EXPLORING MECHANISMS OF REACTIVE BALANCE CONTROL IN FALL-PRONE OLDER PEOPLE WITH NEUROMUSCULAR AND BIOMECHANICAL ANALYSES

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Exploring Mechanisms of Reactive Balance Control in Fall-prone Older People with Neuromuscular and Biomechanical Analyses

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CERTIFICATE OF ORIGINALITY

I hereby declare that this thesis is my own work and that, to the best of my knowledge and belief, it reproduces no material previously published or written, nor material that has been accepted for the award of any other degree or diploma, except where due acknowledgement has been made in the text.

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ABSTRACT

Falls and fall-related injuries adversely affect the community-dwelling older people. Even the multifactorial fall-prevention programs could have insufficient effectiveness in reducing falls among the older people with history of falls (i.e., older fallers). Previous studies mainly used postural sways, gait analyses, or age-related neuromuscular analyses to evaluate balance and gait disorders. The neuromuscular and biomechanical mechanisms underlying reactive balance control were less focused. Identifying these intrinsic deficits of reactive balance control among the fall-prone people is essential to guide a more targeted design of fall-prevention exercises for them.

This PhD project aims to comprehensively investigate the biomechanical and neuromuscular alterations in reactive balance control that can indicate the fall histories/risks among older adults by using synchronized motion capture, electromyographic (EMG), and mechanomyographic (MMG) technologies.

Four observational studies have been conducted, including:

a. Exploring response speed and sequence of multiple lower-limb muscles/joints following unexpected translational moving-platform (Study 1) or waist-pull balance perturbations (Study 3) in healthy young adults.
b. Comparing neuromuscular and kinematic responses following unexpected translational moving-platform perturbations (Study 2) or waist-pull balance perturbations (Study 4) in older fallers vs. older non-fallers.
c. Examining what responses can predict older adults' prospective falls over 1 year (Study 4).

Via a step-by-step approach, all the studies are linked together by the theme of probing the more intrinsic mechanisms of reactive balance control in fall-prone people.

Pilot studies in healthy young adults (Study 1&3) have shown that ankle muscles had the largest activation rate among the examined leg muscles following either anteroposterior or mediolateral sudden balance loss.

Two studies in older adults (Study 2&4) have both revealed that older fallers had insufficient activation of proximal hip muscles for reactive balance control. Study 2 has preliminarily found that fallers had to use the suspensory strategy (e.g., bending knees) to compensate for their slower reaction of ankle/hip strategies following unexpected translational moving-platform perturbations as compared to non-fallers. In Study 4, the reactive balance control induced by unexpected waist-pull perturbations was assessed in 36 fallers vs. 36 non-fallers, and these older adults' prospective falls were tracked. Older fallers were observed with a quicker neuromuscular response following anterior perturbations but slower neuromuscular responses following posterior/medial/lateral perturbations than non-fallers. Additionally, the older adults' prospective falls were predicted by (1) slowed/reduced activation of hip abductor (among the eight investigated leg muscles), (2) altered responses mostly in hip/knee joint (than ankle joint), and (3) alterations mostly in response to the mediolateral perturbations (than anteroposterior perturbations).

In conclusion, the findings of this PhD project support the assessment of hip abductor's activity during reactive balance tasks to complement the current fall-risk assessment, providing insights for a more targeted fall-prevention management for older people. Future work is merited to examine the effectiveness of training that targets the identified fall-related factors on preventing falls among the fall-prone older adults.

PUBLICATIONS ARISING FROM THIS THESIS

A. Journal article publication arising from this thesis:

1) **Zhu RTL**, Lyu PZ, Li S, Tong CY, Ling YT, Ma CZH*. How does lower limb respond to unexpected balance perturbations? New insights from synchronized human kinetics, kinematics, electromyography (EMG) and mechanomyography (MMG) data. Biosensors. (**JCR Q1,** *Category: INSTRUMENTS & INSTRUMENTATION*) 2022 Jun 18;12(6):430.

2) Tong CY, **Zhu RTL (co-first)**, Ling YT, Scheeren EM, Lam FMH, Fu H, Ma CZH*. Muscular and Kinematic Responses to Unexpected Translational Balance Perturbation: A Pilot Study in Healthy Young Adults. Bioengineering. (**JCR Q2,** *Category: ENGINEERING, BIOMEDICAL*) 2023 Jul 13;10(7):831.

3) **Zhu RTL**, Hung TTM, Lam FMH, Li JZ, Luo YY, Sun JT, Wang SJ, Ma CZH*. Older fallers' comprehensive neuromuscular and kinematic alterations in reactive balance control: indicators of balance decline or compensation? A pilot study. Bioengineering. (JCR Q2, Category: *ENGINEERING, BIOMEDICAL*) 2025 Jan 14;12(1):66.

4) **Zhu RTL**, Zuo JJ, Li KJ, Lam FMH, Wong AYL, Yang L, Bai X, Wong MS, Kwok T, Zheng YP, Ma CZH*. Association of lower-limb strength with different fall histories or prospective falls in community-dwelling older people: a systematic review and meta-analysis. BMC Geriatrics. (JCR Q2, Category: GERIATRICS & GERONTOLOGY) 2025 Feb 6;25:83.

5) **Zhu RTL**, Hung TTM, Li S, Zheng YP, Ma CZH*. Probing Neuromuscular and Biomechanical Balance-Control Mechanisms: Fall Prediction and Impact of Falls in Older Adults. Nature Communications. (JCR Q1, *Category: MULTIDISCIPLINARY SCIENCES*) 2024. ***In preparation.***

6) **Zhu RTL**, Huang C, Wong DWC, Ma CZH*. Response speed of reactive balance control in community-dwelling older people with different fall histories or fall risks: A systematic review and meta-analysis. Clinical Rehabilitation. (JCR Q1, *Category: REHABILITATION*) 2024. ***In preparation.***

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C. Invited talk:

1) **Zhu RTL**. In-depth assessment and targeted training of reactive balance: empowering fall-risk prediction and fall prevention. Annual academic conference in 2025 held by Yunnan Association of Rehabilitation Medicine & Yunnan Association of Rehabilitation of Disabled Persons, April 2025, Kunming, China.

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2) **Zhu RTL**, Ma CZH*. Kinetic and Neuromuscular Insights on how Older Adults with History of Falls Maintain Reactive Standing Balance. *World Physiotherapy Congress 2023*, June 2023, Dubai, UAE. *Oral presentation.*

3) **Zhu RTL**, Lyu PZ, Li S, Tong CY, Ma CZH*. How eight major leg muscles respond to unexpected perturbations and maintain standing balance in healthy young adults? *IUPESM World Congress on Medical Physics and Biomedical Engineering 2022*, June 2022, Singapore. *Online oral presentation.*

4) **朱堂龙**, 马宗浩*. 探究社区老年人应对跌倒风险的姿势平衡控制：基于肌电图和生物力学的分析。*中华医学会第二十三次全国物理医学与康复学学术会议 (The 23rd National Physical Medicine and Rehabilitation Congress of Chinese Medicine Association),* September 2023, Shanghai, China. O*ral presentation.*

5) **<u>Zhu RTL</u>**, Zheng YP, Ma CZH*. Impaired Hip Muscle Activation and Kinetic Output Contribute to Poor Balance Control in Older Fallers. *Biomedical Engineering Conference 2023 Healthcare Innovation & Technology: Roles of HK Biomedical Engineers in the GBA,* August 2023, Hong Kong SAR, China. *Poster presentation.*

6) **<u>Zhu RTL</u>**, Zheng YP, Ma CZH*. Kinematics and Muscle co-contractions at the Hip, Knee, and Ankle Joints in Older Adults with History of Falls after Sudden Loss of Balance. *PAIR (The PolyU Academy for Interdisciplinary Research) Conference 2023*, May 2023, Hong Kong SAR, China. *Poster presentation.*

7) Tong CY, **<u>Zhu RTL</u>**, Lyu PZ, Ma CZH*. Lower-Limb Muscle Activities When Maintaining Static Balance. *7th Singapore Rehabilitation Conference & 7th Asian Prosthetics and Orthotics Scientific Meeting (SRC – APOSM 2022)*, October 2022, Singapore. *Online oral presentation.* (**Best Abstract Award (Group)** - Prosthetics and Orthotics Outstanding Capstone Project Award)***.***

8) Luo YY, **<u>Zhu RTL</u>**, Zheng YP, Ma CZH*. Reactive Balance Control of Older Adults with Fall History and Fear of Falling: Probing Neuromuscular and Biomechanical Mechanisms. *The 11th WACBE (World Association for Chinese Biomedical Engineers) World Congress on Bioengineering,* August 2024, Hong Kong SAR, China. *Poster presentation.*

9) Ma CZH*, **<u>Zhu RTL</u>**, Hung TTM, Li S. Prospective Fall Prediction of Older Adults Based on Neuromuscular and Biomechanical Balance-Control Mechanisms. 2024 Rehabilitation Engineering Academic Conference, November 2024, Qingdao, China.

E. Journal publication during the PhD study period (not arising from this thesis):

1) Lyu PZ#, **<u>Zhu RTL (co-first)</u>**#, Ling YT, Wang LK, Zheng YP, Ma CZH*. How paretic and non-paretic ankle muscles contract during walking in stroke survivors? New insight from novel wearable ultrasound imaging and sensing technology. Biosensors. (**JCR Q1,** *Category: INSTRUMENTS & INSTRUMENTATION*) 2022 May 18;12(5):349.

2) Liu W, Wu HD, Li YY, **<u>Zhu RTL</u>**, Luo YY, Ling YT, Wang LK, Wang JF, Zheng YP, Ma CZH*. Effect of ankle-foot orthosis on paretic gastrocnemius and tibialis anterior muscles contraction of stroke survivors during walking: a pilot study. Biosensors. (**JCR Q1**, *Category: INSTRUMENTS & INSTRUMENTATION*) 2024 Dec 4;14(12):595.

3) **Zhu RTL**, Schulte F, Singh NB, Ma CZH*, Awai CE, Ravi DK. Effects of a Single-session Perturbation-based Balance Training with Progressive Intensities in Healthy Older Adults: A Pilot Randomized Controlled Trial. Scientific Reports. ***Under Review.***

4) **Zhu RTL (co-first)**#, Huang C#, He WT, Luo YY, Shou DH*, Ma CZH*. Compliance with external hip protectors in older adults and the associated factors: A systematic review and meta-analysis. ***To be submitted.***

F. Symposium presentation during the PhD study period (not arising from this thesis):

1) Ravi DK, **Zhu RTL**. Quantifying the impact: Movement biomarkers in perturbation-based balance training for fall prevention. *Symposium "Current Research in Sport Science" for Master students in University of Konstanz,* July 2024, German.

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LIST OF ABBREVIATIONS

Abbreviation	Full name
AM	adductor magnus
ANOVA	analysis of variance
ASIS	anterior superior iliac spine
APA	anticipatory postural adjustments
AUC	area under the curve
BESTest	Balance Evaluation Systems Test
BoS	base of support
BBS	Berg Balance Scale
BMI	body mass index
CoM	center of body mass
CoP	center of pressure
PASE-C	Chinese Version of The Physical Activity Scale for the Elderly
CCI	co-contraction index
CPA	compensatory postural adjustment
CI	confidence interval
CEDE	Consensus for Experimental Design in Electromyography
EMG	electromyography; electromyographical
FES-I	Fall Efficacy Scale-International
GMax	gluteus maximus
GMed	gluteus medius
GRADE	Grading of Recommendations, Assessment, Development and Evaluation
HR	hazards ratio
IL	iliopsoas
IPAQ-S	International Physical Activity Questionnaire-Short version

ICC	intraclass correlation coefficients
BF	bicep femoris
MMT	manual muscle testing
MVC	maximal voluntary contraction
MVIC	maximal voluntary isometric contraction
MMG	mechanomyography; mechanomyographical
MG	medial gastrocnemius
Mini-BESTest	Mini Balance Evaluation Systems Test
MFI	multifactorial intervention
OR	odds ratio
POMA	Performance-Oriented Mobility Assessment
PBT	perturbation-based balance training
PSIS	posterior superior iliac spine
PRISMA	Preferred Reporting Items for Systematic Review and Meta-Analyses
PROSPERO	Prospective Register of Systematic Reviews
RFD	rate of force development
RTD	rate of torque development
ROC	receiver operating characteristic
RF	rectus femoris
RR	risk ratio
SA	sartorius
ST	semitendinosus
SPPB	Short Physical Performance Battery
SMG	sonomyography; sonomyographic
SD	standard deviation
SMD	standardized mean difference
SENIAM	Surface Electromyography for the Non-Invasive Assessment of Muscles
TA	tibialis anterior
TUG	Timed Up and Go

Chapter 1 Introduction

1.1 Falls in Older Adults- A Public Health Issue

Falls, based on the definition of World Health Organization (2021), occur when a person inadvertently lands on the ground, floor, or any lower level. They are particularly common among older populations. A recent systematic review reported that the global prevalence of falls among older adults is 26.5% (Salari et al., 2022). While most falls are non-fatal, they can lead to serious physical consequences, such as hip/wrist/elbow fractures, together with psychological effects, including a fear of falling (Asai et al., 2022). Fall-related decline in physical activity or even deconditioning can progressively weaken the older adults (Lusardi et al., 2017). Furthermore, falls impose a heavy burden on society due to direct healthcare costs and indirect productivity losses (Dykes et al., 2023). It is important to identify the modifiable fall-risk factors early for proving targeted fall-prevention management. This may help reduce the fall incidences and related injuries, as well as alleviate the considerable socioeconomic burden.

1.2 Multifaceted Fall-risk Factors

Numerous studies have demonstrated the multifactorial nature of falls among the older population. Falls occurred or may well occur under the complex interaction of various factors. Generally, these fall-risk factors can be categorized into three broad types.

1) <u>the environmental ones</u>

This category of fall-risk factors refer to those hazardous features existing in the environment that can provoke falls like dim lighting, uneven ground, slippery surface and poorly designed public spaces (Kim et al., 2020).

2) <u>the behavioral ones</u>

Behavioral fall-risk factors involve the reduction of physical activity (Gates et al., 2008), fear of falling (Scheffer et al., 2008), ill-suited footwears (Menant et al., 2008), medication use (Huang et al., 2012), alcohol misuse (Mukamal et al., 2004), depression (Kvelde et al., 2013), lack of social support (Chang & Ganz, 2007), etc.

3) the biological ones

Falls have also been reported to link with some biological factors, such as older age (World Health Organization, 2008), female gender, Caucasian ethnicity (Karlsson et al., 2013), decreased function of visual, proprioceptive and vestibular systems (Barker et al., 2009), balance and gait disorders (Seidler et al., 2010), cognitive impairment (Muir et al., 2012), fall histories (J. C. Kim et al., 2017), history of some chronic diseases (Kim et al., 2020), etc. Falls occurred or may well occur under the complex interaction of such factors.

1.3 Fall-prevention Approaches Are Multifactorial

Contingent upon the diversity of fall-risk factors, the intervention programs dealing with falls have also been multidimensional. These preventive or therapeutic approaches for older people can vary from the exercise intervention, education and vitamin D supplementation to the drug-targeted therapy, surgical management, footwear modification and home hazard elimination (Karlsson et al., 2013). Multifactorial intervention (MFI) programs incorporating three or more of such approaches have appeared to be most effective in reducing falls among older individuals, as opposed to the single-factorial or two-factorial treatments (Cheng et al., 2018; Rubenstein, 2006). Exercise intervention, usually served as an indispensable component of MFI program, was also suggested to have the strongest effects on fall prevention when involving balance, strength, flexibility and endurance trainings concurrently (Cadore et al., 2013; Gillespie et al., 2012). Conversely, specific forms of physical activity may increase the likelihood of falling by shifting a person’s center of gravity (Chan et al., 2007).

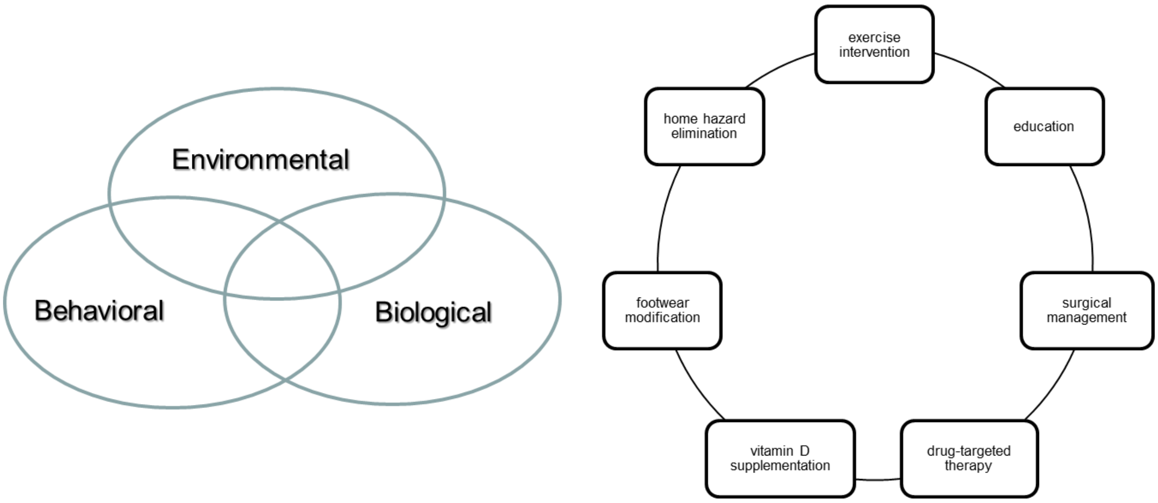

Figure 1-1. Falls and fall-prevention strategies are multifactorial (summarized from Karlsson et al., 2013).

The multifaceted characteristic of these treatments, however, does not necessarily mean that they can be without intervening emphasis. Providing targeted interventions is still of vital importance. For one thing, a more tailored MFI program will be cost-efficient and thus suitable for the occasions when fall-prevention resources are limited (Cheng et al., 2018). For another thing, older adults may be more willing to take part in and adhere to an exercise regimen if similar functional benefits can be obtained with focused but less volume of training (Radaelli et al., 2019).

Nevertheless, even upon joining an individualized MFI program, some older fallers (i.e., older people with fall histories) may still not respond well and fall recurrently. Two randomized controlled trials conducted in Netherlands once studied the effectiveness of MFI programs and usual post-falling injury care, respectively, on the older people who had experienced at least one injurious fall (de Vries et al., 2010; Vind et al., 2009). The MFI programs were designed based on the fall-risk assessment for each individual, involved multidisciplinary medical staff, and consisted of drug intervention, exercise training, education, home hazard reduction etc. But these programs turned out to be no more effective than the usual medical care in reducing older fallers’ rate of future falling, in shortening the time to first fall, or in improving their ability of daily living. In Hong Kong, the 1-year prevalence of recurrent falls (i.e. at least two falls each year) among the community-dwelling older people was about 5%, which means one in every four older fallers would undergo repeated falls (Chu et al., 2007). These recurrent older fallers may suffer more severe disabilities and psychological problems than those who fell once. It seems more intractable to prevent and manage the recurrent falls.

1.4 Possible Factors Undermining Effectiveness of Fall-Prevention Strategies

Leaving aside the implementation issues (e.g., by whom and how the intervention is delivered), limitations of the currently identified fall-risk factors should be considered more in-depth, in terms of why a theoretically ideal multifaceted treatment cannot succeed in fall prevention.

The first limitation is that quite a few biological risk factors of falling are identified as closely associated with degenerative processes. Except for some non-modifiable ones like the race and the history of previous falls, the other modifiable fall-related factors (e.g., the decline of sensorimotor and cognitive capacities) are usually involved with either physiological or pathological degradation. More specifically, ageing and/or the progression of chronic diseases (e.g., the Parkinson’s disease) would counteract the benefits of preventive acts to an extent. Thus, when older community-dwellers’ disabilities do not properly resolve with the treatment of the underlying disorders (e.g. hemiparesis, ataxia, persistent weakness or joint deformities) (Rubenstein, 2006), even the tailored MPI program may appear not that useful to prevent this population from future falls.

The second limitation might be the inappropriate ways whereby some fall-related factors were determined. As reported in some studies, previous fall history is one of the fall-risk factors (Deandrea et al., 2010). This indicates that older fallers, i.e., the older people who have fallen before, are more prone to future falls. Targeting the problems that exist in these high-risk groups may help detect the risk factors of falls that are more related to themselves. However, so far, quite a few biological factors of falling have been determined by comparing between the different age groups only, especially between the older and young adults, without putting more focus on factor of falls by maximally eliminating the influence of the age factor. Based on such study participants and the corresponding identified factors (Palmer et al., 2017; Schettino et al., 2014), some fall-prevention programs have been developed and evaluated (McKinnon et al., 2017). Unfortunately, the previous attempts have generated inconsistent results in terms of the fall-prevention effectiveness (Kobayashi et al., 2016; Piirainen et al., 2014). A possible explanation might be that most previous fall-prevention managements have been designed and developed based on the findings that did not eliminate the factor of age, when the comparisons have mostly been conducted between the young and older people.

Attempts can be made to investigate the mechanisms that are closely correlated with falls 1) by comparing older fallers vs. older non-fallers and/or 2) by investigating the factors that can predict prospective fall incidences among the older adults. It is reasonable to expect that if the identified fall-risk factors are derived in a less age-related way by purely looking at the factor of history of falls and minimizing the confounding factor of age, the effectiveness of the correspondingly developed fall-prevention managements can be less susceptible to the age of the older adults. The convincing method for identifying risk factors associated with falls is through prospective cohort research. Alternatively, distinguishing between fallers and non-fallers provides an economic and feasible way to preliminarily identify the potential fall-related factors. Some of the deficits reflected on older fallers are also good indicators of fall risks, giving valuable guidance to the community-dwelling fallers and non-fallers on remediation and prevention measures, respectively. Together, for one thing, the identification of more potentially fall-related factors merits further study by comparing the older fallers and older non-fallers at similar age; for another thing, the more in-depth fall-risk factors can be determined by directly conducting the prospective cohort studies with confounders (e.g., age) controlled.

1.5 Assessment of Postural Balance Control in Fall-prone Older People

Balance and gait disorders have been consistently identified as the second leading cause for falls in humans (Rubenstein, 2006) and the primary cause for falls in older adults (Salzman, 2010). Generally, there are two types of postural balance control. One is volitional balance control, also called anticipatory or proactive balance control, which means the person has prior knowledge of the incoming perturbation and makes the postural adjustment in advance (Tisserand et al., 2015). The other is the reactive balance control, also called the compensatory or automatic balance control, which means the incoming perturbation is unexpected and the person makes postural adjustments after feeling the feedback of perturbation (Tisserand et al., 2015). Plenty of clinical or instrumented tests have been developed to test the balance performance and identify the older people who are prone to falls.

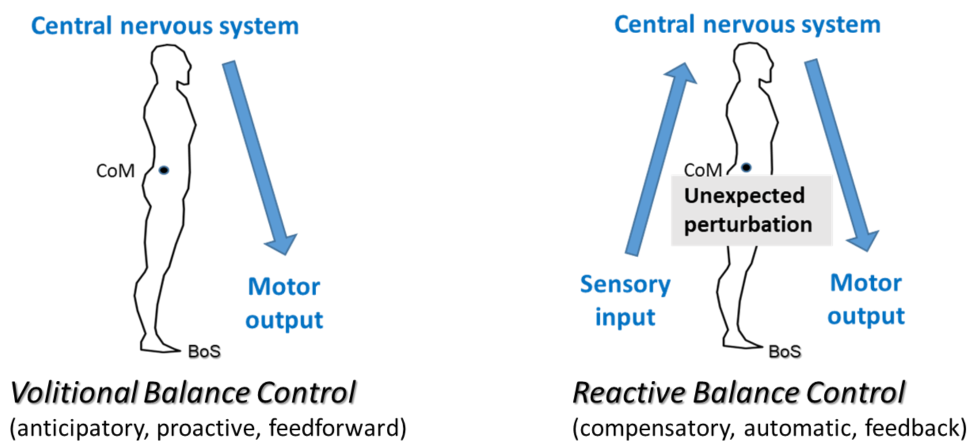

Figure 1-2. Two types of postural control (summarized from Tisserand et al., 2015).

1.5.1 Clinical Assessment of Balance Performance

Most clinical tests/scales assess the client's volitional balance control, whereas there exist only a restricted set of clinical tests or scales designed to evaluate reactive balance control. Examples for assessing volitional balance control are the Timed Up and Go test (TUG), the Short Physical Performance Battery (SPPB), the Berg Balance scale (BBS), and the Performance Oriented Mobility Assessment (POMA). In clinics or hospitals, the examiner typically induces the client's reactive balance control manually. Specifically, one way to induce sudden balance loss is the pull test, which means the client is induced by the examiner's backward pull on the shoulder (Foreman et al., 2011). Another way is the push and release test, which means the client leans on the examiner's hands and the examiner then suddenly removes the hands to induce unexpected balance loss (Valkovič et al., 2008), and the sub-item of assessing reactive balance control in the Balance Evaluation Systems Test (BESTest) (Horak et al., 2009) or the Mini Balance Evaluation Systems Test (Mini-BESTest) is also based on this way (Franchignoni et al., 2010). For both the two ways of assessing reactive balance control, the client's performance is scored based on how many steps the client made or whether the client has a fall after the manually induced perturbation.

Together, the clinical tests/scales for assessing postural balance control are relatively quick tools that enable preliminary screening of older adults' fall risks. However, given the subjective characteristic of the clinical tests/scales, they may still <u>rely much on the examiner's clinical experience</u>. For example, some older adults who are afraid of falling may not dare to lean their body weight fully on the examiner's hands. If the examiner does not identify this or give further instructions to encourage the client, the scoring of the client's reactive balance control will not be accurate. In addition, given the <u>qualitative or semi-quantitative characteristic of the clinical tests/scales</u>, they may not be sensitive enough to detect some fallers with fairly good physical condition. For example, one recent study reported that the Mini-BESTest score could not differentiate older fallers from older non-fallers, due to the ceiling effects (Li et al., 2023). Given the limited approaches and the insufficient sensitivity of the clinical tests/scales that assess reactive balance control, this PhD project attempts to probe the intrinsic mechanisms of reactive balance control among the fall-prone people and enhance fall-risk assessment.

1.5.2 Instrumented Assessment of Balance Performance

In laboratory settings, reactive balance control can be induced by the various perturbation systems, such as the waist-pull perturbation system, moving-platform system, and shoulder-impact system. With the machine, the perturbation magnitude/intensity and the starting time of perturbation are more controllable for inducing reactive balance control (**Chapter 2.3**). Instrumented measurements of balance performance include the quantification of whole-body postural sways, gait parameters or stepping characteristics, and responses of specific joints.

(1) Measurement of whole-body postural sways

The performance of reactive balance control can be quantified by the whole-body postural sways, i.e., the center-of-mass (CoM) displacements, the center-of-pressure (CoP) displacements.

Postural balance is typically described as the capacity to maintain the vertical projection of the CoM within the limits of base of support (BoS) (Winter et al., 1998). With the technology of motion capture and analysis, a person's whole-body segments can be reconstructed, and the CoM displacement can be calculated, serving as a kinematic measure to quantify the person's whole-body postural sway. Older fallers were reported to have a larger CoM displacement than older non-fallers following unexpected mediolateral moving-platform perturbations (Mortaza et al., 2014). In another study, it was found that older fallers exhibited a significantly larger lateral CoM velocity at step landing than older non-fallers after unexpected anterior waist-pull perturbations (Rogers et al., 2001).

By using the force plate(s), laboratory posturography provides an alternative to evaluate the whole-body postural sways through the kinetic measurement of CoP displacements. According to the latest systematic review, some CoP parameters measured during quiet standing not only serve as effective indicators for distinguishing between older fallers from older non-fallers but also demonstrate a strong ability to predict the risk of falls in older adults. (Quijoux et al., 2020). Regarding the reactive balance control, older adults who had prospective falling were observed with significantly later time to peak COP displacement than the older adults without prospective falling following unexpected mediolateral moving-platform perturbations, but no significant difference was found following the anteroposterior ones (Maki et al., 1994). In addition, regarding the average COP velocity following any direction of unexpected moving-platform perturbation, this research discovered that no alteration could distinguish the older adults who had prospective falls from those who did not (Maki et al., 1994). However, for other studies, they failed to identify significant differences in older fallers vs. older non-fallers regarding the onset latency/time to peak COP displacement during unexpected lateral shoulder-impact perturbations (Claudino et al., 2017) or in the peak/average COP velocity following unexpected waist-pull perturbations (Fujimoto et al., 2015).

Together, there have been inconsistent results regarding whether the response speed of whole-body postural sways following sudden balance loss can distinguish the older people with fall histories or high fall risks, a narrative review has been done in **Chapter 2.3** ***Literature Review: Response speed of reactive balance control in older adults with different fall histories or risks***.

(2) Measurement of gait or stepping characteristics

Some gait parameters and stepping characteristics have been reported to be able to identify the older people with fall histories or high fall risks. Spatiotemporal parameters during volitional walking like step length, gait speed, stride length and stance time variability have demonstrated high discriminative power in distinguishing older fallers from older non-fallers (Mortaza et al., 2014). In addition, a recent systematic review synthesized the parameters concerning <u>stepping characteristics</u> and concluded that stepping impairments in both volitional balance control and reactive balance control are notable fall-risk factors (Okubo et al., 2021). Specifically, in their meta-analysis, it was found that older fallers required more recovery steps in contrast to older non-fallers in reactive balance control (Okubo et al., 2021). Besides the number of steps following unexpected perturbations, prior studies have also investigated older fallers and older non-fallers' timing/speed of the reactive stepping responses, such as the step initiation time, step duration, step landing time, average step velocity (Bair et al., 2016; Fujimoto et al., 2015; Mille et al., 2013; Sturnieks et al., 2013; Tantisuwat et al., 2011). However, these speed measures of stepping responses have exhibited inconsistent abilities in identifying the fall-prone older people (**Chapter 2.3**).

(3) <u>Measurement of the biomechanical response at a specific joint</u>

Besides the stepping strategy, humans can have feet-in-place strategies (to maintain postural balance when the perturbation magnitude is not large. Examples are ankle strategy, hip strategy, and suspensory strategy. As summarized in a narrative review, older adults prioritize the use of proximal hip strategy for balance control, whereas young adults prioritize the use of distal ankle strategy for balance control (Osoba et al., 2019). Another study also provided the evidence of joint kinetics and revealed that age-related differences were observed in peak powers of hip and knee joints but not in ankle joints following suddenly forward balance loss induced by a moving-platform system (Hall & Jensen, 2002). In addition, that study found that the use of feet-in-place strategy was energetically more demanding than the stepping strategy during reactive balance control, which could explain why older adults also had more stepping responses that young adults (Hall & Jensen, 2002). However, as stated in **Chapter 1.4**, the age-related alterations cannot be the direct fall indicators. Prior studies have rarely investigated whether the kinematic/kinetic responses of specific lower-limb joints in reactive balance control could indicate the fall histories/risks among older adults. This PhD project therefore attempts to fill in this research gap.

1.6 Assessment of Lower-limb Muscle Function in Fall-prone Older People

Reactive balance control involves the pathways of sensory input (external perturbation feedback), central organization, and motor output (Osoba et al., 2019). Prompt and adequate muscle responses are indispensable in the motor output to produce an effective balance-control strategy (Osoba et al., 2019). Comparing the muscular function between older fallers and older non-fallers might provide more insights into the balance-control mechanisms and fall-prevention strategies.

1.6.1 Measurement of Lower-limb Muscle Strength in Fall-prone Older People

Weakness of lower-limb muscles has been reported as one cause for falls in older people (Aagaard et al., 2010), especially in several agonist/antagonist muscle groups that contribute to the maintenance of balance: hip flexors/extensors, hip abductors/adductors, and ankle plantarflexors/dorsiflexors (Calmels & Minaire, 1995). Attention is typically directed towards the grade of manual muscle testing (MMT) and the peak torque or force values observed during maximal voluntary contraction (MVC) tests as they signify the kinetic output capacity of the muscles in question. However, many daily activities do not actually demand maximal muscle strength. Additionally, generating maximal torque usually requires at least 300 ms (Palmer et al., 2017), which might not be timely in scenarios like recovering balance following an abrupt perturbation, where rapid force production within 100-200 ms is necessary to hold an upright posture (Palmer et al., 2017).

Therefore, in contrast to maximal strength, <u>rapid strength parameters</u> (i.e., rate of torque or force development, power) may have a greater importance on postural control performance to avoid falls. However, the rapid strength characteristics of lower limb muscles have showed inconsistent abilities across the relevant studies in differentiating older adults' fall histories/fall risks, and a systematic review with meta-analyses regarding this topic has been conducted in this PhD project **(see Chapter 2.2)**.

1.6.2 Measurement of Neuromuscular (EMG) and Muscle Contractile Activities (MMG) in Fall-prone Older People

<u>Neuromuscular activation measured by electromyography (EMG) does not equal muscle strength.</u> This has been reflected by the previous study, which found that older fallers showed a higher peak EMG amplitude of knee extensor but had smaller peak torque and peak power values than older non-fallers during the isokinetic strength test (Crozara et al., 2016). The increased activation in fallers could be attributed to a compensatory response to counteract their lower-limb muscle weakness or an impaired neuromuscular efficiency. The requirement for engaging more motor units can induce fatigue and consequently may increase the risk of falls (Crozara et al., 2016). Another study also found that older fallers had a higher co-contraction index (CCI) of the knee flexor-extensor pair than older non-fallers when they were performing the maximal step length task, whereas peak torque and power values at the knee joint did not differ significantly between the two older groups (Schulz et al., 2013). Together, these findings indicated that older fallers had the altered activation in lower-limb muscles in contrast to older non-fallers during the volitional strength or balance tasks, and the observed changes in neuromuscular activation might not align consistently with alterations in muscle strength.

Regarding the neuromuscular activation patterns during reactive balance control, some previous studies observed that fallers had the delayed lower-limb muscle response following sudden balance loss in contrast to older non-fallers (Claudino et al., 2017), while other studies did not observe the difference in neuromuscular activation (Thompson et al., 2018). In **Chapter 2.3,** the analysis of EMG timing parameters has been reviewed to examine whether there is an alteration in the speed of neuromuscular activation following sudden balance loss that can identify the older adults who had fall histories. Some limitations were identified to exist in the previous investigations. Most pertinent studies focused on <u>a limited selection of lower-limb muscles</u>, such as hip abductor (Claudino et al., 2017), knee flexor/extensor (Claudino et al., 2017; Ochi et al., 2014; Thompson et al., 2018), and/or ankle dorsiflexor/plantarflexor (Claudino et al., 2017; Ochi et al., 2014; Studenski & Chandler, 1991; Thompson et al., 2018). In addition, they typically analyzed <u>only one or a few timing and amplitude parameters of EMG signals in each investigation</u> (Claudino et al., 2017; Ochi et al., 2014; Studenski & Chandler, 1991; Thompson et al., 2018). This PhD project therefore has conducted a series of studies to resolve these limitations so that the more rooted neuromuscular response during reactive balance control can be identified to facilitate the early detection of fall-prone people. Firstly, this PhD project attempts to investigate the responses of an expanded set of hip, knee, and ankle muscles crucial for balance control. Secondly, this PhD project attempts to have a more comprehensive analysis of the amplitude and timing parameters of EMG signals. Amplitude parameters include the rate of EMG rise, peak EMG amplitude, and agonist-antagonist CCI. Timing parameters include the EMG onset latency, time to peak EMG amplitude, and EMG burst duration. Thirdly, in addition to studying neuromuscular responses between the older individuals who had fall histories and those who did not, this PhD project attempts to examine what neuromuscular responses following sudden balance loss can predict the future fall risks of older adults.

Mechanomyography (MMG) is a non-invasive and reliable tool that can measure the skeletal muscle vibration, i.e. low-frequency lateral oscillations in the active skeletal muscle fiber (Bos et al., 2016). Unlike EMG recording the electrical activities from the neuromuscular junction, MMG signals provide valuable information on muscle contractile properties. During voluntary or electrically evoked contractions, muscle fibers cyclically shorten and produce pressure waves. MMG is such a technique to detect these “sounds”. Thus, there are three other commonly used terms that refer to MMG, i.e., acoustic myography, vibromyography and phonomyography (Islam et al., 2012). The most common sensor for collecting MMG signals is the accelerometer.

The significance of introducing the measurement of muscle’s contractile activity is that the MMG together with the EMG signals and the force signals can depict the electromechanical delay more intrinsically. Electromechanical delay refers to the time lapse between muscle activation initiation and force generation onset. Within the electromechanical delay, the electrochemical component, i.e., the lagged time between EMG signal onset and MMG signal onset, mainly involves events like the excitation-contraction coupling and the transmission of pressure waves to skin. The mechanical component, i.e., the lagged time between MMG signal onset and force or torque signal onset, mainly indicates the duration required to eliminate slack in the muscle-tendon unit before the force transmits to the tendon insertion becomes effective. As these parameters can reflect the temporal sequence of neuromuscular activation, it’s reasonable to assume that they can reveal more on the reaction time of older people in different levels of fall risks. In this way, whether the fall risk in older population is correlated more with the neural deficits or with the contractile part might be better understood.

However, currently, there is a lack of studies that have proposed to measure the muscle contractile activities during reactive balance control to identify the fall-prone older people. The reason is that the current MMG or SMMG technology has been mostly constrained in the static situation (e.g., isometric contraction). In the dynamic situations, some recent studies have proposed methods to process the collected accelerometry data, trying to partition the MMG signal that is from muscle vibration and the signal that is from movement of body segment during walking (Lyu et al., 2022; Ma, Ling, et al., 2019; Plewa et al., 2018). Therefore, this PhD project also makes attempts to analyze the MMG signals collected from reactive balance control.

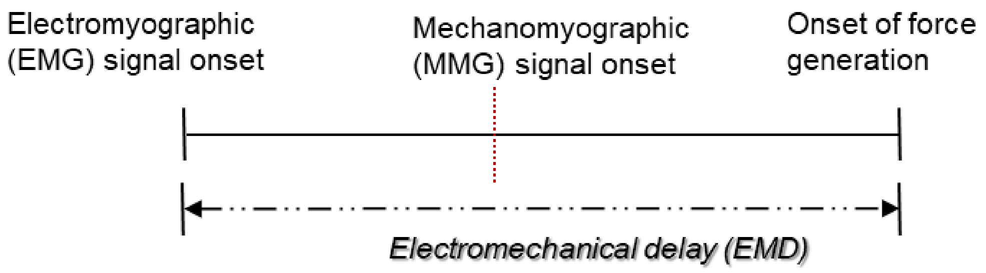

Figure 1-3. A schematic diagram of electromechanical delay.

1.7 Research Gaps

To summarize, there have been three major research gaps in previous relevant studies.

(1) The more intrinsic mechanisms of joint kinematics, joint kinetics, and muscle activities underlying reactive balance control have been unclear.

(2) The joint kinematics, joint kinetics, and muscle activities underlying reactive balance control that can differentiate fall histories have been insufficient and unclear.

(3) The joint kinematics, joint kinetics, and muscle activities underlying reactive balance control that can differentiate prospective falls have been uninvestigated and unknown.

(4) The joint kinematics, joint kinetics, and muscle activities underlying reactive balance control that can predict prospective falls have been uninvestigated and unknown.

1.8 Outline of This PhD Project

This PhD project therefore aims to progressively explore the mechanisms underlying reactive balance control from young adults to older adults, and identify the more in-depth biomechanical/neuromuscular factors that can indicate older adults' fall histories or fall risks. Specifically, this PhD project attempts to answer the research questions below:

(1) How do various hip, knee, and ankle muscles and joints react to sudden balance losses?
(2) What differences do fallers (i.e., older people with at least 1 fall in past one year) exhibit as compared to non-fallers in neuromuscular/kinetic/kinematic responses to sudden balance losses?
(3) What differences do prospective fallers (i.e., older people with 1fall in prospective one year) have as compared to prospective non-fallers in neuromuscular/kinetic/kinematic responses to sudden balance losses?
(4) What neuromuscular/kinetic/kinematic alterations in response to sudden balance losses can predict prospective falls in older adults?

To answer these research questions, this PhD project has the following objectives:

(1) To comprehensively analyze the timing and amplitude characteristics of hip/knee/ankle muscle activities (i.e., EMG signals and MMG signals) and hip/knee/ankle joint responses (i.e., joint powers, moments, and powers) during reactive balance control.
(2) To compare older fallers and older non-fallers' differences in hip/knee/ankle muscle activities and hip/knee/ankle joint responses during reactive balance control.
(3) To compare prospective fallers and prospective non-fallers' differences in hip/knee/ankle muscle activities and hip/knee/ankle joint responses during reactive balance control.
(4) To identify the abilities and accuracies of neuromuscular/kinetic/kinematic alterations during reactive balance control that can predict older people's prospective falls.

According to the findings of prior literature, it is hypothesized as below:

(1) The hip, knee, and ankle muscles/joints would have different response speeds following unexpected balance perturbations.
(2) Older fallers would prominently use hip strategies than older non-fallers following unexpected balance perturbations.
(3) Prospective fallers would prominently use hip strategies than prospective non-fallers following unexpected balance perturbations.
(4) The delayed and reduced neuromuscular/kinetic/kinematic responses following unexpected balance perturbations would predict older adults' prospective falls.

The overall PhD study has been conducted in four phases and can be categorized into two parts based on the methods to induce reactive balance (see **Figure 1-4**).

Study 1 (Moving-platform system validation): Exploring the response speed and sequence of eight major lower-limb muscles/joint motions following unexpected translational moving-platform balance perturbations in healthy young adults (To bridge the research gap 1. See also **Chapter 3**).

Study 2 (Pilot study on moving-platform reactive balance responses in older adults): Exploring neuromuscular and kinematic factors that are closely associated with falls by comparing reactive balance control following unexpected translational moving-platform perturbations in older fallers vs. older non-fallers (To bridge the research gap 1&2. See also **Chapter 4**).

Study 3 (Waist-pull system validation): Exploring the response speed and sequence of eight major lower-limb muscles/joint motions/joint moments/joint powers following unexpected waist-pull balance perturbations in healthy young adults (To bridge the research gap 1. See also **Chapter 5**).

Study 4 (Main study on waist-pull reactive balance responses in older adults): Exploring neuromuscular and biomechanical factors that are closely associated with falls, by comparing reactive standing balance control following unexpected waist-pull perturbations in older fallers vs. older non-fallers and by tracking their falls within future 1 year (To bridge all research gaps. See also **Chapter 6**).

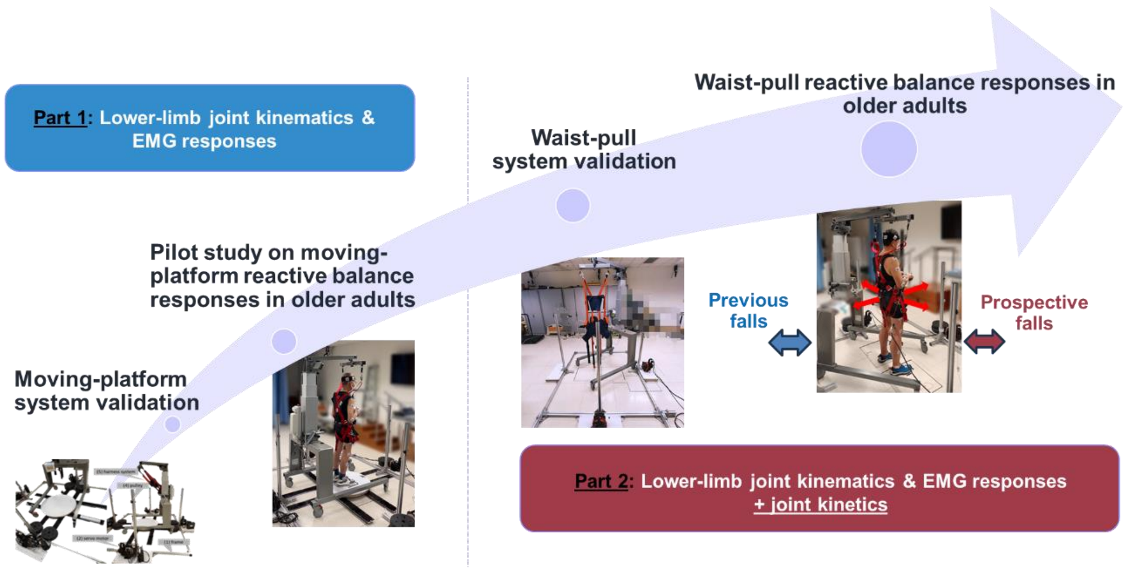

Figure 1-4. Outline of this PhD project.

To summarize, this PhD project attempts to identify the more <u>rooted factors</u> of falls in community-dwelling older adults with the <u>minimal bias of aging and other confounders of falls</u>, thus guiding a more effective fall-prevention program for future implementation in the long run. Towards this goal, the customized perturbation system to induce unexpected translational moving-platform perturbations has been firstly validated and synchronized with the motion capture system and multi-channel EMG and MMG system (**Part 1**). The moving-platform perturbations simulate older adult's taking a bus in daily life. A thorough examination of kinematic, neuromuscular activation, and muscle contractile (i.e., lateral vibration) characteristics has been tried to delve into the intrinsic mechanisms of reactive balance control in young adults (Study 1) and in older fallers vs. older non-fallers (Study 2). However, the experimental set-up of moving-platform system did not incorporate the use of force plates, making kinetic data unavailable. Considering the importance of lower-limb power to differentiate/predict older adults' falls (**Chapter 2**) (Zhu, Zuo, et al., 2025), the perturbation system was modified to enable the inducing of waist-pull perturbations and enable the use of force plates. This PhD project therefore had more focus on the waist-pull responses (**Part 2**). Similarly, the perturbation system has been verified to reliably induce waist-pull perturbations, which simulate the examiner's pull test for assessing reactive balance control in hospitals or clinical settings (Study 3). Then in the main study, a cohort of older fallers and older non-fallers with justified sample size were recruited and their neuromuscular/kinetic/kinematic responses following waist-pull perturbations were compared. After prospective fall tracking, regression analyses were conducted to identify what neuromuscular/kinetic/kinematic parameters during reactive balance control can predict older adults' prospective falls with confounders (i.e., age, sex, body mass index, fall history, balance performance evaluated by Mini-BESTest, physical activity level, degree of fear of falling) adjusted. Diagnostic accuracy tests have been further conducted to determine the accuracies of these neuromuscular/kinetic/kinematic parameters in predicting older adults' prospective falls (Study 4).

1.9 Significance of This PhD Project

Around 1 in 3 older adults falls each year worldwide. Fall-related injuries and deaths are major health problems worldwide and burden society heavily. Older adults who had fall histories (i.e., older fallers) have odds of future falls. Additionally, multifactorial fall-prevention programs are recommended; however, they have been reported to be inadequately successful in reducing fall incidences among the older fallers.

Identifying the more in-depth deficits in balance-control strategies among the fall-prone older people and targeting these deficits may potentially enhance the effectiveness of fall-prevention management. This PhD project therefore has firstly delved deeper into the mechanisms of reactive balance control among the young adults, then retrospectively recruited older fallers and older non-fallers to conduct intrinsic and extrinsic evaluations of reactive balance control among these participants, and finally followed up their fall incidences prospectively to identify the fall-risk factors. The findings of this PhD project could aid in a more precise assessment of fall risk and improve the formulation of targeted fall prevention strategies. To be specific:

By investigating the kinematics and kinetics of hip/knee/ankle joints together with the neuromuscular and contractile properties of multiple lower-limb muscles, the more intrinsic mechanisms underlying reactive balance control can be better understood. These comprehensive analyses can function as multimodal datasets encompassing healthy young adults, older adults at low risk of falls, and older adults at high risk of falls concerning reactive balance control, paving the way for the development of robotic assistive technologies aimed at fall prevention.

By comparing the age-matched older fallers and older non-fallers with relatively good health, the fall-related factors identified in this project might be less susceptible to bias of age and disease, and the targeted interventions may potentially be more successful in mitigating falls among the older adults. By using the synchronized multi-sensing technologies, even the subtle deficits of reactive balance control shall be more easily distinguished, making the detection of more in-depth fall indicators possible.

By examining relationships between the neuromuscular/kinetic/kinematic responses in reactive balance control and the older people’s prospective falls, this project can identify the in-depth alterations underlying reactive balance control that can indicate fall risks in older adults. By determining the diagnostic accuracies and cut-point values of these neuromuscular/kinetic/kinematic parameters to classify fall risks, their clinical utility in fall-risk assessment can be revealed, which may enhance a more sensitive fall-risk assessment among the community-dwelling older adults.

Together, this PhD project highlights its in-depth exploration of neuromuscular/biomechanical mechanisms following sudden balance losses in fall-prone people. The findings of this project will enhance the current assessment of reactive balance control, aiding in the early detection of older adults who have high fall risks and potentially offering clearer guidance for the implementation of targeted fall-prevention programs.

Chapter 2 Literature Review

2.1 Chapter Introduction

This chapter includes literature reviews on two research questions: **1)** Can the lower-limb rapid strength differentiate the community-dwelling older people who had fall histories or had high fall risks? and **2)** Can the response speed of reactive balance control differentiate the community-dwelling older adults who had fall histories or had high fall risks?

The first systematic review with meta-analysis was performed, considering the ability of rapidly generating adequate lower-limb strength is important in maintaining postural balance or preventing experimentally induced tripping. “Rapid strength” has been proposed to reflect this ability, i.e., rate of torque/force development, power. However, there was no consensus on whether the deficiency in lower-limb rapid strength could distinguish the community-dwelling older adults who had fall histories or had high fall risks. To bridge this research gap, this review searched six databases and included 20 relevant observational studies. The meta-analysis results showed that the overall lower-limb rapid strength was smaller in older adults with fall history (SMD = -0.41) and future falls (SMD = -0.21) when in contrast to non-fallers. The average leg-press power and peak sit-to-stand power could discern fall history and predict fall risks among older adults, so as the peak sit-to-stand power. The rate of torque development of a single lower-limb muscle group was unable to predict future falls, and it generally could not distinguish the fall history in older adults. These findings substantiated the quantitative measurements of entire lower-limb power to supplement the existing physical function tests for detecting older adults’ fall risks early, which may potentially facilitate early provision of interventions to prevent falls.

The second narrative review was conducted, considering that there was no consensus on whether the declined reactive balance speed could distinguish the community-dwelling older adults who had fall histories or had high fall risks. To bridge this research gap, this review searched five databases and included 13 relevant observational studies with 859 community-dwelling older adults. Preliminary qualitative analysis showed that the delayed initiation of reactive backward step and the delayed time to peak medial/lateral postural sway could predict the older adults' fall risks, as indicated by the prospective studies with good methodological qualities. The cross-sectional studies with poor to fair methodological qualities also supported the slower response speed following backward/medial/lateral sudden balance losses in older adults with fall histories, while it was the faster response following anterior loss of balance that may indicate the older adults' fall histories.

2.2 Systematic Review: Lower-limb rapid strength in older people with different fall histories or risks

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SYSTEMATIC REVIEW

Open Access

Association of lower-limb strength with different fall histories or prospective falls in community-dwelling older people: a systematic review and meta-analysis

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Abstract

Background Fall is a major health threat to older people. The lower limb power and rate of torque or force development (RTD or RFD) are prominently affected by aging and are crucial for maintaining postural balance. However, there have been inconsistent findings regarding the association of such aspects of lower-limb strength with falls among community-dwelling older adults. Comprehensive synthesis and appraisal are needed to examine what deficits in lower-limb rapid force generation could identify the fallers (i.e., those with a fall history or prospective falls).

Methods This systematic review searched six databases, including PubMed, Web of Science, EMBASE, Scopus, CINAHL, and Cochrane CENTRAL. Meta-analysis was conducted to aggregate standardized mean differences (SMD) or odds ratios (OR). The quality of evidence regarding each strength parameter's ability to identify fallers was assessed using the GRADE approach.

Results Twenty observational studies with 8,231 community-dwelling older adults were included (mean age: 73.5 years; male to female ratio: approximately 6:1). Moderate quality of evidence showed that the lower average leg-press power (SMD & 95% CI: -0.17 [-0.23, -0.12]; OR & 95% CI: 0.84 [0.79, 0.89]) and lower peak sit-to-stand power (Cohen's $d = 0.41$) could predict prospective falls in older adults, especially the injurious/recurrent falls. Low quality of evidence showed that the lower peak sit-to-stand power could also discern fall history (SMD & 95% CI: -0.58 [-0.96, -0.20]). Conversely, low to very low quality of evidence showed that the RTD of a single muscle group could not predict prospective falls and was generally unable to identify fall history in older adults.

Discussions and Conclusion The decline of entire lower-limb power appears a good indicator of prospective falls in community-dwelling older adults. Tests of entire lower-limb power required the cumulative and coordinated contractions of more leg muscles, possibly explaining why they could identify the fallers whereas the RTD or power of a single muscle group could not. Future studies are warranted to determine cut-point values of the entire lower-limb power measurements in fall-risk assessment and explore rapid force generation of a single muscle group in predicting the injurious falls among older adults.

Trial registration Registration No.: CRD42021237091.

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2.2.1 Introduction

Weakness of lower-limb muscles has been widely reported as one cause for falls in community-dwelling and institutionalized older people (Ahmadiahangar et al., 2018; Ikezoe et al., 2009; Yang et al., 2018). To evaluate the degree of weakness, both the maximal strength and the rapid strength characteristics of lower-limb muscles can be tested. Manual muscle testing (MMT) grades the muscle strength from 0 to 5, while the equipment of dynamometry enables a more quantitative record of peak joint torque or peak muscular force during maximal voluntary contractions (MVC) of subjects. These measurements focus on the ability of maximal kinetic output of older people. However, many daily-life activities for older people do not entail the generation of maximal lower limb strength. Rather, developing the muscle force in a very limited time seems to matter more for older people when they confront the unexpected perturbations suddenly in daily life. Occasions like standing on the suddenly accelerating transportation vehicles, unanticipated tripping and/or slipping highly demands the prompt recovery of postural balance, which further implies that the fast generation of adequate lower limb strength would be necessary for older people to avoid the real falls. Several rapid strength characteristics can be used to evaluate both the speed and the amplitude of force production. Muscle power is usually defined as the maximal product of the motion speed and the generated force. Rate of torque development (RTD) or rate of force development (RFD) refers to the average rise of torque or force over a rather short period (Δtorque/Δt or Δforce/Δt). Concerning the above-mentioned relationship between muscle weakness and risk of falls, it is reasonable to assume that rapid strength characteristics of lower limb muscles could be applied in assessing the fall risks of older people.

Decline in rapid strength of lower limb muscles with the advancing of age has been well-documented. This is also evidenced by the increasingly pronounced sarcopenia (loss in muscle quantity and quality) with ages. Varesco et al. found that the normalized RFD at 0-50 and 0-200 ms of maximal isometric knee extension was lower in old men than that in young men (Varesco et al., 2019). Lamoureux et al. had the similar finding that the older people (aged≥70) generated significantly smaller RTD in knee extensors compared with old people at younger age (aged<70) (Lamoureux et al., 2001). For the ankle muscular function, young adults performed better in rapid torque production than older people during isometric MVC tasks of plantarflexion (King et al., 2012; Rice et al., 2019) and dorsiflexion (Cogliati et al., 2020). In addition, Palmer et al. and Inacio et al. reported the reduced RTD variables of hip abductor (Inacio et al., 2019) and extensor (Palmer et al., 2017) voluntary contractions among the older adults. Several other studies have investigated further about the association between the rapid strength characteristics in lower limbs and the balance maintaining capability, as indicated by parameters like maximum recoverable lean angle during tether-release tests (Ochi et al., 2020), peak center of pressure (CoP) displacements during perturbed standing on a moving platform (Zemkova et al., 2017), and peak slip velocity during experimentally induced slips (Wyszomierski et al., 2009). However, the age-related RTD decrease and the association between rapid strength and postural balance were not sufficient to imply the fall risks of older adults.

To more directly elucidate the role that rapid strength plays in differentiating or predicting elders' fall risks, other research inquired into the correlation of rapid strength characteristics with the previous or future falls. However, due to the varied study objectives, only one or two muscles were measured for the rapid torque characteristics (Bruyneel et al., 2018; Gafner et al., 2018; Gafner et al., 2020; Kamo et al., 2019; Morcelli, LaRoche, et al., 2016; Morcelli, Rossi, et al., 2016; Palmer et al., 2017). A few studies even reached contrasting conclusions regarding the capability of specific lower-limb muscles to differentiate older fallers and non-fallers. Thus, the debate persisted on whether rapid strength parameters could distinguish the older people who had fall histories or had high fall risks.

A synthesis and critical appraisal of published articles on lower-limb muscle's rapid and maximal strength characteristics is needed to provide more evidence which muscle characteristic can effectively identify the older adults who had fall histories or had high fall risks. To the best knowledge of the authors, none of the published review articles have focused on muscle rapid strength parameters in distinguishing older adults' risks of falls.

Given the above, the aim of this systematic review and meta-analysis was to answer the questions of: (1) whether the lower-limb rapid strength (including power, RTD, and RFD) could effectively identify the community-dwelling older people who fall histories or had high fall risks; and (2) which rapid strength parameter would show a better ability to quantify the fall risk. The identification of the essential lower-limb rapid strength parameters that can distinguish fall history and/or predict fall incidence is expected to facilitate the current physical function assessments for early detection of the older adults who are vulnerable to falls, which may potentially facilitate the early provision of relevant intervention to prevent falls.

2.2.2 Methods

2.2.2.1 Data source and search strategy

The review protocol was pre-registered in the International Prospective Register of Systematic Reviews (PROSPERO, registration No.: CRD42021237091). Following the PRISMA guidelines, two reviewers (R.T.L.Z. and J.J.J.Z.) undertook the literature search and screening. A complete PRISMA flow chart illustrates the searching strategy and the screened results (**Figure 2-1**).

To identify relevant studies, a three-step search strategy was employed (Ma et al., 2020). Step 1 involved an initial scan of PubMed, where titles and abstracts were reviewed to pinpoint relevant keywords, e.g., "old" AND "fall risk" AND "power" AND "lower limb". Step 2 was conducted by using all the identified keywords to search across six databases: the PubMed, Web of Science, EMBASE, Scopus, CINAHL, and Cochrane CENTRAL. The truncation operators on keywords were used to widen the search, and the Boolean search operators were used to combine the keywords in this review. There were no restrictions on the publishing date. Searching alerts were created to monitor the publication of articles until 31 May 2023. In step 3, reference lists of the identified articles were looked through to find additional relevant studies. The above searching procedures were run first by one reviewer (R.T.L.Z.), and re-run by a separate reviewer (J.J.J.Z.) prior to the screening process. Forward citation tracking was conducted to identify any relevant studies that were published subsequent to the included studies. The corresponding authors were also contacted via e-mails for any accepted relevant articles.

2.2.2.2 Study selection

The inclusion criteria were studies involving: (1) adults chronologically aged 60 years or older living in the community with family or independently; (2) quantitative measurements of rapid strength, i.e., power (in the unit of Watt or Watt·kg^{-1}), RTD (in the unit of Nm/s or Nm/s·kg^{-1}), or RFD (in the unit of N/s, kgf/s, N/s·kg^{-1}, or kgf/s·kg^{-1}), of a single lower-limb muscle group or the whole lower-limb muscles; (3) evaluations of the retrospective fall history or the prospective fall incidence; and (4) effect measures indicating the comparisons (e.g., mean difference), associations (e.g., risk ratio [RR], odds ratio [OR], hazards ratio [HR]), or diagnostic accuracy. There was no restriction on the study design. Exclusion criteria were studies that: (1) focused on older people living in the institutional settings (e.g., nursing homes, hospitals), or older people with a specific neuromuscular, orthopedic, cardiopulmonary or cognitive disease (e.g., stroke, Parkinson's disease, multiple sclerosis, fractures, diabetic foot); (2) measured the rapid strength of the upper-limb or trunk muscle; (3) assessed the fall risks indirectly, i.e., not based on the previous fall history or future falls, such as via the comparison between older and young participants or via balance tests; or (4) were review articles, conference papers, meeting reports (or proceedings), or not written in English.

The three-step literature search identified 574 publications (see **Figure 2-1**). Based on the selection criteria, two reviewers (R.T.L.Z. and J.J.J.Z.) reviewed the titles and abstracts of these publications. Based on the identified abstracts, the full texts were further read based on the identified abstracts to screen the eligible articles.

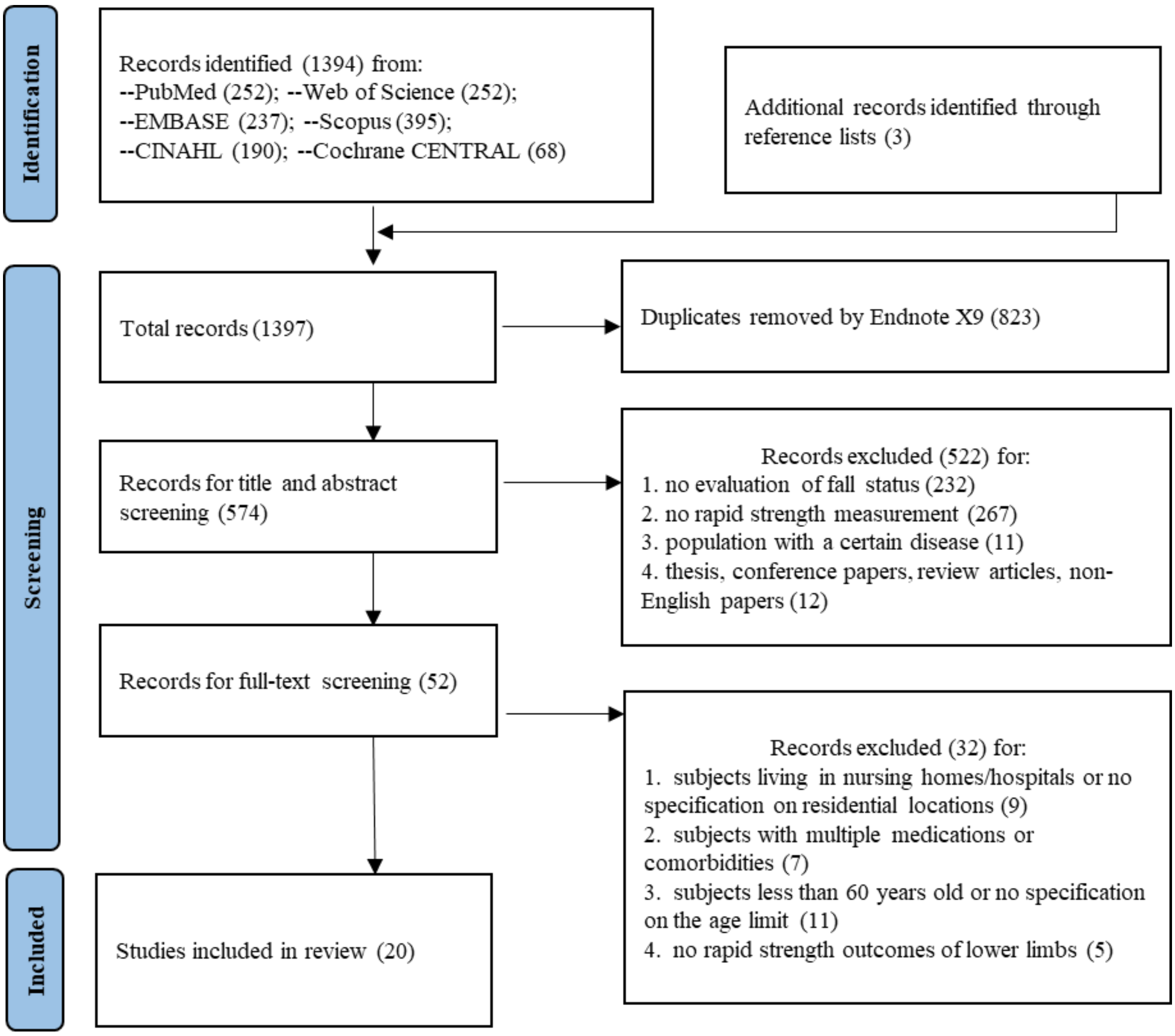

Figure 2-1. Flow chart of study identification and screening.

2.2.2.3 Data extraction

One reviewer (R.T.L.Z.) first extracted the information of study characteristics, participant characteristics, testing conditions, and the values of rapid strength parameters. If data was unavailable in the main text, the supplemental materials were screened for relevant data. If data was not available in the supplemental materials, the reviewer (R.T.L.Z.) contacted the authors via e-mails. All the data was recorded on Excel (Microsoft Office 365) sheets. Another reviewer (K.J.L.) checked the documented data against the original text to ensure the input data was correct.

2.2.2.4 Quality assessment

The 14-item Quality Assessment Tool for Observational Cohort and Cross-Sectional Studies was used to assess the methodological quality of each included study (National Heart Lung and Blood Institute, 2021). When one item was rated as “Yes”, it scored 1 point (Cunha et al., 2019). The item was given 0 point if it was rated as “No”, “not reported” or “not applicable” (Cunha et al., 2019). The Grading of Recommendations, Assessment, Development and Evaluation (GRADE) approach was used to evaluate the quality of evidence regarding each lower-limb rapid strength parameter (Schünemann et al., 2019).

Two reviewers (R.T.L.Z. and J.J.J.Z.) independently conducted the quality assessment. Disagreements over the rating results were first discussed between the two reviewers (R.T.L.Z. and J.J.J.Z.); if agreements persisted, a third reviewer (C.Z.H.M.) made the final decision.

2.2.2.5 Data synthesis and analysis

Meta-analysis was firstly conducted to pool the rapid strength parameters in various muscles and testing tasks to examine the overall effect of lower-limb rapid strength in identifying the older people’s fall history/risks, provided that two or more included studies had the same study design (Slimani et al., 2018; Torres-Costoso et al., 2020; Zhang et al., 2019). Then the meta-analysis was conducted separately for each rapid strength parameter, if two or more included studies had the same study design and the similar testing conditions (Higgins et al., 2022).

Random-effects inverse-variance models were employed to pool the standardized mean differences (SMD), i.e., Hedges' g, of rapid strength between two groups with different fall status in the Review Manager software (Version 5.4.1). Some studies detailed the rapid strength data in non-fallers (i.e., with no fall event), single fallers (i.e., with one fall event), and recurrent fallers (i.e., with two or more fall events). Meta-analysis was conducted separately to compare the lower-limb rapid strength in "fallers (single fallers + recurrent fallers) vs non-fallers", "single fallers vs non-fallers", "recurrent fallers vs non-fallers", and "recurrent fallers vs non-recurrent fallers (single fallers + non-fallers)". The Cochrane's formula was used to merge data from two participant groups into a single participant group, such as aggregating the data of single fallers and recurrent fallers into that of fallers (Higgins et al., 2022). The value of Hedges' g indicates the effect size of "very small" (0-0.2), "small" (0.2-0.5), "medium" (0.5-0.8), and "large" (>0.8) (Cohen, 1992).

2.2.3 Results

2.2.3.1 Types and methodological quality of included studies

This review encompassed twenty articles, all of which were observational studies. Eight of them were prospective cohort studies, and examined the relationship between lower-limb rapid strength and prospective falls (Atrsaei et al., 2021; Chan et al., 2007; Hsieh et al., 2023; Kemoun et al., 2002; Kera et al., 2020; Parsons et al., 2020; Porto et al., 2022; Winger et al., 2023). The remaining 12 cross-sectional studies measured the rapid strength among the older adults who had fall histories in contrast to those who did not.

The methodological quality evaluation indicated a generally moderate risk of bias for the included studies across the studies included, with scores ranging from 3 to 12 points (mean: 7.55 points; median: 7 points; full score: 14 points). The criteria of "clear statement of research question" (item 1; N = 20), "clearly defined exposure measures" (item 9; N = 20), and "clearly defined outcome measures” (item 11; N = 20) were met by all the included studies. Most studies clearly specified the characteristics of study participants (item 2; N = 18), applied the uniform eligibility criteria during participant recruitment (item 4; N = 17), examined the relationships between different levels of exposures and outcomes (item 8; N = 13), and adjusted for the impact of key confounders (e.g., sex, height, weight) on the exposure-outcome relationship (item 14; N = 12). High risk of bias commonly existed in the items of “exposure measured before outcome” (item 6; N = 8), “sufficient timeframe” (item 7; N = 8), “participation rate” (item 3; N = 6), “follow-up rate” (item 13; N = 5), “sample size justification” (item 5; N = 5), “assessors blinded” (item 12; N = 1), and “exposure assessed more than once” (item 10; N = 0).

2.2.3.2 Participants’ demographics and fall status

A total of 8,231 older adults were involved (**Table 2-1**). The sample sizes in the included studies varied from 15 (Palmer et al., 2015) to 5,995 (Chan et al., 2007). The mean age of the older people for each included study spanned from 66 to 80 years. The male to female ratio of the included participants was approximately 6:1. All the included older participants lived in the community and/or lived independently (99.15%) or were specified as healthy (0.85%).

Older fallers (n = 2,058) accounted for approximately 1/4 of all the included older participants (**Table 2-1**). Regarding the definition of “fall”, 13 studies clearly defined it as the event that resulted in a person coming to rest unintentionally on the ground or other lower level (Bento et al., 2010; Cheng et al., 2014; Crozara et al., 2016; Crozara et al., 2013; Dietzel et al., 2015; Ejupi et al., 2017; Kemoun et al., 2002; Palmer et al., 2015; Porto et al., 2022; Ribeiro et al., 2012; Skelton et al., 2002), while the other studies did not outline the definition. As the study designs of the included studies varied, “fallers” in this review referred to participants with fall event(s) that happened either before or after the measurement of lower-limb rapid strength.

The eight prospective cohort studies monitored the future fall incidence via the researchers' monthly telephone calls with participants (Porto et al., 2022) or researchers' questionnaires tri-annually (Chan et al., 2007) or triennially (Winger et al., 2023), the participants' monthly calendar records (Atrsaei et al., 2021; Hsieh et al., 2023) or yearly recalls (Kera et al., 2020; Parsons et al., 2020; Winger et al., 2023), or both the participants' dairy records and the researchers' bimonthly telephone calls (Kemoun et al., 2002). The follow-up period was within 1 year (Atrsaei et al., 2021; Hsieh et al., 2023; Kemoun et al., 2002; Kera et al., 2020; Porto et al., 2022), 2 years (Parsons et al., 2020), 4.5 years (Chan et al., 2007), or 9 years (Winger et al., 2023). The remaining cross-sectional studies retrospectively evaluated the fall histories in older participants, and defined "fallers" as participants experiencing at least one fall in the past one year (Bento et al., 2010; Cheng et al., 2014; Crozara et al., 2016; Crozara et al., 2013; Dietzel et al., 2015; Ejupi et al., 2017; Kamo et al., 2019; Perry et al., 2007), at least three falls in the past one year (LaRoche et al., 2010; Skelton et al., 2002), or at least one fall in the past six months (Ribeiro et al., 2012).

2.2.3.3 Testing tasks and equipment for measurement of lower-limb rapid strength

Older adults' rapid strength has been evaluated in the diverse tests regarding a single lower-limb muscle group and regarding the whole lower-limb muscles. (**Table 2-1**). More details on the testing tasks and devices are described as follows:

<u>Strength tests for a single lower-limb muscle group</u>

Eight included studies evaluated older fallers and older non-fallers' rapid strength of a single lower-limb muscle group. The maximal voluntary isometric contraction (MVIC) tasks were the most frequently used (Bento et al., 2010; Crozara et al., 2013; Kamo et al., 2019; LaRoche et al., 2010; Palmer et al., 2015; Porto et al., 2022), followed by the isokinetic (Crozara et al., 2016) and the submaximal concentric contraction tasks (Ribeiro et al., 2012). The measuring devices involved dynamometers, load cells or force sensors. Participants were instructed to exert force or accomplish a certain joint motion both as hard and as fast as possible. Almost all the major lower-limb muscle groups have been evaluated, including hip flexors/extensors (Bento et al., 2010; Palmer et al., 2015; Porto et al., 2022), hip abductors/adductors (Bento et al., 2010; Porto et al., 2022), knee flexors/extensors (Bento et al., 2010; Crozara et al., 2013; Kamo et al., 2019; LaRoche et al., 2010; Porto et al., 2022), ankle dorsiflexors/plantarflexors (Bento et al., 2010; Crozara et al., 2013; LaRoche et al., 2010; Porto et al., 2022). The RTD parameters were analyzed during the MVIC tasks, while the average power was measured during the isokinetic and the submaximal concentric contraction tasks.

One study also evaluated the lower-limb joint power during the favored-paced walking tasks in fallers vs non-fallers (Kemoun et al., 2002). Participants were instructed to walk at their self-selected pace, and the three-dimensional motion capture system with cameras and force plate(s) was used to capture the kinematic and kinetic data. Based on the inverse dynamics, the hip/knee/ankle joint power in a gait cycle was estimated. Noted that the term, “joint power”, was frequently used in gait analysis. It was the product of the net torques about a joint and the angular velocity of the joint (Richards, 2018). Therefore, the joint power involves the contributions of the muscle power of multiple muscle groups that cross the joint (Neil J. Cronin et al., 2013).

<u>Strength tests for the whole lower-limb muscles</u>

Older people’s rapid strength of the entire lower limb(s) was evaluated during the leg-press tasks, sit-to-stand tasks, stand-to-sit tasks, and jumping tasks. During the leg-press task, the Nottingham Power Rig was used to measure the average power of leg extensors as the participant was instructed to push the pedal down as hard and fast as possible using one leg (Chan et al., 2007; Hsieh et al., 2023; Perry et al., 2007; Skelton et al., 2002; Winger et al., 2023). During the sit-to-stand task, a force plates (Cheng et al., 2014; Dietzel et al., 2015; Kera et al., 2020) or wearable accelerometer (Atrsaei et al., 2021; Ejupi et al., 2017) was used to measure the peak/minimum/average power or the rate of ground reaction force development (RFD). Participants were instructed to stand up until reaching the full knee extension, without any help from their hands or arm support during the movement. In addition, one study analyzed the power parameters of lower-limb muscles during the stand-to-sit process when the participant was performing the five-time sit-to-stand test (Atrsaei et al., 2021). During the jumping test, the peak power was evaluated in older adults as they were instructed to stand on the force plate, bend knees, swing arms, and jump as high as possible (Dietzel et al., 2015; Parsons et al., 2020).

2.2.3.4 Rapid strength parameters to predict the fall risks

Six included prospective cohort studies were eligible for the meta-analysis of overall lower-limb rapid strength (**Figure 2-2**). The synthesized result of various lower-limb muscles in various testing tasks showed that older adults with future fall incidences had significantly smaller lower-limb rapid strength at baseline than those without future fall incidences (SMD = -0.21, I^2 = 0%). For each individual rapid strength parameter, the meta-analysis result showed that the average leg-press power was significantly smaller at baseline in fallers than non-fallers (SMD = -0.17, I^2 = 0%; see **Figure 2-3**).

Apart from the comparisons of rapid strength in fallers and non-fallers, predictive models were used to examine the causal associations between the lower-limb rapid strength and the prospective fall incidence. There was moderate quality of evidence regarding the associations between the future fall incidence and the average leg-press power, sit-to-stand power parameters, or stand-to-sit power parameters (see **Table 2-2**). By using generalized estimating equations, older men with the larger average leg-press power were found to have lower fall risks during a follow-up period of 4.5 years (Chan et al., 2007) and lower risks of injurious falls over a follow-up period of 9 years (OR = 1.19, 95% CI: 1.12-1.26) (Winger et al., 2023). In contrast, by using the logistic regression model, Hsieh et al. (2023) observed no significant association between the average leg-press power and the fall incidence during a follow-up duration of 12 months. By using logistic regression models that adjusted for sex, height and BMI, Atrsaei et al. (2021) demonstrated that the peak power value and the minimum power value in the sit-to-stand task could significantly predict the prospective fall incidences within the ensuing 12 months, but these power parameters during the stand-to-sit task could not. In addition, the diagnostic accuracy of the peak sit-to-stand power in differentiating older fallers from older non-fallers was analyzed (AUC = 0.62), although such discriminative ability was unsatisfactory.

There was low quality of evidence regarding the associations between the future fall incidence and the peak jumping power, the rate of ground reaction force (RFD) in the sit-to-stand test, or the RTD of a single lower-limb muscle group (see **Table 2-2**). By using the logistic regression model, Parsons et al. (2020) found that the greater peak jumping power indicated the decreased odds of falls (OR = 0.91, 95% CI: 0.85-0.98), while Porto et al. (2022) reported that none of the RTD values of the hip, knee, or ankle muscles during the MVIC task could significantly predict the future falls within a follow-up duration of 2 years. Using the Cox proportional hazards regression analysis, Kera et al. (2020) revealed that the RFD value during the sit-to-stand task was unable to significantly predict the future fall incidence within the ensuing 1 year. Regarding the favored-paced walking task, very low quality of evidence suggested that there was no notable difference in the estimated peak power at the hip, knee, or ankle joint at baseline between people with and without future fall incidence within the 1-year follow up (Kemoun et al., 2002) (see **Table 2-2**).

2.2.3.5 Rapid strength parameters to identify the fall history

All the included cross-sectional studies were eligible for the meta-analysis of overall lower-limb rapid strength (**Figure 2-2**). The synthesized rapid strength value of various lower-limb muscles in various testing tasks could significantly differentiate older fallers vs non-fallers (SMD = -0.41, I^2 = 0%), recurrent fallers vs non-recurrent fallers (SMD = -0.36, I^2 = 0%), and recurrent fallers vs non-fallers (SMD = -0.40, I^2 = 0%). There was no significant difference in overall lower-limb rapid strength in recurrent fallers vs single fallers or in single fallers vs non-fallers (see **Figure 2-2**).

For each rapid strength parameter (**Figure 2-3**), older fallers had significantly smaller values than non-fallers in the peak sit-to-stand power (SMD = -0.58, I^2 = 62%) and the average leg-press power (SMD = -0.49, I^2 = 0%). Generally, the RTD of a single lower-limb muscle group during the MVIC task could not significantly identify the community-dwelling older adults who had fall histories, except that the RTD of knee flexors could differentiate fallers from non-fallers (SMD = -0.57, I^2 = 0%) and the RTD of knee extensors could differentiate recurrent fallers from single fallers (SMD = -0.69, I^2 = 0%).

Effect size and quality of evidence for each lower-limb rapid strength parameter (including that was unavailable for meta-analysis) were listed in **Table 2-3**. Regarding the various rapid strength parameters to identify the fall history in older adults, the quality of evidence ranged from very low to low. The sample size concerning a lower-limb rapid strength parameter was commonly small and less than 400, which caused the “imprecision” and downgraded the quality of evidence.

