FIGURE 1 | Differences in balance time at ages 53, 60–64, and 69 years by sex.

DISCUSSION

Main Findings

We quantified associations between a range of risk factors across life and balance performance at ages 53, 60–64, and 69. Individuals with better balance were more likely to be male, have higher SEP in both childhood and adulthood,

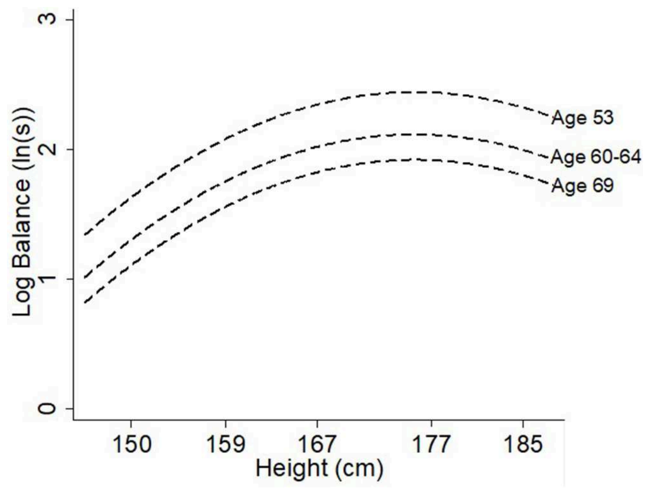

FIGURE 3 | Differences in balance time at ages 53, 60–64, and 69 years by height (cm).

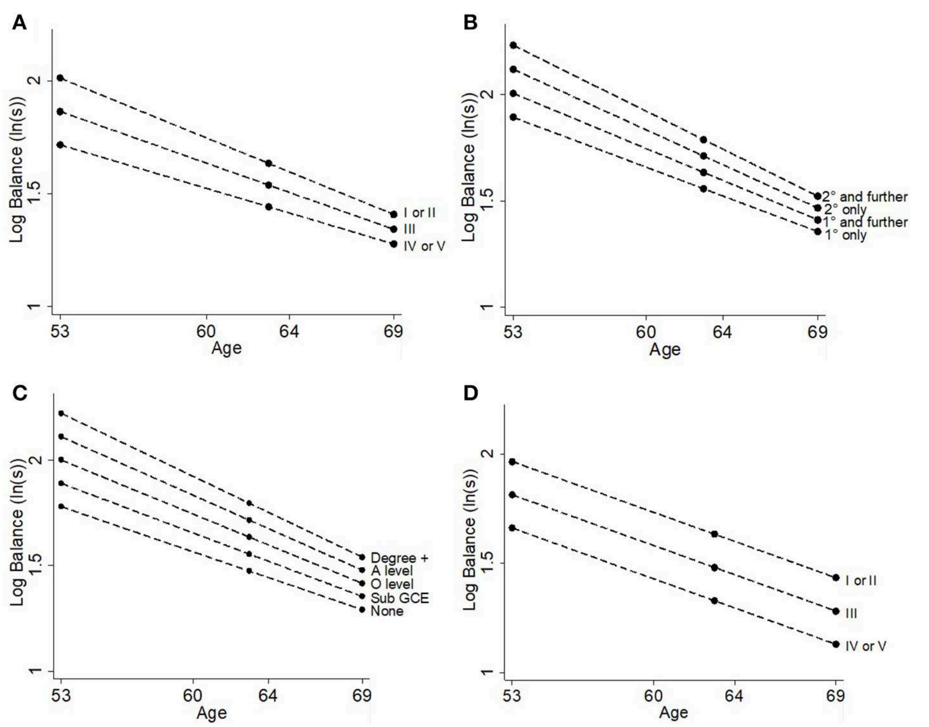

FIGURE 2 | Differences in balance time at ages 53, 60–64, and 69 years by **(A)** paternal occupational class, **(B)** maternal education, **(C)** own education, and **(D)** own occupational class.

be taller, have lower BMI, partake in leisure time physical activity and were less likely to smoke. Individuals with better balance were also more likely to be healthier (no history of diabetes or CVD, not currently experiencing respiratory symptoms or knee pain), less likely to be experiencing symptoms of depression and anxiety, and more likely to have higher verbal memory. In a combined model, the majority of risk factors remained independently associated with balance, indicating that the factors across life that are associated with one-legged balance performance are multifaceted and complex.

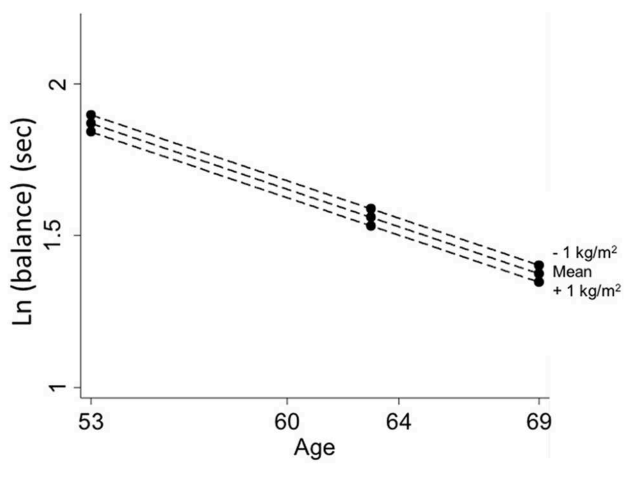

FIGURE 4 | Differences in balance time at ages 53, 60–64, and 69 years by BMI (kg/m^2).

Sex differences in balance performance were not explained by adjustment for other risk factors. Furthermore, there was no evidence to suggest that the associations between these risk factors and balance differed by sex, although several associations did change with age. Associations of balance performance with sex, socioeconomic indicators, physical activity and verbal memory became smaller with increasing age, while associations with anthropometric indicators, smoking, and physical health status stayed constant. Two associations became larger with age; associations of both knee pain and symptoms of anxiety and depression with balance doubled from age 53–69.

Comparison With Other Studies and Explanation of Findings

Socioeconomic Indicators

A systematic review and meta-analysis of over 22 000 individuals from 11 separate studies reported that lower childhood SEP (as indicated by parental occupation and education) was associated with inability to balance with eyes open for ≥ 5 s (Birnie et al., 2011b); adjustment for adult SEP fully attenuated the effect of childhood SEP (paternal occupation used if available). However, maternal education and both indicators of adulthood SEP remained independently associated with balance time in our fully-adjusted model. In addition to differing operationalisations of one-legged balance performance (continuous vs. binary; eyes closed vs. eyes open), a possible explanation for these differing results is that 9 of the 11 studies included in the meta-analysis relied upon retrospective reports of childhood SEP (Birnie et al.,

TABLE 3 | Summary of tests of non-linearity, sex interactions and age interactions of all covariates with balance performance.

	Description of how variable is modeled	Sex interaction *p*-value	Age interaction effect on size of association
Sex (female)	n/a[a]	n/a	↓ with age
Socioeconomic indicators			
Paternal occupational class	Continuously	0.9	Effect ↓ with age
Maternal education	Continuously	0.7	Effect ↓ with age
Education	Continuously	0.5	Effect ↓ with age
Own occupational class	Continuously	0.4	Constant with age
Anthropometry			
Height	Quadratic term	0.9	Constant with age
BMI	Linear term only	0.1	Constant with age
Health behaviors			
Leisure time physical activity	Categorically	0.7	Effect ↓ with age
Smoking	Continuously	0.1	Constant with age
Current health status			
History of diabetes	n/a[a]	0.5	Constant with age
History of cardiovascular events	n/a[a]	0.2	Constant with age
Respiratory symptoms	n/a[a]	0.6	Constant with age
Knee pain	n/a[a]	0.8	Effect ↑ with age
Symptoms of anxiety & depression	Linear term only	0.4	Effect ↑ with age
Other			
Verbal memory	Linear term only	0.2	Effect ↓ with age

[a] *Unable to test non-linearity in dichotomous indicators.*

2011a). A strength of NSHD is that data on SEP and other risk factors were prospectively ascertained and so not prone to recall bias. As previously shown in relation to cognitive outcomes (Kaplan et al., 2001; Guralnik et al., 2006), paternal occupational class and maternal education may have distinctive associations with balance performance; further exploration of these differences are required.

While childhood SEP is hypothesized to be associated with balance ability via a complex pathway of health behaviors, education, adult SEP, cognitive ability and/or health conditions, the association remained when these factors were included in the model. Thus, childhood may also represent a sensitive period of development for balance ability, as previously hypothesized when testing associations of childhood cognition and midlife balance performance in NSHD (Blodgett et al., 2020). Adult SEP may also be associated with balance through a pathway of current physical and cognitive health or health behaviors. That both childhood and adult SEP indicators remained independently associated with balance suggests that accumulation of low SEP across the life course may be a greater risk factor than low SEP at any one particular life stage.

Notably, the relationship between most SEP indicators and balance time weakened with increasing age. This suggests that SEP may be more strongly associated with balance in midlife than at older ages when substantial age-related decline begins and chronic diseases manifest. Nevertheless, the association between the most recent measure of SEP (occupational class at age 53) and balance did not change with age.

Anthropometric Indicators

Higher body mass may influence the stability of an individual and the motor mechanisms involved in the balance process. For example, individuals with higher BMI often require more movement in order to maintain their balance, thus frequently demonstrate high levels of postural sway and reduced balance performance (D'Andréa Greve et al., 2013; Hita-Contreras et al., 2013). Studies have suggested that body stability is inversely related to the height of the center of gravity (D'Andréa Greve et al., 2013) and that shorter individuals are better able to maintain their balance. However, we found that taller individuals had better balance than shorter individuals though this effect appeared to plateau above a certain height.

Previous evidence has suggested that sex differences in balance performance disappear when scores are normalized to height (Maki et al., 1990; Hageman et al., 1995; Era et al., 2002; Bryant et al., 2005), while other studies have shown that anthropometric factors are major determinants of balance performance in women only (Kim et al., 2012). However, we found no sex differences in the association of either height or BMI with balance ability and adjustment for these measures did not explain sex differences (as seen in the combined model). Given that men have higher average strength and mobility compared to women (Miller et al., 1993; Sugimoto et al., 2014; Zunzunegui et al., 2015), further investigation into whether more detailed assessment of body composition (e.g., lean mass, fat mass) explains sex differences is warranted.

Health Behaviors

It is well-recognized that low levels of physical activity (de Rezende et al., 2014; Cooper et al., 2016; Olanrewaju et al., 2016; Schwingshackl et al., 2017) and current or past smoking (Cooper et al., 2016; Daskalopoulou et al., 2018) have negative consequences for an individual's physical capability, including their balance performance. Some studies have shown increasing levels of physical activity are associated with better balance (Powell et al., 2011; Cooper et al., 2015), while others have shown that there is no difference in health benefit between moderately active and maximally active groups (Cooper et al., 2016). In this study, participation in leisure time physical activity was associated with better balance performance. Although there was little additional benefit for balance ability beyond 1–4 times per month at age 53, a graded association between increasing levels of physical activity and balance performance emerged by age 69 (see **Figure 5A**).

Individuals who currently smoked had worse balance performance compared with those who were ex-smokers; ex-smokers also had worse balance compared with those who had never smoked. Although not previously examined in balance ability, this is consistent with increasing severity of poor physical capability seen amongst categories of smoking history (North et al., 2015; Cooper et al., 2016), suggesting that quitting smoking may have a positive association with balance performance.

Current Health Status

The presence of each physical and mental health condition (diabetes, CVD, respiratory symptoms, knee pain, symptoms of anxiety, and depression) was associated with poorer balance performance. This is consistent with the literature on how current health impacts an individual's physical capability or functional decline (Welmer et al., 2012; Kuh et al., 2014; Ryan et al., 2015). Each health condition likely has a direct biological pathway impacting balance. For example, diabetes is related to both peripheral neuropathy (Greene et al., 1992) and age-related visual impairment (Lutty, 2013; Pelletier et al., 2016) while knee pain can have a direct impact on proprioception and musculoskeletal function (Sanchez-Ramirez et al., 2013). Individuals with a history of CVD events or respiratory symptoms often demonstrate shared pathophysiological features common in those with balance impairment including increased postural sway due to physical displacement of breathing (Jeong, 1991), decreased blood flow in specific functional areas (Abate et al., 2009) and decreased musculoskeletal capacity (Crisan et al., 2015). Finally, increased inflammatory markers that are common in arthritis, such as C-reactive protein or nitric oxide (Cepeda et al., 2016) are also more common in individuals with depression than those without. In addition to this inflammation pathway, individuals with depression also tend to restrict their physical activity, have reduced motivation to perform well and exhibit psychomotor impedance such as a slowing in musculoskeletal components (Bennabi et al., 2013); all of these factors can influence balance performance.

The associations of diabetes, CVD and respiratory symptoms with balance performance were constant. However, the associations of knee pain and symptoms of anxiety and

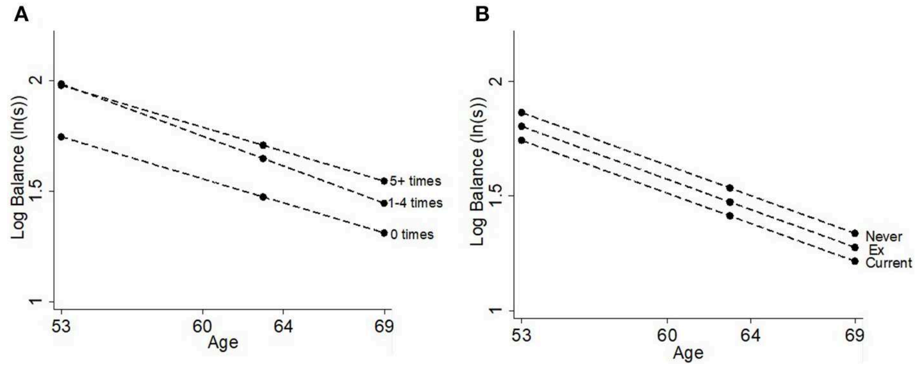

FIGURE 5 | Differences in balance time at ages 53, 60–64, and 69 years by **(A)** leisure time physical activity and **(B)** smoking status.

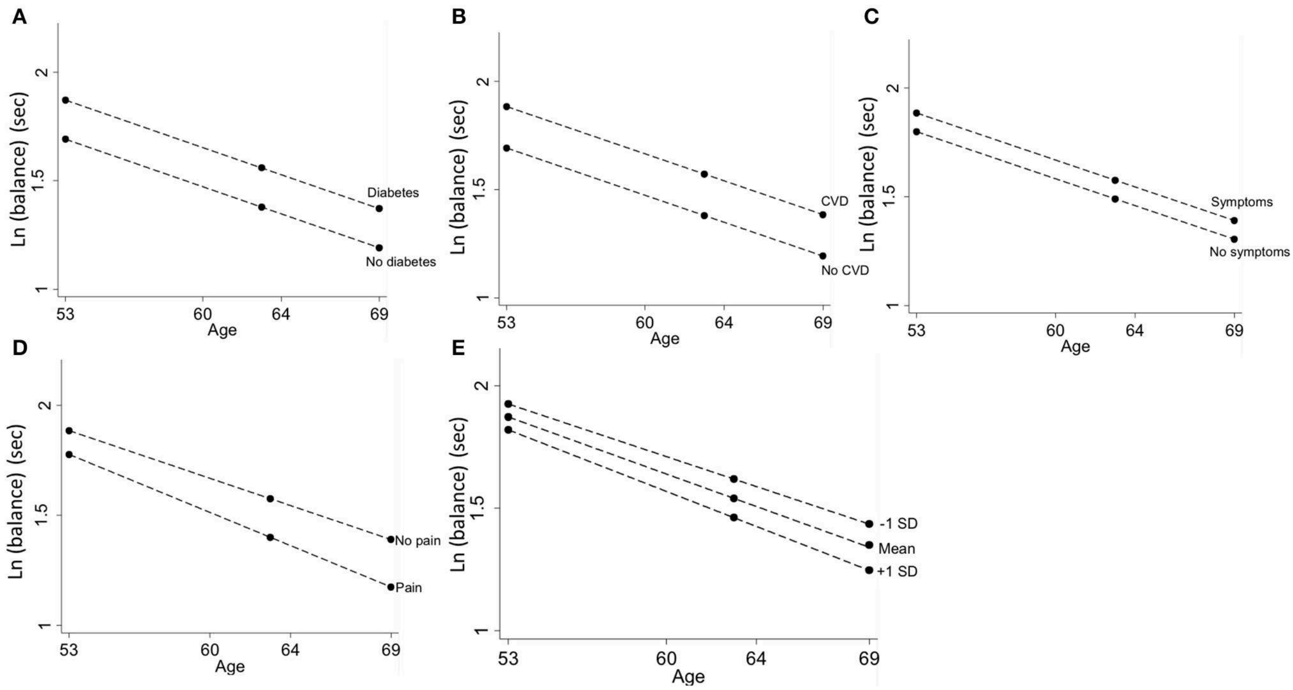

FIGURE 6 | Differences in balance time at ages 53, 60–64, and 69 years by **(A)** diabetes, **(B)** CVD events, **(C)** respiratory symptoms, **(D)** knee pain, and **(E)** symptoms of depression and anxiety.

depression with balance got stronger at older ages. The constant or increasing associations between health conditions and balance ability with age suggests that overall health becomes relatively more important for balance ability in later life; this could in part explain why the strength of associations with many other risk factors decreased with increasing age.

Cognitive Ability

As expected given previous findings in NSHD (Kuh et al., 2009a; Blodgett et al., 2020), higher verbal memory was associated with higher levels of balance performance. Cognitive processing of sensory and motor input is an important component of the balancing process (Li et al., 2018). Previous evidence in NSHD has shown that childhood cognitive ability is associated with adult balance performance; this is primarily via an adult cognition and education pathway that is independent of most of the other risk factors examined here (Blodgett et al., 2020). As suggested above, the decreasing strength of association with age suggests that cognitive ability becomes less important with age, as other factors in the aging process, in particular health conditions, become more important.

Methodological Considerations

A major strength of this paper is the assessment of balance performance at three ages, which facilitated our novel investigation into whether associations between risk factors and balance ability change over 16 years, from mid to later life. A second strength is the comprehensive investigation, in separate and combined models, of the associations between 14 different factors across life and balance performance. These risk factors were all prospectively ascertained which increases reliability of response and limits recall bias. A third strength was the methods used to include those individuals who had missing balance scores for health reasons or because of death or loss to follow up between ages 53 and 69. These combined strengths provided novel evidence on how the associations between these risk factors and balance ability change with age. Finally, the age homogeneity of the sample eliminated any confounding by age that is common when examining physical capability in mid and later life (Seeman and Chen, 2002; Garber et al., 2010).

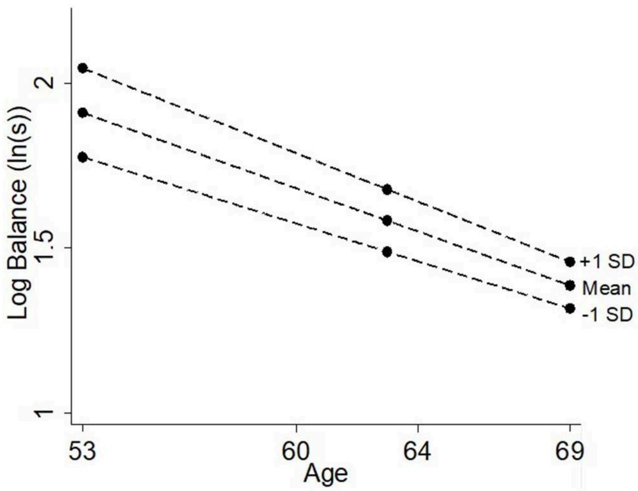

FIGURE 7 | Differences in balance time at ages 53, 60–64, and 69 years by verbal memory.

One potential limitation of our study is that we were unable to include participants in analyses if they had been lost to follow up before age 53 (i.e., the age at which balance was first assessed). Characteristics of study members who were lost to follow up before the first clinical assessment at age 53 were more likely to be male (Stafford et al., 2013), have lower childhood and adulthood occupational class (Stafford et al., 2013; Kuh et al., 2016), demonstrate unhealthy behaviors (smoking, physical inactivity) (Stafford et al., 2013), have lower verbal memory (Stafford et al., 2013) and have poorer overall

TABLE 4 | Combined model of all risk factors and all significant age interactions from individual models additionally adjusting for death and attrition, $n = 2{,}465$ (obs = 5,150).

Risk factors[a]	Mean difference in % balance time at age 53 (intercept)		Age (years)*risk factor interaction	
	Coefficient (%) (95% CI)	***p-value***	**Coefficient (%) (95% CI)**	***p-value***
Sex (female)	−21.7 (−28.7, −14.7)	<0.001	0.9 (0.4, 1.3)	<0.001
Paternal occupational class[b] (per 1 level change)	−2.7 (−6.8, 1.5)	0.21	0.3 (−0.1, 0.7)	0.11
Maternal education[c] (per 1 level change)	−3.9 (−6.7, −1.0)	<0.01	0.1 (−0.2, 0.3)	0.51
Education at age 26[d] (per 1 level change)	−3.8 (−6.2, −1.3)	<0.005	−0.2 (−0.4, 0.02)	0.08
Own occupational class[b] (per 1 level change)	−4.9 (−8.2, −1.7)	<0.005	–	–
Height (m)				
linear term	5.0 (−1.6, 11.6)	0.13	–	–
quadratic term	−0.02 (−0.04, 0.004)	0.12		
BMI (per kg/m^2)	−2.1 (−2.5, −1.7)	<0.001	–	–
Leisure time physical activity[e]				
1–4 times/week	9.1 (1.9, 16.3)	<0.001	0.1 (−0.6, 0.8)	0.28
5+ times/week	5.8 (−0.2, 11.8)		0.3 (−0.3, 0.9)	
Smoking[f] (per 1 level change)	1.4 (−1.6, 4.4)	0.36	–	–
Diabetes[g]	−6.9 (−14.1, 0.5)	0.07	–	–
CVD events[g]	−7.1 (−14.0, −0.3)	0.04	–	–
Respiratory symptoms[g]	−2.5 (−7.0, 2.0)	0.28	–	–
Knee pain[g]	−4.5 (−11.2, 2.1)	0.18	−0.2 (−0.8, 0.4)	0.55
Symptoms of depression/anxiety[h] (per 1 SD)	−3.1 (−5.8, −0.4)	0.02	−0.1 (−0.4, 0.2)	0.46
Verbal memory[h] (per 1SD)	5.9 (2.8, 8.9)	<0.001	−0.4 (−0.7, −0.1)	<0.01

[a]*all models adjusted for sex as no evidence of sex interactions (see* ***Table 3****).*
[b]*ref: I Professional or II Intermediate.*
[c]*ref: Secondary and further education.*
[d]*ref: Degree or higher.*
[e]*ref: none in last 4 weeks.*
[f]*ref: current smoker.*
[g]*ref: individuals with no health condition.*
[h]*SD estimates at each age are provided in* ***Table 1****.*

health (Kuh et al., 2016). Participants who were followed up to age 53 but could not be included in analyses due to missing data on risk factors had similar characteristics to those lost to follow-up before age 53. Many of these characteristics (i.e., low SEP, unhealthy behaviors, lower cognition, and poorer health) were negatively associated with balance ability, thus it is hypothesized that those with lost to follow up before age 53 or with missing risk factor data may have had poorer balance. This likely resulted in an underestimation of the size and strength of associations.

Two further limitations are the assumptions of the model: that the change in balance over time is linear and that all individuals follow the same mean trajectory of decline. Individuals are likely to exhibit heterogonous aging trajectories as they demonstrate different patterns of change in balance performance with age (e.g., steeper decline, delayed decline, maintenance of balance ability). As there were only three time points for balance, it was not appropriate to test for non-linearity in balance trajectories. Identifying polynomial time terms can help identify the age at which decline begins or accelerates. Further research, with at least four measures of balance performance, should address these differences across time and between individuals. We also need to consider the possibility that other factors not considered in our analyses such as alcohol consumption and medication use may also be important and need to be considered in future research.

Although there were multiple comparisons, the risk of Type 1 error remains low as all of the primary associations in **Table 2** were significant at $p < 0.001$ and an alpha of 0.05 was intentionally used for interaction terms to ensure a parsimonious model. Finally, one-legged balance ability measures a specific aspect of static balance that does not directly represent the dynamic balance relied upon in everyday situations (Owings et al., 2000; Mackey and Robinovitch, 2005; Bhatt et al., 2011). Further research should consider if associations between the risk factors identified in this study are consistent for tests of dynamic balance or for more sensitive measures of postural sway, as assessed with a force plate.

Implications and Conclusions

We investigated 14 different factors across life that are associated with balance performance. These findings are important in considering appropriate interventions to minimize balance decline or when identifying high-risk individuals. That multiple risk factors were identified suggests that a multifactorial approach including behavioral, health and cognitive factors (amongst others) may have more benefit than a focus on a single risk factor. As several of these risk factors have different associations with balance at different ages, there may be benefit in targeting different factors at different ages. Knee pain and symptoms of depression and anxiety both appear to become more important with age and may represent important targets for intervention in midlife before their potential association with balance performance increases. While not all of the factors identified (e.g., socioeconomic, height, smoking history) may be easily modified, they are likely to have utility in helping to identify individuals at high risk of future balance difficulties who may require more support than others to maintain balance ability as they age.

In summary, this study identified many anthropometric, behavioral, socioeconomic, health and cognitive risk factors across life that are associated with balance ability. The majority of variables remained independently associated with one-legged balance performance, suggesting that the range of risk factors associated with poor balance ability is diverse and complex. This highlights the importance of considering both type (i.e., multifactorial approach) and timing (i.e., early, mid, and later life) of interventions that target balance performance in adulthood.

DATA AVAILABILITY STATEMENT

The datasets used in this study will not be made publicly available. Access to NSHD data adheres to strict confidentiality guidelines but these data are available to bonafide researchers upon request to the NSHD Data Sharing Committee via a standard application procedure. Further details can be found at http://www.nshd.mrc.ac.uk/data. doi: 10.5522/NSHD/Q101; doi: 10.5522/NSHD/Q102; doi: 10.5522/NSHD/Q103.

ETHICS STATEMENT

At each wave of data collection, relevant ethical approval has been received. Ethical approval for the most recent data collection (2014–2015) was obtained from Queen Square Research Ethics Committee (13/LO/1073) and Scotland A Research Ethics Committee (14/SS/1009). All participants have provided written informed consent.

AUTHOR CONTRIBUTIONS

JB performed statistical analyses and wrote the first draft of the manuscript. All authors contributed to the conception and design of the study, to manuscript revision, read, and approved the submitted version.

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Walking With Ears: Altered Auditory Feedback Impacts Gait Step Length in Older Adults

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Auditory feedback may provide the nervous system with valuable temporal (e. g., footstep sounds) and spatial (e.g., external reference sounds) information that can assist in the control of upright walking. As such, hearing loss may directly contribute to declines in mobility among older adults. Our purpose was to examine the impact of auditory feedback on the control of walking in older adults. Twenty older adults (65–86 years) with no diagnosed hearing loss walked on a treadmill for three sound conditions: Baseline, Ear Plugs, and White Noise. We hypothesized that in response to reduced temporal auditory feedback during the Ear Plugs and White Noise conditions, participants would adapt shorter and faster steps that are traditionally believed to increase mechanical stability. This hypothesis was not supported. Interestingly, we observed increases in step length ($p = 0.047$) and step time ($p = 0.026$) during the Ear Plugs condition vs. Baseline. Taking longer steps during the Ear Plugs condition may have increased ground reaction forces, thus allowing participants to sense footsteps via an occlusion effect. As a follow-up, we performed a Pearson's correlation relating the step length increase during the Ear Plugs condition to participants' scores on a clinical walking balance test, the Functional Gait Assessment. We found a moderate negative relationship (rho = −0.44, p = 0.055), indicating that participants with worse balance made the greatest increases in step length during the Ear Plugs condition. This trend suggests that participants may have actively sought auditory feedback with longer steps, sacrificing a more mechanically stable stepping pattern. We also hypothesized that reduced spatial localization feedback during the Ear Plugs and White Noise conditions would decrease control of center of mass (COM) dynamics, resulting in an increase in lateral COM excursion, lateral margin of stability, and maximum Lyapunov exponent. However, we found no main effects of auditory feedback on these metrics ($p = 0.580$, $p = 0.896$, and $p = 0.056$, respectively). Overall, these results suggest that during a steady-state walking task, healthy older adults can maintain walking control without auditory feedback. However, increases in step length observed during the Ear Plugs condition suggest that temporal auditory cues provide locomotor feedback that becomes increasingly valuable as balance deteriorates with age.

Keywords: locomotion, gait, balance, sound, hearing

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INTRODUCTION

To maintain and control upright walking, the human nervous system continuously processes and responds to multiple streams of sensory feedback. Although research has focused largely on visual (Warren et al., 2001; McAndrew et al., 2011), vestibular (Fitzpatrick et al., 1999; Bent et al., 2000), and somatosensory feedback (Dietz and Duysens, 2000; Sinkjaer et al., 2000), growing evidence suggests that auditory feedback may also play an important role for controlling gait (Menzer et al., 2010; Baram et al., 2016; Camponogara et al., 2016; Shayman and Earhart, 2017; Weaver et al., 2017) (**Figure 1**). Sounds produced during walking may provide temporal cues indicating the timing of important events in the gait cycle. For example, footsteps, generated every time the feet contact the ground, mark the critical transition from leg swing to stance. When footstep sounds are artificially delayed, people modulate their walking speed, suggesting that this form of auditory feedback may be important for controlling step frequency (Menzer et al., 2010). Externally-generated environmental sounds may also provide spatial landmarks that can serve as a reference for controlling aspects of walking such as body posture and orientation (Karim et al., 2018). Although the evidence during walking is limited, several studies have found that auditory cues can improve standing postural balance (Easton et al., 1998; Kanegaonkar and Amin, 2012; Rumalla et al., 2015; Horowitz et al., 2019). In addition, the reflection of self-generated sounds associated with walking (i.e., echolocation) could provide information about an individual's absolute position and velocity (Stoffregen and Pittenger, 1995). Given that people's ability to predict location via echolocation improves with movement (Rosenblum et al., 2000), this form of auditory feedback may take on a heightened role during walking in comparison to standing postural tasks when whole-body movements are minimal. Collectively, auditory feedback may be valuable for controlling both temporal and spatial aspects of walking.

Hearing loss may deprive the nervous system of valuable sensory information for controlling walking. Indeed, hearing loss has been associated with poor mobility, including slower walking speeds, problems walking longer distances, and self-reported walking difficulties (Viljanen et al., 2009a; Li et al., 2013). Additionally, cohort studies have identified a positive relationship between hearing loss and the probability of falling (Viljanen et al., 2009b; Lin and Ferrucci, 2012; Girard et al., 2014) with even mild hearing loss (>25 dB) associated with a three-fold increase in fall rate (Lin and Ferrucci, 2012). The relationship between hearing loss and falls appears to hold even when controlling for age (Lin and Ferrucci, 2012), vestibular function (Lin and Ferrucci, 2012), and genetics (Viljanen et al., 2009b). Although the mechanisms have not been fully evaluated, this prior research suggests that loss of hearing is associated with declines in walking and balance. Because the research supporting this relationship is indirect (Viljanen et al., 2009b; Lin and Ferrucci, 2012; Girard et al., 2014), we do not currently know if hearing loss directly causes balance impairments that contribute to falls.

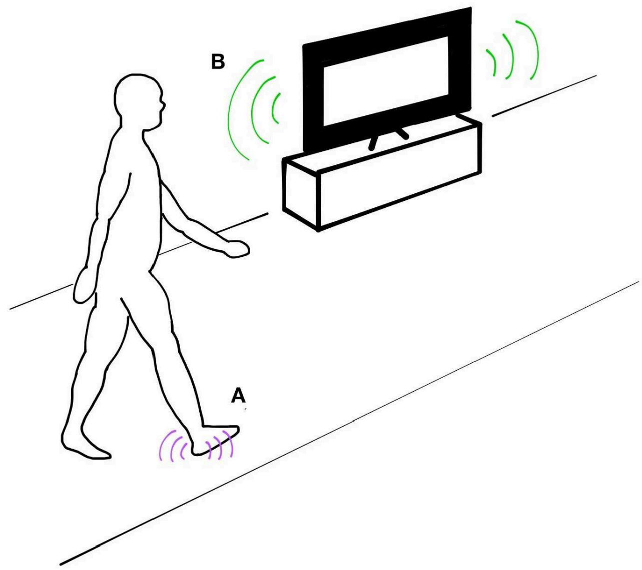

FIGURE 1 | Auditory feedback during walking can provide sensory cues that aid in the control of walking balance. **(A)** Temporal cues such as footsteps provide information about the timing of importance events occurring within the gait cycle. **(B)** External sounds can provide spatial references about an individual's position and velocity.

The potential that hearing loss may negatively impact walking balance has important implications for older adults. Among adults over 65 years old, falls are the leading cause of fatal and non-fatal injuries (Houry et al., 2016). The probability of falling increases linearly as the number of risk factors (e.g., muscle weakness or poor vision) increases (Tinetti et al., 1986). Thus, identifying preventable/modifiable risk factors is an essential component of comprehensive fall prevention programs. Despite evidence linking hearing loss to falls (Viljanen et al., 2009b; Lin and Ferrucci, 2012; Girard et al., 2014), hearing is rarely included in fall risk screening and intervention programs (Stevens, 2013). Considering that some forms of hearing loss are addressable by hearing aids or cochlear implants, there is a need to better understand if, and how, hearing loss impacts walking balance in older adult populations.

If auditory feedback is used to control walking, then observable gait modifications may occur in older adults following hearing loss. First, hearing loss may result in more variable step times. Proprioceptive feedback during walking can provide a clear sensory cue indicating when the foot contracts the ground. When the timing of this proprioceptive feedback is modulated, people will modulate the timing of the transitions between stance and swing phases of gait (Pang and Yang, 2000; Gordon et al., 2009). Given that proprioception declines with aging (Shaffer and Harrison, 2007), supplemental sensory feedback such as the sound of footsteps may improve older adults' ability to detect ground contact. If auditory feedback enhances the ability to detect ground contract, then the loss of hearing may result in more variable step timing. Second, older adults may adapt shorter and faster steps when auditory feedback is reduced. These stepping modifications are believed to improve walking balance by reducing center of mass (COM) peak velocities and increasing the opportunity to make corrective steps (Reimann et al., 2018), and are often observed when people walk in challenging and uncertain environments (McAndrew et al., 2010; Wu et al., 2017). If hearing loss reduces the ability of the nervous system to detect ground contact, thus increasing movement uncertainty, then adapting shorter, faster steps could be a simple compensatory strategy to maintain balance. Finally, hearing loss may particularly impact control of lateral COM motion while walking. During normal gait, the COM naturally oscillates from side-to-side as body weight is shifted over the supporting limb (Saunders et al., 1953). To maintain desired forward walking trajectories, research suggests that control of this frontal-plane motion requires greater involvement of the nervous system than control of sagittal-plane motions, which benefit considerably from the human body's passive dynamics (MacKinnon and Winter, 1993; Bauby and Kuo, 2000; O'Connor and Kuo, 2009; McAndrew et al., 2010). Based on this previous research, it is likely that loss of sensory feedback that provides spatial information will have a greater impact on control of COM motion in the frontal plane than in the sagittal plane. Thus, if older adults use auditory feedback to aid in controlling spatial aspects of walking, then hearing loss might result in lateral COM excursions that are larger, faster, and less stable.

The purpose of this study was to examine the impact of auditory feedback on the control of temporal and spatial aspects of walking balance in older adults. Older adults without diagnosed hearing loss were recruited for this within-subjects design. We selected a within-subjects design to minimize the potential effects of confounding factors. Participants walked on a treadmill under conditions in which auditory feedback was reduced by either a combination of ear plugs and over-ear earmuffs or playing white noise (75 dB) through insertable eartips. We hypothesized that compared to baseline walking, a reduction in auditory feedback would affect control of temporal aspects of gait. Specifically, we predicted that step time would be more variable under conditions of reduced auditory feedback and that individuals would adapt shorter, faster steps. We also hypothesized that reductions in auditory feedback would affect spatial control of COM motions during walking. Specifically, we predicted that under conditions of reduced auditory feedback, individuals would increase lateral COM excursion and peak lateral speed per stride and decrease local orbital stability.

MATERIALS AND METHODS

Participants

Twenty-two older adults who did not have hearing aids, cochlear implants, or a known diagnosis of hearing loss were recruited to participate in the study. Additional inclusion criteria included: 50–90 years of age, normal or corrected vision, and ability to walk continuously for 10 min without undue fatigue or health risk. Exclusion criteria included: musculoskeletal and/or vestibular pathologies that would affect balance and/or stability, current use of medication that might affect proprioception and/or balance, and cognitive deficits that preclude understanding of the instructions required to conduct the test. The Northwestern University Institutional Review Board approved the study protocol and all participants provided written informed consent.

Experimental Setup

We observed participants' gait during a series of walking trials in which auditory feedback was manipulated. Participants walked on a large treadmill, belt size 2.6 m × 1.4 m (Tuff Tread, Willis, TX). For safety, participants wore a trunk harness attached to a passive overhead safety support that allowed free fore-aft motion and was adjusted so it did not restrict small lateral motions or provide bodyweight support (Aretech, Ashburn, VA). Participants did not have access to handrails or other forms of external support during any of the walking trials.

To quantify treadmill walking kinematics, we used a 12-camera motion capture system (Qualisys, Gothenburg, Sweden) to record 3D coordinates of reflective markers placed on the pelvis and feet. Each participant wore a total of 13 reflective markers that we placed bilaterally on the greater trochanters, calcanei, lateral malleoli, 2nd metatarsals, and 5th metatarsals with three additional tracking markers placed on the pelvis. The motion capture system sampled kinematic data at 200 Hz.

Protocol

We performed three clinical assessments before treadmill testing to evaluate participants' hearing, vision, and walking balance. First, to measure hearing sensitivity, we used custom software operable through a standard web browser. Calibrated pure tones at the frequencies of 0.25, 0.5, 1, 2, and 3 kHz were delivered through ER-2 (Etymotic Research Inc., Elk Grove Village, IL) audiometric earphones. Testing was performed in a clinical laboratory space that was quiet but not soundproof. Prior to performing the hearing sensitivity test, we inspected each participant's ear canals with an otoscope to ensure the ear canal was free of excessive cerumen. Participants were excluded if the ear canal had excessive cerumen buildup. Next, we assessed visual acuity with visual correction using a Snellen eye chart. Finally, to assess walking balance, a licensed physical therapist administered the Functional Gait Assessment (FGA). The FGA is a 10-item test that scores participants' performance on a variety of walking tasks that challenge balance and stability. The FGA has been shown to have excellent reliability and construct validity in older adult populations (Walker et al., 2007; Wrisley and Kumar, 2010; Leddy et al., 2011).

Next, we determined participants' preferred treadmill walking speeds using a staircase method of increasing and decreasing the treadmill speed until a desired speed was verbally confirmed by the participant. Participants walked for two additional minutes at the preferred speed to familiarize them with the setup. All future treadmill walking trials were performed at participants' preferred speed.

Participants then performed 6 or 10 individual 3-min treadmill walking trials. Participants were given rest breaks between trials as needed. During the walking trials, we manipulated two variables: auditory feedback and arm swing. The order of all walking conditions was randomized using a custom MATLAB (Mathworks, Natick, MA) script. There were three auditory feedback conditions: **Baseline**, **Ear Plugs**, and **White Noise**. During the Baseline condition, participants were exposed to the normal ambient sounds of the laboratory. No external devices that could alter hearing were worn during the Baseline condition. During the Ear Plugs condition, participants wore a combination of 3M^{TM} E-A-R^{TM} Classic^{TM} Earplugs with a noise reduction rating (NRR) of 29 dB (3M, Maplewood, MN) and 3M^{TM} PELTOR^{TM} Earmuffs (3M, Maplewood, MN) with an NRR of 27 dB. During the White Noise condition, we created a continuous 75 dB SPL white noise auditory stimuli using Max 7 software (Cycle'74, Walnut, CA). Participants listened to the white noise via ER-2 Insert Earphones with 13 mm Eartips (Etymotic Research, Inc., Elk Grove Village, IL). The wires from the earphones worn in the White Noise condition were fixed to the participants' back to restrict their movement and then manually supported by a researcher positioned behind the study participant. The researcher maintained some slack in the earphone wire to reduce the likelihood that the wire position provided haptic feedback. We used these two methods of reducing auditory feedback based on previous research suggesting that headphones or ear plugs could create an occlusion effect that allows perception of footsteps through bone conduction hearing (Durgin and Pelah, 1999; Hamacher et al., 2018). We thus included a White Noise condition as an alternative method to reduce auditory feedback that would mask bone conduction hearing (Studebaker, 1962). A subset of participants also walked during two additional auditory conditions that included a tone panning between the left and right insert earphones at varying frequencies. These panning trials were not included in the current analysis.

Participants repeated each auditory condition during two arm swing conditions: **Arms Free** and **Arms Crossed**. During the Arms Free condition, participants were instructed to let their arms swing naturally. During the Arms Crossed condition, participants folded their arms across their chest. We included these variations in arm swing due to concerns that the effect of changing auditory feedback on gait might be difficult to detect in a homogenous treadmill environment that provides minimal external challenges to balance. Some research has suggested that arm swing may assist in stabilizing gait (Ortega et al., 2008; Delabastita et al., 2016). We theorized that restricting arm swing in older adults may increase the challenge of controlling walking and in turn increase the role of auditory feedback (i.e., reliance on auditory feedback to control COM motion will be greater when the passively stabilizing effect of arm swing is removed). Thus, we included the Arms Crossed condition in an effort to augment the potential impact of the different auditory conditions on gait. However, it should be noted that other research has found conflicting results and suggests that restricting arm swing may not have significant effects on gait stability (Bruijn et al., 2010).

Data Analysis

We processed kinematic marker data using both Visual3D (C-Motion, Germantown, MD) and custom MATLAB scripts. The data were gap-filled (third-order polynomial with a maximum gap of 10 frames) and low-pass filtered (fourth-order Butterworth with a cut-off frequency of 6 Hz). The time of heel strike (HS) and toe-off (TO) events were then identified for each step based on the vertical position of the calcaneus marker, and the fore-aft position of the 2nd metatarsal marker, respectively (Wu et al., 2017, 2019). We visually inspected all gait events to ensure accuracy. COM lateral position was estimated as the midpoint of the lateral positions of the two greater trochanter markers. We then calculated lateral COM velocity as the derivative of COM position.

To assess changes in control of temporal aspects of gait, we calculated means and variabilities of individuals' step times and step lengths for each walking condition. To assess changes in control of spatial aspects of gait, we examined local stability of lateral COM velocity and calculated mean values of the minimum lateral margin of stability (MOS) per step, peak lateral COM speed per stride, lateral COM excursion per stride, and step width per step for each condition. We focused our analysis on variables related to controlling the lateral movements of the COM based on previous research suggesting that walking requires active control to maintain stability in the frontal plane (MacKinnon and Winter, 1993; Bauby and Kuo, 2000).

We calculated step time as the time between the HS of one foot and the following HS of the contralateral foot. Step length was calculated as the fore-aft distance between the calcanei markers at HS. Step width was calculated as the medio-lateral distance between the lateral malleoli markers at mid-stance.

For our metric of gait stability, we calculated local stability of lateral COM velocity using the maximum Lyapunov exponent. We chose lateral COM velocity because velocity, in contrast to position data, is not affected by non-stationarities. The Lyapunov exponent quantifies the average logarithmic rate of divergence of a system after a small perturbation (Rosenstein et al., 1993; Dingwell and Cusumano, 2000). The short-term local divergence exponent (λs) has demonstrated theoretical and predictive validity in simulation and empirical studies of walking (Bruijn et al., 2013). We calculated λs from the last 236 steps per trial. The number of steps analyzed was selected to maximize a consistent number of steps across all participants and trials. For construction of a state space, we used a time delay of 10 samples and an embedding dimension of 5. Within this state space, nearest neighbors were identified, and their divergence tracked. From these distances a log (divergence) curve was calculated, and the local divergence exponent was calculated as the slope of this curve between 0 and 0.5 strides (Stenum et al., 2014).

We calculated the minimum lateral MOS per step as the distance between the lateral extrapolated center of mass position (XCOM), a velocity-weighted lateral COM position, and the base of support (Hof et al., 2005) defined by the lateral position of the 5th metatarsal marker during stance phase. The methods used to calculate minimum lateral MOS have been previously described in detail (Wu et al., 2015). This simple inverted pendulum model of walking provides insight into how individuals control frontal-plane motions during walking. When the XCOM is within the BOS, the inverted pendulum will passively self-stabilize. When the XCOM position exceeds the BOS, corrective actions will be required to remain upright. In addition, the impulse required to move the XCOM beyond the BOS will be proportional to the magnitude of the MOS (Hof et al., 2005).

Statistical Analysis

To investigate how auditory feedback affects gait, we performed separate two-way repeated measures ANOVAs to test for differences in step time, step length, λs, minimum lateral MOS, COM excursion, peak lateral COM speed, and step width between walking trials. The two independent variables were *auditory feedback*, which had three levels—Baseline, Ear Plugs, and White Noise—and *arm swing*, which had two levels—Arms Free and Arms Crossed. We checked each variable for sphericity using Mauchly's sphericity test. If sphericity was violated, we used the Greenhouse-Geisser (GG) correction and p-value to test the main effect of auditory feedback and interaction effect of arm swing. When a significant main effect of auditory feedback was found, Bonferroni-corrected pairwise comparisons were performed to look for differences between the Baseline condition and the other two conditions. When a significant main effect of arm swing was found, a t-test was performed to compare Arms Free and Arms Crossed. When a significant interaction of auditory feedback and arm swing was found, simple effects analysis was conducted (i.e., a Bonferroni-corrected t-test compared Baseline vs. the other two auditory feedback conditions within each arm swing condition). For any measures with distributions that could not be determined as normal, a Friedman's two-way test substituted for the repeated measures ANOVA. The resultant chi-square and p-value were recorded. Significance was set at the $p < 0.05$ level for the repeated measures ANOVAs, pairwise comparisons, and t-tests.

RESULTS

Participants

Of the 22 enrolled participants, 20 completed the study. Two participants were excluded due to significant osteoarthritis with apparent gait impairments ($n = 1$) and excessive cerumen in the ear canal making it unsafe to perform audiometry ($n = 1$). Participants completing the study were 76.9 $\pm$ 6.4 years of age (mean $\pm$ standard deviation), 7 male/13 female, and had FGA scores of 25.7 $\pm$ 2.5. Average performance on the FGA exceeded the established cutoff score of 22/30 for predicting unexplained falls in community-dwelling older adults (Wrisley and Kumar, 2010). Hearing thresholds were recorded at 0.25, 0.5, 1, 2, and 3 kHz. Results of the audiometry testing found participants had hearing thresholds with an average score of 44.9, 39.4, 28.1, 33.8, and 37.4 dB SPL at 0.25, 0.5, 1, 2, and 3 kHz, respectively. Thresholds at 0.25 and 0.5 kHz may have been adulterated due to environmental noise interference. The average hearing thresholds between 1 and 3 kHz were consistent with a gently sloping mild hearing loss. The participants with the worst hearing thresholds exhibited moderate hearing loss. Of the participants completing the full study, all were able to successfully perform the walking tasks without assistance or incidence of falls. Full demographic data can be found in the data repository (https://digitalhub.northwestern.edu/collections/a0deab15-7c16-4c52-86f8-80c96a2fb888).

Temporal Metrics

We found no significant differences in step time variability (repeated measures ANOVA; $p = 0.688$) (**Figure 2A**) or step length variability (Chi-squared; $p = 0.367$) (**Figure 2B**) between auditory feedback conditions.

Mean step time and mean step length were both affected by auditory feedback. We found a significant main effect of auditory feedback (repeated measures ANOVA; $p = 0.032$) on step time (**Figure 3A**). *Post hoc* testing identified that step time was significantly longer (Bonferroni-corrected t-test; $p = 0.026$) for the Ear Plugs condition than the Baseline condition.

We also found a significant main effect of auditory feedback on step length (repeated measures ANOVA; $p = 0.037$). *Post hoc* testing identified that participants took significantly longer steps during the Ear Plugs condition than the Baseline condition (Bonferroni-corrected t-test; $p = 0.047$) (**Figure 3B**). There was also a significant interaction effect between auditory feedback and arm swing for step length (repeated measures ANOVA; $p = 0.036$). *Post hoc* testing identified that within the Arms Free condition, participants took significantly longer steps during the Ear Plugs condition than the Baseline condition (Bonferroni-corrected t-test; $p = 0.031$).

To further examine the changes in step length observed between the Baseline and Ear Plugs condition, we calculated a Pearson's correlation between the changes in step length (Ear Plugs—Baseline condition) and individuals' walking balance as assessed by the FGA (**Figure 4**). We found a moderate negative relationship that approached significance (rho $= -0.44$; $p = 0.055$), indicating that individuals with the poorest balance increased step length the most during the Ear Plugs condition.

Spatial Metrics

We found no significant main effect of auditory feedback on any of the five spatial metrics analyzed; λs (repeated measures ANOVA; GG $p = 0.056$), minimum lateral MOS, lateral COM excursion, peak lateral COM speed (repeated measures ANOVAs; all $p > 0.05$), and step width (Chi-Squared; $p = 0.535$) (**Figure 5**).

We found a significant main effect of arm swing on λs (repeated measures ANOVA; $p < 0.0005$) and COM excursion (repeated measures ANOVA; $p = 0.016$) (**Figure 5**). *Post hoc* testing identified that participants were significantly less stable (t-test; $p < 0.0005$) and had larger COM excursions (t-test; $p = 0.016$) during the Arms Crossed than the Arms Free conditions.

We found significant interaction effects between auditory feedback and arm swing for COM excursion (repeated measures ANOVA; $p = 0.005$) and COM speed (repeated measures ANOVA; $p = 0.024$). However, *post hoc* testing found no significant differences in COM excursion between auditory conditions within either of the arm swing conditions (Bonferroni-corrected t-test; $p > 0.05$). Within the Arms Crossed condition, COM speed was significantly larger during Baseline than the Ear Plugs condition (Bonferroni-corrected t-test; $p = 0.031$).

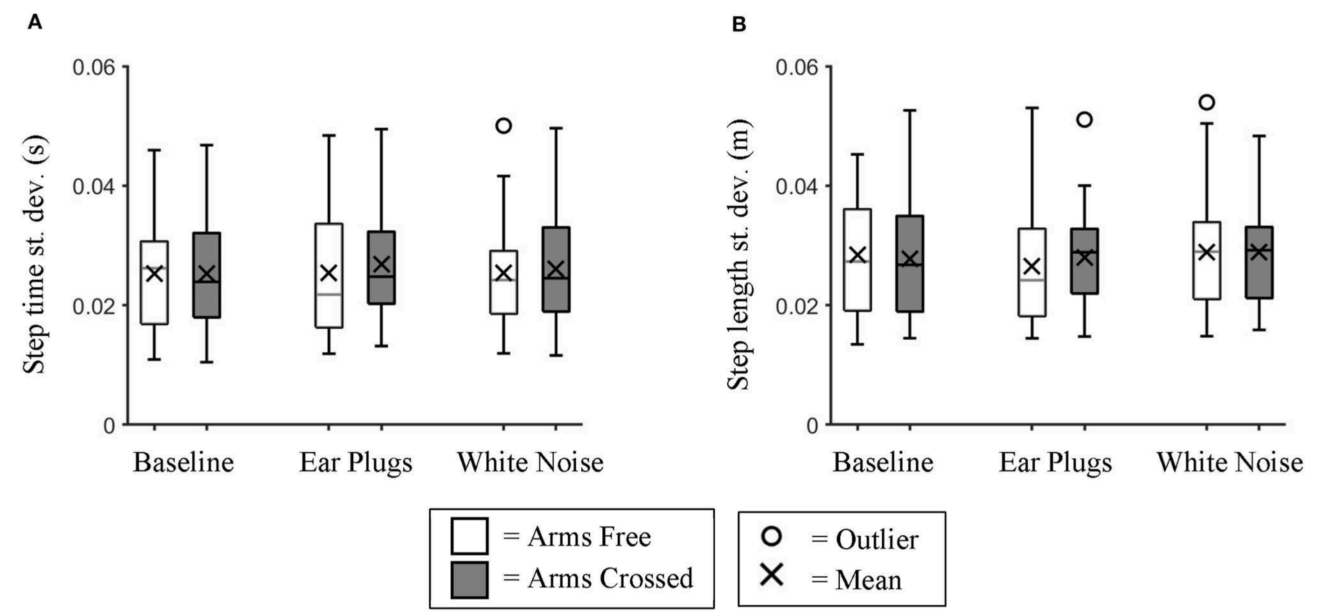

FIGURE 2 | Variability of Step time and Step Length. Box plot data for all participants showing the changes in **(A)** step time variability and **(B)** step length variability between auditory feedback and arm swing conditions. We found no significant differences in variability for either variable between conditions.

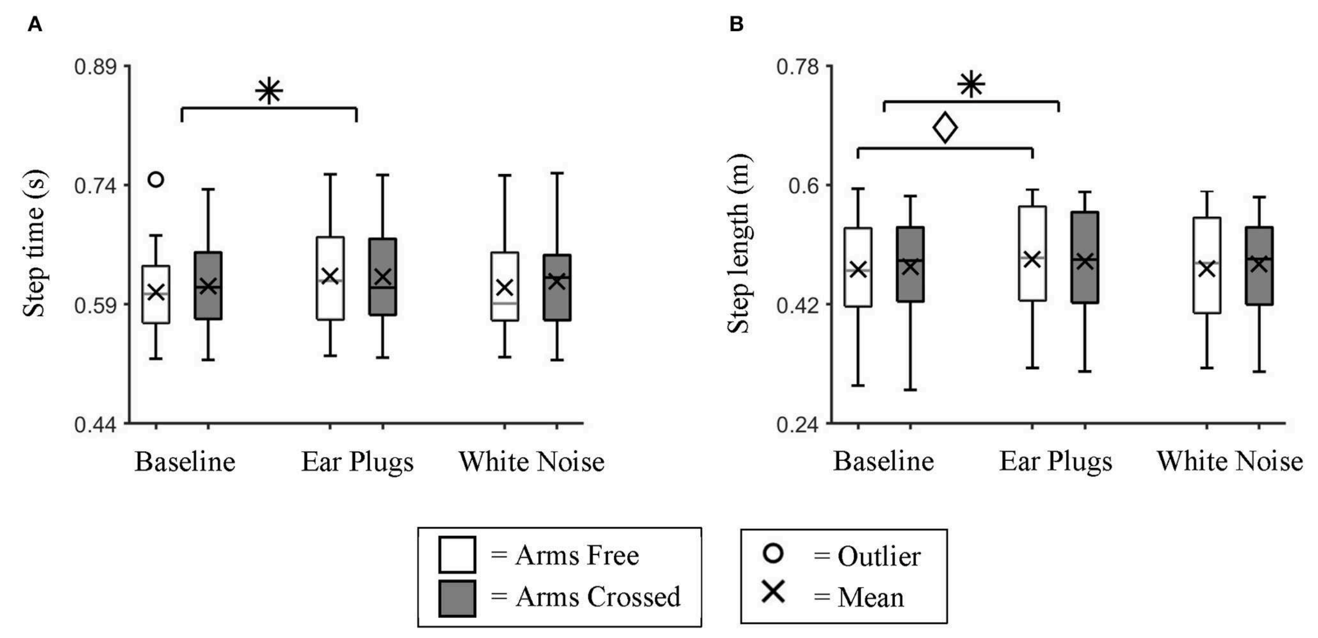

FIGURE 3 | Mean Step time and Step Length. Box plot data for all participants showing the changes in **(A)** mean step time and **(B)** mean step length between auditory feedback and arm swing conditions. We found individuals took significantly slower and longer steps during the Ear Plugs condition than during Baseline. * indicates a significant main effect ($p < 0.05$) between auditory feedback conditions. ◇ indicates a significant simple effect ($p < 0.05$) between auditory feedback conditions during Arms Free walking trials.

DISCUSSION

We examined the impact of auditory feedback on control of temporal and spatial aspects of walking in older adults. We hypothesized that older adults would exhibit more variable step timing and adapt shorter, faster steps during treadmill walking when auditory feedback was reduced with either ear plugs or

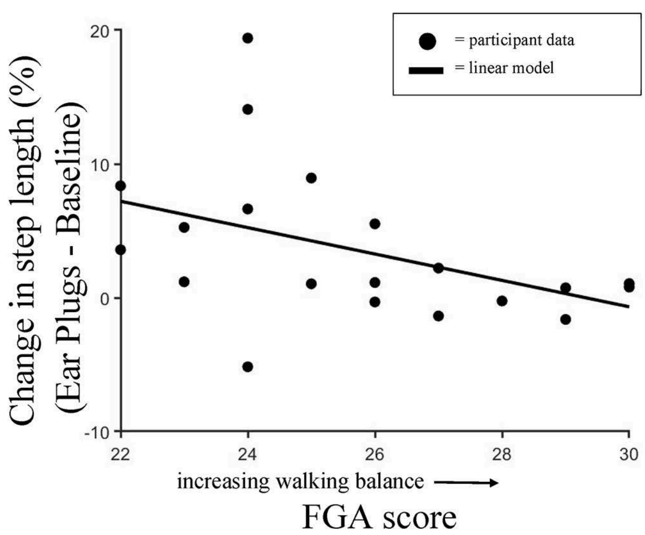

FIGURE 4 | Change in Step Length vs. FGA score. Pearson's correlation showing individual changes in step length observed between the Baseline and Ear Plugs condition and walking balance as assessed by the Functional Gait Assessment score. We found a moderate negative relationship that approached significance (rho = -0.44; $p = 0.055$), indicating that individuals with the poorest balance increased step length the most during the Ear Plugs condition.

white noise. These hypotheses regarding the temporal aspects of gait were not supported. Step time and length variabilities were not affected by changes in auditory feedback, and contrary to our hypothesis, individuals took longer steps when wearing ear plugs. We also hypothesized that spatial control of lateral COM motions during walking would be negatively affected by reductions in auditory feedback. This hypothesis was also not supported. We observed no significant main effects of auditory feedback on lateral COM motion.

Temporal Control

Sounds created by our footsteps can provide information about temporal aspects of walking. Previous research has found that people will modify their walking speed when the sounds of their own footsteps are artificially delayed (Menzer et al., 2010) or altered to resemble different terrains (Turchet et al., 2013). In response to clear auditory feedback of one's own footsteps, individuals with cerebral palsy (Baram and Lenger, 2012), multiple sclerosis (Baram and Miller, 2006), and Parkinson's disease (Baram et al., 2016) adapt longer strides and faster walking speeds. Augmenting footstep sounds has also been found to reduce step length variability in individuals with Parkinson's disease (Rodger et al., 2014). Collectively, these previous findings suggest that people use auditory feedback associated with footsteps to control temporal aspects of walking. In the current study, we sought to identify if the removal of auditory feedback during walking would directly affect temporal aspects of walking in older adults.

We hypothesized that reduction of auditory feedback would result in greater step-to-step variations in step timing. This result was not supported by our findings. Neither step time variability nor step length variability changed when auditory feedback was reduced during the Ear Plugs and White Noise conditions. Our findings are similar to the results of Weaver et al. who also failed to identify differences in stride length variability when bilateral hearing aid or cochlear implant users walked with and without their hearing devices (Weaver et al., 2017). These findings suggest that auditory feedback may have a minimal contribution to ongoing control of steady-state walking. Auditory feedback may not have been necessary for controlling step variability because the nervous system receives multiple streams of sensory information, such as proprioceptive feedback received when the foot contacts the ground each step (Pang and Yang, 2000; Gordon et al., 2009), that can provide redundant temporal information to footstep sounds. It is also possible that if individuals in the current experiment had walked in environments with greater challenges to walking (e.g., uneven surfaces) that the role of auditory feedback for controlling temporal aspects of walking would have been greater. A limitation of the current study was that we examined step time variability during treadmill walking, which is known to reduce gait variability (Hollman et al., 2016).

We also hypothesized that when auditory feedback was reduced, individuals would adapt shorter and faster steps. Adapting shorter and faster steps is a common response when people walk in balance-challenging and unpredictable environments (McAndrew et al., 2010; Wu et al., 2017). This strategy can improve an individual's ability to correct movement errors during walking by reducing their COM velocities and increasing the opportunity to make corrective steps (Reimann et al., 2018). We anticipated that individuals would adapt shorter, faster steps based on the assumption that removal of auditory feedback would hinder the nervous system's capacity to accurately sense ongoing walking dynamics, which would result in greater movement uncertainty. Results of this study did not support this hypothesis. To our surprise, compared to Baseline, individuals adapted longer steps during the Ear Plugs condition. Not only did individuals increase step length, but the participants with the poorest balance made the greatest increases in step length when using ear plugs. Of note, reduction of auditory feedback using white noise did not result in significant modifications to step lengths when compared to Baseline. Our finding was similar to the results of Hamacher et al. who observed that older adults wearing noise-canceling headphones actually improved local gait stability in comparison to normal walking (Hamacher et al., 2018). The authors suggested that this unexpected finding could be a result of the noise-canceling headphones creating an occlusion effect that may have improved individuals' ability to perceive their own footsteps or that the headphones reduced cognitive load. We believe that an occlusion effect may best explain our finding that participants adopted longer steps during the Ear Plugs condition but not during the White Noise condition.