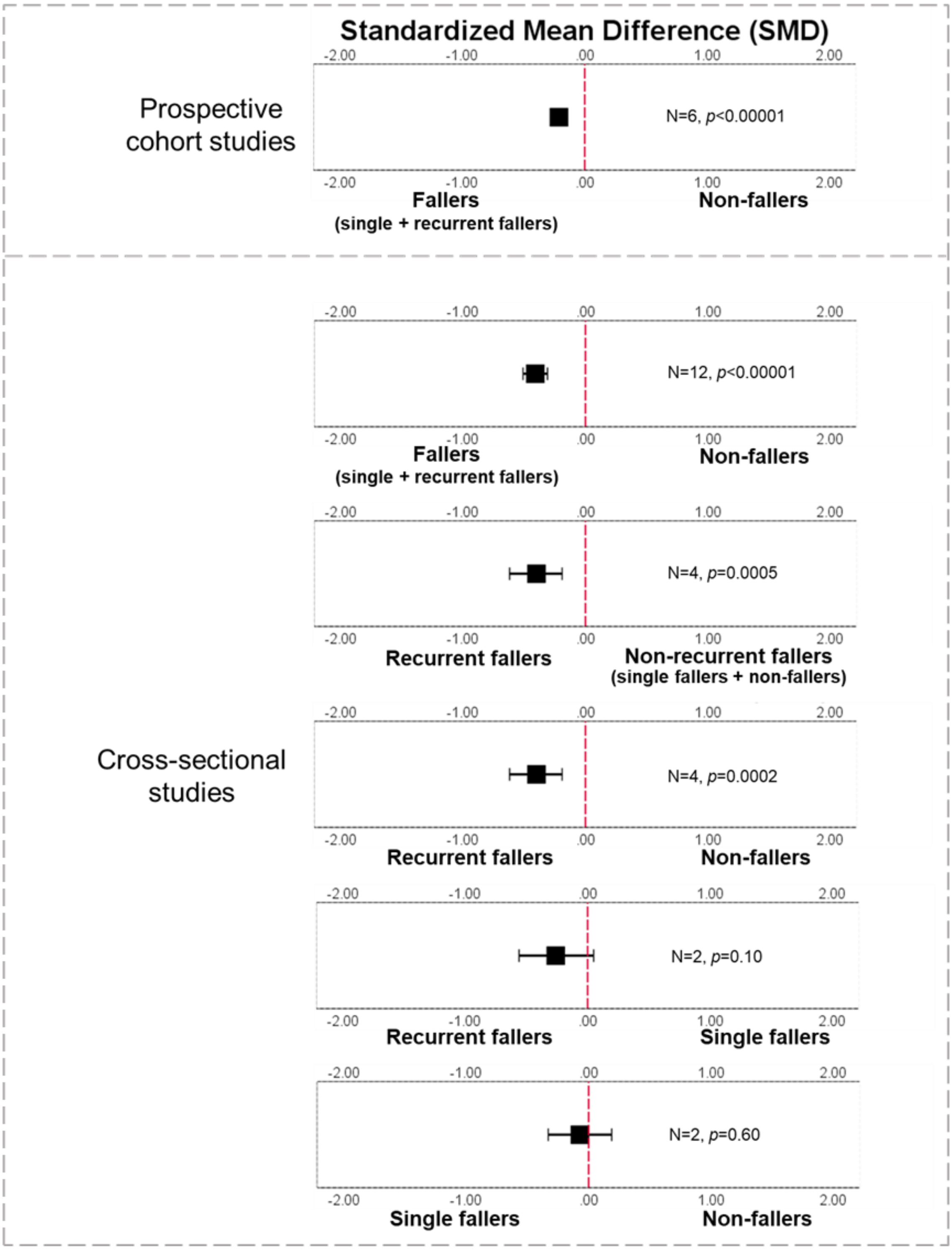

Figure 2-2. Meta-analysis for the comparison of overall lower-limb rapid strength in older fallers vs non-fallers, recurrent fallers vs non-recurrent fallers, recurrent fallers vs non-fallers, recurrent fallers vs single fallers, and single fallers vs non-fallers.

(⊢■⊣ : the SMD with 95% confidence intervals; RTD: rate of torque development; MVIC: maximal voluntary isometric contraction; N: number of studies.)

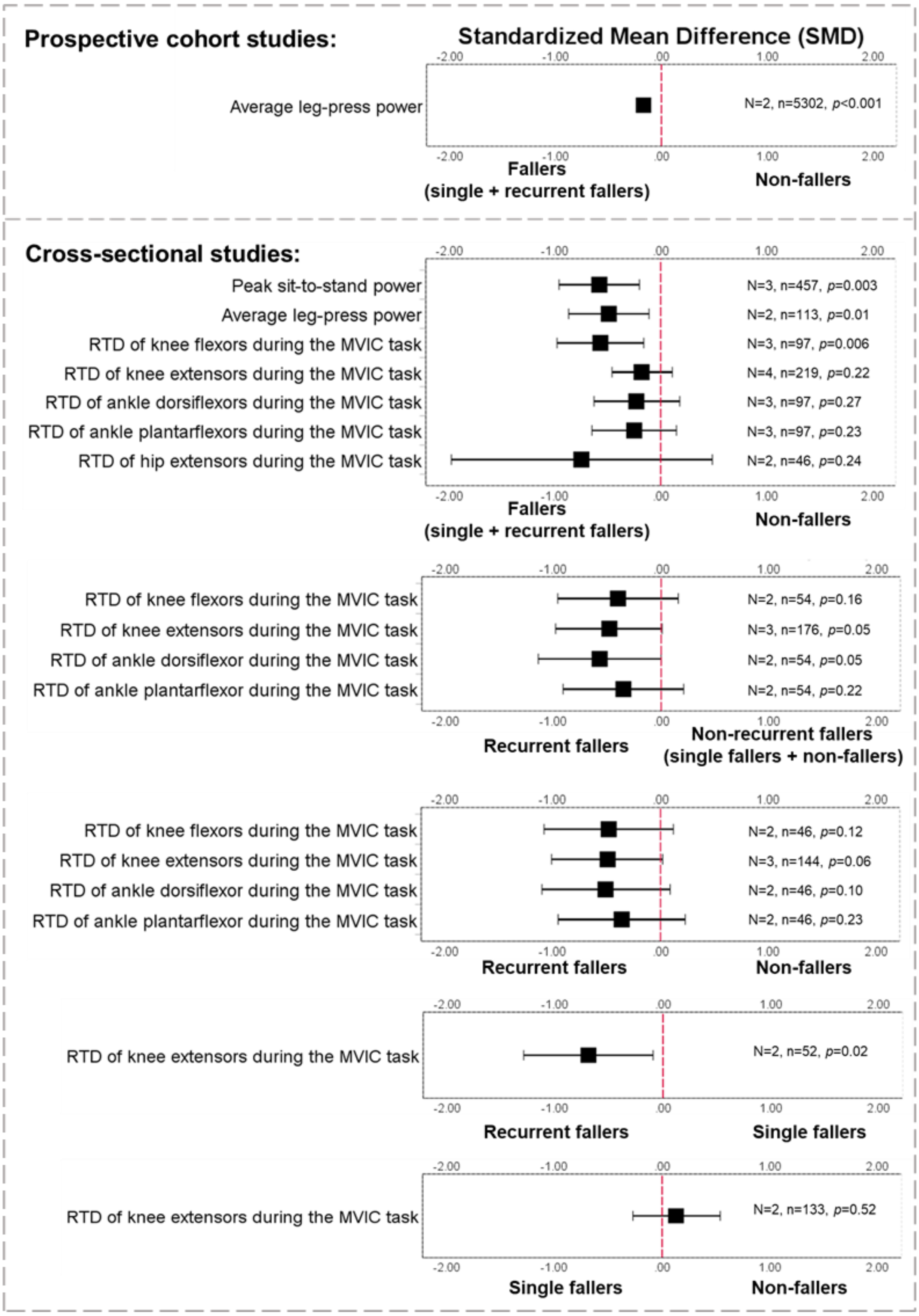

Figure 2-3. Meta-analysis for the comparison of each lower-limb rapid strength parameter in older fallers vs non-fallers, recurrent fallers vs non-recurrent fallers, recurrent fallers vs non-fallers, recurrent fallers vs single fallers, and single fallers vs non-fallers.

(: the SMD with 95% confidence intervals; RTD: rate of torque development; MVIC: maximal voluntary isometric contraction; N: number of studies; n: pooled sample size.)

2.2.4 Discussion

This systematic review and meta-analysis represent the first attempt to consolidate evidence concerning the ability of quantitative measurements of lower-limb rapid strength (power, RTD, and RFD) during muscle strength tests, sit-to-stand/stand-to-sit task, jumping task, and walking task in identifying the history of falls and/or the fall risks in community-dwelling older people. Overall, the lower-limb rapid strength could predict older adults' fall risks and was generally able to identify their different fall histories. Specifically, moderate quality of evidence showed that the average leg-press power and the peak sit-to-stand power could predict the fall risks, and low quality of evidence showed that the peak sit-to-stand power could identify the fall history. These synthesized findings can support the application of lower-limb power measurement for early detection of older people who are prone to falls, and may provide insights on the corresponding fall-prevention intervention for future practices.

2.2.4.1 Methodological quality of the included studies

The included studies showed generally moderate methodological quality. Most studies have attempted to improve the validity when examining the relationships between lower-limb rapid strength and falls among older people. They usually reported detailed characteristics of participants, used uniform eligibility criteria when recruiting older fallers and older non-fallers, and adjusted potential confounding factors (e.g., age-matched fallers and non-fallers, adjusted regression analysis). However, most included studies had the issues of adopting cross-sectional designs, and they had no sufficient long timeframes to show the causality between the lower-limb rapid strength and fall risks. The unjustified sample size, the convenience sampling method, and the retrospective evaluation of fall history (which was prone to recall bias) in most included studies were also the factors compromising the overall methodological quality.

2.2.4.2 Evidence on parameters to predict older adults' fall risks

The current evidence of this review supported that the power parameters of whole lower-limb muscles instead of the RTD parameters of a single lower-limb muscle group could predict the fall risks in community-dwelling older adults. Possible explanations for this difference are as follows.

Effects of the sample size and the follow-up duration for tracking prospective falls may need to be primarily considered. An example was that more than 5,000 older participants were followed up for 4.5 years (Chan et al., 2007) and 9 years (Winger et al., 2023) after the baseline measurement of average leg-press power, while only 100 older participants were followed up for 1 year after the measurements of RTD values in single lower-limb muscle groups (Porto et al., 2022). As fewer fall events and participants were tracked, the latter was more prone to the imprecision in effect estimates (i.e., larger confidence interval) than the former, which may be a reason of why the RTD of a single lower-limb muscle group could not significantly predict the fall incidence.

The effect size of a lower-limb rapid strength parameter in fall prediction may also be influenced by the measured muscles (the whole lower-limb muscles vs the single lower-limb muscle group), the nature of testing task (concentric vs eccentric), and the type of parameter (power vs RTD or RFD). Firstly, the leg-press task, sit-to-stand task, and jumping task all demand the contractions of multiple lower-limb muscle groups. Apart from the cumulative force exertions of multiple leg extensors (i.e., hip extensors, knee extensors, and ankle plantarflexors), these tasks may also require the coordinated contractions of other more leg muscles for postural balance, such as the co-contraction of ankle dorsiflexors and plantarflexors to stabilize the body position after standing from the chair (Atrsaei et al., 2021; Cheng et al., 2014). This may explain why they showed better abilities in detecting the older adult’s fall risks than the rapid strength measurement in a single muscle group. Secondly, the lower-limb rapid strength evaluated in the concentric contraction tasks appeared to be more sensitive to the fall risks in older adults. The capability of quickly generating adequate force is essential for the task demanding concentric strength to accelerate body segments and overcome gravity (Atrsaei et al., 2021; Parsons et al., 2020). For those requiring the eccentric control (e.g., stand-to-sit task) or those not demanding the rapid force generation (e.g., favored-paced walking), older adults could therefore demonstrate the similar lower-limb rapid strength values even if they had different fall risks (Atrsaei et al., 2021; Baltasar-Fernandez et al., 2021; Kemoun et al., 2002). Thirdly, even in the same sit-to-stand task, the peak power and the rate of ground reaction force development (RFD) have shown opposite abilities in fall prediction (Atrsaei et al., 2021; Kera et al., 2020). Although the two parameters both reflect the capability of explosive force generation, their definitions are different. The former was the largest rate of energy generated by lower-limb muscles, while the latter was the rate of force generated by lower-limb muscles. Nevertheless, it remains unclear why the peak power rather than the RFD during the sit-to-stand task could predict the fall risks in older adults. Further evidence is warranted.

In summary, via a single functional task that involves the coordination of multiple lower-limb muscles and/or the postural balance control, the power measurement was able to detect the community-dwelling older adults’ fall risks. This is promising, as a single conventional test for physical function assessment (e.g., the BBS or the TUG test), has usually shown insufficient ability in identifying fall risks (Lima et al., 2018; Omana et al., 2021; Schoene et al., 2013). Nevertheless, the peak sit-to-stand power, average leg-press power, and peak jumping power all showed small effect sizes in predicting the fall risks in older adults (**Table 2-2**). Future studies may be warranted to examine whether the combination of some tests for lower-limb power was better in fall risk prediction.

2.2.4.3 Evidence on parameters to identify older adults with fall history

Community-dwelling older fallers, especially recurrent fallers, had a greater decline in the overall lower-limb rapid strength than non-fallers (**Figure 2-2**). A previous meta-analysis reported that the lower-limb maximal strength was the fall-risk factors among the community-dwelling adults and the effect size was small (OR = 1.66, 95% CI: 1.20–2.29) (Moreland et al., 2004). Our meta-analysis result further indicated that older adults who had fall histories could also exhibit impaired ability of quickly generating adequate force in lower-limb muscles, and the overall lower-limb rapid strength showed a similar small effect size in differentiating fallers from non-fallers. Specifically, single fallers appeared to have no significant difference in the overall lower-limb rapid strength in contrast to recurrent faller or single fallers. One possible reason was that only two included studies with small sample sizes reported rapid strength data of the single faller group, and the difference in rapid strength in single fallers vs recurrent fallers or in single fallers vs recurrent fallers did not reach the statistically significant level. The other possible reason was that the older adults with only one previous fall may not indicate their poorer physical function or poorer balance capability, as those with two or more previous falls were more prone to future falls (Fabre et al., 2010).

Among the various rapid strength parameters, the peak sit-to-stand power showed the higher quality of evidence in differentiating older fallers from non-fallers although the heterogeneity was large (**Table 2-2**). Factors like the chair height, using the arms or not, and the instruction to participants may affect the performance of the sit-to-stand test (Watt et al., 2018). There were also diversities in the types of devices in measuring the power (force plates vs. accelerometer). The power measured by the force plate was calculated as the result of the vertical ground reaction force multiplied by the vertical velocity (Cheng et al., 2014; Dietzel et al., 2015), while that measured by the accelerometer was derived from the vertical net force (ground reaction force minus gravity) multiplied by the vertical velocity (Ejupi et al., 2017).

These factors may explain the large between-study heterogeneity of this meta-analysis. A previous systematic review found that the five times sit-to-stand test time (cut-off point: 12 seconds) could predict the community-dwelling older people's fall risks (Lusardi et al., 2017). In the current meta-analysis, it provided additional kinetic evidence to support the ability of the sit-to-stand performance to distinguish fall history.

Rapid strength parameters of the entire lower limb(s) and of a single lower-limb muscle group appeared to have different abilities in identifying the older adult's fall history (**Figure 2-3**). There was a clear trend for the RTD of a single lower-limb muscle group to be lower in the fallers than non-fallers, but this reached statistical significance only in knee flexors and knee extensors. By contrast, the peak sit-to-stand power and the average leg-press power were able to differentiate older fallers from older non-fallers. This underscores the cumulative effect of the relatively small force decrements across the individual muscles, which could result in the decline of entire lower-limb rapid strength required for functional movements (Perry et al., 2007). Measuring the rapid strength of entire lower limb(s) rather than a single lower-limb muscle group seemed more effective to distinguish the older fallers from non-fallers.

2.2.4.4 Impact and recommendations for future clinical practice

<u>Suggestions for rapid strength measurement to assess fall risks</u>

Measurement of lower-limb power seems necessary to be incorporated into the routine physical function assessment for detecting the community-dwelling older people who had high fall risks. This systematic review and meta-analysis have supported that the evaluation of the entire lower-limb power could be used to identify older adults who had fall histories or had high future fall risks. Specifically, moderate quality of evidence has indicated that the average leg-press power and the peak sit-to-stand power can predict older adults' future falls. Quantitative measurement of the entire lower-limb power during the leg-press test or the instrumented sit-to-stand test is therefore worthy of being promoted. It is expected to complement the current physical function assessments in clinical practice, such as the TUG test and the timed sit-to-stand test (CDC, 2019), and facilitate early detection of fall risks in older adults, especially in those community-dwelling ones with relatively good health.

Implications on future fall-prevention or intervention programs

Given that older people with a fall history or higher fall risks had generally poorer lower-limb power, relevant exercises should be prescribed to reduce the decline in older adults' muscle power and fall incidences. High-velocity resistance training, or power training, has been proposed as a more promising stimulus for improving older adults' physical performance (e.g., sit-to-stand time, walking speed) in contrast with the traditional resistance training (Daly, 2010). Based on a recent meta-analysis, there has been moderate-certainty evidence supporting that the balance and functional exercises (gait, balance, coordination, and functional task training) plus resistance exercises (resistance/power training) can reduce fall incidences among the community-dwelling older people by 34% compared with no exercise (Sherrington et al., 2019). Findings of this study could underscore the importance of encouraging older adults to participate in functional exercises for improving the entire lower-limb power to prevent falls.

2.2.4.5 Perspectives and outlook for future research

It is hard to recommend a cut-point value of lower-limb power to stratify the community-dwelling older people's fall risks, based on the current evidence. Although the peak sit-to-stand power and the average leg-press power have shown small effect sizes to identify older adults with a fall history or higher fall risks, only one included study conducted the diagnostic accuracy analysis (Atrsaei et al., 2021). Knowing the cut-point values can facilitate the judgement and more accurate stratification of fall risks in clinical practice. Future research is warranted to investigate the diagnostic accuracy (e.g., sensitivity, specificity, area under the curve) of lower-limb power parameters in fall-risk differentiation.

More portable devices with the real-time rapid strength values displayed can be developed to facilitate the clinician's judgement on an older client's risk of falls. Portable force plates (Kera et al., 2020) or wearable motion sensors (Atrsaei et al., 2021; Ejupi et al., 2017) have been popular in the rapid strength measurement (**Table 2-1**). They provide a convenient and continuous monitoring of lower-limb rapid strength, making the evaluation not confined to location and time. Such tools may thus be quite useful for the long-term prediction of fall risks in a wider older population. However, most of these portable devices have no real-time display of the rapid strength values to inform the clinicians or the clients. A more uniform standard on the raw data processing is expected to be reached so that a relatively standardized algorithm to calculate the rapid strength value can be included in the testing devices.

2.2.5 Limitations

There are some limitations for this systematic review and meta-analysis. Firstly, as only English articles were finally included, some relevant studies written in other languages might have been excluded. Secondly, this review involved only the community-dwelling older adults with relatively good health. Our results may not be broadly applicable to other older populations, such as individuals who live in nursing homes or have multiple comorbidities. Thirdly, when synthesizing the effect size of overall lower-limb rapid strength, the meta-analysis aggregated the rapid strength measurements of multiple muscles within a study. This method presumed that the measurements in separate muscles were independent of each other. If positive correlations existed, this method may have overestimated the precision of effect (Borenstein et al., 2021). Finally, the associations between lower-limb rapid strength parameters and falls are unavoidably affected by some confounding factors. The causes of falls are multifactorial. Some factors such as the environmental factors have hardly been adjusted in the regression models. This may partly explain why the power parameters have shown small effect sizes in detecting fall risks. Therefore, the quantitative measurement of lower-limb power alone cannot provide a full picture to identify the high fall risks in older adults.

2.2.6 Conclusion

This systematic review and meta-analysis examined whether the overall lower-limb rapid strength and the various rapid strength parameters (power, RTD, and RFD) of lower-limb muscles could effectively identify the community-dwelling adults with the fall history/fall risks. The overall lower-limb rapid strength could distinguish the fall history in older adults. Specifically, the average leg-press power and the peak sit-to-stand power could effectively predict the fall risks as well as distinguish the fall history, while the RTD of a single lower-limb muscle group could occasionally distinguish the older adults who had fall histories but could not predict the older adults' future fall risks. These findings suggest the need of incorporating the lower-limb power measurement into the routine physical function assessment to identify the older adults with higher fall risks early. Further investigations on the diagnostic accuracy of the lower-limb power parameters in fall-risk identification are needed to facilitate the clinical practice.

Table 2-1. Characteristics of included articles (N = 20).

Author (year) / Study design	Definition of fallers	Group	Sample size	Male/ Female	Age (year, mean ± SD)	Device	Task	Rapid strength parameter
Atrsaei et al. (2021) Prospective cohort	≥ 2 falls or 1 injurious fall in future 12 months	NF	350	163/187	74.9±1.4	Accelerometer	Five-time Sit-to-stand test	Peak, minimum, and average power value by multiplying acceleration, body mass and velocity.
		F	108	35/73	74.7±1.4			
Bento et al. (2010) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	13	0/13	67.6±7.5	Load cell	MVIC	RTD from the 20% to 80% of peak torque.
		SF	8	0/8	66.0±4.9			
		RF	10	0/10	67.8±8.8			
Chan et al. (2007) Prospective cohort	≥ 1 fall(s) in future 4.5 years	NF	5995	5995/0	73.7±5.9	Nottingham power rig	Leg press	Average power value calculated based on the final angular velocity of the power rig flywheel. The largest power of nine repetitive trials was used.
		F						
Cheng et al. (2014) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	35	23/12	75.2±6.4	Force plate	Sit-to-stand test	Peak power value by multiplying the vGRF and the vertical upward velocity of center of body mass.
		F	35	22/13	77.5±7.8			
Crozara et al. (2013) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	22	0/22	66.1±6.1	Isokinetic dynamometer	MVIC	RTD over 0-50, 50-100, 100-150, and 150-200 ms (onset: 5% of peak torque).
		F	21	0/21	69.6±7.2			
Crozara et al. (2016) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	23	0/23	66.0±6.0	Isokinetic dynamometer	Isokinetic contraction	Average power value by multiplying the mean torque and the angular velocity.
		F	22	0/22	70.0±7.0			
Dietzel et al. (2015) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	246	131/115	71.4±7.3	Force plate	1) Jumping; 2) Sit-to-stand test	Peak power value by multiplying vGRF and vertical upward velocity of center of body mass.
		F	47	16/31	74.2±7.5			

Continued on next page.

Author (year) / Study design	Definition of fallers	Group	Sample size	Male/ Female	Age (year, mean ± SD)	Device	Task	Rapid strength parameter
Ejupi et al. (2017) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	60	64/30	79.9±6.5	Accelerometer	Sit-to-stand test	Peak power value by multiplying acceleration, body mass and velocity.
		F	34					
Hsieh et al. (2023) Prospective cohort	≥ 1 fall(s) in future 12 months	NF	63	64/60	72.4±6.3	Nottingham power rig	Leg press	Average power value calculated based on the final angular velocity of the power rig flywheel. The largest power of ten repetitive trials was used.
		F	61		73.6±5.9			
Kamo et al. (2019) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	88	45/43	71.3±4.7	Hand-held dynamometer	MVIC	RTD over 0-200 ms (onset: 4 Nm).
		SF	24	13/11	71.2±3.7			
		RF	10	4/6	71.4±2.9			
Kemoun et al. (2002) Prospective cohort	≥ 1 fall(s) in the future 12 months	NF	38	26/12	66.7±4.9	Motion capture system	Favored-paced walking	Peak or minimum power value of a joint in a gait cycle.
		F	16	12/4				
Kera et al. (2020) Prospective cohort	≥ 2 falls in future 12 months	NF	433	170/263	72.3±6.0	Force plate	Sit-to-stand test	Rate of vGRF development between the onset of sit-to-stand motion and the peak force.
		F	23	11/12	72.7±6.7			
LaRoche et al. (2010) Cross-sectional	≥ 3 falls in past 12 months	NF	12	0/12	71.2±6.2	Isokinetic dynamometer	MVIC	The largest value of RTDs calculated for every 50 ms from onset (0.5 Nm) to 200 ms.
		F	11	0/11	71.3±5.4			
Palmer et al. (2015) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	9	0/9	71.4±7.0	Load cell	MVIC	RTD over 0–50 ms (onset: 4 Nm) and 100–200 ms.
		F	6	0/6	72.7±6.9			
Parsons et al. (2020) Prospective cohort	≥ 1 fall(s) in future 2 years	NF	129	/	75.1±2.5	Force plate	Jumping	Peak power value.
		F	40					

Continued on next page.

Author (year) / Study design	Definition of fallers	Group	Sample size	Male/ Female	Age (year, mean ± SD)	Device	Task	Rapid strength parameter
Perry et al. (2007) Cross-sectional	≥ 1 fall(s) in past 12 months	NF	44	15/29	75.9±0.6	Nottingham power rig	Leg press	Average power value calculated based on the final angular velocity of power rig flywheel. The largest power of at least six repetitive trials was used.
		F	34	4/30	76.4±0.8			
Porto et al. (2022) Prospective cohort	≥ 1 fall(s) in future 12 months	NF	72	19/53	66.7±4.5	Isokinetic dynamometer	MVIC	RTD over 30-80 ms, 200-250 ms (onset: 5% of peak torque).
		F	28	4/24	69.8±5.7			
Ribeiro et al. (2012) Cross-sectional	≥ 1 fall(s) in past 6 months.	NF	15	0/15	70.6±6.8	Extension machine + potentiometer	Concentric contraction	Power of knee extensors calculated from the force obtained in 1-RM test, lever arm, and angular displacement (not specifying peak or average power)
		F	11	0/11	68.5±4.3			
Skelton et al. (2002) Cross-sectional	≥ 3 falls in past 12 months	NF	15	0/15	74.0±6.3	Nottingham power rig	Leg press	Average power value calculated based on the final angular velocity of the power rig flywheel. The largest power of at least six repetitive trials was used.
		F	20	0/20	74.5±5.7			
Winger et al. (2023) Prospective cohort	≥ 1 injurious fall(s) in future 9 years	NF	3088	5178/0	72.9±5.6	Nottingham power rig	Leg press	Average power value calculated based on the final angular velocity of the power rig flywheel. The largest power of ten repetitive trials was used.
		F	2090		74.0±5.8			

Note: F: fallers; NF: non-fallers; SF: single fallers, i.e., older people experiencing one fall; RF: recurrent fallers, i.e., older people experiencing two or more falls. SD: standard deviation. RTD: rate of torque development; RFD: rate of force development; vGRF: vertical ground reaction force; RM: repetition maximum.

Table 2-2. Summary of effect size and quality of evidence for each rapid strength parameter in identifying fall risks.

Lower-limb power, RTD, or RFD parameter	Effect measure and size [95% CI]	No. of studies	Study design	Factors downgrading quality: Risk of bias	Inconsistency	Indirectness	Imprecision	Publication bias	Factors upgrading quality	Overall quality of evidence
Average leg-press power *	**Pooled SMD:** -0.17 [-0.23, -0.12] **RR:** Quartile 1: 1.00 Reference Quartile 2: 0.88 [0.81, 0.97] Quartile 3: 0.86 [0.77, 0.95] Quartile 4: 0.82 [0.73, 0.92]	3	-2 (No RCT)	No	No	No	No	NA	+1 (Dose-response gradient)	⊕⊕⊕○ **Moderate**
	***Cohen's d*:**									
Peak sit-to-stand power *	0.41	1	-2 (No RCT)	No	No	No	No	NA	+1 (Dose-response gradient)	⊕⊕⊕○ **Moderate**
Minimum sit-to-stand power *	0.40									
Average sit-to-stand power	0.08									
Normalized peak sit-to-stand power *	0.38									
Peak stand-to-sit power	0.28									
Minimum stand-to-sit power	0.19									
Average stand-to-sit power	0.07									
Normalized peak stand-to-sit power	0.25									
Peak jumping power *	**OR:** 0.91 [0.85, 0.98]	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	+1 (Dose-response gradient)	⊕⊕○○ **Low**

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Lower-limb power, RTD, or RFD parameter	Effect measure and size [95% CI]	No. of studies	Study design	Factors downgrading quality					Factors upgrading quality	Overall quality of evidence
				Risk of bias	Inconsistency	Indirectness	Imprecision	Publication bias		
RFD of entire lower limbs during the sit-to-stand task	**SMD:** -0.33 [-0.75, 0.09]	1	-2 (No RCT)	No	No	No	No	NA	None	⊕⊕○○ **Low**
	OR:	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	+1 (Dose-response gradient)	⊕⊕○○ **Low**
RTD of hip flexors during MVIC task	0.80 [0.40, 1.58]									
RTD of hip extensors during MVIC task	0.77 [0.33, 1.82]									
RTD of hip abductors during MVIC task	1.00 [0.26, 3.80]									
RTD of hip adductors during MVIC task	1.15 [0.34, 3.94]									
RTD of knee flexors during MVIC task	0.41 [0.07, 2.22]									
RTD of knee extensors during MVIC task	0.99 [0.59, 1.68]									
RTD of ankle dorsiflexors during MVIC task	1.35 [0.34, 5.25]									
RTD of ankle plantarflexors during MVIC task	0.82 [0.13, 4.90]									
Minimum hip joint power during the favored-paced walking test Minimum knee joint power during the favored-paced walking test Peak ankle joint power during the favored-paced walking test	**Median Difference**	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**

Note: The quality of evidence was rated based on the Grading of Recommendations, Assessment, Development and Evaluation (GRADE) criteria. * indicates the parameter could significantly predict the fall risks. RTD: rate of torque development; RFD: rate of ground reaction force development; MVIC: maximal voluntary isometric contraction; RM: repetition maximum; RCT: randomized controlled trial.

Table 2-3. Summary of effect size and quality of evidence for each rapid strength parameter in identifying fall histories.

Lower-limb power, RTD, or RFD parameter	Effect measure and size [95% CI]	No. of studies	Study design	Factors downgrading quality					Factors upgrading quality	Overall quality of evidence
				Risk of bias	Inconsistency	Indirectness	Imprecision	Publication bias		
Peak sit-to-stand power *	**Pooled SMD:** -0.58 [-0.96, -0.20]	3	-2 (No RCT)	No	No	No	No	NA	None	⊕⊕○○ **Low**
Average leg-press power *	**Pooled SMD:** -0.49 [-0.87, -0.11]	2	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**
RTD of knee flexors during the MVIC task * RTD of ankle dorsiflexors during the MVIC task RTD of ankle plantarflexors during the MVIC task	**Pooled SMD:** -0.57 [-0.98, -0.16] -0.23 [-0.63, 0.18] -0.25 [-0.65, 0.15]	3	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**
RTD of knee extensors during the MVIC task	**Pooled SMD:** -0.18 [-0.46, 0.11]	4	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**
RTD of hip extensors during the MVIC task	**Pooled SMD:** -0.75 [-1.98, 0.49]	2	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**
RTD of hip flexors during the MVIC task RTD of hip abductors during the MVIC task RTD of hip adductors during the MVIC task	**SMD:** -0.04 [-0.75, 0.67] -0.34 [-1.06, 0.38] -0.24 [-0.95, 0.48]	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**
Peak jumping power (* in female participants)	**SMD:** -0.47 [-0.78, -0.15]	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**

Continued on next page.

Lower-limb power, RTD, or RFD parameter	Effect measure and size [95% CI]	No. of studies	Study design	Factors downgrading quality					Factors upgrading quality	Overall quality of evidence
				Risk of bias	Inconsistency	Indirectness	Imprecision	Publication bias		
Average power of knee flexors during the isokinetic contraction task at 90 °/s *	**SMD:** -0.82 [-1.44, -0.21]	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**
Average power of knee flexors during the isokinetic contraction task at 120 °/s	-0.56 [-1.15, 0.04]									
Average power of knee extensors during the isokinetic contraction task at 90 °/s *	-0.80 [-1.41, -0.19]									
Average power of knee extensors during the isokinetic contraction task at 120 °/s *	-0.73 [-1.33, -0.12]									
Average power of ankle dorsiflexors during the isokinetic contraction task at 90 °/s *	-0.65 [-1.26, -0.05]									
Average power of ankle dorsiflexors during the isokinetic contraction task at 120 °/s	-0.08 [-0.67, 0.50]									
Average power of ankle plantarflexors during the isokinetic contraction task at 90 °/s	-0.22 [-0.81, 0.36]									
Average power of ankle plantarflexors during the isokinetic contraction task at 120 °/s	-0.31 [-0.90, 0.28]									
Power of knee extensors during the concentric contraction task at 70% 1-RM	**SMD:** 0.13 [-0.65, 0.91]	1	-2 (No RCT)	No	No	No	-1 (Sample size < 400)	NA	None	⊕○○○ **Very low**

Note: The quality of evidence was rated based on the Grading of Recommendations, Assessment, Development and Evaluation (GRADE) criteria. * indicates the parameter could significantly identify the different fall histories. RTD: rate of torque development; RFD: rate of ground reaction force development; MVIC: maximal voluntary isometric contraction; RM: repetition maximum; RCT: randomized controlled trial.

2.3 Narrative Review: Response speed of reactive balance in older people with different fall histories or risks

2.3.1 Introduction

Falls and the resulting injuries/deaths have negative impacts on older individuals around the world (World Health Organization, 2021). The early detection of the fall-risk factors that can be modified is important to enable provision of targeted fall-prevention training for older people, alleviating the significant socioeconomic burden of fall consequences.

The speed of balance control and gait is the fundamental component of fall-risk assessment. Considering that balance and gait disorders are the main contributors to falls in older adults (Salzman, 2010) and the fast-twitch fibers of skeletal muscles are more prominently affected with aging as compared to slow-twitch fibers (Gerstner et al., 2017; Reid & Fielding, 2012), assessing the balance/gait and the reaction speed simultaneously appears more effective in detecting older adults' fall risks. The world falls guidelines have used the 0.8 m/s of gait speed or the 15 s of accomplishing the Timed Up and Go test to categorize a client into low and intermediate fall risks (Montero-Odasso et al., 2022). Such testing of balance control speed focuses on the client's volitional balance control. Inadequate focus is put on the client's speed of reactive balance control, which means the incoming perturbation is unexpected and the person makes postural adjustments after feeling the feedback of perturbation. The prompt responses of muscle activations, forces, joint motions, and postural sways are crucial in reactive balance control (Ochi et al., 2020; M. Pijnappels et al., 2005; Pijnappels et al., 2007; Tong et al., 2023; Wyszomierski et al., 2009). The speed of proper reactive balance control should therefore not be ignored. The synthesis of whether it indicates older people's fall risks is merited to provide higher quality of evidence for the fall-risk assessment guidelines.

Quite a lot of studies have compared the response speed of reactive balance control among the older adults who experienced fall event(s), i.e., fallers, versus those who did not experience fall event, i.e., non-fallers. However, there is no consensus yet regarding the relationships between falls and reactive balance control speed in older people. Firstly, several studies reported the opposite results on whether older fallers and older non-fallers have differed in certain parameters depicting reactive balance control speed (Bair et al., 2016; Batcir et al., 2020; Fujimoto et al., 2015; Rogers et al., 2001; Tantisuwat et al., 2011), making it hard to assess the effectiveness/suitability of using such parameters to identify the fall history and fall risks in older people. Secondly, prior studies have employed various temporal (e.g., onset latency, time to peak) or kinematic measures (e.g., velocity, acceleration) to evaluate the speed of a variety of reactive balance control strategies (e.g., feet-in-place strategy, stepping strategy, and reach-grasp strategy), whole-body postural sways (e.g., CoM displacement, CoP displacement), and neuromuscular activations (e.g., electromyographic [EMG] signals). It is essential to conduct a critical appraisal to synthesize evidence on this topic.

Previous systematic reviews have provided synthesized evidence regarding the associations of older adults' falls with the neuromuscular responses during reactive balance control (Phu et al., 2022) or with the stepping impairment (Okubo et al., 2021). Okubo et al. (2021) synthesized the parameters related to the stepping strategy and summarized that stepping impairments in both volitional balance control and reactive balance control can indicate fall risks. Phu et al. (2022) synthesized the impacts of age, fall history, and exercise on the neuromuscular responses following sudden balance loss, while mainly focusing on the effects of age. However, the two similar review articles have involved only part of the reactive balance control responses and have not focused on the speed parameters of reactive balance control. It remains unclear whether older people with high fall risks have a generally quicker or slower reactive balance control. In addition, it is still difficult to know the appropriateness of using what speed parameters of reactive balance control to differentiate the fall histories or fall risks in older adults.

Given the above, the aim of this narrative review was to answer the questions of: (1) whether the overall reactive balance control speed could distinguish the community-dwelling older people who had fall histories or had high fall risks; and (2) which speed parameter of reactive balance control would show a better ability to quantify the fall risk. The identification of the essential speed parameters of reactive balance control that can distinguish fall history and/or predict fall incidence may facilitate the existing balance assessments to detect the older adults who are vulnerable to falls early, which may potentially facilitate the early provision of relevant intervention to prevent falls.

2.3.2 Methods

2.3.2.1 Data source and search strategy

Following the PRISMA guidelines, two reviewers (R.T.L.Z. and C.H.) undertook the literature search and screening. A complete PRISMA flow chart illustrates the searching strategy and screened results (**Figure 2-4**).

To identify relevant studies, a three-step search strategy was employed (Ma et al., 2020). In step 1, a general search of PubMed was done, where titles and abstracts were reviewed to pinpoint relevant keywords, e.g., “old” AND “fall risk” AND “reactive” AND “time”. In step 2, all the identified keywords were used to search across five databases: PubMed, Scopus, CINAHL, Web of Science, and Cochrane CENTRAL. The truncation operators on keywords were used to widen the search, and the Boolean search operators were used to combine the keywords in this review. There was no restriction regarding the publishing date or language. We created searching alerts to prospectively monitor the possibly eligible articles since the authors’ last search on 8 November 2023. In step 3, reference lists of the identified articles were looked through to any other relevant studies.

2.3.2.2 Study selection

Inclusion criteria were the studies with: **(1)** community-dwelling adults aged 60 years; **(2)** evaluations reflecting how fast the reactive balance control is, e.g., time, rate, velocity, acceleration; **(3)** collection of the retrospective fall history or the prospective fall incidence; and **(4)** effect measures that indicated the comparisons, associations, or diagnostic accuracy. There was no restriction regarding the study design. Exclusion criteria were studies with: **(1)** older people living in the nursing homes or hospitals, or older people with a specific disease;

(2) did not specify the ages of older participants; **(3)** indirect determination of fall status, i.e., using the low score of a balance test or the experimentally-induced tripping/slipping to indicate fall risks; or **(4)** were review articles.

The three-step literature search identified 879 publications (see **Figure 2-4**). Based on the selection criteria, two reviewers (R.T.L.Z. and C.H.) reviewed the titles and abstracts of these publications. Based on the identified abstracts, the full texts were further read based on the identified abstracts to screen the eligible articles.

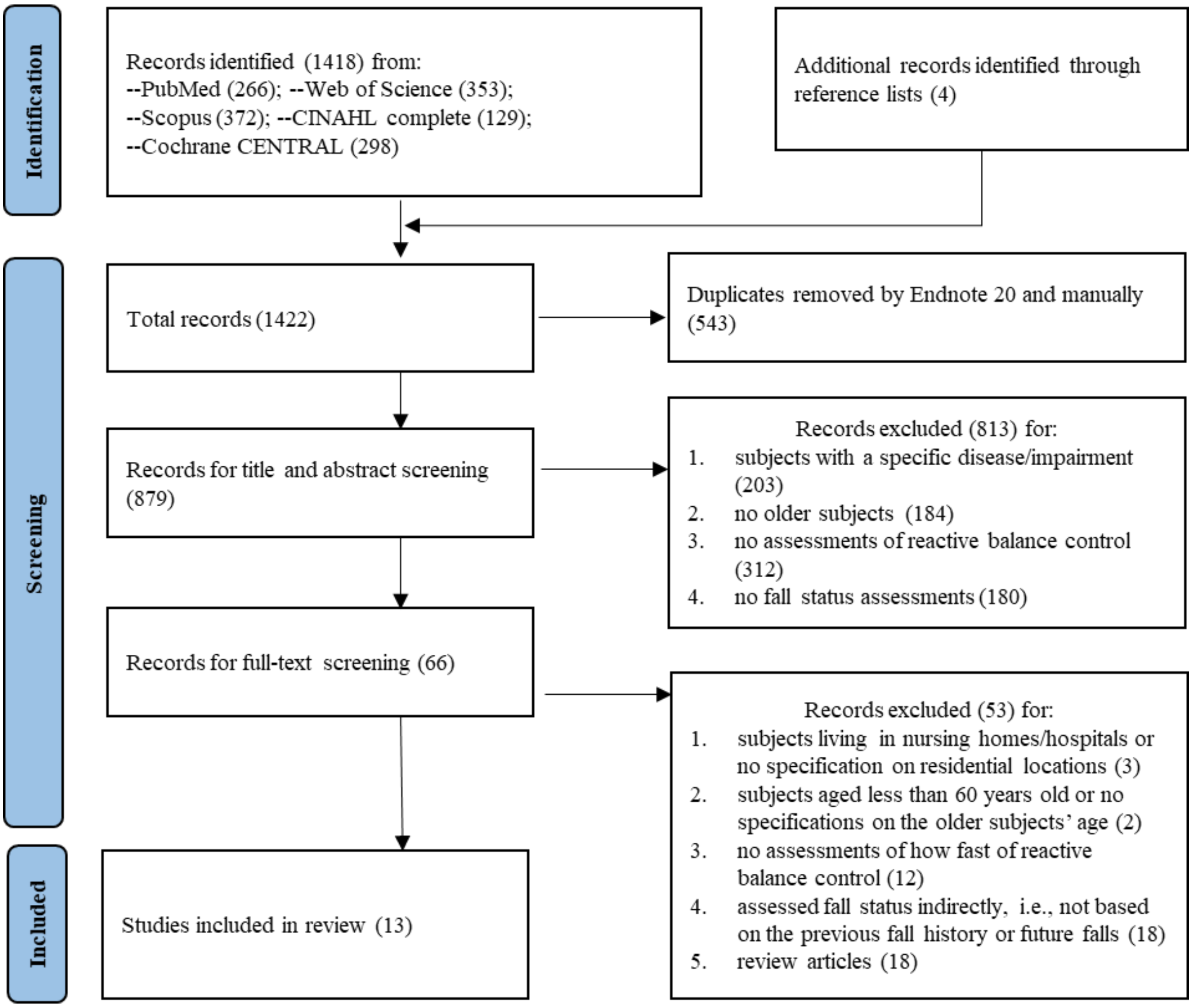

Figure 2-4 The flow chart of study identification and screening.

2.3.2.3 Data extraction

One reviewer (R.T.L.Z.) first extracted information of study characteristics, participant characteristics, fall status assessments, and parameters reflecting the response speed of reactive balance control. If data was not available within the main text, we further screened the supplemental materials for relevant data.

2.3.2.4 Risk-of-bias assessments

We used the 14-item Quality Assessment Tool for Observational Cohort and Cross-Sectional Studies to assess risk of bias for each included study (National Heart Lung and Blood Institute, 2021). Two reviewers (R.T.L.Z. and C.H.) conducted the assessments independently. If disagreements regarding the rating results existed, the two reviewers discussed them first (R.T.L.Z. and C.H.); if agreements did not reach, a third reviewer (C.Z.H.M.) made the final decision.

2.3.3 Results

2.3.3.1 Types and methodological quality of included studies

Thirteen articles were included in this review, which were all observational studies. Three of them were prospective cohort studies, which examined the relationship between the response speeds of reactive balance control and the future falls (Maki et al., 1994; Mille et al., 2013; Sturnieks et al., 2013). The other 10 cross-sectional studies evaluated the response speeds of reactive balance control in older people who had fall histories and those who did not have fall histories (Bair et al., 2016; Batcir et al., 2020; Claudino et al., 2017; Fonseca et al., 2014; Fujimoto et al., 2015; Rogers et al., 2001; Smith et al., 1996; Tantisuwat et al., 2011; Westlake et al., 2016; Zukowski et al., 2023).

The methodological qualities were generally good for the included prospective cohort studies but were poor to fair for the included cross-sectional studies. The scores ranged from 4 to 11 points. The items of “clearly stated research question” (item 1; N = 13), “clearly specified population” (item 2; N = 13), “eligibility criteria applied uniformly to all participants” (item 4; N

= 13), and “clearly defined outcome measures” (item 11; N = 13) were met by all the included studies. Most studies controlled for the confounders (e.g., age, gender) regarding the exposure-outcome relationship (item 14; N = 10). There was commonly high risk of bias regarding the “participation rate” (item 3; N = 0), “sample size justification” (item 5; N = 3), “exposure measured before outcome” (item 6; N = 3), “sufficient timeframe” (item 7; N = 3), “examination of relationships between different levels of exposures and outcomes” (item 8; N = 4), “clearly defined exposures” (item 9; N = 3) , “exposure assessed more than once” (item 10; N = 0), “assessors blinded” (item 12; N = 2), and “follow-up rate” (item 13; N = 3).

2.3.3.2 Participants’ demographics and fall status assessments

A total of 859 older adults were involved (**Table 2-4**). For each included study, the sample size spanned from 23 (Westlake et al., 2016) to 242 (Sturnieks et al., 2013), and the mean age of the older people spanned from 67 to 83 years. The female to male ratio of the included participants was about 2:1. All included participants lived in the community/city/suburban areas, or lived independently, or were specified as healthy.

Older fallers (n = 384) accounted for approximately 1/2 of all the older participants included. Regarding the definition of “fall”, twelve studies clearly defined it as the event that led to a person unintentionally coming to rest on the ground or other lower level, while the other two studies did not mention it. “Fallers” were referred to the participants that experienced fall event(s) either before or after the assessment of reactive balance control. The three prospective cohort studies all monitored the future fall incidences for 1 year via the participants’ fall diaries/postcards on the monthly (Mille et al., 2013; Sturnieks et al., 2013) or weekly (Maki et al., 1994) basis. Telephone calls were made to determine the circumstances if participants had no response in time (Maki et al., 1994; Mille et al., 2013; Sturnieks et al., 2013). The remaining cross-sectional studies evaluated the older participants’ self-reported fall histories, and defined “fallers” as those with ≥ 1 fall in the past one year (Bair et al., 2016; Batcir et al., 2020; Claudino et al., 2017; Fonseca et al., 2014; Fujimoto et al., 2015; Rogers et al., 2001; Westlake et al., 2016), ≥ 2 falls in the past one year (Smith et al., 1996; Zukowski et al., 2023), or ≥ 1 fall in the past six months (Tantisuwat et al., 2011).

2.3.3.3 Assessments of reactive balance control

Older adults' reactive balance control has been induced in the diverse tests. A variety of temporal, kinematic, kinetic, and neuromuscular parameters have been used to evaluate how fast the reactive balance control is (**Table 2-4**). More details are described as follows:

<u>Paradigms/Tests for evaluating response speed of reactive balance control</u>

Twelve included studies investigated the reactive balance control during perturbed standing, while the remaining one included study investigated the reactive balance control during perturbed overground walking. The unexpected perturbations to standing balance were delivered **1)** through the sudden force on the human body such as the waist-pull perturbation (Bair et al., 2016; Fonseca et al., 2014; Fujimoto et al., 2015; Mille et al., 2013; Rogers et al., 2001; Sturnieks et al., 2013) and the shoulder-impact perturbation (Claudino et al., 2017), **2)** or through the sudden movement of the supporting-surface platform such as the translational perturbation (Batcir et al., 2020; Maki et al., 1994; Tantisuwat et al., 2011; Westlake et al., 2016) and the angular perturbation inducing the sudden ankle dorsiflexion (Smith et al., 1996). All these perturbations except the angular perturbation were from the horizontal direction(s). The unexpected perturbation during walking was induced by a near collision that the suddenly moving pedestrian simulated, and the participant needed to perform the collision avoidance walking.

<u>Parameters for evaluating response speed of reactive balance control</u>

The investigations on the response speed of reactive balance control can be categorized into the below four perspectives: **(1)** the whole-body postural sways, **(2)** the lower-limb stepping responses, **(3)** the upper-limb reach-grasp responses, and **(4)** the underlying neuromuscular responses.

Over half of the included studies (N = 7) have investigated the stepping response speed during reactive balance control. Reactive stepping responses were induced mainly by the unexpected waist-pull perturbations (Bair et al., 2016; Fujimoto et al., 2015; Mille et al., 2013; Rogers et al., 2001; Sturnieks et al., 2013), followed by the unexpected translational moving-platform perturbations (Batcir et al., 2020; Tantisuwat et al., 2011). The temporal parameters including the step initiation time (Bair et al., 2016; Batcir et al., 2020; Fujimoto et al., 2015; Mille et al., 2013; Rogers et al., 2001; Sturnieks et al., 2013; Tantisuwat et al., 2011), step duration (Bair et al., 2016; Mille et al., 2013; Rogers et al., 2001; Tantisuwat et al., 2011), step landing time (Batcir et al., 2020; Tantisuwat et al., 2011), and recovery time (Batcir et al., 2020) together with the kinematic parameters including the average step velocity (Sturnieks et al., 2013; Tantisuwat et al., 2011) were measured to evaluate how fast the reactive stepping response occurred.

Five included studies (N = 5) have investigated the response speed of whole-body postural sways in older participants following unexpected balance perturbations. Regarding the rapid responses of CoP displacements, they have been measured to evaluate the feet-in-place balance control strategies following unexpected waist-pull (Fujimoto et al., 2015), shoulder-impact (Claudino et al., 2017), and translational perturbations (Maki et al., 1994). Investigated parameters involved the onset latency of CoP displacement (Claudino et al., 2017), time to peak CoP displacement (Claudino et al., 2017; Maki et al., 1994), peak CoP velocity (Fujimoto et al., 2015), and average CoP velocity (Fujimoto et al., 2015; Maki et al., 1994). Regarding the rapid responses of CoM displacements, they have been evaluated in stepping strategy and dynamic walking (Fujimoto et al., 2015; Rogers et al., 2001; Zukowski et al., 2023). The CoM velocities at step initiation (Fujimoto et al., 2015; Rogers et al., 2001) and step landing (Rogers et al., 2001) following unexpected waist-pull perturbations were investigated, while the other study evaluated the average CoM acceleration, peak CoM acceleration, and the time to peak CoM acceleration in perturbed walking (Zukowski et al., 2023).

Four included studies (N = 4) have analyzed the electromyographic (EMG) onset latency to evaluate neuromuscular response speed in older participants following unexpected waist-pull

(Fonseca et al., 2014), shoulder-impact (Claudino et al., 2017), translational (Westlake et al., 2016), and angular perturbations (Smith et al., 1996). Investigated lower-limb muscles were the ankle dorsiflexor (Fonseca et al., 2014; Smith et al., 1996), ankle plantarflexor (Claudino et al., 2017; Fonseca et al., 2014; Smith et al., 1996), knee flexor (Claudino et al., 2017; Fonseca et al., 2014), knee extensor (Claudino et al., 2017; Fonseca et al., 2014), hip extensor (Fonseca et al., 2014), and hip abductor (Claudino et al., 2017). Investigated trunk muscles were the abdominal muscles and back extensors (Claudino et al., 2017; Fonseca et al., 2014). One included study (N = 1) has also investigated the upper-limb reach-grasp response speed by analyzing the time to handrail contact together with the underlying EMG onset latencies of shoulder flexor and abductor (Westlake et al., 2016).

Table 2-4. Study characteristics (N = 13).

Author (year) / Study design	Definition of fallers	Group	Sample size	Male/ Female	Age (year, mean ± SD)	Perturbation method	Perturbation direction	Parameter(s) indicating response speed of reactive balance control
Bair et al. (2016) Cross-sectional	fall history in previous 12 months	F	16	6/10	73.4 ± 4.6	waist-pull	medial/lateral	step initiation time, step duration
		NF	36	19/17	74.6 ± 7.6			
Batcir et al. (2020) Cross-sectional	≥ 1 fall(s) in previous 12 months	RF	12	2/10	78.5 ± 4.7	translational	medial/lateral	step initiation time, step landing time, step recovery time,
		SF	20	3/17	78.0 ± 5.5			
		NF	51	18/33	79.6 ± 5.1			
Claudino et al. (2017) Cross-sectional	≥ 1 fall(s) in previous 12 months	F	20	12/8	76.0 ± 6.0	shoulder-impact	right	EMG onset latencies of trunk/leg muscles, onset latency of ap/ml CoP displacement, time to peak ap/ml CoP displacement
		NF	20	11/9	73.0 ± 6.0			
Fonseca et al. (2014) Cross-sectional	≥ 1 fall(s) in previous 12 months	F	13	0/13	72.4 ± 8.0	waist-pull	anterior/posterior	EMG onset latencies of trunk/leg muscles
		NF	16	0/16	67.8 ± 6.8			
Fujimoto et al. (2015) Cross-sectional	≥ 1 fall(s) in previous 12 months	F	7	1/6	71.7 ± 4.9	waist-pull	medial/lateral	step initiation time, peak/average CoP velocity before step initiation, CoM velocity at step initiation
		NF	28	15/13	72.8 ± 7.1			
Maki et al. (1994) Prospective cohort	≥ 1 fall(s) in prospective 12 months	F	59	10/49	83.2 ± 6.2	translational	anterior/posterior/ medial/lateral	average CoP velocity, time to peak CoP displacement
		NF	37	7/30	81.6 ± 6.6			
Mille et al. (2013) Prospective cohort	≥ 1 fall(s) in prospective 12 months	F	19	4/15	75.2 ± 7.8	waist-pull	12 directions arranged for every 30° in horizonal plane	step initiation time, step duration
		NF	30	7/23	72.5 ± 5.9			

Continued on next page.

Author (year) / Study design	Definition of fallers	Group	Sample size	Male/ Female	Age (year, mean ± SD)	Perturbation method	Perturbation direction	Parameter(s) indicating response speed of reactive balance control
Rogers et al. (2001) Cross-sectional	≥ 1 fall(s) in previous 12 months	F	18	6/32	74.0 ± 8.0	waist-pull	anterior	step initiation time, step duration, forward CoM velocity at step initiation, lateral CoM velocity at step initiation/landing
		NF	20		71.0 ± 5.0			
Smith et al. (1996) Cross-sectional	≥ 2 fall(s) in previous 12 months	F	32	7/25	74.7 ± 8.5	angular	ankle dorsiflexion	EMG onset latencies of ankle muscles
		NF	30	5/25	74.5 ± 7.6			
Sturnieks et al. (2013) Prospective cohort	≥ 1 fall(s) in prospective 12 months	F (At-home fallers)	106 (54)	57/49 (34/20)	79.8 ± 4.3 (80.3 ± 4.5)	waist-pull	anterior/posterior	step initiation time, step velocity
		NF (Adults except at-home fallers)	136 (188)	75/61 (98/90)	80.2 ± 4.5 (79.9 ± 4.3)			
Tantisuwat et al. (2011) Cross-sectional	≥ 1 fall(s) in previous 6 months	F	36	0/36	/	translational	anterior/posterior	step initiation time, step duration, step landing time, step velocity
		NF	45	0/45				
Westlake et al. (2016) Cross-sectional	≥ 1 fall(s) in previous 12 months	F	12	/	69.8 ± 4.7	translational	medial/lateral	EMG onset latencies of shoulder muscles, time to handrail contact
		NF	11		68.5 ± 4.1			
Zukowski et al. (2023) Cross-sectional	≥ 2 fall(s) in previous 12 months	NF	14	3/11	76.6 ± 9.0	near collision to walking	anterolateral	peak/average AP/ML CoM acceleration, time to peak AP/ML CoM acceleration
		F	15	4/11	77.4 ± 7.6			

Note: F: fallers; NF: non-fallers; RF: recurrent fallers; SF: single fallers; SD: standard deviation. EMG: electromyographic; CoM: center-of-mass; CoP: center-of-pressure; AP: anteroposterior; ML: mediolateral.

2.3.3.4 Response speed of reactive balance control to predict the fall risks

Overall, based on the included prospective cohort studies that examined causal relationships between the response speed of reactive balance control and the fall risks, the later time to peak CoP displacement for maintaining mediolateral feet-in-place balance and the later backward step initiation time for stepping strategy could predict the older people's higher fall incidence (Maki et al., 1994; Mille et al., 2013; Sturnieks et al., 2013). Regarding the step initiation time, neither of two related studies found significant differences between prospective fallers and prospective non-fallers following unexpected waist-pull perturbations (Mille et al., 2013; Sturnieks et al., 2013). However, if specifically looking the at-home fallers, Sturnieks et al. (2013) found that they had significantly later backward step initiation time than the other older participants (RR = 1.89, 95% CI = 1.27–3.19). Regarding the step duration (Mille et al., 2013) and average step velocity (Sturnieks et al., 2013), no significant difference was reported between prospective fallers and prospective non-fallers. Regarding the CoP responses to maintain feet-in-place balance control, fallers were observed with significantly later time to peak CoP displacement in contrast to non-fallers following sudden mediolateral translational perturbations (AUC = 0.68), while no significant difference was found following anteroposterior ones (Maki et al., 1994). In addition, no significant differences were reported between prospective fallers and prospective non-fallers regarding the average CoP velocity following any direction of unexpected translational perturbation (Maki et al., 1994). Further meta-analysis results will be updated.

2.3.3.5 Response speed of reactive balance control to identify the fall history

Fallers seemed to have a faster reactive step following the anteroposterior perturbations, while demonstrating a slower one following the mediolateral translational perturbation than non-fallers. Regarding the step initiation time, three related studies found no significant fall-history-related difference (Bair et al., 2016; Fujimoto et al., 2015; Tantisuwat et al., 2011), one related study found that fallers had significantly earlier step initiation time than non-fallers following the sudden anterior waist-pull perturbations (Rogers et al., 2001), while the other related study found that the recurrent fallers (with two or falls in past 1 year) had significantly later step initiation time in contrast to the single fallers (with one fall in previous 1 year) and non-fallers following the sudden mediolateral translational perturbations (Batcir et al., 2020). Regarding the step duration, two related studies found no significant differences in older fallers vs. older non-fallers (Bair et al., 2016; Rogers et al., 2001), while the other related study found that fallers had a significantly shorter forward step duration than non-fallers following unexpected anteroposterior translational perturbations (Tantisuwat et al., 2011). Regarding the step landing time, one related study observed no significant differences in older fallers vs. older non-fallers (Tantisuwat et al., 2011), while the other related study found that the recurrent fallers had significantly later step landing time than the single fallers and non-fallers (Batcir et al., 2020). Regarding the step recovery time, the recurrent fallers were found to have the later recovery to the original standing place after reactive stepping than the single fallers and non-fallers (Batcir et al., 2020). Regarding the average stepping velocity, fallers had the larger forward step velocity than non-fallers (Tantisuwat et al., 2011). Further meta-analysis results will be updated.

As for the reactive whole-body postural sways, fallers appeared to exhibit quicker responses compared non-fallers in the stepping strategy and walking but not in the feet-in-place strategy. Regarding the rapid responses of CoP, no significant fall-history-related differences were found in the onset latency/time to peak CoP displacement after unexpected lateral shoulder-impact perturbations (Claudino et al., 2017) or in the peak/average CoP velocity after unexpected waist-pull perturbations (Fujimoto et al., 2015). Regarding the rapid responses of CoM, fallers exhibited a significantly larger lateral CoM velocity at step landing in contrast to non-fallers after the sudden anterior waist-pull perturbations (Rogers et al., 2001), and fallers showed a significantly earlier time to peak mediolateral CoM acceleration in contrast to non-fallers after the sudden near collision by a pedestrian during walking (Zukowski et al., 2023).

Regarding the neuromuscular response speed, the fall-history-related difference seemed to be affected by the paradigms for inducing reactive balance control. After sudden shoulder-impact perturbations from the right direction, fallers were observed with later EMG onset latencies in the right abdominal muscle, left hip abductor, and left knee flexor compared to non-fallers (Claudino et al., 2017). In contrast, no significant fall-history-related differences of EMG onset latencies were observed in ankle muscles following the unexpected angular perturbations that induced sudden dorsiflexion (Smith et al., 1996), in dominant-leg and trunk muscles following the unexpected anteroposterior waist-pull perturbation (Fonseca et al., 2014), or in shoulder flexor/abductor for a reach-grasp strategy following the unexpected mediolateral translational perturbations (Westlake et al., 2016). Similarly, no significant fall-history-related difference was found regarding the speed of reach-grasp motion, i.e., time to handrail contact (Westlake et al., 2016).

The discussions and summary below are based on the currently accomplished qualitative analysis. Differences in response speed between older fallers and older non-fallers could vary in the different directions of reactive balance control. Possible mechanisms are as below. Further quantitative meta-analysis will be done to reveal the overall ability of reactive balance control speed in identifying the older people with fall histories or high fall risks and explain the inconsistent results regarding each parameter's fall differentiation ability. Further quality assessment using GRADE will be done to evaluate the evidence level regarding each parameter's fall differentiation ability.

2.3.4 Summary of Narrative Review

Fallers appeared to have a faster response than non-fallers when confronting a sudden threat to forward loss of balance. This was indicated by the fallers' earlier step initiation time and larger lateral CoM velocity at step landing following anterior waist-pull perturbations (Rogers et al., 2001), shorter forward step duration and larger forward step velocity following anterior/posterior translational perturbations (Tantisuwat et al., 2011), and earlier time to peak CoM acceleration following the unexpected near collision during walking in contrast to non-fallers (Zukowski et al., 2023). There were two possible reasons. One could be that fallers had greater concerns or fear of falling, and tended to have anticipation and preplanning in response to the subsequent balance perturbation, even when they were random (Rogers et al., 2001; Sturnieks et al., 2013; Zukowski et al., 2023). Another reason could be that fallers had poorer mediolateral stability in single stance phase when stepping, which could subsequently lead to a quicker forward and lateral step following unexpectedly induced anterior balance loss (Rogers et al., 2001; Tantisuwat et al., 2011).

By contrast, fallers appeared to have a slower response of reactive balance control than non-fallers following unexpectedly induced posterior/medial/lateral loss of balance. This was indicated by the fallers' later step initiation following the posterior waist-pull perturbations (Sturnieks et al., 2013), later time to peak CoP displacement for maintaining feet in place (Maki et al., 1994) and later step initiation/landing/recovery time following the mediolateral translational perturbations (Batcir et al., 2020), and later EMG onset latencies of related trunk and lower-limb muscles maintaining feet in place following the lateral shoulder-impact perturbations in contrast to non-fallers (Claudino et al., 2017). Compared to losing balance in posterior/medial/lateral directions, recovering from anterior loss of balance seemed an easier task. Postural muscles, such as the ankle plantarflexors, can be immediately utilized to eccentrically resist a forward fall as they support the body against gravity during normal standing. Reactive balance control in backward and mediolateral directions could therefore be more challenging for the older adults. The results of this narrative review could support the measurement of posterior/medial/lateral reactive balance control speed for detecting the community-dwelling older people with high fall risks.

In short, the delayed initiation of reactive backward step and the delayed time to peak medial/lateral CoP displacement could predict the older adults fall risks, as indicated by the prospective studies with good methodological qualities. The cross-sectional studies with poor to fair methodological qualities also supported that the backward/medial/lateral reactive balance control speed declined in older people who had fall histories. In contrast, it was the faster response following anterior loss of balance that may indicate the older adults' fall histories. Further meta-analysis and assessment of quality of evidence using GRADE are needed to give confirmed conclusions and recommendations.

2.4 Chapter Summary

In summary, two literature reviews have been conducted in this chapter. The first systematic review and meta-analysis have investigated whether the rapid force generation capabilities (i.e., rate of torque development, rate of force development, power) of lower limbs could effectively differentiate older adults with different fall histories or fall risks. The second narrative review has delved into whether the response speed of reactive balance control (including stepping responses, whole-body postural sways, neuromuscular responses) can differentiate older adults with different fall histories or fall risks.

Together, the two literature reviews in this chapter have suggested that quantitative assessments of the entire lower-limb power during strength tests and the stepping characteristics or whole-body postural sways during reactive balance control could complement the current clinical fall-risk assessments. Nevertheless, the limited number of studies focused on <u>specific lower-limb muscle function</u> restrains comprehensive understanding. These insights lead to the main objective of this PhD project, which is to delve into the intrinsic neuromuscular and biomechanical mechanisms underlying reactive balance control in older adults prone to falls by examining specific hip/knee/ankle joints and muscles.

Chapter 3 Exploring Reactive Balance Control Induced by Translational Perturbations in Young Adults (Study 1)

3.1 Chapter Summary

This chapter includes the contents of study 1 in this PhD project. The study 1 has validated the customized moving-platform system for inducing the translational perturbations which were simulating taking a bus in daily life. Young adults' kinematic/neuromuscular response speed and sequence of multiple lower-limb muscles/joint motions were focused.

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Muscular and Kinematic Responses to Unexpected Translational Balance Perturbation: A Pilot Study in Healthy Young Adults

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Abstract: Falls and fall-related injuries are significant public health problems in older adults. While balance-controlling strategies have been extensively researched, there is still a lack of understanding regarding how fast the lower-limb muscles contract and coordinate in response to a sudden loss of standing balance. Therefore, this pilot study aims to investigate the speed and timing patterns of multiple joint/muscles' activities among the different challenges in standing balance. Twelve healthy young subjects were recruited, and they received unexpected translational balance perturbations with randomized intensities and directions. Electromyographical (EMG) and mechanomyographical (MMG) signals of eight dominant-leg's muscles, dominant-leg's three-dimensional (3D) hip/knee/ankle joint angles, and 3D postural sways were concurrently collected. Two-way ANOVAs were used to examine the difference in timing and speed of the collected signals among muscles/joint motions and among perturbation intensities. This study has found that (1) agonist muscles resisting the induced postural sway tended to activate more rapidly than the antagonist muscles, and ankle muscles contributed the most with the fastest rate of response; (2) voluntary corrective lower-limb joint motions and postural sways could occur as early as the perturbation-induced passive ones; (3) muscles reacted more rapidly under a larger perturbation intensity, while the joint motions or postural sways did not. These findings expand the current knowledge on standing-balance-controlling mechanisms and may potentially provide more insights for developing future fall-prevention strategies in daily life.

Keywords: translational balance perturbation; moving platform; muscle activation; muscle co-contraction; onset latency; time to peak; electromyography (EMG); mechanomyography (MMG)

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1. Introduction

Falls are one of the major public health problems in the world. Approximately one in three older adults fall worldwide [1], and 28.7% of older adults fall in the United States annually [2]. Every year, there are around 684,000 fatal falls, and it is the second leading cause of unintentional injury death [1,3]. For the non-fatal injuries, about one in ten older adults experiences a fall-related injury annually [4]. Falls cause physical and

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3.2 Abstract

Falls and fall-related injuries are significant public health problems in older adults. While balance-controlling strategies have been extensively researched, there is still a lack of understanding regarding how fast the lower-limb muscles contract and coordinate in response to a sudden loss of standing balance. Therefore, this pilot study aims to investigate the speed and timing patterns of multiple joint/muscles' activities among the different challenges in standing balance. Twelve healthy young subjects were recruited, and they received unexpected translational balance perturbations with randomized intensities and directions. Electromyographical (EMG) and mechanomyographical (MMG) signals of eight dominant-leg's muscles, dominant-leg's three-dimensional (3D) hip/knee/ankle joint angles, and 3D postural sways were concurrently collected. Two-way ANOVAs were used to examine the difference in timing and speed of the collected signals among muscles/joint motions and among perturbation intensities. This study has found that (1) agonist muscles resisting the induced postural sway tended to activate more rapidly than the antagonist muscles, and ankle muscles contributed the most with the fastest rate of response; (2) voluntary corrective lower-limb joint motions and postural sways could occur as early as the perturbation-induced passive ones; (3) muscles reacted more rapidly under a larger perturbation intensity, while the joint motions or postural sways did not. These findings expand the current knowledge on standing-balance-controlling mechanisms and may potentially provide more insights for developing future fall-prevention strategies in daily life.

3.3 Introduction

Falls are one of the major public health problems in the world. Approximately one in three older adults fall worldwide (World Health Organization, 2021) and 28.7% of older adults fall in the United States annually (Bergen et al., 2016). Every year, there are around 684,000 fatal falls, and it is the second leading cause of unintentional injury death (Solis-Escalante et al., 2021; World Health Organization, 2021). For the non-fatal injuries, about one in ten older adults experiences a fall-related injury annually (Moreland et al., 2020). Falls cause physical and mental impacts on older adults, and further burden the society heavily (World Health Organization, 2021). Balance and gait disorder is the major cause of falls except accidents (Rubenstein, 2006). The appropriate and timely postural adjustments and lower-limb muscle activities are vital to maintain postural balance. The in-depth investigation into these strategies of how a person reacts to sudden balance perturbations can facilitate our understanding of the underlying mechanisms of falls. This could also provide more insights and inspire the future development of fall-prevention strategies.

Humans have different patterns of lower-limb joint motions in response to varied intensities of balance perturbations. Previous studies have identified three fundamental strategies to maintain the static standing balance, i.e., ankle strategy, hip strategy, and stepping strategy (Rubega et al., 2021). The ankle strategy is reached mainly by the dorsiflexion and plantarflexion of ankle joint, with minimal movement of the other proximal joints (Blenkinsop et al., 2017). This strategy is dominantly employed when no external perturbation exists (Ono et al., 2011). The hip strategy is used when the ankle strategy is not enough to keep the center of mass (CoM) within the base of support (BoS) (Blenkinsop et al., 2017). If the ankle and hip muscles cannot contract sufficiently to compensate for the large balance perturbations, the stepping strategy would be employed (Kochoa, 2016). Sometimes a “mixed strategy” can also be used to maintain balance, with typical characteristics of using the combined above-mentioned strategies depending on the specific situation (Runge et al., 1999).

An investigation of the rapidity and appropriate sequence of movements in the hip, knee, and ankle joints is crucial for understanding the balance controlling strategy, especially during unexpected and intense balance perturbations. Previous studies have primarily focused on the peak responses of postural sways and joint angles in maintaining standing balance (Bair et al., 2016; Chen et al., 2014; Tsai et al., 2014), and only a few studies have analyzed the onset sequence of various lower-limb joint motions in response to balance perturbations in the sagittal plane (Hwang et al., 2009). When specifically looked at the lower-limb responses following the translational perturbations induced by a suddenly forward moving platform, the lower-limb joints commonly reacted with the sequence of ankle dorsiflexion, knee flexion, and finally hip flexion (Hwang et al., 2009). However, research on the timing and speed of lower-limb joint responses to balance perturbations in the frontal plane remains inadequate. Therefore, a study of the temporal parameters, such as onset latency and time to peak, of whole-body postural sways as well as lower-limb joint motions in response to the balance perturbations with various intensities and directions is warranted.

In addition to examining the lower-limb joint motions, analyzing the lower-limb muscle activities can provide greater insights into the underlying balance controlling strategies. Irrespective of the balance controlling strategy utilized, the acceleration of any body segment resulting from a perturbation must be generated by the contraction of the corresponding skeletal muscles. Most previous studies investigated the signals of only one or a couple of lower-limb muscles to maintain balance (Baudry et al., 2012; Błaszczyszyn et al., 2019; Cattagni et al., 2016; Faulkner et al., 2007; Inacio et al., 2019; Mirjam Pijnappels et al., 2005; Sawers et al., 2017). Specifically, most previous studies on static balance control have mainly investigated the EMG signals of ankle dorsiflexor and plantarflexor (Baudry et al., 2012; Błaszczyszyn et al., 2019; Cattagni et al., 2016; Rubega et al., 2021). During the walking task with unexpectedly induced slipping, some previous studies have reported that the older adults who failed to maintain balance tended to have delayed EMG onset in knee flexors/extensors of the slipping legs (Faulkner et al., 2007; Sawers et al., 2017). Some other studies have also reported that the large rate of EMG rise in the dorsal muscles of the stance leg was important to prevent tripping (Mirjam Pijnappels et al., 2005) and that in the hip abductors/adductors was important to make protective stepping (Inacio et al., 2019) in older people. However, till now, there has been insufficient evidence on how fast the major hip, knee, and ankle muscles can react to the balance perturbations with varying intensities and directions.

Meanwhile, one of our recent work has investigated the rapid responses of eight dominant-leg muscles following the unexpected waist-pull perturbations to standing balance, by quantifying the EMG onset latency, time to peak EMG amplitude, and rate of EMG rise of the captioned muscles (Zhu et al., 2022). It has mainly identified that the agonist muscles exhibited quicker activation than the antagonist muscles, and ankle muscles tended to activate faster than the rest six muscles (i.e., knee flexor/extensor, hip flexor/extensor, and hip abductor/adductor) in response to the waist-pull balance perturbation in young adults (Zhu et al., 2022). However, it has remained unclear whether young adults would adopt similar strategies in response to the moving-platform balance perturbation, which mimics a more real-life situation of standing in the buses/trains/boats and merits further study.

Apart from the joint reactions and muscle electrical activities, the response of other events along the motor output pathway during balance control has been less explored. The generation of a joint motion goes through the activation in neuromuscular junctions, the muscle mechanical activities (shortening and lateral vibrations), and the force propagation to tendons (E. Cè et al., 2020). However, it is unknown whether the timing of muscle mechanical activities plays a role in the recovery from balance perturbations of varying intensities and directions. Mechanomyography (MMG) can detect such mechanical activities of a contracting muscle by recording the lateral vibrations that are perpendicular to the muscle fiber direction on skin surface (Woodward et al., 2019). Previous research has reported that the onset of the MMG signal was later than that of the EMG signal during the isometric contraction (Woodward et al., 2019). For dynamic situations, MMG signals have been investigated in the tasks of maintaining walking balance (Lyu et al., 2022; Ma, Ling, et al., 2019) and standing balance (Zhu et al., 2022). The MMG peak timing was found to be later than the EMG peak timing for the ankle plantarflexor during a gait cycle (Ma, Ling, et al., 2019); however, the onset and peak timing of MMG signals was found generally earlier than that of EMG signals following the unexpected waist-pull perturbations, which might be affected by the noise of passive body-segment movements induced by perturbations (Zhu et al., 2022). Therefore, this study made further attempts to explore the timing and coordination patterns of lower-limb major muscles' mechanical activities during balance control.

To bridge the above research gaps, this study aimed to explore how healthy young adults respond to the moving-platform induced balance perturbations with multiple directions and intensities, from the perspectives of postural sways, lower-limb joint motions, and lower-limb muscle activities. The objectives of this study were to examine the differences in speed (onset latency, time to peak, and/or rate of rise) and peak responses of (1) the forward/backward, medial/lateral, and upward/downward CoM displacements; (2) the eight lower-limb joint motions (i.e., hip abduction/adduction, hip flexion/extension, knee flexion/extension, ankle dorsiflexion/plantarflexion); and (3) the eight dominant-leg muscles' electrical and mechanical activities (i.e., hip abductor/adductor, hip flexor/extensor, knee flexor/extensor, ankle dorsiflexor/plantarflexor), under the different perturbation intensities. Such comprehensive analysis of kinematics and muscle activities under the sudden loss of balance that simulates daily scenarios is expected to uncover the more in-depth mechanisms underlying standing balance control. We hypothesized that (1) agonist muscles resisting the induced postural sway would have faster activation than antagonist muscles; (2) involuntary/passive joint motions and postural sways induced by the unexpected perturbation would occur earlier than voluntary/active ones for balance recovery; (3) higher intensities of unexpected perturbations would induce faster muscle activities, joint motions, and postural sways.

3.4 Methods

3.4.1 Study Design and Subjects

This is an observational/cross-sectional and exploratory study. A total of 12 young healthy adults aged between 18 to 24 were recruited through the method of convenience sampling. Ethical approval has been granted by the Institutional Review Board (IRB) of The Hong Kong Polytechnic University (HSEARS20201230002). Subjects satisfied the following inclusion criteria (Gerards et al., 2021):

1) No high-intensity sports within 24 hours before the experiment.
2) No known musculoskeletal or neurological deficits.
3) No history of dizziness, balance or walking disorders.

4) No history of lower limb injuries within a week.

5) No sight or hearing disorders.

6) No medication intake that could affect muscle activities.

3.4.2 Equipment

Moving-platform Perturbation System

A moving-platform perturbation system was developed to induce the unexpected balance perturbations (see Figure 1). It consisted of (1) an aluminum alloy frame, (2) four servo motors (130-07725AS4, Wenzhou Guomai Electronics Ltd., Wenzhou, China), (3) a customized circular wooden platform (diameter: 80.0 cm; thickness: 3.5 cm), (4) four braided polyethylene wires (diameter: 1.2 mm), (5) a set of rails, and (6) a safety harness system (PG-360, Physio Gait Dynamic Unweighting System, Healthcare International Ltd., Langley, WA, USA). An Arduino UNO board (Arduino Uno Rev3, The Arduino Team, Somerville, America) and a customized Arduino program were used to control the servo motors and deliver the unexpected balance perturbations via the wires. Each of the four wires connected the edge of the platform with a motor, and the motor would pull the platform to slide along the rails so as to induce the horizontal balance perturbation in one of the four directions with regard to the subject's body (anterior, posterior, left, and right). Figure 2 shows the flow of generating one perturbation. Firstly, the system delivered a sudden pull to the wooden platform with a randomized direction and intensity. Then, the platform would maintain stationary for 8 s. Finally, the system pulled the wooden platform to return to its original position. The direction, intensity, and starting time of each perturbation were randomized.

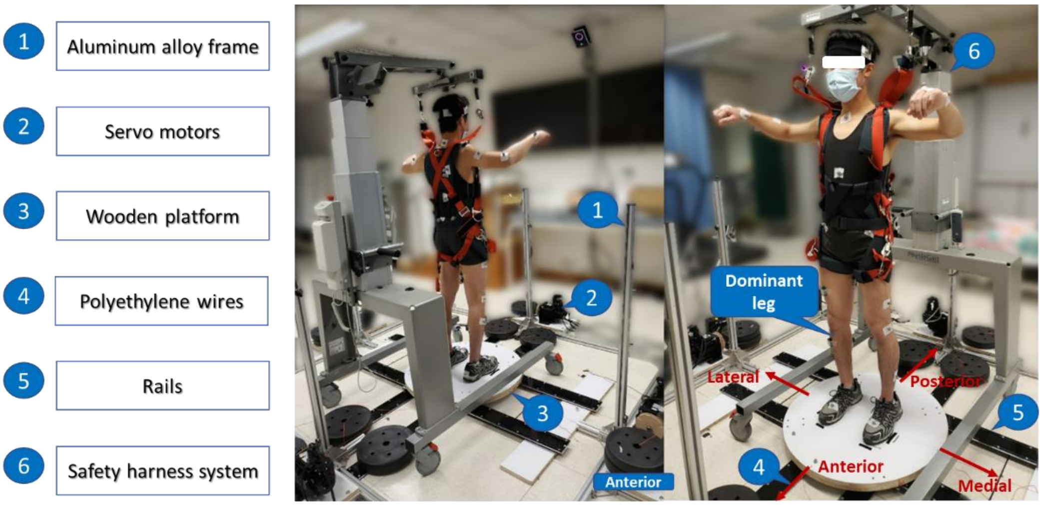

Figure 3-1. The moving-platform perturbation system with a subject.