It is recognized that when wearing ear plugs or inserts, the sound of an individual's own footsteps often appears amplified. This phenomenon has been explained by an occlusion effect; small vibrations of the ear plugs are transmitted to the basilar membrane, where they are received in a manner similar to

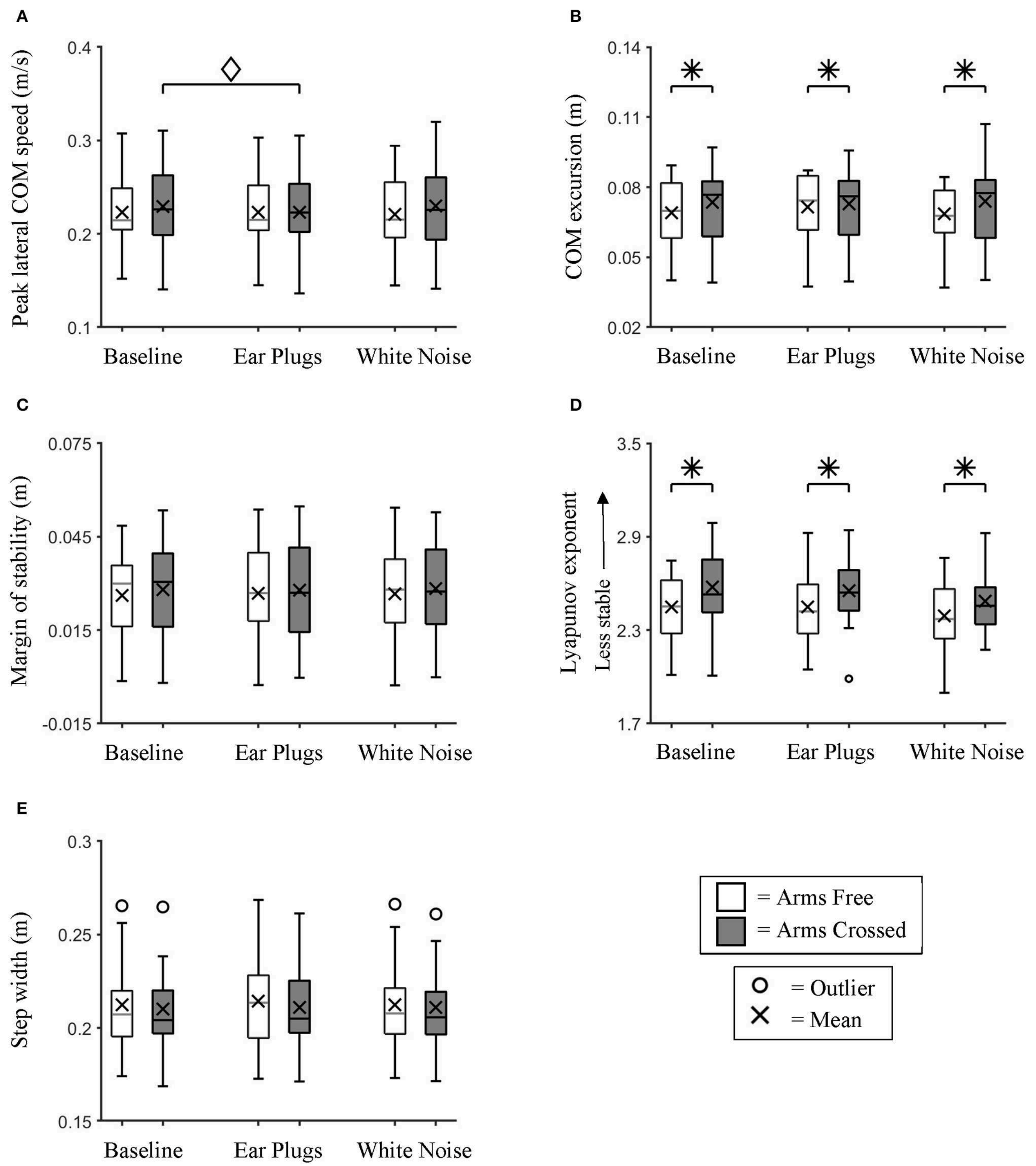

FIGURE 5 | Lateral Center of Mass Dynamics. Box plot data for all participants showing the changes in **(A)** Peak lateral COM speed, **(B)** lateral COM excursion, **(C)** minimum lateral margin of stability, **(D)** λs, and **(E)** step width between auditory feedback and arm swing conditions. We found no significant main effects of auditory feedback on control of lateral COM dynamics. * indicates a significant main effect ($p < 0.05$) between arm swing conditions. ◇ indicates a significant simple effect ($p < 0.05$) between auditory feedback conditions during the Arms Crossed walking trials.

air conduction hearing, via either bone conduction or pressure changes in the ear canal (Stenfelt and Goode, 2005; Stenfelt and Reinfeldt, 2007). During walking, the vibratory excitation of the ear plug is a result of the ground reaction force created when the foot contacts the ground. People can modify the profile of their ground reaction force by modifying their speed or walking style (Galbraith and Barton, 1970; Ekimov and Sabatier, 2006), which should change the perception of footsteps via the occlusion effect. Taking longer steps should increase the magnitude of the anterior-posterior component of the ground reaction force (Martin and Marsh, 1992), which we speculate should amplify the occlusion effect experienced during the Ear Plugs condition. Unfortunately, we were not able to quantify the magnitude of the ground reaction forces or the occlusion effect during the walking trials. While we observed this increase in step length during the Ear Plugs condition, we did not during the White Noise condition. One possible explanation is that the white noise masked any occlusion effect (Studebaker, 1962) created by wearing the insert earphones. Therefore, future studies should focus on auditory manipulation methods employing air conduction hearing, such as playing white noise, to reduce the effects of bone conduction hearing. In addition, future studies should also consider implementing methods to directly assess participants' perception of their footsteps in order to evaluate the efficacy of the intended auditory manipulation.

Cognitive load could have played a role in the manipulated auditory feedback conditions. Executive function and attention are known to affect walking performance (Yogev-Seligmann et al., 2008). Among older adults, the addition of cognitive loads can result in decreases in gait stability and balance (LaRoche et al., 2014). Past research suggests that noise can have variable effects on performance of different tasks (Dalton and Behm, 2007). It is possible that the manipulated auditory feedback conditions may have affected attention. However, in the current study, we did not explicitly assess the effects of the auditory conditions on cognitive load, making it difficult to speculate on any potential interactions between auditory feedback conditions, cognitive load and walking performance. The effects of cognitive load should be considered in future studies.

Spatial Control

Several studies have observed that external environmental sounds can act as a reference for controlling aspects of walking such as body posture and orientation (Karim et al., 2018) and can improve standing postural balance (Easton et al., 1998; Deviterne et al., 2005; Kanegaonkar and Amin, 2012; Gandemer et al., 2014; Rumalla et al., 2015; Horowitz et al., 2019). A considerable body of research has suggested that during walking, control of frontal-plane motion requires greater involvement of the nervous system than sagittal-plane motions (MacKinnon and Winter, 1993; Bauby and Kuo, 2000; O'Connor and Kuo, 2009; McAndrew et al., 2010). As such, we hypothesized that reductions in auditory feedback that may provide spatial information would result in lateral COM excursions that are larger, faster, and less stable. This hypothesis was not supported. We observed no significant differences in lateral COM excursion, peak lateral COM speed, minimum lateral MOS, λs, or step width between auditory feedback conditions. Our findings suggest that continuous auditory feedback may have a limited role in controlling lateral COM dynamics in older adults during a steady-state walking task.

There are several possible reasons why we did not observe changes in control of lateral COM dynamics between auditory conditions. First, although a preferable setup for evaluating steady-state walking, the treadmill environment imposes more restrictions than normal over-ground walking, including an invariable speed, which may have limited our ability to detect gait changes. Additionally, the task may not have sufficiently challenged control of walking balance. To address this issue and introduce a greater challenge to controlling COM dynamics, we added an Arms Crossed condition. While two of our measures, COM excursion and λs, were sensitive enough to detect significant differences between the Arms Crossed and Arms Free conditions, this added challenge was insufficient to detect differences in control between the auditory feedback conditions. It is also possible that other variables, such as body orientation (Karim et al., 2018), may have been more sensitive to the effects of auditory feedback than the measures we selected. We did not measure changes in body orientation because the moving treadmill belt forces individuals to maintain a relatively straight-ahead walking trajectory. Finally, as past research has suggested (Karim et al., 2018), it is possible that the contribution of auditory feedback for controlling walking is increased in situations in which other sensory information is reduced, such as walking in low-light conditions. Future research examining over-ground walking in situations of reduced visibility and auditory feedback would address the limitations of the current study.

Clinical Implications

Although past research has found a positive relationship between hearing loss and falls (Viljanen et al., 2009b; Lin and Ferrucci, 2012; Girard et al., 2014), the results of the current study do not indicate that reductions in auditory feedback have a significant impact on the ability of healthy older adults to control steady-state walking in a low-complexity environment. It is possible that the role of auditory feedback may take on greater importance for controlling walking in more challenging environments; that is to say, auditory feedback may be more important when responding to discrete challenges, such as detecting a change in walking surface, or in situations when other forms of sensory feedback are reduced. However, our finding that older adults increased step length when walking with ear plugs—which may have facilitated perception of footsteps via an occlusion effect—suggests that older adults may be able to utilize enhanced auditory feedback to improve control of walking balance. This finding aligns with results of several studies that similarly found enhanced auditory feedback of one's own footsteps can improve walking speed, stride length, and gait stability in clinical and older adult populations (Baram and Miller, 2006; Baram and Lenger, 2012; Baram et al., 2016; Hamacher et al., 2018). In addition, the trend that individuals with the poorest walking balance made the greatest increases in step length during the Ear Plugs condition may suggest that the temporal feedback received from footstep sounds is a strategy for controlling balance. If so, gait training performed with enhanced audio feedback of one's own footsteps could be used to improve walking balance of older adults.

Conclusions

Overall, our results suggest that during a steady-state walking task, healthy older adults are able to maintain walking control without auditory feedback. However, increases in step length observed during the Ear Plugs condition suggest that temporal auditory cues associated with footsteps may provide feedback for controlling walking balance that may be increasingly valuable as balance declines with age.

DATA AVAILABILITY STATEMENT

The data analyzed for this study can be found at the Northwestern University Feinberg School of Medicine Digital Hub (https://digitalhub.northwestern.edu/collections/a0deab15-7c16-4c52-86f8-80c96a2fb888).

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Northwestern University Institutional Review Board. The participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MW, PS, JS, SD, and KG conceived of and designed the study. JW recruited participants and performed clinical assessments. TC, JW, MW, and BJ collected and analyzed. All authors were involved in interpretation of data. TC, JW, and KG drafted the manuscript. All authors participated in revising the manuscript.

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Balance Training in Older Adults Using Exergames: Game Speed and Cognitive Elements Affect How Seniors Play

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Falls in older adults are a serious threat to their health and independence, and a prominent reason for institutionalization. Incorrect weight shifts and poor executive functioning have been identified as important causes for falling. Exergames are increasingly used to train both balance and executive functions in older adults, but it is unknown how game characteristics affect the movements of older adults during exergaming. The aim of this study was to investigate how two key game elements, game speed, and the presence of obstacles, influence movement characteristics in older adults playing a balance training exergame. Fifteen older adults (74 ± 4.4 years) played a step-based balance training exergame, designed specifically for seniors to elicit weight shifts and arm stretches. The task consisted of moving sideways to catch falling grapes and avoid obstacles (falling branches), and of raising the arms to catch stationary chickens that appeared above the avatar. No steps in anterior-posterior direction were required in the game. Participants played the game for eight 2 min trials in total, at two speed settings and with or without obstacles, in a counterbalanced order across participants. A 3D motion capture system was used to capture position data of 22 markers fixed to upper and lower body. Calculated variables included step size, step frequency, single leg support, arm lift frequency, and horizontal trunk displacement. Increased game speed resulted in a decrease in mean single support time, step size, and arm lift frequency, and an increase in cadence, game score, and number of error messages. The presence of obstacles resulted in a decrease in single support ratio, step size, cadence, frequency of arm lifts, and game score. In addition, step size increased from the first to the second trial repetition. These results show that both game speed and the presence of obstacles influence players' movement characteristics, but only some of these effects are considered beneficial for balance training whereas others are detrimental. These findings underscore that an informed approach is necessary when designing exergames so that game settings contribute to rather than hinder eliciting the required movements for effective balance training.

Keywords: balance training, older adults, exergaming, movement characteristics, game settings

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INTRODUCTION

The rapidly aging population in industrialized countries and concurrent strains on the health care system necessitate more cost-effective treatment and prevention options to counteract age-related decline in functioning. Falls in older adults are among the main causes for hospitalization and institutionalization (Kannus et al., 2005) and significantly impact the cost burden on health care budgets worldwide (Heinrich et al., 2010). Furthermore, both actual falls and fear of falling are associated with reduced activity (Yardley and Smith, 2002; Hornyak et al., 2013), which in turn increases the risk for developing chronic conditions.

There is good evidence that balance training reduces fall risk (e.g., Buchner et al., 1997) and can counteract inactivity caused by fear of falling (Gusi et al., 2012). A cost-effective way to administer additional balance training with good adherence rates is exergaming (Burke et al., 2009; Skjæret et al., 2016). Exergames are videogames that require bodily movements as input to play the game (Brox et al., 2011). The term is used both for vigorous exercises and for less intensive exercises such as balance training or seated upper body exercises. Furthermore, exergames allow for home-based training in elderly, as demonstrated in early studies using force-sensitive matbased stepping exergames (Schoene et al., 2011; Smith et al., 2011). In recent years, exergames have gained popularity both as a complementary tool or as a replacement for traditional exercise and rehabilitation (e.g., Mellecker et al., 2013), with the same or better effectiveness compared to usual care (Skjæret et al., 2016). In addition, exergames have been shown to have positive effects on physical activity in general (e.g., Höchsmann et al., 2016; Rhodes et al., 2017), as well as in specific rehabilitation settings (e.g., Laver et al., 2012; Baltaci et al., 2013). Ongoing developments in game technology, such as videobased motion detection of the player, allow for exergames with more variation in movements compared to mat-based exergame systems.

Correctly executed steps are important to maintain balance and to avoid falls (Robinovitch et al., 2013). In daily life, one is often required to make quick, unanticipated steps to react to changing circumstances in order to avoid a fall (Lord and Fitzpatrick, 2001). This requires both the mental and physical capacity to react to an unknown situation. Caetano et al. (2016) showed that the ability to adapt gait is an important factor in predicting fall risk. Furthermore, they showed that concurrent cognitive tasks resulted in older adults reducing their walking speed and shortening their steps during the stepping and avoidance paradigm used in their studies (Caetano et al., 2017, 2018).

Stepping exergames are well suited for creating artificial situations in which unplanned steps are needed and have been shown to reliably assess fall risk in community-dwelling older adults (Schoene et al., 2011). Furthermore, a recent systematic review of randomized control trials about the effect of balance exergames in older adults (Fang et al., 2020) found improvements in functional performance and balance confidence with respect to dynamic balance, perceived balance, chair stand test, and balance test batteries. Non-significant improvements in static balance and proactive balance were speculated to be caused by ceiling effects in the tests used.

Despite the good evidence for the use of stepping exergames to train balance, there is an overall lack of knowledge on the actual movement characteristics that are elicited by exergames, both in elderly players and in other populations. This knowledge is crucial for the development of evidence-based, targeted exergames if the intention is to provide effective training and rehabilitation in the aging population. This is even more critical when exergames are to be used unsupervised in home-based training. To date, most of the research has focused on the effectiveness of, and adherence to, training programs using exergames compared to usual care, not on whether the intended movements are actually elicited. Without knowledge about the movement characteristics elicited during gameplay, it is difficult to design effective balance training exergames or interpret the effects, or lack thereof, of exergaming interventions in clinical trials.

In order to design effective exergames that achieve high adherence rates over longer periods of time, expertise from multiple disciplines is necessary. On the one hand, health care professionals need to contribute with their knowledge about which exercises and movements are required to train specific functions. On the other hand, the expertise of programmers and video game designers is crucial, both for the technical aspects and to ensure that the exergames are enticing and inherently motivating over longer periods of time. Furthermore, to assess whether the intended movements are indeed elicited during gameplay, the expertise of human movement scientists is required. Such multidisciplinary effort at the intersection of health, movement science, and computer science becomes even more crucial given the lack of knowledge about movement characteristics of elderly exergame users that are elicited by different settings and potential additional cognitive challenges within the exergame.

A critical element for training and rehabilitation programs to be effective is adherence to the program. A common strategy to achieve high adherence during exergaming is to try to keep the player in a so-called flow-zone (Csikszentmihalyi, 1975). In the flow-zone, a player is neither overchallenged nor bored by the exergame. One way to achieve this is by dynamically adjusting the game settings during the game to match the performance of the player. For example, game speed can be changed to adjust the difficulty level of the exergame, thereby personalizing the exergame to the player's abilities and enabling progression (e.g., van Diest et al., 2013). However, very little is known about how a change in game speed affects the movement characteristics of the player, and whether these effects are beneficial or detrimental for the desired training or rehabilitation effect. For example, if a higher game speed would lead to less carefully executed movements, as in a speed-accuracy trade-off (cf. Heitz, 2014), this might not be beneficial to achieve the desired training effects. A better understanding of the effects of game speed on elicited movement characteristics is thus necessary to inform the development of targeted exergames for balance training and prevention of functional decline.

Similarly, additional challenges in the form of cognitive elements are used often in exergames to keep the player in the flow-zone or to add cognitive training to the physical exercise (Anders et al., 2018). These additional cognitive elements can be in the form of extra tasks in the exergame such as counting, matching objects, or avoiding obstacles. Cognitive elements in exergames can create a dual task situation in which the player needs to focus on two or more things simultaneously, thereby training cognitive function as well as physical function. Evidence suggests that dual task training can improve walking performance of older adults (Wollesen et al., 2017). However, the effect of additional cognitive elements on the player's movement characteristics in exergames is rarely explored. In a rare exception, Skjæret-Maroni et al. (2016) found a decrease in the quality of movement characteristics needed to train balance when cognitive tasks were present during exergaming, underscoring the need for better knowledge about how game settings influence elicited movements during exergaming.

The objective of the current study is to address these gaps in our knowledge by investigating the effect of game speed and obstacles on movement characteristics in older adults playing a balance training exergame. As there is good evidence that step-based balance training reduces the risk of falls in older adults (e.g., Okubo et al., 2017), we chose a stepping exergame for balance training and assessed the movement characteristics of older adults playing at two different speed settings and either with or without obstacles to avoid. To the best of our knowledge, there is no previous study that assessed the effects of game speed and obstacles in exergames on movement characteristics in older adults. This knowledge is crucial for both health care professionals and developers of exergames in order to choose between existing or design new exergames that are most beneficial for balance training.

We expected that both game speed and the presence of obstacles would influence movement characteristics, and that some of these effects would be beneficial for balance training but others detrimental. Furthermore, in exergames with higher cognitive load, we expected the participants to make smaller steps.

METHODS

Participants

Fifteen older adults between 65 and 83 years of age participated in an experimental laboratory study at the Norwegian University of Science and Technology (see **Table 1** for participant characteristics). To be included, participants had to be 65 years or older and live independently. Participants were excluded if they had an injury or had undergone surgery of the back or the lower extremities during the last 6 months, or if they were unable to follow instructions given by the researchers.

The protocol was approved by the Regional Committees for Medical and Health Research Ethics, Norway. All participants gave written informed consent before data collection in accordance with the Declaration of Helsinki.

TABLE 1 | Participant characteristics.

	Mean	SD	Range
Age (years)			
Female (*N* = 7)	72.6	4.7	65–79
Male (*N* = 8)	74.8	4.0	70–83
Total (*N* = 15)	73.7	4.4	65–83
Height (cm)			
Female	166.8	3.5	162.8–172.2
Male	175.2	6.1	165.8–183.5
Total	171.3	4.1	162.8–183.5
Weight (kg)			
Female	65.4	4.9	60–74.4
Male	76	4.1	70.2–82.8
Total	71	7	60–82.2

Procedure

All participants played a step-based balance training exergame ("The Fox," SilverfitBV, the Netherlands). The aim of the exergame was to feed a fox-avatar by moving it sideways into the trajectory of falling grapes. The avatar mirrored the lateral movements of the participants, so when the participants placed themselves to the far right of the exergaming area, the avatar would be on the far-right side of the screen. The exergaming area was set to 2 by 2 m as shown in **Figure 1**. Thus, the movement of the avatar from one side of the screen to the opposite side corresponded to a 2 m distance in the physical playing area. The game itself did not determine any required step size, as participants can take several smaller or fewer larger steps to move the avatar the desired distance. Occasionally appearing stationary chickens above the fox-avatar were caught by raising at least one arm, which triggered the fox to jump. Only arm movements that resulted in at least one hand being lifted higher than the head were accepted by the exergame as a trigger for a jump. Other arm movements, e.g. to maintain balance, were ignored by the game. Participants had a time window of 7 s to place themselves under a chicken and raise at least one arm to trigger a jump, before the prey disappeared again.

No movement in the anterior-posterior direction was required to play the exergame. The exergame system used a Microsoft Kinect v2 camera (Microsoft Corporation, Redmond, WA, USA) to capture the movements of the participants. The Kinect v2 was used as input for the exergame only and not for data capturing.

Each participant played eight 2-min exergame trials at two speed settings and either with or without additional obstacles, falling branches, that had to be avoided. The width of a falling branch can be seen in **Figure 1**. The distance between the grapevines on either side corresponded to a physical distance of 2 m, a branch covered approximately 57 cm (35%). Increasing the exergame speed led to an increased number of grapes and branches falling simultaneously on the screen (from 1–3 to 3–5 grapes, and from 1–2 to 1–3 branches) and reducing their time on the screen from 8–10 s to 6–9 s (measured from top to bottom of the screen for missed grapes and branches). Neither the number

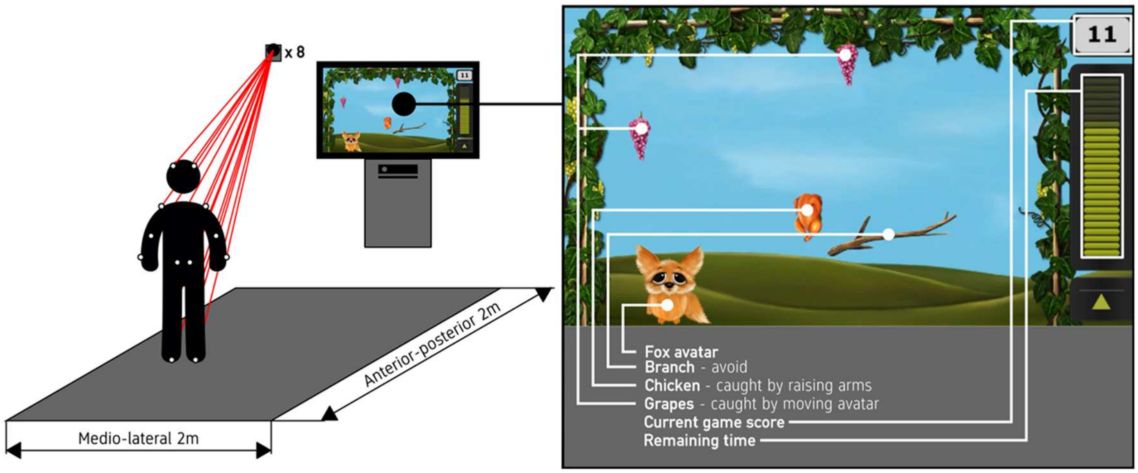

FIGURE 1 | Graphical image of the exergame area (left panel) and a screenshot of "The Fox" exergame (right panel) showing all exergame elements (Fox-avatar, grapes, chickens, branches, game score, and remaining game time).

of chickens nor their time on the screen were affected by changes of the game speed. The four conditions were played twice in counterbalanced order across participants. Participants had a 2-min break between exergame trials. Before the first trial, each participant tested the game at a lower speed setting to ensure that they understood the task and that they were able to perform all required movements. After the participants completed all eight trials, they performed a range-of-motion test to quantify that they were able to perform all required movements such as arm lifts and forward and sideways steps.

To record the participant's movements throughout the trials, 22 retroreflective markers were placed on anatomical landmarks using double-sided tape and a headband. Markers were placed bilaterally on the posterior and anterior side of the head, ulna styloid process, lateral epicondyle of the humerus, acromion, ilium posterior superior, ilium anterior superior, femur lateral epicondyle, lateral malleolus, and lateral distal phalange 1 (big toe), as well as one marker on the sternum and one on the center of the right thigh. A 3D motion capture system (OQUS, Qualisys, Gothenburg, Sweden) consisting of eight cameras was used to record the spatial positions of the markers throughout all exergame trials, with a measurement frequency of 120 Hz. The cameras were wall-mounted to minimize blockage by the TV screen or the Kinect v2 camera. Markers that fell off during a trial were replaced before the start of the next trial.

The output of the exergame on the 55" (139.7 cm) TV screen was captured using Open Broadcaster Software Studio (version 21.0; Open Broadcaster Software Studio, 2018). **Figure 1** shows an exemplary frame including all game elements. The game score displayed on the top right of the screen was explained to the participants, but they were not specifically encouraged to achieve the highest possible score.

If the participant moved outside the predefined exergaming area of 2 by 2 m, the game stopped and an error message appeared on the screen containing information on how to resolve the issue, e.g., "You are too far in the back. Please move closer to the screen." The exergame continued as soon as the participant returned to the exergaming area. The size of the exergaming area was not indicated on the floor in order to mimic a home-based setting.

Data Analysis

A custom Matlab script (version: 9.3.0, Mathworks Inc., Nantick, MA) was used to analyze the movement of the extremities as well as the position of the trunk based on the 3D motion capture data. The start and stop of each step was identified using the same method as in Skjæret-Maroni et al. (2016), which used the position of the markers on the lateral malleolus. A step was defined as a $\geq$ 0.03 m displacement of the toe marker lasting for $\geq$ 0.05 s. A marker velocity of 0 $\pm$ 0.1 ms^{-1} was used for the identification of step initiation and termination. From these, mean duration of single leg support, mean step size, and cadence were calculated for each foot. We chose the term step size rather than step length or step width to capture both sideways and forward aspects of steps, as some participants partly rotated their upper body whereas others remained parallel to the screen while taking steps. Furthermore, the ratio of the total duration of single-leg support (total accumulated single-leg support time divided by trial time) was calculated. Movement data for the arms was used to calculate the number of arm lifts per minute. The start and end of an arm lift was determined by the relationship between the vertical positions of the average height of all head markers compared to the left and right markers on the ulna styloid processes (wrists). The player's position on the playing field was determined by the position of the marker on the sternum. Displacements of the sternum marker were used to create heatmaps that reflected upper body positions of the participants in relation to the screen.

There were slight variations in the duration of each exergame across trials and participants due to error messages that appeared on the screen when a participant left the exergaming area, which temporarily stopped the game but not the 3D motion capture recording. Therefore, we report the ratio for single-leg support and the frequency of arm lifts rather than the total duration of single-leg support and the total number of arm lifts.

Finally, optical character recognition was used on the captured screen frames to identify error messages during each trial, as well as the total game score and the number of chickens caught and branches hit (if present) at the end of each trial.

Statistics

A linear mixed effect analysis in R (R Core Team, 2019) was used to analyze the effects of game speed, the presence or absence of obstacles, gender, and trial repetition, which served as fixed effects on the investigated movement characteristics, using the lme4 package (Bates et al., 2015). Random effects included intercepts for participants as well as by-participant random slope for the effect of body side (right or left arm or foot). Visual inspection of residual plots did not reveal deviations from normality or homoscedasticity. The computation of p-values was based on conditional F-tests with Kenward-Roger approximation for the degrees of freedom (Halekoh and Højsgaard, 2014). Statistical significance was set at $p < 0.05$. The linear mixed effect analysis was chosen to remove individual differences between participants (Schoene et al., 2011) and to account for the slight gender imbalance without losing statistical power.

RESULTS

Game Score

The game score that appeared on the screen after each trial was a combination of grapes and chickens caught and branches avoided (+ 1 point for each grape caught, + 3 points for each chicken caught, and −2 points for each branch hit). The number of grapes presented varied across trials, depending on the chosen game speed, thereby resulting in more opportunities for catching grapes in games with high game speed. In contrast, the number of chickens did not vary with game speed but was constant at 11 chickens per trial. The game score results are shown in **Table 2**. Not surprisingly, the mean game score was higher in the trials at high game speed without obstacles. The presence of obstacles reduced the game score by approximately 20 points for either game speed setting, while the game score was about 20 points higher in games at high speed compared to games at low speed. The percentage of grapes caught was ∼20% higher in low-speed compared to high-speed games. Interestingly, the percentage of grapes caught with or without obstacles was similar, with a small trend to increase when obstacles were present. In contrast, a clear decrease was seen in the percentage of chickens caught in games with obstacles present. Percentage of branches unsuccessfully avoided was very low and slightly higher in games with low speed settings.

Movement Characteristics

Single Leg Support

As standing on one leg is an important strategy to train balance, we investigated both the mean duration of single support events per foot (in seconds) and the ratio of single leg support (accumulated single support time during a trial divided by trial time). The mixed effects model indicated that the mean duration of single support events decreased significantly ($p < 0.001$) with an increase in game speed, whereas the ratio of single support increased slightly but not significantly ($p = 0.189$) (see **Figure 2** and **Table 3**). This indicates that participants stood slightly more often but significantly shorter on one foot when playing at higher speed. Furthermore, the ratio of single support decreased when obstacles were present, indicating that participants accumulated less single support time during a trial when having to avoid obstacles. Neither ratio nor duration of single support were significantly affected by trial repetition, gender, or body side (all p's > 0.10).

TABLE 2 | Mean game score and mean percentage of caught grapes, caught chickens, and unsuccessfully avoided branches for all combinations of game speeds and obstacles.

		Speed		
	Obstacles	*low*	*high*	**Overall**
Game score	*without*	69.8	86	**78.5**
	with	48.7	67.4	**58.3**
	Overall	**59**	**77.1**	**68.5**
Grapes caught	*without*	77.6%	53.7%	**64.8%**
	with	79.8%	59.1%	**69.3%**
	Overall	**78.7%**	**56.3%**	**67%**
Chickens caught	*without*	91.3%	93.1%	**92.3%**
	with	74.8%	76.1%	**75.5%**
	Overall	**82.9%**	**84.9%**	**83.9%**
Branches hit	*with*	5%	3.8%	**4.4%**

The values in bold font represent the average across game speed, presence of obstacles and overall.

Mean Step Size and Cadence

Step size and cadence where both affected significantly by the game speed (mean step size: $p < 0.05$; cadence: $p < 0.001$) and the presence of obstacles (mean step size: $p = 0.001$; cadence: $p < 0.001$). As can be seen in **Figure 3**, higher game speed and the presence of obstacles led to a decrease in mean step size (panel A). Cadence decreased as well when additional obstacles were present but increased with an increase in game speed (panel B). In addition, trial repetition had a significant effect on mean step size ($p < 0.001$), with larger steps when participants played the same condition for the second time. See **Table 3** for all statistical results for mean step size and cadence.

Frequency of Arm Lifts

The frequency of arm lifts decreased with high game speed ($p < 0.001$) and when additional obstacles were present ($p = 0.001$), as shown in **Figure 4**. Trial repetition, gender, and body side had no significant effects on the frequency of arm lifts (all p's > 0.08). See **Table 3** for all statistical results.

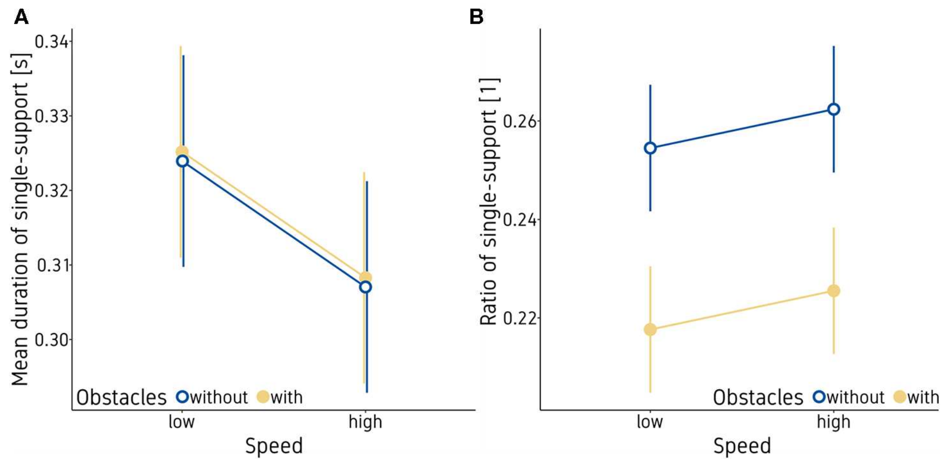

FIGURE 2 | Mean duration of single support events **(A)** and ratio of single support **(B)**. Vertical bars indicate the standard error of the model.

Player's Position

Heatmaps

As described above, the active area in which the exergame could be played was ∼2 by 2 m. The heatmap (see **Figure 5**) shows the number of observations of the sternum marker in the exergaming area (white square) across all participants and all trials, on a 30 by 30 grid. Warmer colors indicate more observations. The starting position was centered in front of the screen, 0 mm in medial-lateral and 500 mm in posterior direction (white circle in **Figure 5**). On average, participants drifted ∼0.5 m closer to the screen throughout an exergame trial, as indicated by the green and yellow squares, indicating more observations of the sternum marker parallel to the screen (white rectangle). There were no systematic differences in the heatmaps between the different conditions.

Moving Outside the Exergaming Area

When a participant moved outside the active exergame area of 2 by 2 m in any direction while playing, an error message would appear on the screen to inform the participant to correct their position in the playing area. We used a custom Matlab script including optical character recognition to detect error messages in the screen recording. Out of 15 participants, only two participants never drifted outside the active exergame area while playing the exergames. The remaining participants moved outside the area between one and nine times across all conditions. On average, 0.38 error messages per exergame trial were observed across all exergame trials that had valid screen captures (113 out of 120 trials). The majority of error messages were triggered by participants who drifted too close to the screen while playing (36 out of 43 error messages in total). The average time to resolve this was 2.44 s. The remaining error messages were divided between being too far to the left (4 times, on average 1.34 s to resolve), too far to the right (once, 0.67 s), and too far back (twice, 1.70 s).

More error messages were triggered in trials at high game speed compared to those at low game speed (see **Table 4**). However, a chi-square test indicated that the presence or absence of obstacles showed no consistent effect on the number of error messages triggered, nor was there a significant interaction between game speed and the presence of obstacles, $\chi^2_{(1)} = 1.203$, $p > 0.05$.

DISCUSSION

The purpose of this study was to investigate the effect of game speed and obstacles on movement characteristics of older adults who played a step-based balance training exergame. According to a meta-analysis by Sherrington et al. (2011), effective balance exercises should include displacement of the center of gravity and reduced base of support. Therefore, we assessed mean step size and cadence, the ratio and duration of single-leg support, as well as the frequency of arm lifts as arm lifts and stretches influence the center of gravity. Both changes in game speed and additional cognitive elements are used regularly in exergames to keep the player in the flow-zone and/or train cognitive and physical functions simultaneously. However, as the results of the current study show, changing the settings of an exergame can have both positive and negative effects on the player's movement characteristics with respect to training balance. Below, we discuss our main findings regarding the effects of game speed, the presence of obstacles, and trial repetition on movement characteristics, the consequences of these effects for balance training, and their relevance for choosing existing or developing new exergames for balance training in older adults.

Game Speed

Higher game speed led to shorter single support events, shorter steps, fewer arm lifts, and increased cadence. Higher cadence is associated with more frequent weight shifts, which are considered beneficial for effective balance training (Sherrington et al., 2011). Furthermore, eliciting faster steps during gaming at higher speed may benefit the training of a quick step to avoid, or recover from, a balance disturbance and imminent fall. On the other hand, participants were more likely to move outside

TABLE 3 | The degrees of freedom (df), confidence intervals (CI), and significance level (p) for the mean duration of single support events, ratio of single support, mean step length, cadence, and the frequency of arm lifts.

	Mean duration of single support (s)			Ratio of single support (1)			Mean step size (mm)			Cadence (min^{-1})			Frequency of arm lifts (min^{-1})		
	df	CI	p	df	CI	p	df	CI	p	df	CI	p	df	CI	p
Speed	229.07	−0.03 to −0.01	**<0.001**	229.09	−0.00 to 0.02	0.189	293.14	−13.87 to −0.45	**0.038**	229.07	5.22 to 8.56	**<0.001**	229.07	−2.27 to −0.99	**<0.001**
Obstacles	229.07	−0.01 to 0.01	0.774	229.09	−0.05 to −0.03	**<0.001**	−7.16	−18.54 to −5.12	**0.001**	229.07	−10.97 to −7.63	**<0.001**	229.07	−1.70 to −0.42	**0.001**
Trial	229.07	−0.01 to 0.01	0.582	229.09	−0.01 to 0.02	0.318	−11.83	5.78 to 19.20	**<0.001**	229.07	−0.80 to 2.55	0.306	229.07	−0.97 to 0.32	0.321
Gender	17.31	−0.05 to 0.06	0.909	17.29	−0.02 to 0.07	0.280	12.49	−42.63 to 3.39	0.113	17.31	−2.12 to 14.10	0.166	17.31	−5.64 to 1.94	0.352
Side	229.07	−0.00 to 0.02	0.102	229.09	−0.01 to 0.01	0.982	−19.62	−7.02 to 6.40	0.928	229.07	−0.64 to 2.70	0.229	229.07	−1.20 to 0.08	0.086

The values in bold font represent significant p values.

the active exergaming area when playing at higher speed, as indicated by more than twice as many error messages. This would cause the exergame to stop, thereby potentially interrupting the participant's flow. Further research is necessary to find an optimal balance between these contrasting effects of game speed when designing or choosing the most beneficial game and game settings for balance training. This is further attested to by our results regarding game score. As higher game speed led to higher scores, this may have important ramifications for exergames where achieving a high score is the focus for the player, as game speed affects the movement characteristics as well.

Obstacles

The presence of obstacles led to shorter steps, reduced cadence, fewer arm lifts, and a decrease in the ratio of single-support time. A likely explanation for these changes in movement characteristics is increased cognitive load caused by the presence of additional exergame elements and potentially conflicting demands on the player. These results indicate that although cognitive elements are often added to the game to increase enjoyment and/or simultaneously train cognitive functions, they can lead to unwanted side effects on elicited movement characteristics that are less favorable for balance training (see also Skjæret-Maroni et al., 2016).

Furthermore, we observed that when participants moved their avatar toward the edge of the screen, the avatar was occasionally trapped there by falling branches that blocked the path back to the middle or the other side of the screen. This resulted in participants waiting for the obstacle to pass before continuing to play the game and try to catch grapes and chickens. The waiting period in which the participants stood still could take up to several seconds. The low percentage of collisions with branches indicates that participants indeed tried to avoid being hit by branches as much as possible, even when that meant having to wait for the branch to pass before continuing to catch chickens and grapes. Although having to avoid obstacles may be positive in terms of adding cognitive training and enjoyment to the exergame, this may also lead to strategies and movements that are considered less beneficial for the training of balance. Waiting for a situation to resolve should be avoided by the game design, for example by allowing the avatar to move forward or backward around the obstacle. An alternative solution could be to introduce e.g., an extra bracing position to protect the avatar from falling branches which could simultaneously challenge balance. Our results underscore that the effects of additional cognitive elements on intended movement characteristics need to be taken into account when designing or using exergames for balance training.

Trial Repetition

Participants played each condition twice, but we did not expect to find learning or fatigue effects with such short exposure. Although most movement characteristics were indeed unaffected by the repetition, we did find larger steps on the second trial compared to the first. A possible explanation for this might be that in the second trial, participants were more familiar and comfortable with playing the exergame in general and

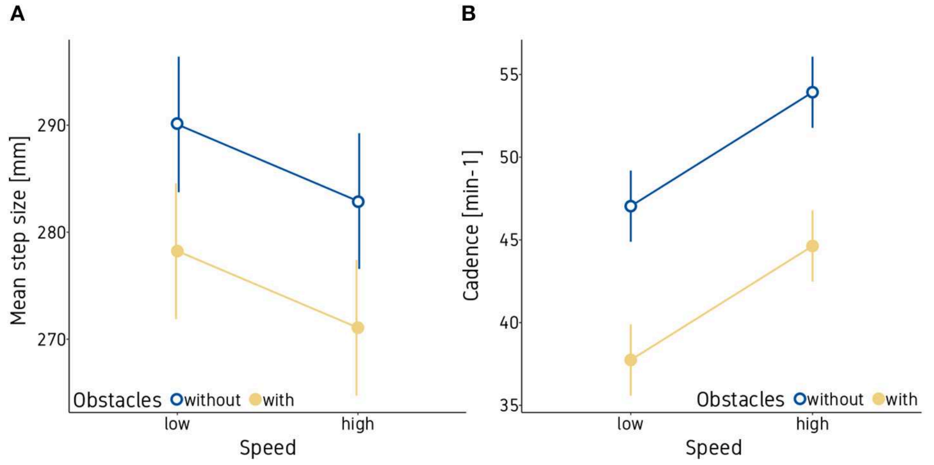

FIGURE 3 | Mean step length **(A)** and cadence **(B)**. Vertical bars indicate the standard error of the model.

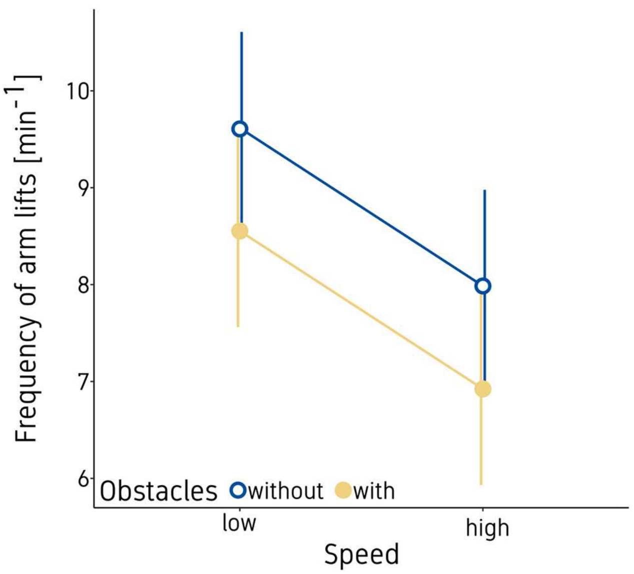

FIGURE 4 | Mean frequency of arm lifts. Vertical bars indicate the standard error of the model.

the movements required to play the exergame in particular. Pure sideways stepping without a forward component to the movement is less common in everyday life. Therefore, it may be speculated whether participants needed a short adaption phase in order to perform this movement with enough confidence to perform larger steps. In that respect, the observed increase in mean step size from trial one to trial two is a positive result, as stepping sideways can be an important strategy to recover postural control after a balance disturbance (Hsiao-Wecksler and Robinovitch, 2007). Furthermore, this result indicates also that exergames might have an immediate short-term learning effect.

Other Observations and Lessons Learned

Although the exergame itself did not distinguish between different ways of raising arms to catch chickens, we observed several distinct playing styles. Some participants raised both arms to catch chickens, whereas others raised one arm only. One participant did not raise the arms at all during the game and did not catch any chickens. These differences did not result from physical limitations, as we checked that all participants were able to perform the required movements with both arms. Some participants also used the jump of the avatar to catch falling grapes faster instead of waiting underneath the trajectory of the falling grapes. These observations indicate that exergame players may perform or play in many different ways that can benefit or hinder the intended balance training. Ideally, the game technology used should be able to distinguish correct vs. incorrect movements in these situations and provide feedback in order for the exergame to function as an effective balance training and rehabilitation tool (cf. Vonstad et al., 2018). Thus, the knowledge gained by investigating different movement characteristics displayed by the players when exergaming should be used in the design and development of new exergames for health benefits.

Throughout each trial, participants moved ∼0.5 m closer to the screen on average. There was no incentive in the exergame to do so as the game was played on a two-dimensional plane parallel to the screen. This gradual forward movement was observed across all conditions and participants. The implemented elements in the game such as the falling branches did not affect this behavior. A possible explanation for the drift toward the screen might be that a pure sideways movement is difficult to achieve and less common in activities of daily living compared to side steps with an additional forwards component to them, as in avoiding an obstacle in the path of progression. Alternatively, it can be speculated that the older adults who served as participants in this study struggled to learn to use the system quickly enough to avoid drifting outside the exergaming area. Younger adults and children, the main customer-base of video game technology, grew

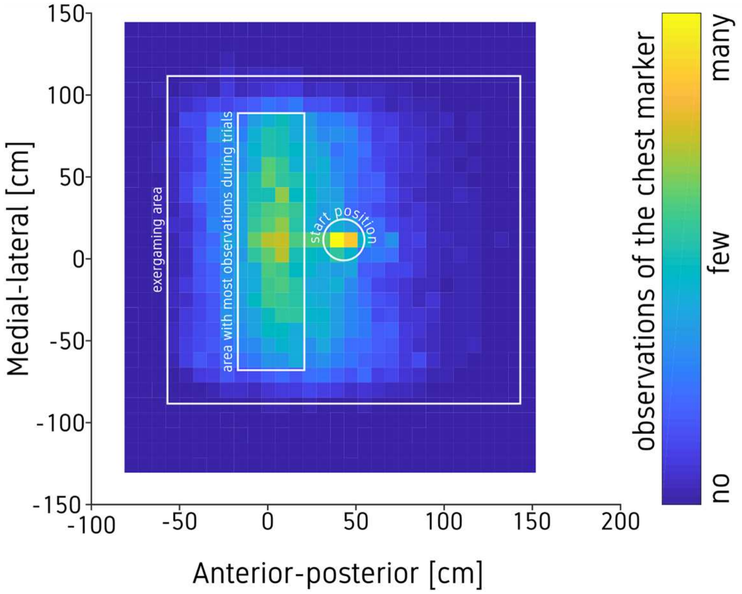

FIGURE 5 | Heatmap of the players' positions in the exergaming area based on the marker on the chest across all conditions and participants. The white square indicates the 2 by 2m active exergaming area. All participants started each trial in the white circle. The white rectangle indicates the area with most observations. The game screen was positioned to the left of the playing area.

TABLE 4 | Error messages triggered by the players moving outside the active exergaming area for each combination of settings.

Error messages		Obstacles		
		without	*with*	**Total**
Speed	*low*	5	8	13
	high	17	13	30
	Total	22	21	43

up using technology such as Microsoft's Kinect or Sony's EyeToy and are therefore more familiar with digital avatars mirroring movements, as well as with the inherent limitations caused by the limited field-of-view of the infrared cameras. Thus, they may be more likely to reposition themselves when they drift away from the center of the observable area. Although we did not mark the exergaming area with tape in order to mimic a more natural home-based setting, proper delimitation might help to reduce the ambiguity concerning one's position within the exergaming area.

Over time, the accumulation of the drift forward resulted in error messages and breaks in the game until participants stepped back into the active exergame area. Though usually of short duration and easily corrected by the player, those error messages and game breaks likely result in a disruption of the flow-zone (Csikszentmihalyi, 1975), with associated potential negative consequences for enjoyment and adherence. However, as we did not collect data on enjoyment of the exergame, this remains an assumption that needs corroboration in further studies. Yet, future development of exergames could take our findings into account by letting the game stimulate the player to move back toward the center of the play area without disrupting the exergame experience by error messages and game breaks.

The current study was designed to observe immediate effects of game settings on movement characteristics in a single exergaming session. But even across only two trial repetitions, steps became significantly larger. In order to achieve lasting improvements in the ability to maintain balance, multiple sessions over a longer period of time are needed. With more repetitions, people who routinely use exergames for balance training might show additional changes in movement characteristics in reaction to changes in game speed or additional challenges. Further research is needed with longer follow-up time to study potential effects over extended playing time.

Limitations

The current study offers several insights into how game settings in exergames influence the movement characteristics of older adults, allowing to provide recommendations for the development and usage of exergames to elicit the movements necessary to train balance in older adults. However, a few limitations should be highlighted as well. First, this study was conducted using a screen-based exergame. Newly developed exergames might use newer technologies such as immersive virtual reality or gamified objects in the environment rather than TV screens. However, we believe that many of the lessons learned may be transferrable to virtual reality exergames, since the basic concepts for eliciting the desired vs. less desirable movements may largely be the same. Furthermore, older adults are a specific subgroup of users who, at the moment, might be less familiar with modern game and simulation technologies than younger generations. However, continued developments in non-immersive virtual reality and increasing tech-savviness of new generations of older adults may contribute to the accessibility of exergame technology for a wider audience. Finally, although no formal tests of the participants' physical and mental capacity were performed in this study, all participants were healthy for their age and living independently. Further research should broaden out to participants with a wider range of functional capacities to investigate how they would react to changes in game speed and additional challenges.

Next Steps

Methods to deliver safe, home-based, low-cost balance training to older adults in order to prevent falls and to maintain independence are an important issue for future research, and the availability of a control system implemented in an exergame that maximizes the likelihood of eliciting the most beneficial balance-training movements is of great interest. There are still several unanswered questions, such as potential changes in movement characteristics when playing over extended periods of time and to what extent comparable findings would be produced using different game settings or different exergames. Furthermore, personalizing the delivery of exergame training to the abilities and preferences of older adults, including providing feedback about the correctness of performance, might help accommodate differences in functioning in this heterogenous age group and potentially lead to additional improvements in balance training outcomes.

CONCLUSION

Together, the results of the current study provide important insights into how the settings of an exergame influence the movement characteristics of older adults when playing a step-based balance training exergame. Higher game speed led to faster-paced movements with shorter duration of single support, whereas additional cognitive elements in the form of obstacles to avoid led to slower movements and smaller steps. Some of these effects on movement characteristics are beneficial for balance training, whereas others likely make the exergame less efficient. Therefore, informed decisions are necessary when designing new, or choosing between existing, balance training exergames if they are to be effective tools for balance training in older adults.

DATA AVAILABILITY STATEMENT

The datasets generated for this study will not be made publicly available. Due to Norwegian legislation, the dataset for this article is not open access. Questions regarding the dataset can be sent to Beatrix Vereijken, beatrix.vereijken@ntnu.no.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Regional Committees for Medical and Health Research Ethics. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

All authors contributed to the conception and design of the study. PA, EB, and KG collected the data. PA and EB processed the data. PA performed statistical analyses. PA and BV drafted the manuscript. All authors contributed to manuscript revision, read, and approved the final version.

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Exercise of Dynamic Stability in the Presence of Perturbations Elicit Fast Improvements of Simulated Fall Recovery and Strength in Older Adults: A Randomized Controlled Trial

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Age-related impairments of reactive motor responses to postural threats and reduced muscular capacities of the legs are key factors for the higher risk of falling in older people. It has been evidenced that a training of dynamic stability in the presence of perturbations has the potential to improve these deficits. However, the time course of training effects during such interventions is poorly understood. The purpose of this parallel-group study was to investigate the temporal adaptation dynamics of the balance recovery performance and leg strength during a dynamic stability training. Forty-two healthy older adults (65–85 years) were randomly assigned to a training ($n = 27$, analyzed $n = 18$) or control group ($n = 15$, $n = 14$). The training was conducted in a group setting for 6 weeks (3×/week, 45 min). The exercises focused on the mechanism of stability control (i.e., modulation of the base of support and segment counter-rotations around the center of mass) during standing, stepping, and jumping on unstable surfaces with a high balance intensity. Before, after 3 and after 6 weeks, the maximum plantar flexion moment and the knee extension moment were assessed. The recovery performance was evaluated by a simulated forward fall (lean-and-release test) and the margin of stability concept. The margin of stability at release decreased significantly after 3 weeks of training (34%, effect size $g = 0.79$), which indicates fast improvements of balance recovery performance. The margin of stability further decreased after week 6 (53%, $g = 1.21$), yet the difference between weeks 3 and 6 was not significant. Furthermore, the training led to significant increases in the plantar flexion moment after weeks 3 (12%, $g = 0.72$) and 6 (13%, $g = 0.75$) with no significant difference between weeks. For the knee extension moment, a significant increase was found only after week 6 (11%, $g = 1.07$). The control group did not show any significant changes. This study provides evidence that a challenging training of dynamic stability in the presence of perturbations can improve balance recovery performance and leg strength of older adults already after a few weeks. Therefore, short-term training interventions using this paradigm may be an effective strategy for fall prevention in the elderly population, particularly when intervention time is limited.

Keywords: fall prevention, aging, dynamic stability training, reactive control, unexpected perturbations and disturbances, randomized controlled trial

INTRODUCTION

The increased risk of falling and associated injuries in older adults (Rubenstein, 2006) makes falls a major source of morbidity and mortality (Rubenstein, 2006; Marks, 2011; Alamgir et al., 2012) in a globally senescent population. There is strong evidence that physical exercise interventions can reduce fall risk and rates in older people (Sherrington et al., 2019) and are therefore promoted by international guidelines and national health bodies as a feasible and cost-efficient prevention tool (Moyer, 2012; Kim et al., 2017; Guirguis-Blake et al., 2018; Sherrington et al., 2019).

The age-related decline of muscular capacities in the lower extremities is one intrinsic key factor for the higher risk of falling (Karamanidis et al., 2008; Pijnappels et al., 2008; Graham et al., 2015) and, for that reason, a classical strength training has the potential to improve balance performance (Pijnappels et al., 2008; Arampatzis et al., 2011; Pamukoff et al., 2014). However, the ability to respond to sudden perturbations and postural threats as the cause of a fall event also strongly relies on the successful application of general mechanisms responsible for the dynamic stability control (Bierbaum et al., 2010), i.e., increase in the base of support and counter-rotating segments around the center of mass (CoM) (Hof, 2007). The application of these control mechanisms of the neuromotor system (including perception, signal processing, and motor control) is not actively exercised during classical strength training and likely requires a different intervention approach that focuses on balance recovery performance (Sherrington et al., 2019). In support of this, adaptations of neural control are different after balance compared to strength training (Beck et al., 2007; Gruber et al., 2007; Taube et al., 2007).

Earlier studies of our group showed that when focusing on exercises that promote the application of stability control mechanisms, the ability of older adults to regain balance after unexpected perturbations (which were not explicitly trained) can be improved (Arampatzis et al., 2011; Bierbaum et al., 2013), even beyond the effects of a mixed training approach including resistance training (Bierbaum et al., 2013). However, since in these studies the relative load on the leg muscles was rather low in the stability training group, a strength gain in the lower extremities was not observed. Consequently, we aimed for developing a training that takes advantage of the dynamic stability control exercises but would also increase leg strength in order to target both fall risk factors (i.e., balance recovery performance and muscular capacities) in one intervention and to improve the efficiency of the intervention. For this purpose, we made use of the presence of perturbations evoked by unstable surface conditions that induce continuous variable and partly unpredictable disturbances in combination with the dynamic stability training approach in a subsequent intervention study (Hamed et al., 2018). It has been shown that the presence of perturbations and surface irregularities leads to increased muscle activation, which may effectively stimulate strength gains over time (Munoz-Martel et al., 2019). Furthermore, there is evidence that the presence of fluctuations and disturbances in the neural processing of sensory inputs to motor outputs can improve motor behavior (Faisal et al., 2008; Sejdić and Lipsitz, 2013) and balance performance (Priplata et al., 2003; Aboutorabi et al., 2018; White et al., 2019), and may facilitate motor learning and adaptation (Faisal et al., 2008; Van Hooren et al., 2019). The results of this intervention showed that after 14 weeks of dynamic stability training in the presence of perturbations by unstable surfaces, both strength and balance recovery performance were significantly improved. The increase in balance ability was even greater when compared to a classical resistance training program (Hamed et al., 2018). An association of the exercise-induced changes of the balance recovery performance and changes of the execution time of the recovery step explained the improvement in the stability performance. We concluded that a training, which includes the application of dynamic stability recovery mechanisms in the presence of perturbations, is very effective to improve age-related impairments of the balance recovery performance in fall-like situations.

Yet, little is known about the time course of adaptation of such a training approach. Intervention studies in this field normally use a training period over several months as in our previous study (Hamed et al., 2018). However, in the face of challenging perturbations, either discrete or continuous, the neuromotor system shows acute control adjustments to cope with the postural threat (Bierbaum et al., 2010; Cronin et al., 2013; Graham et al., 2015; Patikas et al., 2016; Santuz et al., 2018; Munoz-Martel et al., 2019), and this ability seems quite unaffected by age (Bohm et al., 2015). It has been suggested that managing such posture-challenging conditions might be a neural mechanism that triggers adaptations of the balance recovery performance (Munoz-Martel et al., 2019). In fact, retention effects have been documented already in days and a few weeks after an initial exposure to specific perturbations (Trimble and Koceja, 2001; Bhatt et al., 2006; Gruber et al., 2007; McCrum et al., 2018). The potential of short-term adaptations is supported by findings of a high temporal plasticity of the motor control system in response to a general balance training (Taube et al., 2008; Taubert et al., 2010; Patel et al., 2019). This suggests that adaptations in the neuromotor control system may improve balance recovery performance and strength capacity already in a short time of systematic exercising (Penzer et al., 2015).

The purpose of the present randomized controlled trial was to investigate the time course of improvements of the balance recovery performance following a simulated forward fall as well as strength capabilities of the plantar flexors and knee extensors in response to a 3- and 6-week intense dynamic stability training in the presence of perturbations. We hypothesized that repeated exposure to continuously variable and partly unpredictable disturbances during the training sessions will lead to improvements in stability recovery performance following a simulated forward fall and muscle strength already after 3 weeks. Furthermore, we hypothesized that the balance recovery improvement can be partly explained by an association of the exercised-induced changes of the stability performance and changes of the execution time of the recovery step.

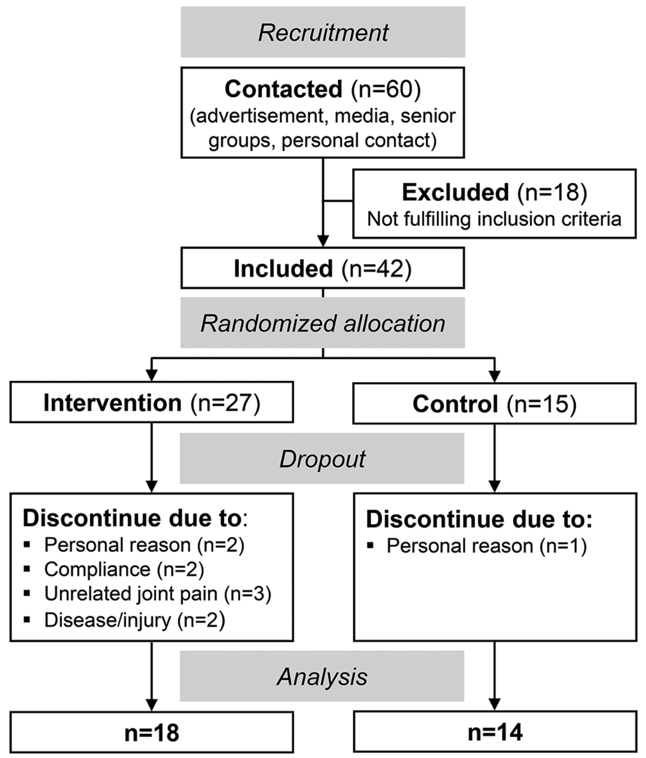

FIGURE 1 | Flow chart illustrating the progress of the intervention trial.

MATERIALS AND METHODS

Experimental Design

Figure 1 illustrates the progress of the randomized controlled trial with a non-blinded parallel group design. Inclusion criteria for the study were an age between 65 and 85 years and no neural, systemic, and musculoskeletal disorders. Forty-two participants were included and randomly assigned to either a training or control group [simple randomization by a random number generator, 2:1 allocation ratio (intervention/control), **Figure 1**]. Finally, 32 older adults (>65 years) were analyzed: 18 in the training group (73 ± 6 years, 12 female) and 14 in the control group (73 ± 7 years, 8 female, **Figure 1**).

The intervention group performed a 6-week training of dynamic stability in the presence of perturbations, based on our earlier study approach (Hamed et al., 2018). The participants of the control group were instructed not to change any of their regular physical activity habits in the intervention period. The primary outcome measures were the balance-recovery performance that was assessed using a simulated forward fall paradigm (lean-and-release test) and the muscle strength of the plantar flexors and knee extensors that was measured on a dynamometer, determined before (week 0), after 3 weeks (week 3), and after 6 weeks (week 6). The ethics committee of the Humboldt-Universität zu Berlin approved the study, and the participants gave written informed consent in accordance with the Declaration of Helsinki. The study followed the CONSORT guidelines (Schulz et al., 2010).