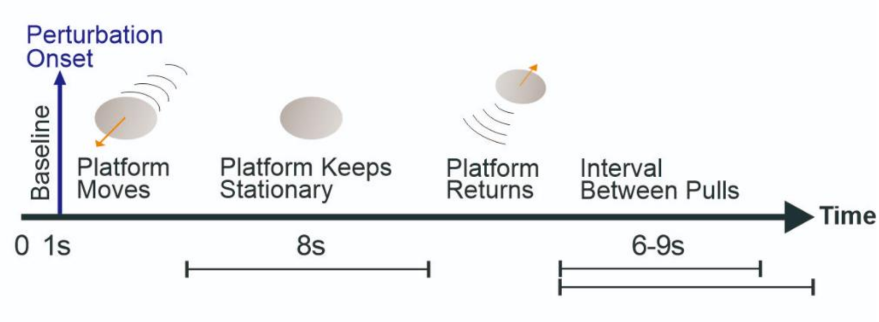

Figure 3-2. The flow of a pulling perturbation.

<u>Data Sampling Equipment</u>

A three-dimensional capture and analysis system, i.e., the Vicon system (Nexus 2.11, Vicon Motion Systems Ltd., Yarnton, UK), with eight cameras was used to capture the whole-body kinematics. The sampling rate was 250 Hz. A total of 39 reflective markers were adhered to the subject's body based on the Full-body Plug-in-gait Dynamic Model, including twelve on the bilateral lower limbs (bilateral thigh, lateral condyle of femur, shank, lateral malleolus, heel, 2nd metatarsal head), four on the pelvis (bilateral anterior superior iliac spine, bilateral posterior superior iliac spine), five on the torso (sternal notch, xiphoid process of the sternum, spinous process of the 7th cervical vertebra, spinous process of 10th thoracic vertebra, right scapula), fourteen on the bilateral upper limbs (bilateral acromion, upper arm, lateral epicondyle of humerus, forearm, radial side of wrist, ulnar side of wrist, 3rd metacarpal head), and four on the head (bilateral temples, bilateral back head) (VICON, 2021). To ensure the reflective markers firmly fixed on the subject's body surface, tight clothes and shorts were provided for subjects to wear during the experiment.

Eight major lower-limb muscles' activities were collected by an eight-channel Trigno Wireless Biofeedback System (SP-W02D-1110, Delsys Inc., Natick, MA, USA). Each Delsys Trigno Avanti sensor (dimension: 37 mm × 27 mm × 13 mm; mass: 14 g) consisted of an EMG sensor (double-differential silver bar electrodes; electrode size: 5 mm × 1 mm; inter-electrode distance: 10 mm; analogue Butterworth filter bandwidth: 20–450 Hz) and a 3-axis accelerometer (range: ± 16 g; resolution: 10 bits) to serve as the MMG sensor (Quam, 2020). The sampling rates of EMG and MMG were 2000 Hz and 250 Hz, respectively. According to the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) recommendation (SENIAM, 2021), the subject's skin was shaved to remove the hair and alcohol wipes were used to clean the skin surface. After the alcohol vaporized and when the skin was dry, the eight EMG and MMG sensors were placed over the investigated eight lower-limb muscles, respectively (see **Table 3-1**), and firmly fixed using double-sided adhesives (Adhesive Interfaces for Trigno Sensors, Delsys, Boston, MA) and pressure-sensitive tapes (Haishi Hainuo Group, Qingdao, China). The Trigno Wireless Biofeedback System has been commercially synchronized with the Vicon system. Infrared light bulbs were connected to the moving-platform perturbation system, and the flash of infrared light indicating each perturbation could be detected by the Vicon system. In this way, the three systems were synchronized during data collection

Table 3-1. Eight investigated muscles and the corresponding EMG/MMG sensor placement.

Muscle	EMG/MMG sensor placement
Ankle dorsiflexor: tibialis anterior (TA)	at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus (SENIAM, 2021).
Ankle plantarflexor: medial gastrocnemius (MG)	on the most prominent bulge of the muscle (SENIAM, 2021).
Knee extensor: rectus femoris (RF)	halfway between the anterior superior iliac spine (ASIS) and the superior boarder of the patella (SENIAM, 2021).
Knee flexor: semitendinosus (ST)	halfway between the ischial tuberosity and the medial epicondyle of the tibia (SENIAM, 2021).
Hip flexor: sartorius (SA)	8 cm distal from the ASIS along the line between the ASIS and the median of the tibial tuberosity (Jiroumaru et al., 2014b).
Hip extensor: gluteus maximus (GMax)	halfway between the sacral vertebrae and the greater trochanter (SENIAM, 2021).
Hip abductor: gluteus medius (GMed)	halfway between the iliac crest to the greater trochanter (SENIAM, 2021).
Hip adductor: adductor maximus (AM)	halfway between the pubic tubercle and the medial femoral epicondyle (Hides et al., 2016).

3.4.3 Protocol

Subjective Assessment

Each subject firstly accomplished the demographic data collection and the subjective assessments. Body mass and height were measured using a standard scale (DETECTO 3P704, Webb city, Missouri, USA), and other anthropometrics like the leg length were measured using a tape measure and a caliper. The International Physical Activity Questionnaire-Short version (IPAQ-S) (Lee et al., 2011) and the Falls Efficacy Scale-International (FES-I) short version (Kempen et al., 2008; Shao et al., 2022; Swanenburg et al., 2013; Zhu et al., 2022) were used to evaluate the physical and mental factors that might affect the balance performance in each subject, respectively. The larger value calculated from the

IPAQ-S reflects the higher level of physical activity in the past 7 days, and the higher score of the FES-I short version (7 items; full score: 28 points) reflects the more concerns over falling (Kempen et al., 2008; Lee et al., 2011). The Mini-Balance Evaluation System Test (Mini-BESTest) was also performed to evaluate the subject's balance capacity in four categories: anticipatory postural control, reactive postural control, sensory organization, and dynamic gait (Franchignoni et al., 2010). The higher score of the Mini-BESTest (14 items; full score: 28 points) indicates the better balance capacity. Finally, the subject's dominant leg was determined using the balance recovery test, ball kick test, and step-up test (Hoffman et al., 1998). The leg used most frequently in nine total trials (3 trials × 3 tests) was considered the dominant leg (Hoffman et al., 1998).

Instrumented Data Collection

Before the perturbation experiment, the reflective markers, EMG and MMG sensors were attached to subjects, with instructions and explanation of the experimental protocol. To simulate the daily situation, the subject was asked to wear his/her usual shoes during the whole perturbation experiment. The subject was instructed to stand naturally with two feet shoulder-width apart in the middle of the platform and knees fully extended. The subject also held a light rod in front of the body at the waist level, so the reflective markers would not be blocked during data collection (Bair et al., 2016). The dark-colored tapes were adhered below the subject's shoes to mark the original foot position on the wooden platform. The subject was instructed to stand still and look forward; when the platform moved, try his/her best to maintain balance without stepping; or return the foot to the original foot position as soon as possible if stepped.

Each subject received a total of 48 unexpected balance perturbations induced by the moving platform (4 directions × 4 intensities × 3 repetitions), and the kinematic, EMG, and MMG data of the subject's responses were collected. These balance perturbations were randomly allocated into four perturbation trials during the experiment, and there was a 5-min rest between the two trials to avoid the effects of fatigue. For each perturbation, the starting time, direction (anterior, posterior, medial, or lateral), or intensity (highest, high, low, or lowest) was random. The direction of a balance perturbation was defined as the moving direction of the platform in reference to the subject's dominant leg. For example, for a subject with the right leg as the dominant leg, pulling the platform toward the left was regarded as a "medial" perturbation while pulling toward the right was a "lateral" perturbation (see Figure 1). Different intensities of balance perturbations were induced by the different displacements and velocities of the moving platform. Based on previous works and our pilot studies, the displacements under the "highest" intensity for the perturbations in anterior, posterior, medial, and lateral directions were set as 4%, 2.67%, 5.33%, and 5.33% of subject's height, respectively (Luchies et al., 1994; Pai et al., 1998; Singh et al., 2017; Zhu et al., 2022). The displacements under the "high", "low", and "lowest" intensities corresponded to the 3/4, 2/4 and 1/4 of that under the "highest" intensity, respectively. Pulling duration of each perturbation was measured from the flash time of infrared light. The displacement, velocity, and acceleration of each perturbation were measured based on the trajectory of the reflective marker fixed on the moving platform. By examining these parameters, the moving-platform perturbation system has shown good reliability of delivering three repetitive balance perturbations with the same direction and intensity. Videos were taken during the entire perturbation experiment to evaluate the subject's stepping strategy following the unexpected perturbations.

3.4.4 Data Processing

The kinematic data, i.e., the whole-body's center of mass (CoM) and the hip, knee, and ankle angles, were firstly processed by the Plug-in-gait Dynamic Model of the Vicon system, and then zeroed to the mean of the 1000-ms baseline values before each perturbation via a customized MATLAB program (MATLAB 2019b, The MathWorks, Inc., Natick, MA, USA). The CoM displacement was further subtracted by the displacement of the platform to obtain the CoM displacement relative to the base of support (BoS).

For the muscle activity data, the raw EMG data were firstly zeroed to the mean values obtained from the whole perturbation trial, then full-wave rectified, and low-pass filtered by a bi-directional 4th order Butterworth filter with a cut-off frequency of 4 Hz to obtain the EMG signal envelopes (Zhu et al., 2022). To extract MMG data, the z-axis accelerometry signals were firstly filtered through an adaptive filter, in an attempt to eliminate the noise caused by limb motion by removing the trajectory of the reflective marker that was close to the MMG sensor. Next, the signals were further band-pass filtered by a 4th order Butterworth filter (5-50Hz) (Lyu et al., 2022; Ma, Ling, et al., 2019), full-wave rectified, and low-pass filtered by a bi-directional 4th order Butterworth filter with a cut-off frequency of 4 Hz to obtain the MMG signal envelopes (Caulcrick et al., 2021). The EMG and MMG signal envelopes were further divided by the 1000-ms baseline mean values before the whole perturbation trial for normalization.

The onset latency, time to peak amplitude, peak amplitude, and/or rate of rise were analyzed for the kinematic, EMG, and MMG data (see Figure 3). The onset and peak points were identified within 2 s after the start of each perturbation. The onset point of a signal was determined as the first time point when the corresponding amplitude exceeded five times of standard deviation (SD) plus the mean of baseline (mean + 5 SD) (Borrelli et al., 2019; Neil J Cronin et al., 2013). The mean of baseline was calculated from the 1000-ms signal values before the start of each perturbation. The onset latency referred to the delayed time between the start of each perturbation and the onset point of a signal. The time to peak referred to the delayed time between the start of each perturbation and the peak point of a signal. The rate of rise was defined as the slope of signal rise over 50 ms after onset. For each parameter (i.e., onset latency, time to peak, peak value, or rate of rise), each subject’s mean of the three values in three repetitive perturbation trials with the same direction and same intensity was used for the statistical analysis.

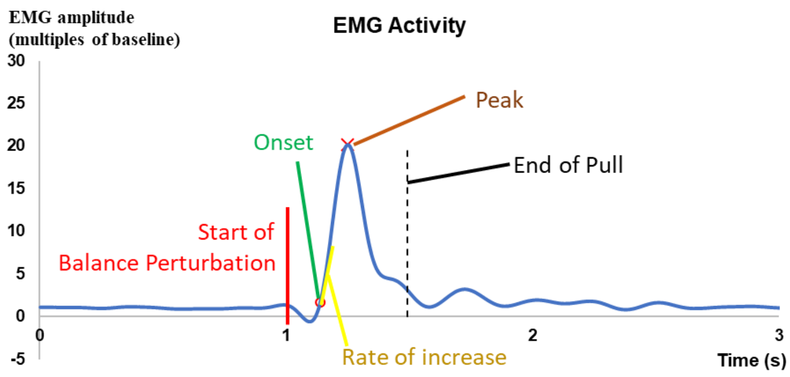

Figure 3-3. **An example of EMG signal showing the examined parameters for one balance perturbation.**

3.4.5 Statistical Analyses

The IBM SPSS version 25 was used for statistical analyses, and the significance level was set as 0.05. For each of the four perturbation directions, two-way ANOVA and post hoc pairwise comparisons with Bonferroni corrections were used to analyze the effects of the below two factors on the onset latency, time to peak, peak amplitude, or rate of rise of the investigated signals:

1. EMG signal difference among the eight different muscles and among the four different perturbation intensities (muscle × perturbation intensity);

2. MMG signal difference among the eight different muscles and among the four different perturbation intensities (muscle × perturbation intensity);

3. Joint angle difference among the eight different joint motions and among the four different perturbation intensities (joint motion × perturbation intensity); and

4. CoM trajectory difference among the six different postural sway directions and among the four different perturbation intensities (postural sway direction × perturbation intensity).

3.5 Results

The demographic data and the subjective assessment results of 12 subjects are shown in

Table 3-2. All subjects had the right leg as the dominant leg. No fall or disastrous event occurred during experiments. All subjects reported that the safety harness system did not restrict their motion while the protection was adequate.

Table 3-2. Demographic data and subjective assessments (mean ± SD) of twelve subjects.

	Male (n=6)	Female (n=6)	Total (n=12)
Age (year)	21.2 ± 1.2	21.5 ± 0.5	21.3 ± 0.9
Height (cm)	174.8 ± 5.8	166.1 ± 4.9	170.4 ± 6.9
Body Mass (kg)	59.2 ± 8.9	56.8 ± 3.6	58.0 ± 6.6
BMI (kg/m²)	19.3 ± 2.2	20.6 ± 1.0	20.0 ± 1.7
Dominant Leg	Right (n=6)	Right (n=6)	Right (n=12)
Leg Length (cm)	88.8 ± 4.6	85.0 ± 3.2	86.9 ± 4.3
IPAQ-S (Kcal/week)	2017.3 ± 1253.3	1238.2 ± 883.6	1627.8 ± 1111.0
FES-I Short Version	10.8 ± 3.4	10.0 ± 2.8	10.4 ± 3.0
Mini-BESTest Score	27.0 ± 0	27.5 ± 0.5	27.3 ± 0.5

Note: BMI: Body mass index. IPAQ-S: International Physical Activity Scale-Short version. FES-I: Fall Efficacy Scale-International. Mini-BESTest: Mini-Balance Evaluation System Test.

Under the lowest, low, and high perturbation intensities, all young subjects were able to maintain balance without stepping or lifting their feet (0/432). Under the highest intensity of perturbations, eight subjects stepped for a total of 14 times following the anterior perturbations (14/36), no subject stepped following the posterior perturbation (0/36), one subject took a step with the non-dominant leg following the medial perturbation (1/36), and one subject took a step with the dominant leg following the lateral perturbation (1/36). Among the stepping responses following anterior perturbations, six subjects took two steps to maintain balance (7/14), three took one step with the dominant leg (4/14), and three took one step with the non-dominant leg (3/14).

3.5.1 CoM Displacements

Figure 4 plots the mean whole-body CoM displacement relative to the BoS of twelve subjects following unexpected balance perturbations with each of the four directions and four intensities (n=12). A red dotted line specifies the start of the balance perturbation (t = 1 s). The mean and standard error (mean ± SE, n=12) values of onset latencies, time to peak, and peak values are also plotted against the different directions of CoM displacement (Figure 5).

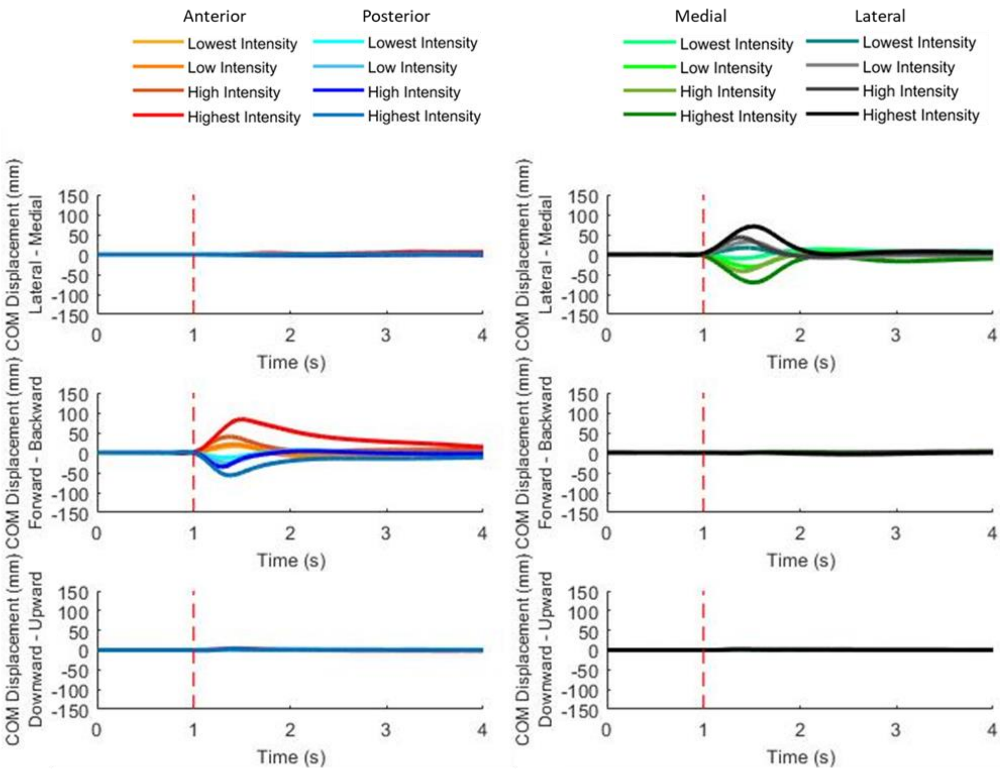

Figure 3-4. **The mean whole-body CoM displacements of twelve subjects following the unexpected anterior, posterior, medial, and lateral perturbations with four intensities (n = 12).**

(Note: CoM: center of mass; Red dotted line specifies the start of the balance perturbation.).

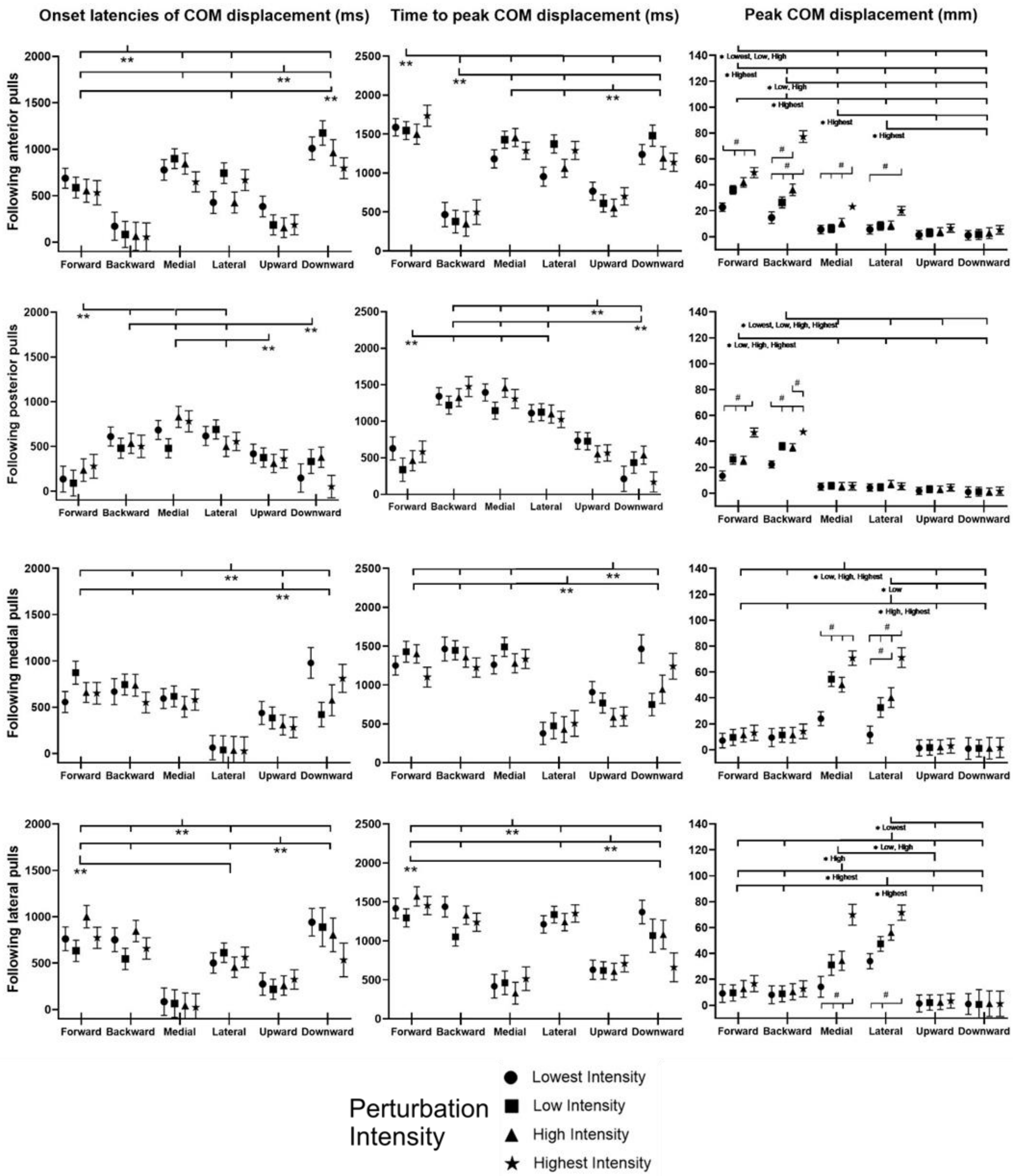

Figure 3-5. **The onset latency of CoM displacement, time to peak CoM displacement, and peak CoM displacement following unexpected horizontal perturbations (mean ± SE, n=12).**

(Note: SE: standard error; ⌐¬ or └┘: pairwise comparison. Significant differences of post hoc pairwise comparisons (p<0.05) were indicated by the ** for the main effect of postural sway factor; * for the simple main effect of postural sway factor; ## for the main effect of intensity factor; # for the simple main effect of intensity factor.)

For each of the four perturbation directions, the initial CoM displacement was found in the direction opposite to the perturbation, followed by the CoM displacement toward the perturbation significantly ($p < 0.05$). Perturbation intensity showed significant effect on the peak CoM displacement ($p < 0.05$), but not on the timing of CoM displacement.

The unexpected perturbation also induced the early response of CoM displacement in the vertical direction. Following anterior perturbations, the CoM displacement had significantly earlier onset and significantly shorter time to peak in upward direction than in downward/forward/medial/lateral directions ($p < 0.05$). Following posterior perturbations, the significantly shorter time to peak CoM displacement was observed in the downward direction compared to the upward/backward/medial/lateral directions ($p < 0.05$). Following medial perturbations, the CoM displacement in upward direction had significantly earlier onset and significantly shorter time to peak than in downward/forward/backward direction ($p < 0.05$). Following lateral perturbations, the CoM displacement had significantly earlier onset and significantly shorter time to peak in upward direction than in downward/lateral/forward/backward direction ($p < 0.05$).

3.5.2 Dominant-leg Joint Motions

Figure 6 plots the mean dominant-leg joint angle changes of twelve subjects following unexpected translational balance perturbations (n=12). Figure 7 shows the mean and standard error (mean ± SE, n=12) values of onset latencies, time to peak, and peak values of the dominant-leg joint angles.

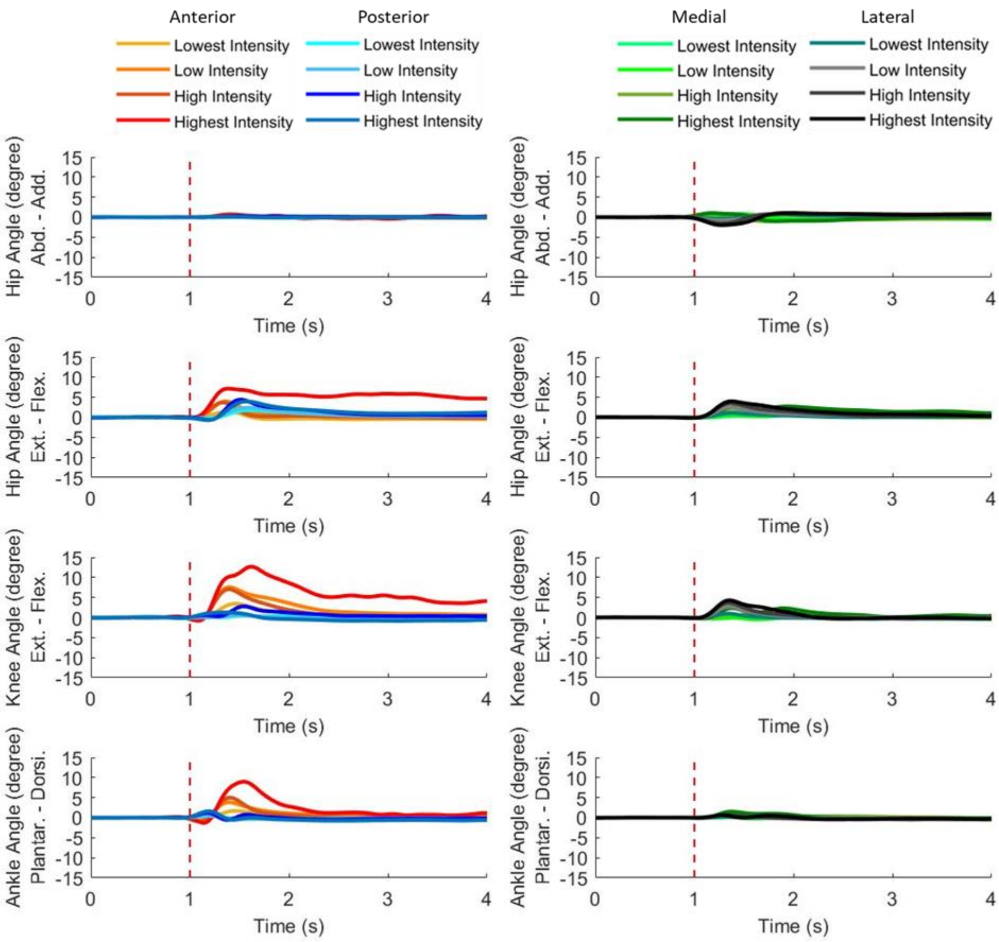

Figure 3-6. The mean dominant-leg joint angle changes of twelve subjects following the unexpected anterior, posterior, medial, and lateral perturbations with four intensities (n = 12).

(Note: Red dotted line specifies the start of the balance perturbation; Add.: adduction; Abd.: abduction; Flex.: flexion; Ext.: extension; Dorsi.: dorsiflexion; Plantar.: plantarflexion.).

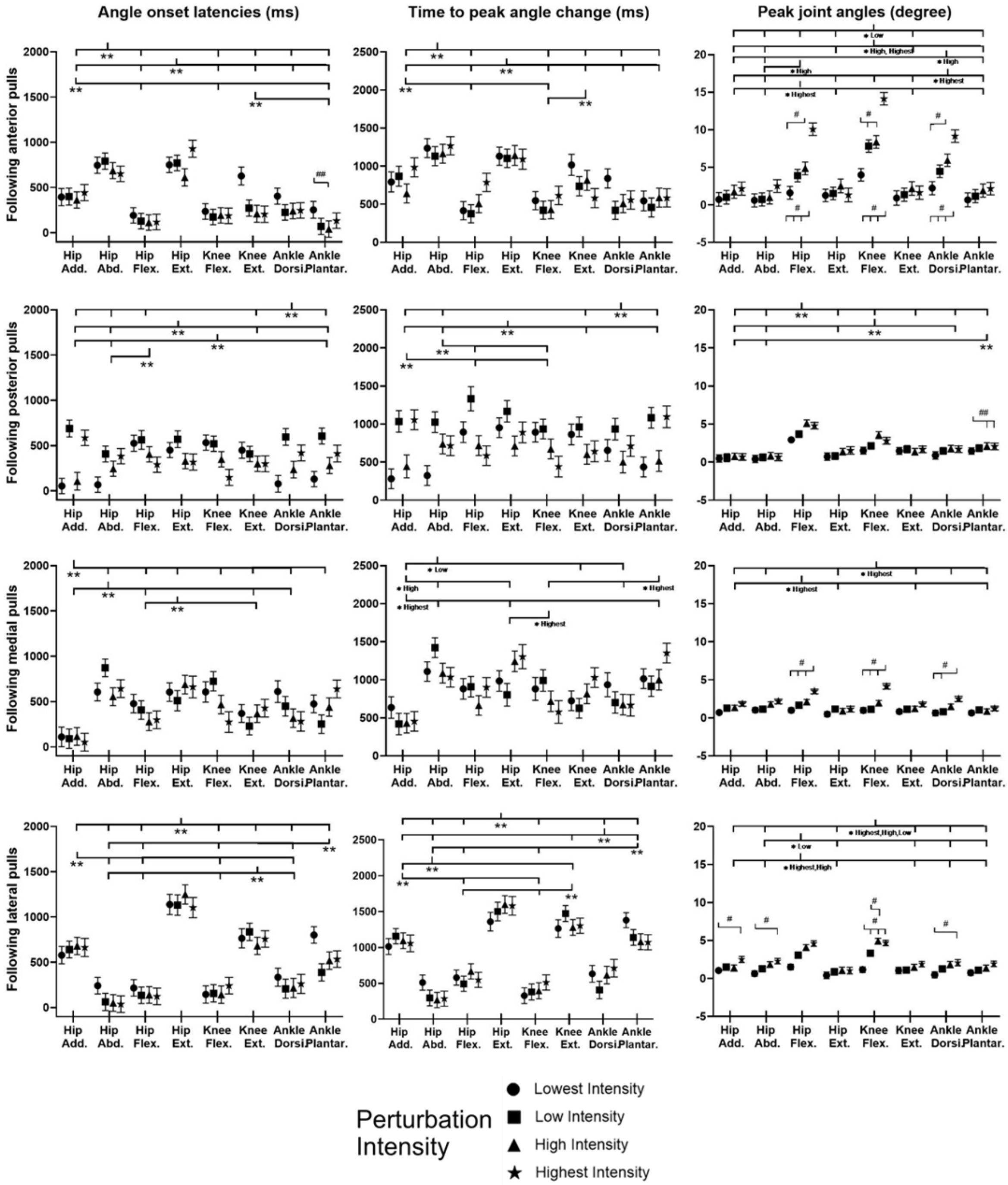

Figure 3-7. The angle onset latencies, time to peak angle, and peak angles of eight lower-limb joint motions following unexpected horizontal perturbations (mean ± SE, n = 12).

(Note: SE: standard error; or : pairwise comparison. Significant differences of post hoc pairwise comparisons (p<0.05) were indicated by the ** for the main effect of joint motion factor; * for the simple main effect of joint motion factor; ## for the main effect of intensity factor; # for the simple main effect of intensity factor.)

When anterior perturbation was induced, significant within-joint differences were observed in the angle onset latencies (hip flexion < hip extension; hip adduction < hip abduction; $p < 0.05$)

and in the time to peak angles (hip flexion < hip extension; hip adduction < hip abduction; knee flexion < knee extension; $p < 0.05$). Generally, the peak angles of ankle dorsiflexion, knee flexion and hip flexion significantly increased with the perturbation intensity ($p < 0.05$).

When posterior perturbation was induced, ankle dorsiflexion had the significantly earlier angle onset and significantly shorter time to peak angle than ankle plantarflexion ($p < 0.05$). Among the eight joint motions, the significantly largest joint motion occurred in hip flexion ($p < 0.05$). Generally, the larger perturbation intensity induced the larger dominant-leg joint motions ($p < 0.05$).

When medial perturbation was induced, hip adduction showed the significantly earliest angle onset among the eight joint motions ($p < 0.05$). Hip adduction also showed significantly shorter time to reach peak angle than hip abduction under the low, high, and highest perturbation intensities ($p < 0.05$); while ankle dorsiflexion showed significantly shorter time to reach peak angle than ankle plantarflexion under the highest perturbation intensity ($p < 0.05$). Generally, the peak angles of ankle dorsiflexion, knee flexion and hip flexion significantly increased with the perturbation intensity ($p < 0.05$).

When lateral perturbation was induced, significant within-joint differences were observed in both the angle onset latency and the time to peak angle (hip abduction < adduction; hip flexion < extension; knee flexion < extension; ankle dorsiflexion < plantarflexion; $p < 0.05$). Under the low, high, and highest perturbation intensities, knee flexion and hip flexion had significantly larger peak angles than the rest joint motions ($p < 0.05$).

3.3. EMG Signals of Eight Dominant-leg Muscles

Figures 8 shows the mean EMG signal of twelve subjects for each of the eight dominant-leg muscles following the unexpected balance perturbations (n=12). Figure 9 plots the mean and standard error (mean ± SE, n=12) values of EMG onset latencies, time to peak EMG amplitude, as well as rate of EMG rise against the eight dominant-leg muscles.

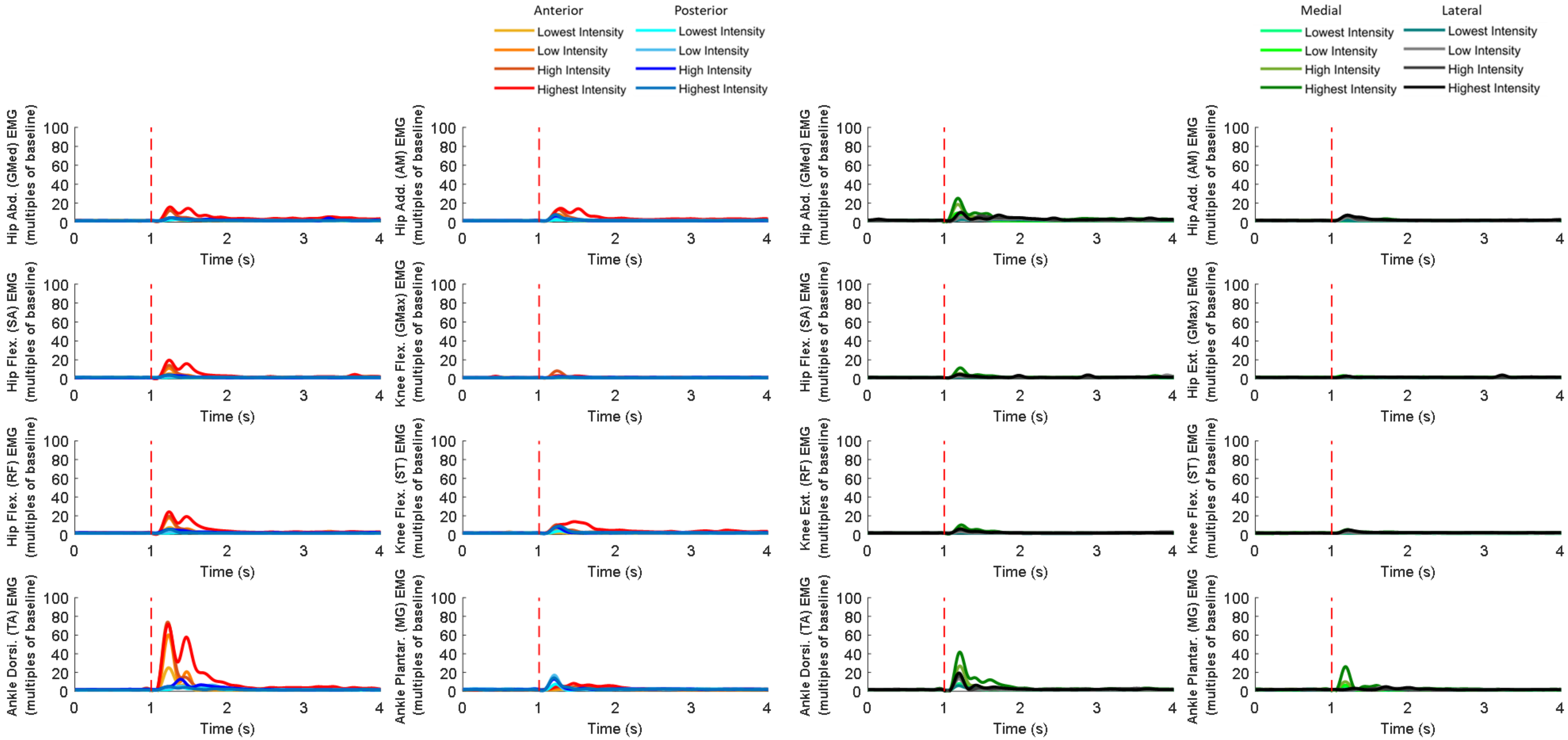

Figure 3-8. **The mean EMG signal changes of twelve subjects for eight dominant-leg muscles following the unexpected anterior, posterior, medial, and lateral perturbations with four intensities (n = 12).**

(Note: Red dotted line specifies the start of the balance perturbation; EMG: electromyography. GMed: gluteus medius; AM: adductor magus; SA: sartorius; GMax: gluteus maximus; RF: rectus femoris; ST: semitendinosus; TA: tibialis anterior; MG: gastrocnemius medialis.).

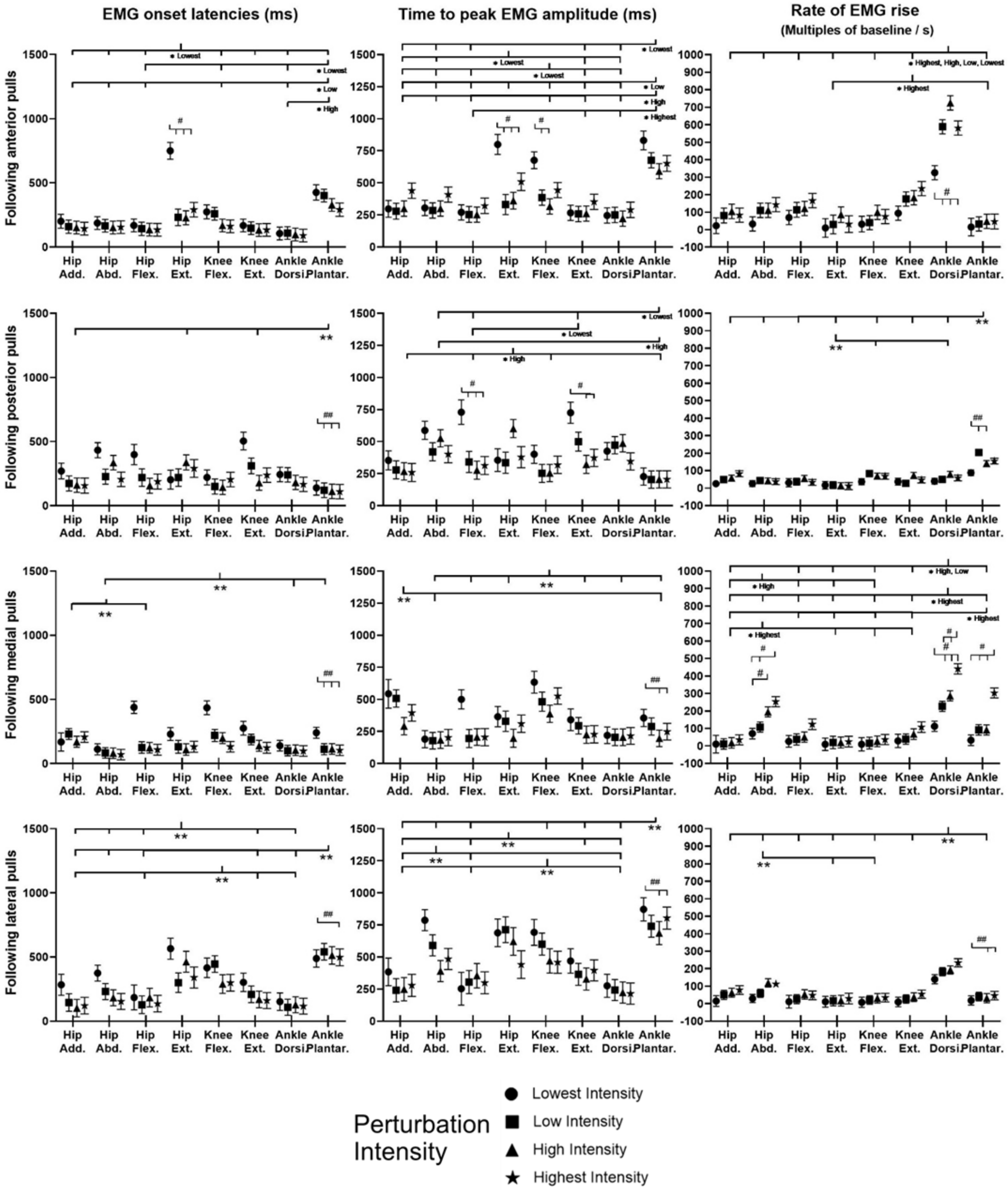

Figure 3-9. The EMG onset latencies, time to peak EMG amplitude, and rate of EMG rise for eight dominant-leg muscles following unexpected horizontal perturbations (mean ± SE, n = 12).

(Note: SE: standard error; ⌐ or ⌙: pairwise comparison. Significant differences of post hoc pairwise comparisons (p<0.05) were indicated by the ** for the main effect of muscle factor; * for the simple main effect of muscle factor; ## for the main effect of intensity factor; # for the simple main effect of intensity factor.)

Following unexpected anterior movement of the platform, significant agonist-antagonist differences were found in EMG onset latency (ankle dorsiflexor < ankle plantarflexor for the lowest, low, and high perturbation intensities; hip flexor < hip extensor for the lowest perturbation intensity; $p < 0.05$) and in time to peak EMG amplitude (ankle dorsiflexor < ankle plantarflexor for all perturbation intensities; knee extensor < knee flexor for the lowest perturbation intensity; $p < 0.05$). The rate of EMG rise was remarkably highest for the ankle dorsiflexor compared to the other muscles ($p < 0.05$).

Following unexpected posterior movement of the platform, ankle plantarflexor had a significantly shorter EMG onset latency compared to knee extensor, hip extensor and hip adductor ($p < 0.05$). Ankle plantarflexor showed the significantly largest rate of EMG rise among the eight dominant-leg muscles ($p < 0.05$). Generally, the larger perturbation intensity evoked the significantly shorter EMG onset latencies ($p < 0.05$) and the significantly larger rate of EMG rise ($p < 0.05$).

Following unexpected medial movement of the platform, significant agonist-antagonist differences existed in EMG onset latency (hip abductor < hip adductor; $p < 0.05$) and time to EMG peak amplitude (hip abductor < hip adductor; knee extensor < knee flexor; $p < 0.05$). Ankle dorsiflexor showed generally the largest rate of EMG rise among the eight dominant-leg muscles under the low, high, and highest perturbation intensities ($p < 0.05$). Generally, the larger perturbation intensity evoked the significantly shorter EMG onset latencies ($p < 0.05$) and the significantly shorter time to EMG amplitude ($p < 0.05$); for ankle dorsiflexor, ankle plantarflexor, and hip abductor, the rate of EMG rise also significantly increased with the perturbation intensity ($p < 0.05$).

Following unexpected lateral movement of the platform, significant agonist-antagonist differences occurred in EMG onset latency (ankle dorsiflexor < ankle plantarflexor; knee extensor < knee flexor; hip flexor < hip extensor; $p < 0.05$) and in time to peak EMG amplitude (ankle dorsiflexor < ankle plantarflexor; hip adductor < hip abductor; hip flexor < hip extensor; $p < 0.05$). Ankle dorsiflexor showed the significantly largest rate of EMG rise among the eight dominant-leg muscles ($p < 0.05$). Generally, the larger perturbation intensity evoked the significantly shorter EMG onset latencies ($p < 0.05$), shorter time to peak EMG amplitude (p <

0.05), and larger rate of EMG rise ($p < 0.05$).

3.5.3 MMG Signals of Eight Dominant-leg Muscles

Figure 10 shows the mean MMG signal of twelve subjects for each of the eight dominant-leg muscles following the unexpected balance perturbations (n=12). Figure 11 plots the mean and standard error (mean ± SE, n=12) values of MMG onset latencies, time to peak MMG amplitude, as well as rate of MMG rise against the eight dominant-leg muscles.

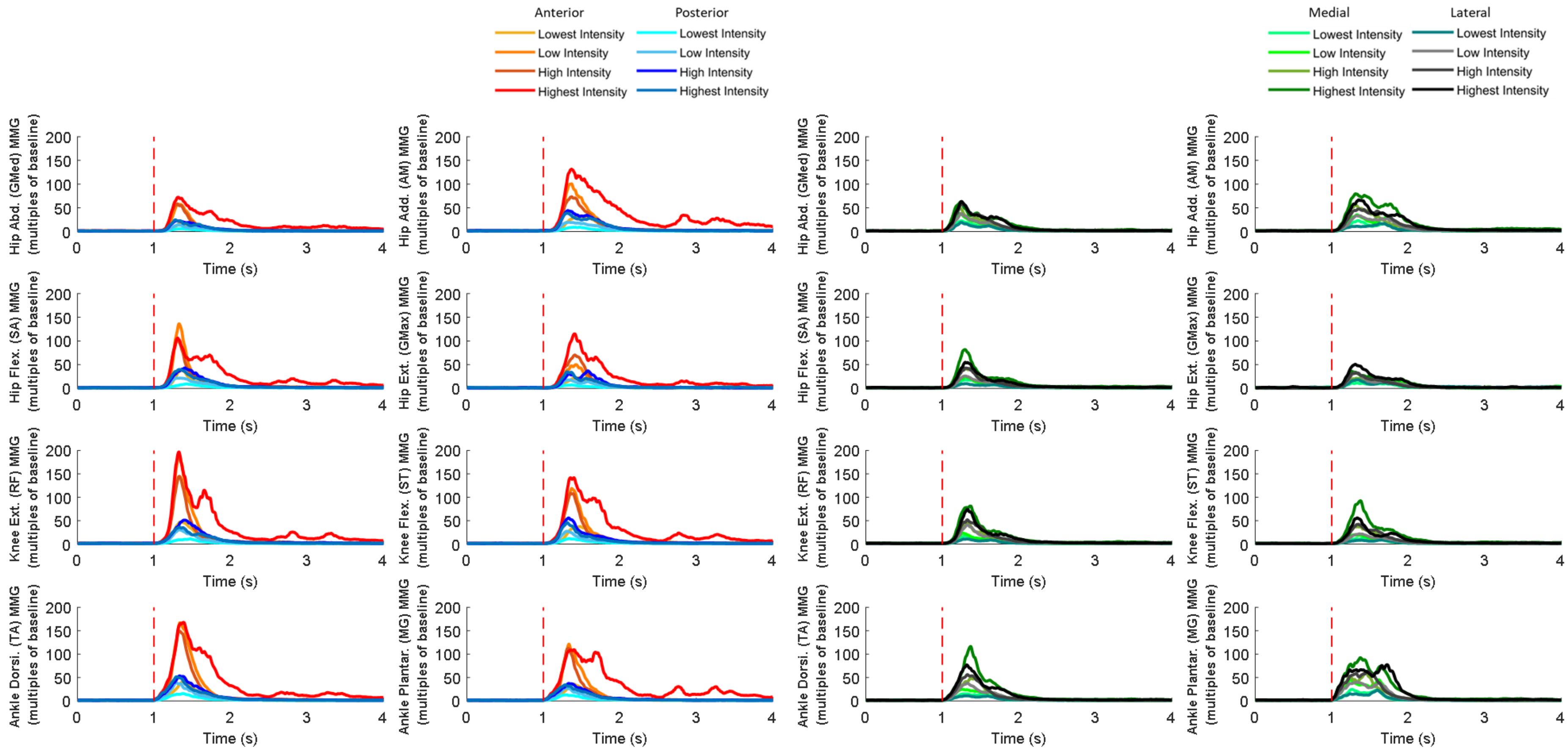

Figure 3-10. The mean MMG signal changes of twelve subjects for eight dominant-leg muscles following the unexpected anterior, posterior, medial, and lateral perturbations with four intensities (n = 12).

(Note: Red dotted line specifies the start of the balance perturbation; MMG: mechanomyography. GMed: gluteus medius; AM: adductor magus; SA: sartorius; GMax: gluteus maximus; RF: rectus femoris; ST: semitendinosus; TA: tibialis anterior; MG: gastrocnemius medialis).

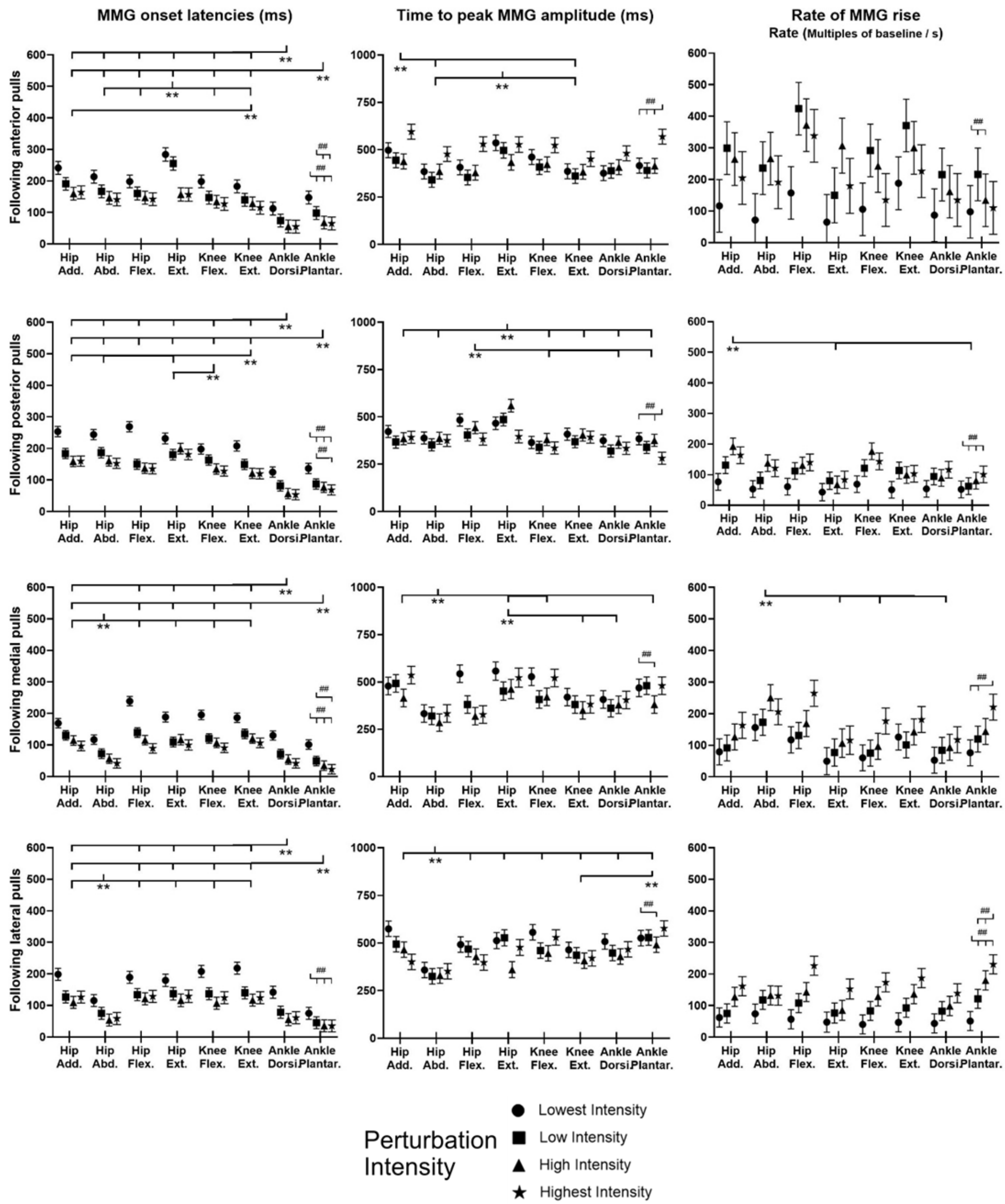

Figure 3-11. The MMG onset latencies and time to peak MMG amplitude for eight dominant leg muscles following unexpected horizontal perturbations (mean ± SE, n = 12).

(Note: SE: standard error; ⌐ or ⌙: pairwise comparison. Significant differences of post hoc pairwise comparisons ($p<0.05$) were indicated by the ** for the main effect of muscle factor; * for the simple main effect of muscle factor; ## for the main effect of intensity factor; # for the simple main effect of intensity factor.)

In general, the onset of MMG signals was significantly earlier in ankle muscles than in knee or hip muscles ($p < 0.05$). Such MMG onset pattern (activation started from distal to proximal lower limb) was observed following all the four directions of balance perturbations. Besides, following the balance perturbations in frontal plane, hip abductor was in the queue with early onset of MMG signal. The larger perturbation intensity evoked the significantly shorter MMG onset latencies ($p < 0.05$), shorter time to peak MMG amplitude ($p < 0.05$), and larger rate of MMG rise ($p < 0.05$).

Following unexpected anterior perturbation, ankle dorsiflexor and ankle plantarflexor had the significantly shorter MMG onset latencies than the other muscles ($p < 0.05$). Significant agonist-antagonist differences existed in MMG onset latency (hip flexor < hip extensor; $p < 0.05$) and in time to peak MMG amplitude (hip abductor < hip adductor; $p < 0.05$). No specific trend was observed when comparing the rate of MMG rise among muscles.

Following unexpected posterior perturbation, ankle dorsiflexor and ankle plantarflexor had the significantly shorter MMG onset latencies than the other muscles ($p < 0.05$).

Following unexpected medial perturbation, ankle dorsiflexor, ankle plantarflexor, and hip abductor had significantly shorter onset latencies than the rest dominant-leg muscles ($p < 0.05$). Significant agonist-antagonist differences existed in both the MMG onset latency and the time to peak MMG amplitude (hip abductor < hip adductor; $p < 0.05$).

Following unexpected lateral perturbation, ankle dorsiflexor, ankle plantarflexor, and hip abductor had significantly shorter onset latencies than the rest dominant-leg muscles ($p < 0.05$). Besides, the hip abductor showed the shortest time to peak MMG amplitude among the eight dominant-leg muscles ($p < 0.05$).

3.6 Discussion

Via the synchronized capture of eight dominant-leg muscles' electrical and mechanical activities (EMG and MMG signals), eight dominant-leg joint motions (angles), and whole-body postural sways (CoM displacements), this study is novel in its comprehensive investigation of the timing and the speed of combined reactions in hip, knee and ankle muscles and joints following the unexpected horizontal/translational perturbations with different intensities and directions. In agreement with our hypothesis 1, this study has observed that (1) Agonist muscles that resisted the perturbation-induced postural sway activated more rapidly than antagonist muscles; and among the eight dominant-leg muscles, ankle muscles' large rate of activation contributed the most to resisting the perturbations in both sagittal and frontal planes. However, our hypotheses 2 and 3 were not entirely supported, as results showed that (2) Fast responses existed not only in those lower-limb joint motions and CoM displacements that were passively/involuntarily induced by the perturbation, but could also occur in those actively/voluntarily generated ones to counteract the perturbation; and (3) A larger perturbation intensity evoked the more rapid muscle activities, but did not induce a faster joint motion or postural sway.

These findings build on our knowledge of how fast the hip/knee/ankle muscles could activate and how the lower-limb joints could coordinate to maintain the reactive standing balance in healthy young adults. This potentially facilitates the understanding of strategies on how humans cope with the varying challenges of losing balance in daily life, which could provide further evidence and guidance for the development of fall-prevention approaches in the future. Details are discussed below.

3.6.1 Faster Activation Existed in Agonist Lower-Limb Muscles, especially Ankle Muscles, to Resist the Induced Postural Sway (hypothesis 1)

The primary finding of this study was that earlier EMG onset and shorter time to peak EMG amplitude occurred in muscles that could resist against the involuntary CoM shift induced by the unexpected moving-platform perturbations. On top of this, this study may highlight the great contribution of ankle muscles' large rate of activation in balance maintenance under the unexpected perturbation in both the sagittal and frontal planes.

Following the unexpected anterior movement of the platform, while the CoM had a firstly posterior shift relative to the BoS, the ventral muscles that could rotate the body forward (ankle dorsiflexor, knee extensor, hip flexor) had generally earlier onset of activation and shorter latency to reach peak activation compared to the dorsal muscles. These observations have been consistent with the previously reported EMG onset patterns (i.e., EMG onset latency of ankle dorsiflexor < ankle plantarflexor; knee extensor < knee flexor) (de Freitas et al., 2010; Hwang et al., 2009; Tsai et al., 2014) and time to peak patterns (i.e., time to peak EMG amplitude in ankle dorsiflexor < ankle plantarflexor; knee extensor < knee flexor) (de Freitas et al., 2010; Krašna et al., 2021). On top of this, this study has also observed that ankle dorsiflexor had the remarkably large rate of activation after onset of perturbation (within 50 ms after onset) under all the perturbation intensities; while the knee extensor showed a large rate of activation under the high perturbation intensity. This indicated the major contribution of the ventral muscles, especially the ankle dorsiflexor, in rapidly shifting the body forward to resist the unexpected anterior movement of the supporting platform. In addition, under the highest perturbation intensity, a second peak was commonly observed for the dominant-leg muscles' EMG signals, which might be related to the subjects' stepping action to recover postural balance. This observation is also in accordance with previous findings that a stepping strategy would be needed to increase the area of BoS and maintain balance, when the ankle and/or hip strategies were insufficient (Kochoa, 2016).

Following the unexpected posterior movement of the platform, the dorsal muscle (ankle plantarflexor) showed an early onset of activation and quick reaching of peak activation to resist the induced forward CoM displacement relative to BoS. This agreed with the previous findings that gastrocnemius and hamstrings had the faster (Runge et al., 1999) and the greater (Baudry et al., 2012; Krašna et al., 2021) muscle activities than the ventral muscles to prevent excessive forward postural sways. In addition, among the eight dominant-leg muscles, this study observed that the ankle plantarflexor had a notably large rising rate at the early phase of muscle activation. This corroborated the previous finding that the rate of ankle plantarflexor's activation played a key role in resisting the forward waist-pull balance perturbations (Zhu et al., 2022) and preventing the forward tripping (Mirjam Pijnappels et al., 2005).

Following the unexpected medial movement of the platform, hip abductor was evoked earlier and reached the peak activation earlier than the hip adductor to resist the induced laterally moving of CoM relative to the BoS. The result consisted with previous studies that reported the significant relations between hip abductor and balance recovery under the lateral waist-pull perturbations (Afschrift et al., 2018; Gilles et al., 1999; Zhu et al., 2022). In addition, this study observed that the ankle dorsiflexor had a generally larger rate of activation than the other dominant-leg muscles; under the high and the highest perturbation intensities, large rate of activation was further evoked in hip abductor and ankle plantarflexor, as the body weight was more quickly transferred to the dominant leg. The dominant-leg distal ankle muscles have been reported in previous studies to provide the immediate joint torque to regain balance under the medial perturbation of platform, followed by the proximal hip muscles with the increasing of perturbation intensity (Freyler et al., 2015; Jeon et al., 2021; Winter et al., 1990). Such kinetic responses could partially be corroborated by the further neuromuscular evidence in this study.

Following the unexpected lateral movement of the platform, the hip adductor reached peak activation earlier than the hip abductor to counteract the suddenly medial moving of CoM relative to the BoS. Besides, among the eight dominant-leg muscles, ankle dorsiflexor had the greatest rate of activation. These findings consisted with the previous studies which indicated the essential contributions of ankle and hip muscles in controlling mediolateral postural balance (Inacio et al., 2019; Jeon et al., 2021; Kovacikova et al., 2015; Zhu et al., 2022). On top of this, in the sagittal plane, this study further observed that the ventral muscles (ankle dorsiflexor, knee extensor, hip flexor) had earlier onset and shorter time to peak activation than the dorsal muscles. These rapid activation patterns might be related to the attempt of trying to counteract the body weight unloading from the dominant leg, and prevent the excessive medial postural sway induced by the unexpected laterally moving of platform (Ma, Chung, et al., 2019).

To the authors' knowledge, no previous study has investigated the activation pattern of all the major hip, knee, and ankle muscles to maintain balance over the unexpected platform movement. By examining the hip adductor/abductor, hip flexor/extensor, knee flexor/extensor, and ankle dorsiflexor/plantarflexor simultaneously, this study identified the essential importance of ankle muscles' rapid activation in resisting the horizontal perturbations for young adults. It is also expected that the more comprehensive and in-depth investigation in eight leg muscles' activities in this study could facilitate the future development of programs and assistive devices to improve balance and prevent falls in older adults (Elhadi et al., 2018; Huang et al., 2022; Ma et al., 2022; Ma, Chung, et al., 2019; Ma et al., 2020; Ma & Lee, 2017; Ma et al., 2015; Ma et al., 2016; Ma, Wong, et al., 2018; Ma, Zheng, et al., 2018; Wan et al., 2016; Wang et al., 2023).

3.6.2 Rapid Kinematic Responses Varied with the Perturbation Direction (hypothesis 2)

The secondary finding of this study was that the rapid responses of lower-limb joint motions and postural sways varied in different directions of balance perturbations. Following the unexpected platform movement, lower-limb joint motions caused by the inertia of body segments were generally the first to appear and considered as passive/involuntary, while sometimes other active/voluntary joint motions could also appear as early as the passive/involuntary ones. Among the six directions of CoM displacement, CoM moving toward the opposite direction of the balance perturbation was generally the earliest because of inertia, which has been consistent with the previous studies' findings (Ma & Lee, 2017; Rietdyk et al., 1999; Runge et al., 1999); while sometimes the CoM displacement in the vertical direction could also occur as early as in the horizontal direction either because of inertia or active/voluntary reactions.

Early onset of joint angles and short time to peak angles were generally observed in the passive/involuntary joint motions induced by the moving platform. Specifically, because of inertia, faster responses of ankle dorsiflexion (compared to ankle plantarflexion) (Runge et al., 1999), hip adduction (compared to hip abduction) (Rietdyk et al., 1999; Zhu et al., 2022), and hip abduction (compared to hip adduction) (Rietdyk et al., 1999; Zhu et al., 2022) were induced under the unexpected posterior, medial, and lateral perturbations, respectively. However, only under the medial perturbation, the onset of passively induced hip adduction was found to be significantly earlier than the rest seven lower-limb joint motions. For the anterior, posterior, and lateral directions, the passive/involuntary joint motion was usually accompanied by the fast responses of other active/voluntary joint motions to resist the balance perturbation.

The timing patterns of active/voluntary lower-limb joint motions were different following the four directions of unexpected balance perturbations. Specifically, under the unexpected anterior perturbation, the passively/involuntarily induced ankle plantarflexion was accompanied by the consistently early onset of hip flexion (compared to hip extension). A previous study also reported that early ankle plantarflexion was followed by the knee flexion, and hip flexion under the sudden forward movement of platform (Hwang et al., 2009). On top of it, this study observed that knee flexion (compared to knee extension) and hip flexion (compared to hip extension) reached peak angles more quickly, which further corroborated the finding in onset of joint motions. Under the unexpected posterior perturbation, no consistently fast response within the knee or hip joint accompanied the passively/involuntarily induced ankle dorsiflexion. However, a previous study reported that young adults also had an earlier hip extension than hip flexion apart from the ankle joint motions following the posterior moving-platform perturbation (Runge et al., 1999). This may be because the perturbation intensity induced by the posterior movement of platform was not as large as that in the previous study (Runge et al., 1999; Zhu et al., 2022). Therefore, this study observed only the consistently later onset and peak in ankle joint motion (ankle dorsiflexion < ankle plantarflexion). The corrective reaction, i.e., ankle plantarflexion, was sufficient to recover from the induced forward loss of balance. Under the unexpected medial perturbation, all the active/voluntary joint motions appeared after the passively/involuntarily induced hip adduction. The hip abduction was corrective as the muscle activity of hip abductor was detected which could work to oppose the passive hip adduction. This reaction was also supported by a previous study which found the corrective response of hip abduction moment following perturbations in frontal plane (Rietdyk et al., 1999; Zhu et al., 2022). Under the unexpected lateral perturbation, the passively/involuntarily induced hip abduction was accompanied by the consistently early onset and short time to peak of hip flexion, knee extension, and ankle dorsiflexion. Similar to the medial perturbation, the hip adductor muscle worked to oppose the passive hip abduction and produce the later corrective hip abduction (Rietdyk et al., 1999; Zhu et al., 2022). In addition, the observations of this study may suggest that when the body weight was unloaded from the dominant leg following lateral perturbations, fast joint motions in sagittal plane were also required to maintain balance; while they were not required when the body weight was loaded on the dominant leg following medial perturbations.

The unexpected horizontal perturbation also induced the consistently faster response of CoM in the vertical direction. As far as the authors know, previous studies have rarely compared the timing of postural sways in horizonal and vertical directions following a balance perturbation. Specifically, in this study, the faster upward CoM displacements (compared to downward direction) were observed following the anterior, medial, and lateral balance perturbations; while the posterior perturbation induced a faster downward CoM displacement. Following the anterior perturbation, the earlier upward CoM displacement (compared to downward direction) seemed to be related to the great proportion of subjects' stepping strategies and toe-rise strategies in the study. Following the medial or the lateral perturbation, the faster responses of hip flexion, knee flexion, and ankle dorsiflexion in the unloaded leg were observed. The fast upward CoM displacement could be caused by the lifting of the unloaded leg. By contrast, following the posterior perturbation, the CoM reached peak displacement more quicky in the downward direction than in the upward direction. This response was considered to be passive, as the downward and the passively induced forward postural sways had similar onset latencies and time to peak. The posterior displacement quickly induced the subject's forward inclined posture, so the CoM had a firstly downward displacement and was followed by a later upward displacement to recover the upward posture. Difference in perturbation intensities may explain why the initial CoM displacement in the vertical direction following posterior perturbations differed from that following anterior/medial/lateral perturbations. Posterior perturbations were set with smaller intensities than anterior/medial/lateral perturbations in this study. The latter ones were challenging enough and induced the faster active/voluntary reactions that elevated the CoM, while the former ones were not.

3.6.3 Larger Perturbation Intensity Evoked Faster Rosponse in Muscle Activation (hypothesis 3)

The tertiary finding of this study was that the larger perturbation intensity induced the earlier EMG onset, shorter time to peak EMG amplitude, and larger rate of EMG rise in the dominant-leg muscles; but not led to faster responses in postural sways or lower-limb joint motions. This may suggest that under a larger unexpected challenge to loss of balance, the lower-limb muscles could activate earlier and more rapidly to restrict the excessive joint motions and prevent excessive CoM shifts out of the BoS.

Results in this study indicated that the response rate of lower-limb muscles could be related to perturbation intensity in general. EMG onset latencies and time to peak EMG amplitude were significantly longer under the smaller perturbation intensities than under the higher ones. Previous studies have reported such phenomenon mainly in the sagittal plane (anterior and posterior directions) (Hwang et al., 2009; Krašna et al., 2021; Lin & Woollacott, 2002; Runge et al., 1999). On top of them, this study suggested that this trend was also applicable to perturbation in the frontal plane (medial and lateral directions). In addition, this study observed the higher EMG rising rate under a larger perturbation intensity. Such effects of perturbation intensity seemed to be more prominent in the agonist muscles that could resist postural sways induced by the unexpected moving-platform perturbations (e.g., ankle dorsiflexor under the anterior perturbations; ankle muscles and hip abductor under the medial perturbations). However, the specific or type of the correlation between the muscle responses and perturbation intensities has remained unclear. It would be interesting to establish some models between the two factors in future studies.

A larger perturbation intensity could evoke the larger responses in lower-limb joint motions and postural sways; however, it may not be able to evoke the faster kinematic responses. For the postural sways, this study found that the larger perturbation intensity generally induced the larger CoM displacement along the line of perturbation direction. An example was that both the forward and the backward peak CoM displacements could increase with the anterior perturbation intensity. This was understandable since the different perturbation intensities in the study were position-controlled. However, a larger intensity of perturbation did not evoke the earlier onset or shorter time to peak CoM displacement. A similar amount of time was required to reach a larger postural sway under the greater perturbation intensity. The result partly agreed with a previous study which showed increasing peak CoM velocity and peak CoM acceleration with perturbation intensity [61]. For the lower-limb joint motions, it was found that ankle dorsiflexion, knee flexion, and hip flexion angles increased with the perturbation intensity for all the four perturbation directions. Such strategies seemed to be able to reduce the additional horizontal excursions when the challenge to loss of standing balance became larger (de Freitas et al., 2010). However, the onset or the time to peak for neither the whole-body postural sways nor lower-limb joint motions could be affected by the perturbation intensity. These may suggest that the earlier and faster lower-limb muscle activation would be adequate for the successful maintenance of balance following a larger balance challenge. In addition, as these findings indicate that the neuromuscular reaction time

(EMG onset latency and time to EMG peak amplitude) could be modulated by the different intensities of perturbations, EMG temporal parameters could potentially be used as the training outcome or biofeedback in perturbation-based balance training (Gerards et al., 2023).

3.6.4 MMG Signals following Balance Perturbations Merit Further Study

Attempts have also been made to process and analyze the MMG signals, aiming to examine the possible delays between the electrical and mechanical muscle activities in response to unexpected balance perturbations. A new processing method has been used upon optimizing the one reported in our previous study (Zhu et al., 2022). However, the MMG signals obtained in the current study may still not fully reflect the exact mechanical activities of muscles, concerning that the onset of MMG signal was not consistently later than that of EMG signal. The observed time delay between the EMG and MMG onset was not comparable with the previous studies, either. The delay in MMG signal after EMG signal observed in previous studies ranged from 7ms to 30ms instead (Emiliano Cè et al., 2020; Cè et al., 2013; Fukawa & Uchiyama, 2016; Smith et al., 2017; Uchiyama et al., 2017). Thus, it should be noted that the following discussion of MMG data might be affected by the current processing method of MMG signals used in this study. Further attempts are still needed to improve the MMG data processing method in future studies.

When comparing among the eight muscles, the onset of MMG signal under all directions of perturbation was significantly earlier at distal than proximal muscles. Such onset might be caused by the perturbation induced from a moving platform supporting the standing, or in other words, at the distal side of subject’s body. This is further supported by the result that the value of each parameter in MMG recorded greatly depended on pulling intensity, but not for muscles or pulling directions. Although MMG showed considerable reliability to detect muscle activities in static conditions (Ling et al., 2020; Smith et al., 2017), it might not be effective in dynamic conditions, as the current processing method of MMG signals could still not effectively eliminate the noise caused by the moving platform and the movement of body segment. Further studies and optimization of set-up are required on MMG technologies to be applied to muscle activity measurement in dynamic situations.

3.6.5 Strengths and Limitations

This study has the below strengths. Firstly, the moving-platform perturbation system has been synchronized with the Vicon system and the Trigno Wireless Biofeedback System, enabling our accurate analyses of the temporal parameters of multiple signals during balance control. Secondly, the perturbation intensity (i.e., pulling displacement) has been set as a percentage of the subject's body height so that the perturbation is a consistent challenge to postural balance across different individuals. Thirdly, this study has comprehensively analyzed how fast the eight major leg muscles' activation and eight lower-limb joint motions could occur during balance control. These findings in healthy young adults can serve as the foundation and reference for not only the further studies of balance control mechanisms in older adults with high fall risks but also the further development of assistive/robotic devices for targeted balance training and fall prevention.

There are some potential limitations of this study. Firstly, only a small number of subjects were recruited for this pilot study. Future studies in older adults or a specific population should justify the sample size to make the findings more representative. Secondly, during the processing of EMG signals, this study used baseline EMG value in normal standing for normalization. Therefore, it should be noted that the rate of EMG rise in this study was based on the muscle activation level in normal standing rather than the capacity of muscle activation. The maximal voluntary contraction (MVC) tests could be carried out in the future to examine if the amplitude normalization methods would affect the finding. Lastly, the current MMG processing method might still be not mature to extract the exact vibrations of lower limb muscles. Further observation and development are needed. More tests could be performed on MMG to evaluate the reliability and validity of reflecting the mechanical muscle activities following the unexpected perturbation.

3.7 Conclusion

To conclude, among the eight dominant-leg muscles, ankle muscles' rapid activation contributed the most to resist the unexpected moving-platform perturbations in both sagittal and frontal planes. Fast responses of the lower-limb joint motions and the vertical postural sways were related to the perturbation direction. Under a larger perturbation intensity, muscles reacted more rapidly, while joint motions or postural sways were not necessarily faster. These findings provide new insights on the sequence or the fast responses of multiple lower-limb muscles/joints to cope with the different levels of balance challenges. The mechanisms underlying reactive standing balance are better understood, which may facilitate future research on developing targeted balance training protocol and/or technology or device in fall prevention for older adults.

Chapter 4 Exploring Reactive Balance Control Induced by Translational Perturbations in Older Adults (Study 2)

4.1 Chapter Summary

This chapter includes the contents of study 2 in this PhD project. As the customized moving-platform system has been validated to induce unexpected translational perturbations in young adults, it was further used in this pilot study for the investigations of neuromuscular/kinematic mechanisms underlying the fall-prone older people's reactive balance control. In this pilot study, the fall-related alterations in lower-limb muscles activities and joint motions following unexpected translational perturbations were focused, by comparing older fallers (i.e., the older adults with fall histories) and older non-fallers. However, this pilot study did not recruit more participants. The reason was that the force plates of Vicon system could not be utilized during the moving-platform experiments, making the collection of kinetic data unavailable. Based on the results of our meta-analysis, the leg-press power and sit-to-stand power could significantly differentiate fall histories or predict prospective falls in older adults. It was inferred that the lower-limb kinetic responses for reactive balance were also important to differentiate older people's fall status. The perturbation system was therefore modified to enable the use of force plates and collection of kinetic data in the later studies (waist-pull studies) of this PhD project.

This study has been published by the author of this thesis as an article in the journal of *Bioengineering-Basel*.

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Article

Older Fallers' Comprehensive Neuromuscular and Kinematic Alterations in Reactive Balance Control: Indicators of Balance Decline or Compensation? A Pilot Study

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Abstract: Background: Falls and fall consequences in older adults are global health issues. Previous studies have compared postural sways or stepping strategies between older adults with and without fall histories to identify factors associated with falls. However, more in-depth neuromuscular/kinematic mechanisms have remained unclear. This study aimed to comprehensively investigate muscle activities and joint kinematics during reactive balance control in older adults with different fall histories. **Methods**: This pilot observational study recruited six community-dwelling older fallers (≥1 fall in past one year) and six older non-fallers, who received unpredictable translational balance perturbations in randomized directions and intensities during standing. The whole-body center-of-mass (COM) displacements, eight dominant-leg joint motions and muscle electrical activities were collected, and analyzed using the temporal and amplitude parameters. **Results**: Compared to non-fallers, fallers had significantly: (a) smaller activation rate of the ankle dorsiflexor, delayed activation of the hip flexor/extensor, larger activation rate of the knee flexor, and smaller agonist-antagonist co-contraction in lower-limb muscles; (b) larger knee/hip flexion angles, longer ankle dorsiflexion duration, and delayed timing of recovery in joint motions; and (c) earlier downward COM displacements and larger anteroposterior overshooting COM displacements following unpredictable perturbations ($p < 0.05$). **Conclusions**: Compared to non-fallers, fallers used more suspensory strategies for reactive standing balance, which compensated for inadequate ankle/hip strategies but resulted in prolonged recovery. A further longitudinal study with a larger sample is still needed to examine the diagnostic accuracies and training values of these identified neuromuscular/kinematic factors in differentiating fall risks and preventing future falls of older people, respectively.

Keywords: community-dwelling; older adults; falls; reactive balance; perturbation; electromyographic (EMG); co-contraction index (CCI); kinematics; postural sways

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1. Introduction

Falls and fall consequences in older adults burden society heavily and are global health issues [1]. Annually, around one in three older adults falls, one in ten older adults

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4.2 Abstract

Background: Falls and fall consequences in older adults are global health issues. Previous studies have compared postural sways or stepping strategies between older adults with and without fall histories to identify factors associated with falls. However, more in-depth neuromuscular/kinematic mechanisms have remained unclear. This study aimed to comprehensively investigate muscle activities and joint kinematics during reactive balance control in older adults with different fall histories.

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Results: Compared to non-fallers, fallers had significantly: (a) smaller activation rate of the ankle dorsiflexor, delayed activation of the hip flexor/extensor, larger activation rate of the knee flexor, and smaller agonist-antagonist co-contraction in lower-limb muscles; (b) larger knee/hip flexion angles, longer ankle dorsiflexion duration, and delayed timing of recovery in joint motions; and (c) earlier downward COM displacements and larger anteroposterior overshooting COM displacements following unpredictable perturbations ($p < 0.05$).

Conclusions: Compared to non-fallers, fallers used more suspensory strategies for reactive standing balance, which compensated for inadequate ankle/hip strategies but resulted in prolonged recovery. A further longitudinal study with a larger sample is still needed to examine the diagnostic accuracies and training values of these identified neuromuscular/kinematic factors in differentiating fall risks and preventing future falls of older people, respectively.

4.3 Introduction

Falls and fall consequences in older adults burden the society heavily and are global health issues (World Health Organization, 2021). Annually, around one in three older adults falls, one in ten older adults has fall-related injuries, and 684,000 fall-related deaths happen worldwide (Moreland et al., 2020; World Health Organization, 2021). However, even the multi-factorial fall-prevention management has shown relatively limited success in fall reduction, especially in older adults with fall histories, i.e., fallers (de Vries et al., 2010). Given that balance and gait disorders are the second leading causes of falls except accidents (World Health Organization, 2021), some in-depth physiological alterations of balance control in older fallers that have remained unidentified may be modifiable to prevent older adults' future falls more effectively.

Several balance control strategies with the involvement of lower limbs have been proposed based on the analyses of kinematics (i.e., postural sways, joint motions) and neuromuscular activities (i.e., electromyographic [EMG] signals) (Kasahara & Saito, 2021). The feet-in-place strategy is commonly employed to keep the whole-body center of mass (CoM) within the base of support (BoS) when external perturbations are not large, which comprises a single or a combination of the ankle strategy, hip strategy, and suspensory strategy (bending knees to lower CoM for stability) (Kasahara & Saito, 2021). The stepping strategy is used to establish a new BoS when the feet-in-place strategy is not enough to overcome the increasing perturbation intensity (Kasahara & Saito, 2021; Tong et al., 2023). Compared to young adults, older adults tended to rely more on the proximal lower-limb joint motions and muscles than the distal ones, and may use the stepping strategy for reactive/compensatory/automatic balance control following unexpected perturbations (Osoba et al., 2019). Apart from the age-related changes in the responses of multiple muscles/joints, prior studies have also shown the interaction effects of age with the perturbation direction and perturbation intensity on the balance control strategies (Allum et al., 2002; Ma & Lee, 2017; Osoba et al., 2019). Nevertheless, the identified age-related kinematic and neuromuscular alterations underlying the reactive balance control may not be directly indicative of fall risks, due to the potential existence of the confounding factor of age. Specific investigations and comparisons of the older adults with and without fall histories (i.e., fallers vs. non-fallers, and excluding the confounding factor of age) are therefore warranted to identify further balance control alterations in older individuals who are prone to falls, and to identify the fall-related factors.

Previous studies have intensively analyzed the stepping strategies and whole-body postural sways to compare the reactive balance control between fallers vs non-fallers (Bair et al., 2016; Batcir et al., 2020; Gerards et al., 2021; Maki et al., 1994; Mille et al., 2013; Sturnieks et al., 2013; Tantisuwat et al., 2011), while the differences in specific joint motions or muscle activities were less focused (Claudino et al., 2017; Studenski & Chandler, 1991; Thompson et al., 2018). Firstly, lower-limb muscle activities during reactive balance control have primarily been examined within a restricted number of lower-limb muscles, i.e., the ankle dorsiflexor/plantarflexor (Claudino et al., 2017; Studenski & Chandler, 1991; Thompson et al., 2018), knee flexor/extensor (Claudino et al., 2017; Thompson et al., 2018), and hip abductor (Claudino et al., 2017). The difference on hip adductor and hip flexor/extensor activation across fallers and non-fallers remained unknown.