Exercise Intervention

The dynamic stability training was conducted in a supervised group setting taking place in the university's sports gym for 6 weeks, three times a week for 45 min, including a 5-min warm-up and cool-down. The concept of the training is grounded on the exercise of the mechanisms responsible for dynamic stability control (Arampatzis et al., 2011; Bierbaum et al., 2013), i.e., adjusting the base of support and counter-rotations of segments around the CoM (Hof, 2007) in the presence of perturbations (Hamed et al., 2018). Soft, unsteady, uneven, and moveable surfaces were used in order to introduce continuously variable, predictable, and unpredictable perturbations to facilitate balance performance and adaptation. The dynamic stability training program was developed as a three-component approach, where we modified (1) the type of exercise, (2) the used equipment, and (3) the level of difficulty of the exercise (**Figure 2**). First, the flexible application of the dynamic stability mechanism was challenged during three exercise blocks of standing, stepping, and jumping (**Figure 2**). Secondly, five different kinds of training equipment [soft pads (Sport-Thieme, balance pad vinyl), wedged soft pads (SoftX, coordination seesaw), posturomed (BIOSWING, Posturomed 202), balance half-balls (Sport-Thieme, balance jumps), and balance cushions (SISSEL, Balancefit)] creating unstable conditions were used to continuously introduce perturbations. Finally, the axis of the level of difficulty of our three-component approach included 11 different modifications of the exercises to increase the balance intensity (e.g., arms crossed, moving body segments, perturbations by partner; **Figure 2**). The combination of all three components (3 exercise blocks, 5 kinds of equipment, and 11 modifications) resulted in a repertoire of 165 different exercises that have been applied during the intervention. The intensity of the dynamic stability exercise, i.e., postural threat, was progressively adjusted to the individual balance ability level of each participant throughout the training period. The criterion for the instructor to increase the intensity by choosing the next-level exercise (vertical axis in **Figure 2**) was that the participant could perform a certain exercise without stepping so often off the device or without taking support from a partner (roughly >1 times every 10 s). The exercises were, therefore, not predetermined but adjusted whenever possible. The participants were encouraged by the instructors to focus on keeping the balance during the exercise blocks as good as possible. The instructors guided and supported during the implementation of an exercise modification. The training itself was organized in 5 stations, each with a different kind of training equipment. On the different stations, each exercise block (standing, stepping, and jumping) was trained consecutively. The participants trained pairwise, while the non-exercising partner always stood with arms upright beside the person that trained to give grasping support or to be able to catch the partner when necessary, and mats were placed around the training stations to account for the appropriate safety of the participants. If persons faced any insecurities or felt uncomfortable, the instructor provided further support. After 60 s, the roles of the partners were changed, summing up to about 3 min of training per participant at one station. After 3 and 6 weeks of training, the participants had exercised for about 135 and 270 min, respectively, the mechanism of dynamic stability control in the presence of perturbations at a high level of balance intensity. Because in our last intervention study (Hamed et al., 2018) the training did not lead to significant increases of knee extensor muscle strength, an additional demand to this muscle group was evoked by keeping the knee flexed in the standing position and by maintaining the lunge and squatting position after stepping and landing for a longer time (up to 20 s). Two introduction and familiarization meetings without systematic exercising within 1 week preceded the intervention. We had three consecutive training groups with nine participants who started in each group. Two instructors were present during the training sessions.

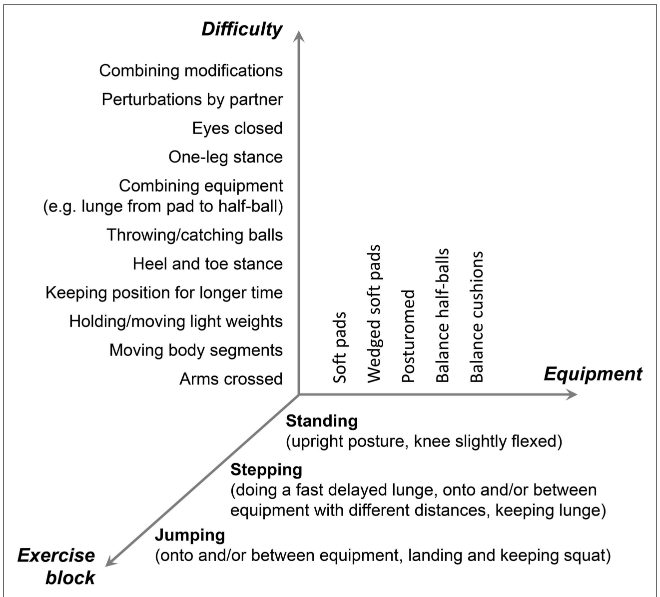

FIGURE 2 | Description of the concept of the training of the dynamic stability in the presence of perturbations. Participants trained each of the exercise blocks under unstable conditions introduced by the different kinds of used equipment. The balance intensity was kept high by modifying the exercises according to the individual performance level to challenge the application of the dynamic control mechanisms (horizontal and vertical axis).

Simulated Forward Fall Paradigm

The participants wore an adjusted upper body harness, which was connected horizontally by a non-elastic rope to an electromagnet mounted on the wall. The magnet was connected in series to a custom-built release system and a force transducer (Megatron 0–5 kN; MEGATRON Elektronik GmbH & Co. KG, Munich, Germany). In this way, the participants were able to lean forward in a straight bodyline, feet hip-width apart and freely hanging arms, while the pulling force expressed as percentage of body weight (BW) was used to control the inclination angle. The initial inclination was set to 8% BW. After receiving a ready cue in the prepared forward lean position, the participants were suddenly released without further warning in an interval of 2–10 s. The participants were instructed to recover a stable lunge with a single step upon the unexpected release. When balance was successfully recovered, the lean angle was increased gradually in 3% BW intervals, until the participants were not able to recover with a single step in three successive attempts. A second rope connected the harness to the ceiling and was adjusted in length, so that the participants would not hit the floor during unsuccessful recovery. The release onset was determined by a 50% reduction in the leaning force signal provided by the force transducer (1,000 Hz). The participants were positioned in such a way that the recovery step by the right leg landed on a force plate (AMTI, BP 400600-2000, 60 cm × 40 cm), and the measured vertical ground reaction force (1,000 Hz) was used for touchdown detection (i.e., increase by ≥5 N).

In order to assess the stability state during release and recovery, we used the concept of the extrapolated CoM proposed by Hof et al. (2005). The extrapolated CoM (X_{CoM}) is calculated as:

$$X_{CoM} = \frac{P_{CoM} + V_{CoM}}{\sqrt{\frac{g}{l}}}$$

were P_{CoM} is the horizontal (anterior–posterior) component of the projection of the CoM to the ground, V_{CoM} is the horizontal CoM velocity in the same direction, and the term $\sqrt{\frac{g}{l}}$ expresses the eigenfrequency of an inverted pendulum of length l (g is the acceleration of gravity and l is the distance between the CoM and the center of the ankle joint in the sagittal plane). The position of the extrapolated CoM in the anterior–posterior direction was then referred to the anterior boundary (U_{max}) of the base of support, expressing the margin of stability (b_x) (Hof et al., 2005) in the anterior–posterior direction (Karamanidis and Arampatzis, 2007) as:

$$b_x = U_{\max} - X_{CoM}$$

Positive values of the margin of stability, i.e., the extrapolated CoM is within the anterior boundary of the base of support, indicate that the body position is stable, while in the opposite, the stability is lost (Karamanidis et al., 2008). The required kinematic data for the CoM calculation were captured by a

Vicon motion capture system (10 cameras at 250 Hz) on the basis of an anatomically referenced set (Bierbaum et al., 2013) of reflective markers (radius 14 mm) that define the respective body segments (foot: calcaneus and second metatarsal bone, shank: lateral femoral epicondyle and lateral malleolus, thigh: greater trochanter and lateral femoral epicondyle, trunk: left and right greater trochanter and C7, upper arm: lateral humeral epicondyle and acromion process, lower arm and hand: midpoint between styloid processes of radius and ulna and lateral humeral epicondyle, head: C7 and a headband with two markers in the front and two in the back). Masses and the locations of the segment CoM were calculated based on the data reported by Dempster et al. (1959), and the body CoM position in the 3D space was calculated according to Winter (1979). The boundaries of the base of support were determined using the vertical projection of the heel marker of the rear foot and the tip of the shoe of the front foot, considering the distance of the metatarsal marker to the anterior boundary of the shoe (measured during preparation).

Muscle Strength Measurements

The strength of the knee extensors and plantar flexors of the right leg (same as the recovery leg in the simulated forward fall test) was examined during maximum voluntary isometric contractions (MVC) on a Biodex dynamometer (Biodex Medical, Syst.3, Shirley, NY, USA). For the knee extensions, the participants were seated on the dynamometer chair with a trunk angle flexion of 85° (trunk in line to the thighs = 0°) and for the plantar flexions with an angle of 70° and with fully extended knee, while the arms being crossed on the chest. Following a standardized warm-up, five MVCs in different joint angles were performed for the knee (between 50 and 75°) and the ankle joint (between 8 and 25° dorsiflexion), respectively. A 3-min rest was given between trials, and the highest value was used for further analysis. The resultant ankle and knee joint moments were calculated using an established inverse dynamics approach in order to account for axis misalignment between the dynamometer and the joint as well as gravitational moments (Arampatzis et al., 2004, 2005). For this purpose, anatomically referenced reflective markers (greater trochanter, medial and lateral femoral epicondyles and malleoli, second metatarsal bone, and calcaneus bone) were captured using a Vicon motion analysis system (Version 1.8, Vicon Motion Systems, Oxford, UK) integrating seven cameras at 250 Hz.

Statistics

A statistical power analysis was performed *a priori* to calculate the required sample size by means of the software G*Power (version 3.1.9.6, Germany). For this purpose, we used the effect size of the margin of stability at release from our previous dynamic stability training intervention study (Hamed et al., 2018). Since the training frequency per week was higher while the intervention duration of the current study was shorter, we assumed a reduced effect (by ∼20%). The power analysis was conducted for the *post-hoc* time point comparison for the intervention group, considering a Bonferroni correction of the *p*-values [$\alpha = 0.0167$ (adjusted), power = 0.8, effect size: 0.86, two-tailed], and revealed a sample size of $n = 17$. We included 27 participants for the intervention group to consider possible dropouts.

An analysis of variance for repeated measures on a linear mixed model was calculated for the strength (normalized to body weight) and stability parameters with the time point as the within-subjects factor (week 0 vs. week 3 vs. week 6) and group as a between-subjects factor (intervention vs. control). In case of time effects or time by group interactions, a Benjamini–Hochberg corrected *post-hoc* analysis was conducted separately for each group (adjusted *p*-values will be reported). Baseline (week 0) anthropometric, strength, and stability parameters were compared between the intervention and control group using the same linear mixed model. The relationship between the changes in the margin of stability at release and the changes in the rate of increase in the base of support from release to touchdown was analyzed by means of a Pearson correlation coefficient. The level of significance was set to $\alpha = 0.05$, and statistical analyses were conducted using R v3.4.1 (R Found. for Stat. Comp.). Effect sizes (Hedges' g) were calculated to assess the strength of the intervention effects, where $0.2 \leq g < 0.5$ indicates small, $0.5 \leq g < 0.8$ indicates medium, and $g \geq 0.8$ indicates a large effect size (Cohen, 1988). Note that due to technical or personal issues, some datasets from the three different measurement time points of the intervention and control groups of the two strength tests (25 from 192) and stability test (15 from 96) were missing; however, a strength of linear mixed models is that they can handle missing data and, thus, the respective participants could be included.

RESULTS

Sixty older adults (>65 years) were contacted and finally 42 meet the inclusion criteria and agreed to participate (**Figure 1**). Twenty-seven of those were randomly assigned to the intervention group and 15 to the control group. Ten participants discontinued the intervention (nine dropouts in the intervention and one in the control group) and, thus, 18 participants of the intervention group and 14 of the control group were included in the final analysis (**Figure 1**). There were no baseline differences of the margin of stability at release and the maximal ankle and knee joint moment between those who completed the training and those who dropped out ($p > 0.05$). The intervention and control group did not show any significant differences with respect to age (intervention: 73 ± 6 years, control: 73 ± 7 years, $p = 0.903$), height (164 ± 11 cm, 169 ± 10 cm, $p = 0.200$), and body mass (70 ± 15 kg, 69 ± 10 kg, $p = 0.970$). A baseline (week 0) group difference was found for the margin of stability at release ($p = 0.018$) and for the rate of increase in the base of support ($p = 0.007$, **Table 1**).

The margin of stability at release showed a significant main effect of time ($p < 0.001$) and no significant time by group interaction effect ($p = 0.098$). The intervention group showed a significant decrease in the margin of stability at release from week 0 to week 3 ($p = 0.013$) and week 0 to week 6 ($p < 0.001$), but no significant change from week 3 to week 6 ($p = 0.258$, **Table 1**). This indicates fast improvements in balance recovery

TABLE 1 | Outcome parameters before (week 0), within (week 3), and after (week 6) the training period for the intervention and control groups.

Parameter	Intervention group			Control group		
	Week 0	Week 3 (g: w0–3)	Week 6 (g: w0–6, w3–6)	Week 0	Week 3 (g: w0–3)	Week 6 (g: w0–6, w3–6)
MoS RS (cm)*	−8.19 ± 6.56	−10.97 ± 6.40 (−0.79)#	−12.55 ± 6.24 (−1.21, −0.44)#	−14.09 ± 6.55’	−14.64 ± 6.55 (−0.16)	−15.97 ± 6.17 (−0.51, −0.36)
MoS TD (cm)	11.44 ± 5.94	10.17 ± 5.88 (−0.22)	8.64 ± 5.81 (−0.49, −0.27)	7.24 ± 5.95	8.21 ± 5.95 (0.18)	7.29 ± 5.75 (0.01, −0.16)
BoS TD (cm)	95.3 ± 14.6	101.1 ± 14.3 (0.62)	100.0 ± 13.9 (0.50, −0.11)	103.9 ± 14.6	103.5 ± 14.6 (−0.05)	106.9 ± 13.8 (0.30, 0.35)
Duration RS-TD (ms)~	524.7 ± 63.5	552.0 ± 63.6 (0.31)	498.0 ± 63.5 (−0.30, −0.60)	479.7 ± 63.6	472.9 ± 63.6 (−0.08)	475.6 ± 63.4 (−0.05, 0.02)
Rate of BoS (cm/s)*~	182.2 ± 32.9	189.9 ± 32.2 (0.34)	200.0 ± 31.5 (0.77, 0.44)#	216.8 ± 32.9’	218.9 ± 32.9 (0.10)	224.5 ± 31.2 (0.33, 0.24)
Moment ankle (Nm/kg)*	1.57 ± 0.44	1.76 ± 0.46 (0.72)#	1.78 ± 0.43 (0.75, 0.08)#	1.65 ± 0.47	1.72 ± 0.47 (0.30)	1.78 ± 0.43 (0.50, 0.22)
Moment knee (Nm/kg)*	2.00 ± 0.45	2.10 ± 0.44 (0.60)	2.21 ± 0.40 (1.07, 0.54)#	2.32 ± 0.43	2.38 ± 0.45 (0.39)	2.39 ± 0.42 (0.38, 0.01)

Mean ± SD of the margin of stability (MoS) at release (RS), margin of stability and base of support (BoS) at touchdown (TD), duration from release until touchdown, rate of increase in the base of support (Rate of BoS), maximum voluntary isometric ankle plantar flexion moment and knee extension moment (normalized to body weight); g, Hedges' g effect size (note that negative values for the MoS RS and Duration RS-TD indicate a positive performance effect).
**Statistically significant time effect ($p < 0.05$).*
 # Statistically significant difference (post-hoc analysis) to week 0 ($p < 0.05$).
~ Statistically significant group effect ($p < 0.05$).
‘Statistically significant difference (post-hoc analysis) to the intervention group ($p < 0.05$).

performance after a short time of training. The control group showed no significant differences between the time points ($p > 0.05$). **Figure 3** illustrates the changes between weeks 0 and 3 as well as weeks 3 and 6 in the margin of stability at release, which were significantly greater in the intervention group compared to the control group ($p = 0.044$).

The base of support of the recovery step in the anterior direction showed no significant time by group interaction effect ($p = 0.158$) and no significant time effect ($p = 0.056$), hence no significant differences were found for both groups between time points ($p > 0.05$, **Table 1**). A significant time effect ($p = 0.008$) but no time by group interaction effect ($p = 0.471$) was observed for the rate of increase in the base of support from release to touchdown. For the intervention group, a significant increase in the rate from weeks 0 to 6 was found ($p = 0.034$, **Table 1**). No further significant differences were observed for the other time points and the control group ($p > 0.05$). There was a significant correlation between the changes in the margin of stability at release and the changes in the rate of increase in the base of support from release to touchdown between weeks 0 and 3 ($r = -0.705$, $p = 0.002$), weeks 0 and 6 ($r = -0.733$, $p = 0.001$), and weeks 3 and 6 ($r = -0.623$, $p = 0.017$, **Figure 4**).

The ankle and knee joint moments showed a significant main effect of time ($p = 0.002$, $p < 0.001$) but no significant time by group interaction ($p = 0.499$, $p = 0.124$), respectively. The ankle joint moment of the intervention group revealed a significant increase from week 0 to week 3 ($p = 0.041$) and week 0 to week 6 ($p = 0.041$), but no significant change from week 3 to week 6 ($p = 0.947$, **Table 1**). The knee joint moment was not significantly different between weeks 0 and 3 ($p = 0.131$) and weeks 3 and 6 ($p = 0.156$) but was significantly different between week 0 and week 6 ($p < 0.001$, **Table 1**). No significant time point differences were observed for the ankle and knee joint moment of the control group ($p > 0.05$, **Table 1**).

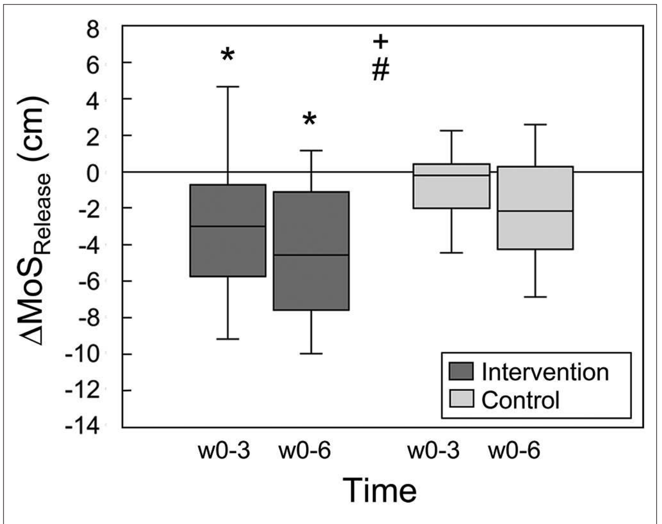

FIGURE 3 | Changes of the margin of stability at release ($\Delta MoS_{Release}$) during the simulated forward falls between weeks 0 and 3 as well as weeks 0 and 6 for the intervention ($n = 18$) and control groups ($n = 14$), respectively. Statistically significant main effect of + time and # group ($p < 0.05$). *Statistically significant difference (*post-hoc* analysis) to baseline, i.e., week 0 ($p < 0.05$).

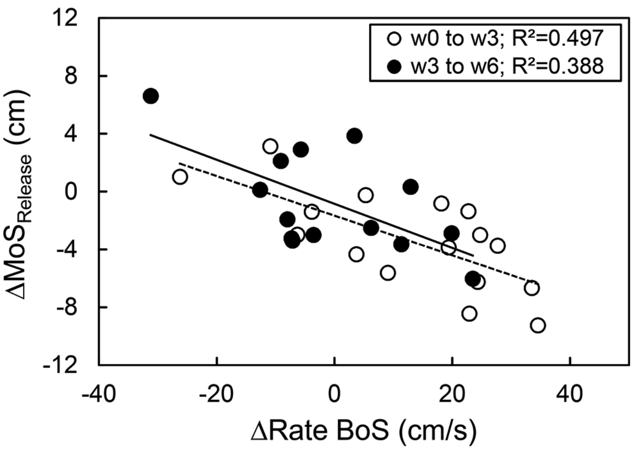

FIGURE 4 | Relationship between the exercise-induced change of the margin of stability at release ($\Delta MoS_{Release}$) and the rate of the base of support increase (ΔRate BoS) from release to touchdown during the simulated forward fall for the time course of week 0 to week 3 (n = 16) and week 3 to week 6 (n = 14).

DISCUSSION

In the present study, we investigated the initial time course of adaptive responses to a challenging training of the dynamic stability in the presence of perturbations, considering the effects on balance recovery performance and lower limb strength capacities of older adults. In agreement with our hypotheses, we found improvements in recovery performance and strength already after 3 weeks of intervention, suggesting fast and effective improvements of critical fall risk factors.

The results showed that a few sessions (nine training sessions) of a stability training in the presence of perturbations, which included the application of mechanisms responsible for dynamic stability control under unstable conditions, were sufficient to improve the reactive stepping behavior during a simulated forward fall test. The recovery performance, defined as the lowest margin of stability at release (or most unstable body position) that could be recovered with a single step, improved by 34% ($g = -0.79$) after week 3 (nine training sessions) and by 53% ($g = -1.21$) after week 6 (18 training sessions) in the intervention group. Reactive stepping performance following a forward loss of balance was shown to be a predictor of the risk of falling (Carty et al., 2015; Okubo et al., 2017). Therefore, the finding of fast improvements in stability performance promotes the application of the current short-term dynamic stability training approach for the prevention of falling in the older population. Furthermore, the training can be applied in a group setting under the supervision of one or two instructors and with cheap and conventional equipment, making this approach very feasible and attractive for a clinical and broader setting.

The improved balance recovery performance in the intervention group was associated with a higher rate of increase in the base of support from release to touchdown, i.e., faster recovery step. Although the gain of the rate of increase in the base of support was only significant after week 6, changes in the rate of increase in the base of support were inversely correlated with changes in the margin of stability at release at all time points. The base of support of the recovery step did not show any significant changes after weeks 3 and 6. This confirms our previous findings underlining the importance of the ability to execute recovery steps in a short time for balance recovery (Karamanidis et al., 2008; Hamed et al., 2018). However, the delayed significant increase in the rate of increase in the base of support (after week 6) compared to the earlier changes of the margin of stability at release (after week 3) may indicate that exercise-induced alterations of other mechanisms (e.g., counter-rotation of segments) also contributed to the improvements of the balance recovery performance, particularly in the initial phase of the intervention (i.e., first weeks).

The fast improvements in the simulated forward fall test following the current training of the dynamic stability in the presence of perturbations, in which this testing task was not explicitly exercised, may rely on the changes of the control by the neuromotor system when coping with postural challenges. Recently, we found that in the presence of perturbations, the basic activation patterns of muscle groups in both balance and locomotion tasks become fuzzier (Santuz et al., 2018, 2020; Munoz-Martel et al., 2019), less unstable, and less complex (Santuz et al., 2020), indicating increased control robustness (i.e., ability to cope with errors; Santuz et al., 2018) in challenging settings. The perturbation-induced modification and modulation of activation patterns might be a neural mechanism that improves dynamic stability performance when repetitively applied such as during the present training of dynamic stability (Munoz-Martel et al., 2019). Furthermore, the presence of continuous disturbances in the neural processing of sensory inputs to motor outputs can improve balance performance (Priplata et al., 2003; Aboutorabi et al., 2018; White et al., 2019), and may stimulate motor adaptation (Faisal et al., 2008; Van Hooren et al., 2019) resulting in the enhanced balance recovery ability.

After 3 and 6 weeks of intervention, we observed significant increases in plantar flexor (g = 0.72, g = 0.75) and knee extensor muscular (g = 0.60, g = 1.07) capacities, respectively. Challenging postural settings change the activation patterns of the lower limb muscles with higher and longer activations to generate compensatory joint moments (Cheung et al., 2009; Cronin et al., 2013; Voloshina et al., 2013; Voloshina and Ferris, 2015; Nazifi et al., 2017; Santuz et al., 2018; Munoz-Martel et al., 2019). During a single training session, the participants were exposed to about 15 min of continuous perturbations at a high balance intensity with body positions that put a great demand on the control and the muscular system. This summed up to about 135 min after 3 weeks and 270 min after 6 weeks. Neural plasticity can take place in a very short time on the spinal, supraspinal, up to the cortical level in response to short-term balance (Taube et al., 2008; Taubert et al., 2010; Patel et al., 2019) and strength training (Carroll et al., 2002, 2011), or a combination of both (Penzer et al., 2015). Thus, we can argue that the presence of perturbations during the training likely triggered changes in the neural drive to the muscle over time and consequently caused the observed strength gains.

The increase in muscle strength of the plantar flexors became significant already after 3 weeks of training (week 3: 12%, week 6: 13%), while for the knee extensors, a significant increase was first observed after week 6 (11%). The earlier gains in the plantar flexors might be due to the fact that the majority of the exercise blocks were performed on soft surfaces (soft pads, balance half-balls, balance cushions) in which predominantly the plantar flexors are involved to maintain anterior–posterior balance. Furthermore, it has been suggested that the plantar flexors are sensitive for surface perturbations due to their morphological design (longer tendons and shorter fascicles) and direct interaction with the ground (Biewener and Daley, 2007; Daley et al., 2007), leading to higher activation levels and presumably faster adaptive responses (Hamed et al., 2018).

In our previous study with the same approach, we were not able to provoke significant increases in the knee extensor strength ($d = 0.41$) (Hamed et al., 2018). Given the importance of the knee extensors for balance recovery (Karamanidis and Arampatzis, 2007; Karamanidis et al., 2008), we modified the exercises for the present intervention to involve this muscle group more in the stabilizing task and to increase the loading demand. This was achieved by introducing more flexed knee angles during the three exercise blocks and by keeping a more lunged/squatted position for several moments after the step and jump on the unstable training utensils. The significant increase in the knee extensor strength after week 6 indicates that the simple modification of the exercises was effective and led to strength gains comparable to the plantar flexors after 6 weeks of intervention, i.e., plantar flexors 13% ($g = 0.75$) and knee extensors 11% ($g = 1.07$).

The balance recovery performance and plantar flexor strength were significantly increased after week 3 with no further significant increase after week 6. This indicates that in the time course of the current training of the dynamic stability in the presence of perturbations, adaptive response rates seem to be higher in the first weeks when compared to the following time period. In our previous intervention study with a similar training approach, the participants exercised twice a week for 14 weeks (Hamed et al., 2018). Despite the lower frequency, the longer intervention period provoked notably greater improvements, i.e., 80% for the forward fall test and 20% for the strength measurement compared to 53% and 13% after week 6 in the present study. Thus, longer training periods seem to be beneficial to promote adaptive effects in older adults.

Although the participants were randomly assigned, we found baseline differences between the intervention and control groups for the margin of stability at release and the rate of increase in the base of support. This was mainly due to two participants of the control group, which performed exceptionally well in the balance recovery test. Both reported to be physically active, which may explain the performance above average. When excluding these two datasets from the analysis, the differences diminished. Since no further group comparisons were made, the participants were included to increase the statistical power. In the current study, we faced a comparably high dropout rate (33%). However, as can be seen in **Figure 1**, the reasons for the dropout were not related to the intervention itself, i.e., unrelated joint pain ($n = 3$), diseases ($n = 2$), and personal reasons ($n = 2$), while two participants were withdrawn because of limited training compliance. For the last two reasons, it should be noted that this was a voluntary participation with no compensation, but the time efforts for the measurements (three times with about 3 h of testing) in addition to the time spent for the training were quite high. Therefore, we can assume that the training itself is indeed feasible and not too difficult or overchallenging. Furthermore, we included only healthy older adults in the intervention and, therefore, any translation of the findings to other populations, e.g., frailty or pathologies, warrants further investigation. No clinical trial registration had been performed.

In conclusion, the current study provides evidence that a challenging stability training, which focuses on the application of mechanisms responsible for dynamic stability control in the presence of perturbations, can improve balance recovery performance and leg strength already after a short time period, i.e., 3 weeks. Therefore, short-term training interventions using this exercise paradigm may be an effective strategy for fall prevention, particularly when intervention time is limited.

DATA AVAILABILITY STATEMENT

The datasets generated for this study are available on request to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics committee of the Humboldt-Universität zu Berlin. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

SB and AA designed the research and drafted the manuscript. SB and MM-L performed the research and analyzed the data. FM and MM-L made important intellectual contributions during revision.

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Stair Gait in Older Adults Worsens With Smaller Step Treads and When Transitioning Between Level and Stair Walking

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Older people have an increased risk of falling during locomotion, with falls on stairs being particularly common and dangerous. Step going (i.e., the horizontal distance between two consecutive step edges) defines the base of support available for foot placement on stairs, as with smaller going, the user's ability to balance on the steps may become problematic. Here we quantified how stair negotiation in older participants changes between four goings (175, 225, 275, and 325 mm) and compared stair negotiation with and without a walking approach. Twenty-one younger (29 ± 6 years) and 20 older (74 ± 4 years) participants negotiated a 7-step experimental stair. Motion capture and step-embedded force platform data were collected. Handrail use was also monitored. From the motion capture data, body velocity, trunk orientation, foot clearance and foot overhang were quantified. For all participants, as stair going decreased, gait velocity (ascent $p_A = 0.033$, descent $p_D = 0.003$) and horizontal step clearance decreased ($p_A = 0.001$), while trunk rotation ($p_D = 0.002$) and foot overhang increased ($p_{A,D} < 0.001$). Compared to the younger group, older participants used the handrail more, were slower across all conditions ($p_A < 0.001$, $p_D = 0.001$) and their foot clearance tended to be smaller. With a walking approach, the older group (*Group x Start* interaction) showed a larger trunk rotation ($p_A = 0.011$, $p_D = 0.015$), and smaller lead foot horizontal ($p_A = 0.046$) and vertical clearances ($p_D = 0.039$) compared to the younger group. A regression analysis to determine the predictors of foot clearance and amount of overhang showed that physical activity was a common predictor for both age groups. In addition, for the older group, medications and fear of falling were found to predict stair performance for most goings, while sway during single-legged standing was the most common predictor for the younger group. Older participants adapted to smaller goings by using the handrails and reducing gait velocity. The predictors of performance suggest that motor and fall risk assessment is complex and multifactorial. The results shown here are consistent with the recommendation that larger going and pausing before negotiating stairs may improve stair safety, especially for older users.

Keywords: stair negotiation, balance control, step going, fall risk, old people

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INTRODUCTION

Aging is a progressive process in which the physical and cognitive abilities deteriorate (Lord et al., 1996; Startzell et al., 2000), with a negative effect on motor performance and confidence whilst performing daily activities. Gait problems are common in old age (Lord et al., 1996; Jahn et al., 2010), and falls are usually associated with some deficits in the locomotor ability (Prince et al., 1997; Begg and Sparrow, 2000). Every year, 1/3 of individuals over 65 years old experience a fall (World Health Organisation, 2007). Indeed, falls are a major cause of morbidity in older people and the primary cause of accidental death (World Health Organisation, 2007; Age, 2012). Older people may experience difficulties because their locomotor pattern can become less efficient, in addition to impairments in their adaptive and recovery mechanisms (Rogers et al., 2003). Gait on stairs is a key example of this difficulty: the task is not only constrained (see below), it also places additional demands on the musculoskeletal and balance control systems, compared to level walking (McFadyen and Winter, 1998; Startzell et al., 2000; Riener et al., 2002; Reeves et al., 2008, 2009). Not surprisingly, a large number of dangerous falls occur during stair negotiation (Svanstrom, 1974; Jacobs, 2016).

Although the recommendations for stair rise height in private and public buildings are specific (170–220 mm), the UK guidelines prescribe goings (i.e., the horizontal distance between two consecutive step edges) between 220 and 400 mm (Government, 2010), which highlights large variation in recommendation. Stair going determines the antero-posterior area for foot placement and stride length, which are critical aspects for locomotion safety and fall risk. For stair safety, going dimensions should allow safe foot placement in descent, reducing foot overhang (that is, the portion of the foot that is not placed on the stair step), and allow the individual to develop an adequate push-off in ascent to propel the body upwards, increasing foot clearance (that is, the distance between edge of the foot and stair step). These are particularly important for older people because they may be less able to react if foot placement is not optimal, and they may be less able to lift the foot to clear a step as their strength reserves may be lower. Changing step going could improve safety on stairs for users, and older individuals specifically. However, understanding the motor adaptations in relation to intact and impaired balance performance is necessary before stricter guidelines can be suggested. For example, stride length is often adapted in level walking (Patla, 2003) in response to deterioration of neuro-musculo-skeletal health, balance ability and general physical wellness. However, stride length is constrained by the step dimensions on stairs. This is more problematic for sedentary older people because their declined neuromuscular and cardiovascular capacities are taxed by stair negotiation which requires moving the body center of mass forward and upward, against gravity (in ascent), and controlling balance and accurate foot placement, especially in descent.

Furthermore, a less efficient locomotor pattern can affect stair negotiation when preceded or followed by level gait. With a walking approach, the nervous system has to quickly respond to a change in the motor task and produce a quasi-feedforward programme (Patla, 2003). As the central and peripheral nervous system may become less efficient with aging, with impaired proprioception and worsened reaction times (Rogers et al., 2003), the need to change a motor programme quickly may introduce an additional motor control difficulty. Taken together, the neuro-musculo-skeletal difficulties and the constraints and demands imposed by stair negotiation could explain why older people are more prone to problems and accidents on stairs (Cavanagh et al., 1997).

In this study, we investigated the effect of changing the going (**Figure 1**) on stair negotiation performance and safety, by measuring key parameters including body orientation, velocity, foot clearance and overhang. We asked: (i) Does going size affect stair negotiation in older, more than in younger participants? (ii) Is the difference in performance of older and younger participants amplified in the case of a transition in motor tasks (i.e., level walking and stair negotiation)?

METHODS

Ethical Approval

Participants gave written informed consent to these experiments, which conformed to the Declaration of Helsinki and were approved by the ethics committee of the Institute for Biomedical Research into Human Movement and Health, Manchester Metropolitan University.

Participants and Procedure

Twenty-one young (thirteen men, eight women; mean ± standard error "SE" 29 ± 1 years; mass 77.2 ± 4.7 kg; height 1.75 ± 0.003 m) and 20 older participants (10 men, 10 women; 74 ± 1 years; 75.2 ± 4.3 kg; 1.66 ± 0.003 m) negotiated a stair (**Figure 1A**) at their self-selected speed. All participants were healthy and were recruited from the local community. Participants were included if they did not report musculoskeletal, neurological or cardiovascular pathologies, which would make stair negotiation risky. Participants were barefoot to minimize the influence of footwear on performance and on walking speed (Menz et al., 2003a), socks were not allowed to standardize friction between the stair and feet. The experiment was performed in a well-lit laboratory, with natural light from windows, and artificial light available when needed, but ambient light level was not controlled. Before the current protocol, each participant performed at least five stair negotiations on a different stair for familiarization. Before the session, participant's left and right leg and foot lengths were measured.

Participants performed four trials in a randomized order: ascent and descent from standing start, and ascent and descent preceded and followed by walking on a 2 m-walkway. Step going of the stair was also randomized for the four goings tested (175, 225, 275, and 325 mm). The experimental staircase was designed without protruding nosings. Stair rise was set at 175 mm, which is within the current recommendations for stair in private and public buildings (170–220 mm) (Government, 2010).

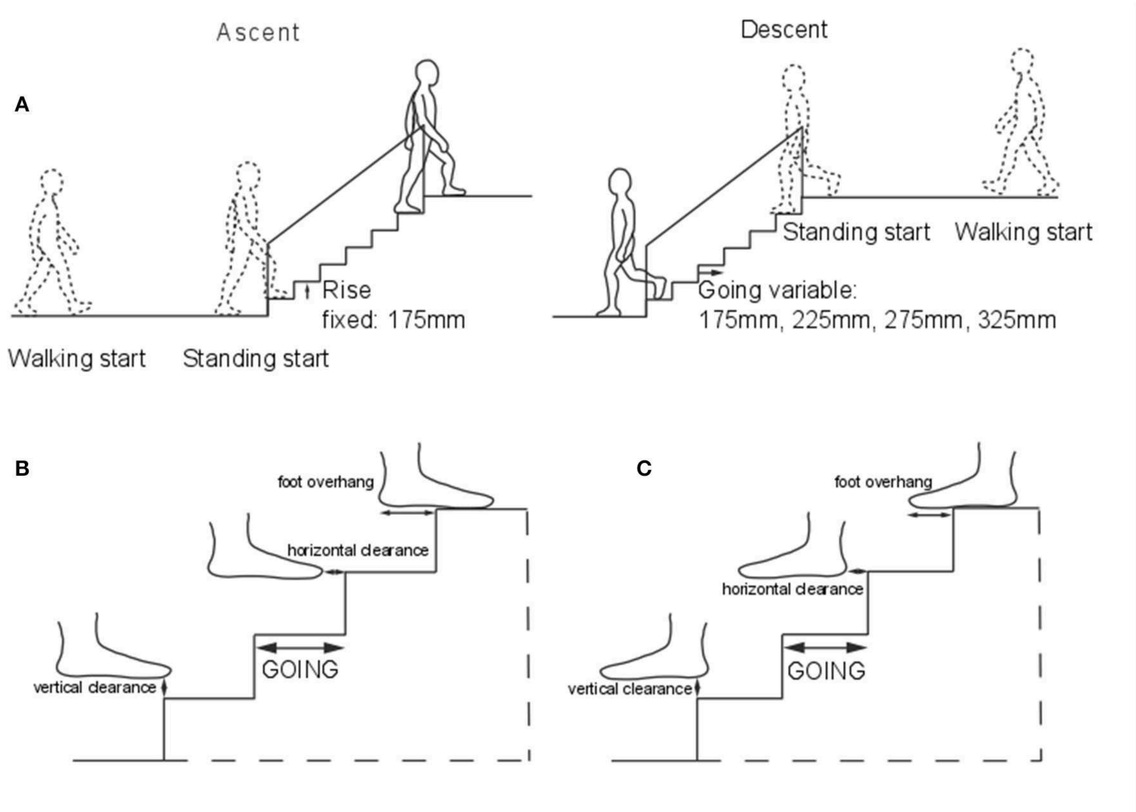

FIGURE 1 | Apparatus. **(A)** The seven-step instrumented stair. Step size: rise (height) 175 mm, going 175, 225, 275, and 325 mm. Four force platforms were embedded in steps 2–5. Handrails were provided on each side. Representation of the vertical and horizontal clearance and overhang in ascent **(B)**, and descent **(C)**.

Apparatus and Measurement

As the average foot size is 260 mm, an adjustable seven-step stair was used with the following going sizes: small that restricted whole foot placement (175 mm), current standard for domestic stairs (225 mm), standard for semi-public buildings (275 mm), and standard for public buildings that allowed comfortable whole foot placement (325 mm) (Roys, 2001).

The stair had four 300 × 500mm force platforms (model 9260AA3, Kistler Instrumente, CH-8408 Winterthur, Switzerland) embedded in the second, third, fourth and fifth steps. The force platforms were used to determine when the foot landed and lifted-off the step. Handrails were provided on both sides of the stair. A safety-harness system suspended from a trolley and girder on the ceiling of the laboratory was secured to the participant. The stair was situated in a volume covered by a 10-camera optoelectronic movement analysis system (Vicon Motion Systems, Oxford, UK). Retro-reflective markers (14 mm) were attached to the participant's skin or tight-fitting clothes at landmarks according to the Plug-In-Gait model, with additional markers on the fifth metatarsal head, the dorsal aspect of the second toe distal tip, on the lateral and medial aspects of the calcaneus and the medial malleoli. Kinematic data were collected at 100 Hz.

As the stair protocol lasted for about 3 h, further data was collected on a second visit. This data included fear of falling questionnaire, although here only an overall score is reported (0 = no fear, 5 = very high fear), self-reported hours of physical activity per week, and total medications taken (Lord et al., 2007). Three tests measured participants' balance using the ground reaction force data sampled at 1,000 Hz (AMTI, OR6-7, Watertown, MA, USA): (1) Standing on the self-selected leg with eyes open for 5 s (used to probe and exacerbate the balance challenges of the single support phase in stair negotiation), (2) quiet standing for 30 s with eyes open (EO), and (3) with eyes closed (EC).

Data Analysis

Each stair trial was visually inspected offline to record handrail use, body orientation and stepping method. Trials were initially assigned a nominal 0 for these indexes. If the participant touched one or both handrails, the trial was given a nominal value of 1 for handrail use. If the body was orientated toward one handrail, the trial was given a nominal value of 1 for change in orientation. If the participant placed both feet on one step, the trial was given a nominal value of 1 for change in stepping method.

The following quantities were calculated using Matlab scripts (Mathworks, Natick, US).

Mean Gait Velocity. The mean antero-posterior velocity of the center of mass of the upper body (trunk and head) over the whole stair. The 3D center of the upper body (four head markers, 7th cervical vertebra, 10th thoracic vertebra, right scapula, sternum and clavicle notch) was determined. The antero-posterior component of its position was extracted and differentiated to compute the velocity. The upper body was chosen to represent gait velocity, because the markers used were less affected by camera visibility obstruction from the stair in the large volume captured.

Trunk Orientation. The angle between the trunk antero-posterior axis and the direction of travel (from the laboratory coordinates), at foot landing (when the force plate signal first crossed a 10N threshold) in the horizontal plane. The average at steady-state (steps with force plates, 2–5) relative to the initial orientation of the trunk when the person was standing still (0–500 ms) was calculated. 0deg indicates no change in trunk orientation.

Foot Overhang. The antero-posterior foot portion landing outside the step (**Figures 1B,C**) as a percentage of the antero-posterior foot length on the step at steady-state (see above). The fore-foot was identified as the geometrical average of the markers placed on the second and fifth metatarsal head and the dorsal aspect of the second toe distal tip. The rear-foot was identified as the geometrical average of the markers placed on the heel and the lateral and medial aspects of the calcaneus. Markers' size was accounted for in the overhang calculations. The coordinates of the step edges were included in the algorithm for the calculations, based on the force platforms positions, included in the motion capture software. Negative values indicate overhang. Left and right feet overhang were averaged to provide a mean per trial.

Foot Clearance. The minimum distance between fore- and rear-foot (as calculated for foot overhang in ascent and descent, respectively, **Figures 1B,C**) and each step edge during swing, in the horizontal and vertical direction and for the lead (landing on the step) and trail limb (landing on the following step). The coordinates of the step edges were included in the algorithm for the calculations, based on the force platforms positions, included in the motion capture software. Clearances were calculated for each step and the average over the central steps (2–5) for each of the four clearances were also calculated. From the individual's step clearance, a coefficient of variation was calculated as an indicator of the repeatability and precision of foot placement.

For the balance tests, the Center of Pressure (CoP) was measured from the point of application of the ground reaction force to evaluate body sway. To evaluate balance abilities we calculated:

Single-Leg Balance. The root mean square (RMS) medio-lateral deviation of the CoP. A lower value indicates better control of balance.

Balance With Eyes Closed vs. Eyes Open (EC vs. EO). The ratio between the antero-posterior RMS CoP from the eyes closed and eyes open tests. A ratio >1 means a higher sway in the eyes closed condition.

Statistical Analysis

For all the statistical tests significance was set at $p \leq 0.05$. Results are reported as mean ± SE. Stair ascent and descent measures were analyzed separately using SPSS (ver.24, IBM). For handrail use, whole body orientation and stepping strategy, we ran Chi-Squared tests, with age-group and going as independent variables. For the other measures, a mixed linear model was used. Age group (2 levels: young, old), going (4 levels: 175, 225, 275, and 325 mm) and start-condition (2 levels: standing, walking) were fixed factors, whereas participant was the random factor. The three-way interactions are not reported here. Least significant difference (LSD) *post-hoc* test was used to investigate significant effects. For the single step clearance differences between age groups for each direction (ascent/descent) and start condition (standing/walking start) were assessed using an ANOVA test.

To compare between groups, left and right leg and foot lengths were averaged for each participant. The group mean and SE for height, leg and foot lengths were calculated. Additionally, the mean and SE were calculated for the data collected on the second visit (balance and questionnaires). The difference between younger and older participants in these quantities was then assessed using a *t*-test.

In order to determine the factors affecting stair performance, regression analyses were run for the younger and older group separately. The analyses were run for foot clearance in ascent, and foot overhang in descent. The factors included in the analyses were chosen to explain the possible influence on stair gait performance. For this reason, balance ability (Svanstrom, 1974; Tinetti et al., 1988; Lord et al., 2007) and hours of physical activity per week (proxy for physical ability) (Svanstrom, 1974; Tinetti et al., 1988; Lord et al., 2007) were used in ascent and descent. Additional factors, such as medications taken (Tinetti et al., 1988; Lord et al., 2007) and fear of falling (Lord et al., 2007) have been shown to be related to performance in stair descent and were included in the analysis. Additionally, mean foot length was included for stair descent to account for differences in anthropometric dimensions, which seem more relevant for foot placement in stair descent.

RESULTS

Stair Negotiation Performance

Ascent

At any given going, older participants used the handrail more often than younger participants (**Figures 2A,F**) in both standing start trials [*Group* Pearson χ^2: Going175 χ^2(df = 1, N = 41)=5.528, $p = 0.019$; G225 χ^2(1,40) = 11.053, $p = 0.001$; G275 χ^2(1,41) = 7.424, $p = 0.006$; G325 χ^2(1,40) = 7.802, $p = 0.005$] and walking start trials, except for 175 mm-going [*Group* G175 χ^2(df = 1, N = 41) = 1.977, $p = 0.160$; G225 χ^2(1,40) = 9.378, $p = 0.002$; G275 χ^2(1,41) = 12.108, $p = 0.001$; G325 χ^2(1,39) = 4.692, $p = 0.030$]. The younger group used the handrail mainly for the smallest going [*Going* standing-start χ^2 (df = 3, N = 84) = 12.205, $p = 0.007$; walking-start χ^2 (3,83) = 19.095, $p < 0.001$]. Body orientation was not affected by group

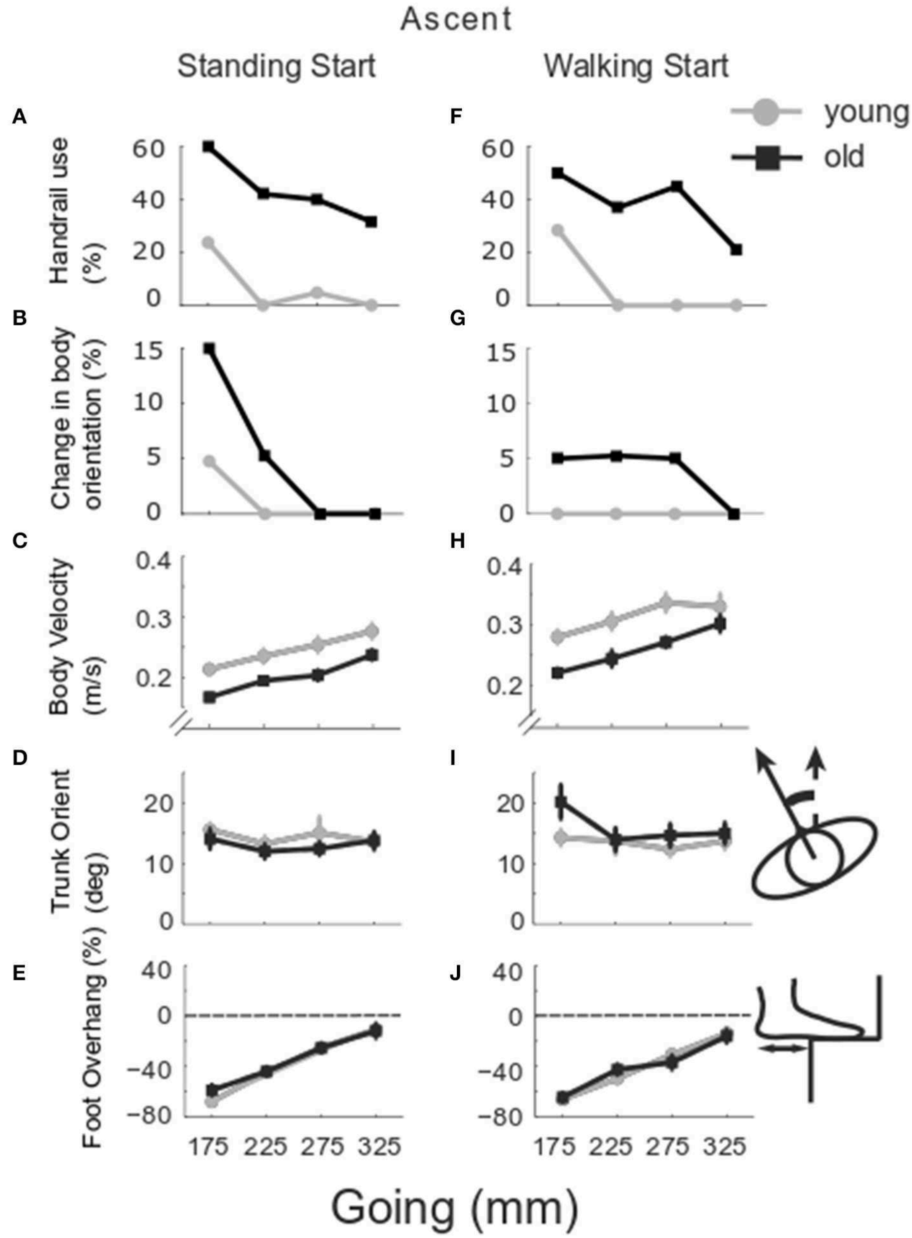

FIGURE 2 | Group stair performance in ascent. The percentage of trials in which the handrail was used **(A,F)** and the whole body turned toward one handrail are reported **(B,G)** for older (black) and younger group (gray). The group mean and SE of body velocity **(C,H)**, trunk orientation **(D,I)** and foot overhang relative to foot length **(E,J)** are reported for the goings investigated −175, 225, 275, and 325 mm from standing (left column) and walking start (right column). Trunk orientation was calculated relative to the trunk position whilst standing at the beginning of the recording and used as the baseline (0deg here). Foot overhang is reported as % of antero-posterior foot length, 0% means that the whole foot is placed on the step, negative values show the percentage of foot outside the step at foot landing.

or going (**Figures 2B,G**). Participants always negotiated the steps with alternate feet.

Gait velocity was slower in older participants (*Group* $p < 0.001$), from a standing start (*Start* $p < 0.001$), and for smaller goings (*Going* $p = 0.033$) (**Figures 2C,H**). Trunk orientation (**Figures 2D,I**) changed with start condition (*Start* $p = 0.045$) and the older group rotated the trunk more in the walking start trials (*Group x Start* $p = 0.011$). Foot overhang (**Figures 2E,J**) increased with smaller goings (*Going* $p < 0.001$).

Lead foot horizontal clearance (**Figures 3A,E**) was smaller in the older group (*Group* $p < 0.001$) and decreased as going decreased (*Going* $p = 0.001$). The older group showed a larger clearance in standing start trials (*Group x Start* $p = 0.046$). Lead vertical clearance (**Figures 3B,F**) was smaller in the older group for the smaller goings (*Going x Group* $p = 0.034$). Trail

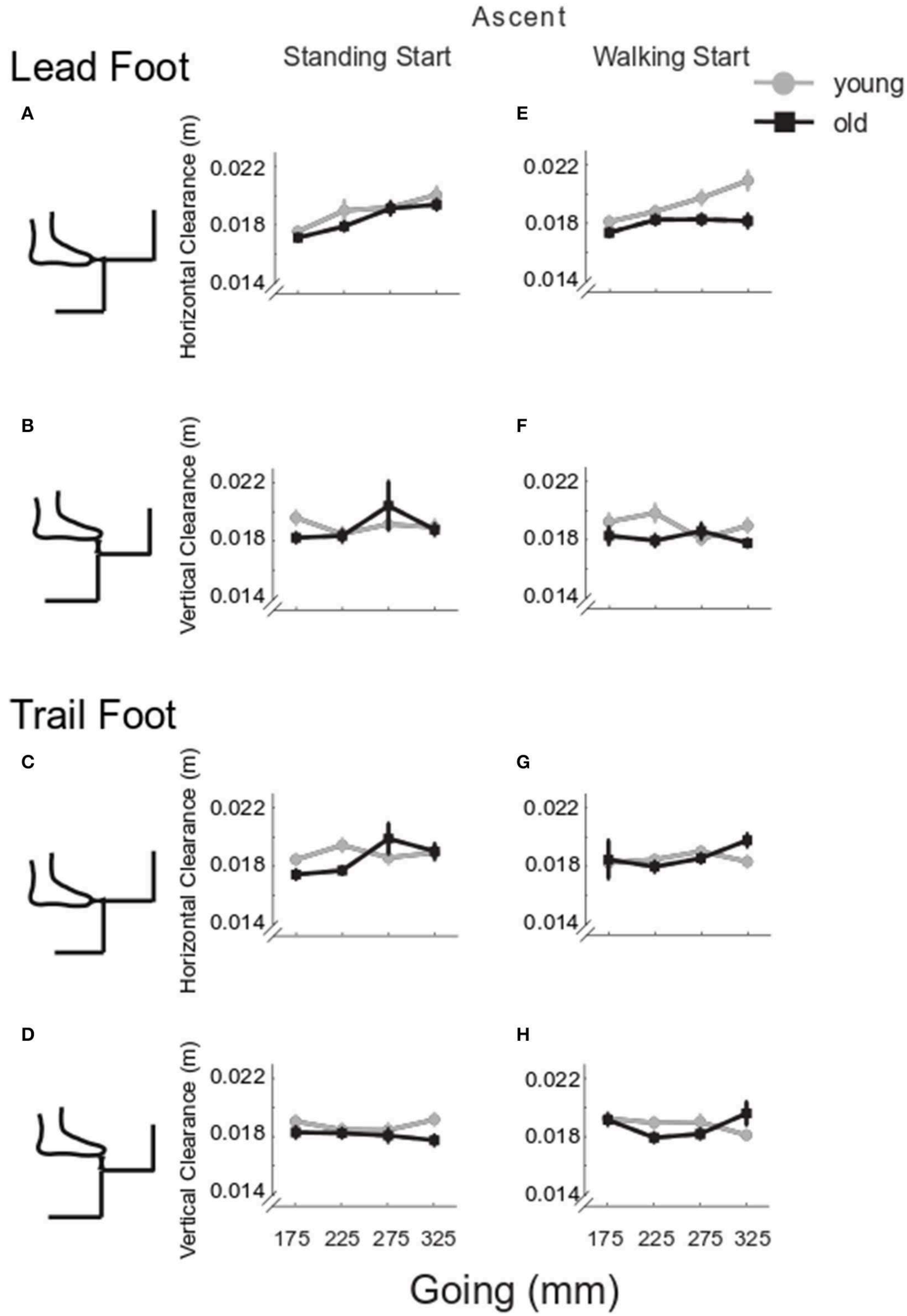

FIGURE 3 | Group clearance in ascent. The group mean and SE for older (black) and younger (gray) clearance in stair ascent was calculated in the horizontal direction (direction of travel) and in the vertical direction for lead and trail foot, at the four goings. The mean values were calculated whilst negotiating the stair from a standing start (left column) and from a walking start (right column). Lead foot clearance in horizontal **(A,E)** and vertical direction **(B,F)**; trail foot clearance in horizontal **(C,G)** and vertical direction **(D,H)**.

foot horizontal clearance was smaller in the older group for the smaller goings (*Going x Group* $p = 0.017$) (**Figures 3C,G**), and the vertical clearance showed a significant *Going x Start* interaction ($p = 0.006$) indicating that the smaller goings differed between the two start conditions (G175 $p = 0.035$; G225 $p = 0.002$; G275 $p = 0.003$) (**Figures 3D,H**). Examining all the configurations for both standing and walking start, when a significant difference in clearance was found at a step, usually the clearance in the younger group was greater than the older group. The central steps (2–4) showed the majority of occurrences of significant differences between the younger and older groups (**Figure 4**, left columns).

The coefficient of variation in foot clearances is shown in **Figure 5**. No significant group differences for horizontal lead foot clearance could be found (**Figures 5A,E**, *Group* $p = 0.219$). However, the coefficient of variation was larger for the standing start condition (*Start* $p = 0.008$). In addition, a going effect was found (*Going* $p < 0.001$) indicating that the 325 mm-going induced a higher coefficient of variation then all the other goings. A *Going x Start* interaction was also found ($p = 0.001$) and *post-hoc* analysis showed that the 325 mm-going induced a higher variation only for the standing start trials. Similar results were found for the horizontal clearance coefficient of variation for the trail leg (**Figures 5B,F**) as clearance variation was larger for the standing start condition (*Start* $p < 0.001$), and for the largest going G325 (*Going* $p < 0.001$), but there was no significant group effect (*Group* $p = 0.284$). A significant *Going x Start* interaction ($p = 0.001$) indicated that the 325 mm-going induced a higher variation only for the standing start trials.

For the vertical clearances (**Figures 5C,D,G,H**), the coefficient of variation did not show any significant group, start condition or going effects for the lead foot (*Group* $p = 0.950$, *Start* $p = 0.078$, *Going* $p = 0.952$) or the trail foot (*Group* $p = 0.951$, *Start* $p = 0.230$, *Going* $p = 0.887$).

The regression analyses (**Figure 6**) for the older participants, showed that vertical clearance in the standing start trials at a going of 175 mm was lower for participants who reported a higher number of hours of physical activity [**Figure 6C**, $F_{(1, 18)} = 9.3613$, $p = 0.0067$, $R^2 = 0.3421$, correlation coefficient $b = -0.7399$], whilst at a going of 325 mm, the clearance was higher for participants who had a higher score in balance EC vs. EO [**Figure 6D**, $F_{(1, 17)} = 10.2991$, $p = 0.0051$, $R^2 = 0.0925$, $b = 2.8092$]. Vertical clearance in the walking start trials was lower for the participants reporting more hours of physical activity per week at a going of 275 mm [**Figure 6H**, $F_{(1, 18)} = 4.4934$, $p = 0.0482$, $R^2 = 0.3457$, $b = -0.9418$] or 325 mm [**Figure 6I**, $F_{(1, 17)} = 5.3088$, $p = 0.0341$, $R^2 = 0.8556$, $b = -0.4810$]. Horizontal clearance in walking start trials was lower for participants reporting fewer hours of physical activity at a going of 325 mm [**Figure 6E**, $F_{(1, 17)} = 5.5676$, $p = 0.0305$, $R^2 = -1.3141$, $b = -0.8736$].

For the younger group, the step-wise regression model showed that the horizontal foot clearance in standing start trials at a going of 175 mm [model $F_{(2, 15)} = 6.7106$, $p = 0.0083$] increased for participants with larger the single-leg balance score (**Figure 6A**, step-1 of the regression $b = 0.0760$, $p = 0.0066$, $R^2 = -0.3833$), or larger balance EC vs. EO score (**Figure 6B**, step-2 of the regression $b = 1.7499$, $p = 0.0193$, $R^2 = 0.1104$). Horizontal clearance in walking start trials was greater for participants with a larger balance EC vs. EO score at a going of 175 mm [**Figure 6F,** $F_{(1, 16)} = 5.0875$, $p = 0.0385$, $R^2 = 0.3181$, $b = -1.8319$), or larger single-leg balance score at a going of 275 mm [**Figure 6G,** $F_{(1, 16)} = 5.2194$, $p = 0.0363$, $R^2 = -0.9008$, $b = -0.0848$).

Descent

At any given going, the older group used the handrail more often than younger participants (**Figures 7A,F**) in both standing start trials [*Group* G175 χ^2 (df = 1, $N = 41$) = 7.411, $p = 0.006$; G225 $\chi^2(1,40) = 11.465$, $p = 0.001$; G275 $\chi^2(1,41) = 14.435$, $p < 0.001$; G325 $\chi^2(1,40) = 12.835$, $p < 0.001$] and walking start trials [*Group* G175 χ^2(df = 1, $N = 41$) = 9.058, $p = 0.003$; G225 $\chi^2(1,40) = 18.947$, $p < 0.001$; G275 $\chi^2(1,41) = 18.814$, $p < 0.001$; G325 $\chi^2(1,39) = 8.980$, $p = 0.003$]. The younger group used the handrail more often for the smallest goings, especially for G175 [*Going* standing-start χ^2(df = 3, $N = 84$) = 18.616, $p < 0.001$; walking-start $\chi^2(3,83) = 22.570$, $p < 0.001$]. Whole body orientation (**Figures 7B,G**) was affected by going for the younger group for both start conditions [*Going* standing-start χ^2(df = 3, $N = 84$) = 11.596, $p = 0.009$; walking-start $\chi^2(3,83) = 11.793$, $p = 0.008$], whilst for the older group, it was only affected in the walking start trials [*Going* χ^2(df = 3, $N = 78$) = 13.585, $p = 0.004$]. At a going of 175 mm, the older group placed both feet on the same step in 10 and 5% of the standing and walking start trials, respectively, and in 5.3% of the standing start trials at a going of 325 mm. No relationship between change in stepping strategy and age group or going was found.

Figures 7C,H show that body velocity was lower in older participants (*Group* $p = 0.001$), from a standing start (*Start* $p = 0.001$) and for smaller goings (*Going* $p = 0.003$). Change in trunk orientation (**Figures 7D,I**) was larger in older participants (*Group* $p = 0.017$) and increased in both groups as going decreased (*Going* $p = 0.002$). The older group rotated the trunk more in walking start trials (*Group x Start* $p = 0.015$). Foot overhang was larger for the smaller goings (*Going* $p < 0.001$) (**Figures 7E,J**). For the trail foot (**Figures 8C,D,G,H**), an interaction was found for the vertical clearance (*Going x Start* $p = 0.039$) indicating that only the 275 mm-going was different between the two start conditions. No significant differences were found for the lead foot (**Figures 8A,B,E,F**). Examining the single step for all the configurations for both standing and walking task (**Figure 4**, right columns), step 1 (at the end of stair negotiation in descent) showed the majority of occurrences of differences between younger and older groups with two instances in which the clearance in the older group was higher than the younger group (horizontal lead foot stand start at 175 mm $p = 0.001$, and vertical trail foot stand start at 275 mm $p = 0.028$).