Secondly, prior investigations of lower-limb muscle activities have examined only one single EMG parameter in each study (Claudino et al., 2017; Studenski & Chandler, 1991; Thompson et al., 2018). Fallers were reported to exhibit longer EMG onset latency of ankle dorsiflexor following anterior translational perturbations during standing (Studenski & Chandler, 1991), longer EMG onset latencies of hip abductor and knee flexor in the weight-bearing leg following lateral shoulder-impact perturbations during standing (Claudino et al., 2017), and no significantly different agonist-antagonist co-contraction index (CCI) of ankle dorsiflexor-plantarflexor or knee flexor-extensor following optical flow perturbations during walking as compared to non-fallers (Thompson et al., 2018). The existing analysis of timing and amplitude characteristics of EMG signals may have been insufficient, since only the delayed muscular reaction was identified to differentiate fallers from non-fallers (Claudino et al., 2017; Studenski & Chandler, 1991). Thirdly, regarding joint kinematics, interestingly, no prior studies seemed to have compared them in fallers vs. non-fallers during reactive balance control to the best of authors' knowledge. Although fallers exhibited decreased range of motion in lower-limb joints than non-fallers (Lin & Woollacott, 2002), it has been unclear whether the lower-limb joint motions during reactive balance control differ between fallers and non-fallers. More comprehensive analyses of lower-limb muscle activities and joint kinematics are needed to facilitate the understanding of older fallers' balance control strategies.

Reactive balance control strategies are influenced by both the perturbation direction and perturbation intensity (Tong et al., 2023; Zhu et al., 2022), while there is still insufficient evidence to determine whether fallers and non-fallers respond differently to diverse directions or intensities of balance perturbations. Regarding the perturbation intensity, a previous study reported that fallers and non-fallers' difference in stepping strategy was more pronounced following a higher intensity of mediolateral perturbation (Bair et al., 2016), whereas another study did not observe an interaction effect of fall history and perturbation intensity on the reactive stepping strategy following unexpected anterior perturbations (Rogers et al., 2001). Regarding the perturbation direction, prior studies also reported inconsistent differences in postural sway between fallers and non-fallers when responding to unexpected anteroposterior (Maki et al., 1994; Rogers et al., 2001) or mediolateral perturbations (Batcir et al., 2020; Claudino et al., 2017; Fujimoto et al., 2015; Maki et al., 1994). The underlying reasons for these inconsistent findings have not been thoroughly understood/explained. Analyzing neuromuscular responses and joint kinematics during reactive balance control can potentially help better explain how the fall-prone older adults respond to varied levels of threats of suddenly losing balance, which may also provide useful insights for clinical assessments of reactive balance.

The main aim of this study was therefore to explore the older fallers' neuromuscular and kinematic alterations of lower limbs during reactive balance control as compared to non-fallers. Specifically, this study had the research question of how EMG/angle signals varied among the eight different dominant-leg muscles/joint motions, different fall histories, directions, and intensities of unexpected translational perturbations. In addition, how the CoM displacements varied among the six different postural sway directions (i.e., forward/backward, medial/lateral, and upward/downward), different fall histories, perturbation directions, and perturbation intensities were investigated. The timing parameters including onset latency, time to peak, and burst duration and the amplitude parameters including the rate of rise, peak amplitude, and/or agonist-antagonist CCI were analyzed for these signals. We hypothesized that the analyzed parameters during reactive balance control would be affected by the interaction of fall history, muscle/joint motion/postural sway direction, perturbation direction, and perturbation intensity. Further for the simple main effects of fall history, based on the previously available findings related to ageing (Allum et al., 2002; de Freitas et al., 2010; Kasahara & Saito, 2021; Lin & Woollacott, 2002; Osoba et al., 2019; Thompson et al., 2018) and fall histories (Claudino et al., 2017; Studenski & Chandler, 1991; Thompson et al., 2018), we extrapolated that fallers would have the delayed timing and larger amplitudes of proximal muscles' activation/joint motions than non-fallers following a high intensity of unexpected anterior or lateral balance perturbation.

4.4 Methods

4.4.1 Study Design and Subjects

This study was a pilot observational cross-sectional study. Subjects were recruited through convenience sampling. Inclusion criteria were: **1)** aged 65 years old or over, **2)** living in the community independently and been able to walk for 400 m without any assistance, and **3)** fallers (with at least one fall within the past one year) or non-fallers (with no fall within the past one year) in matched age and gender. Exclusion criteria were: **1)** being hospitalized or living in nursing homes for more than six months in the past year; **2)** experienced fall(s) due to traffic or occupational accidents; **3)** diagnosed with cognitive impairment or severe systemic disease (e.g., neuromuscular, renal, hepatic, orthopedic, vestibular, or cardiopulmonary disorders) that impacts or limits physical activities; and **4)** participated in any structured exercise training or strengthening exercises within the past 1 year. A total of twelve older participants were finally eligible for this study. Before being tested, each subject has read and signed an informed consent to participate in this study (Ethics approval agency: Institutional Review Board, The Hong Kong Polytechnic

University; Ethical reference number: HSEARS20201230002). Each subject participated in the experiment once, involving subjective assessments and perturbation trials.

4.4.2 Subjective Assessments

The collection of demographic data (e.g., age, gender, height, body mass), medical history, and fall history was first conducted, followed by the assessments using questionnaires/scale. A fall is defined as an event coming to rest inadvertently on the ground or floor or other lower level and not resulting from an intrinsic or overwhelming hazard (Tinetti et al., 1988). The short Falls Efficacy Scale-International (FES-I) and the Chinese Version of the Physical Activity Scale for the Elderly (PASE-C) were introduced to the subject for the measurement of their fear of falling and physical activity level (Ku et al., 2013; Yardley et al., 2005). The Mini-Balance Evaluation System Test (Mini-BESTest) was used to assess the subject's functional balance performance including the anticipatory postural control, reactive postural control, sensory orientation, and dynamic gait (King & Horak, 2013). The Mini-BESTest was selected due to its established reliability and validity (Godi et al., 2013) together with its comprehensive evaluation of various balance dimensions, particularly for the sub-item of reactive balance, making it more relevant to the topic of this study than other clinical assessments (e.g., Timed Up and Go test, Berg Balance Scale). Then the participant’s dominant leg was determined through the use of three tests: the balance recovery test, the ball-kick test, and the step-up test (Hoffman et al., 1998). The leg that was used more frequently for stepping after being nudged forward, kicking a ball, and stepping onto a stair across a total of nine trials (three trials for each test) was identified as the dominant leg (Hoffman et al., 1998). All subjective assessments were conducted by the same examiner.

4.4.3 Perturbation Trials

Experimental Set-Up

A moving-platform perturbation system was used to induce the unexpected translational perturbations (**Figure 4-1**), with technical details reported in a previous study (Tong et al., 2023). Generally, the platform can move horizontally at a random starting time, with random moving direction and random moving distance/velocity/acceleration (related to different intensity) to constitute an unexpected balance perturbation to the subject standing on it. The whole-body kinematics were collected using an 8-camera motion capture system (Nexus 2.11, Vicon Motion Systems Ltd., Yarnton, UK) that sampled at 250 Hz. An eight-channel Trigno Wireless

Biofeedback System (Delsys Inc, Natick, MA, USA) that sampled at 2000 Hz was used to record the muscular electrical activities. The data collection was synchronized for the three systems (Tong et al., 2023).

Protocol of Perturbation Trials

The procedure of perturbation trials was briefed to the subject first. Subjects were informed in advance to wear their daily footwear, except impractical shoes such as sandals, high heels, ballet shoes and slippers. Each subject was given an identical type of tight shirt and shorts, to optimize the Vicon motion capture and the placement of retroreflective markers and EMG sensors. Before the perturbation trials, EMG sensors and retroreflective markers were placed on the subject. The eight wireless surface EMG sensors were placed on the eight dominant-leg muscles according to the recommendation of Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) project (**Table 4-1**) (Hermens et al., 2000). The major muscles relevant to ankle, knee and hip joint motions were selected, including tibialis anterior (TA), gastrocnemius medialis (GM), rectus femoris (RF), bicep femoris (BF), sartorius (SA), gluteus maximus (GMax), gluteus medius (GMed), and adductor maximus (AM). A standard skin preparation procedure included shaving, cleaning and slightly abraded with alcohol wipes before adhering the EMG electrodes. The sensors were applied on the skin with double-sided tapes (Trigno Sensor Adhesive Interface, Delsys, Boston, MA) with medical tapes to enhance fixation. Then a set of 39 retroreflective markers were attached to the bony landmarks of the head, torso, left and right upper limbs, pelvis and left and right lower limbs (Vicon Motion Systems Limited, 2021). All placements were conducted by the same examiner.

The subject was then instructed to stand with two feet wearing shoes and shoulder-width apart on the middle of platform, and hold a light rod at waist level and close to the trunk to keep the arms from blocking the reflective markers. The subject was told to stand naturally and look forward at the beginning, try the best to maintain balance if feeling the perturbation, and then return to the original foot position marked by the dark-colored tapes as quickly as possible if they have moved the foot. A safety harness system (PG-360, Physio Gait Dynamic Unweighting System, Healthcare International Ltd., Langley, WA, USA) was equipped on each subject as a safety measure during the perturbation.

Each subject then experienced four trials (each consisted of 12 random perturbations) covering a total of 48 unexpected balance perturbations (4 directions × 4 intensities × 3 repetitions), with 5 minutes of rest after each trial. The platform moved horizontally in a pre-determined direction and intensity first, then remained stationary for 12 seconds, and was finally pulled back to its original position. The triggering time, directions (anterior, posterior, medial, lateral), and intensities (highest, high, low, lowest) were randomized and blinded to the subject. Based on the human's limits of stability in different directions and our pilot study results in young adults (Tong et al., 2023), the highest intensity for the anterior, posterior, medial, and lateral directions corresponded to the platform's moving distances of 2.67%, 4.00%, 5.33% and 5.33% of each subject's height, respectively. Videos were recorded in real time during all perturbation trials to enable further observation and analysis of the balance control strategies manually.

Table 4-1. Eight investigated dominant-leg muscles and placement locations for EMG sensors.

Muscle	Location of EMG sensor placement
Hip adductor: Adductor magnus (AM)	Halfway between the pubic tubercle and the medial femoral epicondyle
Hip abductor: Gluteus medius (GMed)	Halfway between the iliac crest to the greater trochanter
Hip flexor: Sartorius (SA)	At 8 cm distal from the ASIS towards the medial epicondyle of the femur
Hip extensor: Gluteus maximus (GMax)	Halfway between the sacral vertebrae and the greater trochanter
Knee flexor: Long head of Bicep femoris (BF)	Halfway between the ischial tuberosity and the lateral epicondyle of tibia
Knee extensor: Rectus femoris (RF)	Halfway between the anterior superior iliac spine (ASIS) and the superior border of the patella
Ankle dorsiflexor: Tibialis anterior (TA)	At 1/3 on the line between the tip of fibula and the tip of medial malleolus
Ankle plantarflexor: Gastrocnemius medialis (GM)	On the most prominent bulge of the muscle

4.4.4 Data Processing

The kinematic data including the whole body's center of mass (CoM), the hip, knee, and ankle joint motions were first processed using the Plug-in Gait full body model. Then the kinematic data and raw EMG data were further processed as below in a custom MATLAB program (MATLAB, The MathWorks, Inc., Natick, Ma, USA). The kinematic data were subtracted by the mean signal value of the 1000-ms baseline interval before the start of each perturbation for normalization. To obtain the CoM displacement relative to the base of support (BoS), the CoM displacement was further subtracted by the displacement of the moving platform (Tong et al., 2023). The raw EMG signals were zeroed to the mean value of the entire perturbation trial, full-wave rectified, and low-pass filtered at 4 Hz with a bi-directional 4th order Butterworth filer to obtain the envelope, then further divided by the mean signal value of the 1000-ms baseline interval before the start of perturbation trial for normalization (Tong et al., 2023; Zhu et al., 2022).

Temporal parameters including the onset latency, time to peak, and burst duration, together with the amplitude parameters including the peak amplitude, rate of rise, and/or agonist-antagonist CCI were analyzed through a custom MATLAB algorithm (**Figure 4-2**). Within 2 seconds after the start of each perturbation, the onset was detected as the first point in time when the corresponding signal value exceeded five times of the standard deviation (SD) over the mean baseline value (mean + 5 SD), and the peak was identified as the point after the onset with the maximum signal value (Ling et al., 2020; Tong et al., 2023; Zhu et al., 2022). The reason for detecting onset and peak within 2 s after a perturbation was because participants were observed to recover balance within this duration following perturbations in the pilot study, and the kinematic or EMG reactions within this duration were considered meaningful to resist the sudden balance loss. Within 9 seconds after the start of each perturbation, the offset was identified as the first point in time after the onset when the corresponding signal value dropped below five times the standard deviation over the mean baseline value (mean + 5 SD) (Hesam-Shariati et al., 2017). The baseline for the onset or offset detection was the 1000-ms interval of a signal before the start of each perturbation. The onset latency indicated the time delay from the start of perturbation to the signal onset, the time to peak indicated that from the start of perturbation to the signal peak, and the burst duration indicated that from the signal onset to offset. The rate of rise was determined as the gradient of the signal rise within a 50-ms period following the onset (Tong et al., 2023; Zhu et al., 2022). The agonist-antagonist CCI within the duration from two muscles' later EMG onset to two muscles' earlier EMG offset was calculated based on the formula in **Figure 4-2** (Di Nardo et al., 2022; Falconer, 1985; Thompson et al., 2018). For each parameter, the mean value of the three perturbations with the same direction and intensity was used in further statistical analyses.

4.4.5 Statistical Analyses

The statistical analyses were performed using SPSS (version 25.0) with the significance level set as 0.05. To examine the difference of baseline subjective assessment data between fallers and non-fallers, the independent sample t tests or Mann-Whitney U tests were used based on the data normality for continuous variables, and Chi-square tests were used for categorical variables. For each parameter (i.e., onset latency, time to peak, peak amplitude, burst duration, peak amplitude, rate of rise, and/or agonist-antagonist CCI), a four-way analysis of variance (ANOVA) and post hoc pairwise comparisons with Bonferroni corrections were conducted to examine the effects of two fall histories, four perturbation directions, four perturbation intensities, and six postural sway directions/eight dominant-leg joint motions/eight dominant-leg muscles/four dominant-leg muscle pairs. When the onset of a signal was absent, the onset latency, time to peak, burst duration were filled with 2000 ms, 2000 ms, and 0 ms, respectively; while the peak amplitude, rate of rise, and agonis-antagonist CCI were all filled with 0. With samples of equal size, the ANOVAs were considered robust even when the assumptions of normality and homogeneity were not fully met (Portney, 2020). Given the objectives, this study focused on the interaction effects of fall history with other factor(s), the main effects of fall history when there were no significant interactions, and the simple effects of fall history when there were significant interactions.

4.5 Results

4.5.1 Subjective Assessment Results

No adverse incident happened during all the experiments. There was no significant difference in the number of medications, age, body mass, height, foot length, BMI, short FES-I score or the PASE-C score between the participated older fallers and older non-fallers (**Table 4-2**). Nevertheless, the Mini-BESTest score of fallers was significantly lower than that of non-fallers ($p < 0.05$).

4.5.2 Balance Control Strategies

Fallers were more likely to have stepping responses than non-fallers. The unexpected translational perturbations mainly induced the feet-in-place strategies (567/576, 98.4%), and three subjects (3/12, 25.0%) had stepping responses following nine perturbations (9/576, 1.6%). Specifically, following three highest-intensity medial perturbations, one non-faller had the responses of the non-dominant leg including stepping toward the perturbation direction, performing leg abduction, and elevating the leg (3/576, 0.5%). One faller took a backward step using the non-dominant leg together with several small steps following a highest-intensity anterior perturbation (1/576, 0.2%). The other faller stepped backward using the non-dominant leg following the highest-intensity (2/576, 0.3%) and high-intensity (1/576, 0.2%) anterior perturbations. Additionally, this individual stepped toward the perturbation direction using both legs in response to a low-intensity posterior perturbation (1/576, 0.2%), and with the non-dominant leg in response to a highest-intensity medial perturbation (1/576, 0.2%).

4.5.3 CoM Displacements

The mean changes in CoM displacements over time (n = 12, **Figure 4-3**) together with the onset latency, time to peak, peak amplitude, and burst duration of CoM displacement (mean ± SD, n = 12, **Figure 4-4**) are displayed for each postural sway direction, each perturbation intensity, and each perturbation direction in participated older fallers and older non-fallers.

Four-way ANOVAs showed significant interaction effects of fall history and other factors on the onset latency (fall history × postural sway direction, $p < 0.05$), time to peak (fall history × postural sway direction, $p < 0.05$), and peak amplitude (fall history × direction × postural sway direction, $p < 0.05$) of CoM displacement.

Figure 4-4 illustrates the significant differences between older fallers and older non-fallers. Compared to non-fallers, the fallers' onset latency of CoM displacement was significantly longer in the backward direction, but significantly shorter in the forward and downward directions ($p < 0.05$); the fallers' time to peak CoM displacement was significantly longer in the backward direction, but significantly shorter in the downward direction ($p < 0.05$); the fallers' peak CoM displacement was significantly larger in the forward and downward directions following anterior perturbations, in the backward direction following posterior perturbation, and in the forward direction following both the medial and lateral perturbations ($p < 0.05$).

4.5.4 Dominant-leg Joint Motions

The mean changes of dominant-leg joint motions over time (n = 12, **Figure 4-5**) together with the angle onset latency, time to peak angle, peak angle, and angle burst duration (mean ± SD, n = 12, **Figure 4-6**) are displayed for each joint motion, each perturbation intensity, and each perturbation direction in fallers and non-fallers.

Four-way ANOVAs showed significant interaction effects of fall history and other factors on the angle onset latency (fall history × direction × joint motion, $p < 0.05$), time to peak angle (fall history × direction × joint motion, $p < 0.05$), peak angle (fall history × joint motion, $p < 0.05$), and angle burst duration (fall history × joint motion, $p < 0.05$).

Significant differences between fallers and non-fallers are indicated in the **Figure 4-6**. Compared to non-fallers, the fallers' angle onset latency was significantly longer in the hip adduction, hip extension, and knee extension following anterior perturbations, and in the ankle plantarflexion following medial perturbations ($p < 0.05$); the fallers' time to peak angle was significantly longer in the hip adduction, hip flexion, hip extension, and knee extension following anterior perturbations as well as in the ankle plantarflexion following medial perturbations, but was significantly shorter in the ankle plantarflexion following lateral perturbations ($p < 0.05$); the fallers' peak angle was significantly larger in the hip flexion and knee flexion ($p < 0.05$); the fallers' angle burst duration was significantly longer in the ankle dorsiflexion. ($p < 0.05$).

4.5.5 EMG Signals of Dominant-leg Muscles

The mean changes of EMG signals over time (n = 12, **Figure 4-7**) together with the EMG onset latency, rate of EMG rise, time to peak EMG amplitude, EMG burst duration, and agonist-antagonist CCI (mean ± SD, n = 12, **Figure 4-8**) are presented for each dominant-leg muscle (pair), each perturbation intensity, and each perturbation direction in fallers and non-fallers.

Four-way ANOVAs showed significant interaction effects of fall history and other factors on the EMG onset latency (fall history × muscle, $p < 0.05$), rate of EMG rise (fall history × muscle, $p < 0.05$), and EMG burst duration (fall history × muscle, $p < 0.05$; fall history × direction, $p < 0.05$). The main effect of fall history was observed at the time to peak EMG amplitude ($p < 0.05$) and the agonist-antagonist CCI ($p < 0.05$).

Significant differences between fallers and non-fallers are indicated in the **Figure 4-8**. Compared to non-fallers, the fallers' EMG onset latency was significantly longer for the hip flexor and hip extensor ($p < 0.05$); the fallers' rate of EMG rise was significantly smaller for the ankle dorsiflexor but was significantly larger for the knee flexor ($p < 0.05$); the fallers' time to peak EMG amplitude was significantly longer ($p < 0.05$); the fallers' EMG burst duration was significantly longer for the hip abductor and ankle dorsiflexor, but was significantly shorter for the hip flexor ($p < 0.05$); the fallers' EMG burst duration was also significantly longer following the anterior and posterior perturbations, but was significantly shorter following the medial perturbations ($p < 0.05$); the fallers' agonist-antagonist CCIs were significantly smaller in the investigated muscle pairs ($p < 0.05$).

Table 4-2. Subjective assessment results (categorical variable: ratio; continuous variable: mean ± SD) of twelve subjects.

	Faller (n = 6, 3 male & 3 female)	Non-faller (n = 6, 3 male & 3 female)	Significance (p value)
Number of falls	1.3 ± 0.5	0	/
Number of medications	1.0 ± 1.1	0.3 ± 0.5	0.279
Age (year)	71.5 ± 4.6	69.2 ± 2.9	0.316
Body mass (kg)	55.6 ± 8.4	61.4 ± 13.0	0.381
Height (cm)	157.9 ± 8.7	162.0 ± 7.9	0.406
BMI (kg/m^2)	22.2 ± 2.0	23.3 ± 4.4	0.587
Leg length (cm)	77.3 ± 6.3	80.8 ± 4.6	0.297
Dominant leg (right/left)	5/1	6/0	0.296
Short FES-I (score)	12.2 ± 2.4	11.5 ± 6.0	0.332
PASE-C (score)	139.5 ± 73.2	148.1 ± 34.6	0.802
Mini-BESTest (score)	23.3 ± 1.5	26.0 ± 0.9	**0.004**

BMI: body mass index. **FES-I**: fall efficacy scale-international. **PASE-C**: physical activity scale of elderly-Chinese. **Mini-BESTest**: mini-balance evaluation system test.

Figure 4-1. The moving-platform perturbation system with the subject.

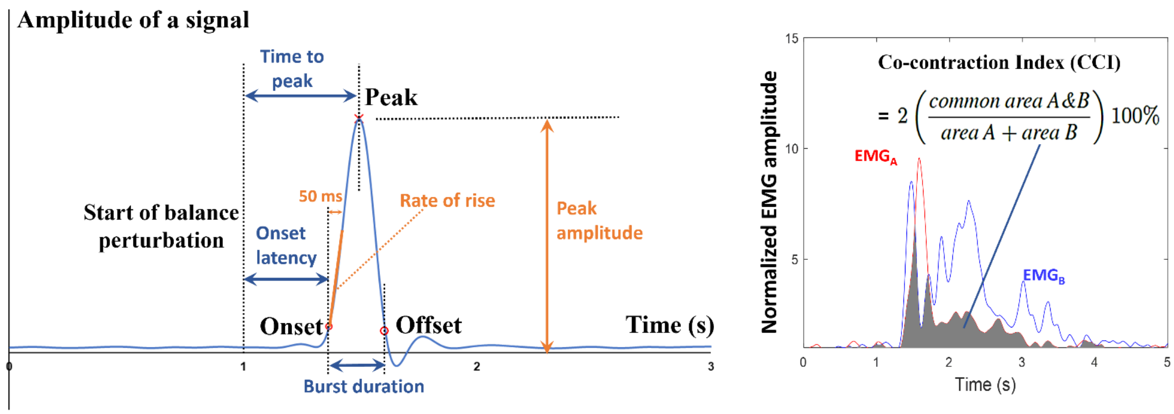

Figure 4-2. Illustrations of the analyzed temporal and amplitude parameters.

EMG: electromyographic.

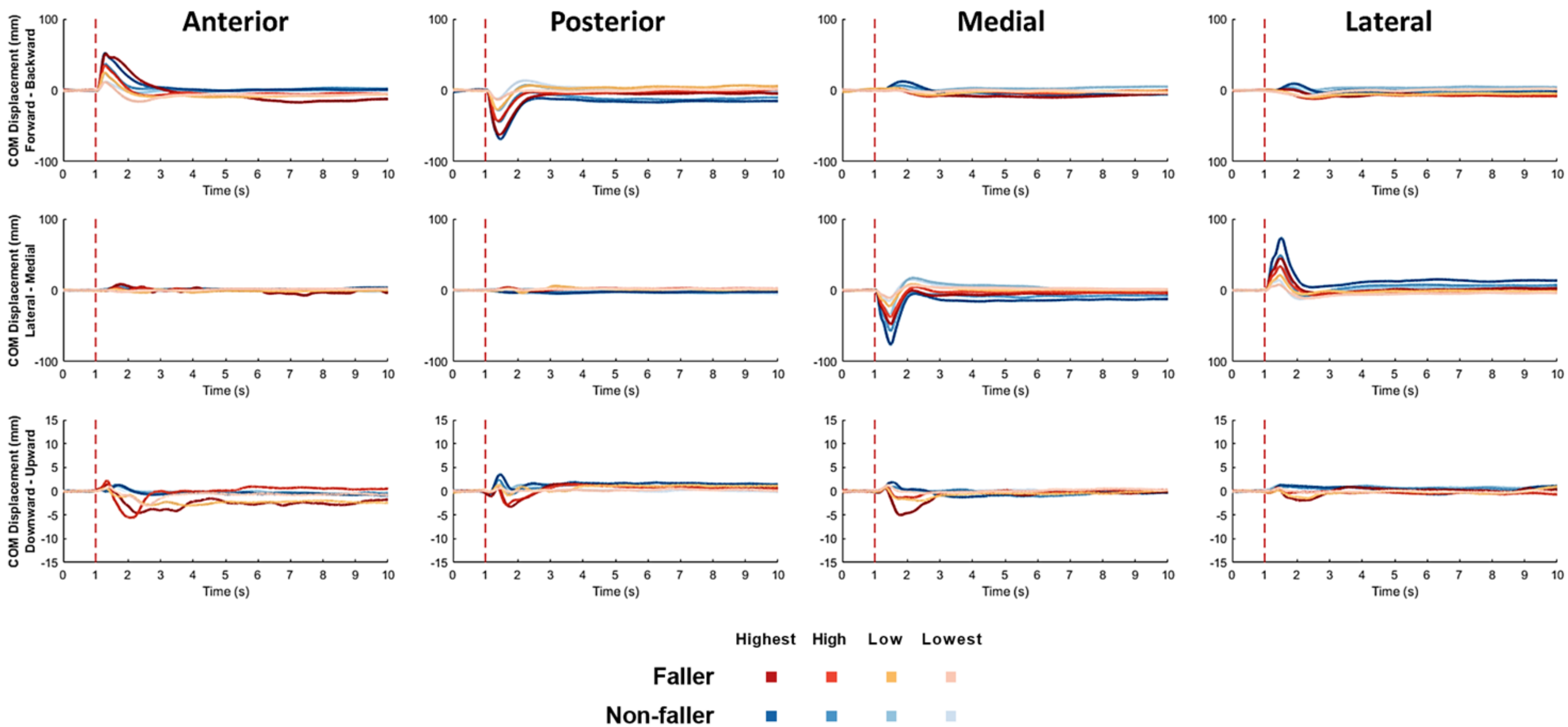

Figure 4-3. The mean forward/backward, medial/lateral, and upward/downward CoM displacements in fallers (n = 6) and non-fallers (n = 6) following perturbations with different directions and intensities.

The red dotted line denotes the start of balance perturbation. CoM: center of mass.

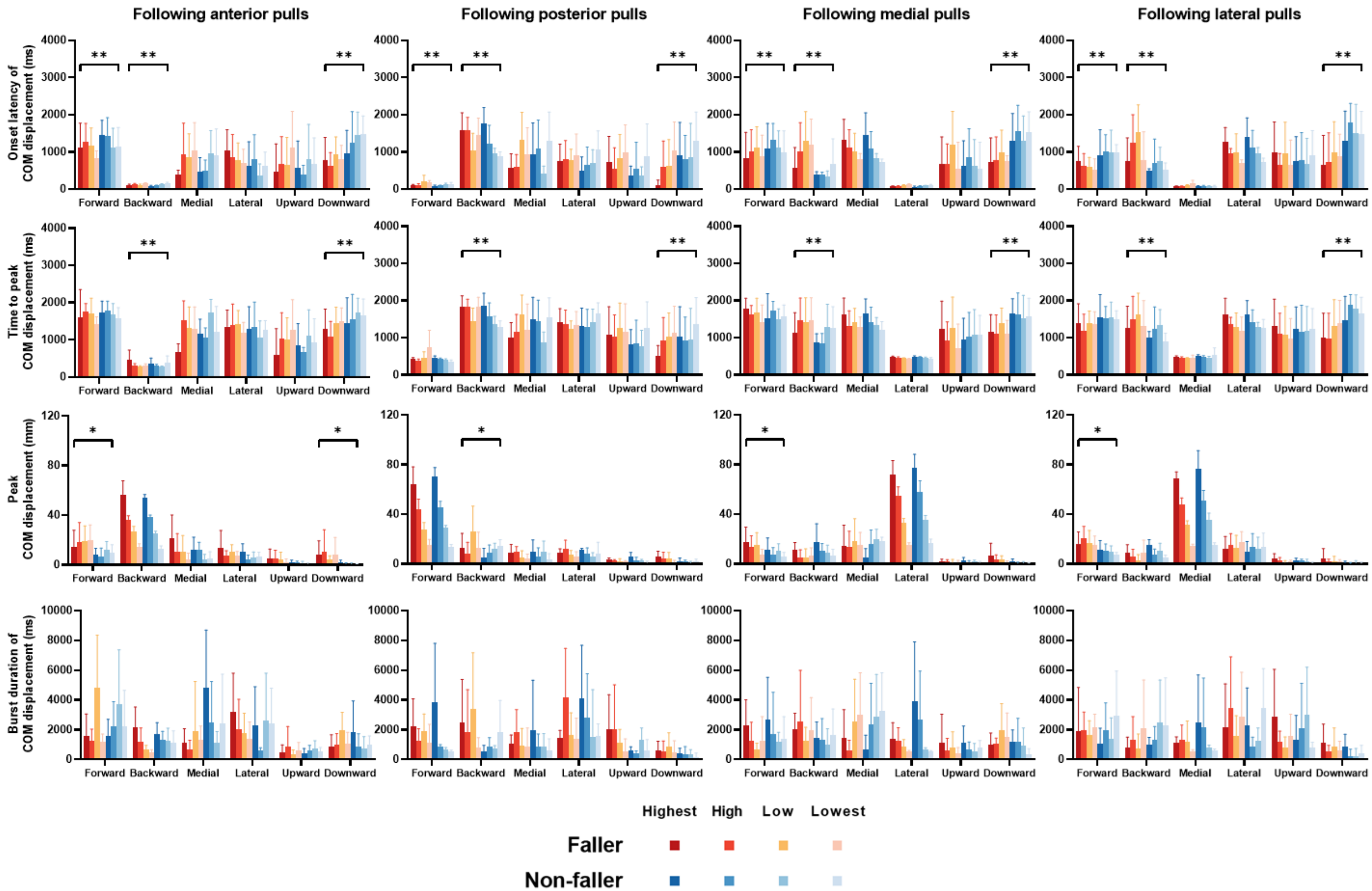

Figure 4-4. Onset latency, time to peak, peak amplitude, and burst duration of CoM displacement (n = 6, mean ± SD).

Significant effects of fall history are indicated by the ⌐¬ (p < 0.05). ** denotes the significant effect of fall history at a certain postural sway direction ("fall history × postural sway direction" interaction). * denotes the significant effect of fall history at a certain postural sway direction and following a certain direction of perturbation ("fall history × postural sway direction × direction" interaction).
CoM: center of mass. SD: standard deviation.

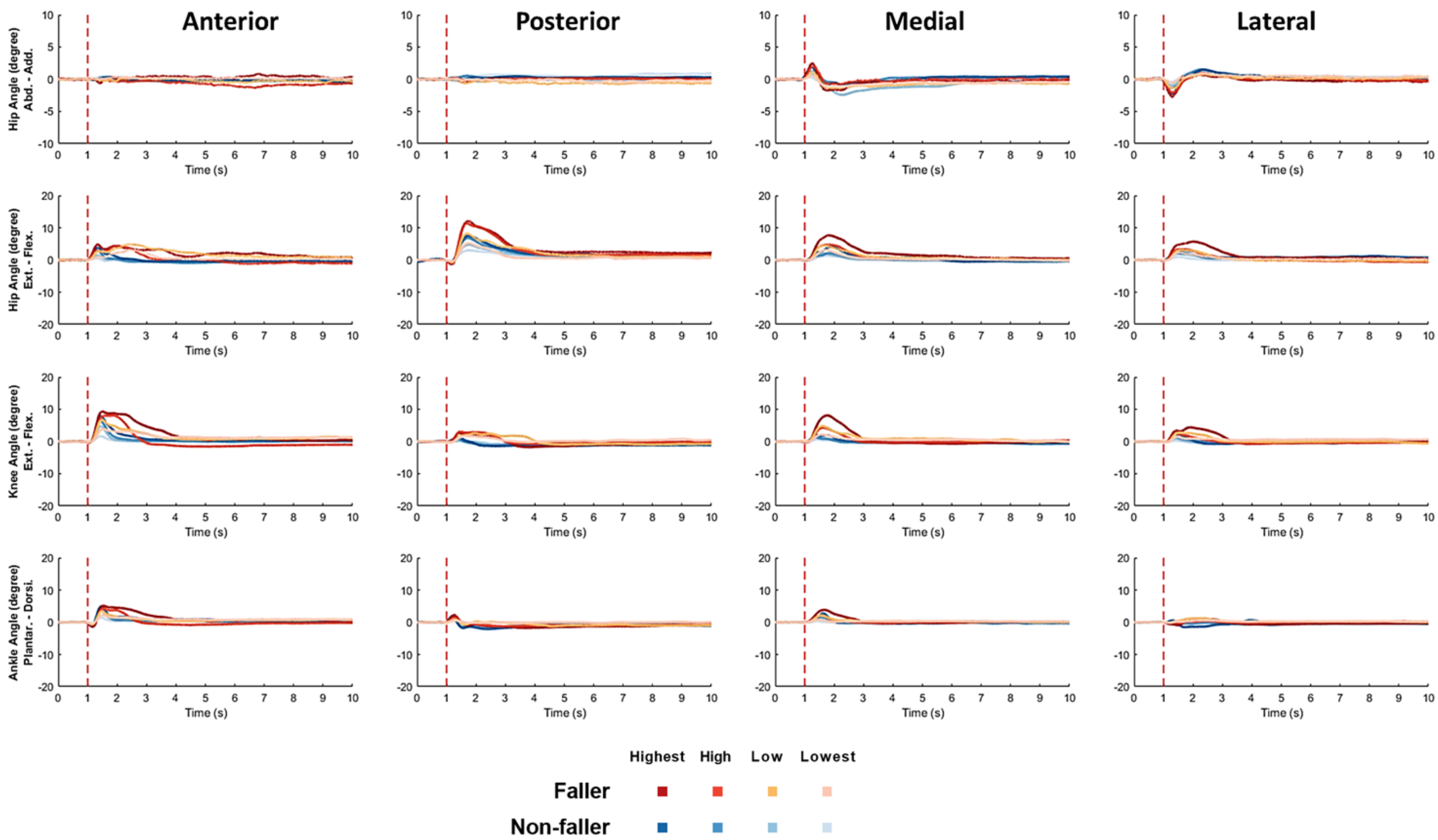

Figure 4-5. The mean change for each of the eight dominant-leg joint motions in fallers (n = 6) and non-fallers (n = 6) following perturbations with different directions and intensities.

The red dotted line denotes the start of balance perturbation. Add: adduction. Abd: abduction. Flex: flexion. Ext: extension. Dorsi: dorsiflexion. Plantar: plantarflexion. SD: standard deviation.

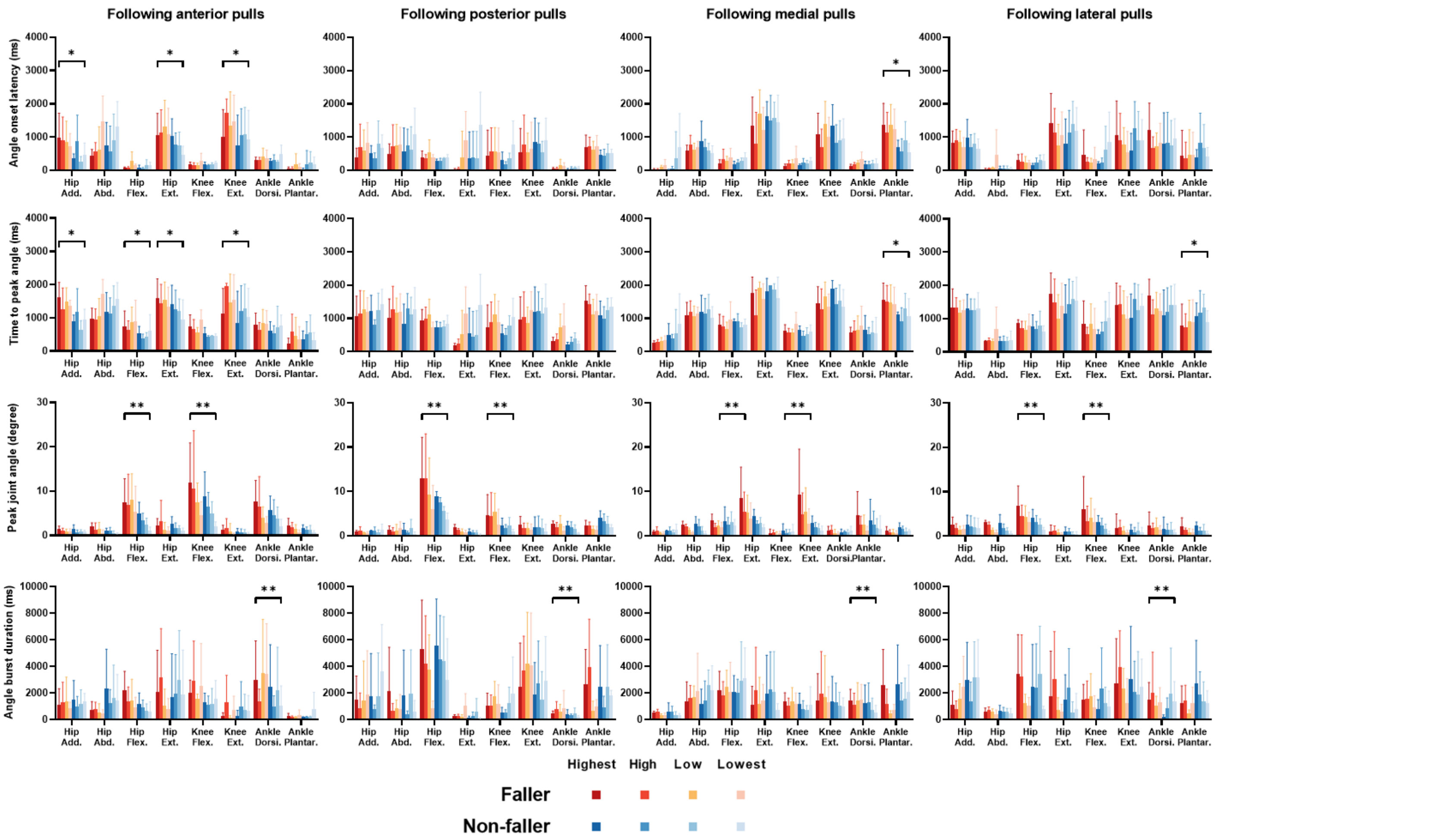

Figure 4-6. Onset latency, time to peak, peak amplitude, and burst duration of joint motion (n = 6, mean ± SD).

Significant effects of fall history are indicated by the ⎴ ($p < 0.05$). * denotes the significant effect of fall history at a certain joint motion and following a certain direction of perturbation ("fall history × joint motion × direction" interaction). ** denotes the significant effect of fall history at a certain joint motion ("fall history × joint motion" interaction). Add: adduction. Abd: abduction. Flex: flexion. Ext: extension. Dorsi: dorsiflexion. Plantar: plantarflexion. SD: standard deviation.

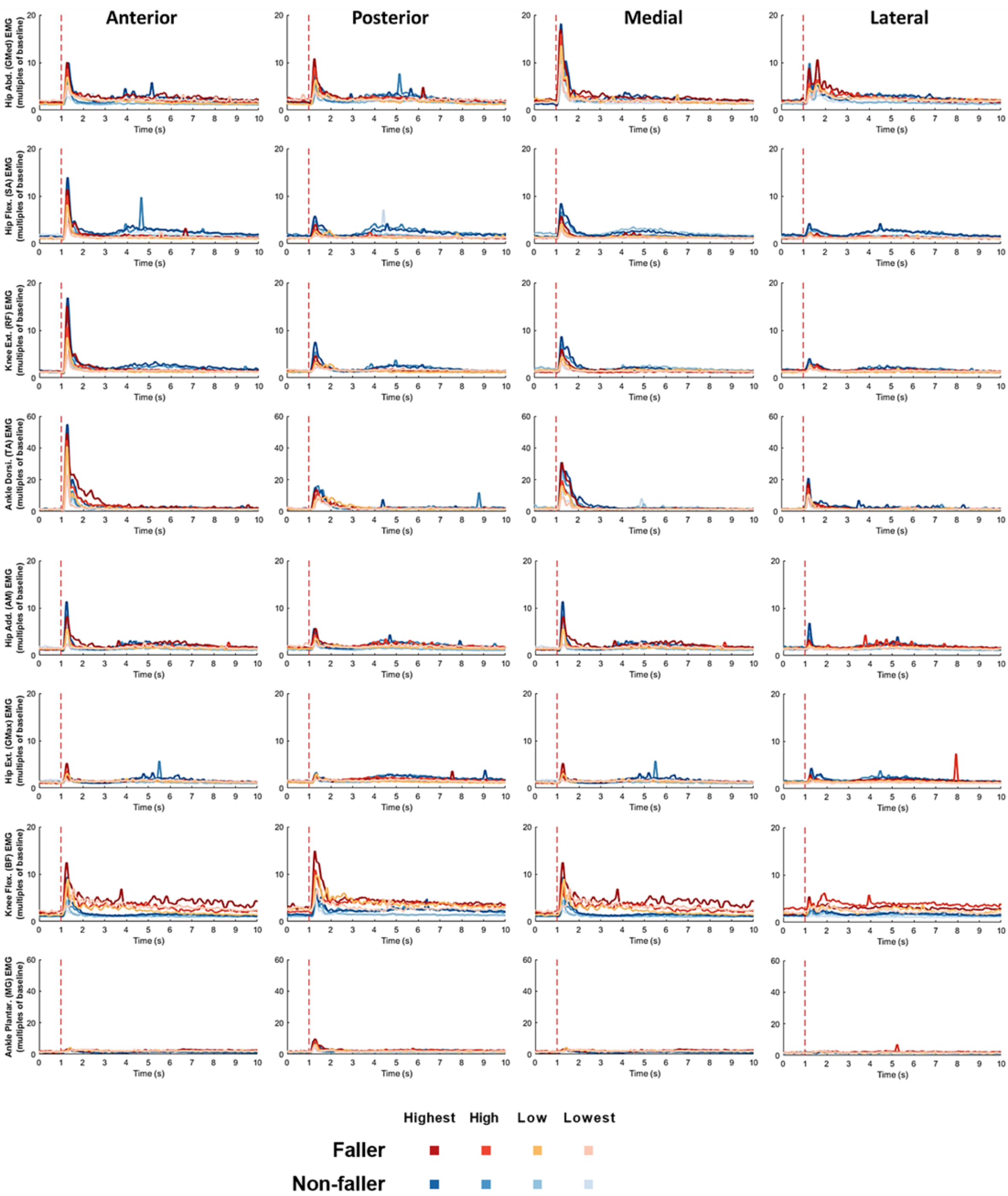

Figure 4-7. The mean EMG signal change for each of the eight dominant-leg muscles in fallers (n = 6) and non-fallers (n = 6) following perturbations with different directions and intensities.

The red dotted line denotes the start of balance perturbation. EMG: electromyographic. CCI: co-contraction index. SD: standard deviation. GMed: gluteus medius. SA: sartorius. RF: rectus femoris. TA: tibialis anterior. AM: adductor magnus. GMax: gluteus maximus. BF: biceps femoris. MG: gastrocnemius medialis. Add: adductor. Abd: abductor. Flex: flexor. Ext: extensor. Dorsi: dorsiflexor. Plantar: plantarflexor.

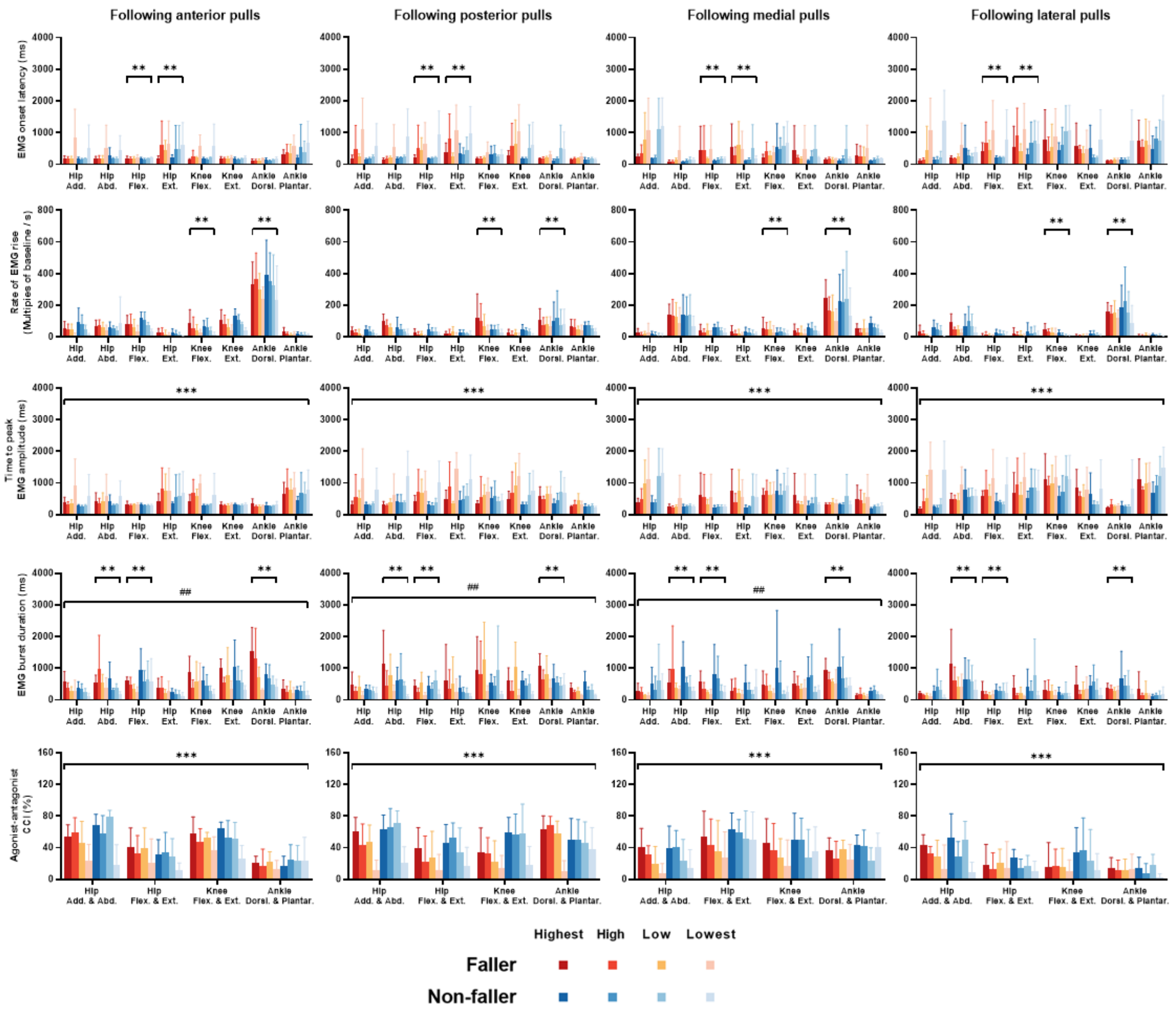

Figure 4-8. Onset latency, rate of rise, time to peak, burst duration, and agonist-antagonist CCI of EMG signal (n = 6, mean ± SD).

Significant effects of fall history are indicated by the ⎴ ($p < 0.05$). *** denotes the significant main effect of fall history (no interaction of “fall history” and any other factor). ** denotes the significant effect of fall history at a certain muscle (“fall history × muscle” interaction). ## denotes the significant effect of fall history following a certain direction of perturbation (“fall history × direction” interaction). EMG: electromyographic. CCI: co-contraction index. SD: standard deviation. Add: adductor. Abd: abductor. Flex: flexor. Ext: extensor. Dorsi: dorsiflexor. Plantar: plantarflexor.

4.6 Discussions

This study comprehensively examined the effects of fall history on reactive standing balance in community-dwelling older adults, by focusing on dominant-leg muscle activities and joint motions. Partially in line with our hypotheses, the effects of “fall history” on the investigated outcomes during reactive balance control interacted with the “muscle/joint motion/postural sway direction” and “perturbation direction” but not with the “perturbation intensity”. Specifically, compared to older non-fallers, older fallers demonstrated slowed activation of ankle/hip muscles while tending to use a suspensory strategy for reactive balance control, as supported by a series of neuromuscular alterations and joint kinematics. These new insights reflect possible reasons for fallers’ decreased balance capability and indicate their utilization of prolonged and enlarged (and even overreacted) compensatory strategies for preserving postural stability (Lin & Woollacott, 2002). Developing some future assessment tools based on the identified parameters may be helpful to screen and identify the fallers from non-fallers in the community-dwelling adults. Furthermore, interventions targeting these identified fall-related alterations may lead to more effective solutions for improving reactive balance control and preventing recurrent falls in older fallers. Details are discussed below.

4.6.1 Fallers Tended to Use Suspensory Strategies Following Unexpected Perturbations: Neuromuscular and Kinematic Mechanisms

The primary finding of this study was that fallers have tended to use the suspensory strategy to maintain standing balance following the unexpected translational perturbations as compared to non-fallers. This strategy has enabled fallers to promptly compensate for their insufficient initiation of ankle and hip strategies, but it has led to their prolonged and overacted balance recovery.

Fallers have exhibited a decreased speed in response to an unexpected threat of losing standing balance, as indicated by their decreased activation rate of ankle dorsiflexor and the delayed EMG onset timing of hip flexor/extensor compared to non-fallers. This could be attributed to the potential degradation in any components along the sensorimotor pathway, including sensory input (feedback from external perturbation), central organization, and motor output (Osoba et al., 2019). Ankle dorsiflexor's activation immediately following the start of perturbation has been in the first line to resist the sudden loss of balance, as this study observed its largest rate of EMG rise among the eight dominant-leg muscles following both anteroposterior and mediolateral perturbations, and our pilot studies in young adults also reported this (Tong et al., 2023). With ageing, humans may shift from a distal-to-proximal strategy to a proximal-to-distal strategy to maintain balance, compensating for the difficulties of generating sufficient ankle torque (Woollacott, 1993). This study further proved that such phenomenon was more pronounced in older fallers than older non-fallers. On the other hand, fallers have shown delayed EMG onset timing of hip flexor and hip extensor, along with reduced EMG burst duration of hip flexor, compared to non-fallers. These alterations could partly restrict the initiation of the hip strategy, which is the second line of defense against the sudden loss of balance. The delayed activation of hip muscles aligns with and may be explained by previous morphological observations that, fallers had reduced density of skeletal muscle fibers and increased intramuscular adipose issues in gluteus muscles compared to non-fallers (Inacio et al., 2014). A prior study also reported delayed neuromuscular activation in reactive standing balance, with fallers exhibiting later EMG onset timing of hip abductor and knee flexor in the loading leg than non-fallers following the unexpected lateral perturbations exerted on the shoulder (Claudino et al., 2017). The discrepancy in the affected muscles could be attributed to the different perturbation methods.

A series of kinematic and neuromuscular alterations in fallers when facing unexpected translational balance perturbations have indicated their prominent use of suspensory strategies as compared to non-fallers. In the absence of sufficient ankle and hip muscle activation, fallers have utilized the suspensory strategy, i.e., the third strategy to resist sudden loss of balance by lowering the CoM to increase limit of stability and absorb the external perturbation (Cheng, 2016; Kasahara et al., 2015; Nijhuis et al., 2007). The increased activation rate of knee flexor, generally decreased agonist-antagonist co-contraction of lower-limb muscles, and larger knee/hip flexion in fallers may have facilitated this strategy. Interestingly, our findings differ from a prior study that reported no differences in postural sway timing or amplitude between fallers and non-fallers following lateral shoulder-impact perturbations (Claudino et al., 2017), suggesting that different body segment perturbations may elicit distinct reactive balance control strategies. Additionally, while previous research linked greater co-contraction to more joint stability and poorer balance control (Falk et al., 2022; Schulz et al., 2013), this study has observed that older fallers even with poorer balance performance than non-fallers (lower Mini-BESTest scores) were able to reduce agonist-antagonist co-contractions of lower-limb muscles and achieve larger knee/hip flexion for a suspensory strategy. On top of them, this study has observed fallers with the longer activation durations of ankle dorsiflexor and hip abductor together with the longer ankle dorsiflexion duration than non-fallers, which may be necessary for maintaining a knee bending posture during the suspensory strategy.

Fallers' balance control strategies in the current study, however, have required prolonged recovery time and caused overreactions. This is evidenced by their neuromuscular and kinematic alterations as below. Firstly, fallers' delayed time to peak activation may have suggested their reduced motor unit recruitment and firing rate in response to external perturbations (de Freitas et al., 2010). Secondly, fallers have shown longer time to peak hip flexion angle following anterior perturbations, longer burst durations of ankle dorsiflexion following all perturbations, and delayed timing of recovery joint motions following anterior/medial perturbations than non-fallers. Thirdly, both fallers and non-fallers have demonstrated the major postural sway that was opposite to the direction of an unexpected translational perturbation because of inertia (Tong et al., 2023), while fallers have shown larger overshooting postural sways when recovering to initial positions following the unexpected anteroposterior perturbations, as indicated by their larger forward peak CoM displacements following anterior perturbations and larger backward ones following posterior perturbations as compared to non-fallers. These findings have indicated that sudden perturbations could pose greater challenges to older fallers. Fallers' more prominent overshoots of backward postural sways, as compared to non-fallers, have also been previously reported following the anterior waist-pull perturbations (Ho & Bendrups, 2002). Additionally, a prior study found that fallers had more variable and delayed recovery steps than non-fallers in perturbed walking (Gerards et al., 2021), which could be corroborated by the observed fallers' larger overshooting postural sways and delayed timing of overshooting lower-limb joint motions in this study. The slowed but exaggerated postural adjustments seemed to reveal the ineffective strategies used by the older adults with fall histories for reactive balance control.

This study also observed large within-group variations in some of the analyzed parameters during reactive balance control. For example, although the faller group was identified to exhibit longer EMG onset latencies in hip muscles than the non-faller group, the within-group variations were large, as indicated by the standard deviations. This reflects the fact that even individuals with the same fall status may respond differently during reactive balance control. On the one hand, a previous study reported that fallers had a higher variation in walking stability than non-fallers after unpredictable perturbations (Gerards et al., 2021). Given its potential in differentiating fall status, the variability of reactive balance performance should not be neglected. On the other hand, analysis of individual variations in reactive balance control within the faller group (e.g., coefficient of variation) could be also crucial for understanding personalized balance intervention requirements, which merits further investigation.

4.6.2 Fallers Had Altered Responses to Different Perturbation Directions

The secondary finding of this study was that fall history showed interaction effects with perturbation direction, but not with perturbation intensity, on the older adults' neuromuscular and kinematic responses during reactive balance control.

Fallers have shown distinct responses to anteroposterior and mediolateral perturbations compared to non-fallers. Regarding the kinematics, a previous study reported the larger CoM path displacement in fallers compared to non-fallers following the mediolateral translational perturbations (Batcir et al., 2020). Our findings have further revealed that fallers' larger postural sways were specifically in the forward direction following the mediolateral perturbations and fallers had overshooting postural sways following the anteroposterior perturbations. These responses could be partly attributed to fallers' delayed timing of recovery joint motions compared to non-fallers following the anterior/medial perturbation. Regarding the neuromuscular responses, fallers have exhibited longer EMG burst durations in dominant-leg muscles following anteroposterior perturbations compared to non-fallers. This could also explain fallers' overshooting postural sways. Additionally, this could explain why fallers had more non-dominant leg stepping following anteroposterior perturbations than non-fallers, as more body weight was loaded on the dominant leg. Conversely, fallers have exhibited shorter EMG burst durations of dominant-leg muscles than non-fallers following medial perturbations, resulting in fallers' fewer non-dominant leg steps following medial perturbations compared to non-fallers.

Notably, this study has found no differences in the responses of fallers compared to non-fallers to varied intensities of unexpected perturbations. Previous studies reported the inconsistent results regarding the interaction effect of fall history and perturbation intensity on the stepping characteristics following waist-pull perturbations (Bair et al., 2016; Rogers et al., 2001). Our finding has further built on evidence following the unexpected translational perturbations and suggested that fallers' neuromuscular/kinematic responses to the different intensities of perturbations, which primarily induced feet-in-place strategies, have been similar to those of non-fallers.

4.6.3 Strengths and Limitations

To the best of authors' knowledge, this study offers a preliminary but more in-depth exploration of the differences in eight major lower-limb muscles' activation or lower-limb joint kinematics during reactive balance control between older fallers and older non-fallers. With the comprehensive analyses of temporal and amplitude characteristics of these investigated signals, this study has built knowledge upon the prior investigations that focused on a limited number of muscles and EMG parameters, and has addressed the gap of limited research on joint kinematics in fallers vs. non-fallers. The mechanisms of fall-prone older adults' decline of reactive balance control and compensatory strategies could be better understood with the findings of this study.

This study has several limitations that shall be acknowledged, and the findings shall be interpreted with caution.

(1) Given the small sample sizes of recruited older fallers and older non-fallers, it should be noted that the findings of this pilot study using four-way ANOVAs can provide only preliminary insights. Results of alternative analyses for the “fall history” factor, using independent sample t-tests for normally distributed data and Mann-Whitney U tests for non-normally distributed data, are presented in the supplementary file of published article (Zhu, Hung, et al., 2025). An example of fall-related kinematic and neuromuscular differences following high or highest intensity of perturbations is presented in **Table 4-3**. Similar to the results of four-way ANOVA and post hoc pairwise comparisons, the results of independent sample t-tests (or Mann-Whitney U tests) have demonstrated that fallers tended to use suspensory strategies to compensate for the delayed/reduced activation of ankle and hip muscles following unexpected moving-platform perturbations.

Table 4-3. Significant fall-related kinematic and neuromuscular differences following high or highest intensity of perturbations.

Direction	Intensity	Muscle / Joint motion / Postural sway direction	Parameter	Unit	Faller (n = 6)	Non-faller (n = 6)	*p* value
Anterior	Highest	Hip adduction	Time to peak angle	ms	1610 ± 455	894 ± 190	0.005
Anterior	Highest	Hip abduction	Peak angle	°	2 ± 1	1 ± 1	0.031
Anterior	Highest	Medial	Time to peak COM displacement	ms	720 (280)	1270 (532)	0.037
Anterior	High	Knee flexion	Time to peak angle	ms	501 (537)	440 (33)	0.004
Anterior	High	Downward	Peak COM displacement	mm	3 (15)	1 (2)	0.016
Anterior	High	Medial	Burst duration of COM displacement	ms	600 (1086)	1603 (2278)	0.037
Posterior	Highest	Ankle plantarflexion	Peak angle	°	2 ± 1	4 ± 2	0.047
Posterior	High	Knee extensor	Rate of EMG rise	baseline/s	17.4 ± 20.2	42.8 ± 15.3	0.034
Posterior	High	Ankle plantarflexion	Time to peak angle	ms	1435 ± 299	979 ± 378	0.043
Posterior	High	Hip extension	Peak angle	°	1 ± 0	1 ± 0	0.031
Medial	Highest	Hip adductor	EMG onset latency	ms	250 ± 75	170 ± 30	0.048
Medial	Highest	Ankle plantarflexor	EMG burst duration	ms	154 ± 37	295 ± 85	0.004
Medial	Highest	Ankle plantarflexion	Angle onset latency	ms	1382 ± 633	710 ± 202	0.048
Medial	Highest	Knee extension	Time to peak angle	ms	1575 (954)	1987 (186)	0.036
Medial	High	Hip extensor	Time to peak EMG amplitude	ms	288 (184)	243 (69)	0.025
Medial	High	Ankle plantarflexor	Rate of EMG rise	baseline/s	27.3 ± 24.5	58.7 ± 23.1	0.045
Medial	High	Hip extension	Time to peak angle	ms	1102 ± 757	2000 ± 0	0.034
Medial	High	Downward	Peak COM displacement	mm	2 (5)	0 (1)	0.046
Lateral	Highest	Hip extension	Time to peak angle	ms	2000 (383)	1247 (1662)	0.049
Lateral	High	Hip flexor	EMG onset latency	ms	407 (677)	209 (63)	0.004
Lateral	High	Knee flexor	Rate of EMG rise	baseline/s	36.4 (20.1)	9.1 (14.4)	0.037
Lateral	High	Knee extensor	EMG onset latency	ms	546 (258)	237 (116)	0.006
Lateral	High	Knee extensor	Time to peak EMG amplitude	ms	574 ± 147	336 ± 100	0.008
Lateral	High	Downward	Onset latency of COM displacement	ms	245 (1400)	2000 (315)	0.021
Lateral	High	Downward	Time to peak COM displacement	ms	623 (1256)	2000 (171)	0.021
Lateral	High	Downward	Peak COM displacement	mm	1 (2)	0 (0)	0.049

Note: For normally-distributed data, "mean ± standard deviation" is used to indicate descriptive statistics, and independent sample t test was used to examine the difference between older fallers and older non-fallers. For non-normally distributed data, "median (interquartile range)" is used to indicate descriptive statistics, and Mann-Whitney U test was used to examine the difference between older fallers and older non-fallers. "

(2) This study did not ensure that the faller and non-faller groups had matched ranges of motion in the investigated joints or muscle strength, which may confound the identified between-group differences in reactive balance performance. Additionally, this study only focused on reactive balance control in fall-prone people. It is important to note that the causes of loss of balance or falls are not confined to this and are multi-factorial (e.g., environmental factors).

(3) Although diagnosis of cognitive impairment was an exclusion criterion for this study and the testers ensured participants understood task instructions, a clinical scale such as the Montreal Cognitive Assessment (MoCA) should be used to better quantify the older participants' cognitive function in future research.

(4) The cut-off frequency used for EMG low-pass filtering in this study may not be optimal for all of the investigated eight leg muscles. Further attempts are warranted to determine the cut-off frequency for processing each respective muscle's EMG signal based on the frequency spectrum analysis. Additionally, this study did not conduct maximal voluntary contraction tests but used the baseline EMG signal value in unperturbed standing for EMG amplitude normalization. It is therefore important to note that the rate of EMG rise and agonist-antagonist CCI in this study reflected the extent to which the perturbation task utilized the activation required for normal standing rather than the maximal activation.

(5) This study did not explore gender-specific variations in balance control by having equal representation of male and female participants in a group. Future studies could consider specifically examining the gender-specific difference in older people with a larger sample size to address this issue.

4.6.4 Implications for Future Clinical Practice

This study has preliminary implications for assessing and training reactive balance control in future applications. On the one hand, the identified fall-related kinematic and neuromuscular factors may inform clinical practice. Reactive balance training may need to be prescribed more for the community-dwelling older adults with fall histories in the future, considering their generally delayed peak activation of lower-limb muscles. A most recent review has reported that the perturbation-based balance training and stepping training can improve the slowed reaction time in confronting with sudden loss of balance (Bhagwat & Deodhe, 2023). This is promising as fallers' degradation in neuromuscular timing can be modified. Our findings further imply that more focus/efforts may need to be put on the ankle and hip muscle power training in older fallers. Although the proximal hip and knee muscles were previously reported to be more affected with aging following unexpected perturbations (Hall & Jensen, 2002), the findings of this study may suggest that the training of ankle muscles should not be ignored, especially in older fallers.

On the other hand, with the advancement of wearable sensors and real-time monitoring systems, future studies could also consider employing these tools to improve balance assessment and training in fall-prone individuals. Real-time monitoring and analysis of the reaction speed (e.g., EMG onset latency, rate of EMG rise) of an older client's ankle dorsiflexor or hip flexor/extensor for reactive balance control may potentially enhance the fall-risk assessment on top of the current reactive balance test. The EMG-based biofeedback may potentially enhance power training for ankle or hip muscles, while the quantitative results on muscle reaction speed may help therapists offer more personalized feedback and guidance during reactive balance training.

Nonetheless, we acknowledge that these implications need validation in future longitudinal studies with larger sample sizes, where the diagnostic accuracy of slower activation in ankle and hip muscles in differentiating older adults' fall risks, as identified in this pilot study, should be examined.

4.7 Conclusion

This pilot study found that older fallers' kinematic and neuromuscular alterations in resisting unpredictable translational perturbations could be indicators of both the decline and the compensation of reactive balance control. Compared to non-fallers, fallers had a decreased activation rate in the ankle dorsiflexor and delayed activation in the hip flexor/extensor, thereby resorting to the suspensory strategy for quickly responding to external perturbations. The increased activation rate of knee flexor, decreased agonist-antagonist co-contraction of lower-limb muscles, enlarged knee and hip flexion, and earlier downward postural sways in fallers could be the basis of their prioritizing of suspensory strategies as compared to non-fallers. However, fallers' balance control strategies required prolonged recovery in lower-limb joint motions and caused overreactions in postural sways. A further longitudinal study with a larger sample is merited to verify these fall-related factors, which could enhance the identification of fall-prone people and provide insights for more targeted fall-prevention strategies.

Chapter 5 Exploring Reactive Balance Control Induced by Waist-pull Perturbations in Young Adults (Study 3)

5.1 Chapter Summary

This chapter includes the contents of study 3 in this PhD project. On top of the findings of literature reviews on the lower-limb rapid strength and the response speed of reactive balance control, this study validated the customized waist-pull system for inducing reactive standing balance in young adults and focused on the response speed and sequence of multiple lower-limb muscles/joint moments/joint powers/joint motions.

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How Does Lower Limb Respond to Unexpected Balance Perturbations? New Insights from Synchronized Human Kinetics, Kinematics, Muscle Electromyography (EMG) and Mechanomyography (MMG) Data

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Abstract: Making rapid and proper compensatory postural adjustments is vital to prevent falls and fall-related injuries. This study aimed to investigate how, especially how rapidly, the multiple lower-limb muscles and joints would respond to the unexpected standing balance perturbations. Unexpected waist-pull perturbations with small, medium and large magnitudes were delivered to twelve healthy young adults from the anterior, posterior, medial and lateral directions. Electromyographical (EMG) and mechanomyographical (MMG) responses of eight dominant-leg muscles (i.e., hip abductor/adductors, hip flexor/extensor, knee flexor/extensor, and ankle dorsiflexor/plantarflexors) together with the lower-limb joint angle, moment, and power data were recorded. The onset latencies, time to peak, peak values, and/or rate of change of these signals were analyzed. Statistical analysis revealed that: (1) agonist muscles resisting the delivered perturbation had faster activation than the antagonist muscles; (2) ankle muscles showed the largest rate of activation among eight muscles following both anteroposterior and mediolateral perturbations; (3) lower-limb joint moments that complied with the perturbation had faster increase; and (4) larger perturbation magnitude tended to evoke a faster response in muscle activities, but not necessarily in joint kinetics/kinematics. These findings provided insights regarding the underlying mechanism and lower-limb muscle activities to maintain reactive standing balance in healthy young adults.

Keywords: balance perturbation; balance control; onset latency; time to peak; electromyography (EMG); mechanomyography (MMG); skeletal muscle; reactive balance response; compensatory postural adjustment (CPA); waist-pulling perturbation

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1. Introduction

Falls and fall-related injuries adversely affect about one-third of the older population globally [1]. To avoid a fall, it is vital to make prompt and proper postural adjustments to maintain or recover balance, i.e., keeping the center of body mass (CoM) within the base of support (BoS) [2]. Reactive balance response, or compensatory postural adjustment (CPA), describes how human beings react to a sudden perturbation. It refers to the postural control and the activation of muscles after the central nervous system detects the balance perturbation [3]. Throughout the pathway of motor output, an in-depth investigation of how the multiple lower-limb muscles and joints react rapidly to maintain standing balance is needed, which can facilitate our better understanding of the mechanisms underlying CPAs and the fall-prevention strategies.

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5.2 Abstract

Making rapid and proper compensatory postural adjustments is vital to prevent falls and fall-related injuries. This study aimed to investigate how, especially how rapidly, the multiple lower-limb muscles and joints would respond to the unexpected standing balance perturbations. Unexpected waist-pull perturbations with small, medium and large magnitudes were delivered to twelve healthy young adults from the anterior, posterior, medial and lateral directions. Electromyographical (EMG) and mechanomyographical (MMG) responses of eight dominant-leg muscles (i.e., hip abductor/adductor, hip flexor/extensor, knee flexor/extensor, and ankle dorsiflexor/plantarflexor) together with the lower-limb joint angle, moment, and power data were recorded. The onset latencies, time to peak, peak values, and/or rate of change of these signals were analyzed. Statistical analysis revealed that: (1) agonist muscles resisting the delivered perturbation had faster activation than the antagonist muscles; (2) ankle muscles showed the largest rate of activation among eight muscles following both anteroposterior and mediolateral perturbations; (3) lower-limb joint moments that complied with the perturbation had faster increase; and (4) larger perturbation magnitude tended to evoke a faster response in muscle activities, but not necessarily in joint kinetics/kinematics. These findings provided insights regarding the underlying mechanism and lower-limb muscle activities to maintain reactive standing balance in healthy young adults.

5.3 Introduction

Falls and fall-related injuries adversely affect about one-third of the older population globally (Kalache et al., 2007). To avoid a fall, it is vital to make prompt and proper postural adjustments to maintain or recover balance, i.e., keeping the center of body mass (CoM) within the base of support (BoS) (Pai et al., 2000). Reactive balance response, or compensatory postural adjustment (CPA), describes how human beings react to a sudden perturbation. It refers to the postural control and the activation of muscles after the central nervous system detects the balance perturbation (Chen et al., 2015). Throughout the pathway of motor output, an in-depth investigation of how the multiple lower-limb muscles and joints react rapidly to maintain standing balance is needed, which can facilitate our better understanding of the mechanisms underlying CPAs and the fall-prevention strategies.

CPAs can be rarely assessed in the subjective balance scales or questionnaires. Most of the clinical tests, e.g., the Berg Balance Scale (BBS), the Performance-Oriented Mobility Assessment (POMA), and the Short Physical Performance Battery (SPPB), evaluate only the anticipatory postural adjustments (APAs) by instructing clients to accomplish some predictable balance challenging tasks (Chen et al., 2015). An exception is the Mini Balance Evaluation Systems Test (Mini-BESTest), which includes the CPA assessment items by suddenly putting the clients' bodies in anterior, posterior, and lateral inclined postures (Franchignoni et al., 2010). The CPAs have been more widely studied in a variety of laboratory equilibrium-disturbing or fall-simulation experiments, where the unexpected perturbations were exerted on different body parts (e.g., shoulder (Kanekar & Aruin, 2014), waist/pelvis (Inacio et al., 2019; Rietdyk et al., 1999; Vlutters et al., 2018), foot (Ma & Lee, 2017; Runge et al., 1999)) to disturb the original postural stability in either static (e.g., perturbed quiet standing (Inacio et al., 2019; Ma & Lee, 2017; Rietdyk et al., 1999; Runge et al., 1999), suddenly tether-released inclined standing (Hsiao-Wecksler, 2008)) or dynamic (e.g., induced slipping (Qu et al., 2012) or tripping (Mirjam Pijnappels et al., 2005; M. Pijnappels et al., 2005) during walking) states. These approaches make it possible to elicit the CPAs and evaluate the reactive balance capability in human beings.

From externally to internally, the whole-body postural sways, the kinematics (e.g., angles) and kinetics (e.g., moments and power) of lower-limb joints, the contraction and activation patterns of lower-limb muscles can all affect how fast the CPAs are made. To quantitatively depict such rapid response, some parameters like the onset latency, the time to peak amplitude, and the rate of change were proposed.

Regarding the whole-body postural sways, previous studies found that the center of pressure (CoP) had larger displacement than the CoM when responding to the unexpected balance perturbation (Kanekar & Aruin, 2014; Rietdyk et al., 1999; Zemkova et al., 2016). In this way, the CoM was kept within the BoS and the standing balance could be maintained. In addition, the time to peak CoM displacement has been reported to vary following the different directions of unexpected platform movements (Ma & Lee, 2017). Regarding the lower-limb joint angles, the onset latencies of hip, knee, and ankle joint motions were studied: (1) during standing, with perturbation induced by a forward-moving (Hwang et al., 2009) or backward-moving platform (Runge et al., 1999) in the sagittal plane, and (2) during walking, with perturbation induced by waist-pulling (Vlutters et al., 2018). Regarding the lower-limb joint kinetics, previous studies analyzed: (1) the joint moment responses in the sagittal plane following balance perturbations induced by a backward-moving platform (Runge et al., 1999), (2) the hip and ankle moment responses (Rietdyk et al., 1999), and the hip power response (Inacio et al., 2019) in frontal plane following waist-pull perturbations. M. Pijnappels et al. (2005) also reported a smaller rate of ankle plantarflexion, knee flexion, and hip extension moment development in the sagittal plane in the stance leg of participants who fell after the experimentally induced tripping. However, previous kinetic analyses have put limited focus on the temporal parameters. It remained unclear how fast the multiple lower-limb joint moments and power would react to unexpected standing perturbations. It is expected that we could have a better understanding of how the hip, knee and ankle joints coordinate to maintain standing balance following perturbations, upon studying the exact time when various lower-limb joints begin to react and reach peaks. Further studies are needed.

Regarding the lower-limb muscle electrical activities, previous studies have investigated the muscle's EMG onset latencies (Bates et al., 2021; Blomqvist et al., 2014; de Freitas et al., 2010; Hwang et al., 2009; Runge et al., 1999) and the time to peak of EMG amplitude (Bates et al., 2021; Blomqvist et al., 2014; de Freitas et al., 2010), following unexpected standing balance perturbations induced by a moving platform. Mirjam Pijnappels et al. (2005) found that in contrast to young people, older people showed increased onset latency and decreased rate of EMG rise in the dorsal muscles of the stance leg after unexpected tripping during walking. Previous studies also reported the age-related reduction in the hip abductors/adductors' rate of EMG rise following unexpected standing balance perturbation induced by the mediolateral waist-pulling (Inacio et al., 2019). However, most of these studies have only investigated the ankle/knee muscles' EMG signals; and very limited previous studies have concurrently evaluated the rapid responses of hip abductor/adductors, hip flexor/extensor, knee flexor/extensor, and ankle dorsiflexor/plantarflexor. It is expected that studying multiple lower-limb muscles' reactions and activation patterns could help further uncover the underlying mechanism of CPAs. In addition to EMG, mechanomyography (MMG) is another technology that can measure the lateral vibration and mechanical activities of skeletal muscles (Orizio, 1993). The onset latencies of EMG, MMG and joint moment signals may enable a more detailed characterization of the motor output pathway, and provide insights on whether the slower balance response is more attributed to the delayed neuromuscular activation, the delayed onset of muscle contraction, or the slower force propagation from muscle to tendon (E. Cè et al., 2020). Thus, MMG may serve as an additional tool to characterize the rapid responses of muscle contractile properties, and merits further studies.

Humans react differently to the varying magnitudes of unexpected balance perturbations. With the increasing perturbation magnitudes, larger lower-limb joint responses and larger amplitudes of muscle activities would be evoked to maintain standing balance (Runge et al., 1999). The larger perturbation magnitude could even alter the EMG onset sequence of lower-limb muscles from distal-to-proximal to proximal-to-distal activation (Manchester et al., 1989), and change the pattern of postural adjustment from the "ankle strategy" to the "hip strategy", "mixed ankle and hip strategy" or "stepping strategy" (Horak & Nashner, 1986; Shumway-Cook & Woollacott, 1995). However, previous studies have mostly reported the effects of different balance perturbations on the choice of responding strategies. It is still unclear whether the faster lower-limb responses are required to resolve a larger balance perturbation, which warrants further investigation.

To fill the above-mentioned research gaps, this study aimed to comprehensively investigate and uncover the more in-depth underlying mechanisms of maintaining standing balance, by investigating how the multiple lower-limb muscles and joints react rapidly following balance perturbations. It would answer the research questions of: (1) how do the onset latencies and the time to peak of the hip, knee, and ankle joints' kinetic and kinematic data respond to the different magnitudes of waist-pull perturbation in sagittal and frontal planes; and (2) how do the onset latencies, the time to peak and the rate of rise of eight lower-limb muscles' EMG and MMG data respond to the different magnitudes of waist-pull perturbation in sagittal and frontal planes. It was hypothesized that both the temporal parameters and the rate of change would be different across the eight lower-limb motions (hip abduction/adduction, hip flexion/extension, knee flexion/extension, and ankle dorsiflexion/plantarflexion), across the eight lower-limb muscles (hip abductor/adductor, hip flexor/extensor, knee flexor/extensor, and ankle dorsiflexor/plantarflexor), and across the three different perturbation magnitudes (small, medium, and large).

5.4 Methods

5.4.1 Participants

A total of 12 healthy young adults aged 18-25 years old (6 males and 6 females) were recruited in this study through convenience sampling. Participants having any neuromuscular, orthopedic, or heart disease were excluded. Ethics approval was granted by the university's Institutional Review Board (reference number: HSEARS20210122001). All participants signed the written informed consent before experiment. The whole experiment was accomplished in the Human Locomotion Laboratory (Department of Biomedical Engineering, The Hong Kong Polytechnic University).

5.4.2 Equipment

As shown in , the waist-pull system for inducing balance perturbations mainly involved: (1) an aluminum alloy frame, (2) four servo motors (130-07725AS4, Wenzhou Guomai Electronics Ltd., China), (3) four pulling strings, and (4) a safety harness. The four servo motors had a rated output power of 2000 W and a rated torque of 7.7 Nm. One end of the pulling string (1.2mm-diameter braided polyethylene wire) was wired around a 40 mm-diameter driving wheel connecting to the servo motor, and the other end went through a turn on the frame and was tied to the belt worn by the participant at pelvis level. A commercially available harness system (PG-360, Physio Gait Dynamic Unweighting System, Healthcare International Ltd., USA) was used to prevent the participant from falling during the experiment.