There was no statistically significant difference in coefficient of variation for horizontal lead foot clearance (**Figures 9A,E**, *Group* $p = 0.400$). However, the coefficient of variation was larger for standing start trials (*Start* $p = 0.034$). In addition, a going effect was found (*Going* $p < 0.001$) indicating that the 325 mm-going induced a higher coefficient of variation then all the other goings. A significant *Going x Start* interaction ($p = 0.045$) indicated that 325 mm-going induced a higher variation only for the standing start trials. Similar results were found for the horizontal trail leg (**Figures 9B,F**), as variation was larger

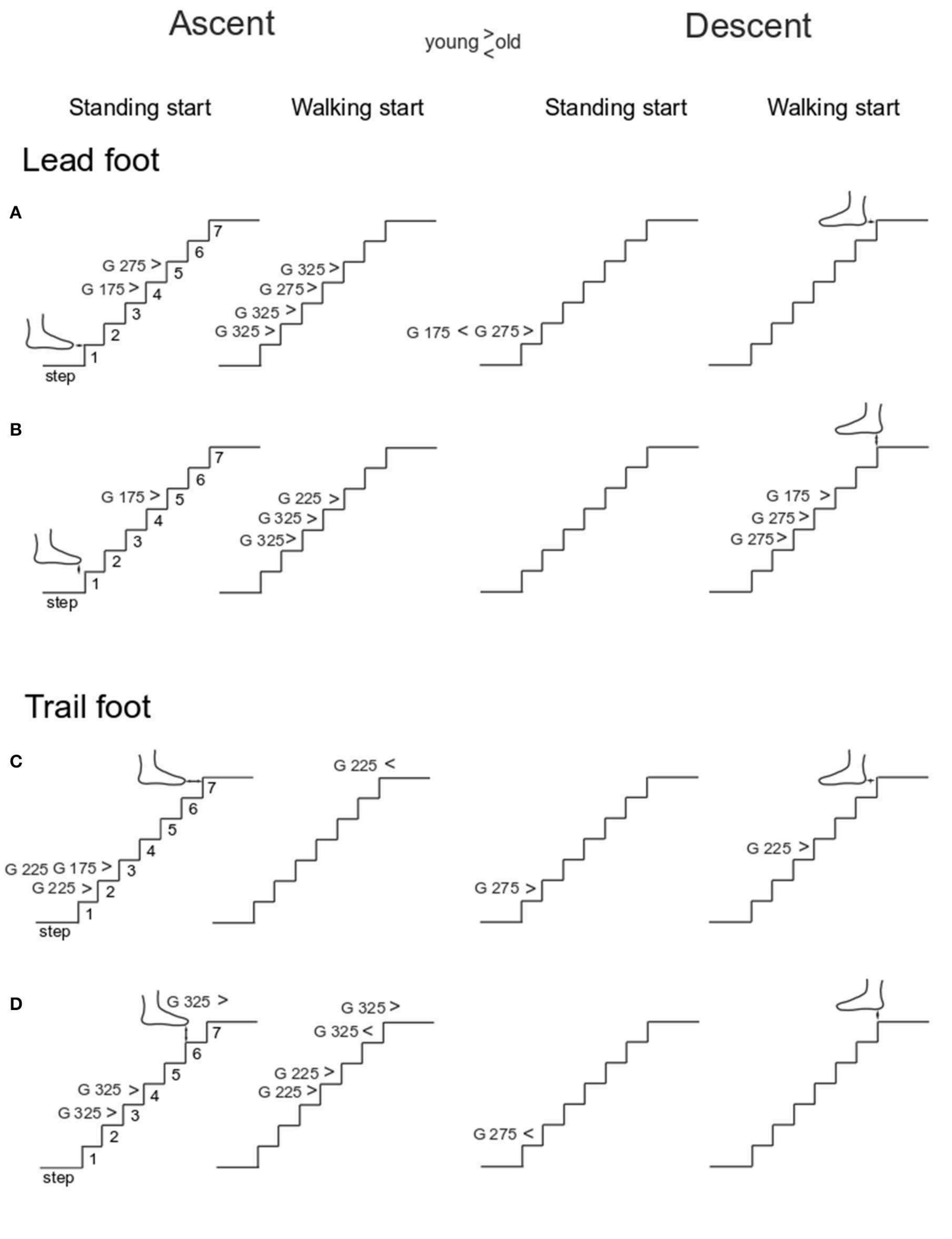

FIGURE 4 | Single step clearance. The group average of foot clearances was calculated for each step and an ANOVA test was run to find differences between younger and older groups for each step in each stair configuration. The results reported here are divided in four columns: ascent standing start, ascent walking start, descent standing start and descent walking start. For each panel, the significant difference ($p < 0.05$) is reported with the indication of which going (G) showed the significant results and whether the younger group's means was higher (>) or lower than older group's mean (<). The four rows show: Lead foot clearance in horizontal **(A)** and vertical direction **(B)**; trail foot clearance in horizontal **(C)** and vertical direction **(D)**.

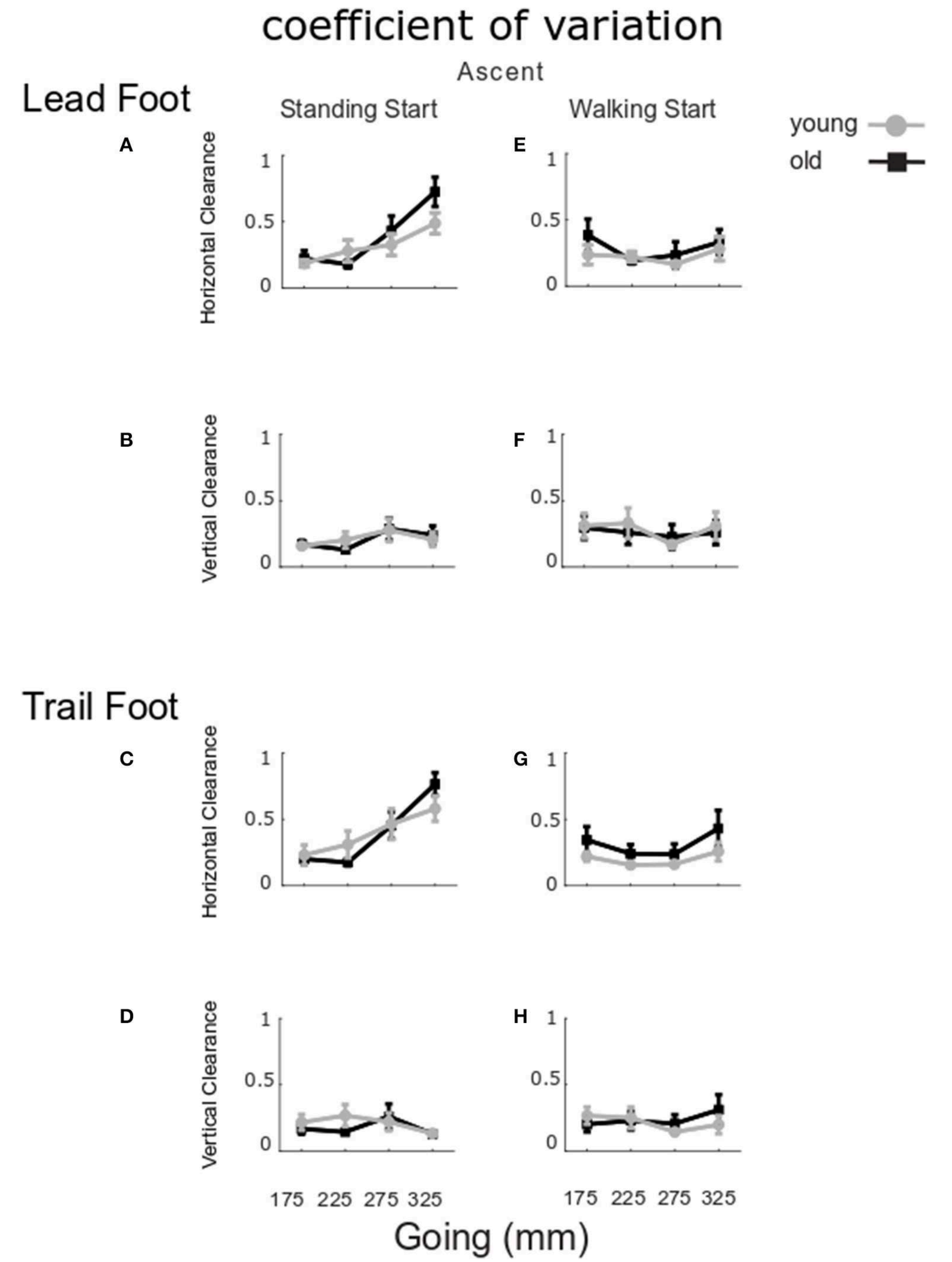

FIGURE 5 | Coefficient of variation for clearance in ascent. The group mean and SE for older (black) and younger (gray) coefficient of variation for clearance in stair ascent at the four goings. The values were calculated for standing start (left column) and walking start (right column). Coefficient of variation for lead foot clearance in horizontal **(A,E)** and vertical direction **(B,F)**; trail foot clearance in horizontal **(C,G)** and vertical direction **(D,H)**.

for the standing start condition (*Start* $p = 0.004$), and for the largest going G325 (*Going* $p < 0.001$), but no differences between groups could be found (*Group* $p = 0.429$). A significant *Going x Start* interaction ($p < 0.001$) indicated that the 325 mm-going induced a higher variation only for the standing start trials.

For the vertical clearances (**Figures 9C,D,G,H**), the coefficient of variation did not show any significant group, start condition or going effects for the lead foot (*Group* $p = 0.523$, *Start* $p = 0.948$, *Going* $p = 0.698$) or the trail foot (*Group* $p = 0.681$, *Start* $p = 0.804$, *Going* $p = 0.446$).

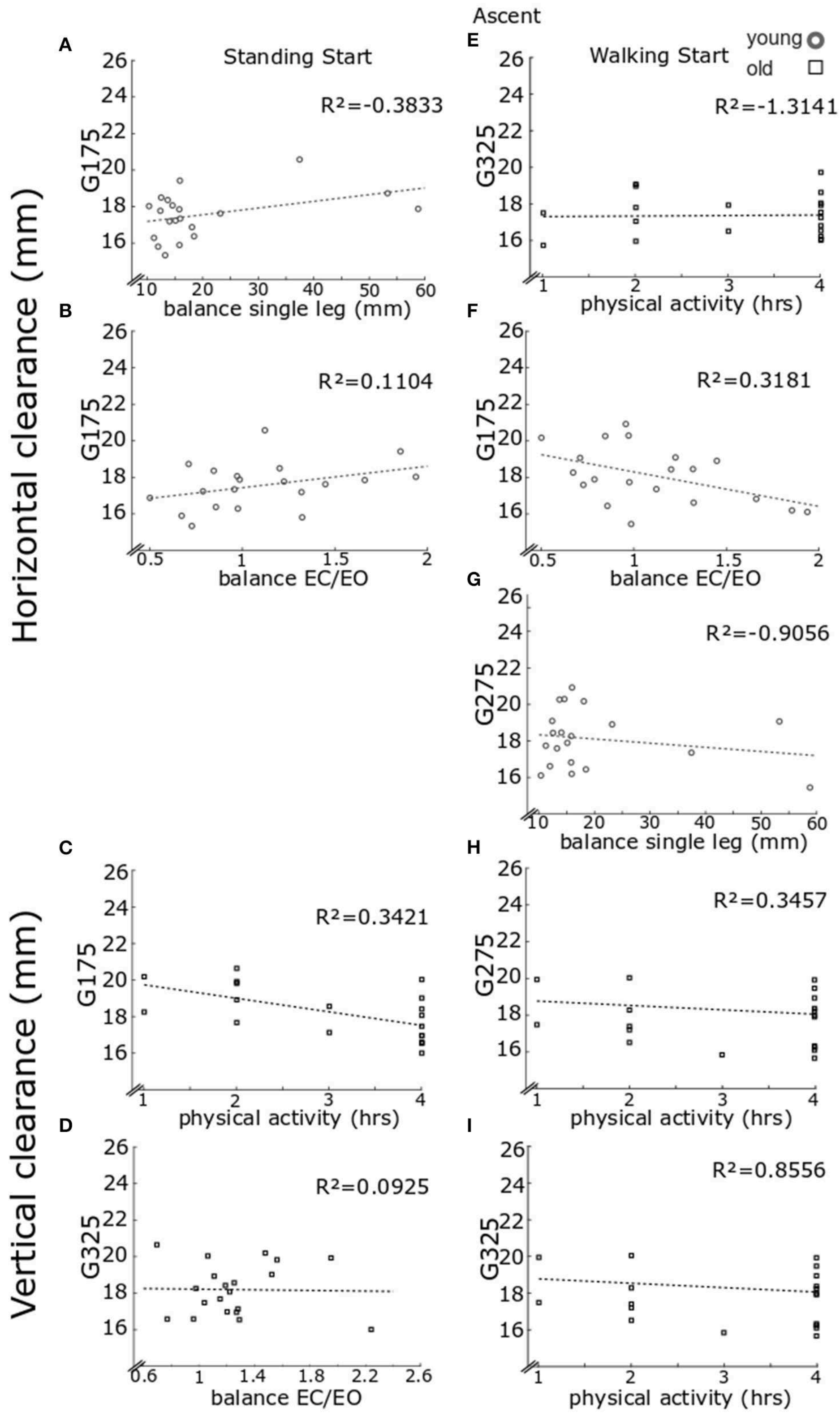

FIGURE 6 | Regression model for clearance in ascent. Lead foot clearance regression model results. Each panel reports individual data for older (black) and younger (gray) clearance in stair ascent in the horizontal direction and in the vertical direction. As only the lead foot showed significant results from the regression models, only

(Continued)

FIGURE 6 | these data are reported. Results are reported for standing start (left column) and walking start (right column). Standing start. Lead foot clearance in the horizontal direction for going of 175 mm for younger participants relative to balance score on one leg **(A)** and balance eyes closed over eyes open **(B)**. Lead foot vertical clearance for going of 175 mm for older participants relative to hours of physical activity **(C)** and for going of 325 mm relative to balance eyes closed over eyes open **(D)**. Walking start. Lead foot clearance in the horizontal direction for going of 325 mm for older participants relative to hours of physical activity **(E)**. Clearance for going of 175 mm for younger participants relative to balance eyes closed over eyes open **(F)** and at going of 275 mm relative to balance score on one leg **(G)**. Lead foot vertical clearance for older participants relative to hours of physical activity for going of 275 mm **(H)** and for going of 275 mm **(I)**. For each panel a least square fit line and the R^2 values are reported.

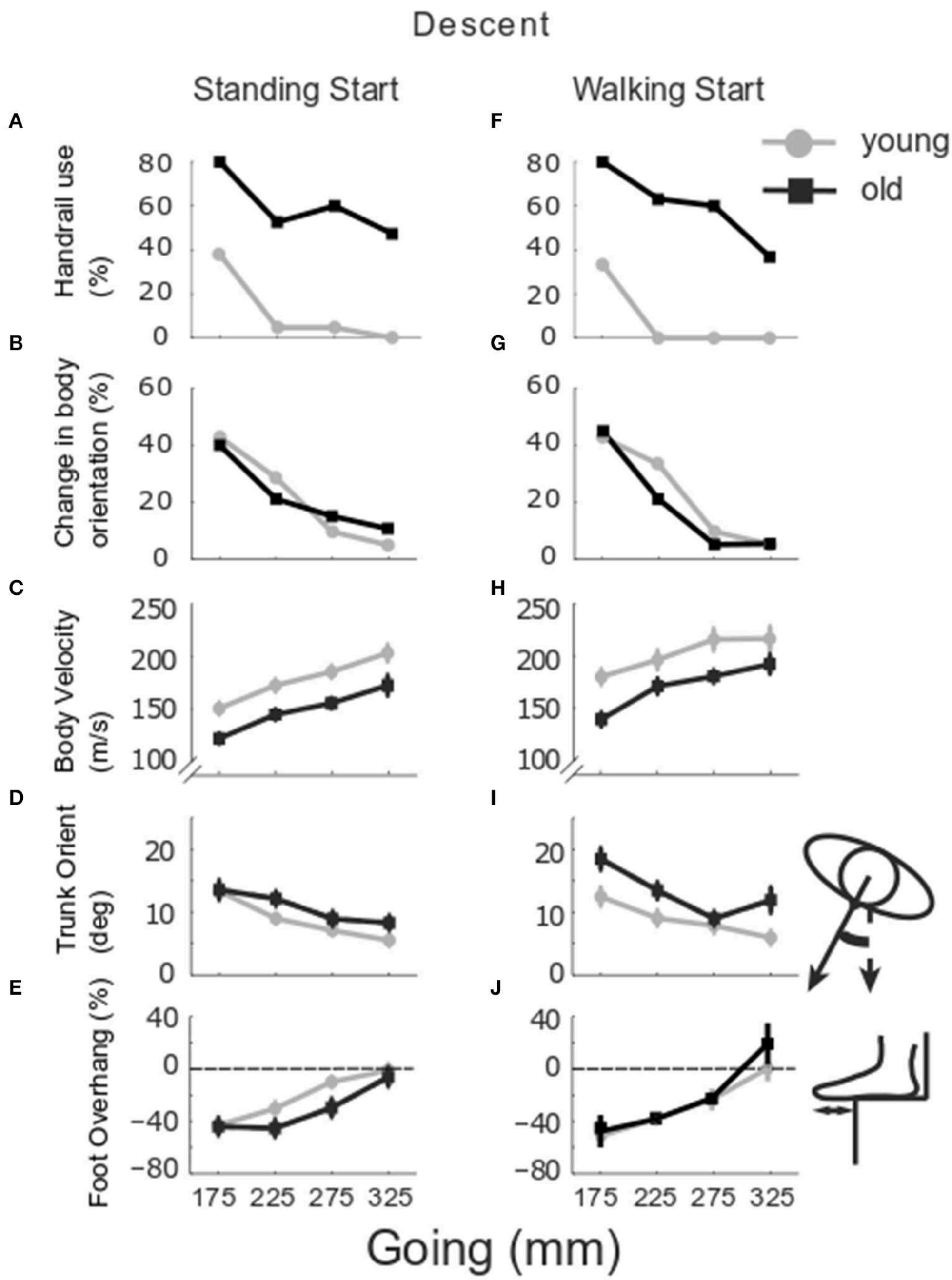

FIGURE 7 | Group stair performance in descent. The percentage of trials in which the handrail was used **(A,F)** and the whole body turned toward one handrail is reported **(B,G)** for older (black) and younger group (gray). Group mean and SE of body velocity **(C,H)**, trunk orientation **(D,I)** and foot overhang **(E,J)** relative to stair going, in standing (left column) and walking start trials (right column).

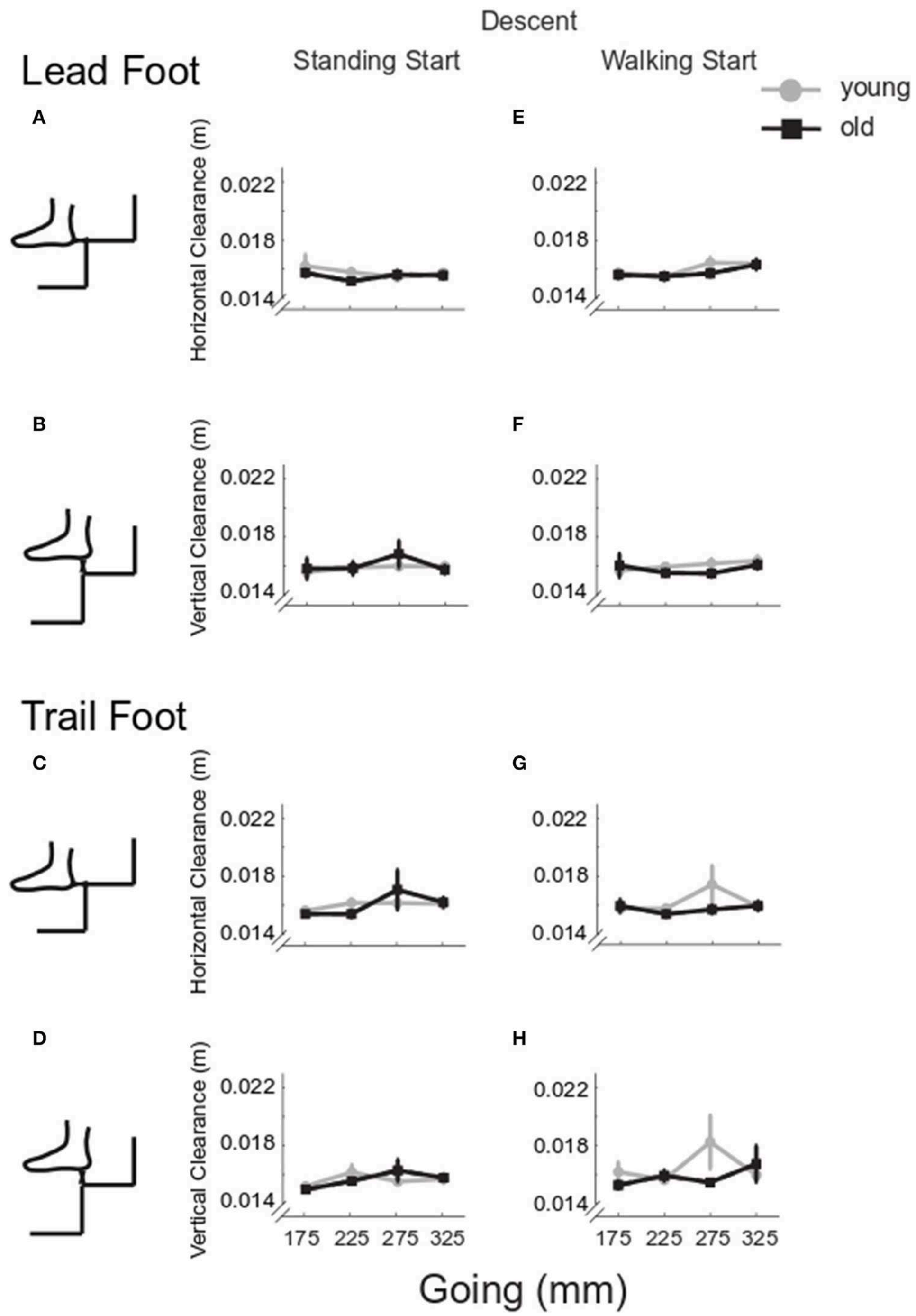

FIGURE 8 | Group clearance in descent. Group mean and SE for older (black) and younger (gray) lead and trail foot horizontal **(A,C,E,G)** and vertical clearance **(B,D,F,H)** in stair descent at the four goings, from a standing (left column) and a walking start (right column).

The regression analyses (**Figure 10**) showed that for the older participants, foot overhang in standing start trials was larger for participants who reported taking fewer medications at a going of 175 mm [**Figure 10A**, $F_{(1,\,16)} = 6.6058$, $p = 0.0205$, $R^2 = 0.2922$, $b = 3.8815$] and at going of 325 mm in walking trials [**Figure 10D**, $F_{(1,\,15)} = 8.3254$, $p = 0.0113$, $R^2 = 0.0956$, $b = 4.2408$]. In walking start trials foot overhang was larger for participants reporting a lower score for fear of falling at a going of 175 mm [**Figure 10C**, $F_{(1,\,15)} = 5.1884$, $p = 0.0378$, $R^2 = 0.2570$, $b = 15.8498$].

For the younger group, foot overhang in standing start trials at a going of 275 mm was larger for participants with a higher score in the single-leg balance test [**Figure 10B**, $F_{(1,\,15)} = 24.6760$, $p < 0.001$, $b = -0.8750$, $R^2 = 0.9199$] and, in walking start trials

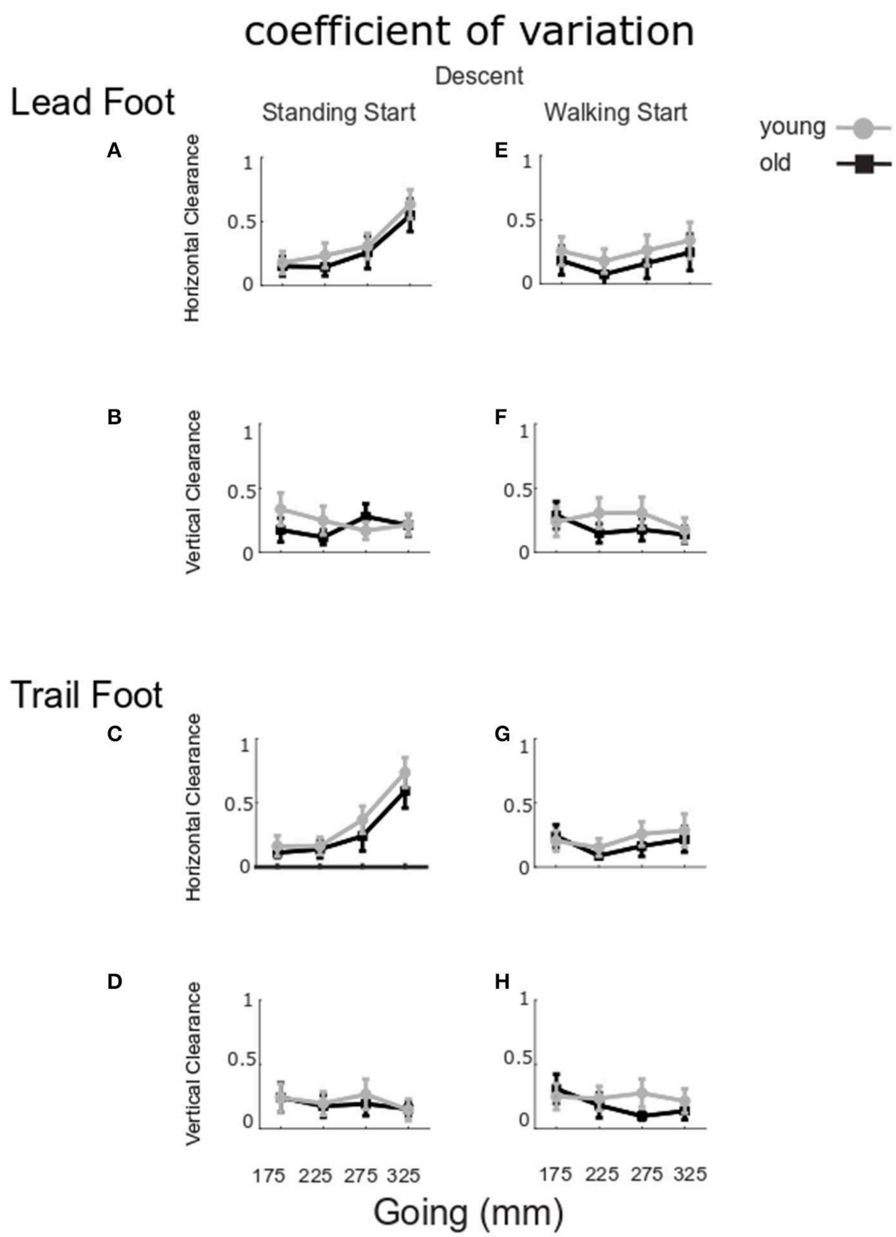

FIGURE 9 | Coefficient of variation for clearance in descent. The group mean and SE for older (black) and younger (gray) of the coefficient of variation for lead and trail foot horizontal **(A,C,E,G)** and vertical clearance **(B,D,F,H)** in stair descent at the four goings, from a standing (left column) and a walking start (right column).

[$F_{(2, 14)} = 7.3830$, $p = 0.0065$] for participants reporting higher hours of physical activity (**Figure 10E**, step-1 of the regression $b = -11.0459$, $p = 0.0128$, $R^2 = 0.3305$) or a higher score in the balance EC vs. EO test (**Figure 10F**, step-2 of the regression $b = -26.8983$, $p = 0.0199$, $R^2 = 0.5461$). Foot overhang at a going of 325 mm in walking trials [$F_{(2, 13)} = 9.6329$, $p = 0.0027$) was predicted by hours of physical activity with larger overhang for higher number of hours (**Figure 10G**, step-1 of the regression $b = -15.2187$, $p = 0.0017$, $R^2 = 0.3841$) or smaller foot length (**Figure 10H**, step-2 of the regression $b = -4.6799$, $p = 0.0276$, $R^2 = 0.5748$).

Functional Capability Assessments

Although older participants' height was lower than the younger group (mean ± standard error "SE" younger 1.75 ± 0.003 m, older 1.66 ± 0.003 m, $p = 0.0206$), no statistical difference between groups was found for leg length (younger 0.854 ± 0.017 m, older 0.833 ± 0.016 m, $p = 0.3745$) or foot length (younger 0.259 ± 0.045 m, older 0.251 ± 0.041 m, $p = 0.2317$).

Single-leg balance. The medio-lateral RMS of CoP was greater in the older group than the young group (older 0.079 ± 0.017 m, younger 0.020 ± 0.003 m, $p < 0.001$) (**Figure 11A**).

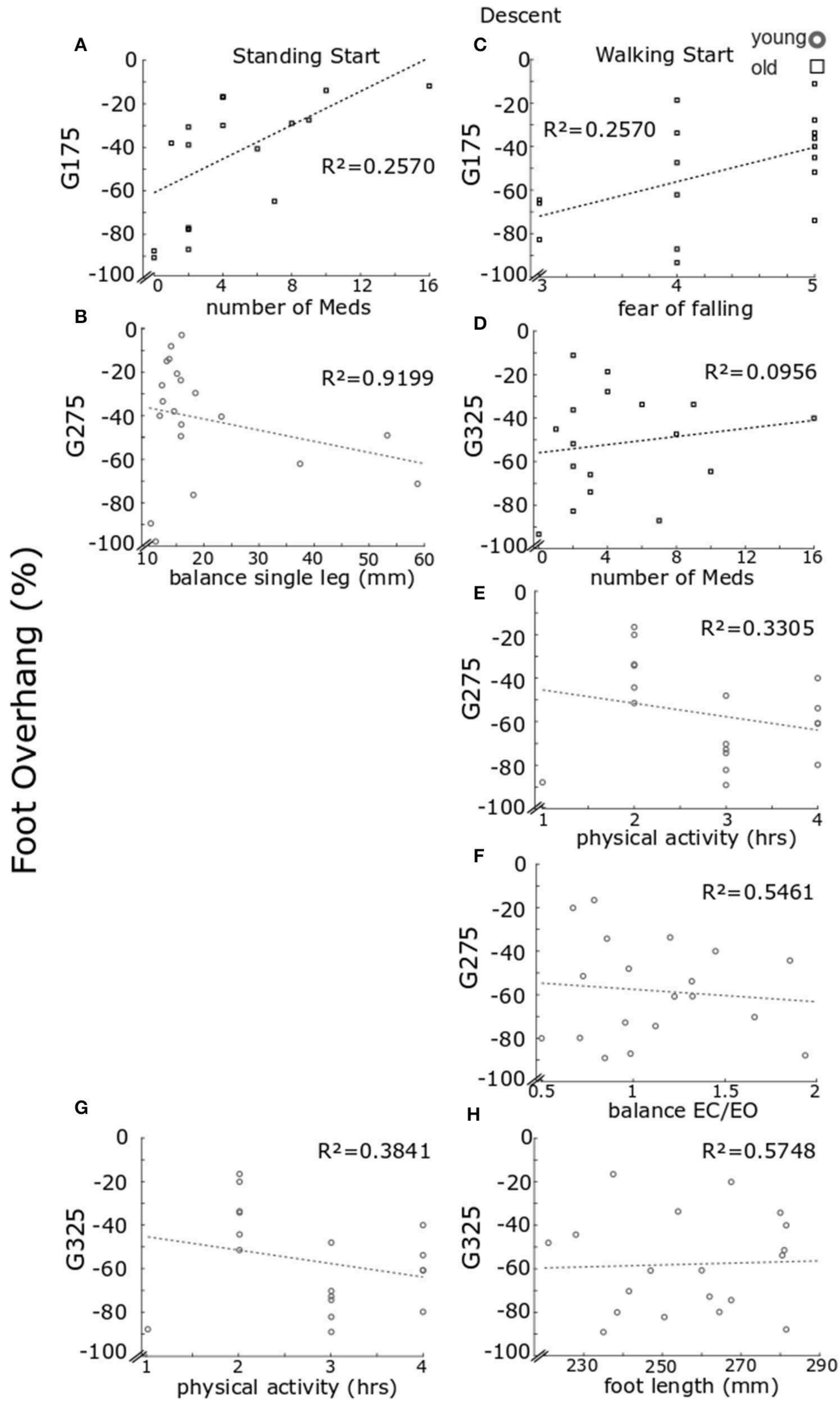

FIGURE 10 | Regression model results for foot overhang in descent. Regression model results for foot overhang. Each panel reports individual data for older (black) and younger (gray) participants. Results are reported for standing start **(A–B)** and walking start **(C–H)**. Standing start. Overhang for going of 175 mm for older participants relative to number of medications taken **(A)** and for younger participants at goings of 275 mm relative to balance on one leg **(B)**. Walking start.

(Continued)

FIGURE 10 | Foot overhang for older participants at going of 175 mm relative fear of falling score **(C)** and at going of 325 mm relative to number of medications taken **(D)**. For the younger individuals, foot overhang at going of 275 mm relative to hours of physical activity **(E)** and balance eyes closed over eyes open **(F)**; while at going of 325 mm relative to hours of physical activity **(G)** and foot length **(H)**. For each panel a least square fit line and the R^2 values are reported.

Balance EC vs. EO. With EC, the antero-posterior RMS of CoP was greater in the older group (older 0.005 ± 0.0005 m, younger 0.0041 ± 0.0002 m, $p = 0.035$). There was no difference between age groups with EO (older 0.004 ± 0.0003 m, younger 0.0041 ± 0.0003 m, $p = 0.907$) (**Figure 11B**) and for the ratio of RMS CoP between EC and EO (older 1.26 ± 0.08, younger 1.10 ± 0.09, $p = 0.398$) (**Figure 11C**).

Questionnaires. Weekly hours of physical activity were similar between groups (older 3.1 ± 0.3 h, younger 2.7 ± 0.2 h, $p = 0.2913$). The older group reported a higher fear of falling (older 0.7 ± 0.2, younger 0.2 ± 0.1, $p = 0.0185$) and a higher number of medications taken (older 4.4 ± 0.9, younger 0.2 ± 0.1, $p < 0.001$).

DISCUSSION

Older and younger participants negotiated an experimental stair with different step going sizes and two different methods of approaching the flight of stairs. Here we discuss the age differences in stair performance in response to going manipulation and start condition.

Differences Between Younger and Older Participants on Stairs: Effect of Aging

Older participants used the handrail more often, had a lower gait velocity and smaller foot clearances. These results are unlikely to be related to the participants' anthropometry because the only group difference was a small difference in height, with no difference in leg or foot length. However, the adaptations shown by the older group may be underpinned by an aging-induced deterioration of musculoskeletal capabilities (Prince et al., 1997; Menz et al., 2003b; Iosa et al., 2014). In ascent, demands on the musculoskeletal system are heightened, because the body mass is moved against gravity. Using the handrail helps in propelling the body upwards. Considering that hours of physical activity was a predictor of older participants' clearance in ascent, it is likely that older participants used the handrail more to compensate for their real or perceived reduced muscle strength. In fact, counterintuitively, the participants that reported a higher level of physical activity showed lower clearances and higher overhang suggesting an increase in confidence or a more tuned strategy, but potentially closer to the risk level. As older individuals tend to employ a higher joint moment on stairs relative to their maximum (Reeves et al., 2008, 2009) compared to younger individuals, it is possible that a higher fitness level is related to a more efficient strategy, without using excessive energy to perform the task successfully.

Older participants also used the handrail more often in descent. The muscle strength demands of descent are lower because no work against gravity is needed, and eccentric muscle strength is better preserved in older people (Roig et al., 2010). However, using the handrails helps with stability, particularly in the single support phase. The need for additional support is consistent with the predictors of older participants' performance in descent which are related to their confidence and overall health level (fear of falling and medications). Consistent with older participants' lower physical abilities and confidence, the older group self-selected stair walking speed was slower, probably to use more time to perform accurate stepping and better cope with the demands of the task, allowing them to produce larger joint forces (relative to their maximum capability) at slower velocities.

Older participants successfully adapted to the changes in environment without accidents. Indeed, the change in going size affected both groups similarly. The smallest going tested here, 175 mm, appeared to be challenging for both groups, as also the younger individuals occasionally used the handrail in both ascent and descent. This may seem at odds with the finding that older people had larger medio-lateral sway in the single leg balance test, particularly as stair negotiation involves periods when the body center of mass is either being lowered or elevated during single stance. These results add to the debate on the relevance of static balance tests as useful predictors of fall risk, as there is no general consensus on the relationship between static balance tests and real-life fall risk (Menz et al., 2003a; Lord et al., 2007; Granata and Lockhart, 2008).

An indication that the system was close to a risk-threshold was displayed by the older group's smaller foot clearance, particularly at smaller goings (Going x Group interaction for horizontal trail foot and vertical lead foot clearances). Changes in step going alter the demands of the task, and may increase the likelihood of accidents and injury (Roys, 2001; Novak et al., 2016), particularly if users are less able to accurately assess these demands, which is a problem for older individuals due to the slowly progressing (rather than acute) decline in neuro-musculo-skeletal capabilities. Although the older group's clearance at 175 mm-going indicated a heightened risk, we showed that this going was the most challenging one for both groups. Unexpectedly, we did not find a reliable effect of the 225 mm or the 325 mm goings. The 225 mm-going is just within the UK guidelines for private buildings and does not allow complete foot placement, on average. For this reason, we expected a potential effect of this going, but our tests only showed a quasi-linear trend for most of the quantities measured, with stair gait becoming less affected as the going was larger. On the other hand, a potential drop in locomotor performance at the largest going (325 mm) could be expected because this going may impose a larger than comfortable stride. However, in this study we showed that 325 mm-going did not seem to worsen performance, but only increased variability in the task as shown by the higher coefficient of variation for horizontal clearance. More

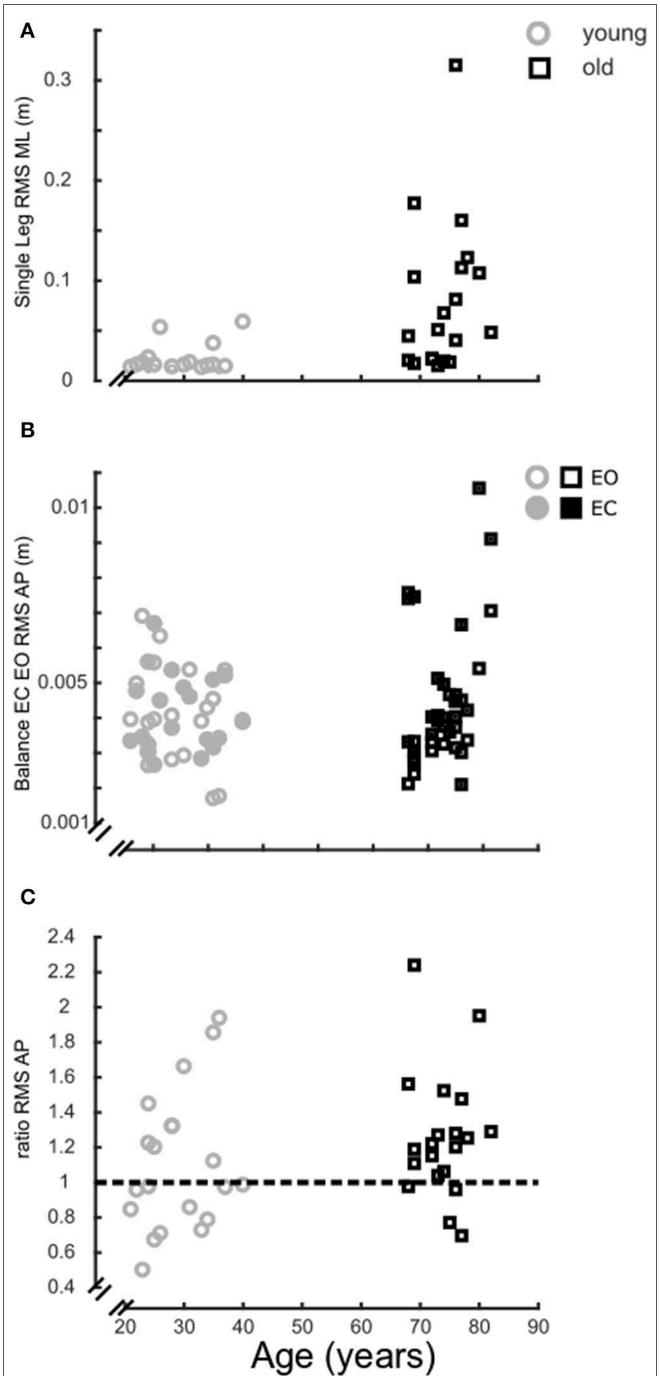

FIGURE 11 | Balance ability. Balance performance over age, grouped in older (black) and younger participants (gray). **(A)**, Medio-lateral root mean square (RMS) of the CoP trace of the single leg balance trial (5 s); **(B)**, Antero-posterior RMS of the CoP trace for the quiet standing trials (30 s) with eyes closed (filled) and eyes open (open); **(C)**, Ratio of the antero-posterior RMS CoP for the quiet standing trials with eyes closed over eyes open trials.

work is needed to determine the precise relationship between going size and participants' anthropometry and the role of an increased variability in efficient motor control. It is also noteworthy that in order to identify reliable predictors of stair falls, precise quantification of performance on the stairs and its relationship with other tests should be investigated, as static standing tests were not able to account for the differences in stair performance tested here. This is needed to investigate the mechanisms of accidents to improve falls prevention, particularly in older people.

Older Group's Increased Difficulty With Walking Start Condition

Here standing and walking start were compared to investigate two different modes of initiation and termination of locomotion, the effect of disrupting the rhythmicity of the task, and the ability to react to changes in environment (Wollacott and Tang, 1977; Menz et al., 2003a). Older participants seemed to experience more difficulties with a walking start (Group × Start interactions); in fact, they rotated their trunk more and had a smaller horizontal lead foot clearance. In this experiment, the transition in tasks could be planned because participants were free to see the staircase before negotiating it and the visual input was not manipulated in any way. For this reason, the older group's adaptations are unlikely to be related to a reduced reaction time in this group. However, with the standing start condition, the initiation of movement requires a process of adjustments and anticipatory reactions and control (Patla, 2003) that destabilizes the system in order to allow movement. This is reflected in the higher coefficient of variation for clearance in ascent and descent. On the other hand, in the walking start condition, a transition between two locomotor processes is needed. This would predict a more optimal strategy in walking start trials. However, the motor tuning necessary to change the motor task may be difficult for older participants, particularly considering the higher body velocity and consequent higher momentum, both of which require control during stair negotiation, but especially in the single support phase in stair descent. This extra level of control may increase the complexity of the overall control in walking start trials, which could explain the increased difficulty for older participants in this study. Therefore, pausing before negotiating a staircase after level walking (either before or in between flights of stairs on a level landing) may make stair walking safer by allowing more time to assess the environment, plan the motor task, and subsequently execute it in a less risky body posture and at lower momentum.

In conclusion, in this study we have shown the difference in stair gait according to going dimension in younger and older individuals. We have found that smaller goings induced significant adaptations in both groups, and healthy older participants showed motor adaptations and strategies consistent with increased difficulty, compared to the younger cohort. We also found that older participants showed additional difficulties when stair negotiation was preceded by level gait, as a transition in motor control was required in the two tasks. This suggests that stair design should allow comfortable gait for everyone, and in particular for older individuals, and that pausing before negotiating a staircase could be a safer strategy.

DATA AVAILABILITY STATEMENT

The datasets generated for this study are available on request to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Institute for Biomedical Research into Human Movement and Health, Manchester Metropolitan University. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

The experiments were performed at the Laboratory of Biomechanics, School of Healthcare Science, Manchester Metropolitan University, Manchester, UK. ID, NR, VB, and CM contributed to the conception of the experiment, data acquisition, analysis, and interpretation. ID wrote the article. All authors contributed to design of the work, contributed to the critical review of the manuscript, and approved the final version.

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The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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ORIGINAL RESEARCH
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Can Smartphone-Derived Step Data Predict Laboratory-Induced Real-Life Like Fall-Risk in Community-Dwelling Older Adults?

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Background: As age progresses, decline in physical function predisposes older adults to high fall-risk, especially on exposure to environmental perturbations such as slips and trips. However, there is limited evidence of association between daily community ambulation, an easily modifiable factor of physical activity (PA), and fall-risk. Smartphones, equipped with accelerometers, can quantify, and display daily ambulation-related PA simplistically in terms of number of steps. If any association between daily steps and fall-risks is established, smartphones due to its convenience and prevalence could provide health professionals with a meaningful outcome measure, in addition to existing clinical measurements, to identify older adults at high fall-risk.

Objective: This study aimed to explore whether smartphone-derived step data during older adults' community ambulation alone or together with commonly used clinical fall-risk measurements could predict falls following laboratory-induced real-life like slips and trips. Relationship between step data and PA questionnaire and clinical fall-risk assessments were examined as well.

Methods: Forty-nine community-dwelling older adults (age 60–90 years) completed Berg Balance Scale (BBS), Activities-specific Balance Confidence scale (ABC), Timed Up-and-Go (TUG), and Physical Activity Scale for the Elderly (PASE). One-week and 1-month smartphone steps data were retrieved. Participants' 1-year fall history was noted. All participants' fall outcomes to laboratory-induced slip-and-trip perturbations were recorded. Logistic regression was performed to identify a model that best predicts laboratory falls. Pearson correlations examined relationships between study variables.

Results: A model including age, TUG, and fall history significantly predicted laboratory falls with a sensitivity of 94.3%, specificity of 58.3%, and an overall accuracy of 85.1%. Neither 1-week nor 1-month steps data could predict laboratory falls. One-month steps data significantly positively correlated with BBS ($r = 0.386$, $p = 0.006$) and ABC ($r = 0.369$, $p = 0.012$), and negatively correlated with fall history ($r_p = -0.293$, $p = 0.041$).

Conclusion: Older participants with fall history and higher TUG scores were more likely to fall in the laboratory. No association between smartphone steps data and laboratory fall-risk was established in our study population of healthy community-dwelling older adults which calls for further studies on varied populations. Although modest, results do reveal a relationship between steps data and functional balance deficits and fear of falls.

Keywords: fall prediction, steps data, smartphone technology, falls, older adults

INTRODUCTION

Falls are a common and a serious problem in older adults aged 60 years and above (Rubenstein, 2006; Carpenter et al., 2019). Even the healthiest community-dwelling older adults are not immune to falls, especially on exposure to external environmental perturbations such as slips and trips which accounts for 60% of outdoor falls among community-dwelling older adults aged 70 years and above (Berg et al., 1997; Luukinen et al., 2000; Crenshaw et al., 2017). Such falls occur due to age-related physiological changes resulting in balance dysfunction, reduced muscle strength, and impaired gait pattern, predisposing older adults to high fall-risk (Ambrose et al., 2013; Zhao et al., 2018). Additionally, with progressing age, older adults experience a decline in physical function and activities of daily living which further increases fall-risk (Smee et al., 2012; Welmer et al., 2017). Falls result in several deleterious physical consequences such as fractures and soft tissue injuries but also lead to fear of falling, thereby resulting in further self-imposed restriction of physical activity (PA) (Pereira et al., 2014; Young and Mark Williams, 2015).

Apart from aging, a fall-risk factor which is non-modifiable, PA which can mitigate age-related declines in muscle strength, balance, and agility is considered a modifiable risk-factor. Due to its adaptability, PA could be systematically monitored and enhanced in the community-living geriatric population to reduce fall-risk. Although sparse, evidence using the self-reported questionnaire, Physical Activity Scale for the Elderly (PASE), demonstrated that fallers had lower PASE scores (less PA) compared to non-fallers and that the low scores were associated with high fall-risk and fear of falling (Roig et al., 2011; Oliveira et al., 2017). Such a questionnaire-based assessment can serve as an inexpensive tool to assess PA among community-dwelling older adults over a period of 7 days (Washburn et al., 1993; Logan et al., 2013; Duray and Genc, 2017). However, it has several limitations. Firstly, self-reported techniques show recall bias, especially when used by older adults with possibly declining memory. Additionally, the need to give socially desirable answers can affect the accuracy of results (Perell et al., 2001). Finally, studies suggest that PASE might have a floor effect because several activities listed in PASE, such as outdoor gardening, yard work, painting, and wall papering, might not be commonly performed activities by older adults (Sallis and Saelens, 2000).

Contrary to self-reported measures, wearable sensor technology comprising of research-based (ActivPal and ActiGraph) and commercially available (FitBit and Apple watch) motion sensors automatically track and store PA and thus effectively combat the issue of recall-bias. Such accelerometer-based wearable sensors are able to identify PA patterns (frequency, duration, and intensity) under both controlled laboratory conditions and uncontrolled, realistic conditions of daily living (Plasqui and Westerterp, 2007; Gomersall et al., 2016; Rosenberger et al., 2016). Additionally, commercially available wearable sensor technology records and stores PA simply in terms of number of steps for several months and years. Thus, commercially available wearable sensors have gained popularity for PA monitoring in both young and older adult populations. The advantage of such technology is that step count can be easily interpreted by older adults themselves or by clinicians and used for comparative analysis by researchers. Among various PA parameters (distance covered, number of steps, and energy expenditure), step data has become the hallmark measure of PA monitoring (Tudor-Locke et al., 2005). One probable reason that steps are being used to represent PA is because walking is one of the most commonly reported forms of activity performed even among sedentary older adults (Paillard et al., 2004). In addition, other than its significant health benefits, walking has become a focus for public health interventions because of its feasibility and acceptability (Li et al., 2005).

Although commercially available accelerometer-based wearable sensors have several advantages, they have demonstrated reduced long-term compliance in older adults (compared to the younger population) due to the need of carrying an extra device and sometimes the cost associated with it (Marschollek et al., 2011; Ferrari et al., 2012). Thus, with the advancement in smartphone technology, the latter have replaced or supplemented wearable sensors for PA monitoring. Smartphones equipped with tri-axial accelerometers can be used in conjunction with inbuilt or freely available smartphone applications for measuring and recording step data. Studies indicate that smartphone applications similar to wearable sensor technology can deliver step data in a user-friendly interface (Higgins, 2016; Lu et al., 2017). For example, Harries et al. reported that participants had greater adherence to using the smartphone applications than wearing their wearable devices because running the smartphone application was quite simple and did not require much effort (Harries et al., 2016). In addition to their easy implementation, studies demonstrated that the accuracy of step data collected using smartphones was just as high as accelerometer-based sensors (Higgins, 2016; Lu et al., 2017).

Although the significance of PA as a modifiable fall-risk factor is established (Chan et al., 2007; Pereira et al., 2008), to the authors' knowledge there is limited evidence of association between daily life ambulation and fall-risk. Studies that demonstrated the association of daily-life ambulation and fall-risk (Rispens et al., 2015; van Schooten et al., 2015), utilized wearable-sensor technology and determined various daily-life ambulatory gait parameters that can predict fall-risk in older adults. However, they did not take number of steps into consideration. Limited studies have considered using number of steps as a parameter to identify fallers. Brodie et al. (2015) found that shorter ambulatory periods with fewer steps recorded using a wearable sensor can identify older adult fallers. However, another study by Weiss et al. (2013) did not find any significant difference between fallers and non-fallers based on the number of steps. Secondly, most fall-risk prediction studies involve retrospective or prospective data of real-life falls collected subjectively via a fall diary. Such a method can introduce a recall bias on the number and type of falls (i.e., the cause of falls). Usually, a large sample of participants are needed to be monitored consistently over a long period of time to collect sufficient fall events. With recent advances in technology, it is possible to reproduce slip and trip-like falls that closely resemble those encountered in daily life (albeit in safe laboratory conditions) to determine the prognostic capacity of various fall-risk measures (Bhatt et al., 2011). Thus, such experimental method allows for an immediate and quantifiable investigation of participants' susceptibility to certain types (i.e., slip and trip) of falls over a short test session.

In summary, free-living steps data, an easily accessible aspect of PA, collected by smartphone could largely increase the feasibility of studies on PA in the geriatric population under realistic community-living conditions. With the ability to reproduce real-life like falls in a laboratory environment, we could explore the association between daily steps data and fall-risk without the need to conduct a longitudinal study with community-based monitoring. If any association between daily steps taken and fall-risk is established, smartphone-based PA monitoring could provide health professionals with a meaningful outcome measure, in addition to existing clinical measurements, to better identify older adults at high fall-risk.

Thus, this study examined whether smartphone-derived steps data either as a single factor or along with other commonly used clinical fall-risk measures could predict laboratory-induced slip or trip related fall-risk in older adults. We also examined if the steps data would correlate with the questionnaire-based PA assessment, the PASE, and with other commonly used clinical fall-risk measures. Additionally, the study included a sub-analysis to determine the prediction capacity of both 1-week step data vs. 1-month step data to better understand the time dependency, if any on the predictive and associative relationships of step data.

METHOD

Participants

Community-dwelling ambulatory healthy older adults were recruited within a 50-mile radius from the laboratory in the city and the neighboring suburbs of the Greater Chicago Area. The study participants were recruited through advertisements via study flyers distributed at different senior centers, community exercise centers, and independent senior living facilities. Participants were included in the study if they were at least 60 years old, weighed <250 pounds, received a cognitive score of >25 on the Folstein Mini Mental Status Exam (MMSE), possessed a smartphone with the "Google Fit" application for Android phones or "Health" application for iPhones for steps data collection, and if they had installed and enabled their respective application for at least 1 month prior to screening. Exclusion criteria included participants with acute (<6 months) musculoskeletal conditions such as back pain or fracture or having a surgical history 6 months prior to the laboratory perturbation test. Seventy-six participants agreed to participate in this pilot study and were included in the initial screening. Participants were excluded if they did not pass the initial screening test ($n = 9$), did not have entire 1 month smartphone data ($n = 8$) or if the smartphone steps data was not recorded even if the participants mentioned that they carried their phones ($n = 5$) or had incomplete laboratory data due to missing markers ($n = 5$). Ultimately, 49 participants were included in the final analysis (**Table 1**). All participants provided written informed consent and this study was approved by the Institutional Review Board.

TABLE 1 | Sample demographics and baseline clinical measurements with the mean and standard deviations.

Variables	Fallers (*N* = 36)	Non-fallers (*N* = 13)
	Mean (*SD*)	Mean (*SD*)
Age (y)	71.72 (5.56)	66.92 (5.15)
Weight (lbs)	160.73 (30.41)	152.84 (30.77)
Height (m)	1.64 (0.81)	1.71 (0.85)
TUG (s)	8.10 (1.23)	7.64 (1.21)
BBS (out of 56)	53.69 (2.05)	53.46 (2.36)
Fall history (%)	47%	38%
ABC (%)	87.80 (11.47)	85.16 (12.55)
PASE	129.57 (62.60)	159.54 (70.29)
MMSE	29.5 (0.77)	28.92 (1.65)
1-week steps	30534 (17637.5)	34286 (18544.5)
1-month steps	131528 (79996.9)	151004 (79328.5)
- <1,00,000 steps	62257.5 (23476.2)	62139 (27319)
- 1,00,000–2,00,000	139899.6 (33210)	145497 (26911)
- >2,00,000 steps	270834 (55744.2)	256759.7 (39458)

Study Design, Protocol, and Outcome Variables

On the initial screening day, all participants underwent various clinical measures to assess their balance [Berg Balance Scale (BBS)], balance confidence [Activities-specific Balance Confidence scale (ABC)], functional mobility [Timed Up-and-Go test (TUG)], and PA [Physical Activity Scale for the Elderly (PASE)]. The BBS assesses balance during functional tasks, and the scores range from 0 to 56 with higher scores indicating better balance and lower fall-risk (Whitney et al., 1998; Steffen et al., 2002). The ABC scale assesses balance confidence across 16 activities, and the scores range from 0 to 100% with higher percentages indicating a higher level of balance confidence (Powell and Myers, 1995; Myers et al., 1998). The TUG score

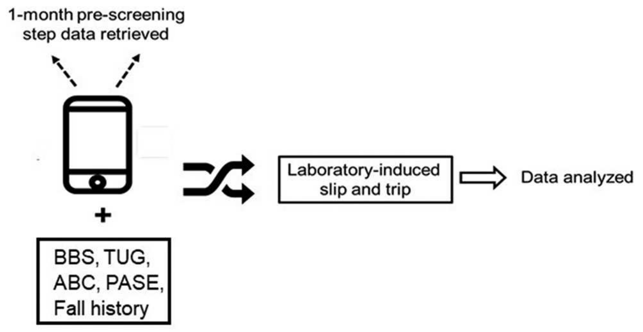

FIGURE 1 | This figure demonstrates the study protocol. Participants were subjected to various fall-risk assessment measures such as Berg Balance Scale (BBS), Activities specific balance confidence scale (ABC), Physical activity scale for elderly (PASE), and previous 1-year fall history. One-month smartphone steps data was also retrieved. Participants were subjected to various fall-risk assessment measures such as the Berg Balance Scale (BBS), Activities specific Balance Confidence scale (ABC), Timed Up and Go (TUG), Physical Activity Scale for Elderly (PASE) and previous 1-year fall history. One week and one-month smartphone steps data was also retrieved.

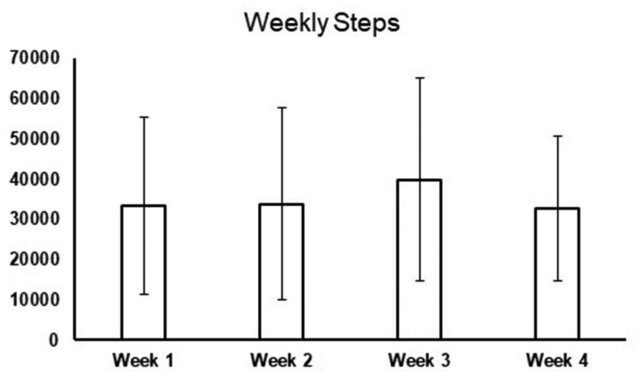

FIGURE 2 | This figure demonstrates the total number of steps walked each week for 4 weeks (1 month) to show that 1-month steps data provided consistent information regarding the participant's physical activity behavior throughout 4 weeks.

represents the time taken to stand up from a chair, walk a distance of 3-m, turn around, and sit back on the chair. Higher scores represent greater time taken to complete the test, indicating poor functional mobility, and high fall-risk (Shumway-Cook et al., 2000; Bohannon, 2006). The PASE scores were calculated from weights and frequency values for each of the 12 types of activities, and a higher score indicates greater PA (Washburn et al., 1993). Self-recalled fall history for the past 1 year was also obtained (**Figure 1**).

In addition, on the day of the initial screening the total steps for the past 1-week and 1-month were retrieved and summed from participants' smartphones. **Figure 2** presents the number of steps walked each week for 4 weeks to indicate that consistent data was collected thereby depicting their consistent PA behavior. Based on the sum of 1-month step data collected for each participant, we classified the data in 3 sub-categories (<1,00,000 steps, 1,00,000–2,00,000 steps, and more than 2,00,000 steps). A qualitative questionnaire including four questions were asked (1) number of hours the participant carries his or her phone, (2) the time of day when the participant is most active, (3) whether the participant carries his or her phone all the time they were active, and (4) whether he or she owns a wearable fitness tracker such as FitBit or Apple watch. All participants were then scheduled to receive the laboratory slip and trip perturbations within 2 weeks of the initial screening date (**Figure 1**).

Laboratory Fall Test

During the laboratory session, participants were assigned to receive a novel slip and trip perturbation in random order. Participants first had to walk 25–35 unperturbed baseline trials to become familiar with the laboratory walking environment before receiving their novel slip or trip perturbation. Each participant received a single slip and a single trip given in a random order. Thus, the participants experienced a total of two perturbations while undergoing the laboratory test. They were informed "a slip or trip may or may not occur during walking." The starting position was adjusted during baseline walking to ensure the upcoming slip or trip trials were induced properly. The slip was induced by a pair of low-friction movable platforms imbedded in a 7-meter walkway. These platforms were mounted on top of low-friction aluminum tracks resting on four force plates (AMTI, Newton, MA). The unannounced release of the platform occurred at heel strike of the perturbed (right) limb, and, following the platform's release, it was free to slide a distance of up to 60 cm (Wang et al., 2019a). Such slip distance has been reported to be enough to induce a fall in older adults (**Figure 3A**). The laboratory trip was induced on the left side by a hinged metal plate imbedded in the same walkway. During regular walking, the plate was locked in a flat position by a pair of electromagnets. For the trip trial, the electromagnets that kept the metal plate in a flat position were powered off to unlock the plate and the springs returned the plate to an upright position to induce a trip when the vertical ground reaction force (GRF) under the unperturbed (right) limb exceeded 80% of the participant's body weight after right heel strike (**Figure 3B**; Wang et al., 2012). All participants were protected by a safety harness connected through a load cell (Transcell Technology Inc., Buffalo Grove, IL) to a low-friction trolley-and-beam system mounted to the ceiling along the walking path. A fall was determined if the load cell detected more than 30% of the participant's body weight after the slip or trip (Yang and Pai, 2011). Additionally, the perturbation outcome was determined to be a fall based on the video recording if the participant was visually observed to have fallen after the novel slip or trip.

Statistical Analysis

Data was summarized using descriptive statistics (means and standard deviations) for all variables including demographics, that is, age, height and weight, fall-risk measurements such as BBS, TUG, ABC, PASE as well as previous 1-year fall history and 1-week and 1-month smartphone steps. In addition, means and standard deviations were also calculated for each sub-category based on 1-month data (<1,00,000 steps, 1,00,000–2,00,000 steps, and more than 2,00,000 steps). Paired t-tests were performed between faller and non-faller groups for overall 1-week and 1-month step data as well as its subcategories to determine whether there is a significant difference in step data between fallers and non-fallers (Lee et al., 2019). Multiple univariate logistic regression analyses were performed to individually identify variables that could best predict laboratory fall-risk (induced by a slip and trip trial) in older adults. Laboratory fall outcome was treated as a binominal variable with the outcome for participants who fell in either one or both of the two perturbations being denoted as 1 or else with a 0. Hours of phone carriage was also inputted into the logistic regression as a covariate. Based on the univariate logistic regression results, variables with a significance of $\leq$0.1 were included in the multivariate logistic regression analysis (Bursac et al., 2008; Sperandei, 2014). A multivariate logistic regression analysis using a backward stepwise method was performed to generate a model with variables that could best predict laboratory-induced slip or trip induced falls in older adults. A receiver operating characteristic (ROC) curve was used to determine the cut-off scores (a score with the highest sensitivity and specificity) of significant variables in the univariate logistic regression and to determine the area under the curve (AUC) for the overall model predicted by the multivariate logistic regression. Pearson correlations were conducted to examine the relationships between participants' demographics, total 1-week and 1-month smartphone collected steps, fall-risk measurements, and PASE. Point biserial correlation was applied to examine relationships between steps data and the dichotomous fall histories.

RESULTS

Based on the results of the qualitative questionnaire, 12 older adults used Android phones whereas 37 used an iPhone. On average, participants carried their phones 9 h (9.06 ± 5.6 h) per day. Of all the participants, 38 reported they were active most of the time they carried their phones, 6 reported they were active even during the time they did not carry their phones, and 5 were unable to recall or answer the question. Only 11 participants owned a wearable device such as a Fitbit.

Out of the 49 participants, 35 participants fell on at least one perturbation during the laboratory fall test. Thirty-two participants fell only on the slip perturbation, 18 participants fell only on the trip perturbation, and 14 participants fell on both slip and trip perturbations. **Table 1** indicates the means of

TABLE 2 | Variables and their significance (p-value) and R square value based on univariate logistic regression results.

Variable	p-value	R square value
Age	0.011	0.147
Weight	0.592	0.006
Fall History	0.056	0.075
MMSE	0.177	0.036
ABC	0.313	0.022
BBS	0.820	0.001
TUG	0.043	0.100
PASE	0.320	0.020
1-week steps	0.863	0.003
1-month steps	0.198	0.040

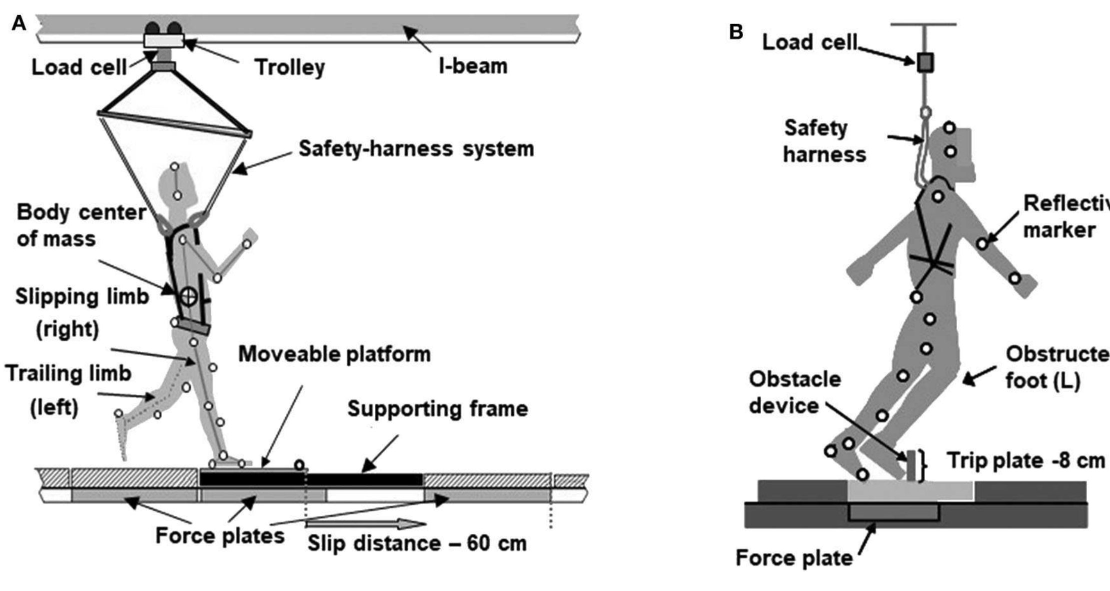

FIGURE 3 | (A) Schematic diagram that indicates the experimental slip set-up demonstrating the participant in a safety harness attached to the load cell and reflective markers attached to the anatomical landmarks. As the right heel of the participant land on the moveable platform, it is unlocked to slide up to a distance of 60 cm. **(B)** The experimental trip set-up demonstrating the trip plate in its upright position as an obstacle to induce a trip.

TABLE 3 | Overall model predicted based on multivariate logistic regression results along with the sensitivity, specificity, overall accuracy, and the area under the curve (AUC) found using the Receiver Operating Curve (ROC).

Model	Variable	Significance	Sensitivity	Specificity	Overall accuracy	Overall significance	AUC
1	Age	0.006	97.1	41.7	83.0	0.002	0.807
	Fall history	0.0065					
2	Age	0.012	94.3	58.3	85.1	0.002	0.831
	TUG	0.169					
	Fall history	0.122					

demographic data and outcome measures of fallers and non-fallers groups. Based on the results of the paired t-test there was no significant difference in overall 1-week and 1-month step data or its sub-categories between fallers and non-fallers ($p > 0.05$). Based on the univariate regression analyses, age, and TUG were the only significant variables with the significance value set at $p = 0.05$ (**Table 2**) and fall history having a near significant value of $p = 0.056$. Based on the results of the ROC curve, we established that the variable age with a cut-off score of 69.5 had a sensitivity of 63.9% and specificity of 61.5%, indicating that older adults above 69.5 years had a greater fall-risk. Similarly, for the variable TUG, the cut off score of 7.49 had a sensitivity of 80% and specificity of 58.3%, indicating that older adults who took longer than 7.49 s to complete the TUG test were at greater fall-risk.

Furthermore, variables with a significance of $\leq$0.1 for the univariate logistic regression analysis were included in the multivariate logistic regression analysis using the backward stepwise method (**Table 2**). The multivariate regression analysis revealed an overall model (Model 1) including age and fall history with an overall accuracy of 83% to predict laboratory-induced falls with a sensitivity of 97.1% and specificity of 41.7% ($p = 0.002$). The area under the curve (AUC) for the model was 0.807. A model before the final model included TUG in addition to variables age and fall history (Model 2). Addition of TUG in the final model (Model 2) improved the overall accuracy to 85.1% with a sensitivity of 94.3% and a better specificity of 58.3% ($p = 0.002$). The AUC of this model increased to 0.831 (**Table 3**). **Figure 4** indicates the AUC for both the models.

Thus, participants who were older, had a higher TUG score (took longer to complete the test), and had a history of falls were more likely to fall during the laboratory-induced perturbations. The best logistic regression model which predicted immediate laboratory fall is represented as Predicted Logic of (Laboratory fall) = 18.175 + (0.207)*Age + (0.525)*TUG + (1.324)*History of fall. Based on this, the odds of falling for a person who had a fall history in the previous year would be 3.75 times more than those with no falls. Similarly, for each 1 s increase in TUG score (slower), the odds of experiencing perturbation-induced falls would be 1.69 times higher. Further, for every 1-year increase in age, the odds of falling on laboratory perturbation would be 1.22 times higher.

Pearson correlations revealed a significant negative correlation between age and hours of phone carriage ($r = -0.300$, $p = 0.046$), indicating that the older participants carried their phones for less time than the younger ones. One-week steps data did not correlate with any clinical fall-risk measure and history of fall. However, 1-month steps data positively correlated with higher BBS ($r = 0.386$, $p = 0.006$) and ABC ($r = 0.369$, $p = 0.012$) scores and negatively correlated with previous 1-year fall history ($r_{pb} = -0.293$, $p = 0.041$). Additionally, hours of phone carriage positively correlated with steps data ($r = 0.327$, $p = 0.028$) and, hence, was inputted as a covariate in logistic regression. There was no significant correlation between steps data and PASE score.

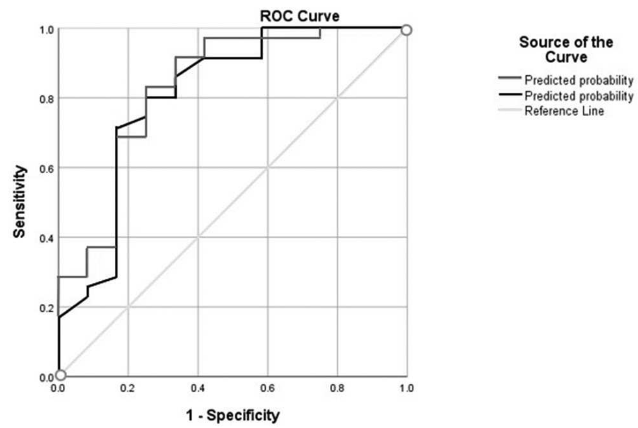

FIGURE 4 | This figure demonstrates the area under the curve (AUC) for model 1 (area under the black line) comprising of variables age and fall history and model 2 (area under the gray line) comprising of variable age, fall history, and Timed Up and Go (TUG) found using the Receiver Operating Curve (ROC).

DISCUSSION

This study explored the relationship between smartphone steps data and other commonly used clinical fall-risk measures and determined their ability to predict laboratory-induced slip or trip falls in healthy community-dwelling older adults. Univariate logistic regression analyses predicted age and TUG as individual significant predictors of laboratory falls with fall history having a near significance value. Multivariate logistic regression determined a model, which included age and fall history that best predicted laboratory-induced slip or trip fall-risk. Addition of TUG to the final model improved the overall prediction capacity of the model. Neither 1-week nor 1-month steps data could predict laboratory-induced fall-risk as a single predictor or in the multivariate model. A weak but significant positive correlation was noted between 1-month smartphone steps and BBS and ABC scores and a negative correlation was seen with previous 1-year fall history. No correlation was found between smartphone steps and PASE scores or other fall-risk measures.

Based on the multivariate logistic regression results, the final model predicted included age and fall history (Model 1) with a sensitivity of 97.1%, specificity of 47.1%, overall accuracy of 83%, and AUC of 0.807, indicating that older adults with a previous history of fall were predisposed to a greater fall-risk during laboratory-induced perturbations. Our results are consistent with the previous literature indicating that older adults with a fall history have difficulty maintaining postural control and thus are at a higher fall-risk (Horak et al., 1989; Ambrose et al., 2013). Similarly, the relationship between aging and falls is consistent with previous literature which indicated that fall rates increase with aging (Ageing Life Course family Community Health World Health Organization, 2008). For example, Pai et al. (2010) demonstrated reduced stability control in older adults compared to young adults, thereby predisposing them to a greater risk for falls during laboratory-induced slip perturbations. Furthermore, aging also has an effect on the recovery stepping response which is critical for establishing a new functional base of support following a perturbation thereby further increasing fall-risk (Tseng et al., 2009).

A model before the final model included TUG in addition to variables age and fall history (Model 2) with similar sensitivity of 94.3%, a better specificity of 58.3%, overall accuracy of 85.1%, and a higher AUC of 0.831. Additionally, TUG as a single factor had significant prediction for laboratory-induced falls, indicating that participants who took longer to finish the test had greater risk for laboratory-induced falls. Our results were similar to a previous study, wherein TUG was able to independently predict 60% of slip-induced falls (Bhatt et al., 2011), and these falls resulted in the center of mass (COM) moving behind the forwardly sliding base of support (BOS). Thus, a lower TUG score could indicate a superior ability of the participant to rapidly relocate the COM over the displaced BOS resulting in improved COM state stability against slip-induced balance loss (Pai and Iqbal, 1999). Conversely, participants would experience forward instability following a trip perturbation due to the forward shift of both COM velocity and displacement with respect to the BOS. Therefore, a faster walking speed could increase forward instability, resulting in a greater risk for trip-induced falls (Pavol et al., 2001; Wang et al., 2019b). The TUG test scores could thus have had opposing predictive effects for slip vs. trip perturbation. In spite of the possibility of such an opposing relationship between TUG scores and fall-type, TUG emerged as a significant fall-risk predictor probably due to the fact that there were greater slip falls ($n = 32$) than trip falls ($n = 18$). However, in spite of being a significant predictor in the univariate regression analyses, it was not included in the final multivariate logistic regression model comprising of age and fall history probably due to the opposite effect of walking speed on the recovery of slip and trip explained above (Model 1). Even though TUG was included in the second model, surprisingly the individual significance of the variables fall history and TUG was lost. This might possibly be because in Multivariate logistic regression analysis, the two variables are basically "competing" with each other for explaining laboratory falls.

Although smartphones are prevalent and can be readily used in the community, steps data collected were unable to accurately predict laboratory falls. Additionally, our results indicated no difference in total 1-week, 1 month, or sub-category step data between fallers and non-fallers. Thus, indicating that either there was no association between daily steps, a single aspect of PA, and fall-risk (given the steps collected by smartphones were accurate) or smartphone is an insufficient tool in collecting steps data under free-living condition. Our study survey indicated that only 22.44% (11/49) of participants in the current study had access to an extra wearable PA tracker (Fitbit). Thus, older adults majorly relied on smartphones for PA monitoring. However, a few study participants mentioned in the survey that they only carried their phones when going outside. Thus, we might have missed out on data when older adults were physically active but not carrying their phones, especially when they were walking inside their homes while performing daily routine activities. Moreover, participants were not given any instruction by the experimenter on the way they should carry their smartphone. Thus, the participants could have carried their smartphones in many different manners, and the phones could have recorded steps differently based on their position and orientation (Carter et al., 2018; Funk and Karabulut, 2018). A recent study also suggested that smartphone updates and different application versions might potentially change the outcomes of smartphone-based assessments (Brodie et al., 2018). It is also unclear whether the type of smartphone application matters, and if there was a difference between the "Health" and "Google Fit" applications since both were used in the study.

PA assessed using PASE was not a significant predictor of fall-risk. The PASE questionnaire might not accurately predict fall-risk in older adults as it only monitors PA over a span of 7 days, which might not be enough to provide an overall view of older adults' PA. As mentioned earlier, recall bias involved in such self-report technique might also limit the accuracy of PA measurement. Additionally, studies have demonstrated reporting bias for subjective questionnaires like PASE, such as individuals overestimating the time spent on strenuous activities or underestimating the time spent on activities that require less exertion (Bolszak et al., 2014). Also, such subjective questionnaires might lead to participants giving socially desirable answers. As PA is encouraged in older adults, participants might overestimate their PA to attain social approval. Such behavior would provide inaccurate data, thus further limiting PASE's sensitivity for fall prediction. Furthermore, PASE involves scoring an individual based on their frequency scores obtained for moderate to strenuous activities which may not be commonly performed by older adults. Thus, PA assessments should comprise of activities performed frequently by older adults to provide better fall-risk prediction. Measurement of these activities might also explain why PASE scores did not correlate with smartphone steps, as PASE considered the overall PA of older adults, including everyday household, recreational, and occupational activities, whereas smartphones only considered steps data.

BBS was not a significant predictor of falls in healthy older adults, which is consistent with the previous literature (Mancini and Horak, 2010; Bhatt et al., 2011). Previous studies suggest that individuals with a BBS cut-off score of 45 and above are high-functioning and at a lower risk of falls (Berg et al., 1992a, 1997). However, despite the average BBS score in our study sample being 53.57, a score much higher than the threshold suggesting a low risk of fall (Berg et al., 1992b), 36 participants out of 49 still fell during at least one laboratory-induced perturbation and 22 had previous fall histories, indicating the limited sensitivity of BBS. One probable reason might be that BBS assesses volitional balance control and does not account for or measure impairments in reactive balance. Secondly, it has shown to have a ceiling effect in healthy older adult population as it rates performance mostly during standing tasks (Newton, 1997; Langley and Mackintosh, 2007). Such tasks might not be challenging enough to assess dynamic balance control during daily living functional tasks in our population of community-dwelling, healthy older adults. Thus, the tasks performed and tested in BBS lack task-specificity to assess fall-resisting skills, thereby indicating its limited sensitivity for predicting fall-risk upon exposure to real-life like large external perturbations.

Psychosocial factors assessed in terms of balance confidence and fear of falling are crucial for fall prediction. However, balance confidence measured using ABC was not a significant fall-risk predictor. Previous studies done using ABC showed inconsistent results for ABC's ability to predict fall-risk (Lajoie and Gallagher, 2004; Schepens et al., 2010). While few studies demonstrated that ABC scores could differentiate fallers from non-fallers, with fallers having a lower ABC score (Mak and Pang, 2009; Hadjistavropoulos et al., 2011), other studies did not demonstrate a link between ABC scores and falls. The study results indicate the mean ABC score for our study population was 85.79%. However, despite having a high balance confidence score, suggesting a low risk of fall, over half of participants fell upon experiencing a slip or trip perturbation, indicating that ABC might demonstrate a ceiling effect as individuals may overestimate their balance abilities. Thus, there could be a potential mismatch between individuals' self-perception of their own balance abilities and their actual functional mobility, balance, and gait impairment, thus limiting ABC's sensitivity for fall prediction.

Our results indicated no correlation between 1-week steps data and commonly used clinical measurements for risk factors of falls, however a positive correlation was noted between 1-month steps data and BBS and ABC scores and a negative correlation with previous 1-year fall history. The correlation of step data with BBS could be expected considering the close association between mobility and stability and the BBS known to be a gold standard for assessing balance control in the older adults (Berg et al., 1992a; Santos et al., 2011). Previous studies have reported that enhancing one's mobility via isolated walking programs improves static and dynamic balance as well as overall postural stability (Brooke-Wavell et al., 1998; Paillard et al., 2004). Although there was a moderate positive correlation between these two variables, both variables were not selected as fall-risk predictors as discussed above.

The correlation between the ABC scale could also be expected and justified. It is known that any improvements in balance and stability may aid in reducing both fall-risk and the subsequent fear of falling in older adults (Gusi et al., 2012). Previous studies suggested that increased walking is associated with good balance perception (Yang and Hsu, 2010). As the ABC scale determines a person's own perception of balance activities, those who were more ambulatory and had more steps could have had an enhanced balance perception of themselves and vice versa. This might explain the paralleled finding of the positive relationship between steps and balance confidence. However, with the correlation between steps data and BBS and ABC being very modest ($r < 0.3$), the potential of smartphones to be used as a screening tool to identify older adults with reduced balance and balance confidence needs further investigation.

Additionally, the results of the sub-analysis found that long-term step monitoring yielded better association with commonly used clinical measures. Several studies have utilized and suggested that short-term monitoring for 1 week is adequate for PA monitoring and fall-risk (Tudor-Locke et al., 2005; Huberty et al., 2015). For example, a study done using wearable sensors in middle-aged and older women indicated that 24 h monitoring over a span of 1 week is a feasible approach for monitoring activity behavior (Huberty et al., 2015). However, there are several studies indicating that collecting long-term baseline data might be more accurate in yielding stochastic predictions (Mathie et al., 2004; Yang and Hsu, 2010). For example, a review article on PA monitoring suggested that long term monitoring could enable better understanding of PA behavior (Taraldsen et al., 2012). It is postulated that long-term data collection enables monitoring of day to day variability thereby providing a better understanding of consistent and habitual steps data of older adults and could thus show a better correlation with clinical fall-risk measures. While 1-week monitoring might be more feasible and could increase compliance, our results similar to few other studies suggest that 1-month monitoring might yield better results.

STUDY LIMITATIONS

This study has several limitations. In the current study, the results could have been affected based on the hours and ways of phone carriage by participants as no strict instruction was given regarding phone usage. Further the software applications inbuilt or installed on the phone varied (e.g., iPhone vs. Android) which could have affected the step data accuracy. However, these factors could not be controlled due to the design of the study aiming at maximally collecting data in a natural manner and environment. Future studies could conduct studies with a uniform type of hardware (smartphone) and software (application) and additionally use wearable motion sensors to validate the association between daily steps data and older adults' fall-risk in response to laboratory-induced, real-life like external environmental perturbations. Lastly, study participants were among the healthiest community-dwelling older adults with a good physical performance and scores on clinical measurements. Hence, it is unknown whether current findings would also apply to frail older adults who are more susceptible to falls.

CONCLUSION

The study revealed no association between smartphone steps data and laboratory fall-risk in a group of community-dwelling older adults with good physical performance. However, being the first of its kind, the current results could be leveraged to design further studies intending to use smartphone step data for fall-risk prediction. Further, the study reinforced previous findings that, older participants with fall histories and higher TUG scores were more likely to fall in the laboratory.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Institutional Review Board at the University of Illinois at Chicago. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

YW, RG, and TB made substantial contributions toward conception, design, and execution of the study. YW, RG, LK, and EW organized the data and performed statistical analysis. YW, RG, TB, EW, LK, and AS contributed in drafting the manuscript and revising it to create a final version for submission. All authors contributed to the article and approved the submitted version.

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Interactions Between Different Age-Related Factors Affecting Balance Control in Walking

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Maintaining balance during walking is a continuous sensorimotor control problem. Throughout the movement, the central nervous system has to collect sensory data about the current state of the body in space, use this information to detect possible threats to balance and adapt the movement pattern to ensure stability. Failure of this sensorimotor loop can lead to dire consequences in the form of falls, injury and death. Such failures tend to become more prevalent as people get older. While research has established a number of factors associated with higher risk of falls, we know relatively little about age-related changes of the underlying sensorimotor control loop and how such changes are related to empirically established risk factors. This paper approaches the problem of age-related fall risk from a neural control perspective. We begin by summarizing recent empirical findings about the neural control laws mapping sensory input to motor output for balance control during walking. These findings were established in young, neurotypical study populations and establish a baseline of sensorimotor control of balance. We then review correlates for deteriorating balance control in older adults, of muscle weakness, slow walking, cognitive decline, and increased visual dependency. While empirical associations between these factors and fall risk have been established reasonably well, we know relatively little about the underlying causal relationships. Establishing such causal relationships is hard, because the different factors all co-vary with age and are difficult to isolate empirically. One option to analyze the role of an individual factor for balance control is to use computational models of walking comprising all levels of the sensorimotor control loop. We introduce one such model that generates walking movement patterns from a short list of spinal reflex modules with limited supraspinal modulation for balance. We show how this model can be used to simulate empirical studies, and how comparison between the model and empirical results can indicate gaps in our current understanding of balance control. We also show how different aspects of aging can be added to this model to study their effect on balance control in isolation.

Keywords: aging, balance, modeling, neuromechanic, vision, muscle strength, cognition, walking

1. INTRODUCTION

Walking on two legs is inherently unstable and requires continuous control to keep the body upright. Learning to do so is a major developmental milestone for infants. At the opposite end of the age spectrum, it is well-known that older adults are at increased risk of falling, with a high probability of falls resulting in injury (Herdman, 1997; Kannus et al., 1999). In the US alone, 3.2 million falls occur each year leading to medical treatment, with health care costs exceeding $30 billion. The risk of falling increases with age (Rubenstein, 2006), and the injuries resulting from falls limit mobility and impair the ability to perform daily tasks, leading to a decline in quality of life (Fuller, 2000; Stevens et al., 2006).

We know that the tendency to fall more often during walking is associated with a number of factors (Osoba et al., 2019), such as weaker muscles (Pijnappels et al., 2008a), slow gait (Jerome et al., 2015), and cognitive decline (Lamoth et al., 2011), and all of these factors also tend to change with age. But we do not understand the causal relationship between these factors, and a decline in upright balance control. One major hurdle is that all of these changes happen simultaneously over a long time. As we get older, our muscles tend to slowly get weaker, we walk slower, and lose mental acuity. But what is the causal relationship, if any, between these factors? Do our weakening muscles cause us to walk slower, which is harder to control and causes increases in fall risk? Or is the slower walking a coping mechanism to account for longer cognitive processing time for executive function?

The sensorimotor control loop for walking integrates dynamic processes ranging over sensation, neural processing, integration with an overall motor plan, transformation into descending motor commands, reflex arcs in the spinal cord, muscle physiology, and force generation and biomechanical interaction with the environment. The processes at all of these levels are dynamically coupled and can potentially interact and affect each other. Furthermore, slow changes in one factor could drive adaptive changes in other processes, like preferring a slower walking speed, to offset the increased risk of falls from longer reaction times, or to account for increased fatigability (Finsterer and Mahjoub, 2014). This integration on a developmental time scale makes it hard to isolate the mechanisms of how each single factor affects balance control and fall risk.

Here we argue that the appropriate tool to solve this problem of isolating and understanding the effect of individual fall risk factors on balance control is to develop a computational model that encompasses the dynamics of the neuromechanical processes at each level. In the following we will frame balance control as a sensorimotor control problem that is solved by the central nervous system, and summarize recently published experimental results from studies with sensory perturbations that measure how this neural control system works (section 2). We will then discuss correlates for deteriorating balance control in older adults, with a focus on muscle weakness, slow walking, cognitive decline, and increased dependency on visual information and review evidence for how they affect fall risk from the general literature, combined with a review of recent empirical findings on how some of these factors affect neural feedback control mechanisms (section 3). We will make the case that the appropriate method to understand the interaction between these different factors and balance control is to use predictive, computational models of balance control, and introduce an existing neuromechanical model of human walking that can be physiologically "aged" to serve as a basis for understanding age effects. We close by performing a simulation study to demonstrate how such predictive models can be used to understand interactions between age-related factors and balance control and to test the functional validity of hypotheses that are hard to evaluate experimentally (section 4).

2. SENSORIMOTOR CONTROL OF BALANCE IN WALKING

A walking human is a mass moving through space, accelerated by forces from muscles, gravity and interaction with the ground. The general walking pattern of moving forward and setting one leg in front of the other emerges passively to some degree, from the mechanical structure of the body. This has been demonstrated by passive walkers, legged mechanical devices that spontaneously generate stable walking patterns, requiring only mechanical energy to off-set losses from friction, usually from walking down an inclined plane (Collins, 2005). While the human body shares some general characteristics with passive walkers, mechanical analysis has shown that it is not mechanically stable. A passively walking human body will generally fall over sideways after a few steps (Kuo, 1999). Stable walking requires a regulating process that actively maintains upright balance. Here we focus on the frontal plane, where balance control is more demanding than in the sagittal plane (Bauby and Kuo, 2000).

Active control of upright balance during walking requires a sensorimotor control loop that collects sensory information about the movement of the body through space, detects deviations from the upright posture, and generates appropriate muscle forces to correct these deviations. To detect deviations from the upright posture, the nervous system uses mainly the proprioceptive, vestibular and visual systems (Shumway-Cook and Horak, 1986). Research in standing balance control has shown that information from these different sensory modes is combined, or "fused," into an estimate of the mechanical state of the whole body in space. If this estimate detects a deviation from the upright, the control loop sends descending motor commands that change muscle activation to generate a corrective force. In quiet standing, this force is usually generated mainly by the ankle musculature that pull on the body as a single, rigid rod, but in situations with substantial sway, the hip joint gets involved as well (Horak and Nashner, 1986). The body behaves, essentially, as an inverted pendulum with one or two links that is fixed to the ground and rotates around the ankle and hip joints.

The biomechanical effect of a muscle activation during walking is highly dependent on the point in the gait cycle (Reimann et al., 2019). The walking body is mechanically complex, with arms and legs moving largely independent from each other, though highly coordinated (Punt et al., 2015; Thompson et al., 2017). The general function of the sensorimotor loop for balance control is the same as in standing (Peterka, 2002), but generating force to correct a detected deviation from the upright posture is less straightforward in walking. In standing, the *gastrocnemius* muscle will always pull the body backward, but in walking, its effect changes drastically based on the gait cycle. During late double stance when the leg is trailing, the *gastrocnemius* will increase the push-off force and move the body forward (Klemetti et al., 2014; Hsiao et al., 2015). During early double stance, when the leg is leading, the *gastrocnemius* will extend the ankle and knee, pushing backward against the body (Hof and Duysens, 2018). During swing, the *gastrocnemius* only moves the foot in the air and does nothing for the whole body. Since the result of a *gastrocnemius* activation depends so strongly on the point in the gait cycle, the appropriate motor response to a detected deviation from the upright posture has to be equally phase-dependent.

What motor responses do humans use when they detect a deviation from the upright, depending on the point in the gait cycle? To answer this question experimentally, we have previously developed a paradigm that perturbs a sensory system to induce artificial fall sensations in walking humans and observes the motor response using kinematics, kinetics, and electromyography (EMG). In this paper, we will summarize results from studies that used this paradigm to investigate the effect of different age-related factors on balance control. All data discussed here has been published elsewhere before. Our platform consists of an instrumented treadmill (Bertec Inc, Columbus, OH, USA) surrounded by a virtual reality (VR) environment projected onto a domed screen (see **Box 1** for details). Artificial fall stimuli are triggered on heel-strike and induce the sensation of falling sideways, rotating in the frontal plane around the stance foot ankle joint during single stance. This rotation around the ankle implies a lateral translation of the whole-body center of mass (CoM). **Figure 1** illustrates the motor responses to these artificial fall stimuli. The overall response is that people move their body in the opposite direction of the perceived fall (**Figure 1A**). After a sensory stimulus induces the sensation of a fall to the right (purple arrow), participants move their CoM (orange line) to the left over the course of the following steps, compared to how the CoM usually moves without a sensory stimulus (gray line). This is the expected response to a fall stimulus. The neural control system detects a deviation from the upright in the form of the artificial fall stimulus to the right and reacts by moving the body back to what it estimates to be upright, i.e., leftward. Since the detected deviation was not real, but artificial, the result is a leftward shift of the whole body CoM in space. By artificially inducing the sensation of a lateral fall in a controlled, repeatable manner, we can observe how the neural controller generates this whole-body leftward movement. The lower panels of **Figure 1** show three different biomechanical mechanisms to modulate the lateral ground reaction force and generate a lateral force against the ground to the right that accelerates the body to the left. In the following paragraphs we will explain each mechanism according to the example of a fall stimulus to the right triggered by a right heel-strike, as illustrated in **Figure 1**, but note that all three mechanisms are used regardless of direction and triggering foot.

2.1. Ankle Roll

Ankle roll is an active ankle inversion torque at the stance leg in single stance (**Figure 1B**). This torque inverts the ankle by pulling the foot segment and the rest of the body together. The foot segment rolls on the ground, shifting the CoP to the right. The upper body is accelerated in space to the left. This mechanism is analogous to the ankle strategy in standing balance control (Horak and Nashner, 1986). This mechanism was first noted in walking by Hof et al. (2007) and later confirmed as an active mechanism (Hof et al., 2010; Reimann et al., 2017, 2018b; Hof and Duysens, 2018). The roll torque is generated by an activation increase in the medial ankle muscles (*tibialis anterior, gastrocnemius medialis*) and a decrease in the lateral muscles (*peroneus longus*) (Reimann et al., 2018b; Fettrow et al., 2019). We quantify this mechanism by integrating the difference in the subtalar joint angle between the perturbed and unperturbed steps over single stance.

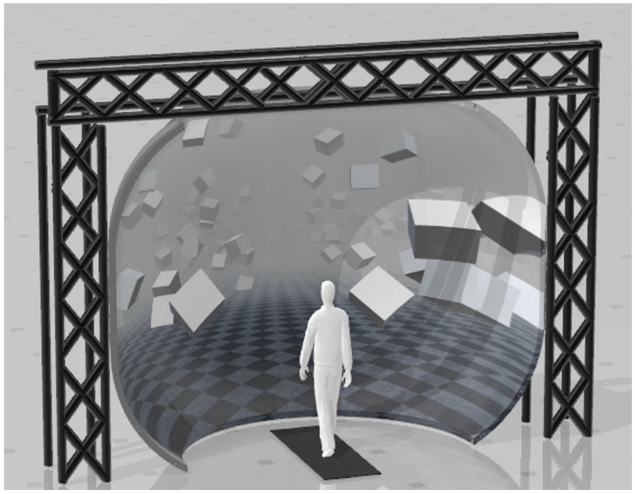
2.2. Foot Placement

Foot placement is an active shift of the lateral foot placement location at heel-strike (**Figure 1C**). This shift in foot position changes the lever arm of the gravitational force acting on the body through the new stance leg during the following step. When detecting a fall to the right, the foot placement is shifted to the right, so gravity pulls the body mass more to the left. This mechanism was first introduced in robotics by Townsend (1985), who showed that foot placement modulation is already sufficient to control upright balance in a walking humanoid. It was discussed later both in robotics (Kuo, 1999; Pratt et al., 2006) and human motor control (Bauby and Kuo, 2000; Hof, 2008) and is now widely accepted to be one of the dominant mechanisms for human balance control during walking (Wang and Srinivasan, 2014; Bruijn and van Dieën, 2018; Reimann et al., 2018a). The lateral shift of the left foot before heel-strike can be generated by a left hip abduction, but also by a combination of internal rotation of the stance leg knee and external rotation of the swing leg hip joint, and we have found evidence for both (Reimann et al., 2018b; Fettrow et al., 2019). We quantify the foot placement shift by calculating the difference between the perturbed foot placement and the predicted foot placement based on the CoM position and velocity at mid-swing using a linear model fitted to the unperturbed steps (for details see Wang and Srinivasan, 2014; Bruijn and van Dieën, 2018; Reimann et al., 2018b).

2.3. Push-Off Modulation

Push-off modulation is a change in the plantar-dorsiflexion angle of the trailing leg during double stance, starting in late single stance (**Figure 1D**). After the right heel-strike triggers a fall stimulus to the right, the right ankle plantarflexion increases, pushing more strongly in the subsequent double stance. This increased push-off shifts the body weight between the two stance legs, in a direction that is largely forward, but also to the left. Push-off is a well-known mechanism for balance in the sagittal plane, used mainly for trip recovery (Pijnappels et al., 2005, 2008b). For medial-lateral

BOX 1 | Virtual walking environment and artificial fall stimuli.



Virtual Walking Environment. Participants walk on a treadmill in a virtual reality (VR) environment projected onto a domed screen. The screen covers almost the complete field of vision of the participant walking on the treadmill. The participant's point of view in the virtual environment is linked to the head position in real time, measured by the motion capture system, creating a motion parallax effect. The speed of the treadmill is linked to the participant's pelvis position, using a non-linear PD-controller to keep the subject centered on the treadmill along the sagittal axis. This user-driven mode allows the participants to walk at a self-selected speed and spontaneously speed up or slow down at any time. The forward progression in the VR environment is also linked to the treadmill speed in real time. In combination, these components create an immersive VR experience, where participants walk through a virtual environment without the secondary task of matching their speed to the treadmill, and where the visual information available to the participants is determined almost exclusively by the virtual environment.

Artificial Fall Stimuli. To induce the sensation of a fall to the side, we stimulate either the visual or the vestibular system of the participants walking in the virtual environment. A **visual** fall stimulus consists of the virtual world rotating around the sagittal axis through the center of the treadmill. This rotation generates optical flow on the participant's retina that is similar to the optical flow of falling sideways by rotating around the stance foot ankle joint. The velocity of this rotation increases at a constant rate of 45–90° s^{-2}, depending on experimental paradigm, for 600 ms, resulting in a horizon tilt of ≈7–15°. The scene remains fixed at that tilt for 2 s, then resets at a constant rate over 1 s to prepare for the next stimulus. A **vestibular** fall stimulus is induced using Galvanic vestibular stimulation (Fitzpatrick and Day, 2004). A light electric current is delivered between two electrodes attached to the mastoid processes behind the ears. The current affects the vestibular nerve and induces the feeling of swaying to the side, with the direction depending on the polarity of the signal. We use a square wave signal with 0.5–1 mA amplitude and 600–1,000 ms duration, depending on the experimental paradigm. Fall stimuli are generally triggered on heel-strike and are followed by a wash-out period of variable length.

balance, this mechanism was first discussed by Kim and Collins (2013), who later showed that an ankle prosthesis using this control principle can reduce the metabolic cost of walking (Kim and Collins, 2015). We observed this mechanism in healthy young humans (Reimann et al., 2018b). The increased plantarflexion in the trailing leg is preceded by increased activity in the *gastrocnemius medialis* in late single stance (Fettrow et al., 2019). We quantify this mechanism by integrating the difference in the ankle plantarflexion angle between the perturbed and unperturbed steps over double stance.

Ankle roll, foot placement shift, and push-off modulation are three biomechanical mechanisms to change the ground reaction force and push the body to the left in response detecting a fall to the right. They become available at different times during the gait cycle and temporally coordinated by the neural control system, which shifts the response between mechanisms as they become available to generate a functional, whole-body balance response to a detected fall (Reimann et al., 2019). In section 3.5, we will review how these balance mechanisms interact with age-related factors affecting fall risk.

3. EFFECTS OF AGING ON BALANCE CONTROL

People tend to fall more often as they get older, and the probability that a fall results in injury is increased with age (Herdman, 1997; Kannus et al., 1999). In the US alone, 3.2 million falls occur each year leading to medical treatment, with health care costs exceeding $30 billion. The risk of falling increases with age (Rubenstein, 2006), and the injuries resulting from falls limit mobility and impair the ability to perform daily tasks, leading to a decline in quality of life (Fuller, 2000; Stevens et al., 2006). Many studies have identified risk factors that predict falls in older adults to some degree (Osoba et al., 2019). In this section, we review how cognitive function, muscle weakness, walking speed and increased dependency on visual information are associated with fall risk, and how these factors are related to the sensorimotor control of balance.

3.1. Cognitive Function

Much of the age-related cognitive decline literature has been focused on cognitive (Reuter-Lorenz et al., 2010) or general

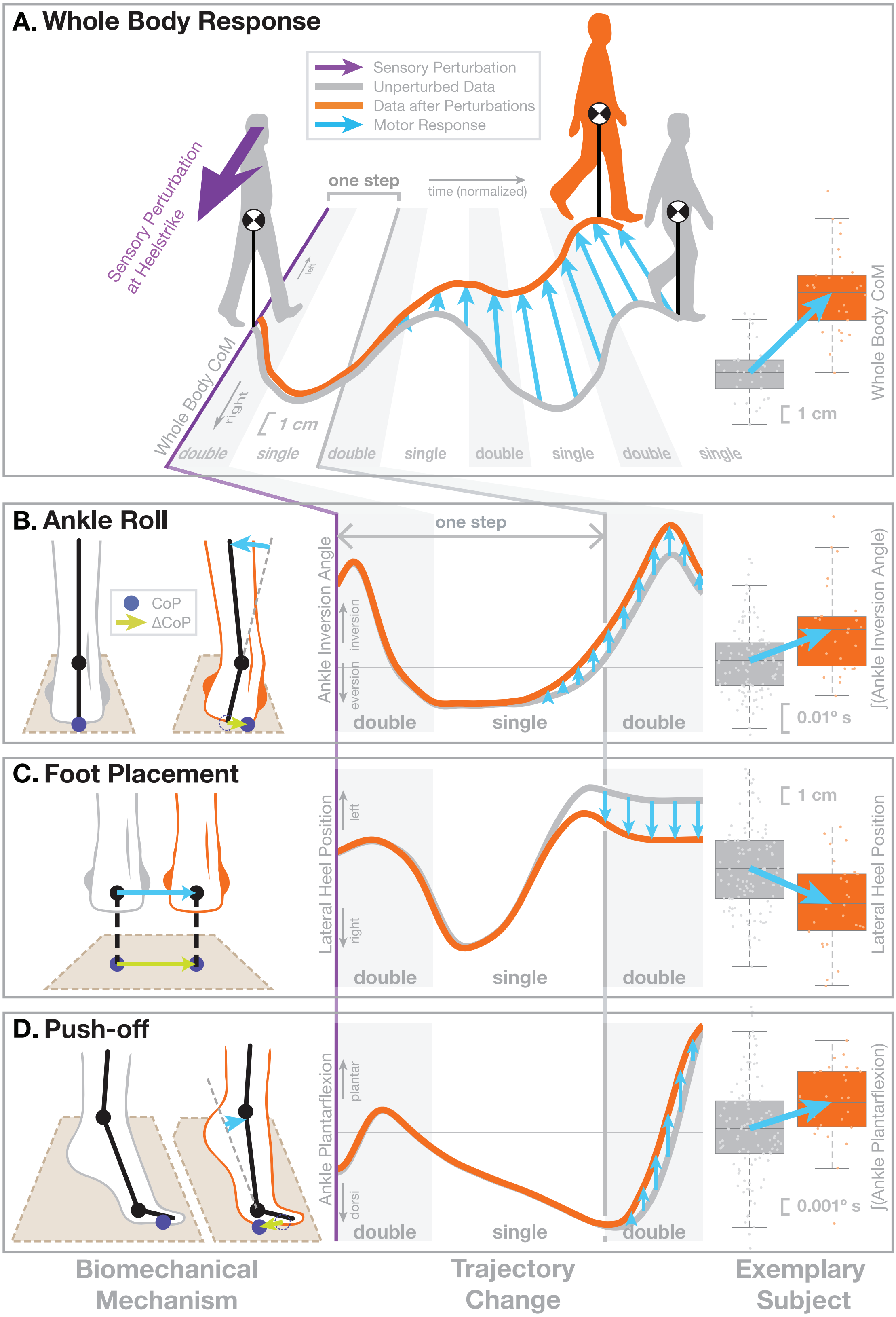


FIGURE 1 | Continued.

FIGURE 1 | Illustration of the motor response to sensory fall stimuli. In this figure we demonstrate the biomechanical action of the different balance mechanism. We selected a non-representative sub-set of previously published data (Reimann et al., 2018b) to clearly illustrate the biomechanics. The subject receives a visual stimulus that induces the sensation of falling to the *right* (purple arrow). **(A)** This results in an overall shift of the whole-body CoM toward the *left*, i.e., the direction opposite to the perceived fall, over the next four steps. The central panel shows the average horizontal CoM trajectory of an exemplary subject for normal, unperturbed steps (gray line). In response to the visual stimulus representing a fall to the right (purple), the average CoM shifts toward the left over the following four steps (orange line). The horizontal axis represents normalized time, with gray areas indicating double stance, white areas single-stance period. Blue arrows show the difference between the two trajectories representing the motor response to the stimulus. The panel on the right shows the total CoM shift after four steps for the same subject, where boxes show the means and 25th–75th quartiles, and whiskers show the data range and dots show the individual data points. The lower part shows the three balance mechanisms that modify the ground reaction force to generate this change in whole-body CoM movement, **(B)** Ankle Roll, **(C)** Foot Placement, and **(D)** Push-Off. In each row, the **left column** shows the biomechanical changes represented by these mechanisms, illustrating how each mechanism affects the forces against the ground. The dark blue dots indicate the CoP position and the green arrows show how the CoP changes as a result of the motor response. Each mechanism results in a CoP shift to the *right*, and this force difference leads to an acceleration of the whole-body CoM to the *left* that results in the CoM trajectory shown in part **(A)**. The **middle column** illustrates changes in how a relevant kinematic variable for each mechanism. The graphs start on the heel-strike triggering the perturbation and the horizontal axis shows normalized time, ranging over the first two double stance periods (gray) and first single stance period (white), which is the period of initial response to the artificial stimulus. The gray line show averages over the normal, unperturbed trajectories for the same subject. The orange lines represent the kinematic changes observed for each mechanism, but since these changes are generally small, we manually modified the curves here to exaggerate them, for the purpose of clearly demonstrating the function. The **right column** shows the actual, unmodified data for the same exemplary subject, where the response is represented by a single aggregate variable. For Ankle Roll, the aggregate variable is the integral of the stance leg ankle inversion angle over the first post-stimulus single stance period (white in the left panel). For the Foot Placement, the aggregate variable is the lateral heel position of the leading leg at the first post-stimulus heel-strike, relative to the trailing leg heel position. For the Push-off, the aggregate variable is the integral of the trailing leg ankle plantarflexion angle over the second post-stimulus double stance period (gray in the left panel). The blue arrows in all panels indicate how each variable changes in response to the stimulus.

motor function (Carson, 2018). The disparity of research linking cognition and mobility may be a result of perspective of the researchers in their respective fields, with little overlap between fields. In the field of robotics, bipedal locomotion can be reproduced with passive dynamics (Kuo, 1999), calling into question the need for neural control, let alone the role of supraspinal circuits. Observation of patients in the medical field yield a different perspective, where locomotion typically deteriorates in the event of a stroke (Dean and Kautz, 2016) or Parkinson's Disease (Curtze et al., 2016) leading to the conclusion that supraspinal control is critical to the task of balance and locomotion. Postural control requires active modulation of muscle activity even in standing (Morasso and Sanguineti, 2002), and there is a wide range of evidence that cognitive processing is required.

Dual-task paradigms have been the main experimental methodology linking cognition to the control of balance and locomotion (Woollacott, 2000; Li and Lindenberger, 2002; Horak, 2006). Cognitive tasks performed during standing or walking generally interact with balance control, often leading to impaired performance in balance (Barra et al., 2006) or the cognitive task (Andersson et al., 2002). Since the 1980s, dual-task paradigms have been used to assess the role of cognition during standing (Cordo and Nashner, 1982; Stelmach et al., 1990). The results of these early studies of dual-task postural control reveal that, in general, postural stability is prioritized over the secondary task. This interaction between balance and secondary cognitive tasks is consistently stronger in older adults compared to younger (Teasdale and Simoneau, 2001; Redfern et al., 2002; Melzer and Oddsson, 2004; Lamoth et al., 2011; Schaefer et al., 2014; Li et al., 2018), as well as in populations with neuromotor impairments (Camicioli et al., 1997; Lapointe et al., 2010; Bahureksa et al., 2016). The degree to which performance in the secondary task diminishes (dual-task cost), is dependent on the perceived risk of injury (Shumway-Cook et al., 1997), indicating a complex relationship between task prioritization, perception of self, and the environment (Wrightson and Smeeton, 2017). Moreover, the specific kind of secondary task influences attentional demands, with visual or arithmetic tasks having a stronger effect than verbal or auditory tasks (Beauchet et al., 2005a,b).

Interventions targeting cognition are also insightful for understanding the role of cognition in the control of balance and locomotion. Balance and strength training while performing simultaneous cognitive tasks can improve performance in dual-task protocols (Hiyamizu et al., 2012). Performing cognitive training in isolation shows transfer to balance related outcomes (Li et al., 2010). Reduced prefrontal brain activity in a walking task after a dance intervention in healthy older adults (Eggenberger et al., 2016) and a treadmill dual-task training intervention for people with Parkinson's (Maidan et al., 2018) indicates less attentional resources are dedicated to the task after the intervention. These results can help confirm a link between cognition and control of balance and locomotion.

In general, these results suggest walking becomes less automatic as age increases, shifting from spinal level control to supraspinal control (Clark, 2015). The shift of control is typically observed as prefrontal over-activation during steady state walking in older adults and other populations with hindered mobility, such as stroke (Mihara et al., 2007), Parkinson's disease (Maidan et al., 2016), and multiple sclerosis (Hernandez et al., 2016). Tasks with a cognitive aspect during walking, such as a precision stepping, have been found to increase activity in the prefrontal cortex, as measured by functional near-infrared spectroscopy (fNIRS, Koenraadt et al., 2014). Increased prefrontal activation is also observed in older adults, compared to younger adults, when increasing the difficulty of walking by adding obstacles to the environment (Chen et al., 2017; Mirelman et al., 2017). When encountering an obstacle during walking, the normal, steady state motor plan must be inhibited, requiring the nervous system to plan a new trajectory that avoids the obstacle (Potocanac et al., 2014a). This process is time-sensitive (Potocanac et al., 2014b) and older people generally perform worse (Potocanac et al., 2015). The direct assessment of supraspinal components during the actual task of interest (walking, walking over obstacles, walking on different terrain) provides the most compelling evidence that supraspinal circuits contribute to the task of balance.

3.2. Muscle Weakness

One of the most consistent risk factors for falling is muscle weakness (Rubenstein, 2006; Pijnappels et al., 2008b), especially in the lower limbs (Moreland et al., 2004). Declines in lower-extremity muscle strength become apparent in the 5th or 6th decade of life (Murray et al., 1980; Lindle et al., 1997), with estimated rate of strength loss of 2–3% per year in adults over age 65 (Skelton et al., 1994). Age-related strength losses are most severe at faster muscle contraction velocities (Callahan and Kent-Braun, 2011), potentially limiting older adults' ability to reposition limbs or generate force fast enough to prevent a fall in response to an unexpected perturbation.

Muscle weakness has repeatedly been correlated with both greater incidence of falls and greater incidence of factors thought to be risk factors for falls. Adults over age 65 have been found to have weaker lower-extremity muscles and greater postural instability (Hurley et al., 1998), with greater weakness being associated with poorer stability (Hasson et al., 2014; Menant et al., 2017; Gadelha et al., 2018b). Further, weaker older adults have a higher incidence of falls than their stronger counterparts (Menant et al., 2017; Gadelha et al., 2018a; Yeung et al., 2019). However, the extent to which muscle strength directly counters fall risk is unclear.

While muscle strength begins to decline in the 40s, increased incidence of falls is generally not reported as a major health concern until after age 65 (Peel, 2002). This suggests that there is either a minimum threshold of strength needed to maintain balance or that, beyond a certain point, the parallel decline of multiple physiological systems makes avoiding falls difficult. Decreased muscle strength can certainly contribute to increased fall risk, as acute muscle fatigue induces gait changes indicative of poorer stability or greater fall risk even in young healthy adults (Barbieri et al., 2014). Acute fatigue of healthy older adults leads to changes in gait and posture toward patterns seen in fall-prone older adults (Helbostad et al., 2007; Egerton et al., 2009; Foulis et al., 2017). These increases in markers of fall risk with acute muscle weakness support some direct role of muscle strength in balance.

Interventions designed to improve muscle strength can increase scores on clinical balance tests (Hess and Woollacott, 2005), reduce fear of falling (Gusi et al., 2012) and decrease fall risk (LaStayo et al., 2003), especially as part of a multifactorial approach (Sherrington and Tiedemann, 2015). Exercise interventions as a whole reduce the rate of falls by 23%, with multifaceted interventions (i.e., balance, functional, and resistance exercise) reducing the rate falls by more than 30% (Sherrington et al., 2019). However, resistance training alone may not lead to reductions in falls (Sherrington et al., 2019). Because muscle weakness is likely one of several factors leading to increased fall risk with age, there is still limited evidence of the power of muscle weakness to predict and of strength training to prevent falls (Pizzigalli et al., 2011; Granacher et al., 2013). An understanding of the mechanisms behind changes in strength and the interaction of changes in strength with other physiological systems is needed.

3.3. Walking Speed

Reduced walking speed has also been tied to incidence of falls (Abellan Van Kan et al., 2009; Verghese et al., 2009; Middleton et al., 2016; Geerse et al., 2019) and fear of falling (Callisaya and Verghese, 2018; Geerse et al., 2019; van Schooten et al., 2019) in older adults. Slowed gait speed is a common characteristic of aging (Himann et al., 1988; Nigg and Skleryk, 1988; Bohannon, 1997; Jerome et al., 2015). Further, declines in walking speed are correlated to declines in muscle strength (Bassey et al., 1988; Bendall et al., 1989; Rantanen et al., 1998).

Despite correlations between walking speed and falls risk, we do not understand how walking speed influences stability, balance, or falls. Greater variability and instability are thought to be indicators of greater fall risk in older adults, and many studies have tested the effects of age, walking speed, or age and walking speed on these parameters. Older adults generally have greater variability and instability than young adults (Hausdorff et al., 2001; Kang and Dingwell, 2008a,b; Verghese et al., 2009) and, because older adults typically walk slower than young adults, it has been thought that these measures of fall risk are mechanistically related to walking speed. However, associations between slow gait speed and greater fall risk may be largely a byproduct of the fact that age-related decreases in gait speed and increases in fall risk occur in parallel.

Gait variability and stability change similarly for young and older adults with increases or decreases in speed despite older adults having greater average variability and instability (Kang and Dingwell, 2008a,b). Faster walking speed itself has not been associated with decreased stability in young adults (Bruijn et al., 2009) and some measures of stability may actually increase with faster walking speed in young adults (England and Granata, 2007; Hak et al., 2013). When only older adults are examined, contrasting results show greater variability in those who walk slower (Verghese et al., 2009) as well as in all older adults regardless of speed (Hausdorff et al., 2001; Dingwell et al., 2017; van Kooten et al., 2018). In a study of individuals with and without diabetic neuropathy, slower speed was a predictor of greater stability in adults with diabetic neuropathy (Dingwell et al., 2000), suggesting that slower walking speed may be used to improve stability in the presence of additional physiological deficits.

The presence of increased gait variability or instability both with and without slowed gait speed in older adults may suggest that, as with decreased muscle strength, slow gait is only used to improve stability once a critical threshold of overall physiological decline is reached. Since walking speed is correlated with muscle strength and cognitive function in older adults (Lauretani et al., 2003; Holtzer et al., 2006), it is unclear whether reduced walking speed itself has a negative effect on balance control, or whether there is a common cause underlying both phenomena. Examining the interacting effects of cognitive function, strength, and walking speed on balance may provide a more complete picture of falls risk.

3.4. Increased Visual Dependency

Older adults tend to depend more on visual information compared to younger adults (Osoba et al., 2019). Lord and Webster (1990) first showed that older adults that had experienced a fall recently performed significantly worse on the rod and frame test, indicating increased visual dependence. Jeka et al. (2010) support this result, finding that older adults show increased responses to visual perturbations in standing, with even higher responses in a fall-prone older adult group. Yeh et al. (2014) studied the effect of a secondary cognitive task and time delay in the visual feedback, and found that older adults tend to prioritize visual feedback over proprioception even with a disruptive time delay. More recently, Lee (2017b) studied the relationship between visual dependency, categorized by the rod and disc test, with clinical assessments of balance and vertigo, finding significant differences between visually dependent older adults compared to both young adults and visually independent older adults. A similar study, however, using different clinical tests, found no significant effect of visual dependency, but noted that these tests largely lacked visual components (Lee, 2017a). In contrast, Almajid et al. (2020) do find that performance in the timed up and go test of visually dependent older adults is more affected by a visual perturbation than the performance of visually independent older adults.

In walking, research on the use of visual information has mostly focused on high-level effects like navigation (Warren et al., 2001) and speed control (Lamontagne et al., 2007). Optical flow is also used for balance control during walking (Reimann et al., 2018b), and older adults tend to depend on it more than young adults. Anderson et al. (1998) removed optic flow during walking by occluding a short stretch of the walkway, which led to a significant increase in gait velocity and step length in older adults, but not in young adults. Perturbing optical flow with filtered white noise has a destabilizing effect on both young and older adults, but the effect is generally much stronger in older adults, where it can significantly disrupt step placement (Franz et al., 2015). This difference in the effect of visual perturbations between age groups is substantially more prominent than the effect of a secondary cognitive task or walking with narrow step width (Francis et al., 2015). Qiao et al. (2018a) investigated the effect of perturbed optical flow at joint level and found a general increase in variance that was larger in older compared to young adults. Surprisingly, Qiao et al. (2018b) found a negative relationship between local dynamic instability measures and responses to optical flow perturbations in young adults, and failed to establish any significant relationship in older adults. Kazanski et al. (2020) used optical flow perturbations in a similar paradigm, but failed to find significantly increased visual dependency in the older adult group.

The evidence is relatively clear that visual older adults are more affected by visual perturbations, but there is no clear picture *why* that is the case or *how* these phenomena are linked. One possible explanation is that age-related decreases in muscle strength compromise balance in older adults, and the CNS adapts the gains of the sensorimotor control loop to re-establish robustness. This might explain why Qiao et al. (2018b) found that young people who responded more strongly to visual perturbations tended to have higher local dynamic stability measures. But this pattern did not show up in older adults in the same study, possibly because of the confounding influence of other age-related factors affecting postural stability.

3.5. Interactions Between Age Effects and Sensorimotor Control of Balance

There is solid evidence, reviewed above, that cognitive function, muscle strength, and walking speed are correlated with increased fall risk, and that older adults generally depend more on visual information for balance control compared to young adults. But we know little about the causality behind this relationship, of *how* these factors are related, or *why* older adults rely more on vision. Here we briefly review some findings about how walking speed, specifically cadence, and cognition affect the sensorimotor control of balance.

Older adults tend to walk with increased cadence compared to young adults (Judge et al., 1996), even at a reduced average gait speed (Ko et al., 2010). We have investigated the interaction between stepping cadence and sensorimotor control of balance by assessing the effect of artificial fall stimuli in healthy young adults walking while matching their cadence to a metronome at 80 (*low* cadence) or 110 (*high* cadence) beats per minute. Here we summarize the relevant findings from this study. For details, please refer to Fettrow et al. (2019). **Figure 2** shows the resulting responses in the whole-body sway and the three balance mechanisms ankle roll, foot placement and push-off. The overall effect of the perturbations is very similar between the two cadence conditions, with no significant difference in the whole-body CoM excursion (**Figure 2A**). There are substantial differences, however, in the underlying biomechanics of how this whole-body sway is generated by the balance control system. In the *low* cadence condition, participants relied on the ankle roll mechanism significantly more than in the *high* cadence condition (**Figure 2B**). This relationship was the opposite for the foot placement mechanism, which participants used heavily in the *high* cadence condition, but did not use at a statistically significant level in the *low* cadence condition (**Figure 2C**). The push-off mechanism, in contrast, was used to a similar degree in both cadence conditions (**Figure 2D**).