The 3D motion capture and analysis system (Nexus 2.11, Vicon Motion Systems Ltd., Yarnton, UK) was used to collect the pelvic and lower-limb kinematics and kinetics during the experiment. The sampling rates of the eight cameras (Vicon Vantage 5, Vicon Motion Systems Ltd., Yarnton, UK) and the two floor-mounted force plates (OR6, Advanced Mechanical Technology, Inc., Watertown, MA, USA) were 250 Hz and 1000 Hz, respectively. Based on the Plug-in Gait Full-body Model, a total of 39 reflective markers were attached to each participant’s anatomical landmarks, including four on head (bilateral front head, bilateral back head), five on torso (spinous process of the 7th cervical vertebra, spinous process of 10th thoracic vertebra, right scapula, sternal notch, xiphoid process of the sternum), four on pelvis (bilateral anterior superior iliac spine, bilateral posterior superior iliac spine), fourteen on each of bilateral upper limbs (bilateral acromion, upper arm, elbow, forearm, radial side of wrist, ulnar side of wrist, 3rd metacarpal head), and twelve on each of bilateral lower limbs (bilateral thigh, knee, shank, lateral malleolus, heel, 2nd metatarsal head) (Vicon Motion Systems & UK, 2022). The waist-pull system and the Vicon system were synchronized.

The synchronized eight-channel Trigno Wireless Biofeedback System (SP-W06-016, Delsys Inc., USA) was used for EMG and MMG data collection. Each Trigno Avanti Sensor (dimension: 37 mm × 27 mm × 13 mm; mass: 14 g) comprised an EMG sensor (double-differential silver bar electrodes; electrode size: 5 mm × 1 mm; inter-electrode distance: 10 mm; common mode rejection ratio > 80 dB; amplifier gain: 909; analog Butterworth filter bandwidth: 20-450 Hz) and a 9-axis inertial measurement unit which involved a 3-axis accelerometer to serve as the MMG sensor. Based on the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) guideline (Hermens et al., 1999), eight EMG and MMG sensors were placed longitudinally over the eight lower-limb muscles after skin preparation. The EMG and MMG signals were sampled at 2000 Hz and 250 Hz, respectively.

Table 5-1. Investigated muscles and locations of EMG sensor placement.

Muscle	Location
Ankle dorsiflexor: tibialis anterior (TA)	at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus.
Ankle plantarflexor: medial gastrocnemius (MG)	on the most prominent bulge of the muscle.
Knee extensor: rectus femoris (RF)	halfway between the anterior superior iliac spine (ASIS) and the superior boarder of the patella.
Knee flexor: semitendinosus (ST)	halfway between the ischial tuberosity and the medial epicondyle of the tibia.
Hip flexor: iliopsoas (IL)	at 3–5 cm distal from the ASIS (Jiroumaru et al., 2014a).
Hip extensor: gluteus maximus (GMax)	halfway between the sacral vertebrae and the greater trochanter.
Hip abductor: gluteus medius (GMed)	halfway between the iliac crest to the greater trochanter.
Hip adductor: adductor maximus (AM)	halfway between the pubic tubercle and the medial femoral epicondyle (Hides et al., 2016).

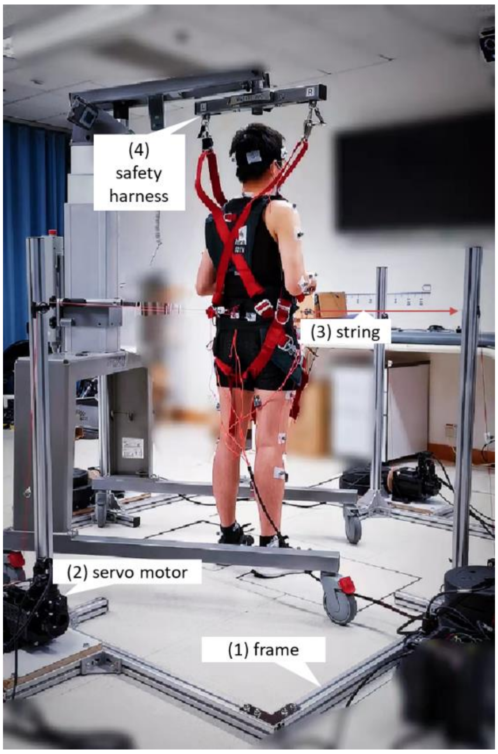


Figure 5-1. Experimental setup and the waist-pull system for inducing sudden balance perturbations in anterior, posterior, medial, and lateral directions.

5.4.3 Protocols

2.3.1. Demographic Data Collection

Each participant's demographic data, including age, height, and body weight, was collected. Physical activity in the past 7 days and the degree of concerns over falling were evaluated via the International Physical Activity Questionnaire-Short version (IPAQ-S) (Lee et al., 2011) and the Fall Efficacy Scale-International (FES-I) short version (Kempen et al., 2008), respectively. The participant's dominant leg was determined as the leg that made a step more often, in response to the six forward and backward shoulder nudges/pushes performed by the researcher (Hoffman et al., 1998).

2.3.2. Instrumented Data Collection

During preparation, the waist-pull system and hardness system were set up on each participant, ensuring that: (1) each participant stood with the two feet in shoes and shoulder-width apart on the two force plates separately, (2) the belt was tied just above the height of posterior superior iliac spine (PSIS), and (3) the harness jacket would not restrict anteroposterior or mediolateral postural responses within a certain range (Bair et al., 2016). Each participant's foot positions were then marked with dark-colored tapes, and the length/height of the harness system was fixed. Then, each participant was given five minutes to sit down and rest to avoid fatigue in the following formal perturbed standing trials.

Before the balance perturbation, each participant was required to hold a light rod in front of the body at the waist level to make sure the reflective markers were detectable. They were instructed to "stand still and look forward; when perturbed by the pulling, try best to maintain postural balance without making steps; if the foot moves, try to return to the initial/original place marked by the dark-colored tapes as soon as possible." Each participant was also instructed that after the start of pulling, their hands could respond freely (Bair et al., 2016).

Each participant accomplished three perturbed standing trials with a total of 36 waist-pulls (3 magnitudes x 4 directions x 3 repetitions). The magnitudes (small, medium, and large), the directions (anterior, posterior, medial, and lateral), and the interval time between two pulls (12–15 s) were pseudo-randomized for each participant. Participants were also blinded to the sequence of the waist-pulls during the experiment. Based on the results from our pilot study and the published literature, the maximal anterior, posterior, medial, and lateral pulling displacements were set as the 6% (Pai et al., 1998), 4% (Luchies et al., 1994), 8%, and 8% (Singh et al., 2017) of each participant's height, respectively. The small, medium, and large pulling magnitude corresponded to the 1/3, 2/3, and 3/3 of the maximal pulling displacement, respectively. Each pull's duration, displacement, and velocity were measured based on the flash time of infrared light, and the movement of the reflective markers fixed on the strings. Videos were taken to record each participant's behavioral performance during the experiment.

5.4.4 Data Processing

The kinematic and kinetic data (i.e., joint angles, joint moments, joint power, CoM and CoP) were processed by using the Plug-in Gait Dynamic model of the Vicon system (Kadaba et al., 1989). The CoP and joint moments were further filtered using a low-pass 4th order Butterworth filter with a 15 Hz cut-off frequency (Laudani et al., 2021). The CoM, CoP, joint angle, and joint moment data was zeroed to the mean of the 1000-ms baseline value before each separate pull.

The muscle activity data as measured by the EMG and MMG sensors was processed by the MATLAB program (MATLAB 2019b, The MathWorks, Inc., Natick, MA, USA). The EMG data were zeroed to the mean values obtained from the whole perturbed standing trial, full-wave rectified, and low-pass filtered using a 4th order and bi-directional Butterworth filter with a cut-off frequency of 4 Hz (Kim & Hwang, 2018). To extract MMG data, the accelerometry signals perpendicular to the skin, i.e., the z-axis components, were used and processed. The signals were firstly band-pass filtered using a 4th order Butterworth filter (5-50 Hz), then full-wave rectified, and further smoothed via a moving-average filter of temporal window of 0.1 s (Plewa et al., 2018). The EMG or MMG signal envelope was then normalized to the 1000-ms baseline mean value at the beginning of each perturbed standing trial.

The start of balance perturbation was defined as the time point when the motorized waist-pull system started running. The onset and peak points of various signals that reached the corresponding peak values were identified within 2 seconds after the start of balance perturbation. The onset time of body CoM, body CoP, joint angle, joint moment, joint power, muscle EMG, and muscle MMG data was defined as the first time point when the corresponding normalized signal/data value went beyond five times of the standard deviation (SD) from the baseline value (mean + 5SD) (Ling et al., 2020). The baseline value was calculated as the mean over the 1000-ms interval before the start of balance perturbation. As shown in ., the onset latencies were referred to as the time delays between the start of balance perturbation and the onset of the corresponding signals. The time to peak was referred to as the delayed time between the start of balance perturbation and the peak of the corresponding signal. The rate of EMG rise was referred to as the slope of EMG signal rise within the 50-ms interval after its onset.

For each outcome (i.e., onset latency, time to peak, peak value, or rate of change), the three values following the three repeated perturbations in same direction and magnitude were calculated and averaged for each participant. These mean values of the 12 participants were used for the following statistical analysis.

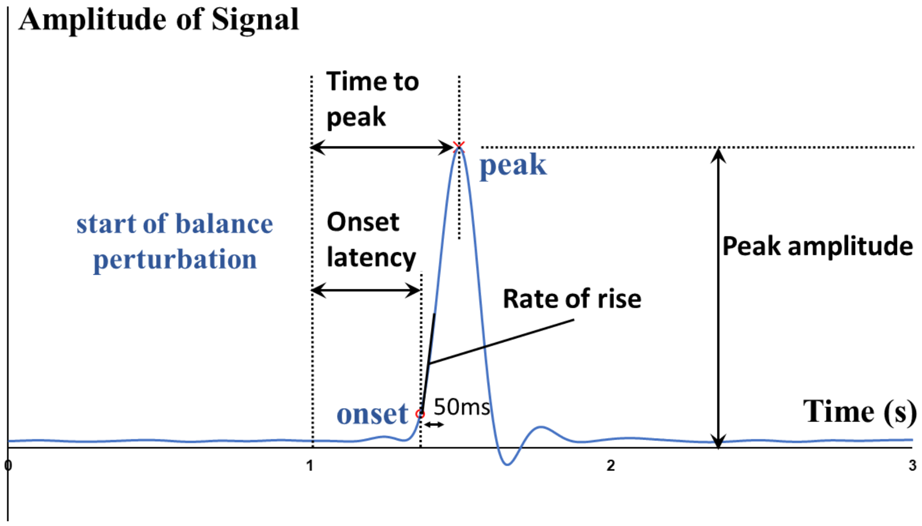


Figure 5-2 Illustration of the definitions of outcome measures.

5.4.5 Statistical Analyses

Statistical analyses were performed using IBM SPSS version 25. Intraclass correlation coefficients (ICCs) were calculated to examine the test-retest reliability of three pulls with the same direction and magnitude. Two-way factorial analysis of variance (ANOVA) and post hoc pairwise comparisons with Bonferroni corrections were used to separately examine: (1) the effects of three different perturbation magnitudes and the spatial difference of eight muscles' electrical and mechanical activities (for EMG and MMG signals; "magnitude" x "muscle"); (2) the effects of three different perturbation magnitudes and the lower-limb joint difference (for angle, moment and power data; "magnitude" x "joint motion"); and (3) the effects of three different perturbation magnitudes and the CoM-CoP difference ("magnitude" x "postural sway"); on the measured onset latency, the time to peak, the specific peak values, and/or the rate of increase. The significance level was set as 0.05.

5.5 Results

A total of 12 healthy young adults (age: 20.9 ± 0.7 years; gender: 6 males and 6 females; height: 169.9 ± 6.9 cm; weight: 58.3 ± 6.2 kg) participated in this study. No fall or other adverse events occurred during experiments, and participants all reported that the harness system did not restrict their movements. As shown in , the ICC values demonstrated good test-retest reliability of the pulling duration, displacement, and velocity of the waist-pull system in this study. The mean and standard error values across the 12 participants are presented in the figures to illustrate the signal changes (i.e., CoM, CoP, angle, moment, power, EMG or MMG) following perturbations.

Table 5-2. Demographic data (mean ± SD) of twelve participants.

	Female (n=6)	Male (n=6)	Total (n=12)
Age (year)	21.0 ± 0.6	20.8 ± 0.8	20.9 ± 0.7
Height (cm)	165.3 ± 4.8	174.5 ± 5.6	169.9 ± 6.9
Body mass (kg)	57.3 ± 2.6	59.3 ± 8.7	58.3 ± 6.2
BMI (kg/m2)	21.0 ± 1.2	19.4 ± 2.0	20.2 ± 1.8
Dominant leg	Right (n=6)	Right (n=6)	Right (n=12)
Leg length (cm)	85.2 ± 3.2	88.8 ± 4.6	87.0 ± 4.3
IPAQ-S (Kcal/week) [1]	2089.4 ± 1965.5	2451.5 ± 1994.6	2270.5 ± 1897.4
FES-I short version [2]	10.2 ± 1.2	11.2 ± 4.8	10.7 ± 3.3

[1]International Physical Activity Scale - Short version. [2]Fall Efficacy Scale - International.

Table 5-3. Mean and ICC values of pulling parameters examining the reliability of the waist-pull system (n=12).

		Duration (s)		Max. Displacement (cm)		Normalized Max. Displacement (%height)		Max. Velocity (m/s)	
Direction	Magnitude	Mean	ICC	Mean	ICC	Mean	ICC	Mean	ICC
	Large	0.396	0.978*	11.4	0.987*	6.7%	0.971*	0.349	0.908*
Anterior	Medium	0.373	0.993*	7.4	0.991*	4.4%	0.983*	0.238	0.995*
	Small	0.347	0.984*	3.8	0.962*	2.2%	0.927*	0.127	0.778*
	Large	0.263	0.981*	7.0	0.927*	4.1%	0.847*	0.338	0.829*
Posterior	Medium	0.247	0.978*	4.7	0.964*	2.8%	0.909*	0.239	0.857*
	Small	0.231	0.968*	2.4	0.983*	1.4%	0.956*	0.124	0.994*
	Large	0.531	0.984*	16.0	0.950*	9.5%	0.681*	0.351	0.716*
Medial	Medium	0.498	0.993*	10.5	0.956*	6.2%	0.730*	0.252	0.759*
	Small	0.465	0.984*	5.3	0.964*	3.1%	0.807*	0.136	0.902*
	Large	0.530	0.954*	15.1	0.998*	8.9%	0.863*	0.317	0.982*
Lateral	Medium	0.498	0.967*	9.9	0.967*	5.8%	0.931*	0.240	0.943*
	Small	0.465	0.914*	4.9	0.995*	2.9%	0.869*	0.128	0.715*

ICC: intraclass correlation coefficient. * Significant difference existed in the intraclass correlation coefficient test ($p<0.05$).

Under the small perturbations, all participants were observed to be able to keep their feet in place. Under the medium perturbations, the stepping of the dominant leg occurred once following the posterior pulls (1 out of totally 36 pulls; 1/36), the elevation of the dominant leg occurred following the medial pulls (1/36), and the elevation of nondominant leg occurred following the lateral pulls (1/36). Under the large perturbations, the stepping or elevation of the dominant leg occurred in one participant following the anterior pulls (3/36), in two participants following the posterior pulls (2/36), in seven participants following the medial pulls (15/36), and in two participants following lateral pulls (3/36); the stepping or elevation of nondominant leg occurred in one participant following the posterior pulls (1/36), in three participants following the medial pulls (8/36), and in five participants following lateral pulls (10/36).

5.5.1 Whole-body CoM and CoP Displacement

As shown in , the whole-body CoM and CoP displacements mainly moved toward the direction of waist-pull perturbation. As shown in , following the unexpected anterior perturbations, CoP showed significantly shorter onset latency of displacement, shorter time to peak displacement, and larger peak displacement than CoM ($p<0.05$). The larger perturbation magnitudes evoked significantly longer time to peak displacement and larger peak displacements ($p<0.05$).

Following the unexpected posterior perturbations, CoP showed significantly shorter onset latency, shorter time to peak displacement under the medium and the small magnitudes, and larger peak displacement than CoM ($p<0.05$). The larger perturbation magnitudes evoked significantly larger peak displacements ($p<0.05$).

Following the unexpected medial perturbations, CoP showed significantly shorter onset latency of displacement under the medium magnitude, shorter time to peak displacement under the large and the medium magnitudes, and larger peak displacement than CoM ($p<0.05$). The larger perturbation magnitudes evoked significantly shorter onset latency of CoM displacement, longer time to peak displacement, and larger peak displacements ($p<0.05$).

Following the unexpected lateral perturbations, CoP showed significantly shorter onset latency, shorter time to peak displacement, and larger peak displacement than CoM ($p<0.05$). The larger perturbation magnitudes evoked significantly shorter onset latency of displacement, longer time to peak displacement, and larger peak displacements ($p<0.05$).

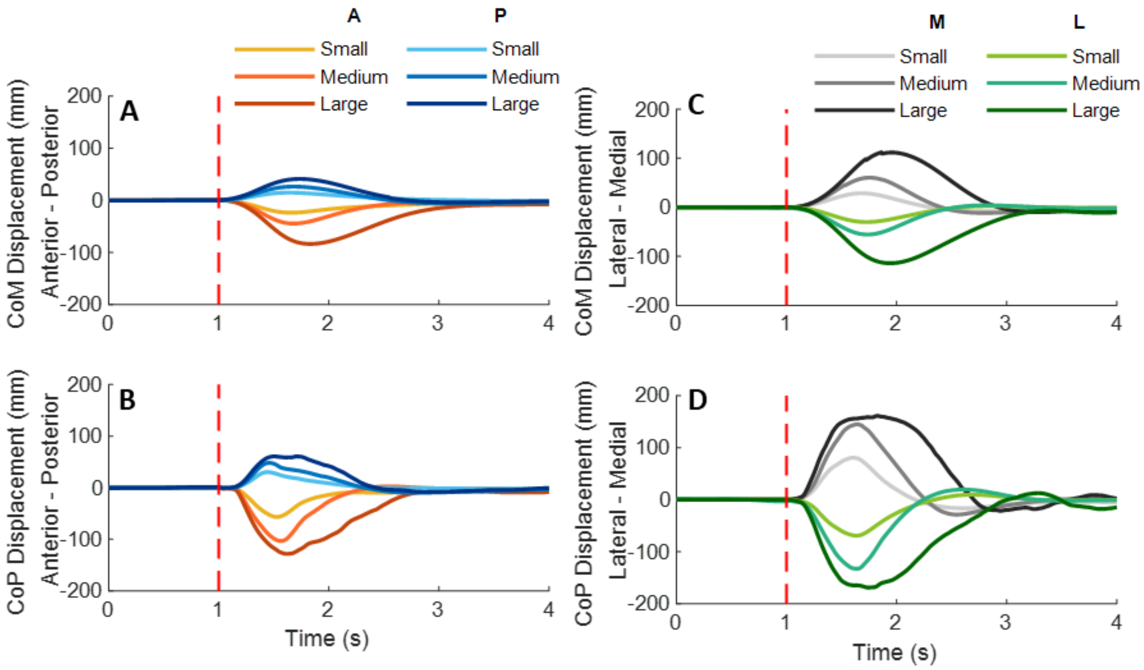


Figure 5-3. The mean whole-body CoM and CoP displacements of twelve participants following the unexpected anterior, posterior, medial, and lateral perturbations with three magnitudes (n=12).

Mean CoM (A) and CoP (B) displacements following anterior and posterior perturbations; Mean CoM (C) and CoP (D) displacements following medial and lateral perturbations. (Note: The red dotted line indicated the start of pulling perturbation. CoM: center of mass; CoP: center of pressure. A: anterior; P: posterior; M: medial; L: lateral.)

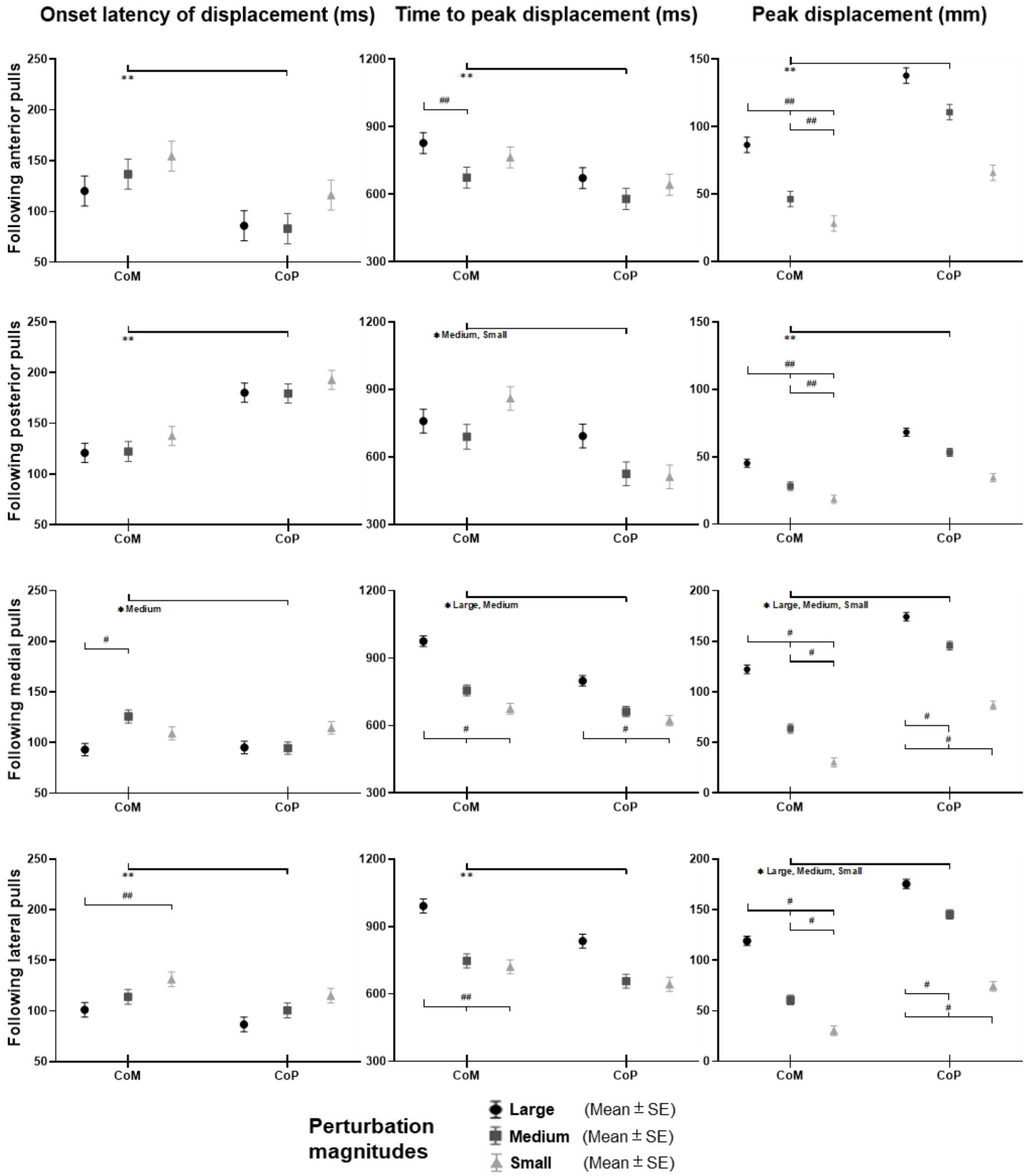


Figure 5-4. The onset latency of displacement, time to peak displacement, and peak displacement of whole-body CoM and CoP following unexpected horizontal perturbations (mean ± SE, n=12).

(Note: CoM: center of mass; CoP: center of pressure; SE: standard error; ⌐ or ⌙: pairwise comparison. Significant differences in post hoc pairwise comparisons (p<0.05) were indicated by the: ** for the main effect of postural sway factor; * for the simple main effect of postural sway factor; ## for the main effect of magnitude factor; # for the simple main effect of magnitude factor.)

5.5.2 Lower-limb Joint Angles and Joint Power

shows the dominant-leg joint angle changes following the unexpected waist-pull perturbations. As shown in , following the unexpected anterior perturbations, the hip extension angle showed significantly shorter onset latency and time to peak angle than the hip flexion angle ($p<0.05$). Peak angles were not significantly different among the eight joint motions. The larger perturbation magnitudes evoked significantly larger peak angles ($p<0.05$).

Following the unexpected posterior perturbations, significant within-joint differences were observed in the angle onset latency (knee flexion<extension; hip adduction<abduction; $p<0.05$) and the time to peak angle (knee flexion<extension; hip flexion<extension; $p<0.05$). The larger perturbation magnitudes evoked significantly larger peak angles in ankle dorsiflexion, knee flexion, and hip flexion ($p<0.05$). Under the large magnitude, peak angles of these three joint motions were significantly larger than the other five joint motions ($p<0.05$).

Following the unexpected medial perturbations, the hip abduction angle showed significantly shorter onset latency than the hip adduction angle irrespective of perturbation magnitudes ($p<0.05$). Under the medium and the small magnitudes, significant within-joint differences were observed in the angle onset latency (hip flexion<extension; $p<0.05$) and the time to peak angle (hip abduction<adduction; hip flexion<extension; $p<0.05$). Under the medium magnitude, the knee flexion angle showed significantly shorter onset latency than the knee extension angle ($p<0.05$). The larger perturbation magnitudes evoked significantly larger peak angles in ankle plantarflexion, knee flexion, hip flexion, and hip abduction ($p<0.05$).

Following the unexpected lateral perturbations, significant within-joint differences were observed in the angle onset latency (hip flexion<extension; hip adduction<abduction; $p<0.05$) and the time to peak angle (hip flexion<extension; hip adduction<abduction; $p<0.05$). Under the large magnitude, peak angles of ankle dorsiflexion, knee flexion, and hip flexion were significantly larger than the other five joint motions ($p<0.05$).

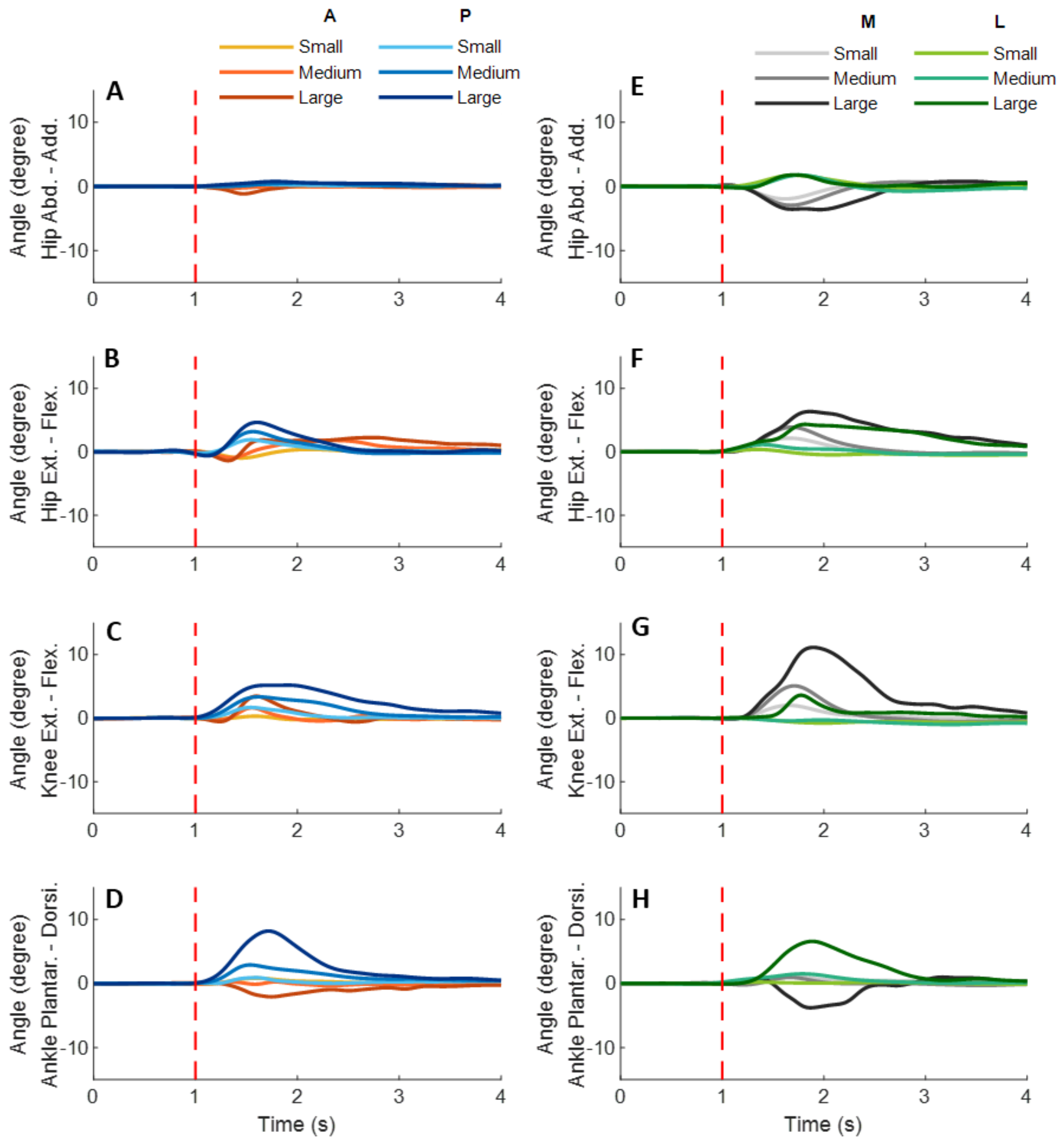


Figure 5-5. The mean dominant-leg joint angle changes of twelve participants following the unexpected anterior, posterior, medial, and lateral perturbations with three magnitudes (n=12).

Mean hip adduction-abduction (A), hip flexion-extension (B), knee flexion-extension (C), and ankle dorsiflexion-plantarflexion (D) angle changes following anterior and posterior perturbations; Mean hip adduction-abduction (E), hip flexion-extension (F), knee flexion-extension (G), and ankle dorsiflexion-plantarflexion (H) angle changes following medial and lateral perturbations. (Note: The red dotted line indicated the start of pulling perturbation. Add.: adduction; Abd.: abduction; Flex.: flexion; Ext.: extension; Dorsi.: dorsiflexion; Plantar.: plantarflexion. A: anterior; P: posterior; M: medial; L: lateral.)

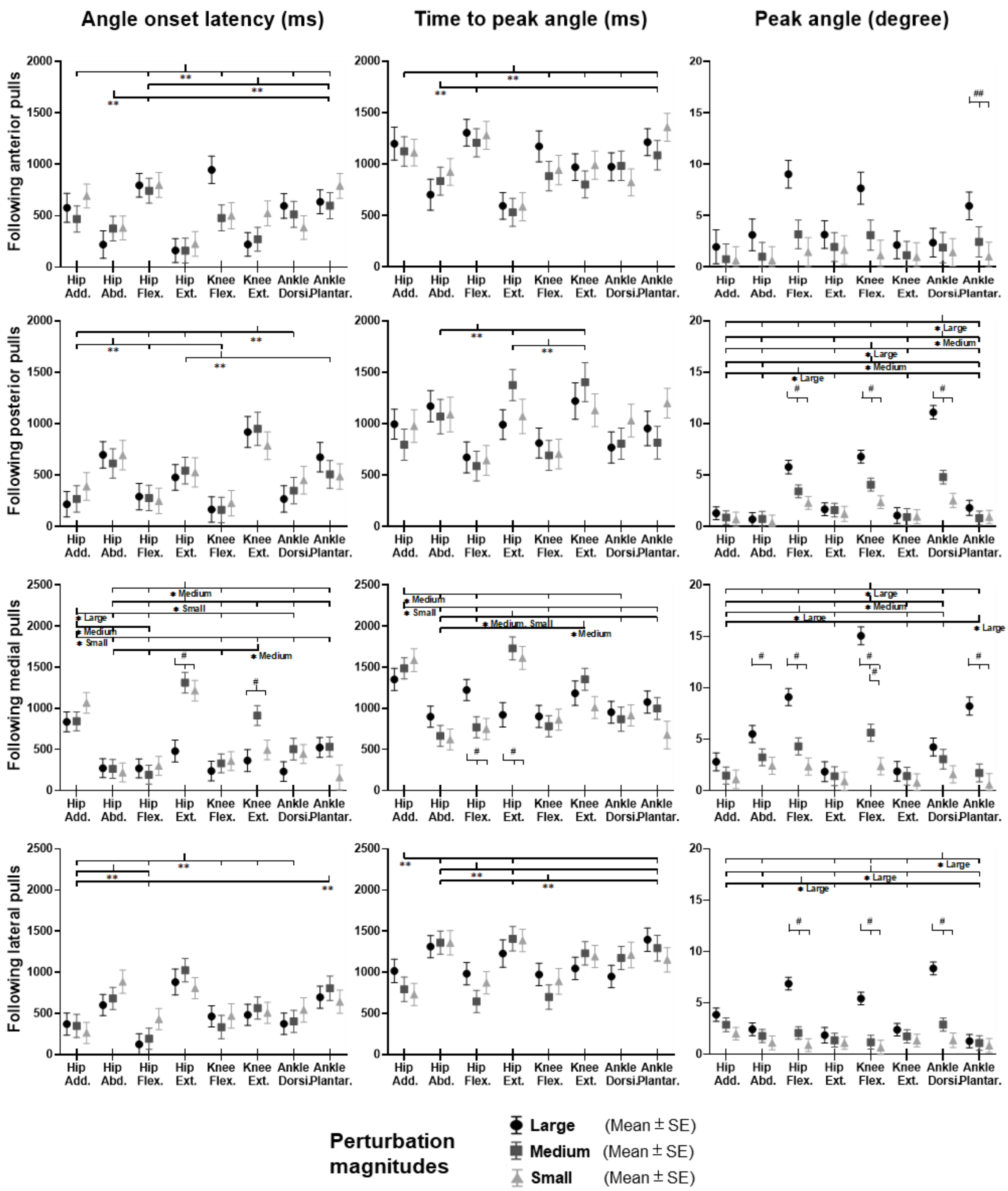


Figure 5-6. The angle onset latencies, time to peak angle, and peak angles of eight lower-limb joint motions following unexpected horizontal perturbations (mean ± SE, n=12).

(Note: SE: standard error; ⌐ or ⌙: pairwise comparison. Significant differences in post hoc pairwise comparisons (p<0.05) were indicated by the: ** for the main effect of joint motion factor; * for the simple main effect of joint motion factor; ## for the main effect of magnitude factor; # for the simple main effect of magnitude factor.)

shows the dominant-leg joint power changes following the unexpected waist-pull perturbations. As shown in , following the unexpected anterior perturbations, significant within-joint differences existed in the power onset latency (hip power absorption<generation; $p<0.05$) and the time to peak power (hip power absorption<generation; knee power generation<absorption; $p<0.05$). Peak power responses in the hip, knee, and ankle joints were not significantly different. The larger perturbation magnitudes evoked significantly larger peak power responses ($p<0.05$).

Following the unexpected posterior perturbations, significant within-joint differences existed in the power onset latency (hip power generation<absorption; knee power absorption<generation; $p<0.05$) and the time to peak power (hip power generation<absorption; knee power absorption<generation; $p<0.05$). Peak power generated in the hip joint was significantly larger than the absorbed ($p<0.05$). The larger perturbation magnitudes evoked significantly larger peak power responses ($p<0.05$).

Following the unexpected medial perturbations, significant within-joint differences were observed in the power onset latency (hip power generation<absorption; knee power absorption<generation; $p<0.05$). Hip power generation showed the shortest time to peak among the six lower-limb joint power responses ($p<0.05$). Peak power responses in the hip, knee, and ankle joints were not significantly different. Generally, the larger perturbation magnitudes evoked significantly shorter power onset latency, longer time to peak power, and larger peak power ($p<0.05$).

Following the unexpected lateral perturbations, significant within-joint differences in power onset latency were observed under the small (hip power absorption<generation; $p<0.05$) and the large magnitudes (ankle power absorption<generation; $p<0.05$). Knee power absorption showed a significantly shorter time to peak than a generation ($p<0.05$). Peak power absorbed in the hip joint was significantly larger than the peak power responses in knee and ankle joints ($p<0.05$). The larger perturbation magnitudes evoked significantly larger peak power responses ($p<0.05$).

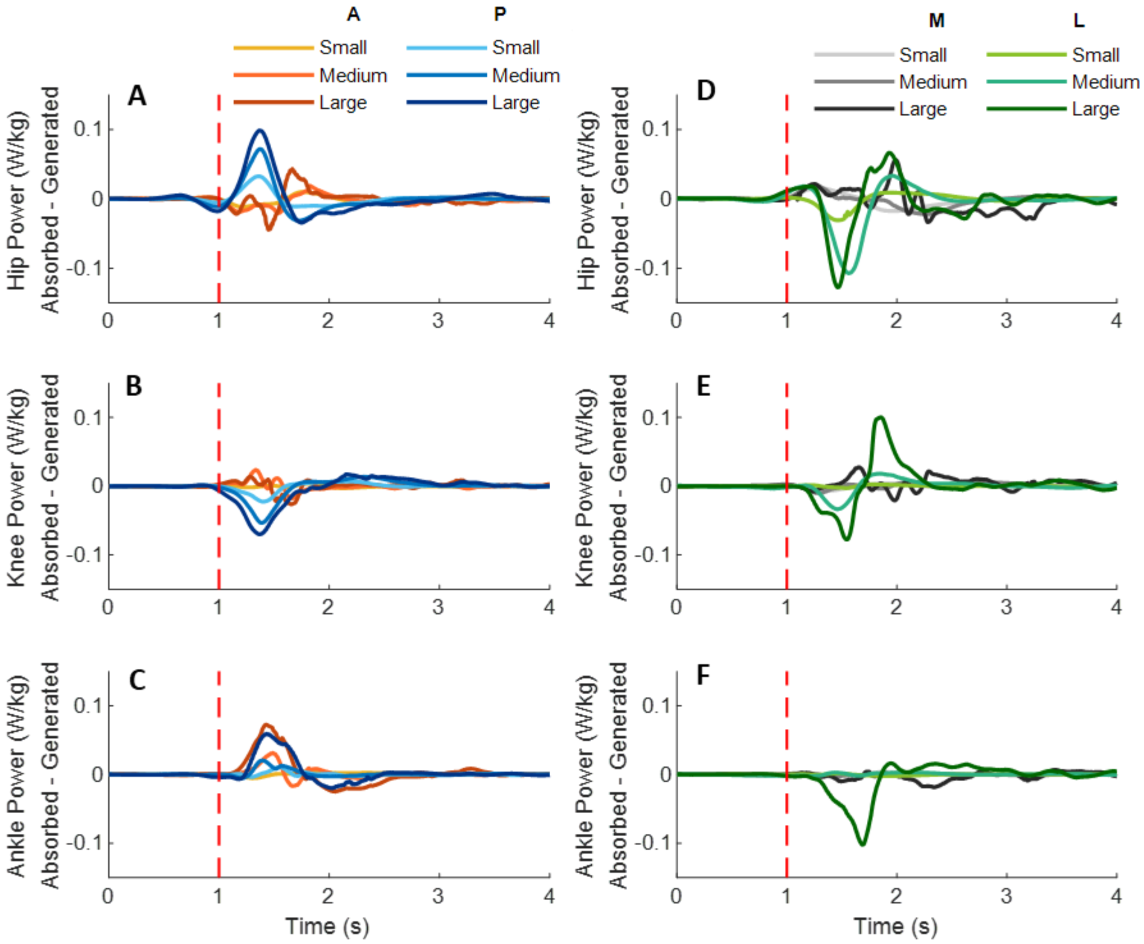


Figure 5-7. The mean dominant-leg joint power changes of twelve participants following the unexpected anterior, posterior, medial, and lateral perturbations with three magnitudes (n=12).

Mean hip (A), knee (B), and ankle (C) power generation and absorption following anterior and posterior perturbations; Mean hip (D), knee (E), and ankle (F) power generation and absorption following medial and lateral perturbations. (Note: The red dotted line indicated the start of pulling perturbation. A: anterior; P: posterior; M: medial; L: lateral.)

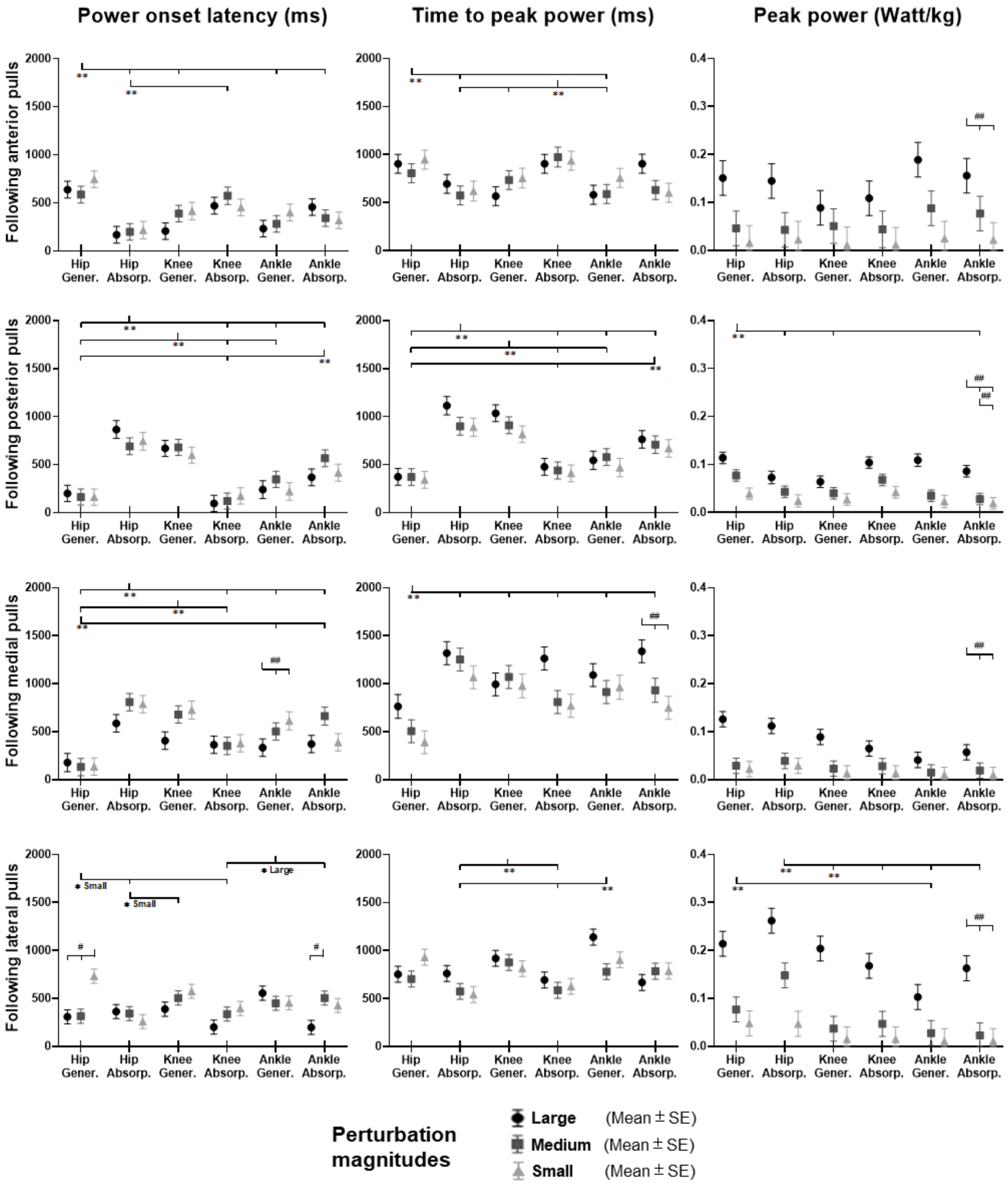


Figure 5-8. The power onset latencies, time to peak power, and peak power responses in lower-limb joints following unexpected horizontal perturbations (mean ± SE, n=12).

(Note: Gener.: power generation; Absorp.: power absorption; SE: standard error; ⌐¬ or └┐: pairwise comparison. Significant differences in post hoc pairwise comparisons (p<0.05) were indicated by the: ** for the main effect of joint motion factor; * for the simple main effect of joint motion factor; ## for the main effect of magnitude factor; # for the simple main effect of magnitude factor.)

Lower-limb Joint Moments

shows the dominant-leg joint moment changes following the unexpected waist-pull perturbations. As shown in , following the unexpected anterior perturbations, significant within-joint differences existed in the moment onset latency (ankle dorsiflexion<plantarflexion; knee extension<flexion; hip flexion<extension; hip adduction<abduction; p<0.05) and the time to peak moment (ankle dorsiflexion<plantarflexion; knee extension<flexion; hip flexion<extension; p<0.05). Peak moments in ankle dorsiflexion, knee extension, and hip flexion were significantly larger than those in the other five joint motions (p<0.05). Particularly, under the medium and large magnitudes, the peak moment of ankle dorsiflexion was the largest among the eight joint motions (p<0.05).

Following the unexpected posterior perturbations, eight joint motions showed no significantly different moment onset latencies. Significant within-joint differences existed in the time to peak moment (ankle plantarflexion<dorsiflexion; knee flexion<extension; hip extension<flexion; hip abduction<adduction; p<0.05). The peak moment of ankle plantarflexion was significantly larger than that of ankle dorsiflexion irrespective of perturbation magnitudes (p<0.05). Knee flexion showed a significantly larger peak moment than knee extension under the medium and the large magnitudes (p<0.05).

Following the unexpected medial perturbations, significant within-joint differences existed in the moment onset latency (hip abduction<adduction; knee extension<flexion; hip flexion<extension; p<0.05). Hip abduction showed the shortest time to peak moment among the eight joint motions (p<0.05). Besides, significant within-joint differences existed in the time to peak moment (hip flexion<extension, knee extension<flexion; ankle plantarflexion<dorsiflexion; p<0.05) and the peak moment (hip flexion>extension; hip adduction>abduction; p<0.05).

Following the unexpected lateral perturbations, significant within-joint differences existed in the moment onset latency (hip adduction<abduction; hip extension<flexion; knee

flexion<extension; ankle dorsiflexion<plantarflexion; p<0.05) and the time to peak moment (hip adduction<abduction; hip extension<flexion; knee flexion<extension; p<0.05). Under the medium and large magnitudes, the peak moment of hip adduction was the largest among the eight joint motions (p<0.05). In the sagittal plane, significant differences of peak moments were observed under the large (hip extension>flexion; knee flexion>extension; ankle dorsiflexion>plantarflexion; p<0.05) and the medium (hip extension>flexion; p<0.05) magnitudes. summarized in what joint motions the more rapid moment response would occur following the four directions of waist-pull perturbations.

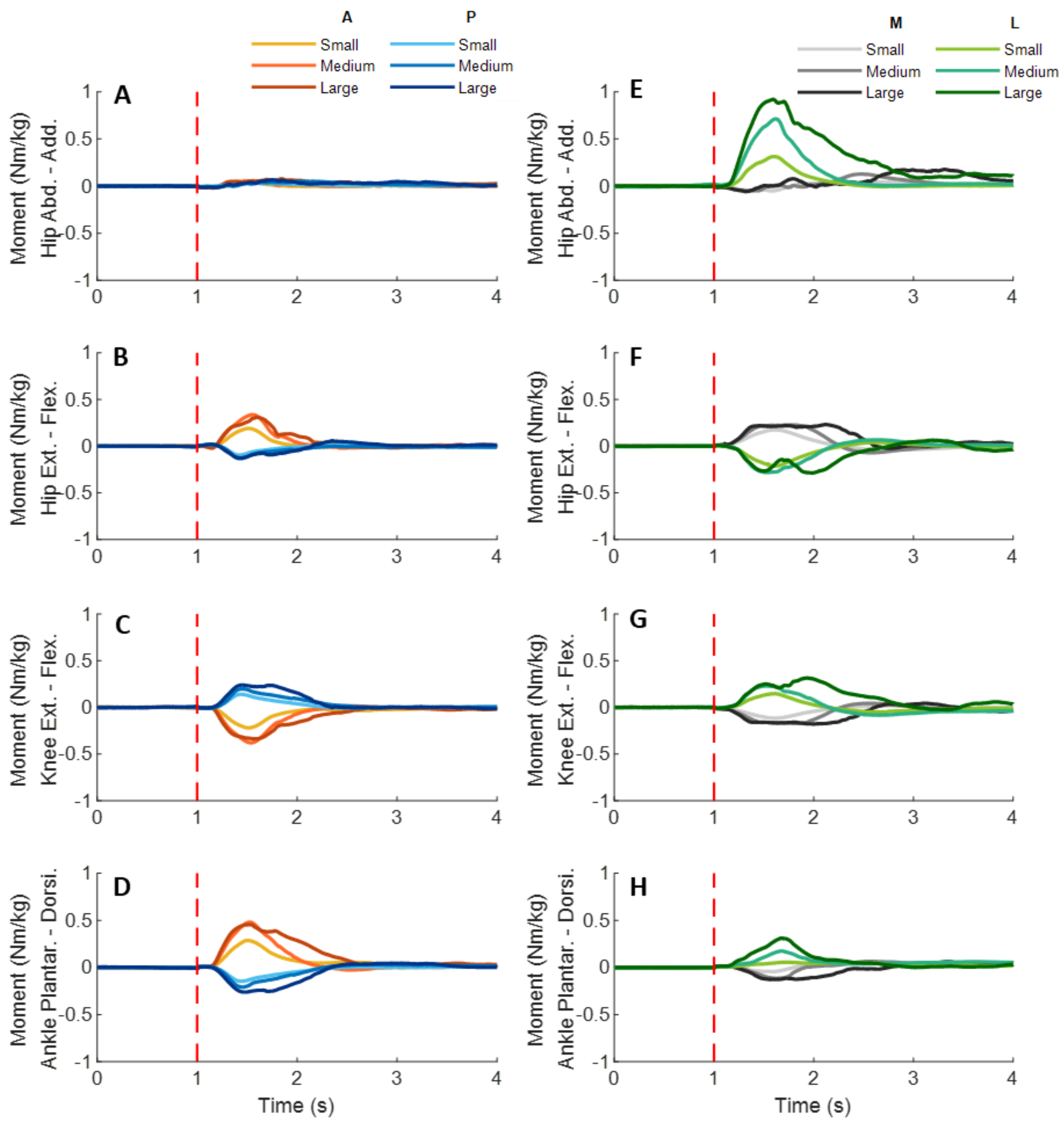


Figure 5-9. The mean dominant-leg joint moment changes of twelve participants following the unexpected anterior, posterior, medial, and lateral perturbations with three magnitudes (n=12).

Mean hip adduction-abduction (A), hip flexion-extension (B), knee flexion-extension (C), and ankle dorsiflexion-plantarflexion (D) moment changes following anterior and posterior perturbations; Mean hip adduction-abduction (E), hip flexion-extension (F), knee flexion-extension (G), and ankle dorsiflexion-plantarflexion (H) moment changes following medial and lateral perturbations. (Note: The red dotted line indicated the start of pulling perturbation. Add.: adduction; Abd.: abduction; Flex.: flexion; Ext.: extension; Dorsi.: dorsiflexion; Plantar.: plantarflexion. A: anterior; P: posterior; M: medial; L: lateral.)

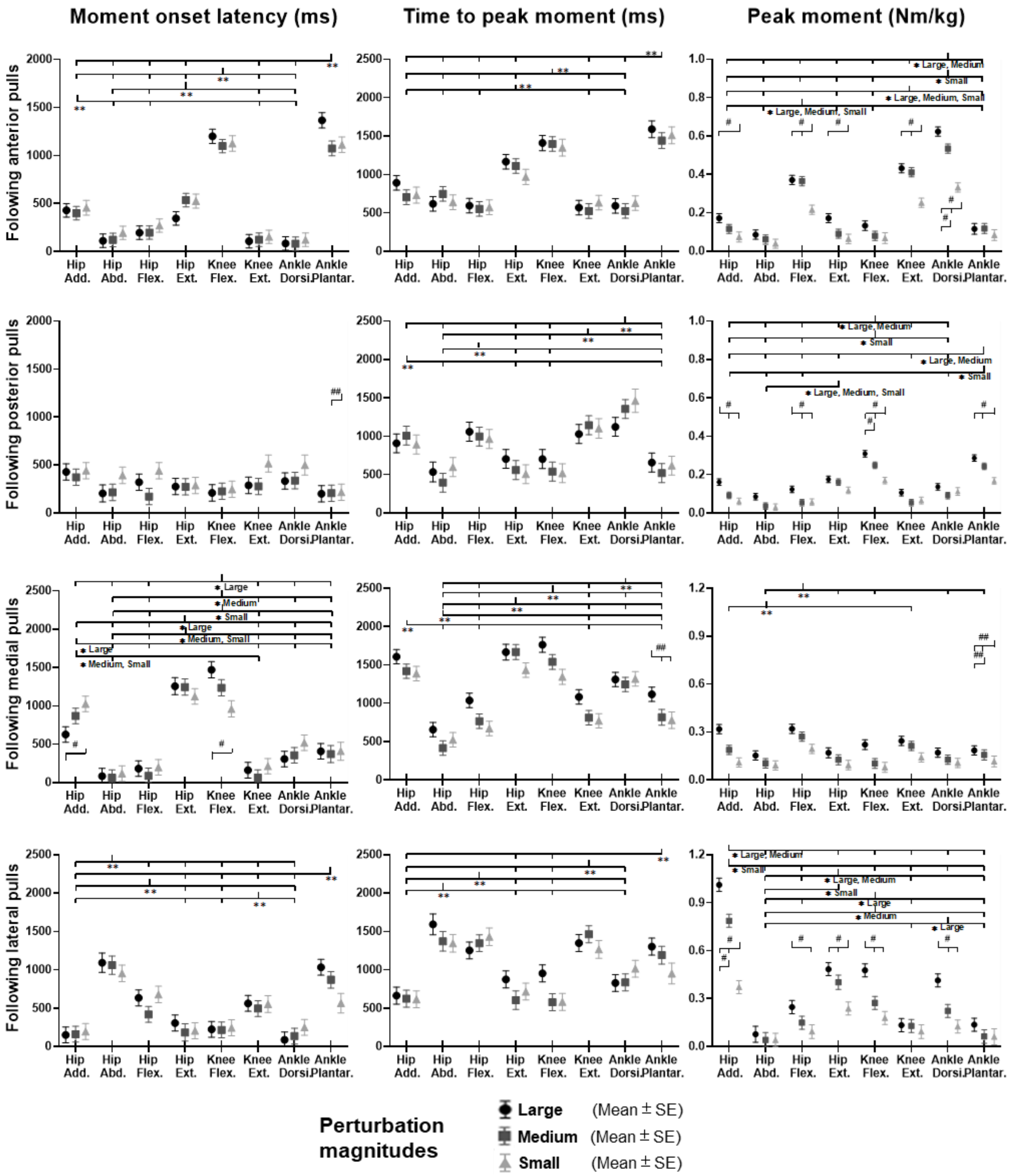


Figure 5-10. The moment onset latencies, time to peak moment, and peak moments of eight lower-limb joint motions following unexpected horizontal perturbations (mean ± SE, n=12).

(Note: SE: standard error; ⌐ or ⌙: pairwise comparison. Significant differences in post hoc pairwise comparisons (p<0.05) were indicated by the: ** for the main effect of joint motion factor; * for the simple main effect of joint motion factor; ## for the main effect of magnitude factor; # for the simple main effect of magnitude factor.)

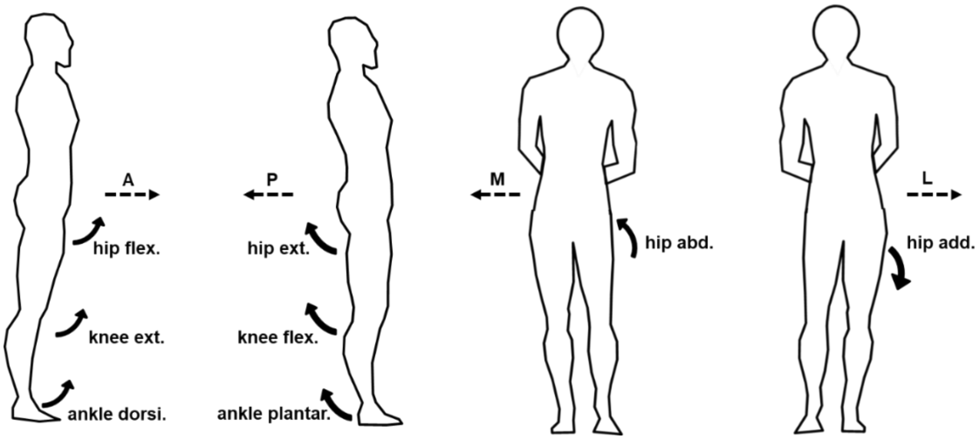


Figure 5-11. Rapid lower-limb joint moment responses evoked by the unexpected waist-pull perturbations.

(Note that the right leg was the dominant leg. A: anterior pulls; P: posterior pulls; M: medial pulls; L: lateral pulls.)

5.5.3 EMG Signals of Eight Lower-limb Muscles

demonstrates the dominant-leg muscles' EMG signal changes following the unexpected waist-pull perturbations. As shown in , following the unexpected anterior perturbations, the ankle plantarflexor, ankle dorsiflexor, and knee flexor were in the queue with short EMG onset latencies, and the ankle plantarflexor showed a significantly shorter EMG onset latency than the other five muscles ($p<0.05$). Significant agonist-antagonist differences existed in the EMG onset latency (knee flexor<extensor; $p<0.05$) and the time to peak EMG amplitude (ankle plantarflexor<dorsiflexor; knee flexor<extensor; $p<0.05$). Ankle plantarflexor showed the largest rate of EMG rise among the eight lower-limb muscles ($p<0.05$). The larger perturbation magnitudes evoked significantly shorter EMG onset latencies and shorter time to peak EMG amplitude ($p<0.05$).

Following the unexpected posterior perturbations, the ankle dorsiflexor, knee extensor, and hip abductor were in the queue with short EMG onset latencies, and the ankle dorsiflexor showed a significantly shorter EMG onset latency than the other five muscles ($p<0.05$). Significant agonist-antagonist differences were observed in the EMG onset latency (ankle dorsiflexor<plantarflexor; knee extensor<flexor; $p<0.05$) and the time to peak EMG amplitude (ankle dorsiflexor<plantarflexor; knee extensor<flexor; $p<0.05$). The ankle dorsiflexor showed the largest rate of EMG rise among the eight lower-limb muscles ($p<0.05$). The larger perturbation magnitudes evoked significantly shorter EMG onset latencies ($p<0.05$).

Following the unexpected medial perturbations ankle dorsiflexor, hip adductor, hip abductor and knee flexor were in the queue with short EMG onset latencies, and the ankle dorsiflexor showed a significantly shorter EMG onset latency than the remaining four muscles ($p<0.05$). Significant agonist-antagonist difference existed in the EMG onset latency (ankle dorsiflexor<plantarflexor; $p<0.05$) and the time to peak EMG amplitude (ankle dorsiflexor<plantarflexor; $p<0.05$). Except for the hip abductor, the ankle dorsiflexor muscle showed a significantly larger rate of EMG rise than the other six muscles ($p<0.05$). The larger perturbation magnitudes evoked significantly shorter EMG onset latencies, longer time to peak EMG amplitude, and a larger rate of EMG rise ($p<0.05$).

Following the unexpected lateral perturbations, significant agonist-antagonist differences existed in the EMG onset latency (hip abductor<hip adductor; knee extensor<knee flexor; $p<0.05$) and the time to peak EMG amplitude (hip abductor<hip adductor; knee extensor<knee flexor; $p<0.05$). Except for the hip abductor, the ankle dorsiflexor showed a significantly larger rate of EMG rise than the other six muscles ($p<0.05$). In the frontal plane, the hip abductor showed a significantly larger rate of EMG rise than the hip adductor ($p<0.05$). The larger perturbation magnitudes evoked significantly shorter EMG onset latencies, longer time to peak EMG amplitude, and a larger rate of EMG rise ($p<0.05$).

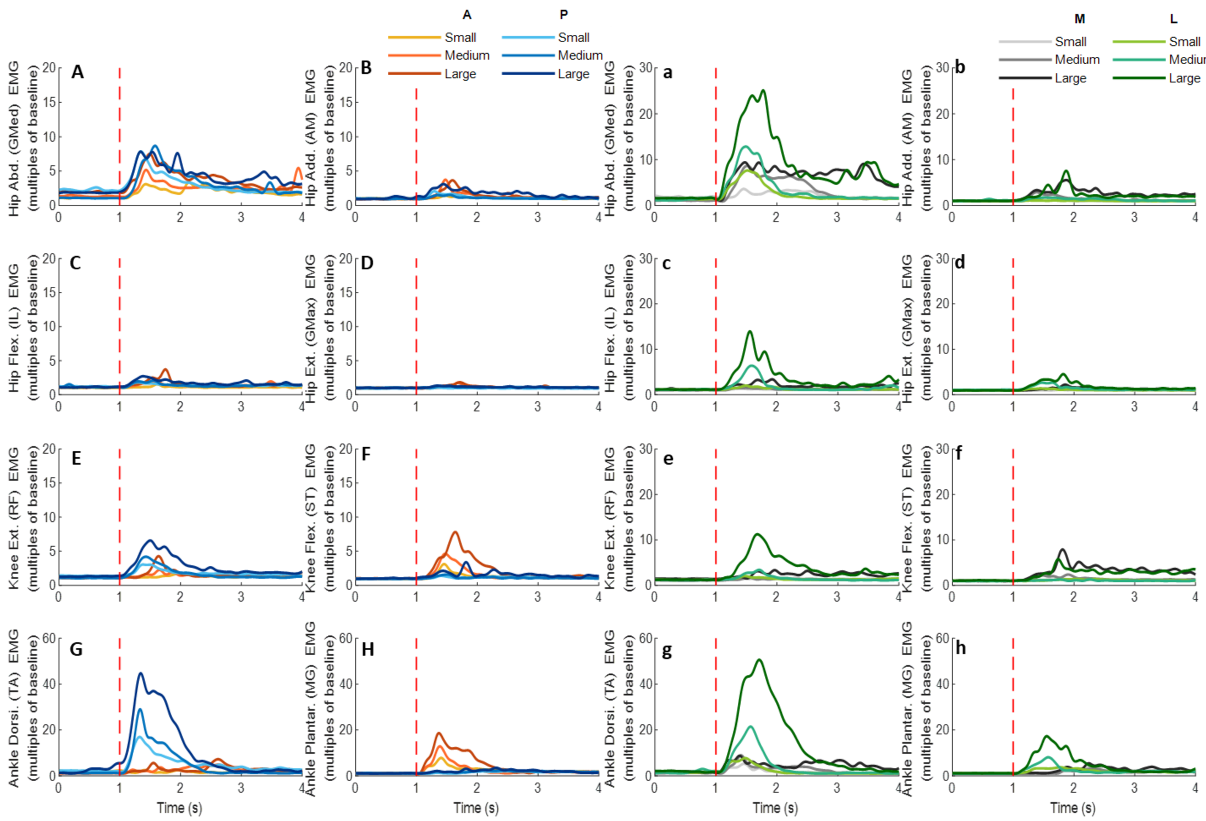


Figure 5-12. The mean EMG signal changes of twelve participants for eight dominant-leg muscles following the unexpected anterior, posterior, medial, and lateral perturbations with three magnitudes (n=12).

Mean EMG signal changes for hip abductor-adductor (A-B), hip flexor-extensor (C-D), knee extensor-flexor (E-F), and ankle dorsiflexor-plantarflexor (G-H) following anterior and posterior perturbations; Mean EMG signal changes of hip abductor-adductor (a-b), hip flexor-extensor (c-d), knee extensor-flexor (e-f), and ankle dorsiflexor-plantarflexor (g-h) following medial and lateral perturbations. (Note: The EMG amplitude values were multiples of the 1000-ms baseline mean value before a pulling perturbation. The red dotted line indicated the start of pulling perturbation. EMG: electromyography. GMed: gluteus medius; AM: adductor magus; IL: iliopsoas; GMax: gluteus maximus; RF: rectus femoris; ST: semitendinosus; TA: tibialis anterior; MG: gastrocnemius medialis; A: anterior; P: posterior; M: medial; L: lateral.)

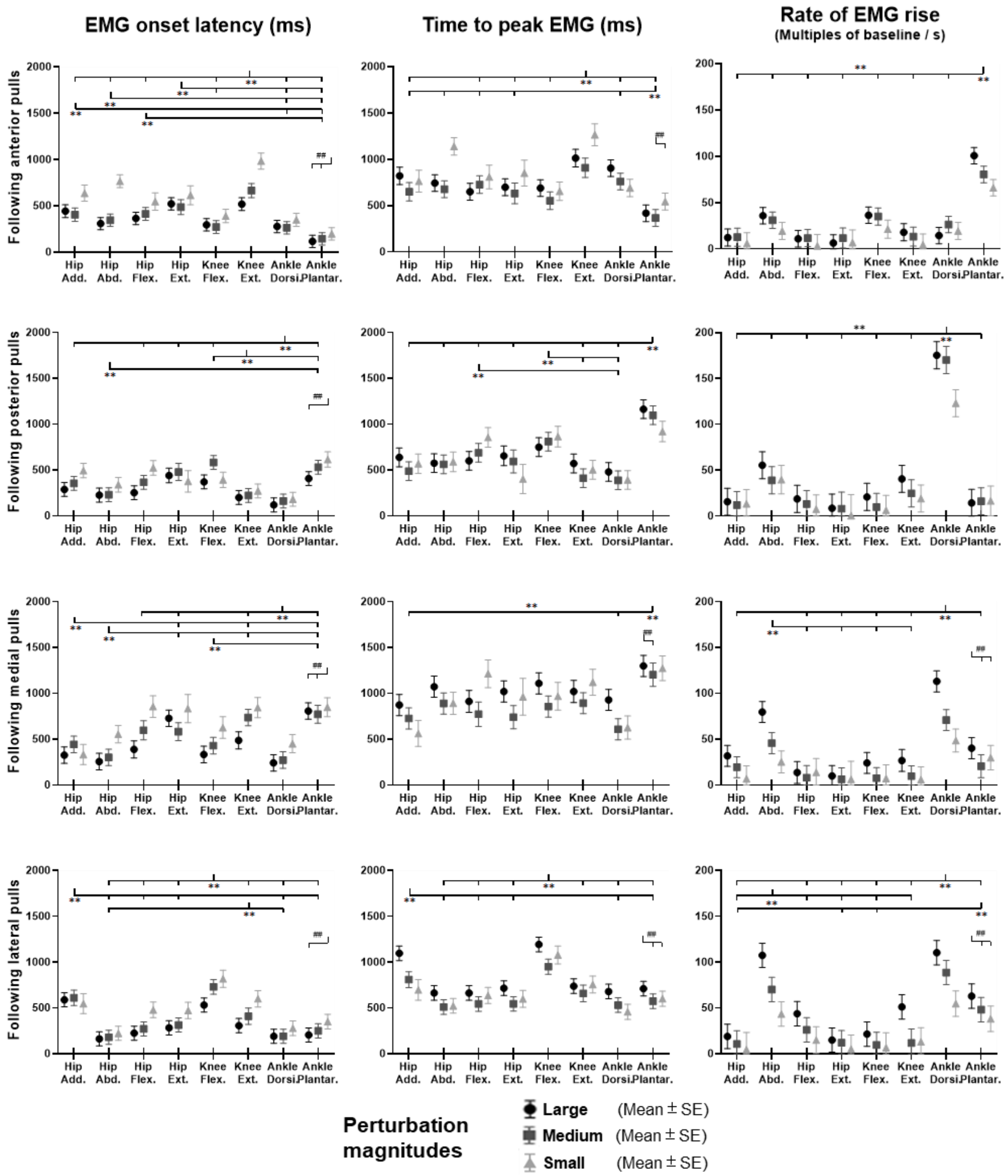


Figure 5-13. The EMG onset latencies, time to peak EMG amplitude, and rate of EMG rise for eight dominant-leg muscles following unexpected horizontal perturbations (mean ± SE, n=12).

(Note: Hip Add.: adductor magus; Hip Abd.: gluteus medius; Hip Flex.: iliopsoas; Hip Ext.: gluteus maximus; Knee Flex.: semitendinosus; Knee Ext.: rectus femoris; Ankle Dorsi.: tibialis anterior; Ankle Plantar.: gastrocnemius medialis. SE: standard error; ⌐¬ or └┐: pairwise comparison. Significant differences in post hoc pairwise comparisons ($p<0.05$) were indicated by the: ** for the main effect of muscle factor; ## for the main effect of magnitude factor.)

5.5.4 MMG Signals of Eight Lower-limb Muscles

demonstrates the eight muscles' MMG signal changes following the unexpected waist-pull perturbations. As shown in , following all the four directions of unexpected perturbations, the hip abductor, hip flexor, and hip extensor were in the queue with short MMG onset latencies. Significant agonist-antagonist differences in MMG onset latencies were observed (hip abductor<adductor; $p<0.05$) following anterior, posterior, and lateral perturbations. The larger perturbation magnitudes evoked significantly shorter MMG onset latencies for all the four pulling directions ($p<0.05$).

Regarding the time to peak MMG amplitude, significant agonist-antagonist differences were observed following anterior (hip abductor<adductor; $p<0.05$), posterior (hip abductor<adductor; $p<0.05$) and lateral (hip abductor<adductor; hip flexor<extensor; $p<0.05$) perturbations. The larger perturbation magnitudes evoked a significantly longer time to peak MMG amplitude following all the four directions of unexpected perturbations ($p<0.05$).

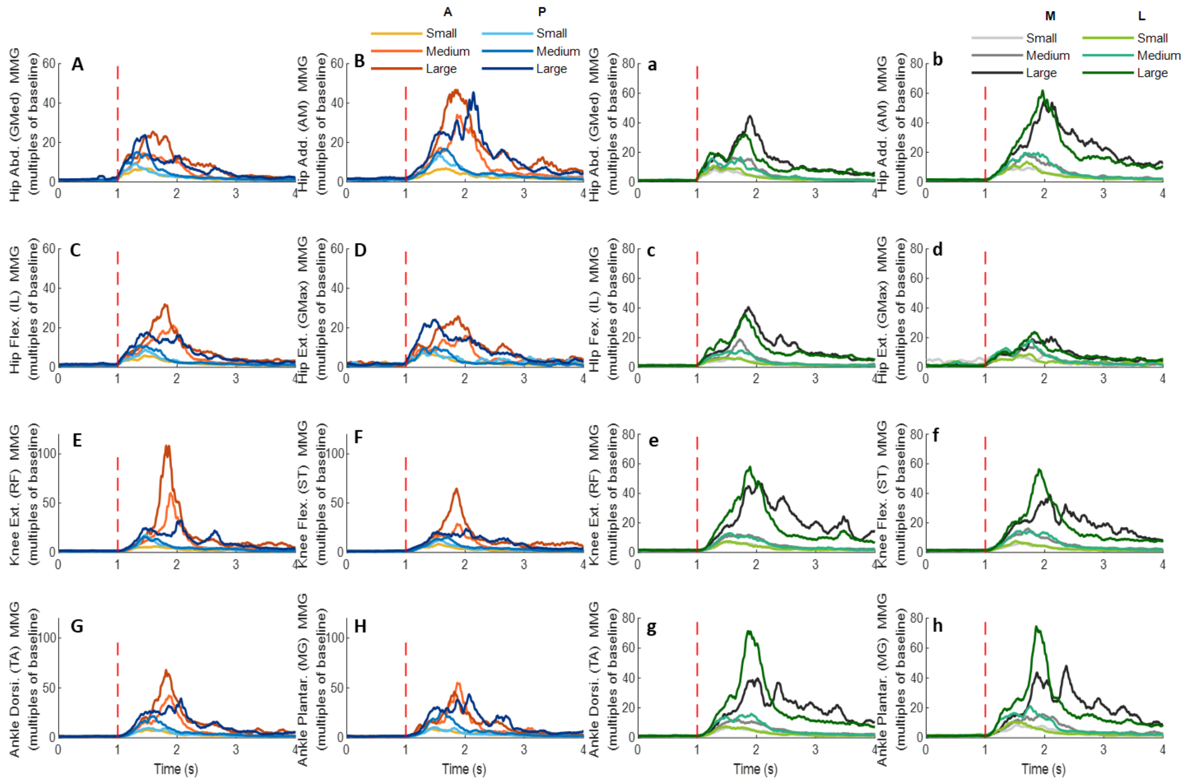


Figure 5-14. The mean MMG signal changes of twelve participants for eight dominant-leg muscles following the unexpected anterior, posterior, medial, and lateral perturbations with three magnitudes (n=12).

Mean MMG signal changes for hip abductor-adductor (A-B), hip flexor-extensor (C-D), knee extensor-flexor (E-F), and ankle dorsiflexor-plantarflexor (G-H) following anterior and posterior perturbations; Mean MMG signal changes of hip abductor-adductor (a-b), hip flexor-extensor (c-d), knee extensor-flexor (e-f), and ankle dorsiflexor-plantarflexor (g-h) following medial and lateral perturbations. (Note: The MMG amplitude values were multiples of the 1000-ms baseline mean value before a pulling perturbation. The red dotted line indicated the start of pulling perturbation. MMG: mechanomyography. GMed: gluteus medius; AM: adductor magus; IL: iliopsoas; GMax: gluteus maximus; RF: rectus femoris; ST: semitendinosus; TA: tibialis anterior; MG: gastrocnemius medialis; A: anterior; P: posterior; M: medial; L: lateral.)

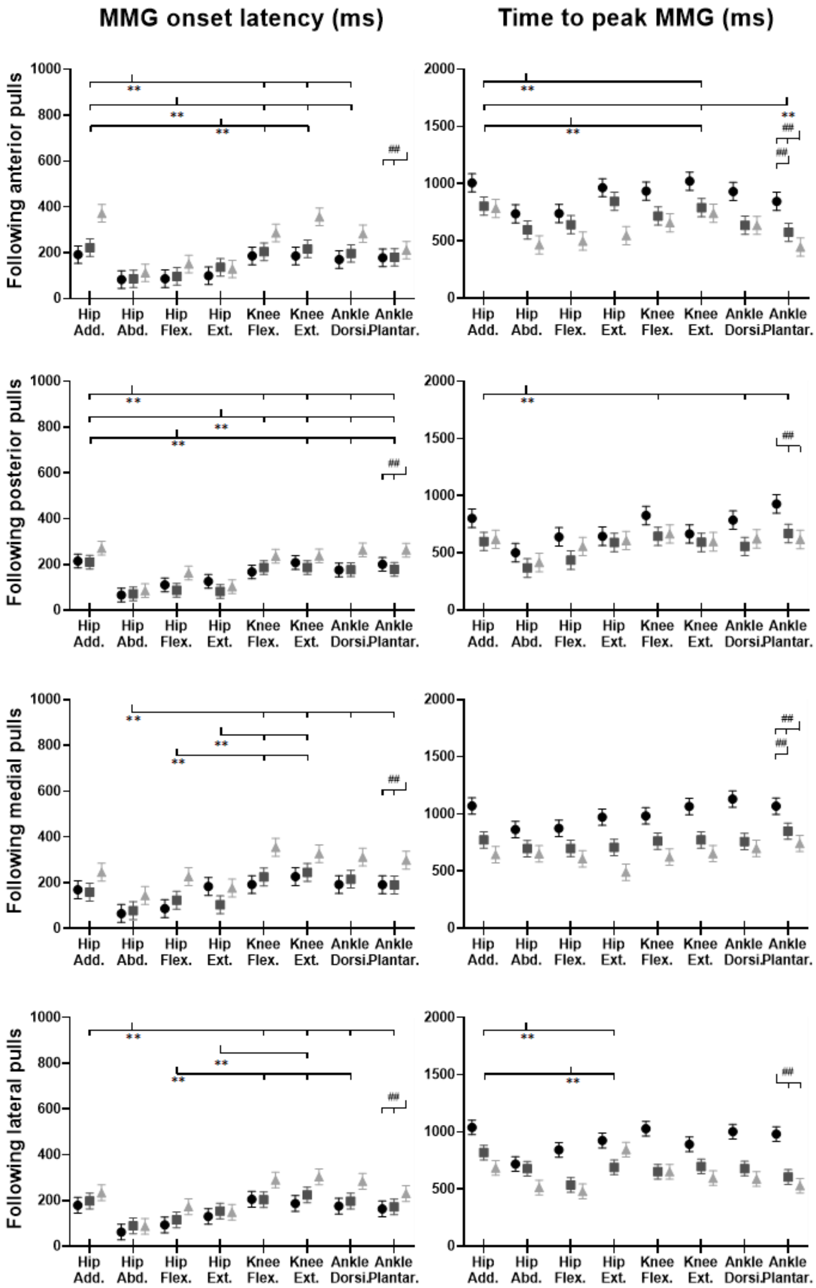


Figure 5-15. The MMG onset latencies and time to peak MMG amplitude for eight dominant-leg muscles following unexpected horizontal perturbations (mean ± SE, n=12).

(Note: Hip Add.: adductor magus; Hip Abd.: gluteus medius; Hip Flex.: iliopsoas; Hip Ext.: gluteus maximus; Knee Flex.: semitendinosus; Knee Ext.: rectus femoris; Ankle Dorsi.: tibialis anterior; Ankle Plantar.: gastrocnemius medialis. MMG: mechanomyography; SE: standard error; ⌐ or ∟: pairwise comparison. Significant differences in post hoc pairwise comparisons ($p<0.05$) were indicated by the:** for the main effect of muscle factor; ## for the main effect of magnitude factor.)

5.6 Discussion

With the innovatively synchronized measurement of postural sway, joint kinetics and kinematics, and muscle EMG and MMG activities, this study comprehensively investigated and uncovered how hip, knee, and ankle muscles and joints reacted to the unexpected perturbations in sagittal and frontal planes. Generally, this study observed that: (1) agonist muscles that resisted the perturbation had more rapid activation than the antagonist muscles; (2) among all agonist muscles resisting the perturbation, ankle muscles had the earliest and largest rate of activation in the sagittal or frontal plane; (3) CoP and lower-joint moments that followed the perturbation had faster increase; and (4) the larger magnitude of perturbations tended to induce faster responses in muscle activities, but not necessarily in joint motions. These findings not only build on our knowledge of how lower-limb muscles and joints respond to balance perturbations, but also facilitate future applied research on developing the targeted balance exercise program and/or the (robotic) assistive technologies/devices to prevent falls of older people and patients. More details can be found below.

5.6.1 Faster Activation Occurred in Muscles Resisting Perturbations, especially for Ankle Muscles

The primary finding of this study is that more rapid activation existed in the agonist muscles that resisted the pulling perturbations, as compared to the antagonist muscles; and ankle muscles appeared to have the earliest and most rapid activation in response to the perturbations in either sagittal (anterior & posterior) or frontal (medial & lateral) plane.

This study observed that for anterior perturbation, muscles moving the body posteriorly (ankle plantarflexor, knee flexor) had early activation and reached the peak neuromuscular activation early. This is consistent with the previous finding that dorsal leg muscles (gastrocnemius, hamstrings) had earlier onset of reflexive activities than ventral muscles, following the unexpected perturbations induced by a backward-moving platform (Runge et al., 1999). The ankle plantarflexor also showed the largest rate of neuromuscular activation among the eight muscles in this study. The rate of EMG rise in the early phase (50 ms after the EMG onset) has been reported as one key determinant of rapid force generation (Folland et al., 2014), and a large rate of dorsal leg muscles' activation was important for preventing tripping (Mirjam Pijnappels et al., 2005). This study further suggested that among the eight lower-limb muscles, the ankle plantarflexor had the most rapid increase of muscle activities to resist the excessive anterior pulling perturbations.