These results show that humans can flexibly choose which balance mechanisms they recruit, depending on a constraint on their gait pattern. In the *high* cadence condition, steps are so frequent that foot placement modulation can be used as the dominant balance mechanism, which might be more metabolically efficient, since muscle force is only used to move the swing foot and the energy for adjusting the motion of the body mass comes from the gravitational field (Kuo, 1999). In the *low* cadence condition, the duration of each stance phase is so long that waiting for the next step to adjust foot placement is less feasible, so the ankle roll mechanism is recruited during single stance. This requires more muscle force than the foot placement mechanism (Kuo, 1999). Furthermore, the force must

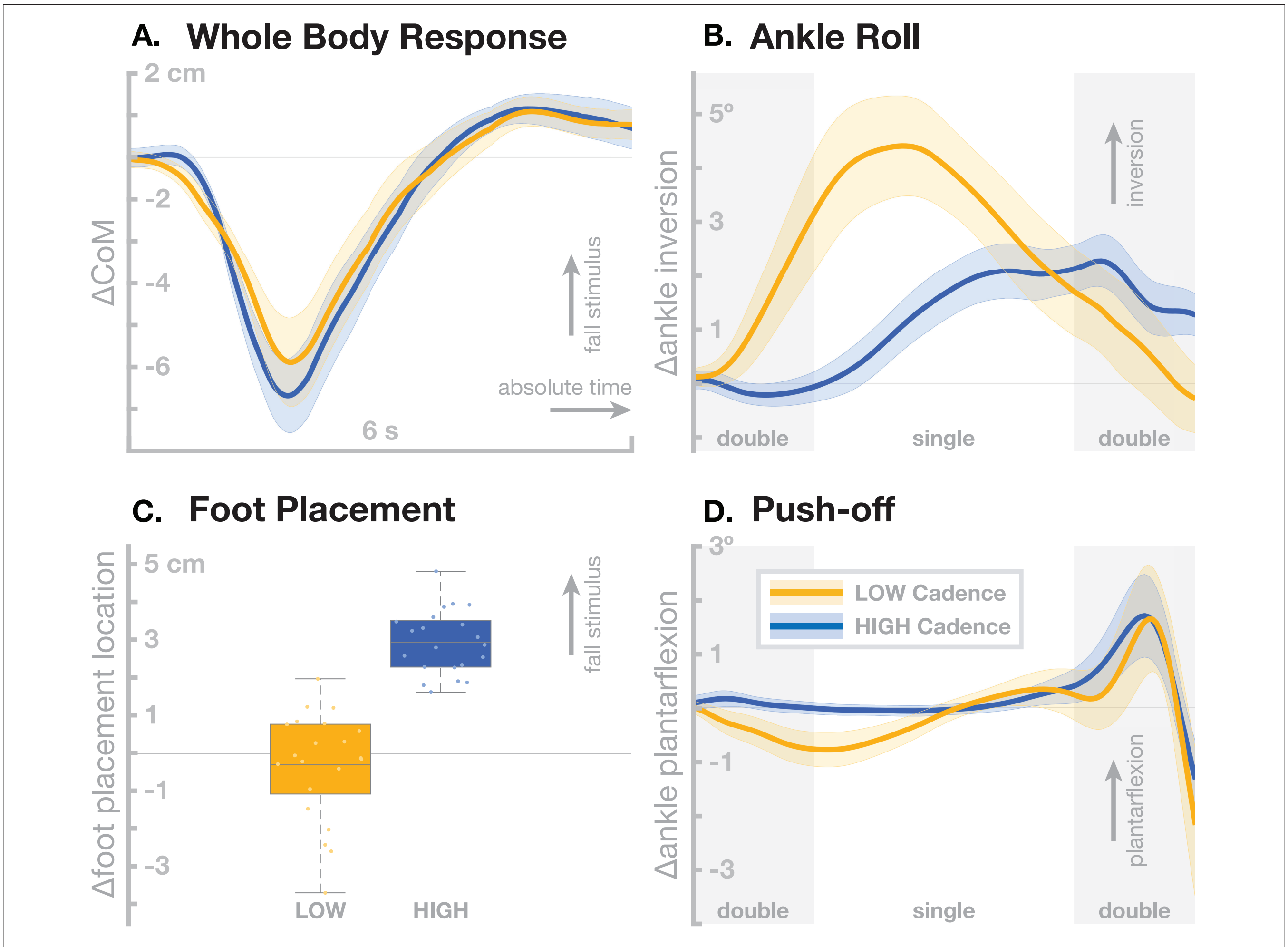


FIGURE 2 | Differences in sensorimotor control of balance depending on stepping cadence. Orange represents walking at *low* cadence (80 bpm), blue at *high* cadence (110 bpm). **(A)** Changes in whole-body CoM over six seconds after the stimulus in absolute time. The other panels show responses in the three balance mechanisms. **(B)** Responses in the ankle roll mechanism over the first step and second double stance after the stimulus in normalized time, with shaded areas marking double stance. **(C)** Differences in foot placement at the first post-stimulus step. **(D)** Responses in the push-off mechanism over the first step and second double stance after the stimulus in normalized time, with shaded areas marking double stance. Color-shaded areas give 95% confidence intervals across subject means, box plots for foot placement cover the inter-quartile ranges, with means marked by horizontal lines, whiskers showing the upper and lower adjacents and dots the individual subject means.

be generated at the distal ankle joint, rather than the proximal hip joint for foot placement, which older adults tend to favor (Tang and Woollacott, 1999). This cadence effect indicates that the tendency in older adults to walk at higher cadence might be related to their generally reduced muscle strength. By this hypothesis, older adults choose to walk at increased cadence in order to favor the foot placement mechanism for balance control over the less efficient ankle roll mechanism.

Age-related cognitive decline also affects fall risk (Lamoth et al., 2011). Walking requires attention to navigate, steer around obstacles and other people in the environment. Balance control is not considered a primarily cognitive task, the vestibular pathway for balance control might even bypass the cortex completely (Stiles and Smith, 2015). On the other hand, there is a consistent effect of secondary cognitive tasks on balance performance (Horak, 2006). Since cognition is complex, so is, necessarily, the interaction between cognition and sensorimotor control of balance. To begin investigating how cognition affects the balance control system in the control of walking, we combined our balance assessment with a virtual constraint, using a head-mounted display. We added a path to the virtual environment that contained *No-Step* zones, marked in red, and asked participants to not step on the red area. The *No-Step* zones alternated between the left and right of the path, with *Neutral* zones in gray on the other side of the path, so that any fall stimulus would induce a sensation of either falling toward a *No-Step* zone, or a *Neutral* zone. Methods and results from this study are published in detail in Fettrow et al. (2020), and here we briefly summarize the relevant findings.

Figure 3 shows the resulting responses in the whole-body sway and the three balance mechanisms ankle roll, foot placement, and push-off. The no-step zone has a statistically significant effect on the whole-body sway (**Figure 3A**), with fall stimuli toward the *No-Step* zone leading to a slightly larger CoM shift than fall stimuli toward the *Neutral* zone. This difference in the whole-body sway is largely generated by a small, statistically significant increase in the foot placement response (**Figure 3C**). There is was no significant difference in the ankle roll and push-off mechanisms (**Figures 3B,D**). While the ankle roll is slightly larger for fall stimuli toward the *No-Step* zone, this difference was not statistically significant.

These results show that there is a clear effect of a cognitive task on the sub-conscious balance responses. The overall balance response is larger when the artificial fall stimulus corresponds to a detected deviation of the body toward the no-step zone, bringing it in conflict with the cognitive task. This difference might be due to a dynamic re-weighting of the balance response gains depending on the current location of the no-step zone, or due to an active cognitive component that is superposed over the normal response.

4. UNDERSTANDING THE EFFECTS OF DIFFERENT AGE-RELATED FACTORS ON BALANCE CONTROL

Cognitive ability and muscle strength decline with age, and older people tend to walk slower and rely more on visual information. Each individual factor correlates with age and balance problems, but the factors also correlate with each other. The causal relationship between age, muscle strength, preferred walking speed, visual dependency and balance control is not well-understood. Experimentally modifying individual factors to identify their role in this causal relationship is not straightforward.

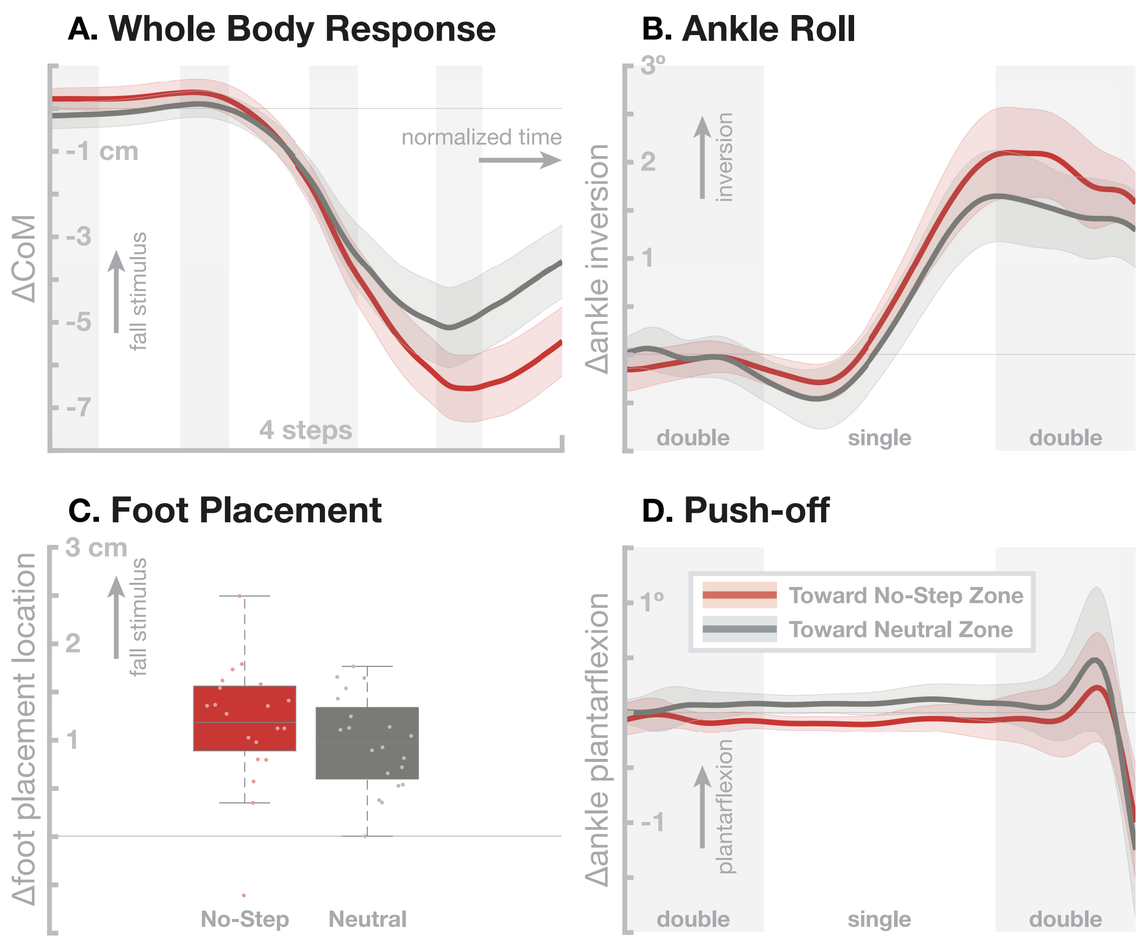


FIGURE 3 | Effect of an added virtual no-step zone on one side of the walking path on the sensorimotor control of balance. Red represents fall stimuli toward a *No-Step* zone, gray represents fall stimuli toward a *Neutral* zone. **(A)** Changes in whole-body CoM over four steps after the stimulus. The other panels show responses in the three balance mechanisms. **(B)** Responses in the ankle roll mechanism over the first step and second double stance after the stimulus, with shaded areas marking double stance. **(C)** Differences in foot placement at the first post-stimulus step. **(D)** Responses in the push-off mechanism over the first step and second double stance after the stimulus, with shaded areas marking double stance. Color-shaded areas give 95% confidence intervals across subject means, box plots for foot placement cover the inter-quartile ranges, with means marked by horizontal lines, whiskers showing the upper and lower adjacents and dots the individual subject means.

One option to understand the effect of each individual factor on the overall behavior is to develop a computational model of the whole system. In such a model, we can then modify individual factors in isolation by changing the specific parameters that describe them, and conduct simulation experiments to observe the result of these modifications on the overall behavior and stability of the system. Developing such a model is, of course, also not straightforward, though feasible in a way that some experimental manipulations are not.

In this section, we introduce an existing neuromechanical model of human walking spanning biomechanics, muscle physiology, spinal reflexes and vestibular control. We then show how this model can be "aged" to study the effect of age-related balance factors, and report the results of a simulation study to isolate the results of common age-related changes in muscle physiology, including reductions in muscle strength.

4.1. A Neuromechanical Model of Human Walking

Our neuromechanical model of walking was first introduced for the sagittal plane by Geyer and Herr (2010) and extended to 3d by Song and Geyer (2015). The model activates muscles based on a list of 10 explicit reflex modules that directly link various sensory inputs to muscle activation. Walking movement patterns emerge from the interaction of the muscle forces from these reflex modules and the reaction forces from ground contact. **Figure 4** shows an overview of the model architecture and a sample walking pattern generated by the model. The model has previously been used to predict human perturbation responses against different perturbations (Song and Geyer, 2017) and to model age-related changes in human walking performance (Song and Geyer, 2018).

The mechanics consists of seven rigid body segments (trunk, thighs, shanks, and feet) and eight degrees of freedom (hip pitch and roll, knee pitch, ankle pitch). The rigid body mechanics interact with the muscle dynamics through geometric conversion of joint angles (θ) into muscle lengths (Δl) and of muscle forces (F) into joint torques (τ). The instantaneous moment arms of the muscles are captured in the matrix R(θ). The matrix is diagonal except for a few off-diagonal terms accounting for the biarticular nature of some leg muscles. The length changes Δl together with the muscle stimulations s form the input for the computation of the muscle tendon dynamics $F(s, \Delta l)$, which are modeled as Hill-type muscles. Hill-type muscles models combine an active, force-generating element with passive parallel and serial elastic elements, generating dynamics similar to muscle-tendon units. We model major leg muscles that are involved in human gait, including the monoarticular soleus, tibialis anterior, biceps femoris short head, vastus group, gluteals, and combined hip flexors, as well as the biarticular gastrocnemius, hamstrings, rectus femoris, and the hip abductors and adductors, for a total of 11 functional muscles at each leg. Besides the forces F, the contractile elements (CE) of the muscle tendon dynamics also generate proprioceptive signals from the muscle spindles

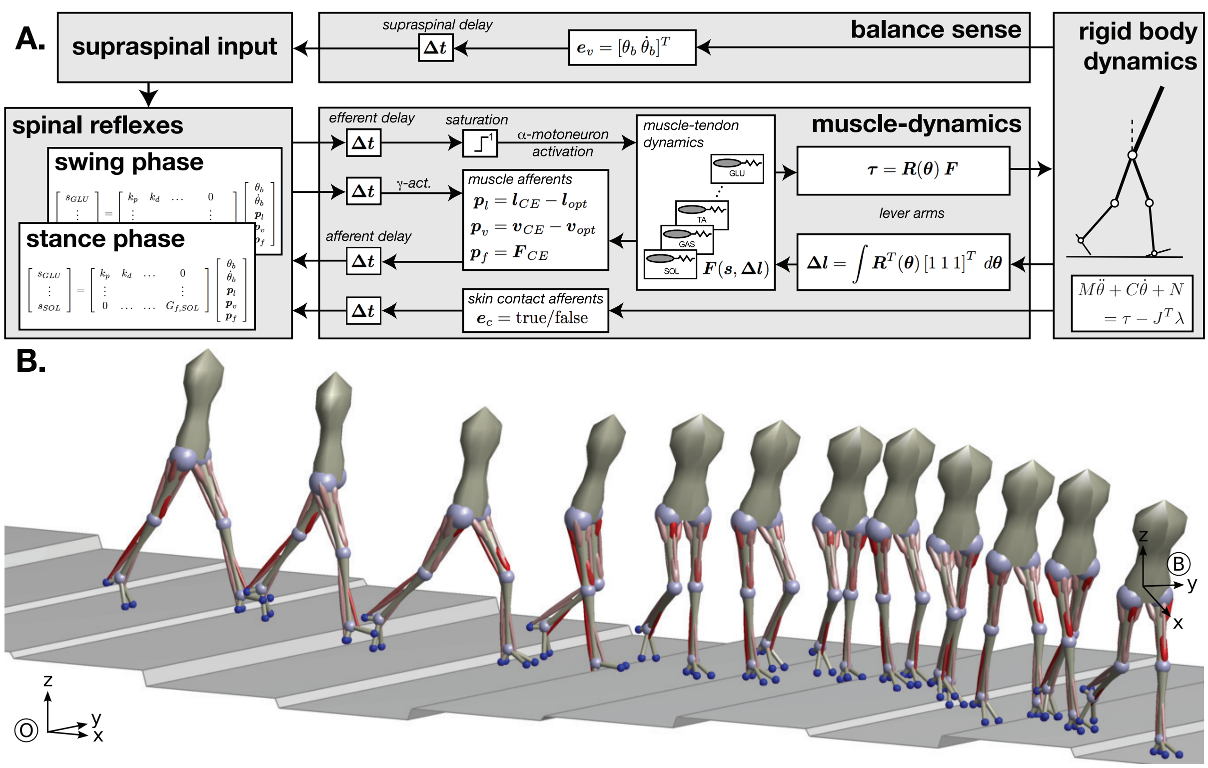


FIGURE 4 | Neuromechanical model of human walking. **(A)** Overview of the model architecture. **(B)** Example of walking behavior generated after neural controller optimization. Red cylinders indicate the leg muscles used in the model, with color saturation signaling the level of muscle activation generated by the controller.

(p_l and p_v) and the Golgi tendon organs (p_f) carrying information about the muscle length, velocity, and force. Although more complex models of these sensory organs exist, they are reduced to proportional signals with offsets for the length and velocity in this particular model.

The reflex control layer receives a range of sensory inputs and generates the muscle stimulations (α-motorneurons) and fusimotor drives (γ-motorneurons). The sensory inputs include the proprioceptive signals from the muscle dynamics and exteroceptive signals from the rigid body mechanics (e_v and e_c). The latter represent the high-level balance sense based on the visual and vestibular systems providing information about the upper body orientation (θ_b) and the mechanoreceptors providing information about the environment interaction in the form of contact detection and ground reaction forces. Note that in neuromechanical gait models the sensory organs for exteroception are generally modeled with less detail than proprioceptors. The estimate of the upper body estimation determines the target angle of the swing leg at heel-strike, generating larger steps when the body leans more. This feedback principle is similar in effect to the balance control scheme proposed by Hof (2008), but with more a complex body geometry, segment angles are more robust state variables than foot location and is more commonly used in robotics (Yin et al., 2007; Wang et al., 2012). The sensory pathways as well as the motor pathways interfacing the spinal α-motorneuron pools and the mechanical layers are time delayed (Δt), mimicking the signal transmission delays in the sensory and motor axons.

The synaptic interconnections between sensory inputs and motor outputs that form the reflex control of the different muscles in the spinal α-motorneuron pools are based on ten functional control modules which embed key functions of legged systems. The modules are organized functional groups that control the stance and swing legs. Key functions of the stance leg control modules are the generation of compliant, spring-like leg behavior, the prevention of knee hyperextension, balancing of the trunk, compensation of the swing leg interactions and flexion of the ankle to prevent ankle overextension. Swing leg modules provide ground clearance of the swing foot and move the leg to a specific target configuration. Individual swing leg modules generate ankle flexion, hip swing, and knee stabilization during the early swing phase, and decelerate and stabilize the leg in the late swing phase. In **Figure 4**, the synaptic interconnections in these reflex modules are represented by matrix multiplications. While this linear representation is accurate for many of the modeled reflexes, the model has more complex interconnections as well. For instance, some reflexes use multiplication of several inputs similar to presynaptic inhibition. Other reflexes include nonlinear effects, such as the switching between stance and swing reflex connections due to input from the mechanoreceptors.

4.2. Modeling the Effects of Aging on Sensorimotor Control of Balance

Here we demonstrate how such a mechanistic, predictive model can be used to test hypotheses about the effect of specific factors on balance control. Specifically, we use the model to investigate the effect of a list of known age-related physiological changes, most prominently loss of muscle strength, on sensorimotor control of balance. As seen in section 3.2, loss of muscle strength is associated with increased fall risk, but we do not know if loss of muscle strength directly causes increased fall risk, or if the two phenomena are only correlated. Here we demonstrate how a neuromechanical model can be used to investigate the causal relationship between muscle strength and fall risk. For simplicity, we analyze the much narrower hypothesis that decreased muscle strength causes changes in the sensorimotor feedback control of balance. This hypothesis predicts that the model with weaker muscles would show systematic differences in the feedback law mapping sensory input to motor output for balance. To test this prediction, we ran a model simulation study using two populations of young and old models.

4.2.1. Age-Related Physiological Changes

We modify the parameters representing properties of the skeleton, muscles and the nervous system to represent an ≈80 year old human, compared to the basis model that represents a ≈20 year old human. This "aging" process is done by adapting a sub-set of the physiological parameters in the model, following Song and Geyer (2018) and explained in more detail below. The model used here is a combination of different versions of the model published previously by Song and Geyer (2015, 2018). Both of these models build upon the earlier work by Geyer and Herr (2010), which showed that simple reflex modules can generate walking movement in the sagittal plane. The Song and Geyer (2015) work extended this to 3d, adding lateral degrees of freedom at the hip and an associated list of reflex modules. The Song and Geyer (2018) work extended the model by "aging" the physiological parameters and adding noise, mainly to investigate the relationship between age and walking speed, metabolic cost and fatigue. It did not contain balance control in 3d, since it was still constrained to the sagittal plane. The study reported here combines these two directions in a model that includes both balance control in 3d and "aged" physiology to investigate interactions between these.

The most prominent age-related change we model is a reduction of muscle strength by 30% and muscle contraction velocity by 20%. Beyond that, we increased eccentric force enhancement by 30% and excitation-contraction coupling time by 20%. Skeletal modifications comprise changes in the body mass distribution and in the range of motion the hip. To model the loss of leg muscles and gain of body fat, we reduced leg mass by 10% and increased trunk mass accordingly to keep the total body weight unchanged. We reduced hip motion range by 20% due to muscle contracture. Neural modifications include an increase of the transmission delays by 15%. Further implementation details can be found in Song and Geyer (2018).

4.2.2. Optimization

We obtain control parameters using the covariance matrix adaptation evolution strategy (Hansen, 2006). For each individual model, we optimize the behavior with respect to the cost function that consists of three parts:

$$J = \begin{cases} 2c_0 - x_{\text{fall}} & \text{if fall} \\ c_0 + d_{\text{steady}} & \text{if non-steady walk} \\ 100||v_{\text{avg}} - v_{\text{tgt}}|| + d_{\text{steady}} & \text{else,} \end{cases} \quad (1)$$

with $c_0 = 10^3$ and d_{steady} as a steadiness measure summing up the differences of relative positions of the body segments at touchdown. The first part of the cost function generates basic walking without falling. The second part generates steady locomotion and the third part adjusts the steady walking to a desired movement speed. A more detailed description of the optimization process can be found in Song and Geyer (2015).

Since the optimization process is stochastic, each optimization results in a different individual model. We repeated the optimization to generate populations of multiple *young* and *old* models, with a size of $N = 9$ individuals in each group, targeting a walking speed of 1.3 m s^{-1}. The sample size was limited by the processing time of the evolutionary parameter optimization, which took two to three days for a single individual.

4.2.3. Simulation Study

The optimization process resulted in two populations of *young* and *old* models. The only explicit difference between the models was in the list of age-related changes to physiological parameters specified above. It is possible, though, that the optimization process results in additional implicit differences between the populations that emerge from the evolutionary pressure encoded in the cost function (Equation 1).

We simulate a study using artificial fall stimuli on *young* and *older* participants. We model the fall stimulus in the form of a bias to the sensory signal for the trunk roll angle of 0.05 rad for a duration of 0.6 s. The magnitude of the modeled stimulus was chosen to generate a foot placement response that was approximately in the same range as the average response of human subjects to visual stimuli of $60^{\circ}s^{-1}$ (Reimann et al., 2018b) The perturbation is applied at left heel strike. We then compare the behavior of the perturbed with the unperturbed models and measure the responses to the perturbations by subtracting the unperturbed from the perturbed trajectories, analogous to the processing of experimental data (Reimann et al., 2018b).

We analyze the effect of aging on sensorimotor control of balance in the model statistically in a similar way to how we analyze experimental data (Reimann et al., 2018b). We use t-tests on the difference between the whole-body CoM at the end of the four steps and the foot placement change at the first post-stimulus step, with $\alpha = 0.05$. We additionally performed an F-test of equality of variances for the foot placement response, to test the hypothesis that older adults show increased variability under sensory perturbations.

4.2.4. Results

All individual models were able to withstand the effect of the sensory perturbation for at least four steps without falling, after which we stopped the simulations. **Figure 5** shows the motor responses generated by the *young* and *old* model populations. The shift in whole-body CoM over four steps was slightly larger in the *young* model group (**Figure 5A**), but this difference was not statistically significant, $t_{(8)} = 1.4795, p = 0.1773$. The models showed active responses in the foot placement mechanism, shown in (**Figure 5B**) Visual inspection shows that both the whole-body and the foot placement responses are similar between the two groups, and the *t*-test support this by not reporting statistical significance for the foot placement [$t_{(8)} = 0.0630, p = 0.9513$]. The *old* model population shows increased variance in the foot placement mechanism compared to the *young* models, but this difference also failed to reach statistical significance [$F_{(8,8)} = 3.4613, p = 0.0983$].

5. DISCUSSION

Older adults fall more often than young adults, but the causes for this increase are still not well-understood. One knowledge gap is our lack of understanding of the neural processes for the sensorimotor control of balance during walking, that map sensory information to appropriate motor responses to maintain balance. Another knowledge gap is that we do not understand how multiple different age-related factors that are correlated with balance control and fall risk, interact with each other and the neural control process for walking.

Here we reviewed research that attempts to overcome these knowledge gaps. Our own recent work uses artificial fall stimuli to characterize the sensorimotor control system for balance during walking by observing how it maps sensory input to motor output. In a literature review, we gathered knowledge about how the age related factors of cognitive decline, muscle weakness, gait speed and increased dependency on visual information is correlated with balance control and fall risk. We argued that the appropriate method to understand how all of these factors interact is to use computational models that formalize the existing knowledge. These models allow concrete predictions, that can then be tested experimentally. We introduced one such model and used it for a simulation study that isolates the effect of several age-related neurophysiological changes, most prominently muscle weakness.

The simulation study resulted in no statistically significant differences between the *young* and *old* model populations. This result does not support the narrow hypothesis that age-related muscle weakness causes differences in the sensorimotor feedback control of balance in walking. The more general hypothesis that age-related muscle weakness causes increased fall risk is, of course, harder to analyze and would require, among other things, a functional measurement of "fall risk" for the model. While it seems straightforward that the sensorimotor control balance affects fall risk in some way, we do not know the details of this relationship, and more research is needed to understand it.

5.1. Limitations in the Experimental Approach

While measuring how the sensorimotor control system for balance responds to fall stimuli is an important first step to understand this system, it is limited to observing the kinematic, kinetic, and electromyographic responses at the surface, and does not attempt to observe the underlying neural dynamics. We essentially treat the nervous system as a black box. This

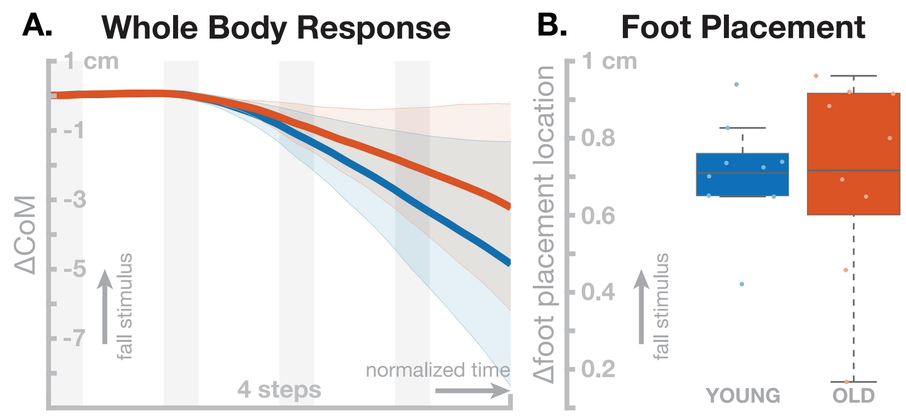


FIGURE 5 | Effect of aging in the neuromechanical model parameters on the sensorimotor control of balance. Blue represents the *young* model, red the *old* model. **(A)** Changes in whole-body CoM over four steps after the stimulus. **(B)** Differences in foot placement at the first post-stimulus step. Color-shaded areas in **(A)** give 95% confidence intervals across the individual parameter sets, box plots in **(B)** cover the inter-quartile ranges, with means marked by horizontal lines, whiskers showing the upper and lower adjacents and dots the individual data points.

is a practical limitation, as most imaging techniques have constraints that severely limit their application to study walking. Other researchers, however, have started to chip away at these constraints. Motion artifacts in EEG systems can be avoided by both software and hardware approaches (Peterson and Ferris, 2019). While movement of any kind is still a challenge for FMRI recordings, a number of studies have attempted to circumvent this obstacle and image brain activity during balance-related activities (Papegaaij and Hortobágyi, 2017; Wittenberg et al., 2017). These promising efforts should be seen as valuable alternate approaches that are complemented by the biomechanics and motor control techniques described here.

5.2. Limitations in the Model Approach

The current neuromechanical model is, of course, limited. Biomechanically, it lacks several important degrees of freedom, most notably the subtalar joint for ankle inversion and eversion, which is required for the ankle roll mechanism. Also lacking is internal rotation at the hip, a bendable spine, and arms. While the role of these degrees of freedom is more subtle, they are undeniably used in balance control.

On the neural control level, the rhythmic activation patterns in the model are generated by dedicated reflex modules, with a large number of 82 parameters that are determined using evolutionary optimization. Reflexes are without a doubt important in maintaining stability, particularly force feedback has been shown to be important for generating spring-like compliant behavior during the stance phase in many animals (Duysens et al., 2000). Other reflex modules are less directly inspired by physiological observations and were introduced based on functional biomechanical principles to generate walking behavior (Geyer et al., 2006). While the resulting model does successfully walk, it largely lacks the flexibility to adjust the walking movement patterns in goal-directed ways. The current model cannot freely modulate its walking speed, change direction, or modulate step width or length. While it is possible to add other structures that solve some of these issues, like central pattern generators (Dzeladini et al., 2014; Van der Noot et al., 2018), the hard-wired, dedicated reflex modules solving one particular task are fundamentally at odds with the seemingly effortless flexibility of human movement (Duysens and Forner-Cordero, 2018). While the current model is an impressive demonstration of how far purely reflexive movement generation can go, successfully modeling the full range of human motor control will likely require a different, more flexible approach.

The limitations in the model presented here are in contrast with models used in studies to investigate the effect of weakness, contracture and activation limits of specific muscles associated with neuromotor impairments such as stroke or cerebral palsy (e.g., Steele et al., 2012; Knarr et al., 2013; Pitto et al., 2017; Fox et al., 2018; Kainz et al., 2018). These models are generally physiologically more detailed, with more biomechanical degrees of freedom and muscles actuating them. The model is then used to estimate the muscle activation that generated experimental behavior, recorded by a combination of motion capture, force plate, and electromyographical data, using inverse kinematics, inverse dynamics and optimization (Thelen et al., 2003). This modeling approach does not deal with balance, since the balance problem was already solved by the human during the experimental session where the data was recorded. More generally, this model type does not ask questions of sensorimotor feedback control, since the approach of fitting previously recorded experimental data means that the control is by definition open-loop and cannot use sensory data to modify the control signal. This direction of modeling is, in that sense, orthogonal to the work we presented here, with both approaches are designed to answer specific questions. Ideally these two directions can be combined at some point to models that are biomechanically and physiologically detailed and capable of both generating behavior using closed sensorimotor feedback control loops and explaining observed experimental data by fitting underlying control laws and motor plans to data.

5.3. The Role of a Computational Model for Generating Knowledge

What have we learned from the simulation study performed here, and what can we learn from computational models in principle? The simulation failed to show any differences between the *young* and *old* model populations. Arguing that this is a failed experiment, however, would miss the larger point of understanding a complex system with many moving parts. We analyzed the narrow hypothesis that age-related decline in muscle strength causes changes in the sensorimotor feedback control of balance. Extending this result to fall risk would require a link between fall risk and sensorimotor control of balance. One possibility for such a link is that muscle weakness drives adaptive changes in feedback control, where weaker muscles decrease stability and the feedback control system adapts by increasing gains to re-establish robust balance. The adaptation process is implicitly modeled by the parameter optimization process, which has a measure of stability as part of the cost function (see Equation 1). The simulation study did not result in increased sensorimotor feedback gains in the *older* model, refuting the hypothesis within the limits of the assumptions. One such limit is that the stability requirement in the cost function is relatively mild, consisting only of a term that rewards body similarity in body configurations between gait cycles. A more robust accounting for stability could add various perturbations and reward robust responses. This hypothesis would also imply that sensorimotor feedback gains in older adults are actually increased. While there is a body of corroborating evidence, this prediction should still be directly tested.

This hypothesis makes intuitive sense, since reduced muscle strength limits the range of perturbations one can successfully recover from. However, rehabilitation programs for balance control targeting muscle strength have had mixed success, with best results for programs that include muscle strength training in a multi-factor approach (Horlings et al., 2008). These results imply that other factors also play a role. The simulation study here had, essentially, the same result, that muscle weakness *alone* fails to explain the observed differences in balance control between young and older adults.

A model is a form of formalized reasoning. While this model is limited, it is important to point out that other ways of generating conclusions are also limited (Smaldino, 2017). The argument chain that (A) older people have weaker muscles, and (B) weaker muscles increase fall risk, so we understand why (C) older people have increased fall risk is clearly not air-tight. This argument chain is also a model, in the sense that it draws conclusions about underlying causes from observable facts. However, such a verbal model contains vague definitions and implications. Formalized, computational models also have underlying assumptions, but they have the benefit that it is possible to bring these assumptions out into the open and discuss them explicitly. Often the attempt to formalize a complex system in a model forces us to go through this process of making our hidden assumptions explicit, allows us to study their validity and effect on the system and ultimately improves our understanding of the system as a whole.

DATA AVAILABILITY STATEMENT

The datasets generated for this study are available on request to the corresponding author.

AUTHOR CONTRIBUTIONS

HR, RR, JH, TF, and JJ wrote the manuscript. RR, HR, and HG developed the model. RR and HR designed and performed the simulation studies.

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BRIEF RESEARCH REPORT
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Indoor vs. Outdoor Walking: Does It Make Any Difference in Joint Angle Depending on Road Surface?

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Measurement of the joint angle during walking in real-world environments facilitates comprehension of the adaptation strategy corresponding to road surfaces. This study investigated the differences between the joint angles in the lower limb when walking on flat road surfaces in indoor and outdoor environments. Ten healthy young males who walked on a carpet-lined corridor in the indoor environment and on an interlocking block pavement surface in the outdoor environment participated in the study. The joint angles of their lower limbs were measured using seven inertial measurement units, and the average and coefficient of variation (%CV) of the joint angular excursion in the two environments were evaluated. The %CVs of the ankle plantar flexion excursion in the early stance was 45% higher in the outdoor environment compared with that in the indoor, although the spatiotemporal parameters and joint angular excursion of the proximal joints showed no difference between the environments. Though the road surfaces were flat from a macroscopic point of view, the interlocking block pavement had stiffer and more irregular characteristics. The variability in the ankle plantar flexion motion in the early stance may be most likely affected by these surface characteristics in the real-world outdoor environment.

Keywords: ankle plantar flexion excursion, walking, outdoor environment, inertial measurement unit (IMU), motion capture (Mocap)

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INTRODUCTION

In daily life, people walk on various types of road surfaces, since walking is indispensable for promoting social life and health (Jacobs et al., 2008). However, in Japan, several falls occur owing to environmental factors such as the surface characteristics of outdoor environments (Niino et al., 2003). Thus, analyzing the physical behavior of walking on various road surfaces is important for understanding the walking strategy necessary to adapt to the different surface characteristics of outdoor environments.

Studies on gait analysis have primarily been performed in laboratories—where the pathways are clean, flat, and short—using optical motion capture systems (MoCap) (Winter, 1984; Kadaba et al., 1990). To simulate the effects of various terrains, the joint kinematics during walking have been evaluated on walkways with randomly placed wooden blocks beneath artificial grass (Thies et al., 2005a,b; Menant et al., 2009), compliant foam (MacLellan and Patla, 2016), and loose rock surface (Gates et al., 2012) constructed for research purpose in laboratories. When walking on such surfaces, the variability in the step width and stride time (Thies et al., 2005a,b), the peak joint angle, and the standard deviation (SD) of the hip, knee, and ankle across the gait cycle (Gates et al., 2012) increased, while the walking speed and stride length (Thies et al., 2005b; Menant et al., 2009) decreased. These studies revealed that the kinematic profiles during walking are adapted to the corresponding irregular road surface. However, the pathways in the laboratory were not as long as those in the outdoor environments, and the surface characteristics differ from those of the real-road surfaces because they were constructed for research purposes. Therefore, it was unclear whether the kinematic change during walking according to the irregular road surfaces obtained in these studies reproduce those in real outdoor environments.

Recent studies have attempted to quantitatively evaluate walking in an outdoor environment using inertial measurement units (IMUs). Specifically, the cadence and speed of daily walking (Weiss et al., 2013; Fasel et al., 2017), and the variability and stability of the acceleration waveform in outdoor conditions (Iosa et al., 2012a,b; Tamburini et al., 2018) have been evaluated from time-series acceleration data measured by IMUs. These studies have reported that the spatiotemporal parameters were affected by the walking environment (Iosa et al., 2012a,b; Fasel et al., 2017; Tamburini et al., 2018). These previous studies focused on the evaluation of walking in the outdoor environment by analyzing of the acceleration data, but the joint angles in the lower extremity were not evaluated when walking in the outdoor environment.

To overcome the limitation of optical MoCap systems, MoCap systems using IMUs attached to each body segment have been developed (Roetenberg et al., 2009; Seel et al., 2014). Maruyama et al. (2018) developed a real-time MoCap system using IMUs that can measure the position of the subjects as well as the joint angles. In previous studies, the accuracy of the IMU-based MoCap systems was evaluated, and it was confirmed to be excellent for hip, knee, and ankle joint angles in the sagittal plane during walking on a flat surface (Al-Amri et al., 2018; Maruyama et al., 2018). Therefore, by using an IMU-based MoCap system, the joint motion in the sagittal plane during walking in environments other than the laboratory can be investigated; this was not possible using conventional optical MoCap systems.

Measurement and analysis of the joint motion during walking on different types of road surfaces in the outdoor environment can facilitate comprehension of realistic adaptation strategy corresponding to various types of road surfaces. Hence, this study investigated the difference between the joint angles when walking on indoor and outdoor road surfaces by an IMU-based MoCap system. In daily life, people often walk on paved flat surfaces, and rarely walk on irregular road surfaces as constructed for research purposes in the previous studies (Gates et al., 2012; Blair et al., 2018; Dixon et al., 2018). Therefore, we hypothesized that the joint motions in the lower extremity were not affected by the flat road surface in real outdoor environments.

METHOD

Participants

Ten healthy young males (age: 24.1 ± 1.9 years, height: 1.70 ± 0.05 m, weight: 61.4 ± 8.3 kg) participated in this study. None of the subjects had any history of neuromuscular diseases, trauma, or orthopedic diseases. The experimental protocol was approved by the local ethical committee, and all the participants provided written informed consent before participating.

Data Collection

Each participant had seven IMUs (MTw; Xsens Technologies Inc., Enschede, Netherlands) attached to the sacrum, bilateral thigh, shank, and foot (**Figure 1**). Before the walking session, the participants were asked to adopt a reference pose for calibrating the IMU-based MoCap system, in which the IMU orientation relative to the corresponding body segment was determined. The subjects walked along a straight carpet-lined corridor in the indoor environment and on an interlocking block pavement surface in the outdoor environment. This is because the IMU-based MoCap system used in this study had been validated only in a laboratory with flat floor surfaces. The slope in the progression direction of the outdoor walkway was $<1^\circ$ as measured by a three-dimensional laser scanner (FOCUS^s 70; FARO Inc., Lake Mary, USA) (Yang et al., 2013). Although the road surfaces are flat from a macroscopic point of view, the interlocking block pavement is stiffer than the carpet, and have small irregularities due to the misalignment of the blocks (Hata et al., 2003).

The walking distance was ∼90 m which was equal to the maximum length of the corridor in the indoor environment (**Figure 2**). All walking sessions were conducted at a self-selected preferred walking speed and with the same shoes (BioTF 02; Moonstar Inc., Fukuoka, Japan). The order of the two walking sessions was randomized. During these sessions, the data from the IMUs were sampled at 60 Hz, and the longest measurement duration was <5 min. Within this duration, the drift error of the IMU is negligible, as reported previously (Robert-Lachaine et al., 2017; Paulich et al., 2018). The errors of the angles relative to those measured using the optical MoCap system ranged from $2.0^\circ \pm 0.3^\circ$ (ankle) to $10.9^\circ \pm 4.0^\circ$ (hip) in the sagittal plane (Maruyama et al., 2018). The waveform similarities were also evaluated using the cross-correlation coefficient and were confirmed to range from 0.86 (ankle) to 0.97 (knee) under this measurement condition.

Data Analysis

The joint angles of the hip, knee, and ankle in the sagittal plane, and the position of the center of mass (CoM) of the whole-body model were calculated using a posture-reconstruction plugin (Maruyama et al., 2018) running on DhaibaWorks—our self-developed motion analysis software (Endo et al., 2014). This plugin reconstructed the lower limb motion by combining the orientation

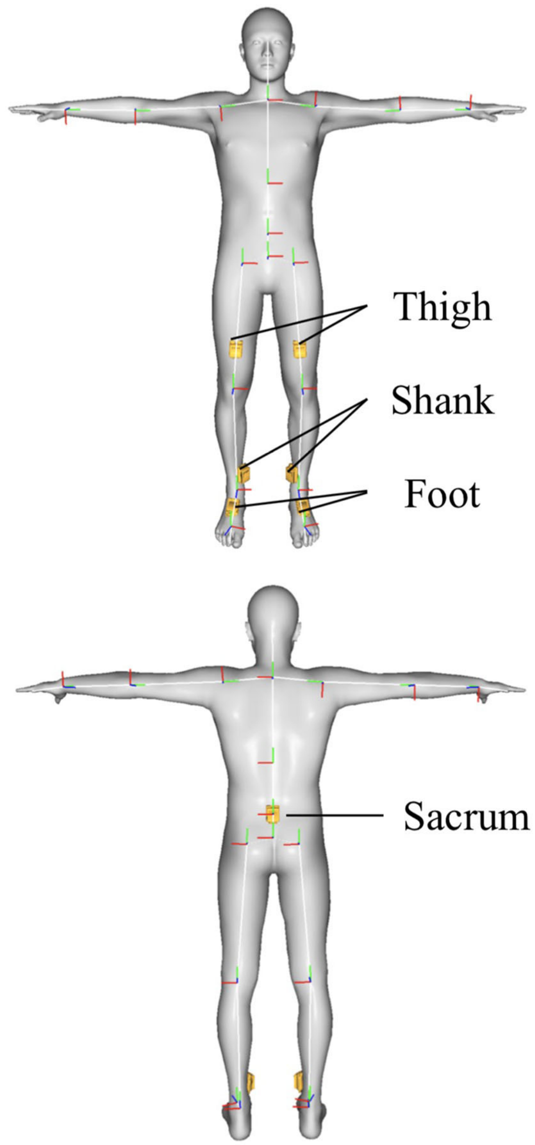

FIGURE 1 | Placement of inertial measurement units on the whole-body model.

FIGURE 2 | Photographs depicting **(A)** the corridor in the indoor environment and **(B)** the walkway in the outdoor environment where the measurements were performed. Subjects walked on a level surface and over an interlocking block pavement surface. The slope angles in the progression and lateral direction of the outdoor walkway were <1°.

data of each IMU and the individual body model with a link structure. The dimensions of the body model were estimated statistically from the participant's height and weight, based on the database of Japanese body dimensions (Endo et al., 2014).

Data for 30 strides during steady-state walking were extracted, as similarly performed in a previous study to analyze stride-to-stride kinematic variability (Dingwell and Cavanagh, 2001). In addition, the joint angular excursions were calculated from the amplitude of the displacements between the key points in a gait cycle (**Figure 3**). The mean and SD values were calculated across the gait cycle, and the coefficient of variation (%CV) was calculated as an index of the variability of the joint angular excursion, as follows:

$$\mathbf{\%CV} = \frac{\mathbf{SD}}{\mathbf{Mean}} \times \mathbf{100}$$

The walking speed (m/s) and cadence (step/min) were calculated using the Euclidean distance of the position of the CoM in the horizontal plane and using one gait cycle time, respectively. These values were calculated for 30 gait cycles and subsequently averaged.

The spatiotemporal parameters, joint angular excursion, and %CV were calculated using MATLAB R2018a (MathWorks Inc., Natick, USA).

Statistical Analysis

Differences in mean and %CV of the spatiotemporal parameters and joint angular excursions of the hip, knee, and ankle of

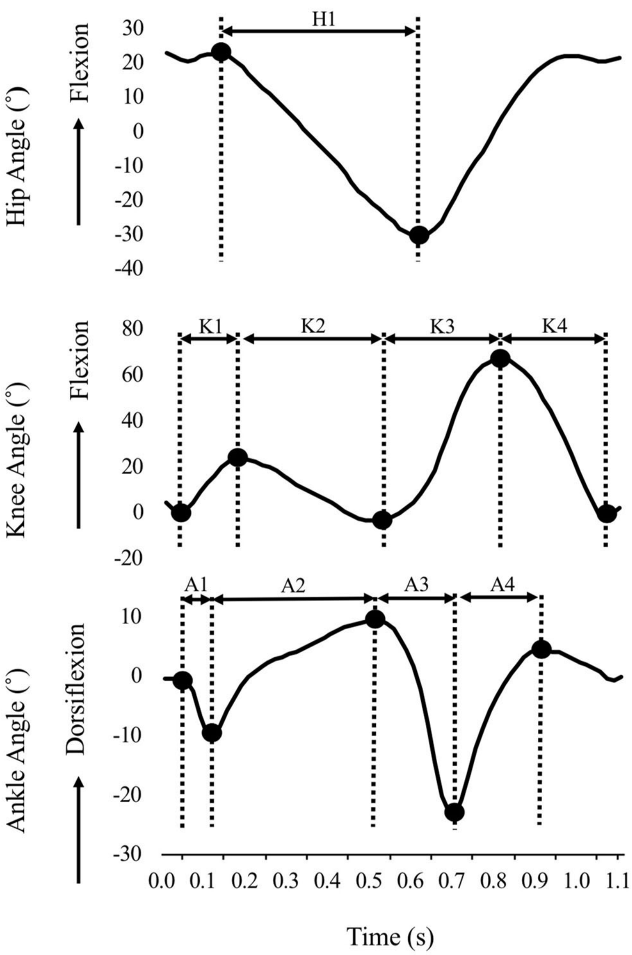

FIGURE 3 | Graphical representation of joint angles of the hip, knee, and ankle in time series. Black dots indicate the key points in a gait cycle. The joint angular excursions were calculated from the amplitude of the displacements between these values. H1: Hip joint excursion; K1: Knee flexion excursion in early stance; K2: Knee extension excursion in mid-stance; K3: Knee flexion excursion in late stance; K4: Knee extension excursion in swing; A1: Ankle plantar flexion excursion in early stance; A2: Ankle dorsiflexion excursion in mid-stance; A3: Ankle plantar flexion excursion in late stance; A2: Ankle dorsiflexion excursion in swing.

the subjects, between indoor and outdoor road surfaces, were analyzed using Wilcoxon signed ranks tests. Values of $p < 0.05$ were considered statistically significant. All data were analyzed using SPSS Statistics version 25.0 (SPSS Inc., Chicago, USA). The r values were calculated as effect sizes that are the magnitudes of the differences between the environments. The amplitudes of these values were classified as small ($0.1 \leq r < 0.3$), moderate ($0.3 \leq r < 0.5$), and large ($0.5 \leq r$).

RESULTS

Table 1 shows the analysis results. The mean values of the spatiotemporal parameters and the joint angular excursions of the hip, knee, and ankle did not differ significantly between the environments.

The %CV of the plantar flexion excursion of the ankle in the early stance in the outdoor environment was 45% higher than that in the indoor environment. This difference yielded large effect size ($r = 0.76$). However, no statistically significant differences were observed in %CVs of the spatiotemporal parameters and the hip and knee joint angular excursions between the environments.

DISCUSSION

This study examined the differences in the joint angles of the lower extremity when walking on flat road surfaces in indoor and outdoor environments. Nevertheless, the %CV of the ankle joint angular excursion in early stance was confirmed to be higher in the outdoor environment, without any changes to the spatiotemporal parameters and the joint angular excursions of the hip and knee joints. These results did not support our hypothesis.

The difference observed in the %CV value between the indoor and outdoor environments indicates that the variability in the ankle plantar flexion excursion increases when walking in the outdoor environment. On the contrary, the walking environment did not influence the amplitude of the angular excursions of the hip, knee, and ankle joints. Previous studies performed in the laboratory reported that the joint angles of the hip, knee, and ankle increased when walking on a destabilizing loose rock surface (Gates et al., 2012) and an uneven surface (Blair et al., 2018; Dixon et al., 2018). In addition, the vertical CoM movement decreased with a large flexion motion of the trunk and lower extremity (Gates et al., 2012). These kinematic changes reflect motor control strategies to overcome perturbations imposed by the uneven road surface. The results of our study did not completely conform to those of these previous studies because the interlocking block pavement was flat compared with the previous studies (Gates et al., 2012; Blair et al., 2018; Dixon et al., 2018), although it had small irregularities due to the misalignment of the blocks. Nevertheless, we found that compared with the indoor environment, the interlocking block road in the outdoor environment leads to the increase in the variability in the ankle plantar flexion excursion in the early stance without affecting the variability of the joint angular excursion of the hip and knee.

The ankle plantar flexion motion in the early stance provides the contact of the foot with the ground. Therefore, adapting the plantar surface of the foot to the walking surface through this motion is important for stable walking on uneven terrain (Gates et al., 2013). In this study, subjects walked on the carpet-lined corridor in the indoor environment and the inter-rocking block pavement in the outdoor environment. Although both road surfaces were flat from a macroscopic point of view, the interlocking block pavement was stiffer than the carpet and had small irregularities due to the misalignment of the blocks (Hata et al., 2003). Thus, the variability of the ankle plantar flexion excursion, which provides the initial contact between the foot and the ground in early stance becomes large to adapt to

TABLE 1 | Mean and coefficient of variation (%CV) of the spatiotemporal parameters and joint angular excursions.

Variable	Joint	Motion	Phase	Indoor	Outdoor	*p*-value	Effect size
Mean value							
Walking speed (m/s)				1.48 (1.35–1.58)	1.52 (1.35–1.60)	0.51	0.21
Cadence (step/min)				117.1 (111.6–118.4)	117.9 (111.2–119.7)	0.20	0.40
Angular excursion (°)	Hip	Excursion		48.1 (47.3–48.5)	50.8 (46.5–51.8)	0.58	0.18
	Knee	Flexion	Early stance	19.9 (18.2–22.9)	22.4 (20.3–24.1)	0.33	0.31
		Extension	Mid-stance	20.0 (17.7–22.2)	23.3 (21.8–24.7)	0.11	0.50
		Flexion	Late stance	64.8 (60.9–68.9)	66.2 (59.7–71.7)	0.33	0.31
		Extension	Swing	66.9 (63.2–67.1)	67.3 (62.8–71.3)	0.96	0.02
	Ankle	Plantar flexion	Early stance	9.3 (8.5–10.2)	8.0 (6.7–10.4)	0.45	0.24
		Dorsiflexion	Mid-stance	17.3 (15.9–17.4)	17.5 (14.2–17.9)	0.45	0.24
		Plantar flexion	Late stance	37.5 (34.5–38.2)	37.8 (34.0–39.4)	0.88	0.05
		Dorsiflexion	Swing	30.4 (26.8–35.4)	28.3 (25.3–28.6)	0.14	0.47
%CV (%)							
Walking Speed				3.1 (2.2–4.1)	3.0 (2.4–3.9)	0.95	0.02
Cadence				1.9 (1.5–2.1)	1.7 (1.4–1.9)	0.57	0.16
Angular excursion	Hip	Excursion		2.6 (2.4–2.9)	2.5 (2.3–2.6)	0.96	0.02
	Knee	Flexion	Early stance	8.5 (7.0–9.6)	10.1 (8.4–10.6)	0.96	0.02
		Extension	Mid-stance	8.6 (7.1–9.1)	8.5 (7.0–8.7)	0.45	0.24
		Flexion	Late stance	2.5 (2.2–3.0)	3.3 (2.9–3.6)	0.06	0.60
		Extension	Swing	2.4 (1.5–2.8)	2.8 (2.2–3.0)	0.24	0.37
	Ankle	Plantar flexion	Early stance	13.6 (11.3–16.9)	19.8 (17.4–21.2)	0.02*	0.76
		Dorsiflexion	Mid-stance	9.9 (8.0–10.8)	10.7 (9.6–12.3)	0.20	0.40
		Plantar flexion	Late stance	4.5 (3.1– 6.1)	5.0 (4.0–7.3)	0.29	0.34
		Dorsiflexion	Swing	6.4 (5.1–7.9)	9.2 (6.8–12.8)	0.14	0.47

Values: central value (Lower quartile-Upper quartile).
**Significant difference between the indoor and outdoor environments ($p < 0.05$).*

the road surface in outdoor environments. We speculated that the proximal joints and the other phases were not affected by the surface characteristic using this adaptation in the healthy young subjects. On the other hand, the people with the ankle-foot orthosis and prosthetic foot (Gates et al., 2013) and the older people with peripheral muscle weakness (Menz et al., 2004) cannot be variably varied the ankle plantar flexion motion in the early stance. Even walking on the relatively flat surface in the outdoor environment, their proximal joints were likely to increase in the variability during the entire stance phase.

In addition to the surface characteristics, visual information also differed between the indoor corridor and outdoor open-field environments. The visual perturbation affected the step length (Iosa et al., 2012a,b) and width (Franz et al., 2015) and their variabilities (Thompson and Franz, 2017). In particular, these effects were larger among older adults than young people. In this study, the only variability of the ankle plantar flexion in the loading response that had a small association with the step width and length (Van Hedel et al., 2006) were affected by the walking environments. In young people, the effect of the difference in the visual information between the environments was small. We speculated that the difference in the variability in the ankle plantar flexion was due to the surface characteristics.

Despite the new revelations, our study has certain limitations. First, the walking motion was analyzed over only a single surface type in an outdoor environment, though in real life, people walk on various types of surfaces, from flat asphalt to irregular gravel. In this study, it is unclear which factors of the surface characteristics—stiffness or irregularity—affected the increase in the variability of the ankle plantar flexion excursion during loading response. Further research is necessary to evaluate the walking strategy corresponding to the characteristics of terrains in real outdoor environments. Second, all participants in this study were healthy young males. Previous studies clarified that the joint kinematics of the elderly were affected by the road surface more significantly than those of the young (Blair et al., 2018; Dixon et al., 2018). Further studies are needed to investigate the variability in the joint angular excursion during real outdoor walking for the elderly. Third, the small sample size and many comparison tests may produce a type I and II error, respectively. Nevertheless, the difference in the %CV of the ankle plantar flexion excursion between the environments had a large effect size that is a quantitative measure of the magnitude of the difference. Therefore, we speculated that the walking environments affected the %CV of the ankle plantar flexion excursion in the early stance.

To our knowledge, this is the first study that has investigated the differences in the joint motion during walking between the indoor and the real-world outdoor environments. The variability in the ankle plantar flexion in the early stance phase increased when walking in the outdoor environment, although the spatiotemporal parameters and joint angular excursion of the hip and knee joints were not different between the two walking environments. The measurement and analysis of the joint motion during walking in the real-world environment make it possible to reveal a more realistic adaptation strategy corresponding to the outdoor road surface. This study suggests that the variability of the ankle plantar flexion excursion during loading response becomes large to adapt to the road surface in the outdoor environment, without affecting the joint angular excursion of the hip and knee.

DATA AVAILABILITY STATEMENT

Datasets are available upon request to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by National Institute of Advanced Industrial Science and Technology. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

HT, TM, and MT conceived and designed the experiments and interpreted data. HT and TM performed the experiment. HT conducted data analysis and drafted the manuscript. TM and MT edited and revised the manuscript and approved the final version. All authors contributed to the article and approved the submitted version.

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Slow but Steady: Similar Sit-to-Stand Balance at Seat-Off in Older vs. Younger Adults

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Many older adults suffer injuries due to falls as the ability to safely move between sitting and standing degrades. Unfortunately, while existing measures describe sit-to-stand (STS) performance, they do not directly measure the conditions for balance. To gain insight into the effect of age on STS balance, we analyzed how far 8 older and 10 young adults strayed from a state of static balance and how well each group maintained dynamic balance. Static balance was evaluated using the position of the center-of-mass (COM) and center-of-pressure (COP), relative to the functional base-of-support (BOS). As the name suggests, static balance applies when the linear and angular velocity of the body is small in magnitude, in the range of that observed during still standing. Dynamic balance control was evaluated using a model-based balance metric, the foot-placement-estimator (FPE), relative to the COP and BOS. We found that the older adults stay closer to being statically balanced than the younger participants. The dynamic balance metrics show that both groups keep the FPE safely within the BOS, though the older adults maintain a larger dynamic balance margin. Both groups exhibit similar levels of variability in these metrics. Thus, the conservative STS performance in older adults is likely to compensate for reduced physical ability or reduced confidence, as their dynamic balance control does not seem affected. The presented analysis of both static and dynamic balance allows us to distinguish between STS performance and balance, and as such can contribute to the identification of those older adults prone to falling, thus ultimately reducing the number of falls during STS transfers.

Keywords: sit-to-stand, balance, static, dynamic, foot placement estimator, coordination, elderly, stability

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1. INTRODUCTION

The ability to get up from a chair is a fundamental prerequisite to perform daily activities and function independently. Unfortunately, difficulty moving from a sit to a stand (STS) is common among older adults, and affects the lives of over 6% of those living independently as well as over 60% of long-term care residents (Jeyasurya et al., 2013). Poor STS performance can make ambulatory older adults prisoners in their chairs and can result in falling. More specifically, more falls occur during STS (including stand-to-sit) compared to walking, especially in residents with higher fall frequency (Rapp et al., 2012; Pozaic et al., 2016; van Schooten et al., 2017). As falls are the number one cause of injuries in older adults over the age of 65, it is imperative to understand and improve their STS performance (Janssen et al., 2002; Millor et al., 2014).

STS is considered the most mechanically demanding task of common daily activities, requiring leg muscle strength, coordination, and balance control (Riley et al., 1997; Millor et al., 2014). Limitations in any one of these factors is suggested to cause poor STS ability, resulting in STS attempts that fail, leaving older adults to sit back down or take a step if possible, potentially creating an even more unstable situation. Surprisingly, metrics to analyze balance during STS have yet to be found to identify those at risk of falling.

Clinical tests commonly measure the duration of, or ability to perform, a number of STS movements, which yields little information regarding any underlying problems (Bohannon, 2012; Silva et al., 2014). Biomechanical studies have described the STS motion, different STS compensatory strategies, and evaluated the effect of factors, such as chair height and foot placement on STS difficulty (Aissaoui and Dansereau, 1999; Janssen et al., 2002; Millor et al., 2014; Boukadida et al., 2015). Only a few studies have aimed to evaluate balance during STS using metrics such as transfer duration, body or trunk dynamics around seat-off, or the location of the body's center of mass or pressure relative to the ankle at seat-off (Moxley Scarborough et al., 1999; Åberg et al., 2010; Akram and McIlroy, 2011; Fujimoto and Chou, 2014). However, these metrics are unable to discern the difference between movement and balance. In addition, most of these studies measure STS with the arms crossed at the chest, while this might affect balance and does not reflect STS movement in daily life.

Recently, a model-based balance metric called the foot placement estimator (FPE), originally developed to study balance in bipedal robotics, was applied to assess dynamic balance during walking in healthy adults, children and patients with movement disorders (Wight et al., 2007; Millard et al., 2009, 2012; Bruijn et al., 2013). The 3D FPE (Millard et al., 2012) uses a 3D inverted pendulum model to calculate where the center-of-pressure (COP) should be placed with respect to the center-of-mass (COM) so that the participant can passively reach a statically balanced standing position (Millard et al., 2012). Although the FPE is similar to the capture-point (Pratt et al., 2006) and the extrapolated center of mass (Hof, 2008), it improves upon these methods by taking both linear and angular momentum into consideration. The FPE has been validated for analyzing gait but the metric has not yet been applied to quantify dynamic balance during STS. The FPE can express the dynamic balance margin relative to the base-of-support (BOS), taking into account specific foot placement and dynamics of the STS movement, to predict how close individuals come to taking a corrective step. In addition, the relative placement of the COP to the FPE can be used to evaluate how well individuals control their direction of travel and speed.

The aim of this paper is to compare how accurately older adults control their balance during STS compared to young adults, by analyzing the motion of the COM, COP, and FPE relative to the BOS of individuals. STS transfers were performed with both more natural arm position (hanging at the side) and arms crossed at the chest to allow for comparison to literature. First, we evaluated static balance by calculating how far each participant was from meeting the conditions for static balance at seat-off, by calculating how far the COM ground projection (COM_{GP}) is from the BOS, the distance between the COM_{GP} and the COP, the COM speed, and finally the average angular speed of the body. Second, we used the FPE to evaluate how well dynamic balance was controlled by evaluating the distance between FPE and BOS as well as how accurately participants tracked the FPE with their COP. We expect that older adults who struggle with STS motions will stay closer to being statically balanced but will struggle to remain dynamically balanced and thus will come closer to taking a compensatory step than younger adults. We also expect older adults to display more variability in all static and dynamic balance measures.

2. METHODS

Sit-to-stand transfers were performed in two different conditions: with arms relaxed at the side (Side) as a more natural, primary condition and with arms crossed at the chest (Chest) for comparison to other STS studies in literature. For the Side condition, participants were instructed to start each STS with arms at their side, but no further instructions were given on arm movement during the task. Participants were asked to perform five consecutive STS movements at their own pace, starting and stopping each cycle with about 2 s of sitting and standing still. Still standing trials were also performed so that the bounds on the various static balance metrics, described in section 2.3, could be established.

A stool (used for marker visibility reasons) was placed at one force plate. As seat height is known to affect STS difficulty (Janssen et al., 2002), it was set to the height of the knee epicondyles, which was similar between groups (Y: 49 $\pm$ 4 cm, E: 50 $\pm$ 4 cm). Participants were instructed to position their feet at their preferred position on the second force place and to not move their feet during the entire experiment. It is important to note that the foot placement relative to the stool, which is known to affect STS difficulty, was similar between young and older adults (**Supplementary Figure 1**) (Janssen et al., 2002).

Data were simultaneously collected using the two ground-embedded force plates at 900 Hz (Bertec, Columbus, OH, USA) and a passive motion capture system at 150 Hz (type 5+ cameras, Qualisys, Gothenburg, Sweden). Motion capture markers (14 mm) were placed on each participant according to the IOR full body model (Cappozzo et al., 1995; Leardini et al., 2011), with extra iliac crest and greater trochanter markers to guarantee tracking throughout the STS motion. Marker and force data were filtered with a bi-directional low-pass Butterworth filter (with a cut-off frequency of 10 Hz). The body COM position, COM velocity, moment-of-inertia about the COM, and angular momentum about the COM was calculated using the IOR full body human model, with a separate upper and lower trunk, in Visual3D (v6, C-motion, Inc., Germantown, MD, USA). As most arm markers were covered up during the Chest condition, the weight of the arm segments was added to the upper trunk in this condition.