Similarly, this study observed that for posterior perturbation, muscles moving the body anteriorly (ankle dorsiflexor, knee extensor) had earlier activation and reached the peak neuromuscular activation earlier than their antagonist muscles. Such results are consistent with the previous studies that found shorter EMG onset latencies (de Freitas et al., 2010; Hwang et al., 2009; Tsai et al., 2014) and time to peak EMG amplitude (de Freitas et al., 2010) existed in the ventral leg muscles (TA and RF), following the unexpected perturbations induced by a forward-moving platform. Furthermore, this study also observed that the ankle dorsiflexor had a significantly larger rate of neuromuscular activation than the other seven lower-limb muscles. While limited previous studies investigated the rate of neuromuscular activation following balance perturbations, the findings of this study suggested that the ankle dorsiflexor was activated most rapidly in response to the posterior perturbation.

This study also observed that for medial perturbation, the hip adductor and hip abductor had earlier activation; and for lateral perturbation, more lower-limb muscles, including the hip abductor, had earlier activation since more body weight was transferred to the dominant leg. This supported the previous studies' finding that the declined rate of hip abductor/adductor activation correlated with a lower incidence of protective stepping following the unexpected lateral waist-pulls (Inacio et al., 2019). On top of this, this study further found that the ankle dorsiflexor's rapid neuromuscular activation is essential for maintaining the mediolateral standing balance.

5.6.2 Postural sway and Joint Moment Response Followed Perturbations

The secondary finding of this study is that the CoP took less time to reach the peak displacement and had a larger peak displacement than the CoM, and the joint moments that resisted the perturbation had an earlier and faster increase following the perturbation.

This agrees with the inverted pendulum assumption that the distance between CoP and CoM displacements was correlated to the CoM acceleration (Rietdyk et al., 1999; Winter et al., 1998). By moving the CoP quickly in the same direction as the sudden CoM displacement, the change in CoM would be decelerated and kept within the BoS (Rietdyk et al., 1999). Based on the current findings, it is also anticipated that the onset sequence of CoM and CoP may depend on the pulling direction. The anterior, medial, and lateral perturbation induced earlier onset of CoP, and the posterior perturbation induced earlier onset of CoM. This may be because the posterior perturbation is less anticipated and poses a higher risk of uncertainty/falls for participants, as compared to the other three directions. The previous study also reported earlier CoM displacement following the unexpected standing perturbations, and earlier CoP displacement following the anticipated perturbations (Santos et al., 2010). Concerning this, the findings of this study may further suggest that more reaction time is needed for making the compensatory postural adjustment (CPA) following the posterior perturbation or backward loss of balance.

The observation that quicker and larger joint moments occurred to comply with the perturbation direction further supported the above-mentioned postural sway trend (i.e., CoP and CoM displacements). Specifically, for anterior perturbation, the ankle dorsiflexion, knee extension and hip flexion moments showed earlier onset, reached peaks faster, and reached larger peaks. Consequently, these joint moments drove the pelvis, thigh, and shank anteriorly, resulting in anterior CoP displacement. This is contrary to a previous study reporting earlier responses of ankle plantarflexion, knee extension, and hip extension moments following perturbation induced by a backward-moving platform, which caused also the sudden forward CoM displacement with respect to the BoS (Runge et al., 1999). This may be due to the different perturbation methods and magnitudes. The waist-pull perturbation in this study exerted perturbation at the proximal body part (at the pelvis), while that of using a moving platform generated perturbation at the distal body part (at the foot). Further studies are needed to compare the two perturbation methods and verify this.

Similarly, this study observed that posterior perturbation induced a quicker response in ankle plantarflexion, knee flexion, and hip extension moments to move lower limbs posteriorly. This finding is comparable to a previous study reporting earlier responses of ankle plantarflexion, knee extension, and hip extension moments, following posterior perturbation induced by a backward-moving platform (Runge et al., 1999). The different reaction at knee joint may be explained by the strategy in participants, where they may try to further lower the CoM by flexing the knee joints. The findings of this study provide evidence of the joint moment changes, in response to the posterior standing perturbations and sudden backward CoM displacement, which may have been unclear/unavailable previously.

For medial perturbation, this study observed an earlier increase and earlier reaching of the peak for hip abduction moment, and a larger peak moment for hip adduction. This is consistent with a previous study, which observed sinusoidal response of hip adduction/abduction moment following inward pushes of the pelvis (Rietdyk et al., 1999). The firstly appeared increase of hip abduction moment may contribute to the quick medial CoP displacement, while the latter increase of hip adduction moment may be functioned to restore the CoP laterally and back to the dominant leg. The observed earlier/quicker moment increase in hip flexion, knee extension, and ankle plantarflexion of the dominant leg may add more evidence on the joint moment responses of the sagittal plane to the medial perturbations.

For lateral perturbation, this study observed an earlier, quicker, and larger increase of hip adduction moment, leading to lateral CoP displacement. This echoes the previous study that reported increased corrective hip abduction moment after lateral pushes on the pelvis (Rietdyk et al., 1999). Additionally, this study observed the earlier and quicker increase of hip extension, knee flexion, and/or ankle dorsiflexion moments in the sagittal plane. While the faster response of knee flexion moment occurred in both the posterior and the lateral perturbation directions in this study, future studies are needed to verify if the knee flexion moment has functioned to lower the CoM and maintain standing balance by investigating the superior-inferior or vertical movement of CoM following a perturbation. These findings build on our knowledge and understanding regarding the detailed CoP, CoM, and joint moment reactions immediately after the perturbations.

5.6.3 Lower-limb Responses Tended to be Affected by the Varying Perturbation Magnitudes

The tertiary finding of this study is that in general, the rapid responses of lower-limb muscle activities tended to be proportional to the perturbation magnitude levels. More specifically, this study observed that the larger magnitude of perturbations evoked earlier onset of lower-limb muscle EMG and MMG activities, following all four directions of waist-pull perturbations. This was consistent with the previous finding that the increasing magnitude of forward (Hwang et al., 2009; Lin & Woollacott, 2002) and backward (Runge et al., 1999) moving-platform perturbation could result in shorter EMG onset latencies of leg muscles, but was contrary to another study that found no effects of varying perturbation magnitudes on the leg muscles' EMG onset latencies (anterior & posterior) (Szturm & Fallang, 1998). The disparity may be caused by the different range of velocities used for perturbation magnitudes. On top of the previous findings, this study supported that in the frontal plane (medial & lateral), earlier onset of muscle activities may also be evoked by the larger perturbation magnitude. Further, this study observed that the larger magnitude of perturbations evoked a larger rate of EMG activation, and a longer time to peak muscle EMG and MMG activities, following medial and lateral perturbations. To the knowledge of the authors, previous studies reported little on the rate of EMG rise and the time to peak muscle activity in response to the different levels of balance perturbations. These results collectively suggested that for young adults, the lower-limb muscle activities appeared to have the below responses to accommodate a larger magnitude of waist-pull balance perturbation: starting earlier, increasing faster immediately after start, and keeping in activation for a longer time.

Similar to the previous findings (Runge et al., 1999; Szturm & Fallang, 1998; Zemkova et al., 2016), the peak responses of CoM displacement, CoP displacement, lower-limb joint moments, power, and angles were observed to be proportional to the perturbation magnitudes in this study. By contrast, this trend was not observed for the rapid responses of these parameters which appeared to vary for different perturbation directions. The perturbation magnitudes were position- and velocity-controlled in this study, and the pulling durations of “small”, “medium” and “large” magnitudes were set to be the same. This may explain why the onset latencies and time to peak following some directions of pulls were not proportional to the perturbation magnitudes. Nevertheless, following the medial perturbation, lower-limb joint rapid responses were all found to be affected by the different perturbation magnitudes. This may account for why the stepping strategies and the foot elevations were more frequently observed under the large magnitude of the medial perturbation. Future studies can be conducted to verify this.

In addition, this study could be innovative in using the balance perturbations that were tailored to the participant’s stature. Some previous studies have attempted to normalize the force of perturbation to the bodyweight (Vlutters et al., 2018). However, regarding the position-controlled perturbations, very few attempts have been made to minimize the possible confounding effects of body height. The different perturbation magnitudes, i.e., pulling displacements, were divided by the participant’s height in this study, which may make the finding of different perturbation magnitudes’ effects on balance response more reliable and generalizable.

5.6.4 Rapid Power and Angle Responses Were Consistent in Proximal Joints

Regarding the joint angle and power responses, this study observed that the unexpected waist-pull perturbations would evoke the rapid power and angle responses more consistently in hip and/or knee joints, which are proximal lower-limb joints. Joint power was calculated by multiplying the angular velocity with the joint moment. Thus, the power generation would indicate a joint’s accelerating motion, and the power absorption would indicate a joint’s decelerating motion. The onset latency and time to peak results of this study may thus suggest that the anterior, posterior, medial, and lateral perturbations would evoke an earlier hip decelerating extension motion, earlier knee decelerating flexion motion, earlier hip accelerating abduction motion, and earlier abduction and flexion motions, respectively. Previous studies have rarely reported the onset sequence or the sequence of reaching a peak in hip, knee, and ankle joint motions following waist-pull perturbations. One study reported that the suddenly forward-moving platform evoked early joint motions of ankle plantarflexion, knee extension, and hip extension (Hwang et al., 2009), which has been different from the early onset of joint motions following anterior/posterior waist-pulls in this study. Such differences may be caused by the different perturbation locations. Consistent rapid response of joint angle and power at the proximal lower-limb joints maybe because the pulling perturbations were exerted on the pelvis. The proprioceptive receptors in hip and/or knee joints may detect the perturbation signal earlier, leading to more consistent compensatory responses than the ankle joint. Further studies are needed to verify this.

5.6.5 Rapid Response of MMG Signals Occurred in Hip Muscles

This study applied the MMG technology, in an attempt to preliminarily investigate the muscle mechanical activities in response to the sudden perturbations. The detected MMG onset latencies were earlier than those of EMG signals, which did not adhere to the temporal sequence that the onset of electrical activity measured by EMG should precede the onset of muscle vibration measured by MMG (E. Cè et al., 2020; Ling et al., 2020). This indicates that the detected rapid response of MMG signals in this study may not be generated by the active and voluntary muscle contraction, but by the passive and involuntary muscle movement following the waist-pull perturbation instead. This is further supported by the observed earliest MMG onset latencies at hip muscles, which have been the closest to the perturbation location in this study. While previous studies have reported the reliable use of MMG to reflect the onset of muscle's voluntary isometric or concentric contractions in sitting and static positions (Ling et al., 2020; Smith et al., 2017), this study preliminarily applied it in standing and dynamic situations. However, it should be noted that the current processing method of MMG signals was not able to exclude the noise of passive body-segment movements caused by waist-pull perturbations, and the presented results were not generated by the active and voluntary muscle contraction in response to the sudden waist-pull perturbation. Previous studies have also reported that the location of the sensor influenced the captured MMG signals (Cescon et al., 2004). Further optimization of the algorithm and experimental set-up is needed to identify an optimal sensor location and achieve the accurate estimation of lower-limb muscles' active and voluntary rapid contractile responses during dynamic standing situations in the future. The findings of this study on MMG data may serve as a steppingstone and inspire future studies. It may also help to apply some ultrafast imaging technologies to visualize the muscle activity from outside to inside of the human body (Ling et al., 2020; Lyu et al., 2022; Ma, Ling, et al., 2019).

5.6.6 Limitations

There are several limitations of this study. Firstly, this study normalized the EMG or MMG signals with reference to the baseline value during unperturbed standing. After carefully reviewing the Consensus for Experimental Design in Electromyography (CEDE) recommendations (Besomi et al., 2020) and the current study's protocol, the current practice of amplitude normalization may be acceptable. However, considering the leg muscles' rapid activation, e.g., rate of EMG rise, would be affected by the normalization method, future efforts should be made to identify an optimal normalization procedure of the EMG/MMG signals in balance-perturbation-related studies.

Secondly, the EMG sensor placement in this study was based on clinical practice and somewhat obsolete. Future studies shall optimize the EMG electrode locations based on the innervation zone of each muscle (Barbero et al., 2012). It is also possible that the crosstalk between the EMG of the investigated muscles may exist in this study, although such crosstalk shall be minimal, since the anatomical positions of investigated muscles, the locations of EMG sensor placement, and the design of EMG sensors have been carefully reviewed and determined based on the available guidelines in this study.

Thirdly, it appeared that the processed onset latencies of MMG signals in this study were due to the inertia and involuntary muscle movement following the sudden waist-pulling passively, rather than the active and voluntary muscle contraction in response to the perturbation. More efforts are needed to look into how to distinguish and extract the MMG signals generated from the active and voluntary muscle contraction from those generated from the passive and involuntary muscle movement in the future.

Another limitation is that a small number of healthy young participants were recruited in this pilot study. It should be noted that the range of 12 participants' lower-limb responses was generally large for the captured signals, except for the postural sway signals. The large range could partly be caused by the sampling error of small sample size. A larger sample size will be needed to reduce the effects of between-individual difference on the outcomes. In addition, the specific sudden pulling direction and magnitude was randomized and blinded to each participant during the experiment, and the mean value of the three repeated perturbation trials was used for statistical analysis in this study. It is so far unclear how the first trial reaction may influence the results and may be investigated in the future.

5.7 Conclusions

This study observed that the agonist muscles resisting perturbation had more rapid activation than the antagonist muscles; among all agonist muscles resisting the perturbation, the ankle muscles had the earliest and largest rate of activation in the sagittal or frontal plane; the postural sway and joint moments that followed the perturbation had earlier and faster increase; and larger magnitude of perturbations tend to induce earlier responses in muscle activities, but not necessarily in joint motions in healthy young adults. These findings enriched our knowledge of how multiple lower-limb muscles and joints coordinated to quickly make compensatory postural adjustments (CPA) and highlighted the important role of ankle muscles' rapid response in maintaining reactive standing balance.

Chapter 6 Exploring Reactive Balance Control Induced by Waist-pull Perturbations in Older Adults (Study 4)

6.1 Chapter Summary

This chapter includes the contents of study 4 in this PhD project. In the previous studies of this PhD project, the customized waist-pull system has been validated in young adults (study 3), and the neuromuscular/kinematic responses following sudden balance loss induced by the moving-platform system have been preliminarily investigated in a relatively small size of older fallers and older non-fallers (study 2). On top of them, the study 4 further investigated the neuromuscular/kinematic/kinetic responses in reactive balance control induced by the unexpected waist-pull perturbations in a justified sample size of older fallers and older non-fallers, and examined what responses could predict the older adults' prospective falls.

6.2 Abstract

The causal relationships between the neuromuscular/kinetic/kinematic responses following sudden balance loss and the falls remain unknown. This study retrospectively assessed the reactive balance in 72 community-dwelling older adults (i.e., 36 fallers vs 36 non-fallers), and prospectively tracked their fall incidence over 1 year.

In cross-sectional analyses, the older fallers have utilized increased ankle muscles' activation to compensate for the insufficient activation of hip abductor and hip extensor following sudden balance loss; however, this seemed not to be an effective strategy as they required enlarged lower-limb joint moments/powers/motions and postural sways during balance recovery in contrast to the older non-fallers.

In prospective cohort analyses, older adults' high fall risks have been predicted by the hip abductor's insufficient activation, especially the reduced rate of EMG rise, after sudden balance losses in both sagittal and frontal planes. The more absorbed hip/knee joint powers and insufficient hip/knee joint motions in response to sudden balance loss, especially mediolateral balance loss, have also indicated the older adults' fall risks.

Together, these findings supported the measurement of hip abductor's activity during reactive balance control to enhance a more sensitive fall-risk assessment among the older people. Findings of this study also implied that exercises or rehabilitation training could be targeted on the proximal leg joints/muscles, especially the hip abductor, to enhance the fall-prevention effects

in older adults. Moreover, the identified neuromuscular/kinematic/kinetic parameters in this study could serve as the multimodal dataset for the design of robotic assistive devices (e.g., the powered lower-limb exoskeleton) to provide support for the fall-prone older population to have the effective reactive balance control strategies.

6.3 Introduction

Falls and the resulting injuries/deaths have negative impacts on older individuals around the world (World Health Organization, 2021). Each year, about one in three older people experiences a fall; in addition, it was estimated that 37.3 million severe falls required medical attention worldwide (Moreland et al., 2020; World Health Organization, 2021). Despite the implementation of multi-dimensional fall-prevention strategies, the effectiveness of reducing falls, particularly in older people who had fall histories (i.e., fallers), has been limited (de Vries et al., 2010). Fallers also have higher odds of future falls than non-fallers (Deandrea et al., 2010). Considering that balance and gait disorders are the primary major contributors to falls in older people (World Health Organization, 2021), a sensitive and accurate assessment of balance control is essential to identify the older adults that are prone to falls early. However, most clinical tests are assessing the volitional balance control. Only a limited number of clinical tests are available to assess reactive balance control, and their evaluations are based on the clinician's observation/scoring of how many steps the client makes after being suddenly released/pulled. Some more intrinsic but modifiable factors influencing reactive balance control in older fallers, or those predictive of future falls, may remain undiscovered. Targeting them may enhance a more effective fall-prevention management.

Through biomechanical and neuromuscular analyses in humans, various strategies for reactive balance control involving the lower limbs have been identified to avoid a real fall (Kasahara & Saito, 2021; Tong et al., 2023; Zhu et al., 2024; Zhu et al., 2022). In contrast to young adults, older adults could demonstrate a greater reliance on proximal leg muscles and joints over distal ones (Hall & Jensen, 2002; Osoba et al., 2019), and could have more stepping responses for reactive balance control following unexpected perturbations (Mille et al., 2013; Mille et al., 2005). Nevertheless, age-related changes following sudden balance loss do not necessarily equate to fall-risk factors due to the potential confounding influence of age. It is therefore essential to conduct specific comparisons between the age-matched older adults who had fall histories and those who did not have fall histories (i.e., fallers vs. non-fallers) or conduct direct investigations into balance control alterations that could predict future falls with age controlled.

For the previous studies investigating how the older adults' fall histories or fall risks were related to the reactive balance control performance, they mostly analyzed stepping responses and whole-body postural sways (Bair et al., 2016; Batcir et al., 2020; Gerards et al., 2021; Maki et al., 1994; Mille et al., 2013; Sturnieks et al., 2013; Tantisuwat et al., 2011), while the specific joint motions (Zhu et al., 2024), joint moments, joint powers, or muscle activities (Claudino et al., 2017; Ochi et al., 2014; Studenski & Chandler, 1991; Thompson et al., 2018; Zhu et al., 2024) were less focused. Regarding the older adults' prospective fall risks, they could be predicted by the slower center-of-pressure (CoP) displacement for maintaining mediolateral feet-in-place balance (Maki et al., 1994) and the delayed initiation for a backward step (Sturnieks et al., 2013). Regarding the older adults' fall histories, fallers could have more unloaded-leg stepping responses (Bair et al., 2016) and slower stepping responses (Batcir et al., 2020) following mediolateral perturbations but could have faster stepping responses following anteroposterior perturbations (Tantisuwat et al., 2011). However, the more rooted biomechanical and neuromuscular fall-related factors have been less investigated. Firstly, previous investigations comparing fallers and non-fallers have analyzed lower-limb EMG signals and joint motions following sudden balance loss, but these studies have been insufficient. Many of these studies focused on a limited set of muscles [i.e., ankle dorsiflexor/plantarflexor (Claudino et al., 2017; Ochi et al., 2014; Studenski & Chandler, 1991; Thompson et al., 2018), knee flexor/extensor (Claudino et al., 2017; Ochi et al., 2014; Thompson et al., 2018), and/or hip abductor (Claudino et al., 2017)], with the only one (Claudino et al., 2017; Studenski & Chandler, 1991; Thompson et al., 2018) or a few (Ochi et al., 2014) timing and amplitude parameters of electromyographic (EMG) signals analyzed in each study. A full picture on fallers and non-fallers' activation rate, peak activation, and agonist-antagonist co-contraction of hip, knee, and ankle muscles during reactive balance control was still unclear. Further, our previous study with a small sample size of fallers and non-fallers preliminarily investigated the lower-limb joint motions and the responses of more lower-limb muscles following unexpected moving-platform perturbations, by involving a more detailed analysis of the timing and magnitude characteristics (Zhu et al., 2024). Fallers were observed with a series of neuromuscular and kinematic alterations, indicating a preference for utilizing the suspensory strategy to compensate for inadequate initiations of ankle strategy and hip strategy compared with non-fallers (Zhu et al., 2024). Such investigation in a larger sample size of fallers and non-fallers is merited to provide more convincing evidence. Secondly, to the best of authors' knowledge, the alterations of specific lower-limb joint powers and moments during reactive balance control that were related to fall histories have been scarcely reported. More importantly, while the previous prospective cohort studies reported the whole-body postural sways and stepping characteristics that could predict fall risks (Maki et al., 1994; Sturnieks et al., 2013), the causal relationships between older adults' responses of lower-limb muscles/joints following sudden balance losses and their prospective falling have remained unknown, let alone the feasibility of using the identified neuromuscular/biomechanical factors for classifying fall risks in clinical practices. Comprehensive investigations are warranted to delineate these relationships and enable a more sensitive fall-risk assessment among the older adults.

This study therefore aims to probe the in-depth neuromuscular and biomechanical mechanisms underlying reactive balance control strategies that could indicate fall risks or fall histories in older adults. Specifically, in the cross-sectional analyses for examining impact of fall histories, this study attempted to delve into fallers' alterations in the EMG signals of eight lower-limb muscles, lower-limb joint powers/moments/angles, and whole-body CoM displacements after sudden waist-pull perturbations by comparing with non-fallers. Timing and amplitude parameters were thoroughly examined for these signals. In the prospective cohort analyses, this study aimed to identify what the above-mentioned parameters of signals could predict the older adults' falls in prospective 1 year and examine the abilities of these parameters in classifying different prospective fall status. Drawing from existing findings related to aging (Allum et al., 2002; de Freitas et al., 2010; Kasahara & Saito, 2021; Lin & Woollacott, 2002; Osoba et al., 2019; Thompson et al., 2018) and fall histories (Claudino et al., 2017; Studenski & Chandler, 1991; Thompson et al., 2018), we hypothesized that fallers would have varied but generally delayed and reduced responses in lower-limb muscles and joints after sudden loss of standing balance comparing with non-fallers, and these changes would be the fall-risk factors.

6.4 Methods

6.4.1 Subjects

This study was an observational study with both the cross-sectional and prospective cohort analysis (registration number: ChiCTR2100047113). A total of 36 fallers (≥1 fall in previous one year) and 36 non-fallers (no fall in previous one year) with matched age and sex were recruited through convenience sampling. Based on the data of first 12 older fallers and first 12 older non-fallers recruited in this study, the difference in CoM displacement between faller group and non-faller group showed the effect size (Cohen's d) of 0.67, which was the mean value of three magnitudes and four directions. The estimated sample size for a statistical power of 0.80 to run the two-tailed unpaired t-test at the 0.05 significance level was 36 subjects for each group (G∗Power Version 3.1.9.4).

Inclusion criteria were individuals: **1)** ≥65 years old, and **2)** who lived independently in the community and can walk for 400 m without assistive tools. Exclusion criteria were individuals: **1)** who had been resided in nursing homes for over 6 months in the previous one year; **2)** who had fall(s) caused by work-related or traffic accidents; **3)** who were diagnosed with a neurological or vestibular disease, diabetes, or cognitive impairment; **4)** who were diagnosed with a severe orthopedic, visual, or cardiopulmonary disease that impacts daily standing and walking; and **5)** engaged in structured strengthening exercises in the previous year. Totally, 72 older subjects met the eligibility criteria for this study (Ethical reference number: HSEARS20220409002-02).

Each subject took part in the experiment once, which included subjective assessments and waist-pull perturbation trials. Then each subject's prospective fall status within one year was tracked.

6.4.2 Subjective Assessments including 1-year Fall History

An examiner with a medical education background carried out the below subjective assessments. The process began with the collection of demographic/anthropological data, medical history, and number of falls in the previous 1 year, followed by the assessments using scales or questionnaires. Each subject was presented with the short Falls Efficacy Scale-International (FES-I) to assess the fear of falling, and presented with the Chinese Version of the Physical Activity Scale for the Elderly (PASE-C) to assess the level of physical activity (Ku et al., 2013; Yardley et al., 2005). Subsequently, the Mini-Balance Evaluation System Test (Mini-BESTest) was employed as a clinical test to assess balance performance (King & Horak, 2013). Following this, the examiner determined the subject's dominant leg, which would be used for the EMG sensor placement later (Tong et al., 2023).

6.4.3 Assessment of Reactive Balance Control in Waist-pull Perturbation Trials

6.4.3.1 Experimental Set-Up

A waist-pull perturbation system was utilized to deliver sudden horizontal perturbations (**Figure 6-2**), with detailed technical specifications outlined in a prior study (Tong et al., 2023). In short, a string attached to the belt at the subject's waist level can initiate a random pull, featuring random starting time, directions, and distances/velocities/accelerations (varying in magnitude) to induce an unforeseen balance disturbance during normal standing. Whole-body kinematics were captured using the motion capture system (Nexus 2.11, Vicon Motion Systems Ltd., Yarnton, UK) sampling at 250 Hz. Additionally, an eight-channel Trigno Wireless Biofeedback System (Delsys Inc, Natick, MA, USA) sampling at 2000 Hz was employed to collect neuromuscular activities.

Data acquisition was synchronized across the three systems (Tong et al., 2023).

6.4.3.2 Protocol of Waist-pull Perturbation Trials

Before the perturbation trials, all placements of sensors and markers were performed by another examiner who had the educational background of physiotherapy. Eight wireless surface EMG sensors were firstly positioned on the dominant-leg muscles based on the SENIAM guidelines (**Table 4-1**) (Hermens et al., 2000). Key muscles crucial for hip, knee, and ankle joint movements were selected, including gluteus medius (GMed), adductor magnus (AM), sartorius (SA), gluteus maximus (GMax), rectus femoris (RF), long head of bicep femoris (BF), tibialis anterior (TA), and medial gastrocnemius (MG). Additionally, 39 retroreflective markers were affixed to the subject's body based on the Plug-in-Gait Model (Vicon Motion Systems Limited, 2021).

The whole process of perturbation trials was explained to each subject beforehand. For each perturbation trial, the subject was directed to stand with both feet clad in shoes, positioned shoulder-width apart on two separate force plates. The subject was guided to grasp a light rod near the trunk (to ensure the arms not obstruct the retroreflective markers), adopt a natural stance, and gaze forward initially; make their best effort to maintain balance when experiencing perturbation, and promptly return to the original foot placement marked by dark-colored tapes if they shifted their feet. A harness was worn by the subject to avoid a real fall during perturbation.

The subject then underwent five trials (with 12 perturbations in each trial and a 5-minute break following each trial), which included a sum of 60 unexpected balance perturbations encompassing five repetitions for each of the four directions and three magnitudes. For each perturbation, a pre-tensioned string pulled the subject horizontally in a direction and magnitude predetermined by the examiner, then slackened for 12 seconds before becoming tensioned again. The starting time, magnitude (large, medium, small), or direction (anterior, posterior, medial, lateral) were random. Based on findings from our preliminary study in young adults (Zhu et al., 2022), the largest magnitudes for the anterior, posterior, medial, and lateral directions corresponded to pulling distances of 6%, 4%, 8% and 8% of the subject's body height, respectively, which could challenge the subject's limit of stability.

6.4.3.3 Data Processing

Kinematic data (i.e., CoM, joint motions) and kinetic data (joint moments, joint powers) were initially processed in Vicon Nexus using the Plug-in-Gait full body model. Joint moments and joint powers were further smoothed with a 15 Hz low-pass filter using a zero-phase 4th order Butterworth filter. To normalize the data, the kinematic data and joint moment data subtracted the baseline signal value, i.e., the mean signal value over the 1000-ms period preceding each perturbation. The raw EMG signals first subtracted the mean value of the entire perturbation trial, then underwent full-wave rectification, got smoothed with a zero-phase 4th order Butterworth filter to generate the envelope, and were further normalized by dividing them by the mean signal value over the 1000-ms period preceding the start of the perturbation trial (Tong et al., 2023; Zhu et al., 2022).

Amplitude parameters (i.e., the peak amplitude for each signal, the rate of rise for joint moment and EMG signals, and the agonist-antagonist CCI for EMG signals) and temporal parameters (i.e., onset latency, time to peak, and burst duration for each signal) were analyzed (**Figure 4-2**). Within 2 seconds following perturbation start, the onset was identified as the first point with its signal value exceeding five times the standard deviation (SD) over the mean baseline value (mean + 5 SD), and the peak determined as the point with the highest signal value following the onset (Ling et al., 2020; Tong et al., 2023; Zhu et al., 2022). Within 9 seconds following perturbation start, the offset was recognized as the first point with its signal value going below five times the SD over the mean baseline value (mean + 5 SD) (Hesam-Shariati et al., 2017). The onset or offset detection utilized the 1000-ms period preceding each perturbation as the baseline. Further, the peak amplitude was defined as the value of the identified peak signal value. The rate of rise was calculated as the slope of signal rise over a 50-ms period after the onset of a signal (Tong et al., 2023; Zhu et al., 2022). The CCI for an agonist-antagonist muscle pair was calculated between the later EMG onset point of two muscles and the earlier EMG offset point of the same muscles, according to the formula in **Figure 4-2** (Di Nardo et al., 2022; Falconer, 1985; Thompson et al., 2018). The onset latency represented the lagged time from the perturbation start to the signal onset. The time to peak denoted the lagged time from the perturbation start to the signal peak. The burst duration represented the lagged time from the signal onset to offset. Each parameter's mean value following the five repetitive perturbations that were in same direction and magnitude was utilized for subsequent statistical analyses.

The stepping response was determined by identifying whether the foot liftoff occurred following a perturbation, i.e., the vertical ground reaction force on one leg was reduced to 0. Based on the number of steps, stepping responses were categorized into with multiple steps, with a single step, and with no step. Based on the stepping leg, stepping responses were categorized into with a first dominant-leg step, with a first non-dominant-leg step, and with no step. Noted that firstly making a dominant-leg step following medial perturbations or making a non-dominant-leg step following lateral perturbations also indicated the unloaded-leg stepping strategy, whereas firstly making a non-dominant-leg step following medial perturbations or making a dominant-leg step following lateral perturbations indicated the loaded-leg stepping strategy. The ratio of subjects and the ratio of perturbations with the above different stepping responses were calculated for each fall status group following each magnitude and direction of perturbations.

6.4.4 Tracking of Prospective 1-year Fall Status

After participation of the perturbation experiment, each subject's fall status and fall-related consequences within a 12-month prospective duration were followed up. The subject was instructed to record each day's fall status in a monthly calendar and return a picture of the record to a tester at the end of the month via the social media application or email (Hirase et al., 2020). When a subject's monthly record was not returned within one week, the tester called the subject to obtain fall data. When a fall event was reported, the tester also called the subject to inquire about the circumstance in which the fall occurred and the fall consequence, such as any fall-related injury or following handling (Hirase et al., 2020).

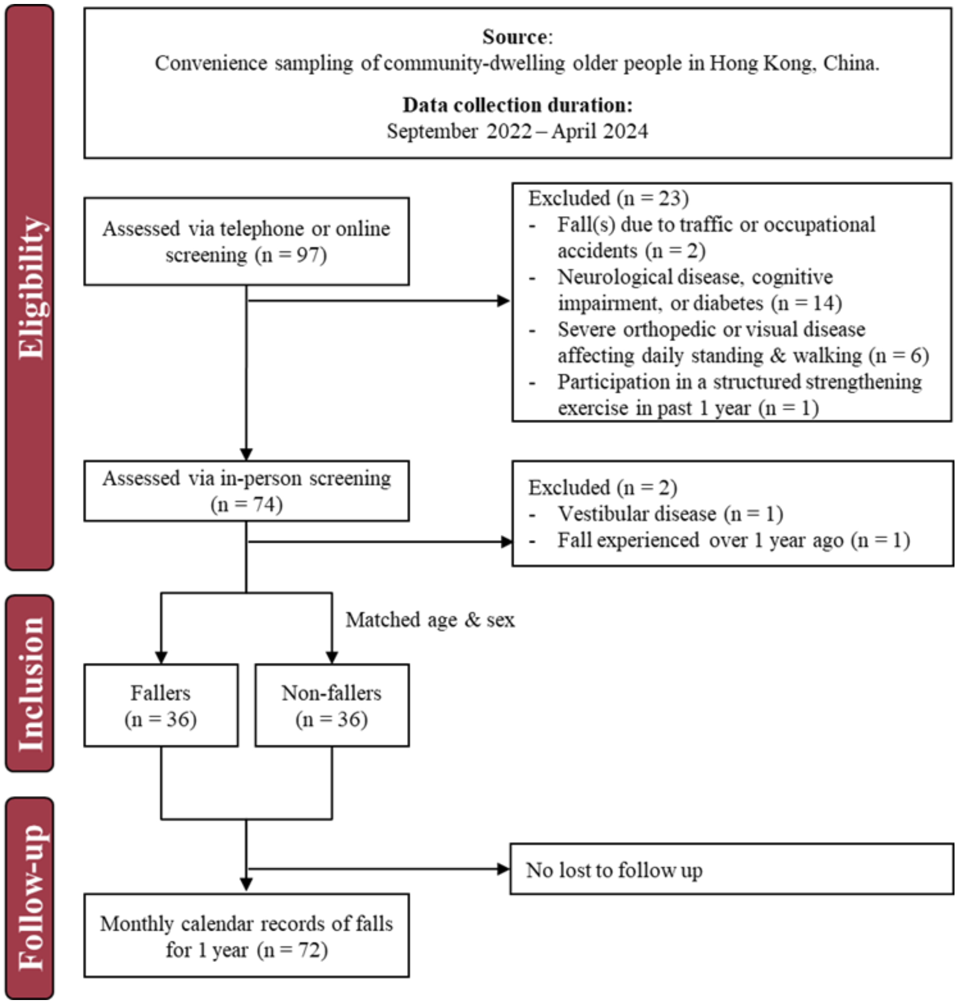


Figure 6-1. The STROBE flow chart.

6.4.5 Statistical Analyses

Statistical analyses were conducted using SPSS and the two-tailed significance level was set at 0.05 (version 25.0). To compare subjective assessment data between older adults with different fall status (i.e., fallers vs. non-fallers, with vs. without 1-year prospective falling), the independent sample t tests, Mann-Whitney U tests, and Chi-square tests were conducted for the normally distributed continuous data, non-normally distributed continuous data, and categorical data, respectively. The Chi-square tests and post hoc pairwise comparisons with Bonferroni corrections were conducted to examine how the ratios of varied stepping responses (“stepping” factor) differed between fall status groups (“fall” factor) following large and medium magnitude of perturbations, considering the ratios of stepping responses following small magnitude of perturbations were mostly less than 10% (Bair et al., 2016).

6.4.5.1 Cross-sectional Analyses

For each amplitude/temporal parameter of the examined CoM displacements, joint motions, joint moments, joint powers, and EMG signals in fallers vs. non-fallers, the independent sample t test or Mann-Whitney U test was used based on the data normality. Partial Pearson correlation tests with age, sex, and body mass index (BMI) adjusted were conducted separately between each examined biomechanical/neuromuscular parameter following large magnitude of perturbations and the number of falls in previous one year. The correlation coefficient, *r*, represents the very small (<0.1), small (0.1-0.3), medium (0.3-0.5), or large effect size (>0.5) (Portney, 2020).

6.4.5.2 Prospective Cohort Analyses

The Logistic regression analyses with or without confounders adjusted were conducted to model the causal relationship between each examined biomechanical/neuromuscular parameter in reactive balance control and the older adults' prospective fall status (≥1 fall vs. no fall within 1-year tracking). Model 1 did not adjust any confounder. Model 2 adjusted the subjects' age, sex, and BMI (Y. Kim et al., 2017). Model 3 adjusted the subjects' fall history status, FES-I short version score, Mini-BESTest score, and PASE-C score besides the confounders in Model 2. The odds ratio (OR) indicates the effect size categorized as large, medium, small, or very small (Chen et al., 2010).

The receiver operating characteristic (ROC) analysis was conducted to examine the diagnostic accuracy of a biomechanical/neuromuscular parameter during reactive balance control in classifying prospective fall status, when any of the three Logistic regression models showed its significant causal relationship with the prospective fall status. The area under the ROC curve (AUC) was obtained to quantify the classification ability as poor (0.5-0.7), acceptable (0.7-0.8), excellent (0.8-0.9), or outstanding (0.9-1.0) (Hosmer Jr et al., 2013). The optimal cut-point value of a biomechanical/neuromuscular parameter during reactive balance control in differentiating the older adults' 1-year prospective fall status was determined as the point on a ROC curve with the maximum sum of sensitivity and specificity (Fluss et al., 2005). The sensitivity and specificity of the optimal cut-point value were also extracted.

6.5 Results

6.5.1 Results of Subjective Assessments and Prospective Fall Status

No significant difference was observed in the age, sex ratio, body mass, height, BMI, leg length, leg dominance, short FES-I score, PASE-C score, or the Mini-BESTest score between the older adults who had fall histories and those who did not (fallers vs. non-fallers) (**Table 6-1**). No older subject dropped out of the follow-up of prospective falls within 1 year after the instrumented assessment of reactive balance control. A total of 26 subjects experienced the 1-year prospective falling, i.e., at least 1 fall during the follow-up period, among whom seven subjects experienced 2 or more prospective falls, and 46 subjects did not experience prospective falling. Regarding the above-mentioned subjective assessment contents, there was no significant difference between the older adults who had prospective falling and those who did not, either (**Table 6-1**).

Table 6-1. Subjective assessment results in older adults with different fall status.

	Fallers (n = 36)	Non-fallers (n = 36)	With 1-year prospective falling (n = 26)	Without 1-year prospective falling (n = 46)
Number of falls in past/prospective 1 year	1.3 ± 0.7*	0*	1.3 ± 0.5*	0*
Age (year)	68.7 ± 3.5	69.3 ± 3.2	68.7 ± 3.0	69.2 ± 3.5
Sex ratio (female/male)	31/5	31/5	20/6	30/16
Body mass (kg)	58.3 ± 8.4	58.7 ± 9.0	59.1 ± 7.0	58.1 ± 9.5
Height (cm)	158.7 ± 7.2	157.5 ± 15.7	159.1 ± 7.1	159.3 ± 9.1
BMI (kg/m^2)	23.2 ± 2.9	23.0 ± 2.5	23.4 ± 2.5	22.9 ± 2.8
Leg length (cm)	79.4 ± 3.1	80.6 ± 5.3	79.6 ± 4.5	80.2 ± 4.4
Leg dominance (right/left)	25/11	25/11	3/23	7/39
Short FES-I (score)	14.5 ± 5.9	12.6 ± 4.2	14.9 ± 5.2	12.8 ± 5.1
PASE-C (score)	120.0 ± 56.3	124.0 ± 57.0	119.1 ± 57.6	123.6 ± 56.1
Mini-BESTest (score)	23.6 ± 2.7	24.4 ± 1.9	23.4 ± 2.3	24.3 ± 2.4

Note: Continuous data are displayed as mean ± standard deviation. Categorical data are displayed as the ratio. * indicates significant difference between older subjects with and without falling in past/prospective 1 year. **BMI**: body mass index. **FES-I**: fall efficacy scale-international. **PASE-C**: physical activity scale of elderly-Chinese. **Mini-BESTest**: mini-balance evaluation system test.

Figure 6-2. The waist-pull perturbation system.

6.5.2 Stepping Responses in Groups of Different Fall Status

Fallers exhibited a higher frequency to have stepping responses, especially the multiple steps and unloaded-leg stepping responses, in contrast to non-fallers following unexpected mediolateral perturbations (**Table 6-2**). The ratio of perturbations with no step following lateral perturbations was significantly smaller in fallers than that in non-fallers ($p < 0.05$), indicating that fallers had more stepping responses than non-fallers. The significantly larger ratio of subjects/perturbations with multiple steps and smaller ratio of making a single step following mediolateral perturbations in fallers indicated that they tended to have multiple steps in contrast to non-fallers ($p < 0.05$). The significantly larger ratio of perturbations with a first dominant-leg step following medial perturbations and a first non-dominant-leg step following lateral perturbations in fallers indicated that they tended to have unloaded-leg stepping responses in contrast to non-fallers ($p < 0.05$).

In contrast to those without 1-year prospective falling, older subjects with 1-year prospective falling had more stepping responses following unexpected anterior perturbations and fewer stepping responses, especially the loaded-leg stepping responses, following unexpected lateral perturbations (**Table 6-2**). The significantly larger ratio of perturbations with a single step following large magnitude of anterior perturbations as well as the significantly larger ratio of participants with multiple steps and larger ratio of perturbations with a first non-dominant-leg step following medium magnitude of anterior perturbations in older subjects with 1-year prospective falling indicated that they were more likely to have stepping responses following forward loss of balance in contrast to the older subjects without 1-year prospective falling ($p < 0.05$). Following medium magnitude of lateral perturbations, the significantly smaller ratio of perturbations with multiple steps and smaller ratio of perturbations with a first dominant-leg step in older subjects with 1-year prospective falling indicated that they used less loaded-leg stepping strategy than older subjects with 1-year prospective falling ($p < 0.05$).

Table 6-2. Stepping responses following unexpected waist-pull perturbations (n = 72).

Perturbation magnitude	Perturbation direction	**Ratio of subjects making multiple steps, a single step, and no step**	**Frequency of making multiple steps, a single step, and no step**	**Frequency of firstly making a dominant-leg step, a non-dominant-leg step, and no step**	**Ratio of subjects making multiple steps, a single step, and no step**	**Frequency of making multiple steps, a single step, and no step**	**Frequency of firstly making a dominant-leg step, a non-dominant-leg step, and no step**
		Fallers (n = 36, 180 perturbations)			**Non-fallers (n = 36, 180 perturbations)**		
Large	Anterior	14/36, 13/36, 9/36	29/180, 63/180, 88/180	62/180, 30/180, 88/180	15/36, 12/36, 9/36	27/180, 54/180, 99/180	44/180, 37/180, 99/180
	Posterior	6/36, 5/36, 25/36	9/180, 17/180, 154/180	15/180, 11/180, 154/180	6/36, 7/36, 23/36	11/180, 20/180, 149/180	17/180, 14/180, 149/180
	Medial	**20/36***, **16/36***, 0/36	**49/180***†, **126/180***, 5/180†	137/180, 38/180, 5/180	**10/36***, **26/36***, 0/36	**22/180***†, **148/180***, 10/180†	130/180, 40/180, 10/180
	Lateral	**17/36***, 19/36, 0/36	**38/180***†, 135/180$^{*\#}$, **7/180**$^{\dagger\#}$	24/180, **149/180**$^{\#}$, **7/180**$^{\#}$	**9/36***, 25/36, 2/36	**15/180***†, 139/180$^{*\#}$, **26/180**$^{\dagger\#}$	32/180, **122/180**$^{\#}$, **26/180**$^{\#}$
Medium	Anterior	8/36, 4/36, 24/36	13/180, 18/180, 149/180	14/180, 17/180, 149/180	7/36, 5/36, 24/36	8/180, 17/180, 155/180	17/180, 8/180, 155/180
	Posterior	5/36, 1/36, 30/36	5/180, 4/180, 171/180	5/180, 4/180, 171/180	2/36, 3/36, 31/36	2/180, 8/180, 170/180	7/180, 3/180, 170/180
	Medial	9/36, 23/36, 4/36	14/180, 107/180, 59/180	**109/180**, 12/180, 59/180	5/36, 27/36, 4/36	10/180, 99/180, 71/180	**89/180**, 20/180, 71/180
	Lateral	6/36, 20/36, 10/36	**17/180**†, 72/180, 91/180†	11/180, **78/180**, 91/180	2/36, 22/36, 12/36	**6/180**†, 72/180, 102/180†	20/180, **58/180**, 102/180
Small	Anterior	2/36, 1/36, 33/36	2/180, 1/180, 177/180	2/180, 1/180, 177/180	2/36, 3/36, 31/36	2/180, 5/180, 173/180	4/180, 3/180, 173/180
	Posterior	0/36, 0/36, 36/36	0/180, 0/180, 180/180	0/180, 0/180, 180/180	1/36, 1/36, 34/36	1/180, 3/180, 176/180	4/180, 0/180, 176/180
	Medial	0/36, 3/36, 33/36	0/180, 10/180, 170/180	10/180, 0/180, 170/180	3/36, 3/36, 30/36	3/180, 6/180, 171/180	1/180, 8/180, 171/180
	Lateral	0/36, 5/36, 31/36	0/180, 7/180, 173/180	2/180, 5/180, 173/180	2/36, 4/36, 30/36	2/180, 6/180, 172/180	4/180, 4/180, 172/180
		With 1-year prospective falling (n = 26, 130 perturbations)			**Without 1-year prospective falling (n = 46, 230 perturbations)**		
Large	Anterior	9/26, 11/26, 6/26	17/130, **52/130**, 61/130	44/130, 25/130, 61/130	20/46, 14/46, 12/46	39/230, **65/230**, 126/230	61/230, 43/230, 126/230
	Posterior	4/26, 4/26, 18/26	4/130, 16/130, 110/130	10/130, 10/130, 110/130	8/46, 8/46, 30/46	16/230, 21/230, 193/230	19/230, 18/230, 193/230
	Medial	13/26, 13/26, 0/26	29/130, 95/130, 6/130	99/130, 25/130, 6/130	17/46, 29/46, 0/46	42/230, 179/230, 9/230	185/230, 36/230, 9/230
	Lateral	11/26, 15/26, 0/26	19/130, 101/130, 10/130	12/130, 108/130, 10/130	15/46, 29/46, 2/46	34/230, 173/230, 23/230	23/230, 184/230, 23/230
Medium	Anterior	**9/26**, 3/26, 14/26	10/130, 13/130, 107/130	8/130*, **15/130**$^{*\#}$, 107/130$^{\#}$	**6/46**, 6/46, 34/46	11/230, 22/230, 197/230	23/230*, **10/230**$^{*\#}$, 197/230$^{\#}$
	Posterior	3/26, 2/26, 21/26	3/130, 5/130, 122/130	6/130, 2/130, 122/130	4/46, 2/46, 40/46	4/230, 7/230, 219/230	6/230, 5/230, 219/230
	Medial	7/26, 17/26, 2/26	7/130, 75/130, 48/130	76/130, 6/130, 48/130	7/46, 33/46, 6/46	17/230, 131/230, 82/230	134/230, 14/230, 82/230
	Lateral	1/26, 17/26, 8/26	**2/130***†, 54/130*, 74/130†	**2/130**†, 54/130, 74/130†	7/46, 25/46, 14/46	**21/230***†, 90/230*, 119/230†	**17/230**†, 94/230, 119/230†
Small	Anterior	1/26, 1/26, 24/26	1/130, 2/130, 127/130	2/130, 1/130, 127/130	3/46, 3/46, 40/46	3/230, 4/230, 223/230	5/230, 2/230, 223/230
	Posterior	0/26, 1/26, 25/26	0/130, 3/130, 127/130	3/130, 0/130, 127/130	1/46, 0/46, 45/46	1/230, 0/230, 229/230	1/230, 0/230, 229/230
	Medial	1/26, 5/26, 20/26	1/130, 11/130, 118/130	8/130, 4/130, 118/130	2/46, 1/46, 43/46	2/230, 5/230, 223/230	3/230, 4/230, 223/230
	Lateral	2/26, 0/26, 24/26	2/130, 0/130, 128/130	1/130, 1/130, 128/130	0/46, 9/46, 37/46	0/230, 13/230, 217/230	4/230, 9/230, 217/230

Note: The bold texts with shading indicate the significant differences between different fall status groups ($p < 0.05$, simple main effect of “fall” factor). The same superscript symbols indicate the pairwise comparison of stepping responses with significant difference. ($p < 0.05$, simple main effect of “stepping” factor).

6.5.3 Neuromuscular/Biomechanical Responses from Cross-sectional Analyses

6.5.3.1 Fallers vs. Non-fallers

As shown in **Table 6-3**, in response to anterior perturbations, fallers exhibited significantly shorter times to peak knee flexion angles, shorter onset latencies for ankle dorsiflexion, shorter burst durations for ankle plantarflexion moments, and longer onset latencies for backward COM displacement than non-fallers ($p < 0.05$).

In the case of posterior perturbations, fallers tended to utilize the ankle strategy as compared to non-fallers. This was indicated by the fallers' significantly higher co-contraction index (CCI) of the ankle dorsiflexor-plantarflexor, shorter onset latencies for ankle sagittal power generation ($p < 0.05$), and exhibited slower knee joint responses, as evidenced by significantly smaller rates of EMG rise in the knee extensor, longer times to peak for knee sagittal power generation, and longer onset latencies for knee flexion ($p < 0.05$) as compared to non-fallers.

Following medial perturbations, fallers demonstrated significantly longer EMG burst durations in the hip flexor, increased peak hip joint powers, larger peak hip abduction and adduction angles, longer burst durations in knee flexion, and greater medial center of mass (CoM) displacement than non-fallers ($p < 0.05$). Additionally, fallers and non-fallers exhibited significant differences in lateral, backward, and downward CoM displacements ($p < 0.05$), suggesting that fallers required more body adjustments to execute multiple steps than non-fallers.

After lateral perturbations, fallers showed significantly longer times to peak EMG amplitude in the hip adductor, longer onset latencies for knee flexion, and longer times to peak and burst durations for lateral CoM displacement than non-fallers ($p < 0.05$).

Table 6-3. Significant differences following the large magnitude of unpredictable perturbations between fallers (n = 36) and non-fallers (n = 36).

Direction	Parameter		Unit	Fallers (n=36)	Non-fallers (n=36)	p value
A	Hip Abductor	Time to peak EMG amplitude	ms	667 (415)	942 (521)	0.024
A	Hip Abduction	Peak angle	°	3 (3)	2 (2)	0.049
A	Knee Flexion	Time to peak angle	ms	852 (492)	1033 (555)	0.014
A	Ankle Plantarflexion	Moment burst duration	ms	481 (386)	688 (497)	0.049
A	Ankle Dorsiflexion	Angle onset latency	ms	109 (337)	228 (593)	0.046
A	Backward CoM Displacement	Onset latency	ms	2000 (318)	1765 (414)	0.03
P	Hip Adductor	EMG onset latency	ms	238 (96)	212 (61)	0.031
P	Hip Abduction	Time to peak angle	ms	933 (663)	1235 (623)	0.013
P	Knee Extensor	Rate of EMG rise	multiples of baseline$\cdot s^{-1}$	15.8 (13.8)	24.5 (24.7)	0.019
P	Knee Sagittal Power Generation	Time to peak	ms	805 ± 332	660 ± 246	0.039
P	Knee Flexion	Angle onset latency	ms	204 (90)	173 (59)	0.02
P	Ankle Dorsiflexor-Plantarflexor	CCI	/	39% (36%)	23% (33%)	0.039
P	Ankle Sagittal Power Generation	Power onset latency	ms	150 (307)	323 (355)	0.012
P	Ankle Dorsiflexion	Moment onset latency	ms	1058 ± 292	890 ± 332	0.025
P	Downward CoM Displacement	Burst duration	ms	799 (839)	1151 (947)	0.034
M	Hip Flexor	EMG burst duration	ms	548 (733)	339 (389)	0.039
M	Hip Frontal Power Absorption	Peak power	$W\cdot kg^{-1}$	0.11 (0.11)	0.05 (0.06)	0.01
M	Hip Sagittal Power Generation	Peak power	$W\cdot kg^{-1}$	0.09 (0.16)	0.05 (0.06)	0.043
M	Hip Sagittal Power Absorption	Peak power	$W\cdot kg^{-1}$	0.09 (0.13)	0.07 (0.05)	0.044
M	Hip Adduction	Peak angle	°	4 (3)	2 (3)	0.044
M	Hip Abduction	Peak angle	°	7 (5)	4 (3)	0.015
M	Knee Flexion	Angle burst duration	ms	2633 (2256)	1751 (1413)	0.047
M	Medial CoM Displacement	Peak CoM Displacement	mm	147 ± 46	121 ± 40	0.014
M	Medial CoM Displacement	Burst duration	ms	2466 (992)	1878 (1178)	0.01
M	Lateral CoM Displacement	Onset latency	ms	1792 (381)	1595 (458)	0.029
M	Lateral CoM Displacement	Time to peak	ms	2000 (160)	1869 (286)	0.04
M	Backward CoM Displacement	Time to peak	ms	1417 (287)	1261 (276)	0.01
M	Backward CoM Displacement	Peak CoM Displacement	mm	40 (43)	21 (26)	0.011
M	Backward CoM Displacement	Burst duration	ms	1804 (1798)	1278 (889)	0.009
M	Downward CoM Displacement	Onset latency	ms	917 ± 506	1171 ± 491	0.034
L	Hip Adductor	Time to peak EMG amplitude	ms	755 (325)	633 (245)	0.044
L	Knee Flexion	Angle onset latency	ms	373 (379)	288 (265)	0.024
L	Lateral CoM Displacement	Time to peak	ms	945 (207)	851 (152)	0.033
L	Lateral CoM Displacement	Burst duration	ms	2268 (1727)	1771 (847)	0.019
L	Medial CoM Displacement	Time to peak	ms	1963 (201)	1868 (315)	0.041

Note: A: anterior perturbation. P: posterior perturbation. M: medial perturbation. L: lateral perturbation. CoM: center of mass. EMG: electromyographic.

6.5.3.2 Partial Correlation Analysis Results

Table 6-4 presents the results of partial correlation tests between the number of 1-year previous falls and the examined responses of CoM, joint motions, joint moments, joint powers, and EMG signals after large magnitude of unexpected perturbations.

The more previous falls in older subjects were related to their delayed but enlarged reactions as well as their elongated recovery of CoM displacement. There were significantly positive correlations in small to medium effect sizes between the number of 1-year previous falls and the onset latency of backward CoM displacement after posterior perturbations (r = 0.303, p < 0.05), the time to peak lateral CoM displacement after lateral perturbations (r = 0.281, p < 0.05), the peak medial CoM displacement after medial perturbations (r = 0.334, p < 0.05), as well as the burst duration of forward, medial, or lateral CoM displacement after anterior (r = 0.298, p < 0.05), medial (r = 0.324, p < 0.05), or lateral perturbations (r = 0.244, p < 0.05), respectively.

Correlations between the number of 1-year previous falls in older subjects and their responses of lower-limb joint motions showed small effect sizes. The more previous falls were related to the shorter time to peak knee flexion angle after anterior perturbations (r = -0.245, p < 0.05), the longer time to peak knee flexion angle after posterior perturbations (r = 0.246, p < 0.05), the larger peak hip adduction angle after medial perturbations (r = 0.271, p < 0.05), as well as the shorter onset latency of knee extension (r = -0.257, p < 0.05), longer onset latency of ankle dorsiflexion (r = 0.295, p < 0.05), shorter time to peak knee extension angle (r = -0.238, p < 0.05), and longer time to peak knee flexion angle (r = 0.260, p < 0.05) and ankle dorsiflexion angle (r = 0.273, p < 0.05) after lateral perturbations.

There were significantly positive correlations in small to medium effect sizes between the number of 1-year previous falls and some responses of lower-limb joint moments. The correlations varied for the responses after different direction of perturbations. After posterior perturbations, the more previous falls were related to the elongated reaction as indicated by longer time to peak ankle plantarflexion moment (r = 0.261, p < 0.05) and related to the elongated recovery as indicated by longer onset latency of ankle dorsiflexion moment (r = 0.287, p < 0.05). After medial perturbations, the more previous falls were related to the delayed but faster moment reactions as indicated by the longer onset latency of hip abduction (r = 0.254, p < 0.05), hip flexion (r = 0.318, p < 0.05), as well as knee extension moment (r = 0.250, p < 0.05) but larger rate of rise of hip abduction (r = 0.304, p < 0.05), hip flexion (r = 0.335, p < 0.05), as well as knee extension moment (r = 0.382, p

< 0.05); the more previous falls were also related to the elongated recovery as indicated by the longer burst duration of ankle plantarflexion moment (*r* = 0.246, $p < 0.05$). After lateral perturbations, the more previous falls were related to the elongated recovery as indicated by the longer burst duration of hip flexion moment (*r* = 0.272, $p < 0.05$).

Small effect sizes showed that the more previous falls were significantly related to the lager peak hip frontal power generation (*r* = 0.283, $p < 0.05$) and absorption (*r* = 0.277, $p < 0.05$) after anterior perturbations, and the shorter onset latencies of ankle sagittal power generation after posterior perturbations (*r* = -0.245, $p < 0.05$).

With small to medium effect sizes, the more 1-year previous falls were related to the decreased hip muscle activation and increased ankle muscle activation during standing without perturbations, while the correlations between the number of previous falls and the EMG responses after perturbations varied for different perturbation directions. Regarding the baseline EMG responses before perturbations, the more previous falls were related to the smaller baseline EMG amplitude of hip abductor after posterior perturbations (*r* = -0.297, $p < 0.05$) as well as the larger baseline EMG amplitudes of ankle plantarflexor after posterior (*r* = 0.244, $p < 0.05$), medial (*r* = 0.353, $p < 0.05$), and lateral (*r* = 0.247, $p < 0.05$) perturbations. Regarding the EMG responses after perturbations, the more previous falls were related to the delayed reaction of agonist muscle as indicated by the longer EMG onset latency of ankle plantarflexor after anterior perturbations (*r* = 0.329, $p < 0.05$), related to the larger ankle dorsiflexor-plantarflexor CCI after posterior perturbations (*r* = 0.263, $p < 0.05$), related to the larger peak EMG amplitude of knee flexor (*r* = 0.240, $p < 0.05$) together with shorter EMG burst duration of ankle plantarflexor (*r* = -0.254, $p < 0.05$) after medial perturbations, and related to the elongated muscle activation as indicated by the longer time to peak EMG amplitudes of hip flexor (*r* = 0.259, $p < 0.05$) and ankle dorsiflexor (*r* = 0.264, $p < 0.05$) as well as the longer EMG burst duration of hip extensor (*r* = 0.284, $p < 0.05$) after lateral perturbations.

Table 6-4. Correlation coefficient (r) for the partial Pearson correlation test between number of previous falls and each examined parameter after large magnitude of perturbations.

		Onset latency				Time to peak				Peak amplitude				Burst duration				Rate of rise				CCI			
		A	P	M	L	A	P	M	L	A	P	M	L	A	P	M	L	A	P	M	L	A	P	M	L
CoM	Forward Displacement	0.129	0.203	-0.015	0.141	0.056	0.079	0.092	0.016	0.177	-0.034	**0.249***	0.091	**0.298***	-0.164	-0.003	0.034								
CoM	Backward Displacement	**0.238***	**0.303***	-0.012	0.045	0.11	0.044	0.216	0.015	-0.139	-0.004	**0.381***	-0.069	-0.016	0.015	**0.29***	-0.121								
CoM	Medial Displacement	0.031	0.002	-0.062	**0.278***	-0.024	0.13	0.147	**0.264***	0.06	-0.094	**0.334***	-0.073	0.14	0.049	**0.324***	-0.122								
CoM	Lateral Displacement	0.203	-0.001	**0.296***	-0.064	0.108	0.089	**0.24***	**0.281***	**0.345***	-0.151	-0.062	0.145	0.143	0.082	0.103	**0.244***								
CoM	Upward Displacement	0.025	0.003	0.128	0.11	0.033	0.192	0.224	0.169	0.061	0.067	-0.148	0.175	-0.006	**0.347***	-0.122	-0.027								
CoM	Downward Displacement	0.071	0.035	-0.232	-0.077	0.099	0.12	-0.169	-0.05	0.064	-0.024	**0.28***	0.004	0.19	**-0.262***	0.053	0.073								
Joint motion	Hip Adduction	-0.012	0.033	0.078	-0.141	0.057	0.086	0.142	-0.031	0.082	-0.102	**0.271***	0.186	-0.017	0.059	-0.076	0.063								
Joint motion	Hip Abduction	-0.083	0.076	-0.053	0.081	-0.063	-0.106	-0.019	0.096	0.213	-0.071	0.171	0.227	0.036	0.151	0.01	-0.023								
Joint motion	Hip Flexion	0.029	-0.13	-0.112	0.029	0.014	0.14	0.109	0.081	0.128	-0.009	0.207	0.204	0.188	0.058	0.176	0.147								
Joint motion	Hip Extension	-0.072	0.082	0.055	0.087	-0.044	-0.187	-0.044	0.015	0.113	-0.005	-0.152	-0.106	0.03	-0.158	-0.159	-0.104								
Joint motion	Knee Flexion	-0.206	0.135	-0.147	0.222	**-0.245***	**0.246***	0.158	**0.26***	0.073	-0.038	0.08	0.006	0.13	-0.042	0.221	0.019								
Joint motion	Knee Extension	0.067	0.067	-0.03	**-0.257***	0.118	0.083	-0.08	**-0.238***	-0.121	0.033	0	0.136	-0.073	-0.118	0.048	-0.137								
Joint motion	Ankle Dorsiflexion	-0.057	0.058	0.037	**0.295***	-0.073	-0.031	0.055	**0.273***	0.022	0.085	0.109	-0.01	-0.087	-0.106	0.027	0.05								
Joint motion	Ankle Plantarflexion	0.025	-0.033	0.054	-0.154	0.122	0.002	0.052	-0.194	0.022	0.127	-0.02	0.1	-0.034	-0.217	-0.073	-0.148								
Joint moment	Hip Adduction	-0.156	0.012	-0.162	0.111	0.078	-0.094	-0.049	0.143	-0.117	-0.164	0.021	-0.202	-0.108	-0.129	0.045	0.032	-0.143	-0.129	0.055	-0.114				
Joint moment	Hip Abduction	0.002	0.156	**0.254***	-0.078	-0.123	0.198	-0.108	-0.03	0.235	0.068	0.048	0.137	0.094	0.075	-0.187	-0.119	0.105	-0.003	**0.304***	0.022				
Joint moment	Hip Flexion	0.062	0.184	**0.318***	-0.092	0.142	0.156	0.097	0.039	0.152	-0.075	0.208	0.212	0.063	-0.03	-0.099	**0.272***	0.087	-0.079	**0.335***	0.044				
Joint moment	Hip Extension	-0.071	0.085	-0.114	0.089	-0.021	0.161	0.014	0.058	0.14	-0.106	0.192	0.008	-0.026	0.025	**0.247***	0.023	**0.262***	-0.184	-0.152	0.005				
Joint moment	Knee Flexion	-0.041	0.184	-0.096	**0.262***	-0.079	0.191	-0.003	**0.243***	0.099	-0.123	0.181	-0.068	-0.109	-0.033	**0.394***	-0.07	0.088	-0.122	0.117	0.137				
Joint moment	Knee Extension	0.018	0.102	**0.25***	-0.077	0.123	0.057	0.045	-0.045	0.092	-0.083	0.162	**0.315***	-0.111	-0.122	-0.072	-0.021	-0.052	0.04	**0.382***	0.132				
Joint moment	Ankle Dorsiflexion	0.005	**0.287***	0.124	-0.069	-0.015	0.214	0.1	0.11	0.079	-0.213	0.04	0.152	-0.071	-0.019	-0.096	0.125	-0.055	-0.067	0.024	0.16				
Joint moment	Ankle Plantarflexion	-0.185	0.106	-0.037	0.18	-0.188	**0.261***	0.138	0.163	0.173	-0.092	0.135	0	-0.115	0.038	**0.246***	0.063	0.091	-0.156	0.08	-0.003				
Joint power	Hip Frontal Generation	-0.015	0.083	-0.166	-0.034	0.009	0.018	0.02	0.147	**0.283***	-0.003	0.156	0.131	-0.174	0	0.126	0.193								
Joint power	Hip Frontal Absorption	-0.153	0.034	-0.165	0.096	-0.216	-0.105	-0.016	0.18	**0.277***	-0.118	0.146	0.008	-0.112	-0.034	0.023	-0.156								
Joint power	Hip Sagittal Generation	0.077	0.236	-0.022	0.066	0.127	0.221	0.098	0.186	-0.001	-0.127	0.164	0.101	0.097	0.141	0.004	-0.005								
Joint power	Hip Sagittal Absorption	-0.021	0.014	-0.043	-0.026	-0.088	0.016	-0.026	0.04	0.084	-0.134	0.181	0.145	0.097	0.055	-0.017	0.091								
Joint power	Knee Sagittal Generation	-0.031	0.144	-0.014	-0.101	-0.088	0.157	0.096	-0.07	0.179	-0.006	0.137	0.101	0.038	-0.048	0.021	-0.008								
Joint power	Knee Sagittal Absorption	-0.134	0.21	-0.071	-0.044	0.107	0.086	-0.073	-0.012	0.121	-0.041	0.152	0.018	-0.066	0.065	-0.053	-0.157								
Joint power	Ankle Sagittal Generation	0.04	**-0.245***	0.221	-0.094	0.149	-0.022	0.075	-0.034	0.051	0.061	0.203	0.084	-0.033	-0.076	-0.056	0.047								
Joint power	Ankle Sagittal Absorption	-0.118	-0.075	-0.06	0.164	0.057	0.031	0.097	0.043	0.038	0.055	0	0.014	-0.188	0.017	-0.004	0.028								
EMG signal	Hip Adductor	-0.081	0.069	-0.1	0.058	0.023	0.087	-0.026	0.163	0.11	-0.074	0.213	0.088	0.186	-0.082	0.052	0.154	-0.033	-0.102	-0.171	0.034	0.039	0.087	0.057	0.072
EMG signal	Hip Abductor	-0.162	0.05	-0.019	-0.067	-0.199	0.113	0.073	0.12	-0.107	-0.179	-0.103	-0.142	0.008	0.087	0.106	0.117	-0.158	-0.181	-0.179	-0.152				
EMG signal	Hip Flexor	-0.1	0.067	-0.153	0.055	0.122	0.126	-0.139	**0.259***	0.132	-0.147	0.19	-0.001	0.107	-0.025	0.235	0.121	0.017	-0.188	0.063	-0.179	-0.124	-0.166	-0.039	0.059
EMG signal	Hip Extensor	0.039	0.121	-0.099	0.018	0.229	0.091	0.203	0.181	-0.036	-0.002	-0.041	-0.05	0.063	0.031	0.035	**0.284***	-0.094	-0.159	-0.116	-0.13				
EMG signal	Knee Flexor	-0.02	0.152	-0.019	-0.119	0.004	0.088	0.104	0.17	0.01	-0.134	**0.24***	-0.029	0.063	0.146	0.054	0.059	0.061	-0.128	0.004	-0.004	0.063	-0.108	-0.004	-0.025
EMG signal	Knee Extensor	0.035	0.072	0.149	0.044	0.137	0.139	0.18	0.101	-0.009	-0.158	0.097	-0.072	-0.043	0.022	-0.027	0.024	-0.14	-0.208	0.067	-0.058				
EMG signal	Ankle Dorsiflexor	-0.038	0.001	-0.144	-0.068	-0.169	0.163	0.09	**0.264***	0.043	0.033	0.127	0.07	0.026	0.197	0.203	0.135	0.076	-0.101	0.005	-0.068	0.089	**0.263***	0.08	0.018
EMG signal	Ankle Plantarflexor	**0.329***	0.11	-0.089	0.083	0.128	0.045	0.034	0.141	0.002	-0.024	0.095	0.034	-0.027	0.018	**-0.254***	0.037	-0.036	-0.02	0.05	-0.015				

Data are from all subjects (n = 72). Correlations are adjusted for age, sex, and BMI. * indicates the partial correlation is significant (two-sided, $p < 0.05$). CoM: center of mass. EMG: electromyographic. CCI: agonist-antagonist co-contraction index. A: anterior perturbations. P: posterior perturbations. M: medial perturbations. L: lateral perturbation.

6.5.4 Neuromuscular/Biomechanical Responses from Prospective Cohort Analyses

Table 6-5 firstly presents the biomechanical and neuromuscular parameters after unexpected large magnitude of perturbations that could significantly predict the older subjects' 1-year prospective fall status in at least one Logistic regression model. Regarding these parameters, cut-point values and diagnostic accuracies (AUC, sensitivity, and specificity) are also displayed in **Table 6-5**. The mean changes of CoM displacements (**Figure 6-3**), lower-limb joint angles (**Figure 6-4**), lower-limb moments (**Figure 6-5**), lower-limb powers (**Figure 6-6**), or EMG signals of lower-limb muscles over time (**Figure 6-7**) in older adults with and without prospective falling are presented.

After unexpected anterior perturbations, the activation rate, peak activation, and activation duration of hip abductor could significantly predict the older subjects' 1-year perspective fall status with very small effect sizes and showed significantly poor abilities in classifying older subjects' 1-year perspective fall status. For each 0.1 s increase of the time to peak hip flexion angle after anterior perturbations, older subjects' odds of having at least one fall within prospective 1 year increased by 19% OR (95% CI); however, such significant causal relationship did not exist after controlling the confounders. In Model 3 which adjusted the subject's age, sex, BMI, fall history status, Mini-BESTest score, short FES-I score, and PASE-C score, the one-unit increase of rate of EMG rise, peak EMG amplitude, and EMG burst duration of hip abductor after anterior perturbations could decrease the odds of 1-year prospective falling by 6%, 23%, and 12%, respectively. These parameters, however, showed poor abilities in classifying the older subjects with and without 1-year prospective falling as indicated by the AUCs ranging from 0.64 to 0.68.

After unexpected posterior perturbations, the knee sagittal power responses, hip frontal power responses, and hip abductor activation could significantly predict older subjects' 1-year perspective fall status. The odds of 1-year prospective falling were increased by the earlier and larger responses of knee sagittal power absorption, as well as the delayed timing of knee sagittal power and hip frontal power generation after posterior perturbations. In addition, as indicated by Model 3, the one-unit increase of the rate of EMG rise and the peak EMG amplitude of hip abductor after posterior perturbations could decrease the odds of 1-year prospective falling by 7% and 35%, respectively. These parameters did not show or showed significantly poor abilities in classifying the older subjects with and without 1-year prospective falling as indicated by the AUCs ranging from 0.60 to 0.69.

After unexpected medial perturbations, older subjects' 1-year perspective fall status could be predicted by the anteroposterior CoM displacements, the kinetic responses of hip and knee joints in sagittal plane, and the hip abductor activation. The odds of 1-year prospective falling were increased by the earlier onset of hip sagittal power generation and larger peak knee sagittal power generation, the earlier onset of knee extension moment and shorter duration of hip flexion moment, the delayed timing of knee flexion moment and increased rate of rise and peak value of hip extension moment, the shorter duration of ankle dorsiflexion, as well as the delayed onset of backward CoM displacement and enlarged forward CoM displacement after medial perturbations. Additionally, in Model 3, the one-unit increase of rate of EMG rise and peak EMG amplitude of hip abductor after medial perturbations decreased the odds of 1-year prospective falling by 10% and 29%, respectively. The onset latency of backward CoM displacement (AUC and 95% CI: 0.72 [0.60-0.84]; cut-point value: 618 ms), the rate of EMG rise of hip abductor (AUC and 95% CI: 0.73 [0.62-0.85]; cut-point value: 12 times of baseline EMG amplitude$\cdot s^{-1}$), and the peak EMG amplitude of hip abductor (AUC and 95% CI: 0.73 [0.62-0.85]; cut-point value: 6 times of baseline EMG amplitude) showed significantly acceptable abilities in classifying the older subjects with and without 1-year prospective falling; however, the other parameters after medial perturbations did not show or showed significantly poor classification abilities as indicated by the AUCs ranging from 0.57 to 0.69.

After unexpected lateral perturbations, a series of biomechanical or neuromuscular responses underlying the delayed initiation and insufficient hip abduction and knee extension could significantly predict older subjects' 1-year perspective fall status. In Model 3, the odds of 1-year prospective falling were increased by the elongated duration of knee sagittal power absorption and delayed timing of knee sagittal power generation, the delayed timing of knee extension moment and angle, and the reduced peak knee extension moment; in addition, the increased odds of 1-year prospective falling resulted from the delayed and decreased rate of activation of hip abductor, the delayed onset of hip frontal power generation and earlier timing of hip frontal power absorption, the decreased rate of rise and elongated duration of hip abduction moment, as well as the decreased duration and decreased peak value of hip abduction.

Table 6-5. ROC analyses for the parameters after large magnitude of unexpected waist-pull perturbations that showed significant Logistic regression with prospective fall status.