To characterize the older participants, we administered established clinical tests and questionnaires. First we determined the participant's frailty level according to the Clinical Frailty Scale (Rockwood et al., 2005) and cognitive function using the Mini-Mental State Examination (Folstein et al., 1975; Tombaugh and McIntyre, 1992) as part of the screening process. After the STS evaluation and a break, participants performed the short physical performance battery (SPPB) (Guralnik et al., 1994) to assess functional ability. At the end of the session, we administered the Barthel Index of activities of daily living (Mahoney and Barthel, 1965) as another measure of functional ability and evaluated fear of falling using the falls efficacy scale international (FES-I) (Hauer et al., 2010) and asked for the number of falls in the last year (see **Table 1**).

2.1. Participants

Ten able-bodied younger adults (28 ± 5 years, 4 females, 24 ± 2 kg/m^2 BMI) and eight older adults (79 ± 8 years, 6 females, 29 ± 6 kg/m^2 BMI) were included (see **Table 1**). Younger participants had to be between 18 and 45 years of age and older participants above 65 years. Participants were excluded from either group if they had neurological, cardiovascular, metabolic, visual, auditory, mental or psychiatric impairments or injuries that made the participant unable to independently perform the sit-to-stand task or walk short distances in the lab. Older participants were excluded if their frailty level was more than moderately frail (Clinical Frailty Scale score $\geq$6) (Rockwood et al., 2005) and if they were severely cognitively impaired (Mini-Mental State Examination $\leq$17) (Folstein et al., 1975; Tombaugh and McIntyre, 1992). It should be noted that participant O6 presented with chronic hemiplegia, but was included as the participant was able to independently perform STS. Since stroke is the leading cause of long-term disability in older adults, the inclusion of O6 contributed to creating a more representative sample of older adults. The protocol was approved by the Institutional Review Board of the medical faculty of Heidelberg University and all participant provided written informed consent.

2.2. Movement Segmentation

Most of our analysis focuses at the time of seat-off which is also the time associated with the fastest movements during STS. As such our analysis is sensitive to how accurately and consistently the time of seat-off is identified. Given the considerable variability among participant's movements, we developed a two-stage algorithm to automatically identify STS transfers and to consistently identify the times of initiation, seat-off, and standing.

In the first stage of the algorithm we used the k-means++ algorithm (Arthur and Vassilvitskii, 2007) to identify candidate motion sequences by clustering both the COM height and the vertical force recorded by the stool's force plate into three clusters. The three COM height clusters were identified and labeled using the mean value of each cluster: the cluster with the lowest mean height is the seated-height cluster, the cluster with the highest mean height is the standing-height cluster, while everything else is in the crouching-height cluster. Similarly the three force plate clusters were identified and labeled using the mean values of each cluster: the cluster with the largest vertical force reading is the seated-force cluster, the cluster with the lowest vertical force reading is the standing-force cluster, and everything else is in the transition-force cluster.

This process yields two vectors in which every point in time has two labels: one label from the COM height clusters, and another from the force plate clusters. Candidate motion sequences were identified by finding all time sets with standing-height label (at least 0.5 s in length) that are connected backwards in time to a seated-force label (also at least 0.5 s in length) with only a single transition from the transition-force cluster to the standing-force cluster in between. Sequences which met this criteria but included times at which the feet broke contact with the ground (at least 3 of the 6 foot markers must move by $<$15 mm in the vertical direction to be acceptable) were rejected.

In the second stage the event times of the motion sequences were refined. The beginning of the STS transfer was identified as the point in time at which the COM speed was 1 cm/s higher than the minimum value observed in the sitting-force set. The time of seat-off is defined by the point in time in which the vertical force of the stool force plate is within 1 N of the value it registers during the standing-height cluster (when the participant is not in contact with the stool). Stance is defined by the point in which the participant's COM is within 1 cm of the median height of the standing cluster and moving upwards with a speed of $<$1 cm/s.

This combination of clustering and adaptive thresholding allowed us to consistently identify movement segments from even the most variable of our participants while discarding as little data as possible. In the Side condition we had to reject 1 trial from O1 (sit-back), 1 trial from O5 (short still sitting), and 7 trials from O6 who presented with hemiplegia (1 with short still standing and 6 with a compensatory step). Since the difficulties O6 faced were clear during the data collection we measured additional trials until 5 successful STS transfers were recorded. In the Chest condition we had to reject 2 trials from O6 (1 with short still sitting and 1 sit-back) and 1 trial from O7 (short still sitting). While our analysis focused on the moment of seat-off, data on the transfer between seat-off to stance, and the median stance values are presented in the figures for context.

2.3. Balance Analysis

The definition of what constitutes balanced movement is task dependent: when a gymnast performs a tumbling routine the pelvis may contact the ground during balanced movement, in contrast, during STS transfer from a chair this would indicate that a fall has occurred. To analyze STS transfers we will adopt the following definitions:

Definition: The participant has *fallen* if any part of their body other than the bottoms of their feet touch the ground.

Definition: The participant is *statically balanced* if they have not fallen, and the linear and angular speed of their body is small. Here the term 'small' is taken to mean within the bounds observed during still standing.

TABLE 1 | Older adult (O) characteristics.

	Age	Frailty	Cognition	Functional ability			Falling	
	(years)	CFS (impair:+)	MMSE (impair:-)	Barthel I (impair:-)	SPPB (impair:-)	Aid	falls/years	FES-I (afraid:-)
O1	80–85	3	29	85	12	Cane	1	16
O2	90–95	5	29	95	10	None	1	11
O3	75–80	2	29	100	10	None	0	11
O5	80–85	3	27	100	7	Stick*	1	13
O6	65–70	3	24	95	8	Cane	0	17
O7	80–85	4	21	95	8	Rollator	0	12
O8	65–70	1	29	100	11	None	0	9
O9	75–80	3	29	100	12	Cane*	0	9

*With the age range indicated for anonymity; CFS the Clinical Frailty Scale testing the frailty level with scores 1–9 (Rockwood et al., 2005); MMSE the Mini-Mental State Examination testing the cognitive function with scores 0–30 (Folstein et al., 1975; Tombaugh and McIntyre, 1992); Barthel I the Barthel Index of activities of daily living with scores 0–100 (Mahoney and Barthel, 1965); SPPB the short physical performance battery with scores 0–12 (Guralnik et al., 1994); Assistive device (aid) used in daily life (note: most of these devices were only used for longer distances outside the house, no assistive devices were used in the study and * these assistive devices are only used occasionally; number of falls in the last year; FES-I the falls efficacy scale international, evaluating the fear of falling with scores 7–28 (Hauer et al., 2010). For each scale it is indicated if more impaired is a higher or a lower value. For brevity a Nordic walking stick is referred to as a "stick".*

Definition: The participant is *dynamically balanced* if they are not statically balanced but eventually become statically balanced without falling.

An STS transfer contains a mixture of static and dynamic balance: the movement begins and ends in a state of static balance and between participants may be dynamically balanced. Accordingly we have defined measures to let us determine if a participant is statically balanced, and if not, to analyze their dynamic balance. The static balance measures quantify the conditions that must be met for an object to be in a state of static balance. The conditions necessary to analyze dynamic balance cannot be measured directly but can be interpreted using a model. As STS transfers include appreciable amounts of both linear and angular momentum, we make use of the FPE (Millard et al., 2012) because this model takes both of these components into account. The following text will briefly cover mathematical details of the static and dynamic balance metrics used in this analysis. For further details see (Wight et al., 2007; Millard et al., 2009, 2012).

We will use a short-hand notation to refer to the various points of interest: the COM is represented by point C, the $\mathrm{COM_{GP}}$ by point G, the COP by point P, the point of BOS polygon that is nearest to a hypothetical point A is B(A). A position vector (r) from point G to P is given by ${}_{\mathrm{G}}r_{\mathrm{P}}$ (${}_{\mathrm{from}}r_{\mathrm{to}}$) while a position vector from the inertial frame to the COM is given by r_{C} (the inertial frame is omitted from the subscript labeling). Velocity vectors follow the same subscript convention with v used for linear velocity and ω used for angular velocity. A signed scalar is given by d while strictly positive scalars (such as speeds and distances) are indicated by applying the Euclidean norm operator ($||\cdot||_2$) to the vector of interest.

To see if a balance metric has values that are consistent with static balance we compared the results to the range spanned during still standing. To this end, we analyzed 10.0 s of quiet standing data in a sub-set of the participants (Y1, Y2, Y6, O1, O2, O6, O7, O9) and calculated the maximum and minimum values of all metrics, averaged across subjects, observed during the still standing trials. For signed metrics we use the larger of the two bounds to result in a symmetric region.

2.3.1. Base of Support

We define the functional BOS as the convex polygon that contains the area in which a participant can place at least half their body weight while keeping their feet flat on the ground. This polygon defines the area which the COP must stay in during the STS experiments if the participant is to complete the movement without rolling the foot excessively or taking a compensatory step. To establish the functional BOS we measured the foot movements and ground forces of two younger adult participants while they purposely moved their COP over the full functional range without taking a step. Measurements were made with shoes (one in light hiking shoes the other in Espadrilles) while standing on one foot, two feet, and also during STS trials. This data was used to create a normalized functional BOS polygon (**Figure 1A**) that is scaled to match each participant's left and right foot. Since both feet remain in contact with the ground throughout the STS movement we have defined the BOS as the convex hull of the ground projection of the left and right foot BOS polygons. Unless otherwise indicated we will refer to the functional BOS as simply the BOS for brevity.

The template BOS polygon was created by resolving each participant's COP profiles into a foot-fixed frame and calculating the convex hull that surrounds all COP data points in the $\hat{x}-\hat{y}$ plane (**Figure 1B**). Only COP points in which the foot can be considered flat and that have a normal force greater than half of the participant's body weight are included. Using a $\mathrm{X}(\psi_X)-\mathrm{Y}'(\theta_Y)-\mathrm{Z}''(\Psi_Z)$ Euler-axis decomposition the foot is considered flat if $\mathrm{w}\sin\psi_X \leq 1.5$ cm and $\ell\sin\theta_Y \leq 1.5$ cm where w and ℓ are the width

$$\mathrm{w} = \hat{x}\cdot\left(\frac{1}{2}(r_{\mathrm{TAM}}-r_{\mathrm{FAL}})+\frac{1}{2}(r_{\mathrm{FM1}}-r_{\mathrm{FM5}})\right) \tag{1}$$

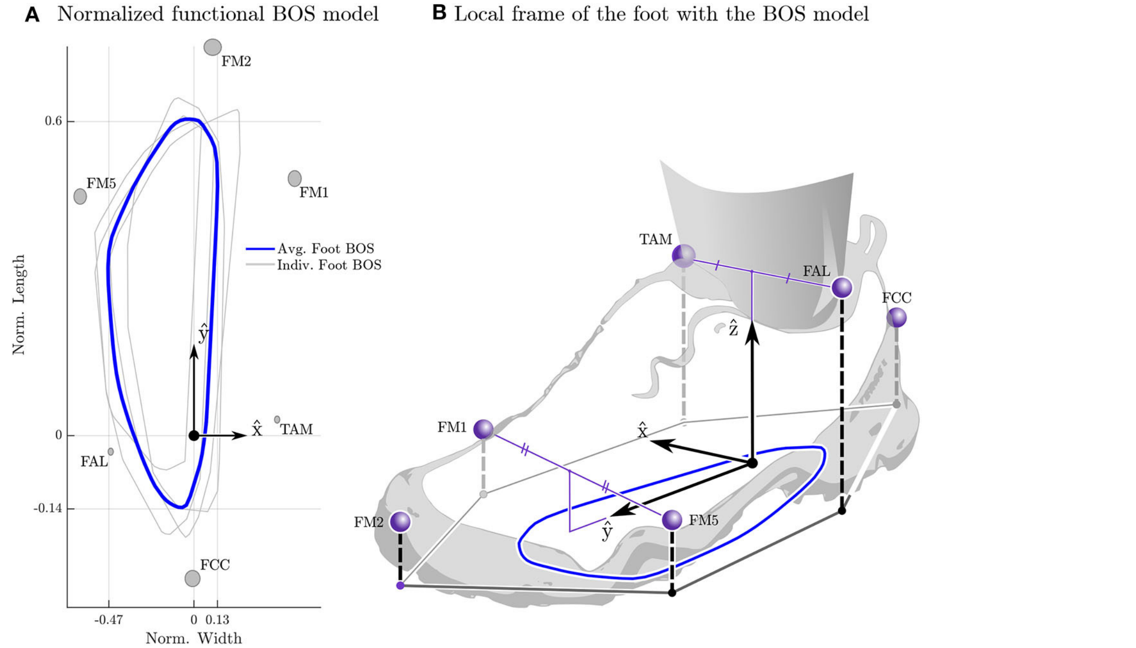

FIGURE 1 | Average normalized functional base-of-support (BOS) polygon. The four BOS polygons used to build the generic functional BOS template are shown in light gray (from the left and right feet of two participants) while the average BOS polygon is shown in blue **(A)**. The BOS from each participant's right foot has been reflected about the $\hat{y}$ and averaged with the left foot's BOS the to ensure that the final BOS polygon is symmetric. The gray ellipses are centered at the mean position for each marker resolved in the foot-fixed frame with the radius of the ellipse in the $\hat{x}$ and $\hat{y}$ set to the standard deviation of the marker position. The y coordinates have been normalized by the length of the foot ($\ell = \hat{y} \cdot (r_{\mathrm{FM2}}) - r_{\mathrm{FCC}})$) while the x coordinates has been normalized by the average width of the foot ($w = \hat{x} \cdot (\frac{1}{2}(r_{\mathrm{TAM}} - r_{\mathrm{FAL}}) + \frac{1}{2}(r_{\mathrm{FM1}} - r_{\mathrm{FM5}})))$. The foot-fixed frame **(B)** is constructed with the origin between the two ankle markers $\frac{1}{2}(r_{\mathrm{FAL}} + r_{\mathrm{TAM}})$, the $\hat{y}$ points from $\frac{1}{2}(r_{\mathrm{FAL}} + r_{\mathrm{TAM}})$ to $\frac{1}{2}(r_{\mathrm{FM1}} + r_{\mathrm{FM5}})$ while the $\hat{x}$ is the component of $(r_{\mathrm{TAM}} - r_{\mathrm{FAL}})$ that is perpendicular to $\hat{y}$, and the $\hat{z}$ is given by the cross product of $\hat{x}$ and $\hat{y}$. This frame is then transformed so that during quiet standing the origin is on the ground plane and $\hat{z}$ points upwards.

and length

$$\ell = \hat{y} \cdot (r_{\mathrm{FM2}} - r_{\mathrm{FCC}}) \tag{2}$$

of the participant's foot. We assume that together the foot pads and the sole of the shoe can compress by up to 1.5 cm on one side of the foot while the unloaded side is in contact with the ground. The estimate of 1.5 cm of compression comes from the fact that foot pads compress by $\sim$ 1 cm (Cavanagh, 1999; Gefen et al., 2001) during stance and the shoe sole likely compresses by 0.5 cm under body weight. The template is created by normalizing each foot's BOS by the length and width of the foot and then taking the average of the four profiles from the two subjects. The average is taken by choosing a point common to all polygons as the center [in this case $(-0.15, 0.3)$ in the left-foot normalized space], densely sampling the radius of each polygon across the common set of ray angles, and finally computing the average radius along each ray angle.

2.3.2. Static Balance

Here we make use of four easily measured conditions that must be true if a participant is statically balanced: the $\mathrm{COM_{GP}}$ is within the BOS; the $\mathrm{COM_{GP}}$ and COP are closely aligned; the linear velocity of the COM ($||v_{\mathrm{C}}||_2$) is small; and is the angular velocity ($||\omega_{avg}||_2$) is small. Note that we have to use relaxed conditions (using the term *closely* rather than *exactly*, and *small* rather than *zero*) to accommodate for modeling error and the fact that even still standing is accompanied by some movement. To see if a metric has values that are consistent with static balance we compare it to the range spanned during still standing.

To determine if the $\mathrm{COM_{GP}}$ is within the BOS (**Figure 2A**) we measure the distance between $\mathrm{COM_{GP}}$ and the nearest BOS edge (${}_{\mathrm{B(G)}}d_{\mathrm{G}}$). Since both feet remain in contact with the ground throughout STS, we defined the BOS as the convex-hull that enclosed the ground-projection of the scaled BOS polygons that are attached to each foot (**Figure 2B**). Note that a positive distance between a point and nearest edge of BOS indicates that the point is within the BOS (**Figure 2C**). To see how closely the $\mathrm{COM_{GP}}$ and the COP align we evaluate $||{}_{\mathrm{G}}r_{\mathrm{P}}||_2$ and see if the alignment is within the bound observed during still standing: since the mass distribution of the rigid body model of each participant is not perfect this distance will not go to zero even during still standing. The linear speed of the body is evaluated using $||v_{\mathrm{C}}||_2$ while the average angular speed of the body is evaluated using $||\omega_{avg}||_2$. Both $||v_{\mathrm{C}}||_2$ and $||\omega_{avg}||_2$ are compared to the bounds from our measurements of still standing.

The average angular velocity of the body ω_{avg} is evaluated by solving the linear system (Essén, 1993)

$$J_C\, \omega_{avg} = H_C \tag{3}$$

where J_C is the moment-of-inertia of the body about the COM, and H_C is the angular momentum of the body about the COM. The moment of inertia J_C is given by

$$J_C = \sum_{i=0}^{n} J_i + m_i(\mathrm{I}({}_C r_i^{\mathrm{T}}\, {}_C r_i) - ({}_C r_i\, {}_C r_i^{\mathrm{T}})), \tag{4}$$

where n is the number of bodies, J_i is the inertia matrix of the ith body at it's COM, m_i is the mass of the ith body, ${}_C r_i$ is the position vector from the COM of the entire system to the COM of the ith body, and all quantities are expressed in the inertial frame. The angular momentum vector is given by

$$H_C = \sum_{i=0}^{n} J_i \omega_i + {}_C r_i \times (m_i\, {}_C v_i), \tag{5}$$

where ω_i is the angular velocity of the ith body, $\times$ is the cross-product, ${}_C v_i$ is the velocity vector from the COM of the whole system to the COM of the ith body, and all quantities are expressed in the inertial frame. Note that the angular velocity of the entire system is termed the average angular velocity because Equation (3) averages across the angular momentum contributions of each of the i bodies in the same way that v_C averages across all of the velocity contributions of each segment.

2.3.3. Dynamic Balance

Since the body may move rapidly during an STS transfer we analyzed each participant's dynamic balance throughout STS by making use of the FPE (Wight et al., 2007; Millard et al., 2009, 2012). The FPE is evaluated by first computing the state of an equivalent single-body representation of the participant (**Figure 3**), projecting this state onto the vertical $\hat{s}$-$\hat{z}$ plane, and finally by applying the planar FPE (Wight et al., 2007) to this projected state. Here we use the FPE to measure four quantities related to dynamic balance: the error incurred when projecting the body's state onto the $\hat{s}$-$\hat{z}$, the size of the dynamic balance margin; the distance between the COP and the FPE in $\hat{t}$ (which causes turning); the distance between the COP and the FPE in $\hat{s}$ (which accelerates the body forwards and backwards).

The FPE is most accurate in the special case when the state of the body can be projected onto the $\hat{s}$-$\hat{z}$ plane (**Figure 3**) without loss of information. The $\hat{s}$-$\hat{z}$ is a vertical plane with its origin at COM_{GP}. The direction vectors are derived using the whole-body angular momentum vector H_G. The direction vector $\hat{t}$ (**Figure 3**) is in the direction of horizontal component of H_G, and the direction vector $\hat{s}$ is defined using the cross-product of $\hat{t}$ with $\hat{z}$: if H_G has no component in $\hat{z}$ it can be projected onto the $\hat{s}$-$\hat{z}$ plane without loss of information. The vector H_G will have no component in the $\hat{z}$ if $\omega_{avg} \cdot \hat{z} = 0$. Though this ideal has been satisfied to small tolerances when applied to walking and jumping (Millard et al., 2009, 2012), it is not clear if this condition will be met during STS. Therefore, we examined the vertical component of the average angular velocity vector: if $\omega_{avg} \cdot \hat{z} \leq \epsilon$, where ϵ is a bound defined using still standing data, it means the movements of the participant are consistent with the assumption used to extend the planar FPE model to 3D. It is important to note that the condition that $\omega_{avg} \cdot \hat{z} = 0$ must be met exactly by the simple model otherwise the trajectory of the pendulum will exit the $\hat{s} - \hat{z}$ plane and the pendulum's motion will not terminate in a quasi-stable standing pose (**Figure 4C**). In contrast, when the FPE is applied to a multibody system, such as a human, small values of $\omega_{avg} \cdot \hat{z}$ can be tolerated since the extra degrees-of-freedom of the body can be used to compensate for the trajectory errors introduced by non-zero values of $\omega_{avg} \cdot \hat{z}$.

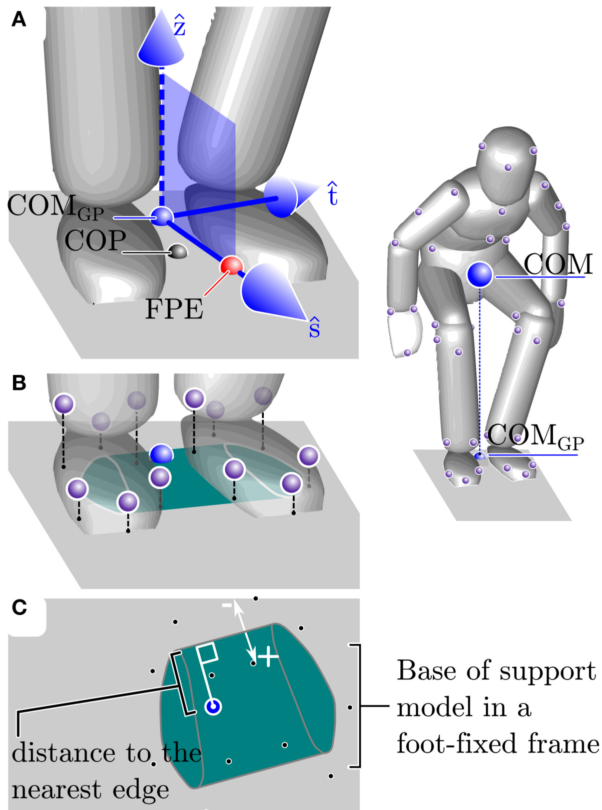

FIGURE 2 | Explanation of the balance metrics. The balance metrics make use of the center-of-mass (COM), COM ground projection (COM_{GP}), the center-of-pressure (COP) location, the foot-placement-estimator (FPE) location, and the base-of-support (BOS). The FPE is a model-based method (see section 2.3.3 for details) that computes the location away from the COM_{GP} along the axis $\hat{s}$ where the COP should be placed so that the body will move into a statically balanced standing pose without any additional input **(A)**. The BOS is defined as the convex hull of the ground projection of each foot's BOS model. The BOS of each foot is defined by a polygon (see section 2.3 and **Figure 1** for details) that is scaled to fit the length and width of each participant's left and right foot and is located in a foot-fixed frame **(B)**. The location of the COM and FPE are measured relative to the nearest edge of the BOS where points within the BOS are given a positive distance and points outside of the BOS have a negative distance **(C)**.

The FPE is a location in the direction $\hat{s}$ away from r_G: steps that are shorter than the FPE will cause the model to fall forwards (**Figure 4A**), steps that are longer will cause it to fall backwards (**Figure 4B**), steps that are exactly on the FPE will allow the model to come to rest passively as the COM passes over the COP (**Figure 4C**). Since steps taken in $\hat{s}$ will preserve the direction of

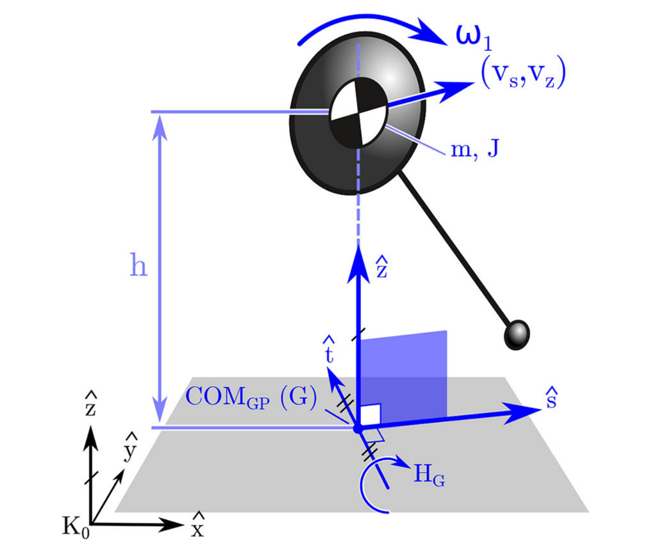

FIGURE 3 | Model used to compute the FPE in 3D. The FPE is applied in 3D by first mapping the state of the participant's body to an equivalent state for a single rigid body. Next the frame centered on G is formed (described in section 2.3) and the single body state is projected onto the $\hat{s} - \hat{z}$ frame. Finally the planar FPE location is evaluated (Wight et al., 2007) and placed along $\hat{s}$ away from G.

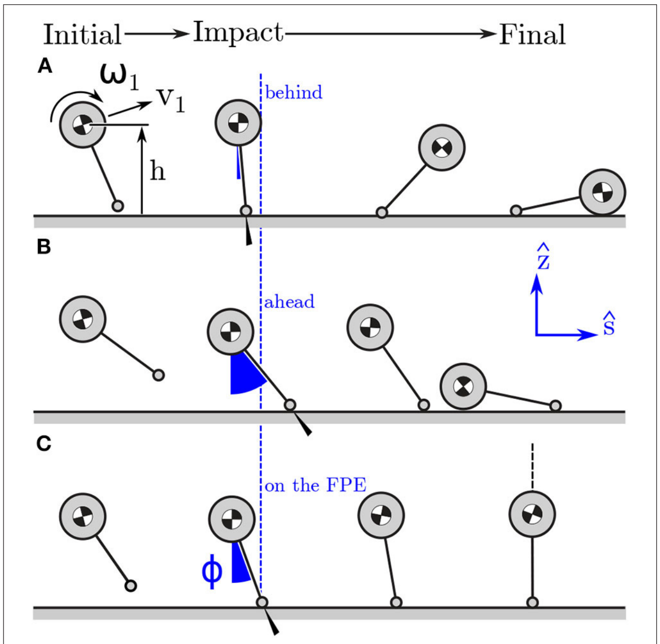

FIGURE 4 | Model stepping behind **(A)**, ahead, **(B)** and on **(C)** the FPE in the $\hat{s} - \hat{z}$ plane. Stepping behind the FPE location will mean that the post-contact kinetic energy of the model cannot be exchanged for potential energy: the model falls forwards. In contrast, if the model has the same initial conditions but steps far ahead of the FPE the model will not have enough post-contact kinetic energy to rotate passively over its foot: the model falls backwards. Stepping on the FPE location will ensure that the post-contact kinetic energy of the model exactly equals the potential energy the model will gain when it is standing over its contact point: in this case the model transitions to a quasi-stable standing position.

$H_{G,1}$ to $H_{G,2}$ this axis is named the straight-step direction. In contrast, steps taken in $\hat{t}$ will change the direction of $H_{G,1}$ to $H_{G,2}$ and cause a turn, and so we name $\hat{t}$ the turning direction. What follows is a derivation of the planar FPE (Wight et al., 2007) so that the dynamic balance metrics computed using the FPE are clear. We begin by assuming that momentum is conserved about the model's contact point

$$H_{G,1} = H_{G,2} \tag{6}$$

before (indicated using $_1$) and after (indicated using $_2$) the contact is made. This expression can be expanded (using $s\phi$ for $\sin\phi$ and $c\phi$ for $\cos\phi$ for brevity) by making use of the assumption that prior to contact the linear ($v_{s,1} = v_C \cdot \hat{s}$ and $v_{z,1} = v_C \cdot \hat{z}$) and angular velocity ($\omega_1 = \omega_{avg} \cdot \hat{t}$) of the model are uncoupled but after contact the model moves in a pure rotation (ω_2) about the contact point on a leg of fixed length ℓ

$$m\ell(v_{s,1}c\phi + v_{z,1}s\phi) + J\,\omega_1 = (m\ell^2 + J)\omega_2 \tag{7}$$

where

$$J = \hat{t}^{\mathrm{T}} J_C\, \hat{t} \tag{8}$$

is the moment of inertia of the COM about $\hat{t}$. For a candidate ϕ it is assumed that the leg length is constant *after* contact, but before contact it is long enough to reach the ground

$$\ell = \frac{h}{c\phi} \tag{9}$$

where $h = r_C \cdot \hat{z}$ is the height of the COM. It is worth noting that the leg length is the distance between the COM and COP which may not correspond to the physical leg length of a human or robot. After substituting Equation (9) into (7) we can isolate ω_2

$$\omega_2 = \frac{mh(v_{s,1}c\phi + v_{z,1}s\phi)c\phi + J\omega_1 c^2\phi}{mh^2 + Jc^2\phi}. \tag{10}$$

The FPE is defined as the contact location such that the model comes to rest just as its COM_{GP} comes into alignment with its COP. This means that the kinetic energy of the model just after contact must be exactly equal to the potential energy the model will gain as its COM moves from an initial height of h to ℓ as it rotates forwards. Since the model is in a pure rotation after contact we arrive at the following energy balance that must hold if ϕ points to the FPE location

$$\frac{1}{2}(J + m\ell^2)\omega_2^2 = mg(\ell - h). \tag{11}$$

Substituting Equation (10) into Equation (11) and simplifying yields a non-linear function

$$f = \frac{(mh(v_{s,1}c\phi + v_{z,1}s\phi)c\phi + J\omega_1 c^2\phi)^2}{mh^2 + Jc^2\phi} + 2mgh\, c\phi(c\phi - 1) \tag{12}$$

that evaluates to 0 when ϕ is at the angle that dynamically balances the model. In practice the value of ϕ that satisfies the constraint $f = 0$ is solved numerically using the bisection method to get close to the solution and Newton's method to polish ϕ to high precision. Trigonometry is then used to find the vector from the inertial frame to point F

$$r_{\mathrm{F}} = r_{\mathrm{G}} + (h \tan \phi)\hat{s}. \tag{13}$$

To apply the FPE to an STS transfer, this set of calculations is repeated for every recorded time sample. Since the single-body equivalent state of the participant and the FPE is re-evaluated at every time sample the assumptions of the method (J is constant, ℓ is constant, and the sum of kinetic and potential energy is constant) introduce small amounts of error in ϕ (at most 1.47° for the younger adults and 3.58° for the older adults, see **Supplementary Material** for details).

The difference between the model's point contact and the comparatively large BOS provided by a typical human foot affect how the FPE is interpreted. The closest physical analog to the model's point contact is the COP location of the BOS: in both cases moments (ignoring spin friction) taken about this point vanish to zero. Thus, if the FPE is within the BOS the participant can remain dynamically balanced by matching the COP location with the FPE. In contrast, if the FPE is outside of the BOS a participant will have to take a physical step that captures the FPE within the BOS in order to eventually transition to a state of static balance. For STS transfers participants keep their feet fixed on the ground throughout the movement and so we can define the *dynamic balance margin* as the displacement between the nearest BOS edge and the FPE (${}_{\mathrm{B(F)}}d_{\mathrm{F}}$). As before a positive sign indicates that point F is within the BOS.

Even though participants are restricted from moving their feet, they can still modulate the location of their COP relative to the FPE: modulating the COP location along the $\hat{t}$ axis will cause the model to rotate about the $\hat{s}$ axis and turn; varying the COP location in $\hat{s}$ will cause the model to fall forwards or backwards (as illustrated in **Figure 4**). Since turning is not a part of the task, we checked the displacement from the COP and the FPE location in $\hat{t}$ (${}_{\mathrm{P}}r_{\mathrm{F}} \cdot \hat{t}$) to see how well participants are able to maintain the direction of travel. To see if participants are modulating their COP in $\hat{s}$, either to help propel them out of the stool or to slow down, we also measure the displacement from the COP to the FPE location in $\hat{s}$ (${}_{\mathrm{P}}r_{\mathrm{F}} \cdot \hat{s}$).

2.3.4. Balance Metrics

In summary, we calculated the following static and dynamic balance metrics:

- Static balance metrics: (1) displacement from the BOS edge to the $\mathrm{COM_{GP}}$ (${}_{\mathrm{B(G)}}d_{\mathrm{G}}$); (2) distance between $\mathrm{COM_{GP}}$ and COP ($||{}_{\mathrm{G}}r_{\mathrm{P}}||$); (3) COM speed ($||v_{\mathrm{C}}||_2$); (4) whole-body average angular speed $||\omega_{avg}||_2$
- Dynamic balance metrics: (1) angular velocity about the vertical axis ($\omega_{avg} \cdot \hat{z}$); (2) displacement from the BOS to the FPE (${}_{\mathrm{B(F)}}d_{\mathrm{F}}$) i.e. the dynamic balance margin; (3) distance from the COP to the FPE location in turning direction (${}_{\mathrm{P}}r_{\mathrm{F}} \cdot \hat{t}$); (4) distance from the COP to the FPE in straight-step direction (${}_{\mathrm{P}}r_{\mathrm{F}} \cdot \hat{s}$).

This set of static and dynamic balance measures allows us to create a rich picture of how each of the participants coordinate their body and the forces acting on it to execute an STS transfer. We expect that participants who are frail, or afraid, will execute STS transfers while staying statically balanced or nearly so. We expect to see that participants who struggle with balance will allow the FPE location to approach the edge of their BOS, while participants who balance effectively can keep their FPE within the BOS and further from the edges. Since the STS transfer does not involve spinning or turning we expect to see all participants maintain small values of $\omega_{avg} \cdot \hat{z}$ and keep their COP as close as possible to the FPE in $\hat{t}$. We expect that older adults who are confident in their movements but struggle to get out of the stool may bias their COP behind of the FPE in the $\hat{s}$ direction to help propel them forwards out of the stool. Finally, we expect the older adults to display larger variation than the younger adults in all of the metrics.

2.4. Statistics

The 8 balance metrics taken at seat-off, generally considered the most unstable moment during STS, was used for analysis (see **Supplementary Table 1** for more information on stance). In addition, we analyzed total STS duration and duration from seat-off to stance. To assess differences in performance between groups, values were averaged over STS repetitions per participant; for differences in variability we took the range over repetitions measured for each individual. As group size was limited, we used non-parametric tests. Primary analysis tested for differences between young and older participant groups in the Side condition using unpaired Wilcoxon rank sum test. Secondary analysis tested for differences between arm conditions Side and Chest using paired Wilcoxon signed rank tests per age group. Reported values represent median and interquartile ranges (from 25 to 75%). Significance was set at $p < 0.05$ and statistical analyses were performed in Matlab (version 2019a, Natick, MA, USA).

3. RESULTS

The four static balance measures all indicated that the older adults stay closer to being statically balanced than the younger participants between seat-off and standing (**Figure 5**). At seat-off, most of the older adults had their $\mathrm{COM_{GP}}$ 1.7 [4.7] cm inside the BOS and kept it there throughout the movement to stance, while nearly all of the younger participants began seat-off with their $\mathrm{COM_{GP}}$ outside of their BOS (−4.0 [3.2] cm, $p = 0.004$, **Figure 5A**). While both groups closely aligned their $\mathrm{COM_{GP}}$ and COP at standing, the older adults began seat-off with a smaller distance of 3.6 [2.0] cm compared to 7.6 [3.6] cm for the younger participants ($p = 0.006$, **Figure 5B**). The maximum speed of the COM of the older participants tended to be lower throughout the movement than their younger counterparts, with 32.1 [5.2] compared to 39.9 [9.6] cm/s at seat-off ($p = 0.068$, **Figure 5C**). There was no significant difference between older and younger adults in the total STS duration (1.9 [0.9] vs. 1.7 [0.2] s, $p = 0.315$) but there might be a trend of older adults requiring more time to move from seat-off to standing (1.0 [0.3] vs. 0.8 [0.1] s, $p = 0.122$). Both older and younger adults had significant angular speeds during STS (**Figure 5D**), with values as high as 56.2 [28.0] and 50.0 [24.8] °/s at seat-off ($p = 0.829$) and lower speeds over the entire movement (18.9 [11.5] and 12.0 [10.2] °/s). There was no detectable difference in within-subject variability (as defined in section 2.4) between the older and younger adults in the duration of the movement nor in any of the four static balance measures ($p = 0.12$–0.95 see **Supplementary Table 2**).

The older adults appeared to control their balance similarly well as the younger adults, while maintaining a larger dynamic balance margin (**Figure 6B**). Both groups almost maintained low angular velocities about the vertical axis (**Figure 6A**), with values of $\omega_{avg} \cdot \hat{z}$ as small as 1.62 [6.2] and −0.95 [4.6] °/s at seat-off ($p = 0.315$) and −2.3 [18.5] and −0.035 [7.2] °/s on average throughout the movement. This indicates that for nearly all participants the FPE should be accurate, except for participant O6 who moved with values of $\omega_{avg} \cdot \hat{z}$ that were much larger than other participants at seat-off (26.5 [56.6] °/s) and throughout the movement (−17.1 [54.9] °/s). Both young and older adults kept the FPE well within the BOS between seat-off and standing, with the older adults maintaining larger dynamic balance margins than younger adults (7.9 [1.8] vs. 5.7 [1.4] cm $p = 0.006$) at seat-off (**Figure 6B**). In addition, both groups maintained the direction of travel, having a distance between FPE and COP in the $\hat{t}$ direction close to zero with a narrow spread: 0.6 [0.8] cm for the older vs. 0.7 [0.5] cm for the younger adults ($p = 0.696$), **Figure 6C**). In the $\hat{s}$ direction, some of the older adults showed a preference for beginning seat-off with a larger distance between the FPE and COP than the younger participants (4.2 [3.4] vs. 2.0 [1.2] cm, $p = 0.055$, **Figure 6D**) presumably as a strategy to help propel them out of a seated position. No differences were found in variability between the older and younger adults for any of the dynamic balance measures ($p = 0.38$–0.97 see **Supplementary Table 2**).

Several static and dynamic balance metrics were affected by the arm conditions, although the effect of age was dominant and preserved between conditions (**Figure 7**). While both total duration and duration from seat-off to stance were not affected by the position of arms in either group ($p > 0.195$), both groups were further from being statically stable at seat-off with arms crossed at their chest compared with arms at the side (**Figure 7**). Specifically, both groups kept their COM_{GP} further from the center of the BOS ($p < 0.016$), as well as from the COP ($p < 0.008$), but they did have lower COM speed ($p < 0.016$). In addition, both groups tended to decrease their dynamic balance margin by placing their FPE closer to the BOS edge (Y: $p = 0.002$, O: $p = 0.109$), as well as closer to the COP in the $\hat{s}$ direction ($p < 0.002$), thus reducing the forward propelling strategy. As these differences between arm conditions were in the same direction for both age groups, the differences in balance control between age groups are dominant over the arm effect. The effect of age at the chest condition was 1.5–3.0 times as large for the static balance variables that were affected by arms condition and 1.2 times for the FPE to BOS (**Figure 7**). Only the effect of arms condition on the FPE-COP excursions in $\hat{s}$ was larger compared with the age effect (0.7 times, see **Supplementary Material**).

4. DISCUSSION

As STS performance degrades, many older adults suffer injuries due to falls (Rapp et al., 2012; Pozaic et al., 2016; van Schooten et al., 2017). Previous investigations into STS balance have examined quantities that do not directly measure the conditions necessary for balance but instead quantities that correlate with performance: including STS duration, COM kinematics and COP to ankle position at seat-off (Moxley Scarborough et al., 1999; Åberg et al., 2010; Akram and McIlroy, 2011; Fujimoto and Chou, 2014). We began our balance assessment by calculating how close each participant is to being statically balanced during their movement: the displacement between COM_{GP} and BOS, the distance between COM_{GP} and COP, the speed of the COM, and the average angular speed of the whole body at seat-off. In addition, we assessed dynamic balance by applying the FPE, allowing us to define an STS as being balanced if the FPE remained within the BOS throughout STS. In addition, we analyzed the distance between FPE and COP in both straight step and turning directions.

Although the FPE allows us to take both linear and angular momentum into account, it comes at the cost of more involved modeling and mathematics. Therefore, we analyzed the necessity and applicability of using the FPE by evaluating the angular speed as well as the assumption of small angular velocity about the vertical axis. Regarding the necessity, we found that all participants moved with a large whole-body angular speed, especially at seat-off. Clearly it is important to use a balance metric, like the FPE, that takes angular velocity into account. Regarding the applicability, we found that the assumption of small $\omega_{avg} \cdot \hat{z}$ was well met by nearly all of the participants except Y5, O6, and O9. This is in contrast to previous work in which the FPE was used to study walking motions Millard et al. (2009); Bruijn et al. (2013) where the condition $\omega_{avg} \cdot \hat{z} = 0$ is satisfied to a small tolerance. Though the values reported in **Figure 6** may seem large, neither Y5 nor O9 visibly struggled with balance during the STS trials presumably because they could compensate for a non-zero $\omega_{avg} \cdot \hat{z}$ after seat-off. In contrast, O6 did visibly struggle with balance during the STS trials. Due to O6's unique pathology (hemiplegia) in our participant group we re-ran the analysis excluding O6 and found no changes to our results. It is striking that the participant who most visibly struggled with balance was also the only participant with large and highly variable values of $\omega_{avg} \cdot \hat{z}$. In all other respects the static and dynamic balance metrics of O6 (**Figures 5**, **6**) are unremarkable: it is as if O6 has retained all faculties to balance except to regulate $\omega_{avg} \cdot \hat{z}$ to small values. A failure to control $\omega_{avg} \cdot \hat{z}$ has consequences beyond the applicability of the FPE: it will make it difficult to control the direction of travel which has a clear impact on maintaining balance. In the future it will be valuable to study $\omega_{avg} \cdot \hat{z}$ in more detail so that it is clear what range is associated with typical movements and what ranges might be indicative of a balance pathology. Further it will be important to determine if others who struggle with balance also exhibit large variations in $\omega_{avg} \cdot \hat{z}$ or if this is a problem specific to people (such as O6) who have an asymmetric pathology like hemiplegia.

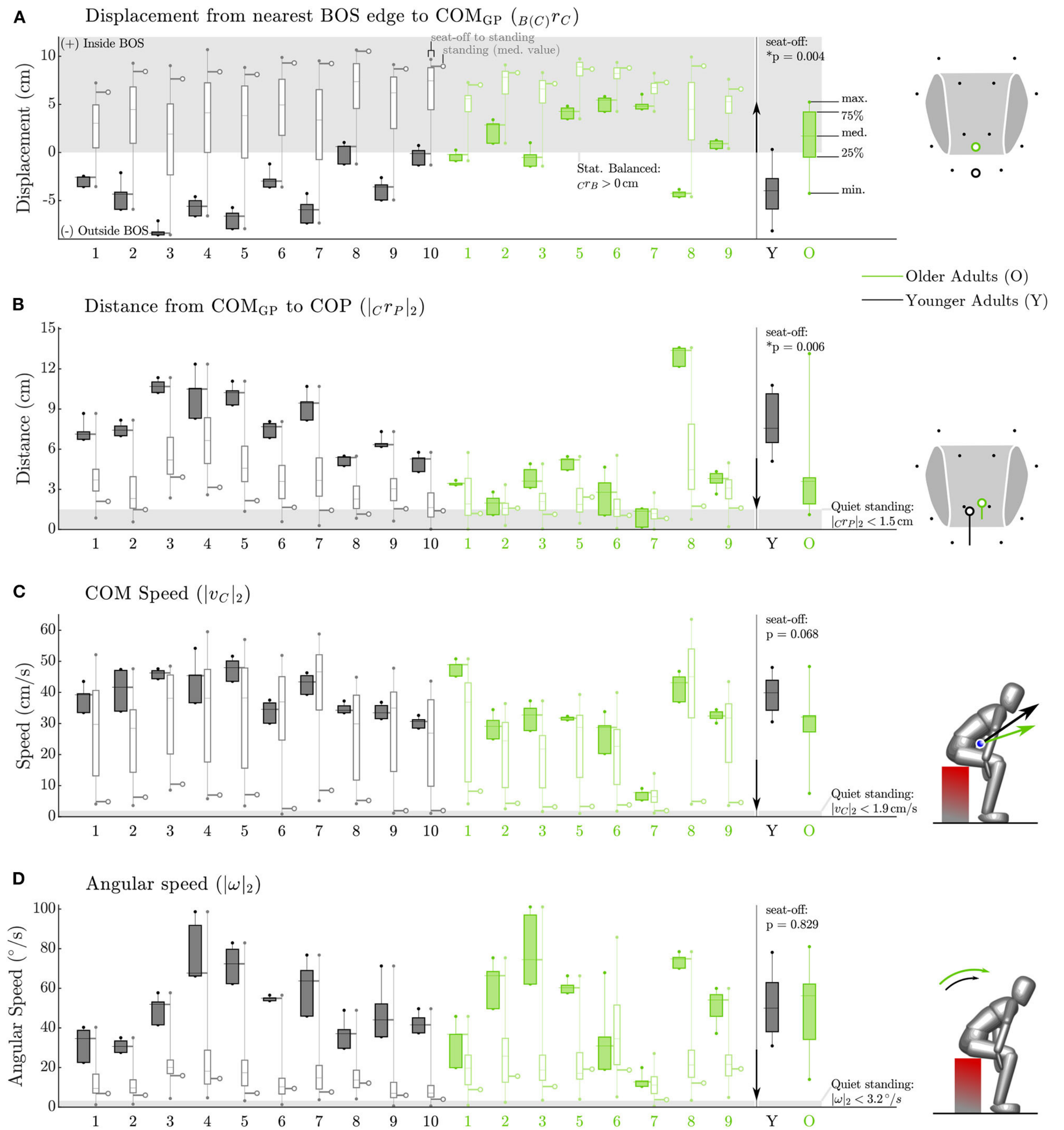

FIGURE 5 | Static balance metrics for the Side condition. Data for each participant appears on the left while group data at seat-off appears on the right. The older participants (green) keep their COM$_{GP}$ within their BOS from seat-off to standing while nearly all of the younger participants (black) begin seat-off with their COM$_{GP}$ outside of the BOS **(A)**. The older adults keep their COP closer to their COM$_{GP}$ than do younger adults **(B)**. The older adults move their COM more slowly between seat-off and standing than do younger adults **(C)**. The average angular speed of both the older and younger participants is quite high at seat-off **(D)**. The gray box indicates the static balance region in **(A)** and values measured during quiet standing in subplots **(B–D)**. The arrow between the individual and group data indicates the direction toward more conservative static balance.

Our approach of analyzing both static and dynamic balance yielded a surprise: although the older adults stay closer to being statically balanced than the younger adults, supporting our first hypothesis, both groups maintained similar dynamic balance margins, refuting our second hypothesis of reduced dynamic balance in older adults. In addition, there was no indication

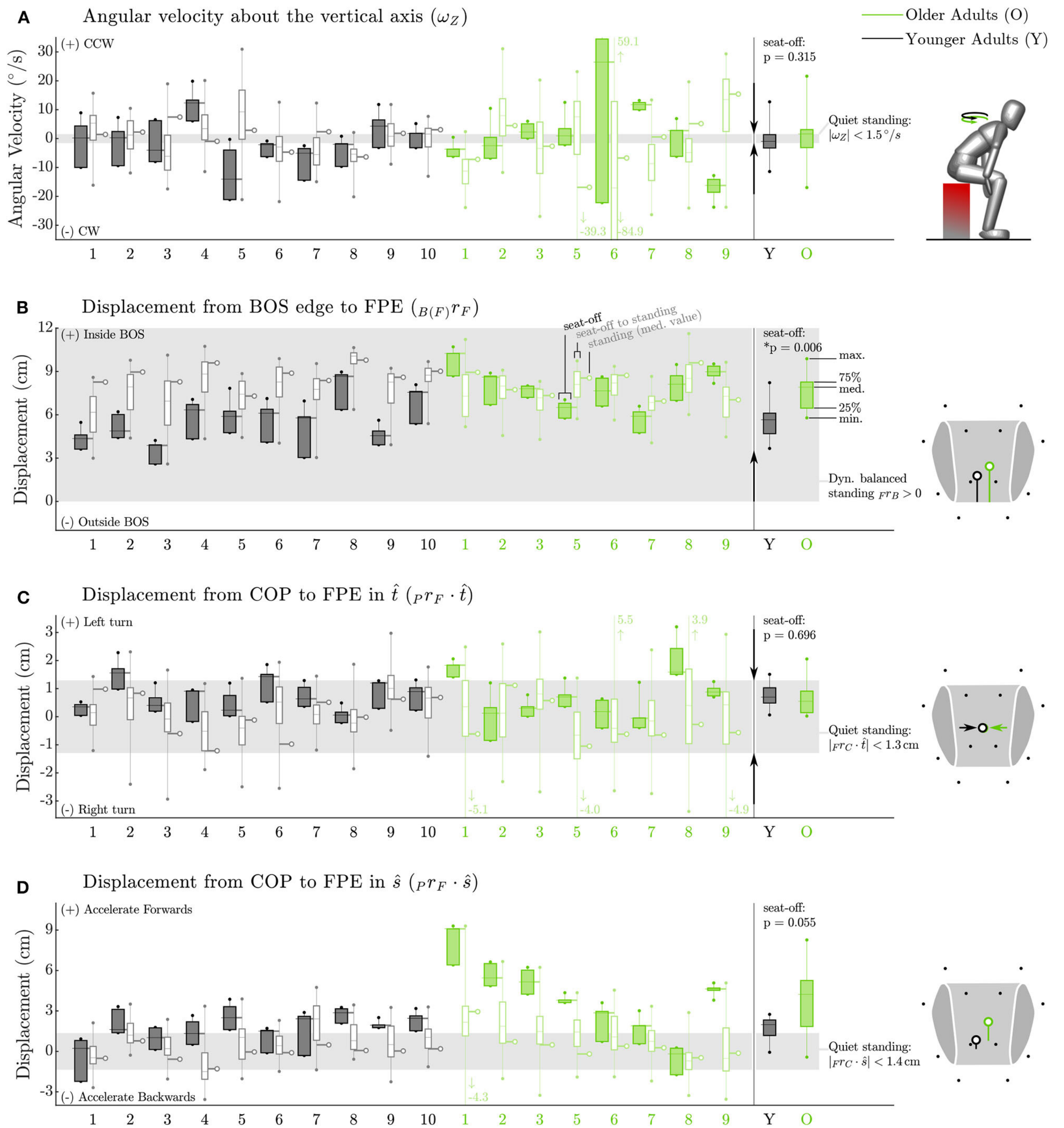

FIGURE 6 | Dynamic balance metrics for the Side condition. Data for each participant appears on the left while group data at seat-off appears on the right. The FPE method assumes a small angular velocity about the vertical axis. While this assumption is reasonable for most participants it is not appropriate for all participants, particularly O6 **(A)**. Both groups (older adults in green, younger adults in black) keep the FPE similarly well within the BOS (shown in gray) between seat-off and standing and **(B)**. Both groups display a tight control of the COP relative to the FPE location in the turning direction $\hat{t}$ **(C)** indicating that neither group is turning on average between seat-off and standing. Some of the older adults bias COP behind the FPE at seat-off **(D)** presumably as a strategy to help propel them off of the stool. The arrows between the individual and group data indicate the direction toward a state of more conservative dynamic balance. No arrow appears in the plot of $_Pr_F \cdot \hat{s}$ **(D)** as deviations in $\hat{s}$ are permitted (and thus indicate individual preference) provided $_{B(F)}d_F$ is positive.

that older adults display more variability in the execution of STS refuting our final hypothesis. As such, the results of the static balance analysis generally echo what has been previously found: older adults are slower, keep their COM_{GP} more anterior, and have an increased rising duration compared to younger participants. It has been reported that older adults place their

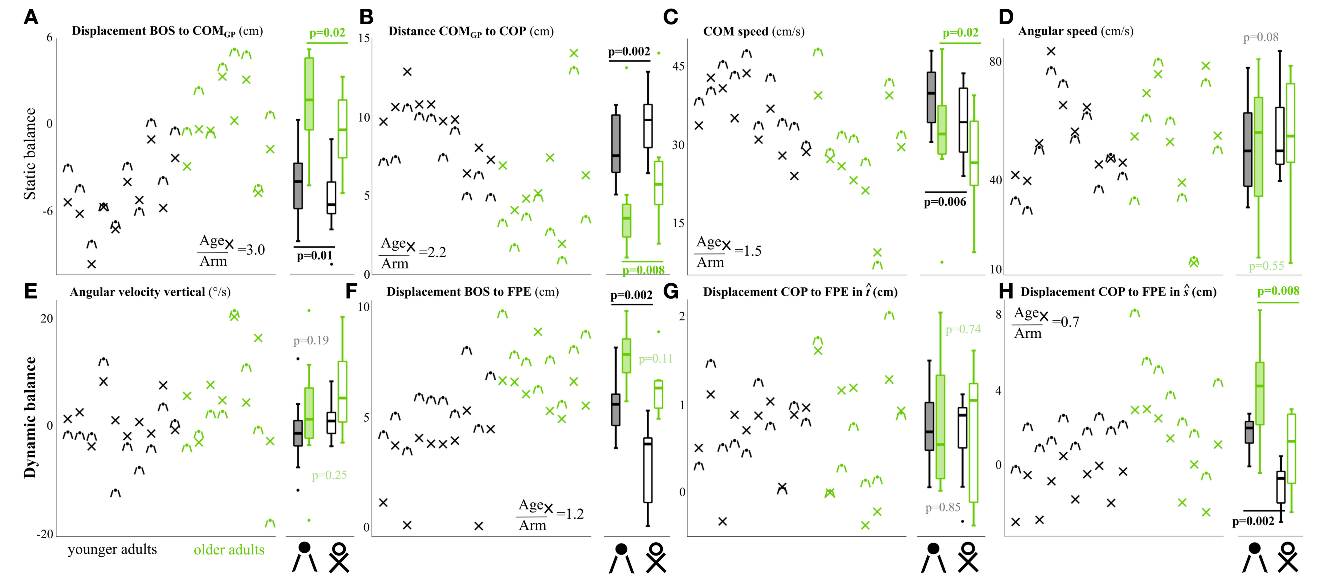

FIGURE 7 | Comparison between Side and Chest conditions for all static **(A–D)**, and dynamic **(E–H)** balance metrics. Individual median data points are shown for younger (black) and older adults (green). Conditions are shown with x-mark for Chest condition and dot (median value) with hanging side arms for Side condition. Although most variables are affected by arm condition, the differences in balance control between the groups are preserved between conditions. *P*-values from the paired non-parametric sign-rank tests are given comparing effect of condition per group, in bold when significant ($p < 0.05$). In addition, the ratio of age effect for the Chest condition and arms condition $(O^C - Y^C)/((O^C + Y^C) - (O^S + Y^S))$ (where *O* and *Y* stand for older and younger, and *C* and *S* stands for Chest and Side) is indicated for the affected variables. Box plots indicate the median, 25th and 75th percentiles of the group data, and the whiskers the extreme values that are not considered outliers (plotted as dots).

COM and COP further forward at seat-off than younger adults, and more so in those with previous falls, impairments or STS difficulty (Schultz et al., 1992; Aissaoui and Dansereau, 1999; Papa and Cappozzo, 2000; Chen et al., 2013; Fujimoto and Chou, 2014). In addition, older adults exhibited lower trunk or body speed, especially those with higher frailty level, fear of falling or failed STS attempts (Riley et al., 1997; Kouta and Shinkoda, 2008; Åberg et al., 2010; Ganea et al., 2011), although compensatory increases in trunk flexion have been reported as well (Papa and Cappozzo, 2000). In accordance with the lower body speed, older adults tended to require more time for the most demanding STS phase, to move from seat-off to stance, aligning with the general notion that they are slower to rise especially when more frail or impaired (Ganea et al., 2011). However, our data shows that while older adults execute STS more slowly and conservative as noted in literature, they are dynamically balancing as well as the younger participants with similar levels of variability between repetitions. Thus, the changes in performance with age do not seem to reflect impaired balance control, but are likely a compensatory mechanism for reduced physical ability or reduced confidence.

Both the static and dynamic balance analysis depend on knowing the geometry of the BOS. To make our analysis as accurate as possible we have developed a BOS model of a shod foot. Since the BOS polygon of the foot has been fitted to data of two younger adults (**Figure 1**) it may not accurately represent every participant. It is important to note that the definition of the BOS does affect some of our results. Using an alternative BOS model (the convex hull of the ground projection of the motion capture markers attached to the feet) the differences between younger and older adults in ${}_{B(F)}d_F$ (**Figure 6B**) are no longer significant ($p = 0.633$), though both groups still maintain positive dynamic balance margins. Though other numerical results change, no other statistical differences are affected by the change in BOS model. We have chosen to present the results obtained using the functional BOS model because this model should be more accurate in principle than the simpler model. In addition the range spanned by ${}_{B(F)}d_F$ between the 25th–75th percentiles is smaller using the functional BOS in comparison to the simpler alternative model: [1.8] vs. [3.8] for the older adults and [1.4] vs. [1.8] for the younger adults. The reduction in the span from the 25th–75th percentile indicates that the BOS model is removing some of the systematic variations in ${}_{B(F)}d_F$ that participants are making to accommodate for the size and shape of their feet. In the future we hope to make the same detailed measurements of the BOS across a larger range of participants to see how the functional BOS varies from person to person.

This study introduced a combination of clustering and adaptive thresholding that allowed us to identify different transitions in the movement, including seat-off and standing. As some of our older adults are quite variable between STS movements, including some sit-back failures, a single threshold did not work well within, let alone between, participants. Although the segmentation algorithm is a little elaborate the alternative has drawbacks: manually identifying these events would have been subjective and not reproducible. We expect similar results would have been obtained had we manually segmented every movement as our analysis focused at the moment of seat-off which can be accurately identified using the force plate data. We also confirmed using a few of test cases that manually identifying the transition to standing yields similar results to the automatic segmentation routine. In contrast, consistently identifying the time of STS initiation and standing manually would be challenging. We hope others find this approach useful to segment movement data while adapting to each participant's characteristics.

In addition, we have also found that the while the arm conditions of the STS transfer affect the movement, many similar observations were made in the Chest compared to the more natural Side condition. Despite this similarity, the differences between the arm conditions are large enough that we feel that STS should be assessed using conditions that are as natural as possible: otherwise trends might be observed which are due to the unnatural lack of arm motion rather than an underlying condition. It should be noted that other conditions might be even more natural to people, such as providing support using the legs or arm rests. The effect of different types of assistance on the STS movement and especially balance will be part of future research.

The main limitation of this study is the relatively small size of the groups that are included. It is possible that differences in dynamic balance control could therefore not be detected, but we were able to detect differences in static balance control between older and younger participants. In addition, we were not able to include older adults with known STS difficulties; however, the clinical metrics show that we have a rather heterogeneous sample, ranging from those who are managing well to those with reduced physical ability. This was reflected by some older adults showing observable difficulty with getting up, resulting in a few failed attempts, as is represented by lower SPPB values. Regardless, as some older adults performed in the range as younger adults, this could have masked differences in dynamic balance and variability that might be characteristics for more frail older adults. As such, our future research will be directed to gathering data in more frail older adults, including those who are (more) dependent on assistance, and also focus on failed STS attempts.

We have made several important contributions in this work: we have analyzed the conditions for static and dynamic balance during STS by applying the FPE for the first time to STS, and by using a geometric model of the BOS. While few studies have used a point of convenience, such as the ankle, as a reference point before (Schultz et al., 1992; Moxley Scarborough et al., 1999; Papa and Cappozzo, 2000; Fujimoto and Chou, 2014), the conditions for even static balance cannot be evaluated without a BOS model. Our work fills a void in the literature since existing studies have not analyzed all of the quantities necessary to assess balance directly but instead have focus on a few isolated metrics: STS duration, COM speed, trunk movement, COM-ankle distance, COP-ankle distance (Moxley Scarborough et al., 1999; Akram and McIlroy, 2011; Jeyasurya et al., 2013; Fujimoto and Chou, 2014). Together, our approach allowed us to show that quantities measured in existing literature that differ between younger and older adults (Aissaoui and Dansereau, 1999; Moxley Scarborough et al., 1999; Janssen et al., 2002; Åberg et al., 2010; Akram and McIlroy, 2011; Fujimoto and Chou, 2014; Millor et al., 2014; Boukadida et al., 2015) may not actually pertain to reduced balance control in the latter group. As is evidenced in our data, it is possible to execute an STS, slowly or quickly, with a lot of COP movement or a little, all while displaying a fine control of dynamic balance.