Perturbation direction	Parameter		Unit	OR (95% CI)			Model 3 OR category	AUC (95% CI)	Cut-point value to predict 1-year fall risk	Sensitivity	Specificity	AUC category
				Model 1	Model 2	Model 3						
Anterior	Hip Abductor	Rate of EMG rise	1 time of EMG baseline$\cdot s^{-1}$	**0.94 (0.89 – 1.00) ***	**0.94 (0.89 – 1.00) ***	**0.94 (0.88 – 1.00) ***	Very small	**0.66 (0.52 - 0.79) ***	< 7 baseline$\cdot s^{-1}$	0.46	0.85	Poor
Anterior	Hip Abductor	Peak EMG amplitude	1 time of EMG baseline	**0.79 (0.63 – 1.00) ***	0.79 (0.63 – 1.00)	**0.77 (0.60 - 0.99) ***	Very small	**0.64 (0.50 - 0.77) ***	< 3 baseline	0.5	0.8	Poor
Anterior	Hip Abductor	EMG burst duration	0.1 s	**0.91 (0.82 - 1.00) ***	0.91 (0.83 - 1.01)	**0.88 (0.77 – 1.00) ***	Very small	**0.68 (0.55 - 0.82) ***	< 384 ms	0.58	0.78	Poor
Anterior	Hip Flexion	Time to peak angle	0.1 s	**1.19 (1.01 - 1.41) ***	1.18 (1.00 - 1.40)	1.19 (1.00 - 1.41)	Very small	**0.64 (0.51 - 0.78) ***	> 1422 ms	0.58	0.72	Poor
Posterior	Hip Abductor	Rate of EMG rise	1 time of EMG baseline$\cdot s^{-1}$	**0.93 (0.88 - 0.99) ***	**0.93 (0.88 - 0.99) ***	**0.93 (0.87 – 1.00) ***	Very small	**0.69 (0.56 - 0.81) ***	< 14 baseline$\cdot s^{-1}$	0.81	0.52	Poor
Posterior	Hip Abductor	Peak EMG amplitude	1 time of EMG baseline	**0.72 (0.52 - 0.98) ***	0.71 (0.50 – 1.00)	**0.65 (0.45 - 0.94) ***	Very small	**0.68 (0.54 - 0.81) ***	< 2 baseline	0.42	0.91	Poor
Posterior	Hip Frontal Power Absorption	Time to peak	0.1 s	**0.84 (0.70 - 1.00) ***	**0.84 (0.70 – 1.00) ***	**0.83 (0.69 – 1.00) ***	Very small	**0.66 (0.52 - 0.79) ***	< 748 ms	0.73	0.61	Poor
Posterior	Hip Frontal Power Generation	Time to peak	0.1 s	**1.20 (1.02 - 1.43) ***	**1.21 (1.02 - 1.45) ***	**1.22 (1.01 - 1.48) ***	Very small	0.60 (0.45 - 0.74)	> 838 ms	0.42	0.83	Poor
Posterior	Knee Sagittal Power Absorption	Onset latency	0.1 s	0.80 (0.60 - 1.07)	0.8 (0.59 - 1.08)	**0.69 (0.48 - 0.99) ***	Very small	**0.68 (0.55 - 0.81) ***	< 232 ms	0.77	0.61	Poor
Posterior	Knee Sagittal Power Absorption	Peak power	0.1 W$\cdot kg^{-1}$	**1.74 (1.07 - 2.83) ***	**1.89 (1.14 - 3.12) ***	**1.99 (1.19 - 3.34) ***	Small	**0.64 (0.51 - 0.78) ***	> 0.13 W$\cdot kg^{-1}$	0.39	0.91	Poor
Posterior	Knee Sagittal Power Generation	Onset latency	0.1 s	**1.23 (1.02 - 1.49) ***	**1.26 (1.03 - 1.55) ***	**1.25 (1.00 - 1.55) ***	Very small	**0.64 (0.51 - 0.78) ***	> 483 ms	0.54	0.76	Poor
Posterior	Knee Sagittal Power Generation	Time to peak	0.1 s	**1.20 (1.01 - 1.42) ***	**1.26 (1.04 - 1.52) ***	**1.24 (1.02 - 1.52) ***	Very small	**0.64 (0.50 - 0.78) ***	> 827 ms	0.54	0.8	Poor
Medial	Hip Abductor	Rate of EMG rise	1 time of EMG baseline$\cdot s^{-1}$	**0.93 (0.88 - 0.98) ***	**0.91 (0.86 - 0.97) ***	**0.9 (0.84 - 0.97) ***	Very small	**0.73 (0.62 - 0.85) ***	< 12 baseline$\cdot s^{-1}$	0.65	0.74	Acceptable
Medial	Hip Abductor	Peak EMG amplitude	1 time of EMG baseline	**0.77 (0.63 - 0.94) ***	**0.76 (0.61 - 0.94) ***	**0.71 (0.55 - 0.92) ***	Very small	**0.71 (0.60 - 0.83) ***	< 6 baseline	0.92	0.46	Acceptable
Medial	Hip Sagittal Power Generation	Onset latency	0.1 s	0.83 (0.68 - 1.01)	0.82 (0.66 - 1.01)	**0.79 (0.62 – 1.00) ***	Very small	0.63 (0.50 - 0.76)	< 379 ms	0.81	0.5	Poor
Medial	Hip Flexion	Moment burst duration	0.1 s	**0.95 (0.90 - 0.99) ***	**0.95 (0.90 – 1.00) ***	0.95 (0.90 – 1.00)	Very small	**0.66 (0.53 - 0.79) ***	< 1504 ms	0.62	0.74	Poor
Medial	Hip Extension	Rate of moment rise	1 N$\cdot$m$\cdot kg^{-1}\cdot s^{-1}$	**2.80 (1.11 - 7.04) ***	**2.81 (1.09 - 7.22) ***	**3.56 (1.25 - 10.14) ***	Medium	**0.64 (0.50 - 0.78) ***	> 0.96 N$\cdot$m$\cdot kg^{-1}\cdot s^{-1}$	0.46	0.83	Poor
Medial	Hip Extension	Peak moment	0.1 N$\cdot$m$\cdot$ kg^{-1}	**1.78 (1.10 - 2.89) ***	**1.81 (1.07 - 3.06) ***	1.67 (0.97 - 2.88)	Very small	**0.66 (0.52 - 0.79) ***	> 0.13 N$\cdot$m$\cdot kg^{-1}$	0.58	0.8	Poor
Medial	Knee Sagittal Power Generation	Peak power	0.1 W$\cdot kg^{-1}$	**1.73 (1.08 - 2.76) ***	**1.74 (1.07 - 2.83) ***	1.61 (0.99 - 2.61)	Very small	0.58 (0.42 - 0.73)	> 0.16 W$\cdot kg^{-1}$	0.42	0.91	Poor
Medial	Knee Extension	Moment onset latency	0.1 s	0.89 (0.78 - 1.01)	0.88 (0.77 - 1.01)	**0.76 (0.63 - 0.93) ***	Very small	**0.68 (0.55 - 0.82) ***	< 250 ms	0.69	0.74	Poor
Medial	Knee Flexion	Moment onset latency	0.1 s	**1.09 (1.00 - 1.19) ***	**1.11 (1.01 - 1.22) ***	**1.15 (1.03 - 1.28) ***	Very small	**0.64 (0.51 - 0.78) ***	> 828 ms	0.62	0.7	Poor
Medial	Knee Flexion	Time to peak moment	0.1 s	**1.20 (1.05 - 1.37) ***	**1.25 (1.07 - 1.46) ***	**1.30 (1.10 - 1.55) ***	Very small	**0.69 (0.57 - 0.82) ***	> 1148 ms	0.85	0.5	Poor
Medial	Ankle Dorsiflexion	Angle burst duration	0.1 s	0.92 (0.84 – 1.00)	0.91 (0.83 – 1.00)	**0.89 (0.80 - 0.99) ***	Very small	**0.65 (0.52 - 0.78) ***	< 1115 ms	0.85	0.41	Poor
Medial	Backward CoM displacement	Onset latency	0.1 s	**1.39 (1.09 - 1.78) ***	**1.42 (1.09 - 1.85) ***	**1.51 (1.12 - 2.04) ***	Very small	**0.72 (0.60 - 0.84) ***	> 618 ms	0.65	0.74	Acceptable
Medial	Forward CoM displacement	Peak amplitude	1 cm	1.36 (0.91 - 2.03)	**1.54 (1.00 - 2.36) ***	1.43 (0.92 - 2.23)	Very small	0.57 (0.43 - 0.72)	> 26 mm	0.35	0.89	Poor
Lateral	Hip Abductor	EMG onset latency	0.1 s	**2.85 (1.24 - 6.53) ***	**2.75 (1.22 - 6.25) ***	**3.12 (1.30 - 7.50) ***	Small	**0.71 (0.59 - 0.83) ***	> 183 ms	0.46	0.89	Acceptable
Lateral	Hip Abductor	Rate of EMG rise	1 time of EMG baseline$\cdot s^{-1}$	**0.97 (0.95 – 1.00) ***	**0.97 (0.95 – 1.00) ***	**0.97 (0.95 – 1.00) ***	Very small	**0.67 (0.55 - 0.80) ***	< 44 baseline$\cdot s^{-1}$	0.92	0.44	Poor
Lateral	Hip Frontal Power Generation	Onset latency	0.1 s	1.20 (0.97 - 1.47)	1.25 (1.00 - 1.56)	**1.28 (1.01 - 1.63) ***	Very small	0.62 (0.48 - 0.76)	> 327 ms	0.58	0.74	Poor
Lateral	Hip Frontal Power Absorption	Time to peak	0.1 s	0.89 (0.73 - 1.09)	0.84 (0.67 - 1.05)	**0.76 (0.59 - 0.97) ***	Very small	0.62 (0.49 - 0.76)	< 1105 ms	0.23	0.83	Poor
Lateral	Hip Abduction	Rate of moment rise	1 N$\cdot$m$\cdot kg^{-1}\cdot s^{-1}$	0.77 (0.51 - 1.16)	0.70 (0.45 - 1.11)	**0.53 (0.30 - 0.94) ***	Small	0.52 (0.39 - 0.65)	< 2.34 N$\cdot$m$\cdot kg^{-1}\cdot s^{-1}$	0.92	0.24	Poor
Lateral	Hip Abduction	Moment burst duration	0.1 s	1.07 (0.98 - 1.18)	1.08 (0.98 - 1.19)	**1.12 (1.00 - 1.24) ***	Very small	0.59 (0.46 - 0.73)	> 396 ms	0.81	0.39	Poor
Lateral	Hip Abduction	Peak angle	5 °	0.58 (0.25 - 1.34)	0.52 (0.21 - 1.32)	**0.32 (0.10 - 0.97) ***	Small	0.55 (0.42 - 0.69)	< 6 °	0.89	0.33	Poor
Lateral	Hip Abduction	Angle burst duration	0.1 s	**0.91 (0.84 - 0.99) ***	**0.9 (0.83 - 0.99) ***	**0.91 (0.83 - 0.99) ***	Very small	0.62 (0.49 - 0.75)	< 1274 ms	0.92	0.39	Poor
Lateral	Knee Sagittal Power Absorption	Burst duration	0.1 s	**1.38 (1.00 - 1.90) ***	1.37 (0.99 - 1.90)	**1.58 (1.10 - 2.28) ***	Very small	**0.65 (0.51 - 0.78) ***	> 374 ms	0.46	0.85	Poor
Lateral	Knee Sagittal Power Generation	Time to peak	0.1 s	**1.22 (1.01 - 1.47) ***	**1.23 (1.01 - 1.50) ***	**1.30 (1.04 - 1.63) ***	Very small	**0.65 (0.52 - 0.79) ***	> 837 ms	0.77	0.57	Poor
Lateral	Knee Extension	Moment onset latency	0.1 s	**1.17 (1.04 - 1.33) ***	**1.19 (1.04 - 1.36) ***	**1.23 (1.07 - 1.42) ***	Very small	**0.69 (0.56 - 0.83) ***	> 550 ms	0.58	0.83	Poor
Lateral	Knee Extension	Time to peak moment	0.1 s	1.10 (0.98 - 1.23)	1.11 (0.98 - 1.24)	**1.14 (1.00 - 1.29) ***	Very small	**0.64 (0.50 - 0.77) ***	> 1204 ms	0.58	0.76	Poor
Lateral	Knee Extension	Peak moment	0.1 N$\cdot$m$\cdot kg^{-1}$	0.84 (0.62 - 1.14)	0.82 (0.59 - 1.12)	**0.64 (0.44 - 0.95) ***	Very small	0.61 (0.47 - 0.75)	< 0.21 N$\cdot$m$\cdot kg^{-1}$	0.69	0.61	Poor
Lateral	Knee Extension	Angle onset latency	0.1 s	**1.12 (1.02 - 1.22) ***	**1.11 (1.01 - 1.22) ***	**1.16 (1.05 - 1.29) ***	Very small	0.64 (0.50 - 0.78)	> 657 mm	0.54	0.85	Poor
Lateral	Knee Extension	Time to peak angle	0.1 s	**1.09 (1.00 - 1.19) ***	1.09 (1.00 - 1.18)	**1.13 (1.02 - 1.25) ***	Very small	0.63 (0.49 - 0.77)	> 1234 ms	0.46	0.85	Poor

Data are from all subjects (n = 72). Model 1: Unadjusted logistic regression model. Model 2: Adjusted for age, sex, and BMI. Model 3: Model 2 + fall history status, Mini-BESTest score, short FES-I score, and PASE-C score. OR: odds ratio; CI: confidence interval. ROC: receiver operating characteristic; AUC: area under the ROC curve. * indicates that the logistic regression model was significant (two-sided, $p < 0.05$) or the AUC value was significantly different from 0.5 (two-sided, $p < 0.05$). CoM: center of mass. EMG: electromyographic.

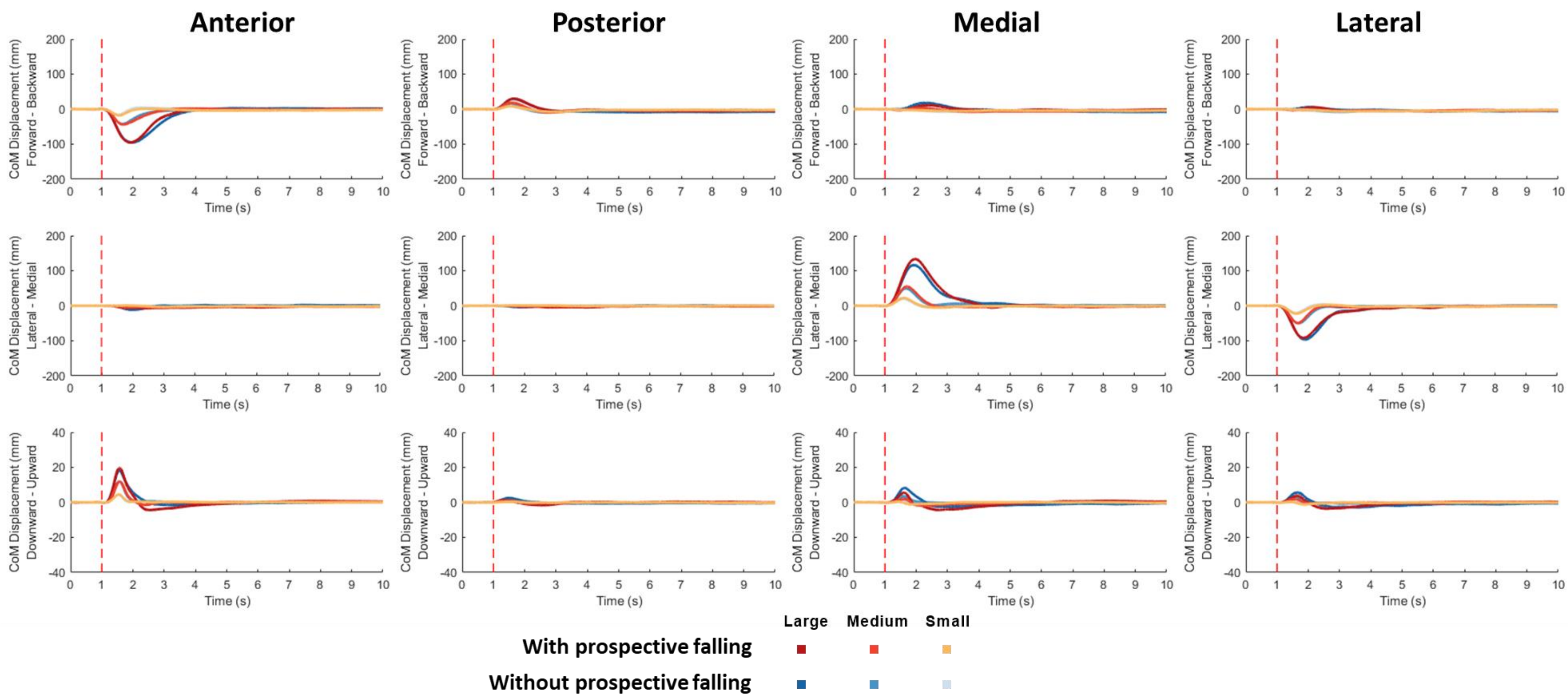


Figure 6-3. The mean CoM displacements in older adults with (n = 26) and without (n = 46) prospective falling after each direction and magnitude of perturbations. The start of balance perturbation is denoted by t = 1s. CoM: center of mass.

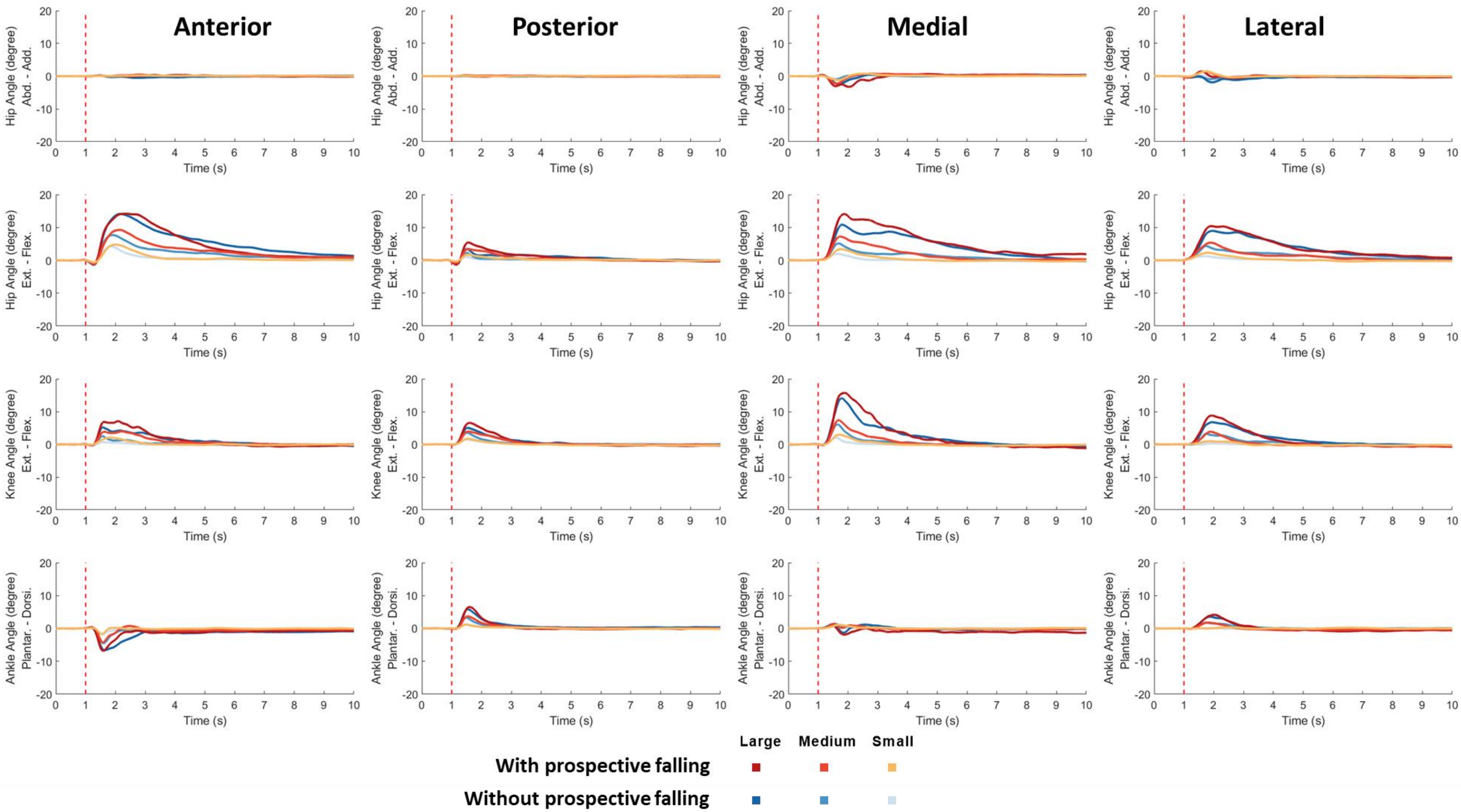


Figure 6-4. The mean lower-limb joint motions in older adults with (n = 26) and without (n = 46) prospective falling after each direction and magnitude of perturbations. The start of balance perturbation is denoted by t = 1s. Add.: adduction. Abd.: abduction. Flex.: flexion. Ext.: extension. Dorsi.: dorsiflexion. Plantar.: plantarflexion.

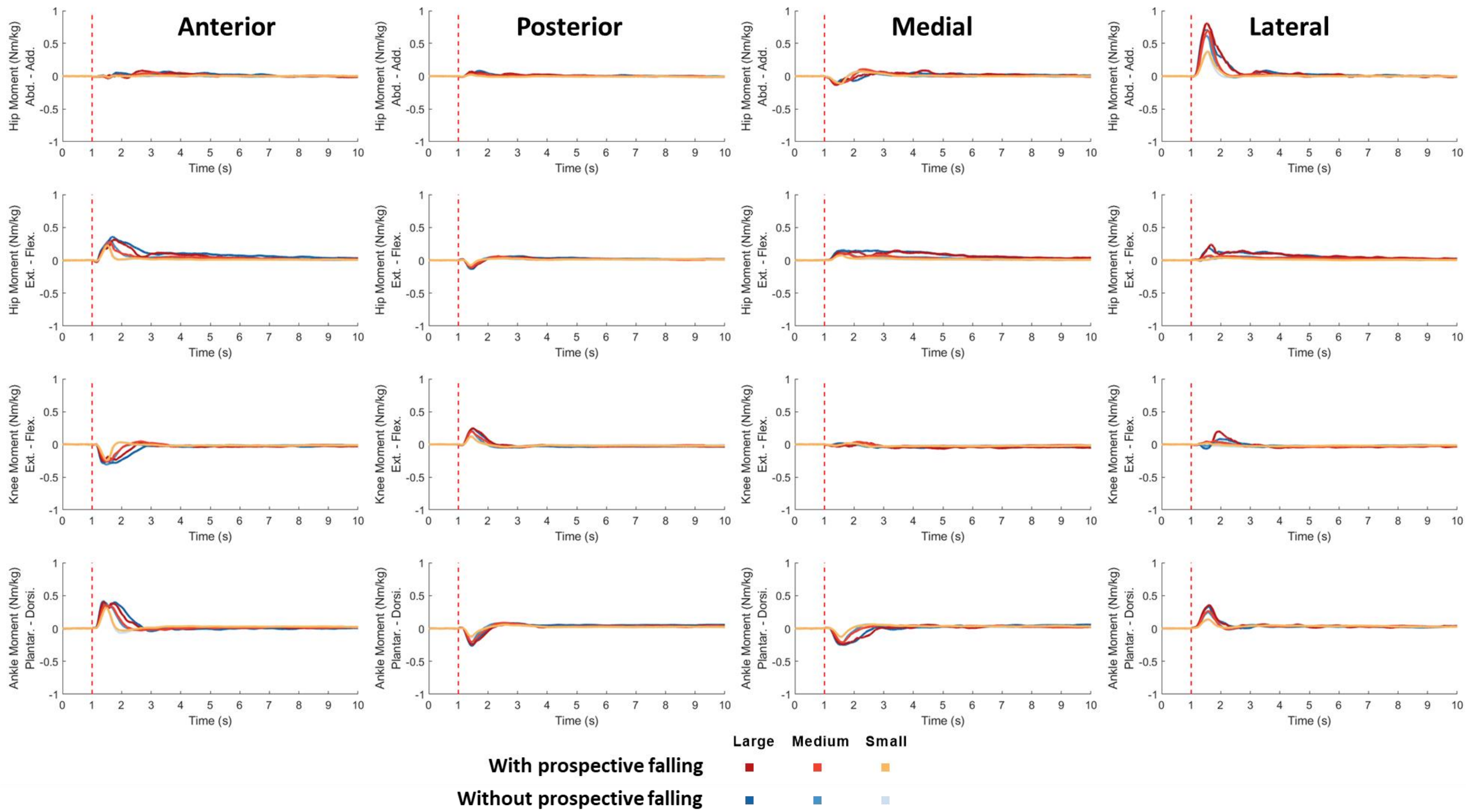


Figure 6-5. The mean lower-limb joint moments in older adults with (n = 26) and without (n = 46) prospective falling after each direction and magnitude of perturbations. The start of balance perturbation is denoted by t = 1s. Add.: adduction. Abd.: abduction. Flex.: flexion. Ext.: extension. Dorsi.: dorsiflexion. Plantar.: plantarflexion.

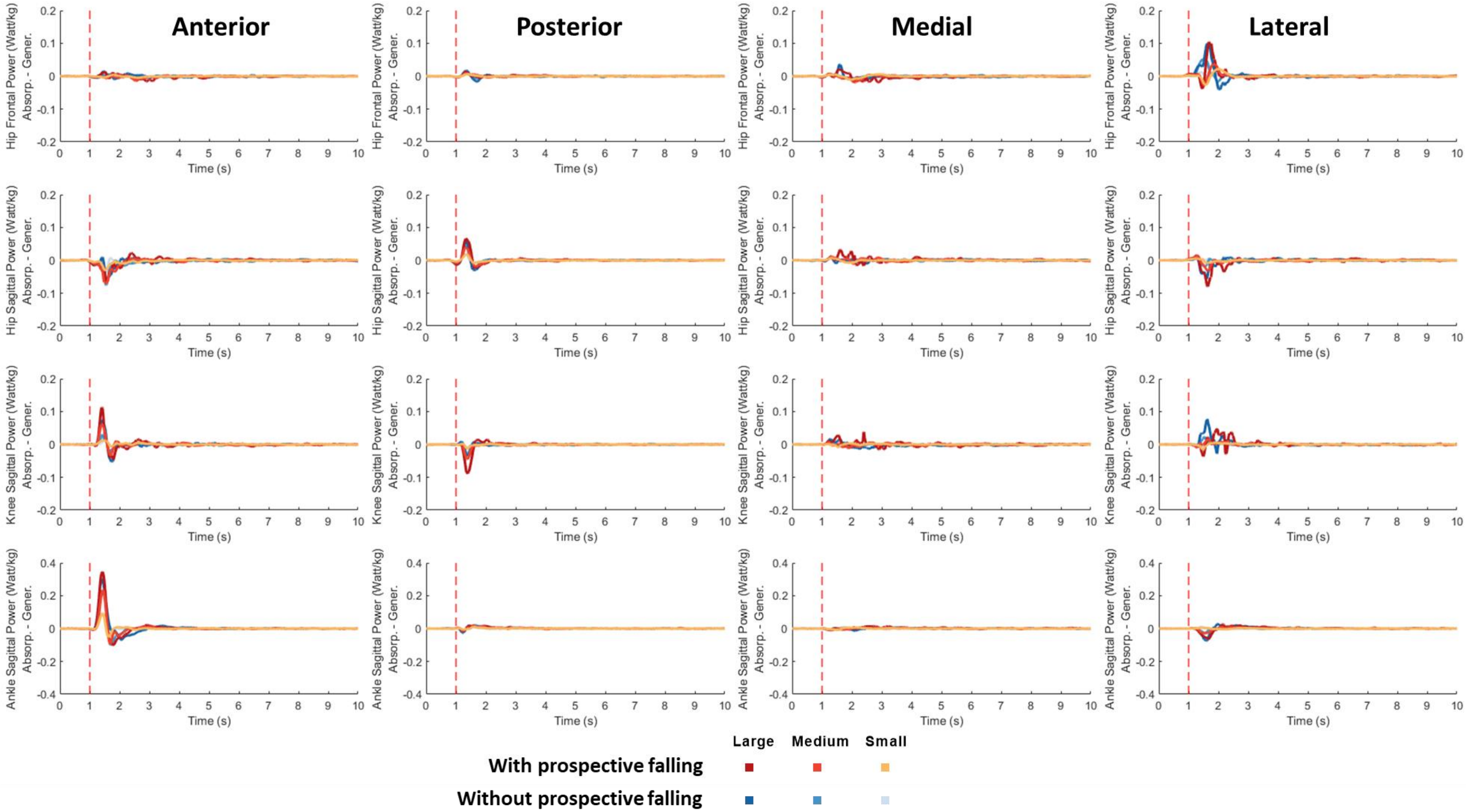


Figure 6-6. The mean lower-limb joint powers in older adults with (n = 26) and without (n = 46) prospective falling after each direction and magnitude of perturbations. The start of balance perturbation is denoted by t = 1s. The red dotted line denotes the start of balance perturbation. Absorp.: absorption. Gener.: generation.

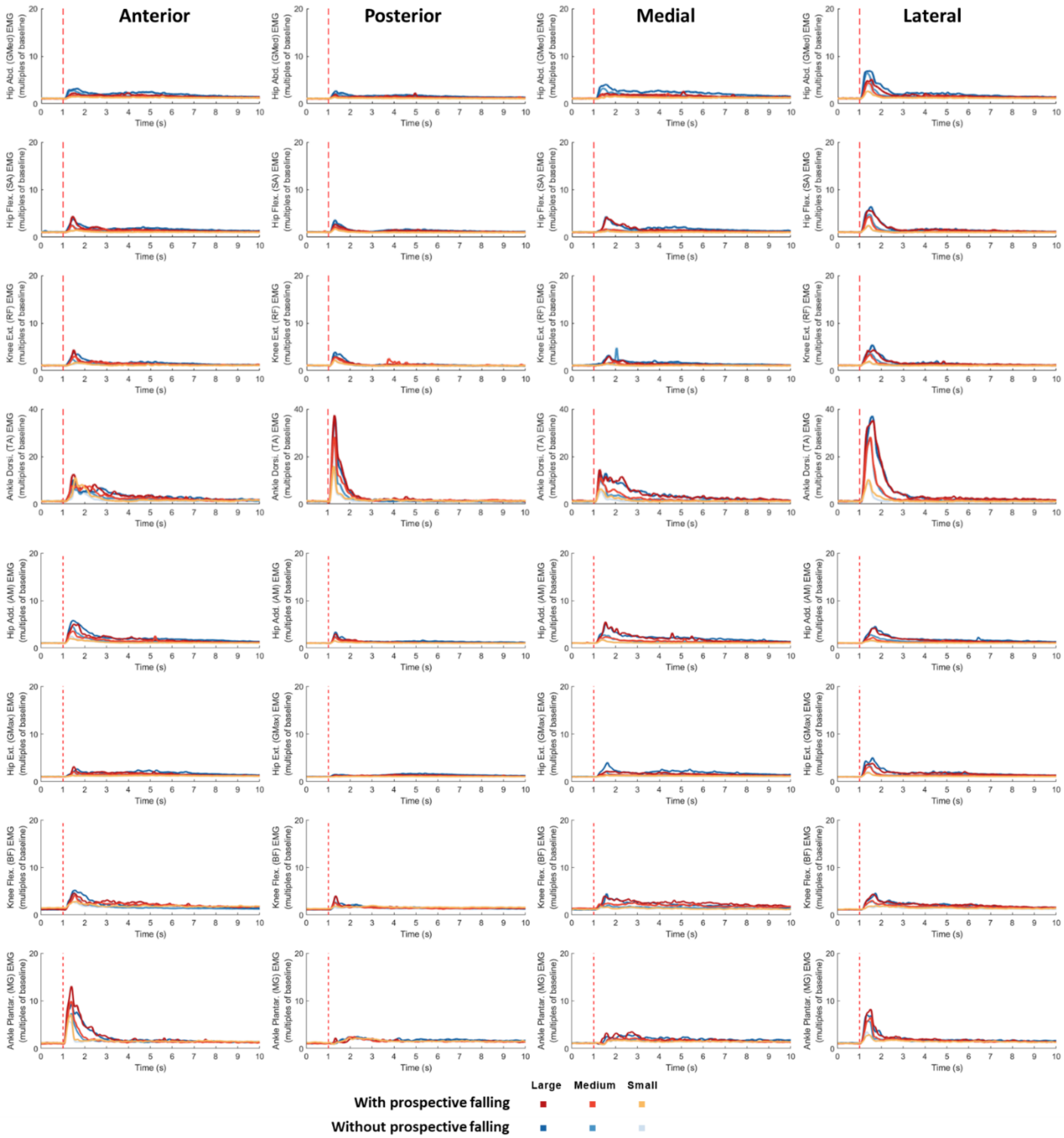


Figure 6-7. The mean EMG signals for eight lower-limb muscles in older adults with (n = 26) and without (n = 46) prospective falling after each direction and magnitude of perturbations.

The start of balance perturbation is denoted by t = 1s. EMG: electromyographic. GMed: gluteus medius. SA: sartorius. RF: rectus femoris. TA: tibialis anterior. AM: adductor magnus. GMax: gluteus maximus. BF: biceps femoris. MG: medial gastrocnemius. Add.: adductor. Abd.: abductor. Flex.: flexor. Ext.: extensor. Dorsi.: dorsiflexor. Plantar.: plantarflexor.

6.6 Discussions

The objective of this study was to delve into alterations during reactive balance control that could predict fall risks or differentiate fall histories among the community-dwelling older people. This study innovatively elucidates the causes of impaired reactive balance control for falls, by controlling relevant confounding factors of falls and by comprehensively analyzing the lower-limb muscle activities as well as joint powers/moments/motions following unexpected waist-pull perturbations. Partly aligning with our hypotheses, the older adults with fall histories have utilized increased activation of ankle muscles to compensate for the insufficient activation of hip abductor and hip extensor following sudden balance loss, which seemed not to be an effective strategy as the subsequently enlarged lower-limb joint moments/powers/motions and postural sways have been required in contrast to non-fallers. However, many of the fall-related factors identified in the cross-sectional analysis could not predict the older adults' prospective falling. The older adults' higher risk of falls has been predicted by the hip abductor's insufficient activation, especially the reduced activation rate, following sudden balance losses in both sagittal and frontal planes. The more absorbed hip/knee joint powers and insufficient hip/knee joint motions in response to sudden balance loss, especially mediolateral balance loss, have also indicated the older adults' fall risks. These findings not only revealed the mechanisms of fall-prone people's ineffective balance-control strategy but also detected the more rooted alterations during reactive balance control that could indicate older adults' high fall risks. The identified fall-risk factors and their cut-point values in this study could further facilitate a more sensitive fall-risk assessment and help detect the older adults with high fall risks earlier. Interventions targeting these identified fall-risk factors may also lead to better effects on enhancing reactive balance control and reducing falls in fall-prone people.

6.6.1 Insufficient Activation of Hip Abductor in Reactive Balance Control Could Predict Fall Risks

The primary finding of this study was that the slowed and reduced activation of hip abductor following unexpected waist-pull perturbations could predict the older adults' prospective falling. Among the eight investigated hip, knee, and ankle muscles, only the activation of hip abductor has been observed with the ability to indicate older adults' fall risks. As far as the authors know, this is the first study delineating such causal relationship.

Older adults' higher risk of falls could be predicted by the smaller activation rate of hip abductor in response to sudden loss of balance. The important role of hip abductor's fast activation for maintaining reactive balance was previously reported (Inacio et al., 2019). In contrast to young adults, older adults were found to have a smaller rate of EMG rise of hip abductor following unexpected lateral waist-pull perturbations (Inacio et al., 2019). Such alteration seemed to have explained why older adults used the loaded-leg stepping strategies less, which were more common in young adults and considered as the more effective stepping responses to avoid a real fall (Inacio et al., 2019; Mille et al., 2005). Our study has further provided the direct evidence of the hip abductor's activation rate following unexpected perturbations as a fall-risk factor independent of age, sex, etc. In addition, that fall risks could be predicted by the reduced activation rate of hip abductor has been true for not only the mediolateral perturbations but also the anteroposterior perturbations. Previous studies once reported how the values of hip abductor's activation rates (unit: multiples of baseline EMG amplitude) varied when suddenly losing balance in different directions (Tong et al., 2023; Zhu et al., 2024; Zhu et al., 2022). Larger activation rate was needed for resisting lateral balance loss, followed by medial or backward balance loss, and followed by forward balance loss (Tong et al., 2023; Zhu et al., 2024; Zhu et al., 2022). In this study, the cut-point values of hip abductor's activation rates following the four horizontal directions of perturbations that could indicate older adults' fall risks have exhibited the similar trend (**Table 6-5**. Anterior: <7 baseline/s; Posterior: <14 baseline/s; Medial: <12 baseline/s; Lateral: <44 baseline/s).

The delayed activation, reduced peak activation, and reduced activation duration of hip abductor following unexpected waist-pull perturbations could also predict older adults' fall risks. On the one hand, apart from the smaller activation rate, the higher odds of prospective falling have been directly indicated by the hip abductor's earlier EMG onset following lateral perturbations, larger peak EMG amplitudes following anteroposterior and medial perturbations, as well as shorter EMG burst duration following anterior perturbations. On the other hand, the insufficient activation of hip abductor during reactive balance control seemed to be accompanied by or be underlying the lower-limb kinetic and kinematic alterations that could also indicate fall risks. In this study, the causal relationships between the neuromuscular/biomechanical alterations during reactive balance control and fall risks have shown generally very small effect sizes. Nevertheless, it is worth noting that the insufficient hip abductor's activation, and the subsequently reduced rate of hip abduction moment rise as well as reduced peak hip abduction angle have been identified with larger effect sizes in predicting the odds of prospective falling and/or better abilities in classifying the prospective fall status (**Table 6-5**), highlighting the importance of assessing hip abductor's activation in reactive balance control assessment to enhance fall-risk detection among the older adults.

6.6.2 Altered Responses of the Proximal Lower-limb Joints in Reactive Balance Control Were More Indicative of Fall Risks

The secondary finding of this study was that older adults' prospective falling could be mostly predicted by their alterations in the hip and knee joints but not in the ankle joint following unexpected waist-pull perturbations. To the best of knowledge, causal relationships between the lower-limb joint kinematics/kinetics following sudden balance losses and the falls were reported little in prior studies. This finding has partially substantiated the previous reports on the age-related alterations of lower-limb responses following perturbations, where older adults were found to exhibit prominently hip or suspensory strategies rather than ankle strategies following unexpected perturbations in contrast to young adults (Hall & Jensen, 2002; Osoba et al., 2019). On top of them, this study has found that older people's higher risk of falls has been directly indicated by the more hip/knee joint power absorption as well as the insufficient hip/knee joint motion during reactive balance control.

(1) Increased Hip and Knee Power Absorption in Reactive Balance Control Indicated Older Adults' Fall Risks

The more hip/knee joint power absorption following unexpected perturbations could predict older adults' prospective falling, whereas the alteration of joint power generation that could indicate fall risks varied across different directions of sudden balance loss (**Table 6-5**). Specifically, the older adults' higher odds of falling have been predicted by the earlier onset and larger peak value of knee sagittal power absorption and earlier peak timing of hip frontal power absorption following posterior perturbations, as well as the earlier peak timing of hip frontal power absorption and longer duration of knee sagittal power absorption following lateral perturbations. These alterations (larger peak value and elongated duration of power absorption) mean that more energy was absorbed in hip and knee joints during reactive balance control for the older people with high fall risks, given that energy equals power multiplying time. Unlike power absorption, the alterations of power generation indicating fall risks have been related to the perturbation directions. The older adults at higher risk of falls had less power generation following posterior and lateral perturbations (i.e., delayed timing of hip frontal power generation and knee sagittal power generation) but had more power generation following medial perturbations (i.e., earlier timing of hip sagittal power generation, larger peak value of knee sagittal power generation). With the more absorbed hip/knee joint power and less hip/knee power generation, older adults might be unable to have prompt reactions that could resist sudden loss of balance and therefore have increased fall risks.

For example, such alterations of powers following posterior perturbations could restrict the quick and adequate generation of knee flexion for a suspensory strategy, which has been reported as an effective balance-control strategy by lowering the CoM height to improve stability (Zhu et al., 2024). By contrast, following medial perturbations, the increased hip/knee sagittal power generation in older adults with high fall risks seemed to be a way to compensate for their insufficient resistance to sudden balance loss due to the hip abductor's activation of unloaded leg, which might have also explained their further responses of joint moments and motions as well as their higher ratio/frequency of unloaded-leg stepping responses (**Table 6-2**).

(2) Insufficient Hip and Knee Joint Motions in Reactive Balance Control Indicated Older Adults' Fall Risks

The delayed and reduced hip/knee joint motions following unexpected perturbations could predict older adults' prospective falling (**Table 6-5**). Specifically, the older adults' higher odds of falling have been predicted by the delayed timing of hip flexion angle following anterior perturbations together with the reduced duration and peak value of hip abduction angle and delayed timing of knee extension angle following lateral perturbations. The alterations in lower-limb joint motions could be the basis of fall-related alterations in stepping responses. For example, following anterior perturbations, the insufficient activation of hip abductor together with the delayed hip flexion might have indicated that the hip strategy in older people with high fall risks could be not enough to maintain feet in place, which might have further explained their significantly higher stepping frequency compared with older people without prospective falling (**Table 6-2**). This has echoed with previous observation that older people with fall histories, i.e., older fallers, had more stepping responses in reactive balance control than older non-fallers (Bair et al., 2016).

(3) Joint Moments Following Different Directions of Unexpected Perturbations Did Not Show Consistent Alterations that Could Indicate Older Adults' Fall Risks

Underlying these alterations of joint motions, the hip/knee joint moments did not show a consistent alteration that could indicate fall risks (**Table 6-5**). Specifically, following medial perturbations, the older adults' higher odds of falling have been predicted by the reduced duration of hip flexion moment and delayed timing of knee flexion moment together with the larger rate of rise and peak value of hip extension moment, which might be related to their higher frequency of unloaded-leg backward stepping responses. Following lateral perturbations, the higher risk of falls has been indicated by the smaller rate of rise and elongated duration of hip abduction moment which might be due to hip abductor's insufficient activation, more absorbed hip frontal power, and delayed hip frontal power generation of the loaded leg. This finding has partially agreed with a previous study, where older adults exhibited a smaller peak value of generated hip abduction power and a smaller peak hip abduction moment than young adults for a loaded-leg step following unexpected lateral waist-pull perturbations (Inacio et al., 2019). On top of the age-related declines of hip joint moment and power in response to lateral loss of balance, our study has further identified the casual relationship between the delayed timing of hip frontal power generation (or earlier timing of hip frontal power absorption) and older adults' fall risks. In addition, in this study, the higher risk of falls has been found to be indicated by the insufficient knee extension moment (delayed timing and smaller peak value) which might be due to the more absorbed knee sagittal power and delayed timing of knee sagittal power generation in the loaded leg.

Generally, these alterations following unexpected mediolateral perturbations have echoed the previous studies reporting stepping characteristics (Bair et al., 2016; Mille et al., 2005). The older adults with fall histories had more unloaded-leg stepping responses and fewer loaded-leg stepping responses in contrast to those without fall histories (Bair et al., 2016). Such alteration in stepping response was also seen in the older adults in contrast to young adults (Mille et al., 2005). Our study has provided the direct evidence indicating fall risks and partially corroborated the previous findings. In this study, the similar alterations of loaded-leg/unloaded-leg stepping responses have been observed following mediolateral perturbations in older adults with prospective falling in contrast to those with prospective falling, but no significant differences existed between the two groups (**Table 6-2**); nevertheless, a series of more in-depth kinetic/kinematic alterations underlying the insufficient loaded-leg responses or exaggerated unloaded-leg responses have been observed to be able predict the older people's fall risks in this study. This may imply that for some community-dwelling older people who have not exhibited the stepping impairment yet, their fall risks could be potentially identified earlier and more sensitively through the measurements of hip/knee joint responses in reactive balance control.

6.6.3 Altered Responses to the Mediolateral Perturbations Could Predict Fall Risks Better

The third finding of this study was that there were more alterations of lower limb following unexpected mediolateral perturbations that have shown better abilities in predicting older adults' fall risks compared with following unexpected anteroposterior perturbations. On top of a prior study reporting that older adults with prospective falling had a higher frequency of multiple steps following mediolateral waist-pull perturbations but not following anteroposterior waist-pull perturbations in contrast to older adults older without prospective falling (Mille et al., 2013), this study has further provided neuromuscular/biomechanical evidence explaining the underlying reason, i.e., the insufficient hip abductor's activation as well as insufficient hip abduction mattered more to maintain mediolateral balance.

More number of lower-limb alterations during sudden mediolateral balance loss have been observed to significantly indicate older adults' fall risks, in contrast to during sudden anteroposterior balance loss (**Table 6-5**). Regarding the abilities of anteroposterior and mediolateral postural sways in differentiating older adults' fall risks, the recent systematic review summarized that anteroposterior center-of-pressure (CoP) features in quiet standing were more discriminatory than mediolateral CoP features. However, our study seemed to have identified the opposite trend regarding the fall-differentiating abilities of reactive balance control, given that this study has observed more fall-related alterations in lower-limb joint kinetics/kinematics following the mediolateral perturbations than following the anteroposterior perturbations. More importantly, in contrast to following anteroposterior perturbations, the hip abductor's activation rate following mediolateral perturbations has shown better ability in classifying older adults' fall risks (**Table 6-5**. Anterior: poor. Posterior: poor. Medial: acceptable. Lateral: acceptable). The mechanism could be that hip abductor's rapid activation has played a more important role in resisting sudden mediolateral balance loss than in resisting sudden anteroposterior balance loss, as suggested by the values of hip abductor's activation rates following the four directions of perturbations in present and previous studies.(Tong et al., 2023; Zhu et al., 2024; Zhu et al., 2022) In addition, despite the "very small" effect sizes of most parameters in fall prediction, the delayed hip abductor's activation and insufficient hip abduction moment/angle following lateral perturbations have shown larger effect sizes (i.e., "small") in predicting older adults' fall risks. The mechanism could be that these alterations have restricted the subsequent initiation of effective loaded-leg stepping strategies to avoid a real fall after a sudden lateral balance loss.(Bair et al., 2016; Inacio et al., 2019; Mille et al., 2005)

6.6.4 Responses that Could Indicate Fall Histories May Not Indicate Fall Risks

The fourth finding of this study was that the identified alterations following sudden balance loss in older fallers (i.e., older adults with fall histories) by comparing with non-fallers may not necessarily the risk factors of falls. On top of the prior preliminary study that examined the neuromuscular/biomechanical responses following unexpected perturbations in a limited size of fallers and non-fallers (Zhu et al., 2024), this study may have provided more valid findings regarding the fall-history indicators given its justified and larger sample size.

This study has found that fallers exhibited a quicker neuromuscular response following anterior perturbations but slower neuromuscular responses following posterior/medial/lateral perturbations as compared to non-fallers. This is consistent with the finding of our narrative review, as fallers were found to have quicker forward stepping response but slower backward/medial/lateral stepping responses (**Chapter 2**). Previous studies observed the delayed or slowed reactions of lower-limb muscles in fallers, e.g., the delayed EMG onset in ankle dorsiflexor following a suddenly backward balance loss induced by a moving platform (Studenski & Chandler, 1991), the delayed timing to peak EMG amplitude in ankle plantarflexor following a suddenly forward balance loss induced by a tether-release test (Ochi et al., 2014), the delayed EMG onset in loaded-leg hip abductor and knee flexor following a suddenly lateral balance loss induced by a shoulder-impact perturbation (Claudino et al., 2017). Nevertheless, the alterations in the specific ankle, knee, or hip muscles have varied across studies, which may be mainly due to the different paradigms used for inducing unexpected perturbations in different studies (Claudino et al., 2017; Ochi et al., 2014; Studenski & Chandler, 1991; Zhu et al., 2024).

The reason was that some fallers did not fall in the follow-up of prospective falling, whereas some non-fallers were found to have experienced prospective falling in this study. In addition, the collection of fall history is more susceptible to recall bias than prospectively tracking falls. These have indicated that the detected fall-related alterations following sudden balance losses by comparing older fallers and older non-fallers could not fully delineate the causal relationships between reactive balance control performance and falls, given that the inquiry of fall history could be susceptible to recall bias and there was a small proportion of recurrent fallers in the faller group for this study (Deandrea et al., 2010; Nastasi et al., 2018).

6.6.5 Implications for Clinical Fields

The findings of this study could facilitate a more sensitive fall-risk assessment in clinical settings and give insights on the prescription of more targeted training/exercises to prevent community-dwelling older people's falls. While clinical tests are mainly based on a client's number of steps to evaluate the performance of reactive balance control, they may have shown ceiling effects among the community-dwelling older adults without significant diseases. This study has found that the older adults' fall risks could be predicted by the hip abductor's slowed and reduced activation following sudden balance loss. Particularly, the hip abductor's activation following mediolateral perturbations has shown acceptable abilities in classifying the older adults' fall risks. The cut-point values in this study could also provide quantitative references for the therapists to interpret the outcomes of reactive balance control. The monitoring of hip abductor's activation in induced balance loss, especially in the mediolateral direction, may therefore help therapists have a more sensitive evaluation of the older adult's fall risks. In addition, the results of this study suggest that exercises or rehabilitation training could focus on the proximal leg muscles and joints, especially the hip abductor and hip abduction, to enhance the fall-prevention effects in older adults, given that one recent study has reported the promising effects of two exercises that targeted hip abductor (i.e., laterally induced stepping training, hip abductor strengthening) on fall prevention among community-dwelling older people (Rogers et al., 2021). Moreover, the identified neuromuscular/kinematic/kinetic parameters in this study could serve as the multimodal dataset for the design of robotic assistive devices (e.g., the powered lower-limb exoskeleton) that can provide support for effective reactive balance control in fall-prone older population (Moreira et al., 2021; Vlutters et al., 2018).

6.6.6 Strengths and Limitations

This is the first study that has investigated neuromuscular alterations and joint kinematics/kinetics during reactive balance control that could predict fall risks in older adults. The reactive balance control has been examined extensively through the investigation of eight major leg muscles and joint powers/moments/motions, and intensively through the analyses of both timing and amplitude parameters for each signal. The prospective falling has been tracked within 1 year through the monthly calendars to minimize the older participants' recall bias. The potential factors that could confound the causal relationships between alterations of reactive balance control and falls have been controlled. The diagnostic accuracies of the identified neuromuscular/biomechanical alterations in classifying fall risks have been examined. With these valid approaches, this study can provide new evidence to enable a more sensitive fall-risk assessment and an earlier identification of the community-dwelling people who are prone to falls. New insights can also be provided by this study regarding the more targeted interventions for fall prevention.

There are two limitations in this study. Firstly, this study did not focus on the older people with a history of recurrent falls or the older people with prospective multiple falls. Given that recurrent fallers had higher odds of future falling than fallers and could be the more typical representation of fall-prone people (Deandrea et al., 2010), some previous studies have compared recurrent fallers and no-fallers in the cross-sectional analysis (Li et al., 2023) or used the multiple falls within the prospective follow-up period as the outcome (Hirase et al., 2020). As the objective of this study was to detect the more in-depth factors underlying reactive balance control for explaining why some older adults with relatively good health are prone to falls, both the single fallers and recurrent fallers were considered suitable for the faller group. Additionally, partly because of the relatively healthy cohort of older people recruited in this study, only a small number of older adults with prospective multiple falls have been observed and thus not suitable as an additional group in the regression analysis (Riley et al., 2019). The identified fall-related factors in this study may therefore have not fully revealed the alterations of reactive balance control in those older adults with higher fall risks. Secondly, this study did not perform a maximal voluntary contraction (MVC) test to determine the maximum EMG amplitude for each of the investigated eight leg muscles. Instead, the baseline EMG signal value during unperturbed standing has been used for EMG amplitude normalization. When interpreting the findings of EMG amplitude parameters in this study, it is therefore worth noting that they were reflecting the extent of muscle activation required for the perturbation task relative to normal standing. Thirdly, it should be noted that the reactive balance involves not only motor output but also the sensory input (**Figure 1-2**). The axons innervating the human arm, sensory axons outnumber motor axons in a ratio of 9:1 (Gesslbauer et al., 2017). Similar work has not been done for the lower extremities but one may speculate that a similar ratio exists for the lower extremities, emphasizing the critical role of sensory input. Therefore, the delayed/slower neuromuscular activation could be attributed to not only the number or firing rate of motor units but also the inputs of proprioceptive/vestibular/visual sensations. Regarding the role of sensory input, this study excluded people with a vestibular disease or significant visual impairment. Based on the score of sensory orientation (including the test disturbing proprioception), a subcategory of the Mini-BESTest, no significant difference was found between fallers and non-fallers or between prospective fallers and prospective non-fallers. Nevertheless, future study may consider directly measuring the proprioception of hip, knee, or ankle joint.

6.7 Conclusion

This study has identified a series of neuromuscular/biomechanical alterations that could indicate fall risks/histories in older adults. The altered responses in proximal lower-limb joints/muscles during reactive balance control, especially the insufficient hip abductor’s activation and the alterations following sudden mediolateral balance loss, can predict fall risks in older adults, although these parameters have just shown poor to acceptable abilities in classifying fall risks.

However, most of the fall-related factors identified by the comparison of older people with and without fall histories have not been the risk factors of falls. These findings highlight the importance of measuring the hip abductor's activity during reactive balance control assessment for a more sensitive and earlier detection of the fall-prone older adults. It is also implied that the exercises or robotic assistive devices targeting hip and knee joints could enhance the fall-prevention effects in older adults.

Chapter 7 Discussions and Implications

7.1 Chapter Summary

Overall, this PhD project has conducted four relevant studies step by step to realize the identification of in-depth neuromuscular/biomechanical mechanisms underlying reactive balance control that can indicate older adults' fall risks. This chapter firstly discusses the link of the four studies conducted in this PhD project. Subsequently, this chapter compares the findings of the four studies and delves into their similarities and differences. Lastly, this chapter discusses the implications for clinical practice and outlines future research directions.

The designs of the four observational studies have been based on the findings of the two conducted literature reviews in this PhD project. The systematic review and meta-analyses on lower-limb rapid strength has suggested that the decline in entire lower-limb power can indicate community-dwelling older people's fall histories/risks, while the rapid strength of a single muscle group shows insufficient distinguishing ability due to the small number of relevant studies (Zhu, Zuo, et al., 2025). In addition, the narrative review regarding the response speed of reactive balance control has suggested that the altered stepping characteristics and delayed whole-body postural sways may indicate community-dwelling older adults' fall histories/risks, while the evidence of neuromuscular responses was insufficient and the evidence of joint kinetics or kinematics was lacking. Given these limitations in prior studies, the four studies have been designed in this PhD project.

Among the four studies, <u>the later study has been designed and optimized based on the outcomes of the preceding study</u>. This has been reflected on how we dealt with the crosstalk issue of EMG and the processing of MMG signal. Regarding the EMG crosstalk, while the iliopsoas was examined as the hip flexor in the pilot study of waist-pull experiments, the sartorius was chosen in subsequent studies due to its more superficial and distinguishable location in contrast to the deeper iliopsoas. Similarly, the semitendinosus served as the knee flexor in pilot studies of young adults; however, in later studies, the long head of the biceps femoris was selected as the knee flexor to minimize EMG signal crosstalk, as it was anatomically distant from the investigated hip adductor. Regarding the MMG signal processing, the pilot studies of waist-pull and moving-platform experiments have tried different filtering methods to isolate the muscle vibration signals from the accelerometry data that were collected in human's dynamic situations. However, neither method of MMG signal processing yielded sensible results, as the time lag between EMG onset and MMG onset did not align with findings from prior studies. Given the immature techniques of

MMG signal processing in human's dynamic situations, the later studies in older adults did not have further analysis of MMG signals during reactive balance control.

As the main aim of this PhD project was to investigate the in-depth mechanisms of reactive balance control in fall-prone older people, previous chapters have separately discussed how the identified neuromuscular/biomechanical alterations could indicate the fall histories and the fall risks in older adults. The main findings were: older adults' prospective fall risks could be predicted by a series of neuromuscular/kinematic/kinetic alterations following unexpected waist-pull perturbations with the very small to medium effect sizes, including the insufficient activation of hip abductor, the more absorbed hip/knee joint power, the insufficient hip/knee joint motions etc.

Beyond discussions on how the reactive balance control strategies were associated with fall histories or fall risks, how the reactive balance control strategies were associated with other factors are discussed below.

7.2 Parameters Indicating Fall Histories vs. Fall Risks

In this PhD project, study 4 not only retrospectively compared the older fallers and older non-fallers but also prospectively examined what factors underlying reactive balance control could predict fall risks in these older participants. As far as the authors know, no prior studies have concurrently reported how the reactive balance control strategies would indicate fall histories and fall risks in a same cohort of older people. This study has found that the altered responses in reactive balance control observed when comparing older fallers to older non-fallers did not demonstrate the ability to predict future fall risks. This suggests the fall-history-based comparisons may not fully capture the causal mechanisms linking reactive balance control to actual fall risk. Since the inquiry of fall history could be susceptible to recall bias and most fallers had a history of only one fall in study 4, the differences of reactive balance control detected in older fallers vs. older non-fallers may not completely reflect the deficits of reactive balance control in older adults with high fall risks (Deandrea et al., 2010; Nastasi et al., 2018).

7.3 Waist-pull vs. Translational Moving-platform Perturbations

Although comparing the reactive balance control strategies following waist-pull perturbations and following translational moving-platform perturbations was not the main aim of this PhD project, some interesting findings have been observed.

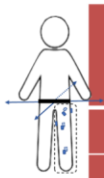
The key difference for the two perturbation methods is that they induced the subject's sudden balance loss in different ways and simulated different scenarios. On the one hand, the two perturbations were delivered to the subject's different body parts (pelvis vs. under foot). This caused that the same direction of waist-pull perturbation and translational moving-platform perturbation would immediately induce the subject's sudden balance loss towards opposite directions. For example, the anterior waist-pull perturbations would firstly induce the subject's forward postural sways, while the anterior translational moving-platform perturbations would firstly induce the subject's backward postural sways. Therefore, it is more sensible to compare the two types of perturbations that induced the sudden balance losses towards the same direction. On the other hand, the two perturbation methods may simulate different real-life scenarios that require reactive balance control. The waist-pull perturbations are like the "pull tests" performed by clinicians to assess a subject's reactive balance control (Foreman et al., 2011). In contrast, the translational moving-platform perturbations mimic the experience of standing in an accelerating or decelerating environment, such as a moving cabinet.

It is interesting to observe some similarities in responses following the two different perturbations. Firstly, regarding the responses across the eight investigated leg **muscles**, ankle muscles consistently exhibited early onset/peak timing of activation and exhibited the largest activation rate following both anteroposterior and mediolateral perturbations. This phenomenon was understandable for the translational moving-platform perturbations, since the perturbations were directly exerted under the feet. However, this phenomenon was also seen following waist-pull perturbations. These suggest that the rapid activation of ankle muscles is paramount in the primary defense against sudden balance loss, regardless of whether the perturbation is applied distally or proximally. Secondly, regarding the responses across the varied **perturbation magnitudes/intensities**, the larger magnitude/intensity of two perturbations could both evoke earlier/faster responses in lower-limb muscles but might not necessarily in lower-limb joint motions or postural sways. This implies that the quicker lower-limb muscle activation is crucial and enough for effective balance maintenance during more challenging situations. Since the parameters such as EMG onset latency, time to EMG peak amplitude, and rate of EMG rise can be modulated by perturbation magnitudes, they may potentially serve as valuable metrics in perturbation-based balance training for biofeedback and training outcome assessment (Gerards et al., 2023).

Beyond the muscle- and perturbation-level similarities, key differences were also found in the **fall-related responses** to the two types of perturbations. Shortly after the translational moving-platform perturbations, fallers tended to use the suspensory strategy (i.e., bending knees) in contrast to non-fallers. In contrast, after the waist-pull perturbations, fallers relied more on the ankle strategy in contrast to non-fallers. However, a common finding was that fallers exhibited the delayed activation of hip extensor in contrast to non-fallers, regardless of perturbation type. The similarities emerged more during the recovery phase, where fallers consistently showed the smaller co-contraction of hip/knee/ankle muscle pairs, elongated ankle dorsiflexor activation, larger hip/knee flexion angles and postural sways than non-fallers. Together, such findings suggest that while the fall-prone older adults' initial reactions to the different types of perturbations may differ, the less effective balance recovery strategies used by fall-prone older adults were similar across perturbation types.

As far as the authors know, there has been limited evidence regarding how the different perturbations methods affect balance-control strategies. One previous study compared reactive balance control strategies following different perturbation methods in the same cohort of human subjects (Verniba & Gage, 2020). The participated young adults had a higher frequency of making multiple steps following anteroposterior translational moving-platform than following anteroposterior shoulder-pull perturbations, which seemed to indicate that translational moving-platform perturbations were more challenging (Verniba & Gage, 2020). This PhD project has observed a similar phenomenon when designing study 1 and study 2. In the attempts before study 2, the large magnitude used in waist-pull perturbations was found to be too challenging for the subjects when applied to the translational moving-platform perturbations. Therefore, the medium magnitude of waist-pull perturbation was used as the highest intensity of moving-platform perturbation in this PhD project (**Figure 7-1**). Additionally, it is worth noting that the same cohort of young subjects participated in study 1 (moving-platform perturbations) and study 3 (waist-pull perturbations). Further statistical analyses are merited to compare the neuromuscular/kinematic differences following the two types of unexpected perturbations, which may provide enriched mechanisms of reactive balance control.

Pulling displacement (% body height) of each magnitude or intensity



Waist-pull Direction (balance loss direction)	Anterior (Forward)	Posterior (backward)	Medial (Medial)	Lateral (Lateral)
Small	2.00%	1.33%	2.67%	2.67%
Medium	4.00%	2.67%	5.33%	5.33%
Large	6.00%	4.00%	8.00%	8.00%

Translational Direction (balance loss direction)	Posterior (Forward)	Anterior (backward)	Lateral (Medial)	Medial (Lateral)
Lowest	1.00%	0.67%	1.33%	1.33%
Low	2.00%	1.33%	2.67%	2.67%
High	3.00%	2.00%	4.00%	4.00%
Highest	4.00%	2.67%	5.33%	5.33%

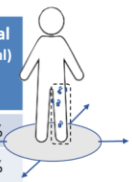


Figure 7-1. The pulling displacements of waist-pull and moving-platform perturbations.

7.4 Older Adults vs. Young Adults

Since investigating the age-related alterations of reactive balance control strategies was not the aim of this PhD project, statistical analysis was not conducted to compare older adults and young adults. Some preliminary observations in older vs young adults are as below.

Key differences were found regarding the stepping responses. Older adults were found to have the stepping strategy more frequently than young adults following the same magnitude of unexpected perturbations in this project. Additionally, in contrast to young people, older people were found to have the stepping strategy following a rather small perturbation. These alterations have aligned with the previous reports (Mille et al., 2013; Mille et al., 2005), which could be attributed to the age-related decline in limits of stability and muscle function. Prior studies also reported that older people used fewer loaded-leg stepping strategies and more unloaded-leg stepping strategies following mediolateral waist-pull perturbations in contrast to young adults (Mille et al., 2005). The unloaded-leg stepping strategy could be an ineffective strategy due to its drawbacks such as the more frequent multiple steps, prolonged single-limb support time, higher chances of limb collisions, and diminished base of support upon initial step landing (Bair et al., 2016; Inacio et al., 2019). This phenomenon has been partially proved by the observation in this PhD project, where older adults have exhibited unloaded-leg stepping strategies more frequently than young adults. Further statistical analysis may be needed to verify these preliminary comparisons.

Some similarities were found regarding the sequence of lower-limb joint/muscle responses. For both the older adults and young adults, this project has observed that ankle muscles had the largest activation rates of ankle muscles among the eight investigated muscles. The response patterns of lower-limb joint moments following unexpected waist-pull perturbations were also similar across age groups. For instance, anterior waist-pull perturbations consistently induced early responses of ankle dorsiflexion moment, knee extension moment, and hip flexion moment for both older and younger adults. For example, the anterior waist-pull perturbations consistently induced the early responses of ankle dorsiflexion moment, knee extension moment, and hip flexion moment. In addition, the difference in lower-limb joint motions was observed between the two age groups in this project. By using independent sample t tests or Mann-Whitney U tests to compare moving-platform responses in young (study 1) vs. older adults (study 2), some significant neuromuscular alterations have been found following the highest intensity of mediolateral perturbations (**Figure 7-2**). It is interesting to find that older and young people also differed in the hip abductor activation for reactive balance control, since the slower response in hip abductor activation has been also observed in prospective fallers as compared to prospective non-fallers (study 4). However, age-related significant neuromuscular differences were not observed only in the hip abductor but also in other hip/knee/ankle muscles. This indicates that the age-related differences in reactive balance were not exactly the fall-related ones. Such finding also echoes a previous study where older people were reported to have slower ankle plantarflexor activation and generally slower leg muscle activation following mediolateral moving-platform perturbations (Jeon et al., 2021). Other statistical analysis results to examine the age-related alterations in each specific lower-limb joint/muscle response will be published in additional journal articles.

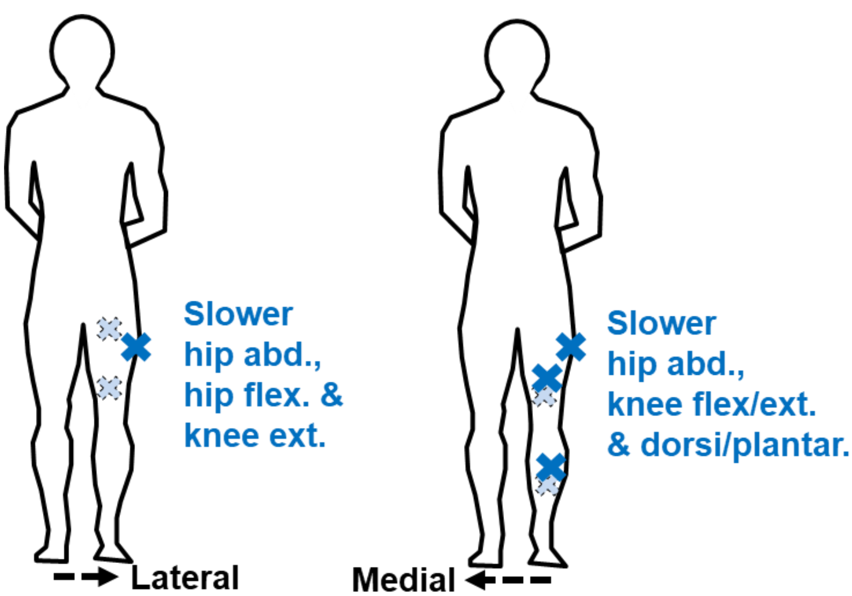


Figure 7-2. Older adults' significant electromyographic (EMG) alterations as compared to young adults following the highest mediolateral moving-platform perturbations ($p < 0.05$).

7.5 Intrinsic vs. Extrinsic Measures of Reactive Balance Control to Indicate Fall Histories/Risks

This PhD project has focused on the intrinsic measures (i.e., lower-limb muscle/joint responses) of reactive balance control and examined their diagnostic accuracies in predicting the older adults with prospective fall risks. The intrinsic measures have exhibited poor to acceptable abilities in distinguishing the older people with prospective falls from those without prospective falls (AUC ranging from 0.64 to 0.73). Among them, hip abductor's reduced activation rate (AUC = 0.73) and peak activation (AUC = 0.71) following medial waist-pull perturbations along with the hip abductor's delayed activation following lateral waist-pull perturbations (AUC = 0.71) have shown acceptable abilities in fall-risk identification. These suggest that <u>insufficient hip abductor's activation</u> can be used in assessment of reactive balance control in <u>mediolateral</u> direction to detect the older people with high fall risks. Furthermore, does this imply that only the hip abductor needs to be measured for fall-risk assessment in future clinical practice?

Take the eight leg muscles' EMG onset latencies following lateral waist-pull perturbations for example (**Figure 7-3**). Firstly, the precision of fall-predictive abilities should be considered. Although only the larger EMG onset latency of hip abductor could significantly predict the prospective falls among the eight muscles, its wide confidence interval should be noticed. The wide confidence interval indicates the imprecision of the effect estimate, namely that the future research is likely to change the effect estimate. In addition, the EMG onset latency of ankle dorsiflexor also exhibited a wide confidence interval of effect estimate, implying a future study with a larger sample size may possibly observe its significant fall-predictive abilities. Secondly, the characteristics of the cohort of older adults recruited in this study should be noted. The eligibility criteria for participating in the waist-pull experiments have excluded the community-dwelling older people with significant diseases or with poor physical function. The recruited older participants were therefore those with relatively good health. Additionally, this cohort of older people was not that old (mean age < 70 years old). The results obtained may not be generalized to the frail older people or those living in nursing homes. In other words, the cohort of older participants may not be a representative sample of the generally older population. Thirdly, based on Hill's criteria for causality, future research is still merited to examine whether the slower activation of hip abductor can consistently predict older people's prospective falls with a strong association (Schünemann et al., 2011).

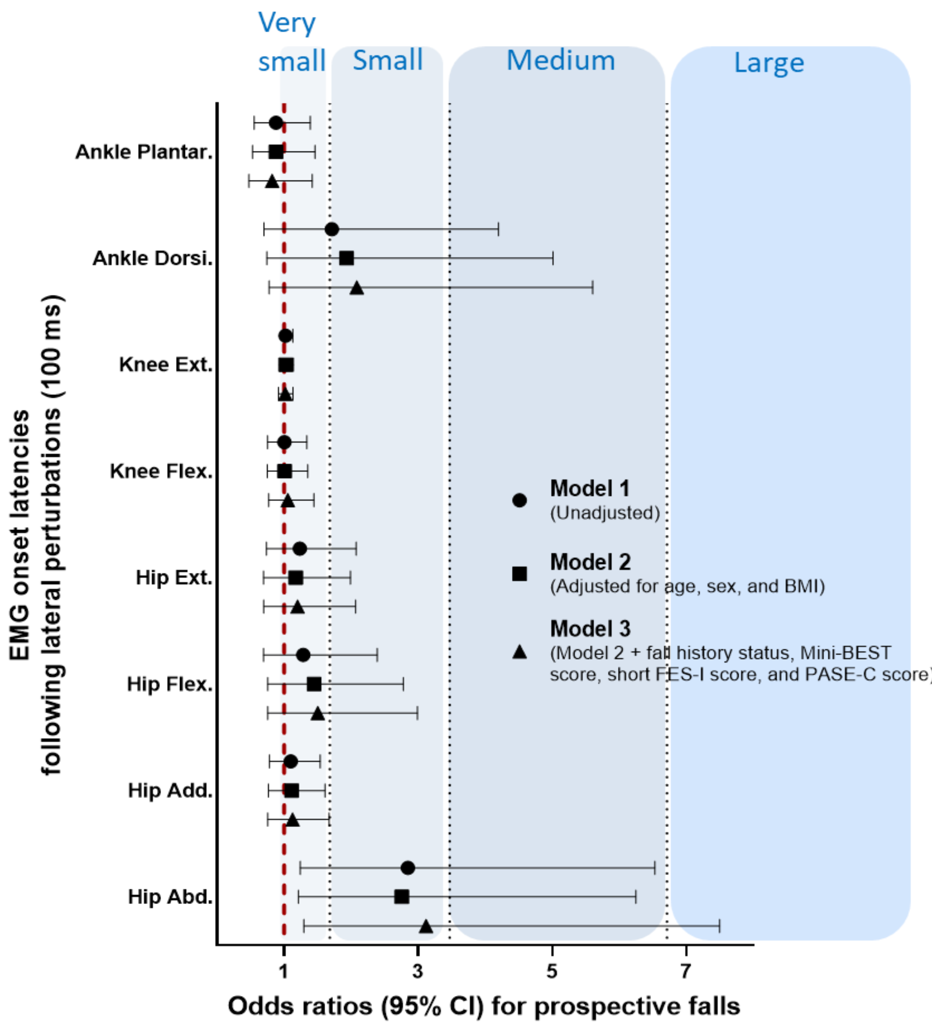


Figure 7-3. Fall-predictive abilities and effect size categories of eight leg muscles' EMG onset latencies following unexpected waist-pull perturbations.

Previous studies have explored extrinsic measures, such as whole-body postural sways and stepping responses, in detecting older adults with fall histories or risks. One prior study reported that the delayed time to peak CoP displacement for maintaining feet-in-place strategies following unexpected mediolateral translational perturbations could identify the older adults with prospective falls with a non-significant poor accuracy (AUC = 0.68, $p > 0.05$) (Maki et al., 1994). Similarly, a recent meta-analysis revealed that a greater number of steps following reactive step tests (including waist-pull perturbations, tether-release tests, and induced slips) showed non-significantly acceptable accuracy in distinguishing the older adults with fall histories/risks, although with a broad 95% confidence interval ranging from 0.47 to 0.90 (AUC = 0.74, $p > 0.05$) (Okubo et al., 2021).

Together, these findings suggest that both the intrinsic neuromuscular/biomechanical measures and the extrinsic measures of reactive balance control may offer similar abilities in identifying fall risks. While intrinsic measures can reflect the fundamental reasons behind declines in reactive balance control among the fall-prone older adults and have significant accuracies in fall prediction, they might not improve the fall-predictive ability or accuracy a lot compared to the traditional step responses or postural sway measurements.

7.6 Suggestions and Implications for Clinical Practice

Firstly, the findings of this project can provide evidence and suggestions for advancing fall-risk assessment. This study has found that the insufficient activation of hip abductor following sudden balance loss can indicate both fall risks and fall histories among the community-dwelling older people. Particularly, the insufficient activation of hip abductor following mediolateral sudden balance loss has exhibited acceptable abilities, which are comparable to the traditional postural sway or stepping response measures, in identifying the older people with high fall risks. Given these, this project suggests that the measurement of hip abductor's activities can complement the current clinical assessments in fall-risk detection. By leveraging the cut-point values derived from this research, therapists can gain valuable insights into interpreting outcomes related to reactive balance control and identify individuals at higher risk of falls, thereby facilitating decision making. Nevertheless, the feasibility and cost-effectiveness of adding an EMG sensor for fall-risk assessment need to be further determined.

Secondly, this project can provide insights for enhancing fall-prevention management. The findings of this project may imply that exercises targeting the specific proximal lower-limb joints and muscles, especially the hip abductor, could potentially yield more benefits in preventing falls among older adults. For example, one recent study reported that both the two exercises that targeted hip abductor, (i.e., reactive step training induced by lateral waist-pull perturbations, hip abductor resistance training) can reduce community-dwelling older people's fall incidences, although the reactive step training did not show additional fall-prevention effects compared to resistance training (Rogers et al., 2021). It is expected that using the hip abductor's activation as the outcome for each session of reactive step training may give clients valuable feedback on their training performance from the previous session and may aid clinicians in tailoring the progression of a client's overall training regimen. Further studies are merited to examine if incorporating hip abductor activation monitoring can enhance the efficacy of reactive step training in preventing falls among older adults.

7.7 Limitations of This Project

Firstly, some limitations existed in EMG and MMG signal processing. Regarding EMG signal processing, this PhD project did not do maximal voluntary contraction (MVC) tests for the eight investigated leg muscles to obtain maximum EMG amplitudes as the references of normalization. Instead, the baseline EMG signal values during unperturbed standing were utilized for normalizing EMG amplitudes. Therefore, when interpreting the EMG amplitude parameters, it is essential to recognize that they represented the level of muscle activation needed for the perturbation task in comparison to standard standing conditions. If time permitting, future research may try to conduct MVC tests for each leg muscle or conduct functional tasks (e.g., limit of standing stability test by leaning forward/backward/sideways) to get the reference EMG signal values for normalization. Regarding the MMG signal processing, it is worth noting that the methods of isolating MMG signal, i.e., muscle vibration signal, from the accelerometry data of dynamic situations that were tried in this project were still immature.

Secondly, for the studies enrolling older adults, this project did not recruit only recurrent fallers as the fall-prone older group. A previous study has reported that in contrast to fallers (including single fallers and recurrent fallers), recurrent fallers had higher odds of prospective falls and could be the more typical representation of fall-prone people. Nevertheless, this project attempted to investigate the more intrinsic mechanisms during reactive balance control that can indicate the fall histories/risks in older people with relatively good health. Therefore, this project did not restrict the enrollment of single fallers. Comparing older recurrent fallers with older non-fallers may possibly reveal a greater number of neuromuscular and biomechanical factors related to falls.

Thirdly, the sample size for the prospective cohort analysis was not adequately justified. In study 4, the sample size of 72 older subjects was estimated from the pilot cross-sectional analysis that compared older fallers and older non-fallers. During the tracking of prospective falls in this cohort of older subjects, only a small percentage experienced multiple falls (7/72). This rendered the investigation of risk factors for multiple falls unsuitable for regression analysis. Further research may be necessary to delve deeper into this aspect.

Lastly, multiple separate statistical analyses were conducted to investigate the ability/accuracy of a single measure of reactive balance in fall prediction, giving rise to type I error; besides, it remains a debate to use a composite score of multiple measures or use a single measure for fall-risk assessment. One the one hand, beyond using independent sample t tests or Mann Whitney

U tests to separately examine the fall-related differences in each amplitude/temporal parameter of a signal, future attempts will be made to try some other statistical methods (e.g., statistical parametric mapping) to compare fall-related neuromuscular/biomechanical time-series data. This way may reduce the number of statistical inferences. On the other hand, it should be noted that based on this PhD project, a single neuromuscular/biomechanical measure of reactive balance showed only a very small to small fall-predictive ability and a poor to acceptable fall-predictive accuracy. This partially reflects the multi-factorial nature of cause for falls (Ma et al., 2024). While identifying a single major cause for falls is essential for targeted intervention to prevent future falls, a composite score of fall risks based on multiple measures may help identify the fall-prone older people without a significant disease/impairment early. Further exploration of the data of this PhD project is merited.

Chapter 8 Future Work

Given the limitation of this PhD project, there are some future research directions as below.

Firstly, the muscle contractile properties that can indicate older adults' fall risks merit further investigation. On the one hand, methods of processing accelerometry could be further optimized to make the detection of muscle vibrations in dynamic situations, e.g., reactive balance control, possible. Besides, with the most recent development of ultrafast ultrasound imaging technology, a more real-time characterization of muscle vibrations became possible. This new MMG approach, called sono-mechano-myo-graphy (SMMG), sampled the isometric muscle contractions at a very high frequency of 20 kHz (Ling et al., 2020). With the EMG and MMG signals, it might be better understood whether the delayed reaction in fall-prone older people is correlated more with the deficits in neural drive or with the deficits in contraction. On the other hand, alternative methods of measuring muscle contractile properties can be tried. Sonomyography (SMG, e.g., ultrasound imaging of muscles) enables visualizing skeletal muscle morphology and more subtle architectural changes of muscle contractions (e.g., muscle thickness, fascicle length, pennation angle) which are assumed as the basis of force generation during reactive balance control (Azizi et al., 2008; Li et al., 2015). With the advancing of ultrasound imaging technology, some wearable SMG devices have been available, making the observation of muscle morphological change in dynamic situations possible (Liu et al., 2024; Lyu et al., 2022; Ma, Ling, et al., 2019). In addition, the SMG can visualize the muscles in a deeper layer, which are assumed important for postural control but cannot be detected by the surface EMG. Given that the identified neuromuscular/biomechanical measures of reactive balance control in this project have shown similar fall-risk detection abilities compared with the traditional extrinsic measures, it is expected that muscle contractile properties detected by SMG could reveal the more intrinsic deficits of fall-prone older people's reactive balance control and provide more sensitive fall-risk assessment. In addition, the SMG technology can avoid the crosstalk issue of EMG. However, the drawbacks of using SMG should be also noted, such as relatively low sampling frequency, no guideline consensus on the ultrasound probe location on a muscle.

Secondly, considering the recurrent or injurious fallers higher risks of future falls, research and fall-prevention management may need to prioritize these groups, especially when resources are limited.

Thirdly, further studies are merited to investigate if incorporating the monitoring of hip abductor activities can enhance the efficacy of perturbation-based balance training (PBT) in preventing falls among older adults. In the last twenty years, PBT has been emerging to target the improvement of reactive balance control (McCrum et al., 2022; Zhu, Schulte, et al., 2025). This kind of training delivers the unexpected perturbations with sufficient intensity to challenge a person's balance control in a controlled and safe environment, during which the person can react and adapt to the sudden loss of balance and learn the balance recovery strategies gradually (McCrum et al., 2022). Nevertheless, the PBT did not show consistent abilities in reducing older adults' real-life fall incidences (Pai et al., 2014; Wang et al., 2022; Zhu, Schulte, et al., 2025). As the reduced and slowed activation of hip abductor during reactive balance control indicate fall risks in older adults, monitoring the hip abductor's activation is expected to provide supplementary feedback on fall-related performance to both clinicians and older adults, which is expected to ultimately improve the effectiveness of PBT in preventing falls among older adults.

It is worth noting that the fall prevalence seemed to have decreased within the past two decades. Globally, the annual prevalence of falls in older adults has reduced from 28%-35% (World Health Organization, 2008) to 26.5% (Salari et al., 2022). In Hong Kong communities, the annual prevalence of falls in older adults has reduced from 19.3% (Chu et al., 2007) to 15.7% (Elderly Health Centre, 2023). These facts can partially reflect that the efforts in fall prevention works. Having built more evidence on using reactive balance test for fall-risk assessment, this PhD project expects to contribute to the design of more targeted reactive balance training for fall prevention in the future.

Chapter 9 Conclusions

This project has found the below enriched neuromuscular/biomechanical mechanisms of reactive balance control in both young and older adults. (1) Ankle muscles exhibited early activation, and had the largest activation rates among the eight investigated leg muscles irrespective of perturbation directions (anterior/posterior/medial/lateral) and methods (waist-pull/moving-platform perturbations). (2) Lower-limb joint responses that followed the direction of waist-pull perturbation had rapid and major responses for both young and older adults. (3) Lager magnitudes of perturbations could evoke faster responses in leg muscle activities but may not in lower-limb joint motions or postural sways, irrespective of perturbation directions and methods.

This project has identified the neuromuscular/biomechanical alterations underlying reactive balance control that were related to older adults' fall histories. (1) In contrast to older non-fallers, older fallers tended to use the suspensory strategy to compensate for the slower initiation of ankle and hip strategies following unexpected translational moving-platform perturbations; however, this led to their elongated and overacted balance recovery. (2) In contrast to older non-fallers, older fallers exhibited a quicker neuromuscular response following anterior waist-pull perturbations but slower neuromuscular responses following posterior/medial/lateral waist-pull perturbations. Such alterations might explain their alterations in stepping responses.

This project has found that older adults' prospective falls could be predicted by the neuromuscular/biomechanical alterations following unexpected waist-pull perturbations as below: (1) slowed and reduced activation of only hip abductor among the eight investigated leg muscles, (2) altered responses mostly in hip/knee joint than in ankle joint, including the more absorbed hip/knee powers and (3) alterations mostly in response to the mediolateral perturbations than anteroposterior perturbations. Taken together, this project suggests the importance of monitoring hip abductor's activation to complement the current assessment of reactive balance control in identifying the community-dwelling older adults with high fall risks, and provides insights on the development of more targeted fall-prevention exercises.

References

Aagaard, P., Suetta, C., Caserotti, P., Magnusson, S. P., & Kjær, M. (2010). Role of the nervous system in sarcopenia and muscle atrophy with aging: strength training as a countermeasure. *Scandinavian Journal of Medicine & Science in Sports*, *20*(1), 49-64. https://doi.org/https://doi.org/10.1111/j.1600-0838.2009.01084.x

Afschrift, M., Pitto, L., Aerts, W., van Deursen, R., Jonkers, I., & De Groote, F. (2018). Modulation of gluteus medius activity reflects the potential of the muscle to meet the mechanical demands during perturbed walking. *Scientific Reports*, *8*(1), 1-11.

Ahmadiahangar, A., Jayadian, Y., Babaei, M., Heidari, B., Hosseini, S., & Aminzadeh, M. (2018). The role of quadriceps muscle strength in the development of falls in the elderly people, a cross-sectional study. *Chiropractic & Manual Therapies*, *26*, Article 31. https://doi.org/10.1186/s12998-018-0195-x

Allum, J. H., Carpenter, M. G., Honegger, F., Adkin, A. L., & Bloem, B. R. (2002). Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man. *Journal of Physiology*, *542*(Pt 2), 643-663. https://doi.org/10.1113/jphysiol.2001.015644

Asai, T., Oshima, K., Fukumoto, Y., Yonezawa, Y., Matsuo, A., & Misu, S. (2022). The association between fear of falling and occurrence of falls: a one-year cohort study. *BMC Geriatrics*, *22*(1), 393.

Atrsaei, A., Paraschiv-Ionescu, A., Krief, H., Henchoz, Y., Santos-Eggimann, B., Bula, C., & Aminian, K. (2021). Instrumented 5-Time Sit-To-Stand Test: Parameters Predicting Serious Falls beyond the Duration of the Test. *Gerontology*. https://doi.org/10.1159/000518389

Azizi, E., Brainerd, E. L., & Roberts, T. J. (2008). Variable gearing in pennate muscles. *Proceedings of the National Academy of Sciences*, *105*(5), 1745. https://doi.org/10.1073/pnas.0709212105

Bair, W.-N., Prettyman, M. G., Beamer, B. A., & Rogers, M. W. (2016). Kinematic and behavioral analyses of protective stepping strategies and risk for falls among community living older adults. *Clinical Biomechanics*, *36*, 74-82. https://www.ncbi.nlm.nih.gov/pmc/articles/PMC5568553/pdf/nihms867635.pdf

Baltasar-Fernandez, I., Alcazar, J., Rodriguez-Lopez, C., Losa-Reyna, J., Alonso-Seco, M., Ara, I., & Alegre, L. M. (2021). Sit-to-stand muscle power test: Comparison between estimated and force plate-derived mechanical power and their association with physical function in older adults. *Experimental Gerontology*, *145*, 111213. https://doi.org/https://doi.org/10.1016/j.exger.2020.111213

Barbero, M., Merletti, R., & Rainoldi, A. (2012). *Atlas of muscle innervation zones: understanding surface electromyography and its applications*. Springer Science & Business Media.