This work provides an important but first step in exploring balance during STS in older adults. As with most metrics that have suggested to quantify balance during movements, further validation to demonstrate how these static and dynamic balance metrics actually relate to falls is needed. Therefore, this analysis should be repeated in older adults that are prone to falling, have a fear of falling, are more frail or have different impairments. More specifically, it would be interesting to contrast the static and dynamic balance values for successful versus failed STS attempts, including sit-back, side-step, and step-forward attempts, and simulated falls. To simulate falls, an important and open question has yet to be answered: how do older adults actually fall during sit-to-stand and stand-to-sit.

5. CONCLUSION

In this work we have shown that while older adults execute STS more slowly and stay closer to being statically balanced than younger adults, they are dynamically balancing as well as the younger participants with similar levels of variability. Our analysis of static and dynamic balance indicates that the reason for this difference is not due to a reduced sense of balance. Thus, the presented approach of using the model-based dynamic balance metric FPE as well as expressing metrics relative to individual's BOS, allows us to distinguish between STS movement (such as duration and COM speed) and balance. Future research is needed to see how the patterns of static and dynamic balance change between balanced and unbalanced motion, and between people who are prone to falling from those who move safely.

DATA AVAILABILITY STATEMENT

The data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the IRB of the medical faculty of Heidelberg University. The participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

LS, CW, and KM designed the experiment. LS and CW collected the data and recruited the participants. MM and LS processed and analyzed the data, interpreted the data, wrote the manuscript, and generated the figures and tables. All authors provided the critical feedback on the manuscript.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2020.548174/full#supplementary-material

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REVIEW
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The Influence of Virtual Reality Head-Mounted Displays on Balance Outcomes and Training Paradigms: A Systematic Review

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Background: Falls are the leading causes of (non)fatal injuries in older adults. Recent research has developed interventions that aim to improve balance in older adults using virtual reality (VR).

Purpose: We aimed to investigate the validity, reliability, safety, feasibility, and efficacy of head mounted display (HMD) systems for assessing and training balance in older adults.

Methods: We searched EBSCOhost, Scopus, Web of Science, and PubMed databases until 1 September 2020 to find studies that used HMD systems for assessing or training balance. The methodological quality was assessed using a modified version of Downs and Black. We also appraised the risk of bias using Risk of Bias Assessment tool for Non-randomized Studies (RoBANS).

Results: A total of 19 articles (637 participants) were included for review. Despite heterogenous age ranges and clinical conditions across studies, VR HMD systems were valid to assess balance and could be useful for fall prevention and for improving postural control and gait patterns. These systems also have the capacity to differentiate healthy and balance-impaired individuals. During VR versions of traditional balance tests, older adults generally acquire a cautious behavior and take more time to complete the tasks.

Conclusion: VR HMD systems can offer ecologically valid scenarios to assess and train functional balance and can be used alone or in addition to other interventions. New norms and protocols should be defined according to participants' age, health status, and severity of their illness when using VR HMD systems for balance assessment and training. For safe and feasible training, attention must be given to display type, VR elements and scenarios, duration of exposure, and system usability. Due to high risk of bias and overall poor quality of the studies, further research is needed on the effectiveness of HMD VR training in older adults.

Keywords: head-mounted display (HMD), posture (MeSH), vestibular, somatosensory, visual, older adults, gender, balance

INTRODUCTION

Maintaining balance while standing and during gait is paramount because stability plays a crucial role in human locomotion (Wodarski et al., 2019). Balance is mostly achieved and maintained by sensorimotor control systems that include sensory inputs from vision, proprioceptive, and vestibular systems (Manchester et al., 1989). Vision uses the information projected on the retina to guide the relationship between the environment and the body. The proprioceptive sensation provides information on the position and joint movements. The vestibular system combines gravity and acceleration inputs to collect information on the position and movement of the head. Changes in any sensory source elicit alterations in postural control. Of the three systems, vision is often considered as the most important factor for maintaining balance during quiet standing and activity (Horiuchi et al., 2017). Vision consists of the optical flow and visual field (central and peripheral). Optical flow deals with the perception of self-motion, and provides information about heading direction (Gibson, 1950). In the peripheral visual field, optical flow also contributes to a better postural sway stabilization (Horiuchi et al., 2017).

Falls affect around one-third of older adults and often put their independent functionality at risk (Al-Aama, 2011). Balance assessment should be an integral part of any fall prevention program to establish the baseline levels and monitor progression. Several methods are being used to assess the risk of falls, including surveys, physical tests, and perturbation-based measurements. Surveys are useful to measure the external risk factors but should be complemented with physical tests for measuring internal risk factors. Questionnaires could also be influenced by the participants' age, sex, and motivation, as well as cognitive and emotional status (Yardley and Redfern, 2001). Physical test batteries are easy to administer but may lack ecological validity and the ability to isolate specific sensory impairments (Saldana et al., 2017). For example, sensory organization test (SOT) is considered as the gold standard in estimating sensory contributions to balance control, but is unable to diagnose clinical disorders, such as vestibular hypofunction, because the postural sway is not a good indicator of underlying pathology (Lubetzky et al., 2018). Other systems that evaluate the isolated role of vision on standing posture have also their limitations. For example, Prism glasses alter a view on sagittal and horizontal planes, but not on the coronal plane (Ohmura et al., 2017). Balance training interventions include various static and dynamic routines for overcoming daily tasks in older adults (Halvarsson et al., 2014). However, such interventions may produce mixed success results as their exercises may not target the neuromuscular skills required for preventing falls (Parijat et al., 2015). The rehabilitation programs are also time-consuming and often lack adherence and compliance (Meldrum et al., 2015). Alternative approaches are needed to customize the interventions to the participants' needs, to objectively measure performance, and to boost participants' motivation while improving the adherence and compliance (Avola et al., 2019).

With the advancement of technology, many researchers are exploring the use of active video games (exergames) and virtual reality (VR) as assessment and rehabilitation tools. These systems use computer interfaces to immerse participants in virtual worlds. VR allows investigating functional balance performance without the risks associated with the previous methodologies (Oddsson et al., 2007). Sensory domes and foams have been used to provide head-fixed visual references as well as inaccurate somatosensory feedback. Tilting floors and moving walls are also used to offer sensory destabilization in the sagittal plane resulting in increased sway in the anteroposterior direction (Alahmari et al., 2014). Under rehabilitation purposes, many older adults show higher interests in playing exergames as compared to real-world activities (Wollersheim et al., 2010). The feasibility and concept of VR systems are linked to the simulation, interaction, and immersion of the existing technologies (Soltani, 2019). While earlier exergame systems were using TV screens to project the game in front of the players, newer systems employ portable head-mounted displays (HMD) with larger fields of view (FOV) and stereoscopic visual fields, that are updated continuously using head position and rotation. By adding depth perception and by blocking external visual information, these systems may offer acceptable, feasible, and ecologically valid results in addition to the previous assessment methodologies.

In the early 2000s, Takahashi and Murata (2001) suggested that VR experience using HMD systems may cause symptoms of motion sickness and poor equilibrium. Since HMD systems block the eyesight, the users may lose their balance by hitting physical objects. The immersive virtual environments (VE) may also cause conflicts between proprioceptive and vestibular sensory systems. This can also reduce users' abilities to regain stability after balance loss (Keshavarz et al., 2015). The illusory sensation of body movement induced by virtual scenarios, such as riding a VR rollercoaster, may also cause loss of balance. Participants might try to accommodate these illusory self-movements with physical body movements, which can cause them to lose their balance. Therefore, it is important to understand how the type of display affects motor behavior, especially if the system is being used as training and rehabilitative tools for older adults. In this systematic review, our goal was to investigate the validity, reliability, safety, feasibility, and efficacy of HMD systems to assess and train balance in older adults.

METHODS

This systematic review was conducted according to the Preferred Reporting Items for Systematic Reviews and Meta-Analysis (PRISMA) guidelines (Moher et al., 2009). There is no registered protocol for this review.

Eligibility Criteria

The eligibility criteria were formulated by using a modified patient population, intervention/indicator, comparator, outcome, and study design (PICOS) framework (Santos et al., 2007). The population comprised older adults of 50+ years of age, and the intervention/indicator was the assessment or training of balance using HMD, which was compared with traditional assessment or other training programs. By using exergames (e.g., Nintendo Wii or Xbox Kinect), participants can still receive contextual information from their surrounding and adjust their balance accordingly. Therefore, articles that used other VR technologies, such as exergames, for presenting the VR scenarios were excluded. The outcomes that we considered were balance control or the ability to maintain balance. We considered full-text articles with different designs that investigated balance assessment or training. While we considered conference papers (if full article was available) for inclusion, those that provided only the abstract (conference abstract) were excluded. We also excluded review articles.

Search Strategy

The EBSCOhost, Scopus, Web of Science, and PubMed electronic databases were searched for articles and conference abstracts that focused on the use of HMD to assess and train balance and were published between 1990 until 1 September 2020. The following search terms and Boolean operators were used: (HMD OR head mount* display OR virtual reality OR artificial environment OR simulated 3D environment OR simulated three-dimensional environment) AND (balance OR posture OR fall) AND (vestibular OR visual OR somatosensory OR context*). **Appendix 1** details the search strategy used. The search was restricted to English peer-reviewed articles. Each author independently screened the titles and abstracts of the identified records for relevance. The full text of the articles that appeared relevant and those without obvious relevance (after inspecting the title and abstract) were retrieved. The full-text records were read, and their eligibility was screened with the criteria mentioned above. The reference lists of all eligible studies were hand-searched to identify additional relevant studies. Any disagreement was resolved by mutual discussion.

Data Extraction

One author (PS) extracted all data from each article into a piloted form. The data extracted comprised the type, number, age, and sex of the participants, HMD type, VR scenario, protocol, main outcomes, and the actual HMD effect. Due to the heterogeneity on the population characteristics, HMD interventions, and outcome assessment tools, a meta-analysis was not pursued.

Methodological Quality and Risk of Bias

The methodological quality of each article was assessed using a custom quality assessment questionnaire adapted from Downs and Black (1998) and Campos et al. (2011). These questionnaires provide scores for study quality, and internal and external validity, in randomized and non-randomized studies. The results from these questionnaires provide insights on the completeness of the reporting which helps future studies on commonly overlooked methodological steps, and how future studies should comprehensively report their methodology. Each article was evaluated on 13 questions that considered the reporting parameters, internal and external validity, and study power (**Table 1**). Each question was answered by "Yes," "No," or "Unable to determine." As this review represents a qualitative summary of using HMD in assessing and training balance, no total score was computed for quality assessment. The risk of bias was appraised using the Risk of Bias Assessment tool for Non-randomized Studies (RoBANS; Kim et al., 2013). The RoBANS tool can be applied to different types of studies and eliminates the need to use several tools for each type of study. We opted to use solely the RoBANS tool to display all the risk of bias judgements in a single tool. The RoBANS tool includes the following domains: selection of the participants (selection bias), confounding variables (selection bias), measurement of exposure (performance bias), blinding of outcome assessments (detection bias), incomplete data outcome (attrition bias) and selective outcome reporting (reporting bias). For intervention studies, we also considered two additional domains to account for bias arising from the interventions, which included planning and implementation of interventions (performance bias) and deviation from intended interventions (performance bias). Each domain was judged as unclear, low risk, or high risk of bias. Any disagreements were discussed until consensus.

TABLE 1 | Quality assessment questionnaire.

	Questions
Q1	Is the hypothesis/objective/aim of the study clearly described?
Q2	Are the main outcomes to be measured clearly described in the Introduction or Methods?
Q3	Are the characteristics of the participants clearly described?
Q4	Are the inclusion/exclusion criteria described and appropriate?
Q5	Are the main findings of the study clearly described?
Q6	Are estimates of the random variability in the data for the main outcomes provided?
Q7	Have actual probability values been reported for the main outcomes?
Q8	Are the participants representative of the entire population from which they were recruited?
Q9	Are the setting and conditions typical for the population represented by the participants?
Q10	Are retrospective unplanned analyses avoided?
Q11	Are the statistical tests used to assess the main outcomes appropriate?
Q12	Are the main outcome measures used accurate (valid and reliable)?
Q13	Is a sample size justification, power description, or variance and effect estimates provided?

RESULTS

Search Results

A total of 1,694 articles were found in the initial search. After duplicated and non-English removal, 385 articles were screened for titles and abstracts. After full text review, a total of 19 articles evaluating the use of VR HMD in balance assessment and training in older adults were included (**Figure 1**).

Methodological Quality and Risk of Bias Assessment

The results of the quality assessment are mentioned in **Figure 2**. Most of the studies properly described their aims, main outcomes, participants' characteristics, inclusion/exclusion

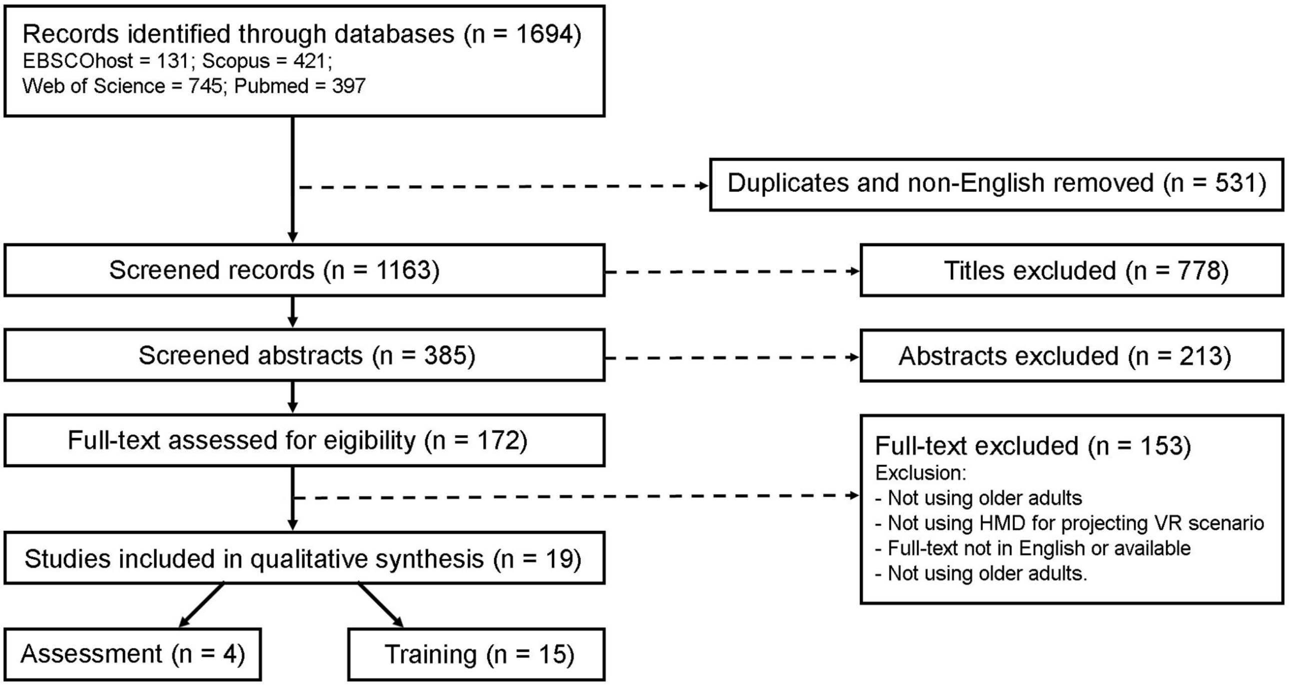

FIGURE 1 | Flowchart of the obtained results, unfolding the search strategy.

criteria, and main findings (Q1 to Q5, respectively). For reporting the information (Q6 and Q7), most of the studies mentioned that their data were either normally distributed or provided standard error, standard deviation, confidence interval, and actual probability values. External validity was measured using Q8 and Q9. Since most of the studies were only using older adults without comparing them with a younger control group, the distribution of the main confounding factors was not the same in the study sample. Finally, internal validity was evaluated using Q10 to Q13. Very few studies were based on a predefined hypothesis or provided justification for their sample size, power description, variance, and effect estimates.

The judgement of each domain for each study, as well as the summary of each domain can be seen in **Figure 3**. Note that in the summary of domain risk of bias there is a high percentage (∼20%) of "not applicable," which is related to the non-intervention studies; this high percentage is due to the higher sample size of these studies that translated into greater weight in the summary plot. All studies showed high risk of detection bias because none of the studies blinded the outcome assessor. While selection bias due to uncontrolled confounding variables was a major concern (74% of studies), selection bias due to selection of participants had low risk of bias all but one study. Attrition and reporting bias were also not common, with only one study judged as high risk of bias attrition bias and none for selective reporting. Performance bias due to measurement of exposure was judged as high risk in 16% of the studies. From the 15 studies that included HMD as an intervention, only two studies were judged to having high risk of performance bias due to planning and implementation of interventions or deviations from intended interventions.

Participants Characteristics

The included studies presented a wide variety of health conditions and age ranges (**Table 2**). Eleven studies involved healthy older adults (58%; $n = 230$), four studies included individuals with vestibular disorder (21%; $n = 193$), four studies included individuals with Parkinson's disease (PD; 21%; $n = 67$), one study included participants with mild cognitive impairment (MCI; 5%; $n = 12$), two studies involved participants with stroke (10%; $n = 8$), and five studies included balance impaired and at risk of fall individuals (26%; $n = 127$).

HMD and VR Protocols

Most of the included studies used the Oculus Rift (DK 1 and 2) as HMD tool (33%). Other devices included HTC Vive (19%), Balance Rehabilitation Unit (BRU; 33%), and other systems (eMagin Z800 3D Visior, Sony Glasstron, Virtual Research VR8, Solo Myvu, Samsung Gear VR, and Revelation 3D; 43%). Twenty-nine percent of the studies did not report the name of the HMD. The VR scenarios were heterogenous across studies. Seven studies (37%) used VEs with adjustable visual perturbances while four studies (21%) used stationary visual scenes. Two studies (11%) used simple avatar VE and one study (5%) used (semi)realistic VEs. In 17 studies (89%), participants had to interact with VR to complete the tasks.

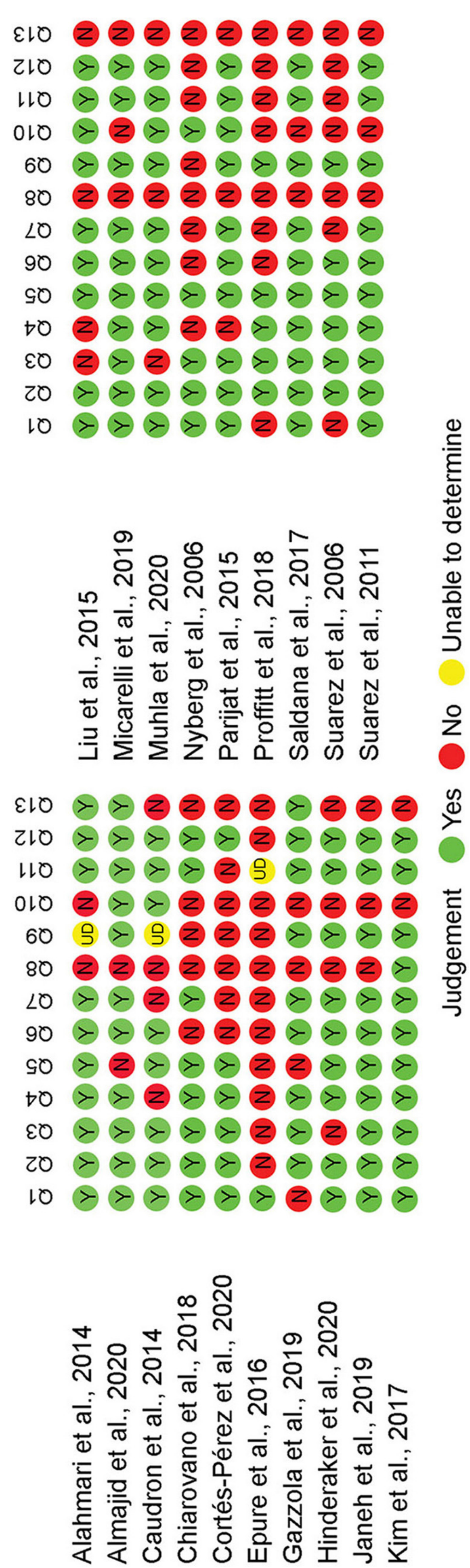

FIGURE 2 | Quality assessment of the included studies.

Outcome Assessment

Sixteen studies (84%) used validated physical test batteries that utilized force plates, plantar pressure, posturography, physiological profile assessment, and human machine interaction. Two studies (11%) used specific perturbation-based protocols either by pulling and pushing, or by including visual perturbations inside the VR scenario. Six studies (32%) used questionnaires including Motion Sickness Susceptibility Questionnaire (MSSQ), Simulator Sickness Questionnaire (SSQ), Falls Efficacy Scale International (FES-I), and Activity-specific Balance Confidence (ABC). Four studies (19%) used surveys including gaming habits, Dynamic Gait Index, Berg Balance Scale (BBS), Tinetti Scale, Subjective Visual Vertical (SVV) test, Romberg, and Motion Sickness Rating. Actual HMD effect varied between studies. Studies that included healthy older adults, measured changes in gait patterns, postural stability, balance, fear of falling, body sway, simulator sickness, and balance strategies. Studies comprising patients with pathological conditions, evaluated both improvement and deterioration of balance and its related metrics.

Can VR Systems Using HMD Be Used for Assessing Balance?

The studies were heterogenous in terms of HMD features, VR scenarios, and comparison outcomes. Therefore, a direct comparison was not possible. Below, we report the results of HMD systems to assess balance in regard to validity, reliability, and consistency of their measurements with gold standard methodologies (**Tables 3**, **4**).

Validity and Reliability of HMD Systems for Assessing Balance

Only one study evaluated the validity and reliability of HMD systems for assessing balance. Alahmari et al. (2014) measured test-retest reliability, concurrent validity, and construct validity of their HMD by comparing it to the center of pressure (CoP) measurement in healthy adults and individuals with vestibular disorders. The CoP area and velocity in anteroposterior and mediolateral directions were correlated with the measurements of SOT, indicating concurrent validity with a standard clinical test for measuring sensory organization abilities in healthy and vestibular patients.

Validity and Performance of HMD Comparing to Traditional Balance Tests

Two studies compared the validity and performance of HMD to a traditional balance test—the Timed Up and Go (TUG) test. These two studies showed that adding the HMD VR component to the TUG, the task was more challenging to complete and that the self-motion perceptions during the task changed. Muhla et al. (2020) contextualized the traditional TUG to make it closer to the real-world scenarios. They observed an increase in the number of steps and the time to complete the VR TUG test. Almajid et al. (2020) added a motor task and visual stimulus to the TUG test and evaluated the effects of age-related visual dependence on motor performance. They showed that while wearing HMD, older adults' egocentric self-motion perceptions decreases which

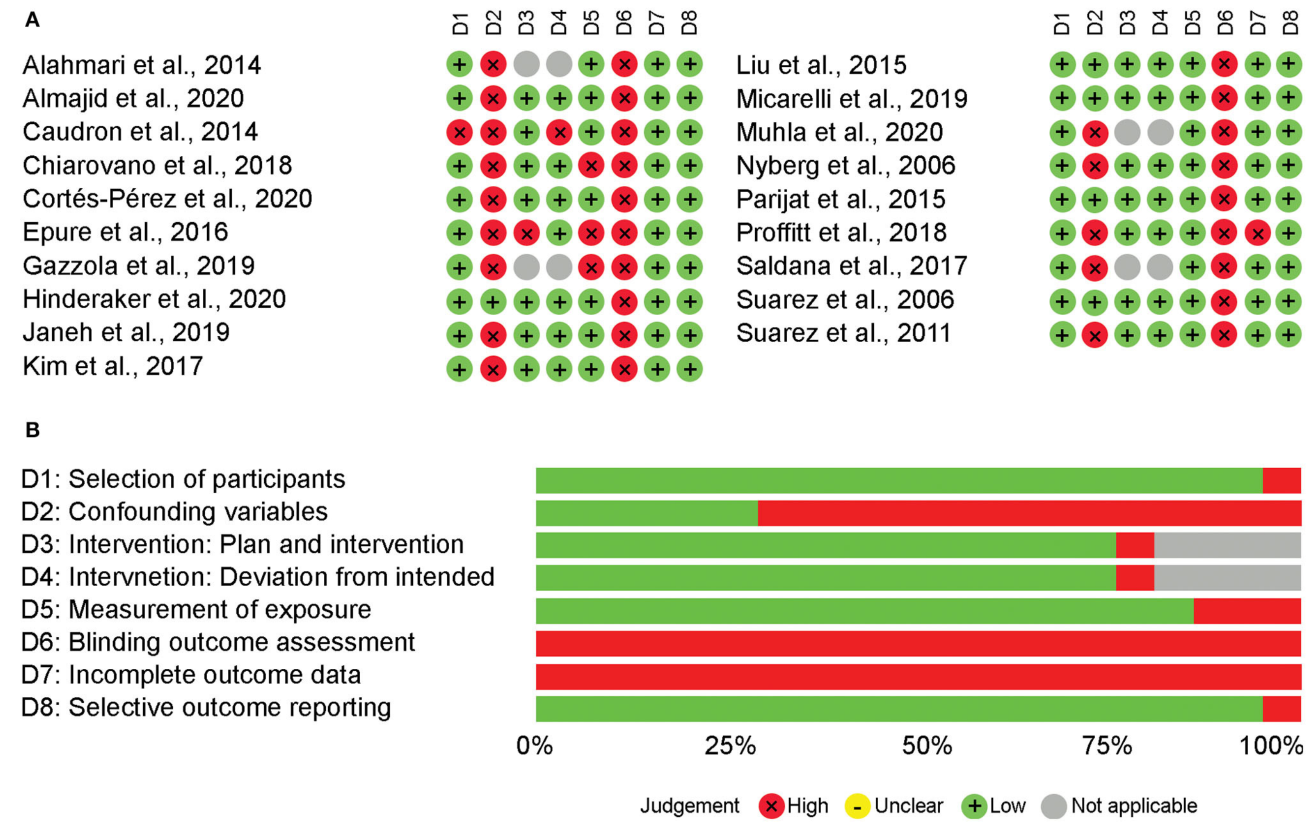

FIGURE 3 | Risk of bias appraisal using RoBANS. Risk of boas domain judgement for each study using RoBANS **(A)**. Summary appraisal of risk of bias domains using RoBANS **(B)**.

could negatively affect motor performance. Wearing HMD also changed biomechanical and perceptual constraints, limited peripheral vision, and created inaccurate subject-to-object and object-to-object distance estimation.

Capacity of HMD Systems in Differentiating Healthy and Balance-Impaired Individuals

Only one study evaluated the capacity of HMD systems to differentiate between healthy individuals and those with balance impairments. To this end, Saldana et al. (2017) assessed the validity and reliability of their VR HMD system by comparing the balance measurements with a force-plate. They showed that patients with risk of falls display faster anteroposterior velocity as compared to the healthy individuals. This finding was associated with increased odds of falling and thus capable of identifying those with increased risk of falls.

Effects of VR Elements and Scenarios

The effects of VR elements and scenarios was investigated in two studies comprising either healthy or balance-impaired individuals. Complexity of VE can also increase the time to complete the tasks in older adults (Nyberg et al., 2006). Similarly, walking patterns including walking speed and stride length, balance reactions, and slips could vary according to the equipment, sensory load, and the VR scenarios (Nyberg et al., 2006). Gazzola et al. (2019) measured the effects of visual, somatosensorial, and visual-vestibular manipulation on postural control in older adults. Vestibular patients with(out) history of falls had lower limit of stability, which is the area where their oscillation is safer. They showed that visual and somatosensory cues could compensate the inaccurate information of the vestibular system for the maintenance of body balance.

Usability of HMD Systems

Only two studies assessed the usability of HMD systems to measure balance. Saldana et al. (2017) measured the acceptability of their VR system using a simulator SSQ. They reported no significant changes in the nausea-subscale score and SSQ overall score after VR exposure. Kim et al. (2017) used longer bouts of walking in VR and showed that symptoms of simulator sickness were higher in patients with PD compared to healthy younger and older adults.

Can VR Systems Using HMD Be Used as a Balance Training Tool?

Safety and Feasibility

Several studies investigated the safety of HMD systems for training balance. Kim et al. (2017) showed that within their setup with simple cityscape and without any turns,

TABLE 2 | List of the studies with the data extracted from each article.

References	Health status or pathological condition	Number	Age	Sex	HMD sensor	VR scenario/Avatar	Protocol
Alahmari et al. (2014)	Healthy	30	77.2 ± 5.0	18F, 12M	BRU, N/R	Stationary visual scene (basketball gym), No avatar	CoP area and velocity, comparison with SOT
Almajid et al. (2020)	Healthy	16	69.0 ± 4.4	8F, 8M	Oculus Rift DK 2	Visual disturbance consisting of falling snowflakes, No avatar	BBS, TUG test with additional motor task (holding a cup)
Gazzola et al. (2019)	Healthy	41	72.51 ± 6.84	24F, 17M	BRU, eMagin Z800 3D Visior	Following random visual targets with different colors and letters, No avatar	CoP, LoS, VoS
Hinderaker et al. (2020)	Healthy	10	71.0 ± 5.0	6F, 4M	HTC Vive	Moving colored spherical particles, No avatar	CoP
Kim et al. (2017)	Healthy	11	66.0 ± 3.0	8F, 3M	Oculus Rift DK2	Walking in VE, No avatar	Mini-BESTest, CoP, SSQ, stress, and arousal
Liu et al. (2015)	Healthy	36	71.24 ± 6.82, 70.54 ± 6.63, 74.18 ± 5.82	18F, 18M	Sony Glasstron	VE with visual perturbations, No avatar	CoM velocity
Muhla et al. (2020)	Healthy	31	73.7 ± 9.0	20F, 11M	HTC Vive	Sit and get up task, No avatar	TUG
Nyberg et al. (2006)	Healthy	4	69.0-80.0	2F, 2M	Virtual Research VR8	Walking in VE, heavy snow vs. tilting world, graphic disturbance and flickering, No avatar	Disturbances of balance, walking patterns
Parijat et al. (2015)	Healthy	24	65.0+	12F, 12M	Sony Glasstron	Exploration scene with visual perturbation, No avatar	Gait
Saldana et al. (2017)	Healthy	8	81.4 ± 6.25	7F, 1M	Oculus Rift DK2	Standing activities with semi realistic graphics, No avatar	SSQ, postural sway
Suarez et al. (2011)	Healthy	19	62.3 ± 12.7	N/R	BRU, N/R	Static visual field with visual optokinetic stimulation, No avatar	LoS, CoP, BFR
Alahmari et al. (2014)	Vestibular	15	66.0 ± 8.0	10F, 5M	BRU, N/R	Stationary visual scene (basketball gym), No avatar	CoP area and velocity, comparison with SOT
Caudron et al. (2014)	Idiopathic PD	17	61.9 ± 8.2	7F, 10M	Solo, Myvu	Simplified avatar of participant's body using motion capture, Simplified avatar	Pull task with eyes open, eyes closed, or visual feedback
Chiarovano et al. (2018)	Vestibular impairment (neuritis), cervicogenic dizziness, general dizziness and imbalance	90	65.0 ± 15.0	51F, 39M	Samsung Gear VR	Stereoscopic visual scene, No avatar	Several visual conditions and unpredicted visual perturbation at several amplitudes of movement
Cortés-Pérez et al. (2020)	Acute stroke	3	45, 50, 53	3M	HTC Vive	Interaction with virtual objects while standing, With animal avatar	BBS and Tinetti scale, SVV, Romberg, gait, ABC, FES-I, and perception of verticality
Epure et al. (2016)	Balance impaired	6	59.0-69.0	N/R	N/R	Skiing game, semi realistic graphics, No avatar	Physical balance
Gazzola et al. (2019)	Chronic vestibular disorder	76	71.90 ± 5.23 and 73.92 ± 6.27	55F, 21M	BRU, eMagin Z800 3D Visior	Following random visual targets with different colors and letters, No avatar	CoP, LoS, VoS
Janeh et al. (2019)	PD	15	67.6 ± 7.0	15M	HTC Vive	Walking in VE, No avatar	SSQ, gait
Kim et al. (2017)	PD	11	65.0 ± 7.0	8F, 3M	Oculus Rift DK2	Walking in VE, No avatar	Mini-BESTest, CoP, SSQ, stress, and arousal
Micarelli et al. (2019)	UVH and MCI	12, 12	74.3 ± 4.7 and 72.5 ± 3.6	6F, 6M and 7F, 5M	Revelation 3D	Track speed racing 3D game, 110° FOV, No avatar	VOR, postural control
Proffitt et al. (2018)	Post-stroke patients	5	56.0 ± 3.0	2F, 3M	Oculus Rift DK2	Recycling game (sorting, filling, and loading), No avatar	Interview, usability

(Continued)

TABLE 2 | Continued

References	Health status or pathological condition	Number	Age	Sex	HMD sensor	VR scenario/Avatar	Protocol
Saldana et al. (2017)	At risk of fall	5	78.4 ± 9.37	3F, 2M	Oculus Rift DK2	Standing activities with semi realistic graphics, No avatar	SSQ, postural sway
Suarez et al. (2011)	Balance disorder	26	73.0-82.0	N/R	BRU, N/R	Virtual scene under different sensory conditions, No avatar	CE, SV
Suarez et al. (2011)	PD	24	66.5 ± 8.5	N/R	BRU, N/R	Static visual field with visual optokinetic stimulation, No avatar	LoS, CoP, BFR

ABC, Activities-specific balance confidence; BBS, Berg balance scale; BFR, Balance functional reserve; BRU, Balance rehabilitation unit; CE, Confidential ellipse; CoB, Center of balance; CoP, Center of pressure; EMG, Electromyography; FES-I, Falls efficacy scale international; FOV, Field of view; fps, Frames per second; HMD, Head-mounted display; LoS, Limit of stability; MCI, Mild cognitive impairment; Mini-BESTest, Mini-balance evaluations systems test; MSSQ, Motion sickness susceptibility questionnaire; N/R, Not reported; PD, Parkinson's disease; SOT, Smart equitest sensory organization test; SSQ, Simulator sickness questionnaire; SV, Sway velocity; SVV, Subjective visual vertical test; TUG, Timed up and go test; UVH, Unilateral vestibular hypofunction; VE, Virtual environment; VOR, Vestibul-ocular reflex; VoS, Velocity of oscillation; VR, Virtual reality.

doorways, or crossing thresholds, most of the patients with PD completed the tests without any discomfort. Proffitt et al. (2018) used their HMD system for telerehabilitation of a group of post-stroke patients. They showed that all patients required assistance for balance and fall prevention that limit the application of HMD systems as telerehabilitation interventions. The duration of VR exposure was an important factor that could affect the feasibility of balance training using HMD systems. Other studies have analyzed the balance performance of individuals with balance-related pathological conditions and explored which strategies can be employed to complete the VR tasks. Nyberg et al. (2006) studied fall tendencies of healthy individuals in functional VR settings, such as walks and turns, and showed that their subjects usually opted for more cautious strategies to prevent a fall incident.

Effects of the HMD Display Features and VR Scenarios

The HMD features have been shown to significantly affect the balance training outcomes. Epure et al. (2016) compared the effects of display type on physical balance in healthy and balance-impaired adults. They showed that both groups had higher stability when using monitor displays compared to HMD. Different VR scenarios yielded different outcomes. Janeh et al. (2019) used VR-based gait manipulation to improve gait symmetry in a group of patients with PD. Their VR task dissociated visual and proprioceptive inputs and increased the patients' step length, cadence, and swing time variabilities for both body sides. VR gaming systems with visual modifications also improved vestibulo-ocular reflex and postural control in individuals with vestibular disorders or mild cognitive impairments (Micarelli et al., 2019).

The way that the VR scenario is perceived (first vs. third-person view) can have an impact on the balance outcome, but the results were individual-dependent. While some post-stroke patients felt more engaged when playing in a first-person view, others felt more in control and at ease when playing in the third-person view (Proffitt et al., 2018). Changes of visual perception was found to be an important feature of HMDs and VR scenarios. Suarez et al. (2011) evaluated the effects of stable and moving visual fields on balance outcomes of patients with PD. They showed that changes in visual information increased the CoP area which triggered balance control. In another study, Suárez et al. (2006) evaluated postural adaptations after vestibular rehabilitation in two different perceptual conditions with static and dynamic visual fields, and found that visual fields with moving targets could elicit postural disturbances. Parijat et al. (2015) induced virtual perturbations similar to slip and observed evoked recovery reactions that could be transferred to the actual slip trials. Hinderaker et al. (2020) evaluated the effects of optical flow speed on brain activity and postural control of younger and older adults. They showed that older adults show higher brain activities at lower optical flow speeds as compared to younger adults, but that the optical flow speed did not affect the postural sway in either the younger or older adults.

TABLE 3 | Outcomes of the studies and the actual HMD effect in healthy individuals.

References	Balance assessment/training protocol	Outcomes	Actual HMD effect
Alahmari et al. (2014)	CoP area and velocity, comparison with SOT	Significant correlation between BRU and SOT (CoP area ICC = 0.64–0.81, velocity ICC = 0.44–0.76), higher CoP area, and velocity for older adults compared to younger adults.	Correlation with traditional test. Need to define new norms for different participants.
Almajid et al. (2020)	BBS, TUG	Non-statistically significant differences between younger and older adults in balance measures and cognitive function, lower BBS in older adults, visually independent older adults perform better that dependent older adults, Lower turning cadence, walking slower, decreased pitch, yaw, and roll peak trunk velocity in all TUG components. Smaller AP and ML acceleration ranges in sit-to-stand and smaller AP acceleration in stand-to-sit.	Wearing HMD negatively affects TUG.
Gazzola et al. (2019)	CoP, LoS, VoS	Higher LoS and lower CoP in control group compared to training groups.	Wearing HMD did not affect balance negatively.
Hinderaker et al. (2020)	CoP	Older adults had maximum amplitude compared to younger adults at the OF speed of 10 m/s, age could affect older adults' abilities to process OF stimulation.	Wearing HMD increased body sway.
Kim et al. (2017)	Force plate, SSQ	Decreased stress after training, lower arousal compared to PD patients (arousal absolute changes in healthy young vs. healthy old = 3 ± 2 vs. 4 ± 4).	Wearing HMD decreased stress, did not change simulator sickness, and static and dynamic balance.
Liu et al. (2015)	CoM velocity	Higher fall reduction and better forward trunk rotations reduction in VR group, better corrections to a slip in traditional methods, lack of efficacy as patients gain experience	Initial positive results but lower efficacy as they gain experience.
Muhla et al. (2020)	TUG	Higher completion times and steps in TUG VR.	Wearing HMD increased completion times and induced more cautious behavior.
Nyberg et al. (2006)	Kinematics, Survey, SSQ	Tilting virtual scene affected balance performance during walking (% increased time to complete task vs. no HMD = 166–283%).	Changes in balance with visual perturbations.
Parijat et al. (2015)	Gait	Increased ankle plantarflexion, knee flexion, and trunk flexion at heel contact in VR compared to overground walking. Early muscle activation was observed in VR compared to treadmill walking.	Wearing HMD induced balance strategies and cautious behavior.
Saldana et al. (2017)	Force plate, SSQ	Lower change of tilt in anteroposterior direction compared to patients with risk of fall (tilt at risk vs. healthy = 0.7°/s vs. 0.4°/s).	Similarity with traditional test. Need to define new norms.
Suarez et al. (2011)	LoS, CoP, BFR	Healthy had lower CoP values compared to the PD group in the static visual field, BFR was reduced significantly in sensory VR scenarios.	Wearing HMD increased risk of falls when viewing sensory VR scenarios.

ABC, Activities-specific balance confidence; AP, Anteroposterior; BBS, Berg balance scale; BFR, Balance functional reserve; BRU, Balance rehabilitation unit; CE, Confidential ellipse; CoP, Center of pressure; DGI, Dynamic gait index; DHI, Dizziness handicap inventory; FES-I, Falls efficacy scale international; FOV, Field of view; HMD, Head-mounted display; ICC, Intraclass correlation coefficient; ML, Mediolateral; MS, Multiple sclerosis; MSSQ, Motion sickness susceptibility questionnaire; OF, Optical flow; PD, Parkinson's disease; PPA, Physiological profile assessment; SOT, Smart equitest sensory organization test; SPT, Static posturography testing; SRF, Static rest frame; SSQ, Simulator sickness questionnaire; SUS, System usability scale; SV, Sway velocity; SVV, Subjective visual vertical test; VR, Virtual reality; TUG, Timed up and go test.

TABLE 4 | Outcomes of the studies and the actual HMD effect in individuals with pathological conditions.

References	Pathological condition	Balance assessment/training protocol	Outcomes	Actual HMD effect
Alahmari et al. (2014)	Vestibular	CoP area and velocity, comparison with SOT	Significant correlation between BRU and SOT (CoP area ICC = 0.64−0.81, velocity ICC = 0.44–0.76), higher CoP area, and velocity for younger vestibular adults compared to younger healthy adults.	Correlation with traditional test. Need to define new norms for different participant.
Caudron et al. (2014)	Idiopathic PD	Angular displacement measurement using kinematics (postural reaction peak, final orientation, stability of performance)	No changes in backward trunk tilt (mean amplitude = 9–11°, patients could recover their initial vertical orientation and had less frequent falls with visual feedback.	Less frequent falls immediately after HMD protocol.
Chiarovano et al. (2018)	Vestibular impairment (neuritis), cervicogenic dizziness, general dizziness and imbalance	DHI, postural sway	No correlation between DIH score and balance measurement, significant correlation between DIH and age, and balance and age.	Wearing HMD provided more possibilities for controlled visual conditions.
Cortés-Pérez et al. (2020)	Acute stroke	BBS, Tinetti scale, SVV, TUG, ABC	Higher functional balance (BBS and Tinetti scale) in VR compared to traditional and no treatment, larger reduction of risk of fall in VR patient.	Wearing HMD improved postural balance and gait and induces cautious behavior.
Epure et al. (2016)	Balance impaired	Survey, Kinect and Wii Balance Board	Higher stability when using monitor display compared to HMD (−20–30° vs. −20–40°).	Wearing HMD increased instability.
Gazzola et al. (2019)	Chronic vestibular disorder	CoP, LoS, VoS	Higher LoS in control group compared to training groups, higher CoP area and VoS in vestibular patients with a history of fall	Wearing HMD did not affect postural ability of vestibular patients without falls negatively, worsens postural ability of vestibular patients with falls, could be used to identify and quantify patients at risk of falling.
Janeh et al. (2019)	PD	SSQ, gait	Higher step width, cadence, gait variability, and gait pattern insecurity for PD patients in VR. Low SSQ and no significant increase over the time of experiment.	Wearing HMD increased gait symmetry with the help of visual scenes.
Kim et al. (2017)	PD	Force plate, SSQ	Decreased stress and higher arousal for PD patients (stress levels pre- and post = −9 ± 4 vs. −11 ± 3; arousal absolute changes in PD patients = 8 ± 7).	Wearing HMD decreased stress, did not change simulator sickness, and static and dynamic balance.
Micarelli et al. (2019)	UVH and MCI patients	SPT, DHI, ABC, DGI, SSQ	Improvement in otoneurological outcome measures, no changes in VOR gain.	Improved balance compared to traditional training.
Proffitt et al. (2018)	Post-stroke patients	Survey	Visual perception played a role in patients' preference, no adverse effects in balance.	Wearing HMD did not cause adverse effects, post-stroke patients need assistance.
Saldana et al. (2017)	At risk of fall older adults	Force plate, SSQ	Higher change of tilt in anteroposterior direction in patients with risk of fall (tilt at risk vs. healthy = 0.7°/s vs. 0.4°/s).	Similarity with traditional test. Need to define new norms.
Suarez et al. (2011)	Balance disorder	CE, SV	Higher reductions in postural responses after visual optokinetic stimulation	Wearing HMD resulted in improvements in postural control.
Suarez et al. (2011)	PD patients	LoS, CoP, BFR	PD had higher CoP values compared to the control group in the static visual field, BFR was reduced significantly in sensory VR scenarios.	Wearing HMD increased risk of falls when viewing sensory VR scenarios.

ABC, Activities-specific balance confidence; BBS, Berg balance scale; BFR, Balance functional reserve; BRU, Balance rehabilitation unit; CE, Confidential ellipse; CoP, Center of pressure; DGI, Dynamic gait index; DHI, Dizziness handicap inventory; FES-I, Falls efficacy scale international; FOV, Field of view; HMD, Head-mounted display; ICC, Intraclass correlation coefficient; MS, Multiple sclerosis; MSSQ, Motion sickness susceptibility questionnaire; PD, Parkinson's disease; PPA, Physiological profile assessment; SOT, Smart equitest sensory organization test; SPT, Static posturography testing; SRF, Static rest frame; SSQ, Simulator sickness questionnaire; SUS, System usability scale; SV, Sway velocity; SVV, Subjective visual vertical test; TUG, Timed up and go test; VoR, Vestibulo-ocular reflex; VoS, Velocity of oscillation; VR, Virtual reality.

Effectiveness of HMD Balance Training vs. Other Current Methods and Training Systems

A total of six studies compared the effectiveness of VR HMD systems with traditional training protocols (**Tables 3**, **4**). In these studies, HMD had good or better outcomes compared to the traditional training programs. In a small cohort, Cortés-Pérez et al. (2020) compared the effects of VR balance training with conventional physiotherapy in three stroke patients. After 2 months, the patient under VR training showed higher improvements of balance and obtained higher walking speed as compared to those with conventional physiotherapy. Chiarovano et al. (2018) in a cohort of balance-impaired individuals, evaluated the relationship between objective and subjective measurements of balance while perturbed visual inputs were introduced using VR. They found no correlations between objective and subjective balance measurements but showed that VR allowed revealing the participants with the real risk of fall. Their system categorized patients in three groups: those with high subjective measurements who passed the objective test, those with low subjective measurements who failed the objective test, and those with a correlation between subjective and objective measurements. Caudron et al. (2014) evaluated whether postural responses of patients with PD could be improved by traditional focus-based instructions or visual biofeedback. Biofeedback visual training improved stabilization and orientation components of postural control as compared to the traditional method of focus-based instructions. They also observed that online visualization of trunk and head orientations improved the stabilization when postural disturbance occurs. Liu et al. (2015) compared the effects of two slip training modalities on reducing fall frequency and reactive recovery. While both moving platform and VR training groups reduced their fall frequencies after training, VR group also reduced their forward trunk rotations which has been shown to bring the center of mass (CoM) of the body within boundaries of stability. Micarelli et al. (2019) compared traditional vestibular rehabilitation to VR balance training. They showed that by modifying visual information and increasing the complexity of their VR protocol, regardless of their health status, they could achieve significantly higher VOR gains as compared to those undergoing traditional vestibular rehabilitation alone.

DISCUSSION

Summary

The purpose of this systematic review was to explore the validity, reliability, safety, feasibility, and efficacy of HMD systems for assessing and training balance in older adults. The combined findings of the included studies show that VR HMD systems offer ecologically valid scenarios to assess and train functional balance. There is however a need to define standardized norms and protocols according to age, health status, and severity of disease. The studies also showed that several parameters of display type, VR scenario, and the duration of exposure can contribute to the safety and feasibility of HMD VR systems for balance training. These features could be adjusted according to participants' needs to ensure safety and efficacy of training. The use of HMD systems was effective in training balance and can be a useful tool to augment previously established interventions. Various visual scenarios can be added, removed, isolated, and manipulated to identify and treat specific balance-related impairments of different clinical conditions. The level of difficulty can also be adjusted to the patients' baseline levels to allow progress or regress when needed.

Validity and Effectiveness of HMD VR Systems for Measuring and Training Balance in Healthy Older Adults

Only one study showed that HMD VR systems were valid and effective in healthy older adults. Those who trained with HMD VR systems showed improved gait parameters and lowered frequency of falls (Giotakos et al., 2007; Parijat et al., 2015; Kim et al., 2017). These improvements are probably due to the challenges that are induced when using VR scenarios. While some researchers reported that older adults have similar balance performance in traditional and VR scenarios (Alahmari et al., 2014), others have argued that participants' behaviors in VR versions of balance tests differ significantly (Muhla et al., 2020). One explanation could be that HMD systems remove the surrounding visual cues that older adults require for maintaining balance. Older adults will have to rely on VR scenarios, presented through HMD systems, which offer different visual stimulation compared to the real world. Another explanation is that HMD VR systems evoke various proactive and reactive postural adjustments (Liu et al., 2015; Parijat et al., 2015). It has been shown that participants generally walk slower, and their performance is lower in VR balance tests as compared to the traditional ones (Nyberg et al., 2006; Almajid et al., 2020). Participants also acquire more cautious behaviors when completing VR tasks in less secure VEs (Muhla et al., 2020) and take more time and steps which are adaptive strategies to ensure balance and avoid falling (Nyberg et al., 2006). Other strategies that healthy adults employed for increasing the stability in unfamiliar situations included a reduced forward displacement of whole-body CoM and lowered height of CoM (Thomas et al., 2016).

Motion sickness is believed to happen when the brain's assumption about sensory information does not match the actual received signals (Reason, 1978). This mismatch could be due to hardware and software limitations of VR HMD systems. The motion-to-photon latency, or the delay between the users' movements and their reflections on the display, can contribute to the motion sickness (Choi et al., 2018). The effects of latency are so important that even the smallest lag degrades the sense of balance compared to the naked eye with a similar FOV (Kawamura and Kijima, 2016). The interactions of gender and sickness history can also influence the risk of motion sickness and instability and should be considered when designing the HMD hardware and software (Kawamura and Kijima, 2016; Munafo et al., 2017). It has been shown that females are more susceptible to motion sickness (Munafo et al., 2017). In pathological conditions, such as PD, the sensory deficits could make the individuals less prone to the simulator sickness and could be less influential on their perceptions of balance (Kim et al., 2017). On the other hand, HMD related head movements pose sensory mismatch to the central nervous system of UVH patients, which later tries to resolve the mismatch (Clendaniel, 2010).

The duration of VR exposure should also be carefully monitored as it can provoke sickness symptoms. Previous research has associated verbal reports of cybersickness severity, as well as relatively high incidences of simulator sickness with the duration of HMD use (Treleaven et al., 2015; Dennison et al., 2016). HMD systems could also cause visual fatigue due to a disruption of the natural relationship between vergence and accommodation (Mon-Williams and Wann, 1998). Various physiological measurements have been used to estimate post-immersion cybersickness (Dennison et al., 2016). Those with greater levels of cybersickness may show less variations in postural sway (Dennison and D'Zmura, 2017). Therefore, the physiological measurements are important for differentiating between motion sickness and actual imbalance (Dennison et al., 2016).

While postural instability increases with time, it only happens when visual perturbations are present. It seems that elements of VR scenarios could also affect the way healthy older adults interact with the systems. Some researchers reported that increasing rotation speed in VR scenarios could increase cybersickness (Dennison and D'Zmura, 2017). Others have shown that speed manipulations in simple and complex VR scenarios do not cause motion sickness (De Keersmaecker et al., 2020). Adding an independent visual background to the scene could also reduce postural disturbance and could reduce the simulator sickness (Prothero et al., 1999; Duh, 2001).

As compared to younger adults, the older population is relatively more egocentric and more accurate when they are in the first-person perspective compared to the third-person perspective (Mattan et al., 2017). Research has also highlighted that a sense of actual presence in VR could be weakened if older participants view the VE from a third-person perspective (Lenggenhager et al., 2007). Therefore, perspective-taking capacity should be considered when designing virtual scenarios for older adults. Concerning the interaction with avatars, only one study included animal avatars (Cortés-Pérez et al., 2020). It seems that other contextual information around the avatars, including ground and objects, could act as visual reference points for the participants to adjust their balance. Naturalness of the movements, or the alignment between visual and proprioceptive senses, is also reduced when using HMD systems (Sander et al., 2006). Mismatch in proprioceptive and visual senses, or the inability to get information from body, as well as the low FOV and stability of the headset, could also reduce the naturalness of the movements. Additional motion tracking sensors, such as Leap Motion, can embed hands' motions directly into HMD systems, which can act as visual reference points for older adults (Scheggi et al., 2015). After a few trials, participants start to adapt to the gaming and protocol mechanics and adjust their walking to the VR and visual perturbations (Liu et al., 2015). As a result, VR might lose its ability to induce perturbations after 2–3 trials. It is thus important that HMD VR trainings use different scenarios and intensities so that they reduce the risk of adaptation to the VR environments and allow progression to optimize the results.

Validity and Effectiveness of HMD VR Systems to Measure and Train Balance in Pathological Conditions

There is no standard protocol for defining postural instability in patients with different pathologies. Therefore, the results of different studies are conflicting regarding both assessment and training paradigms.

Fear of falling and perceived postural threat can affect gait patterns in older adults and could induce stiffing strategies in their joints. VR exposure therapy scenarios could increase self-efficacy beliefs of falling and provide a sense of control over falling (Giotakos et al., 2007). In the case of PD and in addition to imbalance issues, many patients have lower access to visual cues. They might have a fear of imbalance and decreased gait performance, preventing them from benefiting from rehabilitation. Patients with neurological conditions (PD and stroke) show improvement of balance and reduction of the risk of falls after VR training as compared to non-VR training (Caudron et al., 2014, Cortés-Pérez et al., 2020). Vestibular patients who are suffering from chronic balance impairments are likely to benefit from HMD VR training. Those who trained with HMD showed overall improvements in vestibulo-ocular reflex gain and in posturography parameters (Micarelli et al., 2017; Viziano et al., 2019). Vestibular patients without MCI have also shown improvements in otoneurological outcome measures after HMD VR training (Micarelli et al., 2019).

The HMD VR scenarios could also help in identifying different patients within certain pathology. For example, Chiarovano et al. (2018) could identify three types of vestibular patients: those with high dizziness scores who did not fail the balance test, those with low dizziness scores who failed the test, and other participants whom their dizziness scores were correlated with balance measurements. Visually dependent older adults with greater risks of fall show smaller acceleration ranges in VR TUG test compared to visually dependent older adults without any risks (Almajid et al., 2020). Patients with PD also show different CoP and balance functional reserve values in static and dynamic VR scenes (Suarez et al., 2011). The differences in the balance-related measures could be used to identify patients with higher balance impairments, to understand who will benefit the most from HMD VR training and guide users on how to adjust the training scenarios to the patients' specific needs. Nevertheless, there is a need for further studies to establish realiable cut-offs for each of these balance-related metrics and to adjust them to the associated pathological disorders. Then, the benefit of using VR is that sensory loads can be tailored and increased according to individuals' visual perception and threshold values to enhance balance performance (Proffitt et al., 2018). Notwithstanding the results of these studies showing the usability of HMDs to identify balance-impaired individuals, there are also conflicting results. Some researchers have shown that the CoP area and velocity were similar to healthy older adults during traditional and VR tests (Alahmari et al., 2014) and others have shown that CoP values were worse in vestibular patients compared to healthy older adults (Gazzola et al., 2019).

As patients get accustomed to the training, postural responses tend to adapt to the visual scenarios (Suárez et al., 2006). It has been suggested that different VR HMD systems could be more salient by improving stimulus presentation through encouraging participants to explore the VE (Menzies et al., 2016). Instructing participants to follow targets that move out of their direct FOV could also increase their presence. VR tasks should also challenge known sensorimotor deficits of individuals with special clinical and functional needs. For example, in patients with PD, more challenging walking tasks like turning, obstacle handling, or passing through doorways are impaired (Mirelman et al., 2011). Similar VR scenarios should be incorporated in the rehabilitation programs to increase their clinical relevance for their targeted population. Different types of virtual scenarios and frequencies of movement disturbance could also affect participants' stability (Jurkojc et al., 2017). It should be noted that participants get accustomed to these disturbances faster in open (seeing the horizon in 100 m distance) compared to closed sceneries (a room; Jurkojc et al., 2017). It is therefore important that patients are exposed to various and progressive training scenarios. The VR tests are easier to standardize than ordinary tests performed in gymnasiums or rehabilitation training facilities, which is a great advantage. Using VR systems, lighting conditions, room size, and other features can be set according to desired specifications. Each condition can also be manipulated individually to calculate their contribution in balance outcomes (Ferdous et al., 2018) and help to further understanding of complex mechanisms involved in balance.

Challenges and Practical Implications

Older adults may suffer from technological illiteracy. The reluctance of not using VR, stems from common beliefs about required equipment, financial costs, and unfamiliarity with the benefits of VR systems (Schwartzman et al., 2012). An insufficient knowledge of the systems both by the patients and their therapists could also hamper their acceptance. Therefore, VR training systems should have user-centric and user-friendly design. Developers should focus on improving the usability and accessibility of VR training systems, for promoting adherence and motivation. As participants' behaviors in VR differ from traditional environments, new norms should be defined to compare different participants. Different amounts of time spent in the VR rehabilitation programs, as well as motivation and other psychological parameters may also affect the outcomes and practitioners should consider them when interpreting the results.

Researchers, developers, and practitioners should also consider participants' gender and age, as well as sickness and medication history because one system and scenario may not work for everyone. Carefully planned visual scenarios with right frequencies (speed) and duration should be used to prevent visual fatigue and cybersickness. The timing and intensity of introducing new VR elements and scenarios should also be considered when developing training programs to ensure a rate of adaptation that allows continued progression and maximizes training efficacy. Future VR systems should be able to adjust these scenarios and their complexity, as well as sensory load, duration, and game algorithm for different participants. As technology is advancing rapidly, future HMD and processing systems will be faster, smaller, more powerful, and less obtrusive. Hopefully lower latency could be translated in more realistic interactions with VR HMD systems.

Recommendations for Future Research

Almost all the studies display statistical limitations that limit external validity of their findings. Future studies should include more subjects, control confounding variables, and match for sex to enhance the generalizability of their findings. The effects of other features of VR, such as field of regard, display size, head-based rendering, stereoscopy, realism of lighting, and scene update latency, on balance should be further explored. Future research should also target the effects of biofeedback features, including audio and vibrotactile biofeedback, on postural control. It is also important to explore whether participating in VR trainings translates into functional improvements in real life. Despite the increasing amount of studies in this area, there is still much room to improve the current technologies, and the limitations identified in this review, should be considered when planning future studies.

Study Limitations

There are a few study limitations to note. The low samples sizes and the high risk of confounding from most studies are the major concerns, and limit both the internal and external validity of findings. The aims and populations of the included studies were heterogenous, which hinder generalizing the results. Many clinical and balance-related measurements could be influenced by medication and severity of the disease. Inevitably, healthy and unhealthy older adults might have different SSQ or arousal levels because of the medication they were taking. Therefore, it is important to be mindful of such parameters when interpreting the results. Considering that some diseases are more prevalent in specific sex, sample size should be sex-matched to avoid selection bias. Some patients, such as vestibular patients, could also be less susceptible to any form of rehabilitation, possibly due the progressive physiological activity decline. Finally, some questionnaires are only validated in healthy younger adults and therefore should be validated in older adults and in individuals with specific diseases to establish the psychometric properties and thresholds before being used as reliable comparative measures. Unfortunately, there is still not enough evidence in the literature to provide a focused and critical review on the effects of HMD VR systems in a specific population or clinical condition. Still, this systematic review provides a comprehensive view on how HMD VR systems could be used to assess balance, and which are their potential benefits if used for training balance in older adults.

CONCLUSIONS

HMD VR systems offer ecologically valid scenarios to assess and train functional balance that can be used alone or combined with other interventions. Various visual scenarios can be added, removed, isolated, and manipulated to identify and adjust to specific balance-related impairments of different clinical conditions. HMD VR training creates real-based scenarios that can be useful to improve postural control and gait patterns by altering motor planning and muscle activation. HMD VR systems could also augment fall prevention programs of those with higher risk of falling by providing more visual cues on how to handle context-based unexpected events. HMD systems can also be useful in identifying those with balance impairments, but there is still a need for further validation and cut-offs for each balance metric and adjustment to be made according to the pathological conditions. The risk of bias and overall quality of studies is still a major concern. There is a need of high-powered and high-quality studies before any definitive recommendations could be made on the use of HMD VR training to improve balance in older adults. Future studies should focus on establishing new norms and protocols that are adjusted to different individual characteristics, including age, gender, health status, type and severity of clinical condition, and cognitive and physical impairments, for maximizing their benefits.

AUTHOR CONTRIBUTIONS

PS designed the study, performed the database searches, extracted the data, analyzed the results, drafted the manuscript, and performed the risk of bias judgment. RA contributed to the analysis of the results, risk of bias assessment, drafting manuscript, and critical revision. Both authors approved the manuscript.

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Conflict of Interest: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

APPENDIX

Database	Search strategy
PubMed and Web of Science	((HMD OR head mount∗ display OR "virtual reality" OR "immersive reality" OR "artificial environment" OR "simulated 3D environment" OR "simulated three-dimensional environment") AND (balance OR posture OR fall)) AND (vestibular OR visual OR somatosensory OR context∗)
Scopus	((hmd OR head AND mount∗ AND display OR "virtual reality" OR "immersive reality" OR "artificial environment" OR "simulated 3D environment" OR "simulated three-dimensional environment") AND (balance OR posture OR fall)) AND (vestibular OR visual OR somatosensory OR context∗) not INDEX (medline)
EBSCOhost	((HMD OR head mount∗ display OR "virtual reality" OR "immersive reality" OR "artificial environment" OR "simulated 3D environment" OR "simulated three-dimensional environment") AND (balance OR posture OR fall)) AND (vestibular OR visual OR somatosensory OR context∗)

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