Barker, A. L., Nitz, J. C., Low Choy, N. L., & Haines, T. (2009). Measuring Fall Risk and Predicting Who Will Fall: Clinimetric Properties of Four Fall Risk Assessment Tools for Residential Aged Care. *The Journals of Gerontology: Series A*, *64A*(8), 916-924. https://doi.org/10.1093/gerona/glp041

Batcir, S., Shani, G., Shapiro, A., Alexander, N., & Melzer, I. (2020). The kinematics and strategies of recovery steps during lateral losses of balance in standing at different perturbation magnitudes in older adults with varying history of falls. *BMC Geriatrics*, *20*(1), 249. https://doi.org/10.1186/s12877-020-01650-4

Bates, A. V., McGregor, A., & Alexander, C. M. (2021). Adaptation of balance reactions following forward perturbations in people with joint hypermobility syndrome. *BMC Musculoskeletal Disorders*, *22*(1), 123. https://doi.org/10.1186/s12891-021-03961-y

Baudry, S., Lecoeuvre, G., & Duchateau, J. (2012). Age-related changes in the behavior of the muscle-tendon unit of the gastrocnemius medialis during upright stance. *Journal of Applied Physiology*, *112*(2), 296-304.

Bento, P. C., Pereira, G., Ugrinowitsch, C., & Rodacki, A. L. (2010). Peak torque and rate of torque development in elderly with and without fall history. *Clinical Biomechanics (Bristol, Avon)*, *25*(5), 450-454. https://doi.org/10.1016/j.clinbiomech.2010.02.002

Bergen, G., Stevens, M. R., & Burns, E. R. (2016). Falls and fall injuries among adults aged≥ 65 years—United States, 2014. *Morbidity and Mortality Weekly Report*, *65*(37), 993-998.

Besomi, M., Hodges, P. W., Clancy, E. A., Van Dieën, J., Hug, F., Lowery, M., Merletti, R., Søgaard, K., Wrigley, T., Besier, T., Carson, R. G., Disselhorst-Klug, C., Enoka, R. M., Falla, D., Farina, D., Gandevia, S., Holobar, A., Kiernan, M. C., McGill, K., Perreault, E., Rothwell, J. C., & Tucker, K. (2020). Consensus for experimental design in electromyography (CEDE) project: Amplitude normalization matrix. *Journal of Electromyography and Kinesiology*, *53*, 102438. https://doi.org/https://doi.org/10.1016/j.jelekin.2020.102438

Bhagwat, A. P., & Deodhe, N. P. (2023). The Effect of Perturbation-Based Balance Training vs Step Training on Reaction Time in Older Persons: A Review. *Cureus*, *15*(11), e48104. https://doi.org/10.7759/cureus.48104

Błaszczyszyn, M., Szczęsna, A., & Piechota, K. (2019). sEMG activation of the flexor muscles in the foot during balance tasks by young and older women: a pilot study. *International journal of environmental research and public health*, *16*(22), 4307.

Blenkinsop, G. M., Pain, M. T., & Hiley, M. J. (2017). Balance control strategies during perturbed and unperturbed balance in standing and handstand. *ROYAL SOCIETY OPEN SCIENCE*, *4*(7), 161018.

Blomqvist, S., Wester, A., & Rehn, B. (2014). Postural muscle responses and adaptations to backward platform perturbations in young people with and without intellectual disability. *Gait and Posture*, *39*(3), 904-908. https://doi.org/https://doi.org/10.1016/j.gaitpost.2013.11.018

Borenstein, M., Hedges, L. V., Higgins, J. P., & Rothstein, H. R. (2021). *Introduction to meta-analysis*. John Wiley & Sons.

Borrelli, J. R., Zabukovec, J., Jones, S., Junod, C. A., & Maki, B. E. (2019). Age-related changes in the capacity to select early-onset upper-limb reactions to either recover balance or protect against impact. *Experimental Gerontology*, *125*, 110676.

Bos, R. A., Haarman, C. J. W., Stortelder, T., Nizamis, K., Herder, J. L., Stienen, A. H. A., & Plettenburg, D. H. (2016). A structured overview of trends and technologies used in dynamic hand orthoses. *Journal of Neuroengineering and Rehabilitation*, *13*(1), 62. https://doi.org/10.1186/s12984-016-0168-z

Bruyneel, A. V., Gafner, S. C., Ferrari, S., Gold, G., Monnin, D., Terrier, P., Bastiaenen, C. H., & Allet, L. (2018). Intra-rater reliability of hip abductor isometric strength testing in a standing position in older fallers and non-fallers. *European Review of Aging and Physical Activity*, *15*, Article 9. https://doi.org/10.1186/s11556-018-0198-6

Cadore, E. L., Rodríguez-Mañas, L., Sinclair, A., & Izquierdo, M. (2013). Effects of Different Exercise Interventions on Risk of Falls, Gait Ability, and Balance in Physically Frail Older Adults: A Systematic Review. *Rejuvenation research*, *16*(2), 105-114. https://doi.org/10.1089/rej.2012.1397

Calmels, P., & Minaire, P. (1995). A review of the role of the agonist/antagonist muscle pairs ratio in rehabilitation. *Disability and Rehabilitation*, *17*(6), 265-276. https://doi.org/10.3109/09638289509166646

Cattagni, T., Scaglioni, G., Laroche, D., Grémeaux, V., & Martin, A. (2016). The involvement of ankle muscles in maintaining balance in the upright posture is higher in elderly fallers. *Experimental Gerontology*, *77*, 38-45.

Caulcrick, C., Huo, W., Hoult, W., & Vaidyanathan, R. (2021). Human Joint Torque Modelling With MMG and EMG During Lower Limb Human-Exoskeleton Interaction. *IEEE Robotics and Automation Letters*, *6*(4), 7185-7192. https://doi.org/10.1109/LRA.2021.3097832

CDC. (2019). *Resource: algorithm for fall risk screening, assessment, and intervention*. Centers for Disease Control and Prevention. https://www.cdc.gov/steadi/pdf/STEADI-Algorithm-508.pdf

Cè, E., Coratella, G., Bisconti, A. V., Venturelli, M., Limonta, E., Doria, C., Rampichini, S., Longo, S., & Esposito, F. (2020). Neuromuscular versus mechanical stretch-induced changes in contralateral versus ipsilateral muscle. *Med Sci Sports Exerc*, *52*(6), 1294-1306.

Cè, E., Longo, S., Limonta, E., Coratella, G., Rampichini, S., & Esposito, F. (2020). Peripheral fatigue: new mechanistic insights from recent technologies. *European Journal of Applied Physiology*, *120*(1), 17-39. https://doi.org/10.1007/s00421-019-04264-w

Cè, E., Rampichini, S., Agnello, L., Limonta, E., Veicsteinas, A., & Esposito, F. (2013). Effects of temperature and fatigue on the electromechanical delay components [https://doi.org/10.1002/mus.23627]. *Muscle & Nerve*, *47*(4), 566-576. https://doi.org/https://doi.org/10.1002/mus.23627

Cescon, C., Farina, D., Gobbo, M., Merletti, R., & Orizio, C. (2004). Effect of accelerometer location on mechanomyogram variables during voluntary, constant-force contractions in three human muscles. *Medical and Biological Engineering and Computing*, *42*(1), 121-127. https://doi.org/10.1007/BF02351021

Chan, B. K. S., Marshall, L. M., Winters, K. M., Faulkner, K. A., Schwartz, A. V., & Orwoll, E. S. (2007). Incident fall risk and physical activity and physical performance among older men - The Osteoporotic Fractures in Men Study. *American Journal of Epidemiology*, *165*(6), 696-703. https://doi.org/10.1093/aje/kwk050

Chang, J. T., & Ganz, D. A. (2007). Quality Indicators for Falls and Mobility Problems in Vulnerable Elders [https://doi.org/10.1111/j.1532-5415.2007.01339.x]. *Journal of the American Geriatrics Society*, *55*(s2), S327-S334. https://doi.org/https://doi.org/10.1111/j.1532-5415.2007.01339.x

Chen, B., Lee, Y.-J., & Aruin, A. S. (2015). Anticipatory and compensatory postural adjustments in conditions of body asymmetry induced by holding an object. *Experimental Brain Research*, *233*(11), 3087-3096. https://doi.org/10.1007/s00221-015-4377-7

Chen, C.-L., Lou, S.-Z., Wu, H.-W., Wu, S.-K., Yeung, K.-T., & Su, F.-C. (2014). Effects of the type and direction of support surface perturbation on postural responses. *Journal of Neuroengineering and Rehabilitation*, *11*(1), 50. https://doi.org/10.1186/1743-0003-11-50

Chen, H., Cohen, P., & Chen, S. (2010). How big is a big odds ratio? Interpreting the magnitudes of odds ratios in epidemiological studies. *Communications in Statistics—simulation and Computation®*, *39*(4), 860-864.

Cheng, K. B. (2016). Does knee motion contribute to feet-in-place balance recovery? *Journal of Biomechanics*, *49*(9), 1873-1880. https://doi.org/https://doi.org/10.1016/j.jbiomech.2016.04.026

Cheng, P., Tan, L., Ning, P., Li, L., Gao, Y., Wu, Y., Schwebel, D. C., Chu, H., Yin, H., & Hu, G. (2018). Comparative Effectiveness of Published Interventions for Elderly Fall Prevention: A Systematic Review and Network Meta-Analysis. *International Journal of Environmental Research and Public Health*, *15*(3). https://doi.org/10.3390/ijerph15030498

Cheng, Y. Y., Wei, S. H., Chen, P. Y., Tsai, M. W., Cheng, I. C., Liu, D. H., & Kao, C. L. (2014). Can sit-to-stand lower limb muscle power predict fall status? [Article]. *Gait and Posture*, *40*(3), 403-407. https://doi.org/10.1016/j.gaitpost.2014.05.064

Chu, L., Chi, I., & Chiu, A. (2007). Falls and fall-related injuries in community-dwelling elderly persons in Hong Kong: a study on risk factors, functional decline, and health services utilization after falls. *Hong Kong Medical Journal. Xianggang Yi Xue Za Zhi*, *13*(Suppl 1), 8-12.

Claudino, R., Dos Santos, M. J., & Mazo, G. Z. (2017). Delayed Compensatory Postural Adjustments After Lateral Perturbations Contribute to the Reduced Ability of Older Adults to Control Body Balance. *Motor Control*, *21*(4), 425-442. https://doi.org/10.1123/mc.2016-0005

Cogliati, M., Cudicio, A., Toscani, F., Gaffurini, P., Bissolotti, L. M., Orizio, C., & Negro, F. (2020). Normalized maximal rate of torque development during voluntary and stimulated static contraction in human tibialis anterior: Influence of age. *Experimental Gerontology*, *138*, Article 110999. https://doi.org/10.1016/j.exger.2020.110999

Cohen, J. (1992). Quantitative methods in psychology: A power primer. *Psychological Bulletin*, *112*, 1155-1159.

Cronin, N. J., Barrett, R. S., Lichtwark, G., Mills, P. M., & Carty, C. P. (2013). Decreased lower limb muscle recruitment contributes to the inability of older adults to recover with a single step following a forward loss of balance. *Journal of electromyography and kinesiology*, *23*(5), 1139-1144.

Cronin, N. J., Prilutsky, B. I., Lichtwark, G. A., & Maas, H. (2013). Does ankle joint power reflect type of muscle action of soleus and gastrocnemius during walking in cats and humans? *Journal of Biomechanics*, *46*(7), 1383-1386. https://doi.org/https://doi.org/10.1016/j.jbiomech.2013.02.023

Crozara, L. F., Morcelli, M. H., Hallal, C. Z., Marques, N. R., Spinoso, D. H., Goethel, M. F., Vieira, E. R., & Gonçalves, M. (2016). Effect of age and fall status on lower-extremity muscle activation and joint torque and power in physically active women [Article]. *Isokinetics and Exercise Science*, *24*(1), 67-77. https://doi.org/10.3233/IES-150602

Crozara, L. F., Morcelli, M. H., Marques, N. R., Hallal, C. Z., Spinoso, D. H., de Almeida Neto, A. F., Cardozo, A. C., & Gonçalves, M. (2013). Motor readiness and joint torque production in lower limbs of older women fallers and non-fallers. *J Electromyogr Kinesiol*, *23*(5), 1131-1138. https://doi.org/10.1016/j.jelekin.2013.04.016

Cunha, A. I. L., Veronese, N., de Melo Borges, S., & Ricci, N. A. (2019). Frailty as a predictor of adverse outcomes in hospitalized older adults: A systematic review and meta-analysis. *AGEING RESEARCH REVIEWS*, *56*, 100960. https://doi.org/https://doi.org/10.1016/j.arr.2019.100960

Daly, R. M. (2010). Independent and Combined Effects of Exercise and Vitamin D on Muscle Morphology, Function and Falls in the Elderly. *NUTRIENTS*, *2*(9), 1005-1017. https://doi.org/10.3390/nu2091005

de Freitas, P. B., Knight, C. A., & Barela, J. A. (2010). Postural reactions following forward platform perturbation in young, middle-age, and old adults. *Journal of electromyography and kinesiology*, *20*(4), 693-700. https://doi.org/https://doi.org/10.1016/j.jelekin.2009.11.009

de Vries, O. J., Peeters, G. M. E. E. G., Elders, P. J. M., Muller, M., Knol, D. L., Danner, S. A., Bouter, L. M., & Lips, P. (2010). Multifactorial Intervention to Reduce Falls in Older People at High Risk of Recurrent Falls: A Randomized Controlled Trial. *Archives of Internal Medicine*, *170*(13), 1110-1117. https://doi.org/10.1001/archinternmed.2010.169

Deandrea, S., Lucenteforte, E., Bravi, F., Foschi, R., La Vecchia, C., & Negri, E. (2010). Risk factors for falls in community-dwelling older people: a systematic review and meta-analysis. *Epidemiology*, *21*(5), 658-668.

Di Nardo, F., Morano, M., Strazza, A., & Fioretti, S. (2022). Muscle Co-Contraction Detection in the Time-Frequency Domain. *Sensors (Basel)*, *22*(13). https://doi.org/10.3390/s22134886

Dietzel, R., Felsenberg, D., & Armbrecht, G. (2015). Mechanography performance tests and their association with sarcopenia, falls and impairment in the activities of daily living - a pilot cross-sectional study in 293 older adults. *Journal of musculoskeletal & neuronal interactions*, *15*(3), 249-256. https://pubmed.ncbi.nlm.nih.gov/26350943

Dykes, P. C., Curtin-Bowen, M., Lipsitz, S., Franz, C., Adelman, J., Adkison, L., Bogaisky, M., Carroll, D., Carter, E., & Herlihy, L. (2023). Cost of inpatient falls and cost-benefit analysis of implementation of an evidence-based fall prevention program. JAMA Health Forum,

Ejupi, A., Brodie, M., Lord, S. R., Annegarn, J., Redmond, S. J., & Delbaere, K. (2017). Wavelet-Based Sit-To-Stand Detection and Assessment of Fall Risk in Older People Using a Wearable Pendant Device [Article]. *IEEE Transactions on Biomedical Engineering*, *64*(7), 1602-1607, Article 7581085. https://doi.org/10.1109/TBME.2016.2614230

Elhadi, M. M. O., Ma, C. Z.-H., Lam, W. K., & Lee, W. C.-C. (2018). Biomechanical approach in facilitating long-distance walking of elderly people using footwear modifications. *Gait and Posture*, *64*, 101-107.

Fabre, J. M., Ellis, R., Kosma, M., & Wood, R. H. (2010). Falls risk factors and a compendium of falls risk screening instruments. *Journal of Geriatric Physical Therapy*, *33*(4), 184-197.

Falconer, K. (1985). Quantitative assessment of cocontraction at the ankle joint in walking. *Electromyography and Clinical Neurophysiology*, *25*, 135-148.

Falk, J., Strandkvist, V., Pauelsen, M., Vikman, I., Nyberg, L., & Röijezon, U. (2022). Increased co-contraction reaction during a surface perturbation is associated with unsuccessful postural control among older adults. *BMC Geriatrics*, *22*(1), 438. https://doi.org/10.1186/s12877-022-03123-2

Faulkner, K. A., Redfern, M. S., Cauley, J. A., Landsittel, D. P., Studenski, S. A., Rosano, C., Simonsick, E. M., Harris, T. B., Shorr, R. I., & Ayonayon, H. N. (2007). Multitasking: association between poorer performance and a history of recurrent falls. *Journal of the American Geriatrics Society*, *55*(4), 570-576.

Fluss, R., Faraggi, D., & Reiser, B. (2005). Estimation of the Youden Index and its associated cutoff point. *Biometrical Journal: Journal of Mathematical Methods in Biosciences*, *47*(4), 458-472.

Folland, J. P., Buckthorpe, M. W., & Hannah, R. (2014). Human capacity for explosive force production: Neural and contractile determinants [https://doi.org/10.1111/sms.12131]. *Scandinavian Journal of Medicine & Science in Sports*, *24*(6), 894-906. https://doi.org/https://doi.org/10.1111/sms.12131

Fonseca, L. C. S., Karuka, A. H., Crozara, L. F., Spinoso, D. H., Hallal, C. Z., Marques, N. R., Gonçalves, M., & Scheicher, M. E. (2014). Electromyographic reaction time in older female fallers and non-fallers after postural perturbation. Revista Brasileira de Cineantropometria e Desempenho Humano,

Foreman, K. B., Addison, O., Kim, H. S., & Dibble, L. E. (2011). Testing balance and fall risk in persons with Parkinson disease, an argument for ecologically valid testing. *Parkinsonism & Related Disorders*, *17*(3), 166-171.

Franchignoni, F., Horak, F., Godi, M., Nardone, A., & Giordano, A. (2010). Using psychometric techniques to improve the Balance Evaluation Systems Test: the mini-BESTest. *Journal of Rehabilitation Medicine*, *42*(4), 323-331. https://doi.org/10.2340/16501977-0537

Freyler, K., Gollhofer, A., Colin, R., Brüderlin, U., & Ritzmann, R. (2015). Reactive balance control in response to perturbation in unilateral stance: interaction effects of direction, displacement and velocity on compensatory neuromuscular and kinematic responses. *PloS One*, *10*(12), e0144529.

Fujimoto, M., Bair, W. N., & Rogers, M. W. (2015). Center of pressure control for balance maintenance during lateral waist-pull perturbations in older adults. *Journal of Biomechanics*, *48*(6), 963-968. https://doi.org/10.1016/j.jbiomech.2015.02.012

Fukawa, T., & Uchiyama, T. (2016). System Identification of Evoked Mechanomyogram to Clarify Lower Limb Muscle Stiffness in Treadmill Walking. *Advanced Biomedical Engineering*, *5*, 1-6. https://doi.org/10.14326/abe.5.1

Gafner, S. C., Bastiaenen, C. H., Ferrari, S., Gold, G., Terrier, P., Hilfiker, R., & Allet, L. (2018). Hip muscle and hand-grip strength to differentiate between older fallers and non-fallers: a cross-sectional validity study. *Clinical Interventions in Aging*, *13*, 1-8. https://doi.org/10.2147/cia.S146834

Gafner, S. C., Bastiaenen, C. H. G., Ferrari, S., Gold, G., Trombetti, A., Terrier, P., Hilfiker, R., & Allet, L. (2020). The Role of Hip Abductor Strength in Identifying Older Persons at Risk of Falls: A Diagnostic Accuracy Study. *Clinical Interventions in Aging*, *15*, 645-654. https://doi.org/10.2147/cia.S246998

Gates, S., Fisher, J. D., Cooke, M. W., Carter, Y. H., & Lamb, S. E. (2008). Multifactorial assessment and targeted intervention for preventing falls and injuries among older people in community and emergency care settings: systematic review and meta-analysis. *BMJ*, *336*(7636), 130. https://doi.org/10.1136/bmj.39412.525243.BE

Gerards, M., Marcellis, R., Senden, R., Poeze, M., de Bie, R., Meijer, K., & Lenssen, A. (2023). The effect of perturbation-based balance training on balance control and fear of falling in older adults: a single-blind randomised controlled trial. *BMC Geriatrics*, *23*(1), 305. https://doi.org/10.1186/s12877-023-03988-x

Gerards, M. H., Meijer, K., Karamanidis, K., Grevendonk, L., Hoeks, J., Lenssen, A. F., & McCrum, C. (2021). Adaptability to balance perturbations during walking as a potential marker of falls history in older adults. *FRONTIERS IN SPORTS AND ACTIVE LIVING*, *3*, 132.

Gerstner, G. R., Thompson, B. J., Rosenberg, J. G., Sobolewski, E. J., Scharville, M. J., & Ryan, E. D. (2017). Neural and Muscular Contributions to the Age-Related Reductions in Rapid Strength. *Med Sci Sports Exerc*, *49*(7), 1331-1339. https://doi.org/10.1249/mss.0000000000001231

Gesslbauer, B., Hruby, L. A., Roche, A. D., Farina, D., Blumer, R., & Aszmann, O. C. (2017). Axonal components of nerves innervating the human arm. *Annals of Neurology*, *82*(3), 396-408.

Gilles, M., Wing, A. M., & Kirker, S. G. (1999). Lateral balance organisation in human stance in response to a random or predictable perturbation. *Experimental Brain Research*, *124*(2), 137-144.

Gillespie, L. D., Robertson, M. C., Gillespie, W. J., Sherrington, C., Gates, S., Clemson, L., & Lamb, S. E. (2012). Interventions for preventing falls in older people living in the community. *Cochrane Database of Systematic Reviews*(9). https://doi.org/10.1002/14651858.CD007146.pub3

Godi, M., Franchignoni, F., Caligari, M., Giordano, A., Turcato, A. M., & Nardone, A. (2013). Comparison of reliability, validity, and responsiveness of the mini-BESTest and Berg Balance Scale in patients with balance disorders. *Physical Therapy*, *93*(2), 158-167.

Hall, C. D., & Jensen, J. L. (2002). Age-related differences in lower extremity power after support surface perturbations. *Journal of the American Geriatrics Society*, *50*(11), 1782-1788.

Hermens, H., Freriks, B., Merletti, R., Stegeman, D. F., Blok, J. H., Rau, G., Klug, C. D., Hägg, G., Blok, W. J., & Hermens, H. J. (1999). European recommendations for surface electromyography: Results of the SENIAM Project.

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology*, *10*(5), 361-374. https://doi.org/https://doi.org/10.1016/S1050-6411(00)00027-4

Hesam-Shariati, N., Trinh, T., Thompson-Butel, A. G., Shiner, C. T., & McNulty, P. A. (2017). A longitudinal electromyography study of complex movements in Poststroke therapy. 1: heterogeneous changes despite consistent improvements in clinical assessments. *Frontiers in Neurology*, *8*, 340.

Hides, J. A., Beall, P., Smith, M. M. F., Stanton, W., Miokovic, T., & Richardson, C. (2016). Activation of the hip adductor muscles varies during a simulated weight-bearing task. *PHYSICAL THERAPY IN SPORT*, *17*, 19-23.

Higgins, J., Thomas, J., Chandler, J., Cumpston, M., Li, T., Page, M., & Welch, V. (2022). Handbook for Systematic Reviews of Interventions, version 6.3 (updated February 2022). In: The Cochrane Collaboration, London UK.

Hirase, T., Okubo, Y., Menant, J., Lord, S. R., & Sturnieks, D. L. (2020). Impact of pain on reactive balance and falls in community-dwelling older adults: a prospective cohort study. *Age and ageing*, *49*(6), 982-988.

Ho, C. Y., & Bendrups, A. P. (2002). Ankle Reflex Stiffness During Unperceived Perturbation of Standing in Elderly Subjects. *The Journals of Gerontology: Series A*, *57*(9), B344-B350. https://doi.org/10.1093/gerona/57.9.B344

Hoffman, M., Schrader, J., Applegate, T., & Koceja, D. (1998). Unilateral postural control of the functionally dominant and nondominant extremities of healthy subjects. *Journal of athletic training*, *33*(4), 319-322. https://pubmed.ncbi.nlm.nih.gov/16558528

Horak, F. B., & Nashner, L. M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. *Journal of Neurophysiology*, *55*(6), 1369-1381. https://doi.org/10.1152/jn.1986.55.6.1369

Horak, F. B., Wrisley, D. M., & Frank, J. (2009). The balance evaluation systems test (BESTest) to differentiate balance deficits. *Physical Therapy*, *89*(5), 484-498.

Hosmer Jr, D. W., Lemeshow, S., & Sturdivant, R. X. (2013). *Applied logistic regression*. John Wiley & Sons.

Hsiao-Wecksler, E. T. (2008). Biomechanical and age-related differences in balance recovery using the tether-release method. *JOURNAL OF ELECTROMYOGRAPHY AND KINESIOLOGY*, *18*(2), 179-187. https://doi.org/10.1016/j.jelekin.2007.06.007

Hsieh, K. L., Speiser, J. L., Neiberg, R. H., Marsh, A. P., Tooze, J. A., & Houston, D. K. (2023). Factors associated with falls in older adults: A secondary analysis of a 12-month randomized controlled trial. *Archives of Gerontology and Geriatrics*, *108*, 104940.

Huang, A. R., Mallet, L., Rochefort, C. M., Eguale, T., Buckeridge, D. L., & Tamblyn, R. (2012). Medication-Related Falls in the Elderly. *Drugs and Aging*, *29*(5), 359-376. https://doi.org/10.2165/11599460-000000000-00000

Huang, Z.-H., Ma, C. Z.-H., Wang, L.-K., Wang, X.-Y., Fu, S.-N., & Zheng, Y.-P. (2022). Real-time visual biofeedback via wearable ultrasound imaging can enhance the muscle contraction training outcome of young adults. *Journal of Strength and Conditioning Research*, *36*(4), 941.

Hwang, S., Tae, K., Sohn, R., Kim, J., Son, J., & Kim, Y. (2009). The Balance Recovery Mechanisms Against Unexpected Forward Perturbation. *Annals of Biomedical Engineering*, *37*(8), 1629-1637. https://doi.org/10.1007/s10439-009-9717-y

Ikezoe, T., Asakawa, Y., Shima, H., & Ichihashi, N. (2009). PHYSICAL FUNCTION SCREENING OF INSTITUTIONALIZED ELDERLY WOMEN TO PREDICT THEIR RISK OF FALLING.

Japanese Journal of Physical Fitness and Sports Medicine, *58*(5), 489-497. https://doi.org/10.7600/jspfsm.58.489

Inacio, M., Creath, R., & Rogers, M. W. (2019). Effects of aging on hip abductor-adductor neuromuscular and mechanical performance during the weight transfer phase of lateral protective stepping. *Journal of Biomechanics*, *82*, 244-250. https://doi.org/10.1016/j.jbiomech.2018.10.040

Inacio, M., Ryan, A. S., Bair, W.-N., Prettyman, M., Beamer, B. A., & Rogers, M. W. (2014). Gluteal muscle composition differentiates fallers from non-fallers in community dwelling older adults. *BMC Geriatrics*, *14*(1), 37. https://doi.org/10.1186/1471-2318-14-37

Islam, A., Sundaraj, K., Ahmad, B., Ahamed, N. U., & Ali, A. (2012). Mechanomyography Sensors for Muscle Assessment: a Brief Review. *JOURNAL OF PHYSICAL THERAPY SCIENCE*, *24*(12), 1359-1365. https://doi.org/10.1589/jpts.24.1359

Jeon, W., Griffin, L., & Hsiao, H.-Y. (2021). Effects of initial foot position on postural responses to lateral standing surface perturbations in younger and older adults. *Gait and Posture*, *90*, 449-456. https://doi.org/https://doi.org/10.1016/j.gaitpost.2021.09.193

Jiroumaru, T., Kurihara, T., & Isaka, T. (2014a). Establishment of a recording method for surface electromyography in the iliopsoas muscle. *Journal of Electromyography and Kinesiology*, *24*(4), 445-451. https://doi.org/https://doi.org/10.1016/j.jelekin.2014.02.007

Jiroumaru, T., Kurihara, T., & Isaka, T. (2014b). Measurement of muscle length-related electromyography activity of the hip flexor muscles to determine individual muscle contributions to the hip flexion torque. *SpringerPlus*, *3*, 624-624. https://doi.org/10.1186/2193-1801-3-624

Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., Gainey, J., Gorton, G., & Cochran, G. V. B. (1989). Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait [https://doi.org/10.1002/jor.1100070611]. *Journal of Orthopaedic Research*, *7*(6), 849-860. https://doi.org/https://doi.org/10.1002/jor.1100070611

Kalache, A., Fu, D., Yoshida, S., Al-Faisal, W., Beattie, L., Chodzko-Zajko, W., Fu, H., James, K., Kalula, S., & Krishnaswamy, B. (2007). World Health Organisation Global Report on Falls Prevention in Older Age.

Kamo, T., Asahi, R., Azami, M., Ogihara, H., Ikeda, T., Suzuki, K., & Nishida, Y. (2019). Rate of torque development and the risk of falls among community dwelling older adults in Japan. *Gait and Posture*, *72*, 28-33. https://doi.org/10.1016/j.gaitpost.2019.05.019

Kanekar, N., & Aruin, A. S. (2014). Aging and balance control in response to external perturbations: role of anticipatory and compensatory postural mechanisms. *Age (Dordrecht, Netherlands)*, *36*(3), 9621-9621. https://doi.org/10.1007/s11357-014-9621-8

Karlsson, M. K., Vonschewelov, T., Karlsson, C., Cöster, M., & Rosengen, B. E. (2013). Prevention of falls in the elderly: a review. *Scandinavian Journal of Public Health*, *41*(5), 442-454. https://doi.org/10.1177/1403494813483215

Kasahara, S., & Saito, H. (2021). Mechanisms of postural control in older adults based on surface electromyography data. *HUMAN MOVEMENT SCIENCE*, *78*, 102803. https://doi.org/https://doi.org/10.1016/j.humov.2021.102803

Kasahara, S., Saito, H., Anjiki, T., & Osanai, H. (2015). The effect of aging on vertical postural control during the forward and backward shift of the center of pressure. *Gait and Posture*, *42*(4), 448-454.

Kemoun, G., Thoumie, P., Boisson, D., & Guieu, J. D. (2002). Ankle dorsiflexion delay can predict falls in the elderly [Article]. *Journal of Rehabilitation Medicine*, *34*(6), 278-283. https://doi.org/10.1080/165019702760390374

Kempen, G. I. J. M., Yardley, L., Van Haastregt, J. C. M., Zijlstra, G. A. R., Beyer, N., Hauer, K., & Todd, C. (2008). The Short FES-I: a shortened version of the falls efficacy scale-international to assess fear of falling. *Age and Ageing*, *37*(1), 45-50. https://doi.org/10.1093/ageing/afm157

Kera, T., Kawai, H., Takahashi, J., Hirano, H., Watanabe, Y., Fujiwara, Y., Ihara, K., Kim, H., & Obuchi, S. (2020). Association between ground reaction force in sit-to-stand motion and falls in community-dwelling older Japanese individuals. *Archives of Gerontology and Geriatrics*, *91*, Article 104221. https://doi.org/10.1016/j.archger.2020.104221

Kim, D., & Hwang, J.-M. (2018). The center of pressure and ankle muscle co-contraction in response to anterior-posterior perturbations. *PloS One*, *13*(11), e0207667. https://doi.org/10.1371/journal.pone.0207667

Kim, J., Lee, W., & Lee, S. H. (2020). A Systematic Review of the Guidelines and Delphi Study for the Multifactorial Fall Risk Assessment of Community-Dwelling Elderly. *International Journal of Environmental Research and Public Health*, *17*(17). https://doi.org/10.3390/ijerph17176097

Kim, J. C., Chon, J., Kim, H. S., Lee, J. H., Yoo, S. D., Kim, D. H., Lee, S. A., Han, Y. J., Lee, H. S., Lee, B. Y., Soh, Y. S., & Won, C. W. (2017). The Association Between Fall History and Physical Performance Tests in the Community-Dwelling Elderly: A Cross-Sectional Analysis. *Annals of Rehabilitation Medicine*, *41*(2), 239-247. https://doi.org/10.5535/arm.2017.41.2.239

Kim, Y., Triolo, M., & Hood, D. A. (2017). Impact of aging and exercise on mitochondrial quality control in skeletal muscle. *Oxidative Medicine and Cellular Longevity*, *2017*(1), 3165396.

King, G. W., Stylianou, A. P., Kluding, P. M., Jernigan, S. D., & Luchies, C. W. (2012). Effects of Age and Localized Muscle Fatigue on Ankle Plantar Flexor Torque Development. *Journal of Geriatric Physical Therapy*, *35*(1), 8-14. https://doi.org/10.1519/JPT.0b013e318221f53b

King, L., & Horak, F. (2013). On the Mini-BESTest: Scoring and the Reporting of Total Scores. *Physical Therapy*, *93*(4), 571-575. https://doi.org/10.2522/ptj.2013.93.4.571

Kobayashi, Y., Ueyasu, Y., Yamashita, Y., & Akagi, R. (2016). Effects of 4 Weeks of Explosive-type Strength Training for the Plantar Flexors on the Rate of Torque Development and Postural Stability in Elderly Individuals. *International Journal of Sports Medicine*, *37*(6), 470-475. https://doi.org/10.1055/s-0035-1569367

Kochoa, P. (2016). *Balance Strategy Basics*. Retrieved June 2023 from https://www.professionalptandtraining.com/balance-strategy-basics/

Kovacikova, Z., Zemkova, E., Neumannova, K., Jelen, M., Jelen, K., & Janura, M. (2015). The role of lateral preference of lower limbs in a postural stabilization task. *Neuroendocrinol. Lett*, *36*, 91-95.

Krašna, S., Keller, A., Linder, A., Silvano, A. P., Xu, J.-C., Thomson, R., & Klug, C. (2021). Human response to longitudinal perturbations of standing passengers on public transport during regular operation. *Frontiers in bioengineering and biotechnology*, *9*.

Ku, P.-W., Sun, W., & Chen, L.-J. (2013). Reliability and Validity of the Chinese Version of the Physical Activity Scale for the Elderly. *Sport & Exercise Research*, *15*, 309-319.

Kvelde, T., McVeigh, C., Toson, B., Greenaway, M., Lord, S. R., Delbaere, K., & Close, J. C. T. (2013). Depressive Symptomatology as a Risk Factor for Falls in Older People: Systematic Review and Meta-Analysis [https://doi.org/10.1111/jgs.12209]. *Journal of the American Geriatrics Society*, *61*(5), 694-706. https://doi.org/https://doi.org/10.1111/jgs.12209

Lamoureux, E. L., Sparrow, W. A., Murphy, A., & Newton, R. U. (2001). Differences in the neuromuscular capacity and lean muscle tissue in old and older community-dwelling adults. *JOURNALS OF GERONTOLOGY SERIES A-BIOLOGICAL SCIENCES AND MEDICAL SCIENCES*, *56*(6), M381-M385. https://doi.org/10.1093/gerona/56.6.M381

LaRoche, D. P., Cremin, K. A., Greenleaf, B., & Croce, R. V. (2010). Rapid torque development in older female fallers and nonfallers: A comparison across lower-extremity muscles. *Journal of Electromyography and Kinesiology*, *20*(3), 482-488. https://doi.org/10.1016/j.jelekin.2009.08.004

Laudani, L., Rum, L., Valle, M. S., Macaluso, A., Vannozzi, G., & Casabona, A. (2021). Age differences in anticipatory and executory mechanisms of gait initiation following unexpected balance perturbations. *European Journal of Applied Physiology*, *121*(2), 465-478. https://doi.org/10.1007/s00421-020-04531-1

Lee, P. H., Macfarlane, D. J., Lam, T. H., & Stewart, S. M. (2011). Validity of the International Physical Activity Questionnaire Short Form (IPAQ-SF): a systematic review. *The International Journal of Behavioral Nutrition and Physical Activity*, *8*, 115. https://doi.org/10.1186/1479-5868-8-115

Li, J., Zhou, Y., Zheng, Y.-P., & Li, G. (2015). An attempt to bridge muscle architecture dynamics and its instantaneous rate of force development using ultrasonography. *Ultrasonics*, *61*, 71-78. https://doi.org/https://doi.org/10.1016/j.ultras.2015.03.009

Li, K.-J., Wong, N. L.-Y., Law, M.-C., Lam, F. M.-H., Wong, H.-C., Chan, T.-O., Wong, K.-N., Zheng, Y.-P., Huang, Q.-Y., & Wong, A. Y.-L. (2023). Reliability, validity, and identification ability of a commercialized waist-attached inertial measurement unit (IMU) sensor-based system in fall risk assessment of older people. *Biosensors*, *13*(12), 998.

Lima, C. A., Ricci, N. A., Nogueira, E. C., & Perracini, M. R. (2018). The Berg Balance Scale as a clinical screening tool to predict fall risk in older adults: a systematic review. *Physiotherapy*, *104*(4), 383-394. https://doi.org/10.1016/j.physio.2018.02.002

Lin, S.-I., & Woollacott, M. H. (2002). Postural muscle responses following changing balance threats in young, stable older, and unstable older adults. *Journal of motor behavior*, *34*(1), 37-44.

Ling, Y. T., Ma, C. Z.-H., Shea, Q. T. K., & Zheng, Y.-P. (2020). Sonomechanomyography (SMMG): Mapping of Skeletal Muscle Motion Onset during Contraction Using Ultrafast Ultrasound Imaging and Multiple Motion Sensors. *Sensors*, *20*(19), 5513. https://www.mdpi.com/1424-8220/20/19/5513

Liu, W., Wu, H.-D., Li, Y.-Y., Zhu, R. T.-L., Luo, Y.-Y., Ling, Y. T., Wang, L.-K., Wang, J.-F., Zheng, Y.-P., & Ma, C. Z.-H. (2024). Effect of Ankle-Foot Orthosis on Paretic Gastrocnemius and Tibialis Anterior Muscle Contraction of Stroke Survivors During Walking: A Pilot Study. *Biosensors*, *14*(12), 595.

Luchies, C. W., Alexander, N. B., Schultz, A. B., & Ashton-Miller, J. (1994). Stepping responses of young and old adults to postural disturbances: kinematics. *Journal of the American Geriatrics Society*, *42*(5), 506-512. https://doi.org/10.1111/j.1532-5415.1994.tb04972.x

Lusardi, M. M., Fritz, S., Middleton, A., Allison, L., Wingood, M., Phillips, E., Criss, M., Verma, S., Osborne, J., & Chui, K. K. (2017). Determining Risk of Falls in Community Dwelling Older Adults: A Systematic Review and Meta-analysis Using Posttest Probability. *Journal of Geriatric Physical Therapy*, *40*(1), 1-36. https://doi.org/10.1519/jpt.0000000000000099

Lyu, P.-Z., Zhu, R. T., Ling, Y. T., Wang, L.-K., Zheng, Y.-P., & Ma, C. Z. (2022). How Paretic and Non-Paretic Ankle Muscles Contract during Walking in Stroke Survivors: New Insight Using Novel Wearable Ultrasound Imaging and Sensing Technology. *Biosensors*, *12*(5).

Ma, C. Z.-H., Bao, T., DiCesare, C. A., Harris, I., Chambers, A., Shull, P. B., Zheng, Y.-P., Cham, R., & Sienko, K. H. (2022). Reducing Slip Risk: A Feasibility Study of Gait Training with Semi-Real-Time Feedback of Foot–Floor Contact Angle. *Sensors*, *22*(10), 3641.

Ma, C. Z.-H., Chung, A. K.-L., Ling, Y. T., Huang, Z.-H., Cheng, C. L.-K., & Zheng, Y.-P. (2019). A Newly-Developed Smart Insole System with Instant Reminder: Paves the Way towards

Integrating Artificial Intelligence (AI) Technology to Improve Balance and Prevent Falls. *Age and Ageing*, *48*(Supplement_4), iv28-iv33.

Ma, C. Z.-H., Lam, W.-K., Chang, B.-C., & Lee, W. C.-C. (2020). Can Insoles Be Used to Improve Static and Dynamic Balance of Community-Dwelling Older Adults? A Systematic Review on Recent Advances and Future Perspectives. *Journal of Aging and Physical Activity*, *28*(6), 971-986. https://doi.org/10.1123/japa.2019-0293

Ma, C. Z.-H., & Lee, W. C.-C. (2017). A wearable vibrotactile biofeedback system improves balance control of healthy young adults following perturbations from quiet stance. *HUMAN MOVEMENT SCIENCE*, *55*, 54-60. https://doi.org/https://doi.org/10.1016/j.humov.2017.07.006

Ma, C. Z.-H., Ling, Y. T., Shea, Q. T. K., Wang, L.-K., Wang, X.-Y., & Zheng, Y.-P. (2019). Towards Wearable Comprehensive Capture and Analysis of Skeletal Muscle Activity during Human Locomotion. *Sensors*, *19*(1), 195. https://doi.org/10.3390/s19010195

Ma, C. Z.-H., Wan, A. H.-P., Wong, D. W.-C., Zheng, Y.-P., & Lee, W. C.-C. (2015). A vibrotactile and plantar force measurement-based biofeedback system: Paving the way towards wearable balance-improving devices. *Sensors*, *15*(12), 31709-31722.

Ma, C. Z.-H., Wong, D. W.-C., Lam, W. K., Wan, A. H.-P., & Lee, W. C.-C. (2016). Balance improvement effects of biofeedback systems with state-of-the-art wearable sensors: A systematic review. *Sensors*, *16*(4), 434.

Ma, C. Z.-H., Wong, D. W.-C., Wan, A. H.-P., & Lee, W. C.-C. (2018). Effects of orthopedic insoles on static balance of older adults wearing thick socks. *Prosthetics and orthotics international*, *42*(3), 357-362.

Ma, C. Z.-H., Zheng, Y.-P., & Lee, W. C.-C. (2018). Changes in gait and plantar foot loading upon using vibrotactile wearable biofeedback system in patients with stroke. *Topics in Stroke Rehabilitation*, *25*(1), 20-27.

Ma, C. Z.-H., Zhu, R. T.-L., Huang, M., Lee, W. C.-C., Yang, Y., & He, C. (2024). Balance-controlling mechanism and fall-prevention strategy. In (Vol. 15, pp. 1385917): Frontiers Media SA.

Maki, B. E., Holliday, P. J., & Topper, A. K. (1994). A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. *Journal of Gerontology*, *49*(2), M72-84. https://doi.org/10.1093/geronj/49.2.m72

Manchester, D., Woollacott, M., Zederbauer-Hylton, N., & Marin, O. (1989). Visual, Vestibular and Somatosensory Contributions to Balance Control in the Older Adult. *Journal of Gerontology*, *44*(4), M118-M127. https://doi.org/10.1093/geronj/44.4.M118

McCrum, C., Bhatt, T. S., Gerards, M. H., Karamanidis, K., Rogers, M. W., Lord, S. R., & Okubo, Y. (2022). Perturbation-based balance training: Principles, mechanisms and implementation in clinical practice. *FRONTIERS IN SPORTS AND ACTIVE LIVING*, *4*, 1015394.

McKinnon, N. B., Connelly, D. M., Rice, C. L., Hunter, S. W., & Doherty, T. J. (2017). Neuromuscular contributions to the age-related reduction in muscle power: Mechanisms and potential role of high velocity power training. *AGEING RESEARCH REVIEWS*, *35*, 147-154. https://doi.org/https://doi.org/10.1016/j.arr.2016.09.003

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., & Lord, S. R. (2008). Optimizing footwear for older people at risk of falls. *Journal of Rehabilitation Research and Development*, *45*(8), 1167-1181. https://doi.org/10.1682/jrrd.2007.10.0168

Mille, M.-L., Johnson-Hilliard, M., Martinez, K. M., Zhang, Y., Edwards, B. J., & Rogers, M. W. (2013). One step, two steps, three steps more… directional vulnerability to falls in community-dwelling older people. *Journals of Gerontology Series A: Biomedical Sciences and Medical Sciences*, *68*(12), 1540-1548.

Mille, M.-L., Johnson, M. E., Martinez, K. M., & Rogers, M. W. (2005). Age-dependent differences in lateral balance recovery through protective stepping. *Clinical Biomechanics*, *20*(6), 607-616. https://doi.org/https://doi.org/10.1016/j.clinbiomech.2005.03.004

Montero-Odasso, M., van der Velde, N., Martin, F. C., Petrovic, M., Tan, M. P., Ryg, J., Aguilar-Navarro, S., Alexander, N. B., Becker, C., Blain, H., Bourke, R., Cameron, I. D., Camicioli, R., Clemson, L., Close, J., Delbaere, K., Duan, L., Duque, G., Dyer, S. M., Freiberger, E., Ganz, D. A., Gómez, F., Hausdorff, J. M., Hogan, D. B., Hunter, S. M. W., Jauregui, J. R., Kamkar, N., Kenny, R.-A., Lamb, S. E., Latham, N. K., Lipsitz, L. A., Liu-Ambrose, T., Logan, P., Lord, S. R., Mallet, L., Marsh, D., Milisen, K., Moctezuma-Gallegos, R., Morris,

M. E., Nieuwboer, A., Perracini, M. R., Pieruccini-Faria, F., Pighills, A., Said, C., Sejdic, E., Sherrington, C., Skelton, D. A., Dsouza, S., Speechley, M., Stark, S., Todd, C., Troen, B. R., van der Cammen, T., Verghese, J., Vlaeyen, E., Watt, J. A., Masud, T., & the Task Force on Global Guidelines for Falls in Older, A. (2022). World guidelines for falls prevention and management for older adults: a global initiative. *Age and Ageing*, *51*(9), afac205. https://doi.org/10.1093/ageing/afac205

Morcelli, M. H., LaRoche, D. P., Crozara, L. F., Marques, N. R., Hallal, C. Z., Rossi, D. M., Goncalves, M., & Navega, M. T. (2016). Neuromuscular performance in the hip joint of elderly fallers and non-fallers. *Aging Clinical and Experimental Research*, *28*(3), 443-450. https://doi.org/10.1007/s40520-015-0448-7

Morcelli, M. H., Rossi, D. M., Karuka, A. H., Crozara, L. F., Hallal, C. Z., Marques, N. R., Goncalves, M., & Navega, M. T. (2016). Peak torque, reaction time, and rate of torque development of hip abductors and adductors of older women. *Physiotherapy Theory and Practice*, *32*(1), 45-52. https://doi.org/10.3109/09593985.2015.1091870

Moreira, L., Figueiredo, J., Fonseca, P., Vilas-Boas, J. P., & Santos, C. P. (2021). Lower limb kinematic, kinetic, and EMG data from young healthy humans during walking at controlled speeds. *Scientific data*, *8*(1), 103.

Moreland, B., Kakara, R., & Henry, A. (2020). Trends in nonfatal falls and fall-related injuries among adults aged≥ 65 years—United States, 2012–2018. *Morbidity and Mortality Weekly Report*, *69*(27), 875.

Moreland, J. D., Richardson, J. A., Goldsmith, C. H., & Clase, C. M. (2004). Muscle Weakness and Falls in Older Adults: A Systematic Review and Meta-Analysis [https://doi.org/10.1111/j.1532-5415.2004.52310.x]. *Journal of the American Geriatrics Society*, *52*(7), 1121-1129. https://doi.org/https://doi.org/10.1111/j.1532-5415.2004.52310.x

Mortaza, N., Abu Osman, N. A., & Mehdikhani, N. (2014). Are the spatio-temporal parameters of gait capable of distinguishing a faller from a non-faller elderly? *European Journal of Physical and Rehabilitation Medicine*, *50*(6), 677-691.

Muir, S. W., Gopaul, K., & Montero Odasso, M. M. (2012). The role of cognitive impairment in fall risk among older adults: a systematic review and meta-analysis. *Age and Ageing*, *41*(3), 299-308. https://doi.org/10.1093/ageing/afs012

Mukamal, K. J., Mittleman, M. A., Longstreth Jr, W. T., Newman, A. B., Fried, L. P., & Siscovick, D. S. (2004). Self-Reported Alcohol Consumption and Falls in Older Adults: Cross-Sectional and Longitudinal Analyses of the Cardiovascular Health Study [https://doi.org/10.1111/j.1532-5415.2004.52318.x]. *Journal of the American Geriatrics Society*, *52*(7), 1174-1179. https://doi.org/https://doi.org/10.1111/j.1532-5415.2004.52318.x

Nastasi, A. J., Ahuja, A., Zipunnikov, V., Simonsick, E. M., Ferrucci, L., & Schrack, J. A. (2018). Objectively measured physical activity and falls in well-functioning older adults: findings from the Baltimore Longitudinal Study of Aging. *American Journal of Physical Medicine and Rehabilitation*, *97*(4), 255-260.

National Heart Lung and Blood Institute. (2021). *Quality Assessment Tool for Observational Cohort and Cross-Sectional Studies*. https://www.nhlbi.nih.gov/health-topics/study-quality-assessment-tools

Nijhuis, L. B. O., Bloem, B. R., Carpenter, M. G., & Allum, J. H. J. (2007). Incorporating voluntary knee flexion into nonanticipatory balance corrections. *Journal of Neurophysiology*, *98*(5), 3047-3059. https://doi.org/10.1152/jn.01303.2006

Ochi, A., Ohko, H., Hayashi, T., Osawa, T., Sugiyama, Y., Nakamura, S., Ibuki, S., & Ichihashi, N. (2020). Relationship Between Balance Recovery From a Forward Fall and Lower-Limb Rate of Torque Development. *Journal of motor behavior*, *52*(1), 71-78. https://doi.org/10.1080/00222895.2019.1585743

Ochi, A., Yokoyama, S., Abe, T., Yamada, K., Tateuchi, H., & Ichihashi, N. (2014). Differences in muscle activation patterns during step recovery in elderly women with and without a history of falls. *Aging Clinical and Experimental Research*, *26*, 213-220.

Okubo, Y., Schoene, D., Caetano, M. J. D., Pliner, E. M., Osuka, Y., Toson, B., & Lord, S. R. (2021). Stepping impairment and falls in older adults: A systematic review and meta-analysis of volitional and reactive step tests. *AGEING RESEARCH REVIEWS*, *66*, Article 101238. https://doi.org/10.1016/j.arr.2020.101238

Omana, H., Bezaire, K., Brady, K., Davies, J., Louwagie, N., Power, S., Santin, S., & Hunter, S. W. (2021). Functional Reach Test, Single-Leg Stance Test, and Tinetti Performance-Oriented Mobility Assessment for the Prediction of Falls in Older Adults: A Systematic Review. *PHYSICAL THERAPY & REHABILITATION JOURNAL*, *101*(10). https://doi.org/10.1093/ptj/pzab173

Ono, H., Sato, T., & Ohnishi, K. (2011). Balance recovery of ankle strategy: Using knee joint for biped robot. 2011 1st International Symposium on Access Spaces (ISAS),

Orizio, C. (1993). Muscle sound: bases for the introduction of a mechanomyographic signal in muscle studies. *Critical Reviews in Biomedical Engineering*, *21*(3), 201-243.

Osoba, M. Y., Rao, A. K., Agrawal, S. K., & Lalwani, A. K. (2019). Balance and gait in the elderly: A contemporary review. *Laryngoscope Investig Otolaryngol*, *4*(1), 143-153. https://doi.org/10.1002/lio2.252

Pai, Y.-C., Bhatt, T., Yang, F., Wang, E., & Kritchevsky, S. (2014). Perturbation training can reduce community-dwelling older adults' annual fall risk: a randomized controlled trial. *Journals of Gerontology Series A: Biomedical Sciences and Medical Sciences*, *69*(12), 1586-1594.

Pai, Y.-C., Rogers, M. W., Patton, J., Cain, T. D., & Hanke, T. A. (1998). Static versus dynamic predictions of protective stepping following waist–pull perturbations in young and older adults. *Journal of Biomechanics*, *31*(12), 1111-1118. https://doi.org/https://doi.org/10.1016/S0021-9290(98)00124-9

Pai, Y. C., Maki, B. E., Iqbal, K., McIlroy, W. E., & Perry, S. D. (2000). Thresholds for step initiation induced by support-surface translation: a dynamic center-of-mass model provides much better prediction than a static model. *Journal of Biomechanics*, *33*(3), 387-392. https://doi.org/https://doi.org/10.1016/S0021-9290(99)00199-2

Palmer, T. B., Thiele, R. M., & Thompson, B. J. (2017). Age-Related Differences in Maximal and Rapid Torque Characteristics of the Hip Extensors and Dynamic Postural Balance in Healthy, Young and Old Females. *The Journal of Strength & Conditioning Research*, *31*(2). https://journals.lww.com/nsca-jscr/Fulltext/2017/02000/Age_Related_Differences_in_Maximal_and_Rapid.26.aspx

Palmer, T. B., Thiele, R. M., Williams, K. B., Adams, B. M., Akehi, K., Smith, D. B., & Thompson, B. J. (2015). The identification of fall history using maximal and rapid isometric torque characteristics of the hip extensors in healthy, recreationally active elderly females: a preliminary investigation. *Aging Clinical and Experimental Research*, *27*(4), 431-438. https://doi.org/10.1007/s40520-014-0305-0

Parsons, C. M., Edwards, M. H., Cooper, C., Dennison, E. M., & Ward, K. A. (2020). Are jumping mechanography assessed muscle force and power, and traditional physical capability measures associated with falls in older adults? Results from the hertfordshire cohort study [Article]. *Journal of musculoskeletal neuronal interactions*, *20*(2), 168-175. https://www.scopus.com/inward/record.uri?eid=2-s2.0-85085678443&partnerID=40&md5=3e7b34b74144b67876a4cb14e92a3447

Perry, M. C., Carville, S. F., Smith, I. C. H., Rutherford, O. M., & Newham, D. J. (2007). Strength, power output and symmetry of leg muscles: effect of age and history of falling. *European Journal of Applied Physiology*, *100*(5), 553-561. https://doi.org/10.1007/s00421-006-0247-0

Phu, S., Sturnieks, D. L., Lord, S. R., & Okubo, Y. (2022). Impact of ageing, fall history and exercise on postural reflexes following unpredictable perturbations: A systematic review and meta-analyses. *Mechanisms of Ageing and Development*, *203*, 111634. https://doi.org/10.1016/j.mad.2022.111634

Piirainen, J. M., Cronin, N. J., Avela, J., & Linnamo, V. (2014). Effects of plyometric and pneumatic explosive strength training on neuromuscular function and dynamic balance control in 60–70year old males. *Journal of Electromyography and Kinesiology*, *24*(2), 246-252. https://doi.org/https://doi.org/10.1016/j.jelekin.2014.01.010

Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2005). Control of support limb muscles in recovery after tripping in young and older subjects. *Experimental Brain Research*, *160*(3), 326-333. https://doi.org/10.1007/s00221-004-2014-y

Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2005). Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. *Gait and Posture*, *21*(4), 388-394. https://doi.org/10.1016/j.gaitpost.2004.04.009

Pijnappels, M., van der Burg, J. C. E., Reeves, N. D., & van Dieën, J. H. (2007). Identification of elderly fallers by muscle strength measures. *European Journal of Applied Physiology*, *102*(5), 585-592. https://doi.org/10.1007/s00421-007-0613-6

Plewa, K., Samadani, A., Orlandi, S., & Chau, T. (2018). A novel approach to automatically quantify the level of coincident activity between EMG and MMG signals. *Journal of Electromyography and Kinesiology*, *41*, 34-40. https://doi.org/https://doi.org/10.1016/j.jelekin.2018.04.001

Portney, L. G. (2020). *Foundations of clinical research: applications to evidence-based practice*. FA Davis.

Porto, J. M., Freire Júnior, R. C., Capato, L. L., Spilla, S. B., Nakaishi, A. P. M., Braz e Silva, E., Faccio, A. F. F., & Abreu, D. C. C. d. (2022). Rate of Torque Development and Torque Steadiness of the Lower Limb and Future Falls Among Community-Dwelling Older Adults Without Previous Falls: A Longitudinal 1-Year Study. *Journal of Aging and Physical Activity*, *30*(2), 168-176. https://doi.org/10.1123/japa.2020-0442

Qu, X., Hu, X., & Lew, F. L. (2012). Differences in lower extremity muscular responses between successful and failed balance recovery after slips. *International Journal of Industrial Ergonomics*, *42*(5), 499-504. https://doi.org/10.1016/j.ergon.2012.08.003

Quam, S. (2020). Johnson Health Tech: EMG Sensor Holder for Heels and Center of Mass Approximation.

Quijoux, F., Vienne-Jumeau, A., Bertin-Hugault, F., Zawieja, P., Lefèvre, M., Vidal, P. P., & Ricard, D. (2020). Center of pressure displacement characteristics differentiate fall risk in older people: A systematic review with meta-analysis. *Ageing Res Rev*, *62*, 101117. https://doi.org/10.1016/j.arr.2020.101117

Radaelli, R., Brusco, C. M., Lopez, P., Rech, A., Machado, C. L. F., Grazioli, R., Müller, D. C., Tufano, J. J., Cadore, E. L., & Pinto, R. S. (2019). Muscle quality and functionality in older women improve similarly with muscle power training using one or three sets. *Experimental Gerontology*, *128*, 110745. https://doi.org/10.1016/j.exger.2019.110745

Reid, K. F., & Fielding, R. A. (2012). Skeletal Muscle Power: A Critical Determinant of Physical Functioning in Older Adults. *EXERCISE AND SPORT SCIENCES REVIEWS*, *40*(1), 4-12. https://doi.org/10.1097/JES.0b013e31823b5f13

Ribeiro, A. M. P., Gomes, M. M., Rosa, R. C., & de Abreu, D. C. C. (2012). Is the History of Falls an Indicative of Greater Decline in Quadriceps Muscle Function and Postural Sway? *Topics in Geriatric Rehabilitation*, *28*(1), 60-66. https://doi.org/10.1097/TGR.0b013e318249a4f5

Rice, P. E., Needle, A. R., Leicht, Z. S., Zwetsloot, K. A., & McBride, J. M. (2019). Bone health, muscle properties and stretch-shortening cycle function of young and elderly males. *Journal of Musculoskeletal & Neuronal Interactions*, *19*(4), 389-395. <Go to ISI>://WOS:000500257300002

Richards, J. (2018). *The comprehensive textbook of clinical biomechanics* (Second edition. ed.). Elsevier.

Rietdyk, S., Patla, A. E., Winter, D. A., Ishac, M. G., & Little, C. E. (1999). Balance recovery from medio-lateral perturbations of the upper body during standing. *Journal of Biomechanics*, *32*(11), 1149-1158. https://doi.org/https://doi.org/10.1016/S0021-9290(99)00116-5

Riley, R. D., Snell, K. I., Ensor, J., Burke, D. L., Harrell Jr, F. E., Moons, K. G., & Collins, G. S. (2019). Minimum sample size for developing a multivariable prediction model: PART II-binary and time-to-event outcomes. *Statistics in Medicine*, *38*(7), 1276-1296.

Rogers, M. W., Creath, R. A., Gray, V., Abarro, J., Waller, S. M., Beamer, B. A., & Sorkin, J. D. (2021). Comparison of Lateral Perturbation-Induced Step Training and Hip Muscle Strengthening Exercise on Balance and Falls in Community-Dwelling Older Adults: A Blinded Randomized Controlled Trial. *JOURNALS OF GERONTOLOGY SERIES A-BIOLOGICAL SCIENCES AND MEDICAL SCIENCES*, *76*(9), E194-E202. https://doi.org/10.1093/gerona/glab017

Rogers, M. W., Hedman, L. D., Johnson, M. E., Cain, T. D., & Hanke, T. A. (2001). Lateral Stability During Forward-Induced Stepping for Dynamic Balance Recovery in Young and Older Adults. *The Journals of Gerontology: Series A*, *56*(9), M589-M594. https://doi.org/10.1093/gerona/56.9.M589

Rubega, M., Di Marco, R., Zampini, M., Formaggio, E., Menegatti, E., Bonato, P., Masiero, S., & Del Felice, A. (2021). Muscular and cortical activation during dynamic and static balance in the elderly: A scoping review. *Aging Brain*, *1*, 100013.

Rubenstein, L. Z. (2006). Falls in older people: epidemiology, risk factors and strategies for prevention. *Age and Ageing*, *35*(suppl_2), ii37-ii41. https://doi.org/10.1093/ageing/afl084 %J Age and Ageing

Runge, C., Shupert, C., Horak, F., & Zajac, F. (1999). Ankle and hip postural strategies defined by joint torques. *Gait and Posture*, *10*(2), 161-170.

Salari, N., Darvishi, N., Ahmadipanah, M., Shohaimi, S., & Mohammadi, M. (2022). Global prevalence of falls in the older adults: a comprehensive systematic review and meta-analysis. *Journal of Orthopaedic Surgery and Research*, *17*(1), 334.

Salzman, B. (2010). Gait and balance disorders in older adults. *American Family Physician*, *82*(1), 61-68.

Santos, M. J., Kanekar, N., & Aruin, A. S. (2010). The role of anticipatory postural adjustments in compensatory control of posture: 2. Biomechanical analysis. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology*, *20*(3), 398-405. https://doi.org/10.1016/j.jelekin.2010.01.002

Sawers, A., Pai, Y.-C., Bhatt, T., & Ting, L. H. (2017). Neuromuscular responses differ between slip-induced falls and recoveries in older adults. *Journal of Neurophysiology*, *117*(2), 509-522.

Scheffer, A. C., Schuurmans, M. J., van Dijk, N., van der Hooft, T., & de Rooij, S. E. (2008). Fear of falling: measurement strategy, prevalence, risk factors and consequences among older persons. *Age Ageing*, *37*(1), 19-24. https://doi.org/10.1093/ageing/afm169

Schettino, L., Luz, C. P. N., de Oliveira, L. E. G., de Assunção, P. L., da Silva Coqueiro, R., Fernandes, M. H., Brown, L. E., Machado, M., & Pereira, R. (2014). Comparison of explosive force between young and elderly women: evidence of an earlier decline from explosive force. *Age*, *36*(2), 893-898. https://doi.org/10.1007/s11357-013-9612-1

Schoene, D., Wu, S. M., Mikolaizak, A. S., Menant, J. C., Smith, S. T., Delbaere, K., & Lord, S. R. (2013). Discriminative ability and predictive validity of the timed up and go test in identifying older people who fall: systematic review and meta-analysis. *Journal of the American Geriatrics Society*, *61*(2), 202-208. https://doi.org/10.1111/jgs.12106

Schulz, B. W., Jongprasithporn, M., Hart-Hughes, S. J., & Bulat, T. (2013). Effects of step length, age, and fall history on hip and knee kinetics and knee co-contraction during the maximum step length test [Article]. *Clinical Biomechanics*, *28*(8), 933-940. https://doi.org/10.1016/j.clinbiomech.2013.08.002

Schünemann, H., Brożek, J., Guyatt, G., & Oxman, A. (2019). GRADE handbook for grading quality of evidence and strength of recommendations. Updated October 2013. The GRADE Working Group, 2013. *Available from guidelinedevelopment. org/handbook*.

Schünemann, H., Hill, S., Guyatt, G., Akl, E. A., & Ahmed, F. (2011). The GRADE approach and Bradford Hill's criteria for causation. *Journal of Epidemiology and Community Health*, *65*(5), 392-395.

Seidler, R. D., Bernard, J. A., Burutolu, T. B., Fling, B. W., Gordon, M. T., Gwin, J. T., Kwak, Y., & Lipps, D. B. (2010). Motor control and aging: Links to age-related brain structural, functional, and biochemical effects. *Neuroscience & Biobehavioral Reviews*, *34*(5), 721-733. https://doi.org/https://doi.org/10.1016/j.neubiorev.2009.10.005

SENIAM. (2021). *The SENIAM project* Retrieved 20 April 2022 from

Shao, X. R., Wang, Z., Luan, L. J., Sheng, Y. L., Yu, R. N., Pranata, A., Adams, R., Zhang, A. R., & Han, J. (2022). Impaired ankle inversion proprioception during walking is associated with fear of falling in older adults. *Frontiers in Aging Neuroscience*, *14*, Article 946509. https://doi.org/10.3389/fnagi.2022.946509

Sherrington, C., Fairhall, N. J., Wallbank, G. K., Tiedemann, A., Michaleff, Z. A., Howard, K., Clemson, L., Hopewell, S., & Lamb, S. E. (2019). Exercise for preventing falls in older people living in the community. *Cochrane Database of Systematic Reviews*, *1*(1), Cd012424. https://doi.org/10.1002/14651858.CD012424.pub2

Shumway-Cook, A., & Woollacott, M. H. (1995). Theory and practical applications. *Motor Control*.

Singh, H., Sanders, O., McCombe Waller, S., Bair, W.-N., Beamer, B., Creath, R. A., & Rogers, M. W. (2017). Relationship Between Head-Turn Gait Speed and Lateral Balance Function in Community-Dwelling Older Adults. *Archives of Physical Medicine and Rehabilitation*, *98*(10), 1955-1961. https://doi.org/https://doi.org/10.1016/j.apmr.2017.05.006

Skelton, D. A., Kennedy, J., & Rutherford, O. M. (2002). Explosive power and asymmetry in leg muscle function in frequent fallers and non-fallers aged over 65. *Age and Ageing*, *31*(2), 119-125. https://doi.org/10.1093/ageing/31.2.119

Slimani, M., Paravlic, A., & Granacher, U. (2018). A Meta-Analysis to Determine Strength Training Related Dose-Response Relationships for Lower-Limb Muscle Power Development in Young Athletes [Systematic Review]. *Frontiers in Physiology*, *9*. https://doi.org/10.3389/fphys.2018.01155

Smith, B. N., Segal, R. L., & Wolf, S. L. (1996). Long latency ankle responses to dynamic perturbation in older fallers and non-fallers. *Journal of the American Geriatrics Society*, *44*(12), 1447-1454. https://doi.org/10.1111/j.1532-5415.1996.tb04069.x

Smith, C. M., Housh, T. J., Hill, E. C., Johnson, G. O., & Schmidt, R. J. (2017). Dynamic versus isometric electromechanical delay in non-fatigued and fatigued muscle: A combined electromyographic, mechanomyographic, and force approach. *Journal of Electromyography and Kinesiology*, *33*, 34-38. https://doi.org/https://doi.org/10.1016/j.jelekin.2017.01.008

Solis‐Escalante, T., Stokkermans, M., Cohen, M. X., & Weerdesteyn, V. (2021). Cortical responses to whole-body balance perturbations index perturbation magnitude and predict reactive stepping behavior. *European Journal of Neuroscience*, *54*(12), 8120-8138.

Studenski, S., & Chandler, J. (1991). Postural Responses and Effector Factors in Persons with Unexplained Falls: Results and Methodologic Issues [Article]. *Journal of the American Geriatrics Society*, *39*(3), 229-234. https://doi.org/10.1111/j.1532-5415.1991.tb01642.x

Sturnieks, D. L., Menant, J., Delbaere, K., Vanrenterghem, J., Rogers, M. W., Fitzpatrick, R. C., & Lord, S. R. (2013). Force-Controlled Balance Perturbations Associated with Falls in Older People: A Prospective Cohort Study [Article]. *PloS One*, *8*(8), Article e70981. https://doi.org/10.1371/journal.pone.0070981

Swanenburg, J., Nevzati, A., Hager, A. G. M., de Bruin, E. D., & Klipstein, A. (2013). The maximal width of the base of support (BSW): Clinical applicability and reliability of a preferred-standing test for measuring the risk of falling. *Archives of Gerontology and Geriatrics*, *57*(2), 204-210. https://doi.org/10.1016/j.archger.2013.04.010

Szturm, T., & Fallang, B. (1998). Effects of varying acceleration of platform translation and toes-up rotations on the pattern and magnitude of balance reactions in humans. *JOURNAL OF VESTIBULAR RESEARCH-EQUILIBRIUM & ORIENTATION*, *8*(5), 381-397.

Tantisuwat, A., Vachalathiti, R., Hiengkaew, V., & Assantachai, P. (2011). Stepping Responses during Forward and Backward Fall Recovery between Thai Elderly Fallers and Non-fallers. *JOURNAL OF PHYSICAL THERAPY SCIENCE*, *23*(3), 373-379. https://doi.org/10.1589/jpts.23.373

Thompson, J. D., Plummer, P., & Franz, J. R. (2018). Age and falls history effects on antagonist leg muscle coactivation during walking with balance perturbations. *Clinical Biomechanics (Bristol, Avon)*, *59*, 94-100. https://doi.org/10.1016/j.clinbiomech.2018.09.011

Tinetti, M. E., Speechley, M., & Ginter, S. F. (1988). Risk factors for falls among elderly persons living in the community. *New England Journal of Medicine*, *319*(26), 1701-1707. https://doi.org/10.1056/nejm198812293192604

Tisserand, R., Robert, T., Chabaud, P., Livet, P., Bonnefoy, M., & Chèze, L. (2015). Comparison between investigations of induced stepping postural responses and voluntary steps to better detect community-dwelling elderly fallers. *Neurophysiologie Clinique/Clinical Neurophysiology*, *45*(4-5), 269-284.

Tong, C. Y., Zhu, R. T., Ling, Y. T., Scheeren, E. M., Lam, F. M., Fu, H., & Ma, C. Z. (2023). Muscular and Kinematic Responses to Unexpected Translational Balance Perturbation: A Pilot Study in Healthy Young Adults. *Bioengineering*, *10*(7).

Torres-Costoso, A., López-Muñoz, P., Martínez-Vizcaíno, V., Álvarez-Bueno, C., & Cavero-Redondo, I. (2020). Association between muscular strength and bone health from children to young adults: a systematic review and meta-analysis. *Sports Medicine*, *50*, 1163-1190.

Tsai, Y.-C., Hsieh, L.-F., & Yang, S. (2014). Age-related changes in posture response under a continuous and unexpected perturbation. *Journal of Biomechanics*, *47*(2), 482-490.

Uchiyama, T., Hori, Y., & Suzuki, K. (2017, 11-15 July 2017). Estimation of tibialis anterior muscle stiffness during the swing phase of walking with various footwear. 2017 39th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC),

Valkovič, P., Brožová, H., Bötzel, K., Růžička, E., & Benetin, J. (2008). Push and release test predicts better Parkinson fallers and nonfallers than the pull test: comparison in OFF and ON medication states. *Movement disorders: official journal of the Movement Disorder Society*, *23*(10), 1453-1457.

Varesco, G., Espeit, L., Feasson, L., Lapole, T., & Rozand, V. (2019). Rate of force development and rapid muscle activation characteristics of knee extensors in very old men. *Experimental Gerontology*, *124*, Article 110640. https://doi.org/10.1016/j.exger.2019.110640

Verniba, D., & Gage, W. H. (2020). A comparison of balance-correcting responses induced with platform-translation and shoulder-pull perturbation methods. *Journal of Biomechanics*, *112*, 110017.

VICON. (2021). *Full body modeling with Plug-in Gait*. Retrieved 20 April 2022 from

Vicon Motion Systems, & UK. (2022). *Full body modeling with Plug-in Gait*. Vicon Motion Systems, Oxford, UK. Retrieved 29 May from https://docs.vicon.com/display/Nexus213/Full+body+modeling+with+Plug-in+Gait

Vicon Motion Systems Limited. (2021). *Plug-in Gait Reference Guide*. https://docs.vicon.com/display/Nexus212/Plug-in+Gait+Reference+Guide

Vind, A. B., Andersen, H. E., Pedersen, K. D., Jørgensen, T., & Schwarz, P. (2009). An Outpatient Multifactorial Falls Prevention Intervention Does Not Reduce Falls in High-Risk Elderly Danes. *Journal of the American Geriatrics Society*, *57*(6), 971-977.

Vlutters, M., van Asseldonk, E. H. F., & van der Kooij, H. (2018). Lower extremity joint-level responses to pelvis perturbation during human walking. *Scientific Reports*, *8*(1), 14621. https://doi.org/10.1038/s41598-018-32839-8

Wan, A. H., Wong, D. W., Ma, C. Z., Zhang, M., & Lee, W. C. (2016). Wearable vibrotactile biofeedback device allowing identification of different floor conditions for lower-limb amputees. *Archives of Physical Medicine and Rehabilitation*, *97*(7), 1210-1213.

Wang, H., Gao, C., Fu, H., Ma, C. Z.-H., Wang, Q., He, Z., & Li, M. (2023). Automated Student Classroom Behaviors' Perception and Identification Using Motion Sensors. *Bioengineering*, *10*(2), 127.

Wang, Y., Wang, S., Liu, X., Lee, A., Pai, Y.-C., & Bhatt, T. (2022). Can a single session of treadmill-based slip training reduce daily life falls in community-dwelling older adults? A randomized controlled trial. *Aging Clinical and Experimental Research*, *34*(7), 1593-1602.

Watt, A. A., Clark, C., & Williams, J. M. (2018). Differences in sit-to-stand, standing sway and stairs between community-dwelling fallers and non-fallers: a review of the literature. *Physical Therapy Reviews*, *23*(4-5), 273-290. https://doi.org/10.1080/10833196.2018.1470748

Westlake, K. P., Johnson, B. P., Creath, R. A., Neff, R. M., & Rogers, M. W. (2016). Influence of non-spatial working memory demands on reach-grasp responses to loss of balance: Effects of age and fall risk. *Gait and Posture*, *45*, 51-55. https://doi.org/10.1016/j.gaitpost.2016.01.007

Winger, M. E., Caserotti, P., Cauley, J. A., Boudreau, R. M., Piva, S. R., Cawthon, P. M., Orwoll, E. S., Ensrud, K. E., Kado, D. M., & Strotmeyer, E. S. (2023). Lower leg power and grip strength are associated with increased fall injury risk in older men: the osteoporotic fractures in men study. *The Journals of Gerontology: Series A*, *78*(3), 479-485.

Winter, D. A., Patla, A. E., Frank, J. S., & Walt, S. E. (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Physical Therapy*, *70*(6), 340-347.

Winter, D. A., Patla, A. E., Prince, F., Ishac, M., & Gielo-Perczak, K. (1998). Stiffness Control of Balance in Quiet Standing. *Journal of Neurophysiology*, *80*(3), 1211-1221. https://doi.org/10.1152/jn.1998.80.3.1211

Woodward, R. B., Stokes, M. J., Shefelbine, S. J., & Vaidyanathan, R. (2019). Segmenting mechanomyography measures of muscle activity phases using inertial data. *Scientific Reports*, *9*(1), 1-10.

Woollacott, M. H. (1993). Age-Related Changes in Posture and Movement. *Journal of Gerontology*, *48*(Special_Issue), 56-60. https://doi.org/10.1093/geronj/48.Special_Issue.56

World Health Organization. (2008). *WHO global report on falls prevention in older age* (ISBN: 9789241563536). https://www.who.int/publications/i/item/9789241563536

World Health Organization. (2021). *Falls*. World Health Organization. Retrieved Apr., 2023 from https://www.who.int/news-room/fact-sheets/detail/falls#

Wyszomierski, S. A., Chambers, A. J., & Cham, R. (2009). Knee Strength Capabilities and Slip Severity. *Journal of Applied Biomechanics*, *25*(2), 140-148. https://doi.org/10.1123/jab.25.2.140

Yang, N. P., Hsu, N. W., Lin, C. H., Chen, H. C., Tsao, H. M., Lo, S. S., & Chou, P. (2018). Relationship between muscle strength and fall episodes among the elderly: the Yilan study, Taiwan. *BMC Geriatrics*, *18*, Article 90. https://doi.org/10.1186/s12877-018-0779-2

Yardley, L., Beyer, N., Hauer, K., Kempen, G., Piot-Ziegler, C., & Todd, C. (2005). Development and initial validation of the Falls Efficacy Scale-International (FES-I). *Age and Ageing*, *34*(6), 614-619. https://doi.org/10.1093/ageing/afi196 %J Age and Ageing

Zemkova, E., Jelen, M., Kovacikova, Z., Miklovic, P., Svoboda, Z., & Janura, M. (2017). Balance Performance During Perturbed Standing Is Not Associated With Muscle Strength and Power in Young Adults. *Journal of motor behavior*, *49*(5), 514-523. https://doi.org/10.1080/00222895.2016.1241751

Zemkova, E., Kovacikova, Z., Jelen, M., Neumannova, K., & Janura, M. (2016). Postural and Trunk Responses to Unexpected Perturbations Depend on the Velocity and Direction of Platform Motion. *Physiological Research*, *65*(5), 769-776. https://doi.org/10.33549/physiolres.933177

Zhang, L., Quan, M., & Cao, Z.-B. (2019). Effect of vitamin D supplementation on upper and lower limb muscle strength and muscle power in athletes: A meta-analysis. *PloS One*, *14*(4), e0215826.

Zhu, R. T.-L., Hung, T. T. M., Lam, F. M. H., Li, J.-Z., Luo, Y.-Y., Sun, J., Wang, S., & Ma, C. Z.-H. (2024). Older Fallers and Non-fallers' Neuromuscular and Kinematic Alterations in Reactive Balance Control: Indicators of Balance Decline or Compensation?

Zhu, R. T.-L., Hung, T. T. M., Lam, F. M. H., Li, J.-Z., Luo, Y.-Y., Sun, J., Wang, S., & Ma, C. Z.-H. (2025). Older Fallers' Comprehensive Neuromuscular and Kinematic Alterations in Reactive Balance Control: Indicators of Balance Decline or Compensation? A Pilot Study. *Bioengineering*, *12*(1), 66.

Zhu, R. T.-L., Lyu, P.-Z., Li, S., Tong, C. Y., Ling, Y. T., & Ma, C. Z.-H. (2022). How Does Lower Limb Respond to Unexpected Balance Perturbations? New Insights from Synchronized Human Kinetics, Kinematics, Muscle Electromyography (EMG) and Mechanomyography (MMG) Data. *Biosensors*, *12*(6), 430. https://www.mdpi.com/2079-6374/12/6/430

Zhu, R. T.-L., Schulte, F. A., Singh, N. B., Ma, C. Z.-H., Easthope, C. A., & Ravi, D. K. (2025). Effects of Single-Session Perturbation-Based Balance Training with Progressive Intensities on Resilience and Dynamic Gait Stability in Healthy Older Adults. *bioRxiv*, 2025.2001. 2005.631356.

Zhu, R. T.-L., Zuo, J. J.-J., Li, K.-J., Lam, F. M. H., Wong, A. Y. L., Yang, L., Bai, X., Wong, M. S., Kwok, T., & Zheng, Y.-P. (2025). Association of lower-limb strength with different fall histories or prospective falls in community-dwelling older people: a systematic review and meta-analysis. *BMC Geriatrics*, *25*, 83.

Zukowski, L. A., Brinkerhoff, S. A., Iyigun, G., Roper, J. A., Giuliani, C. A., & Plummer, P. (2023). Fall history in older adults impacts acceleration profiles after a near collision with a moving pedestrian hazard. *Aging Clinical and Experimental Research*, *35*(3), 621-631. https://doi.org/10.1007/s40520-023-02345-7