

Medio-lateral gait stability and fall risk in older adults

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Abstract

This thesis explores the use of medio-lateral (ML) jerk to assess gait stability of older adults (OAs) with and without a fall history. Current research is unable to identify individuals at risk of falling prior to their first fall. There is also the identification and summary of common laboratory methods used in gait perturbation and responses assessment.

During a 10 metre over ground unperturbed walk, ML jerk was able to discriminate between 834 OAs with and without a fall history in the previous 12 months. There was a small effect size indicating the differences may have been a result of the large cohort size.

A systematic review was conducted to identify and describe the current methods and outcome variables used to apply and report balance perturbations response during gait. Findings showed floor level perturbations during over ground walking are the most representative of exposure to slips and trips in normal walking conditions.

Data was then collected from 31 participants during over ground perturbed walking to apply ML jerk as an outcome variable. There was a significant effect of age group on ML jerk during gait. Further data collection is needed to show conclusively the effectiveness of ML jerk as a fall indicator.

Early identification of individuals at a higher risk of falls would allow for the development of more effective interventions to reduce the risk of falling and the impact of a fall on the individual. Furthermore, reduced falls rates would improve overall health, quality of life and life expectancy.

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Table of abbreviations

η_p^2	partial eta squared
ACL	anterior cruciate ligament
AJ	accelerating jerk
AP	anterior-posterior
AXIS	appraisal tool for cross-sectional studies
BM	body mass
BMI	body mass index
BoS	base of support
CI	confidence interval
cm	centimetre
CoM	centre of mass
CoP	centre of pressure
DJ	decelerating jerk
DoF	degrees of freedom
DT	dual task
g	grams
HS	heel strike
Hz	Hertz
IMU	inertial measurement unit
kg	kilograms
kg.m ⁻²	kilograms per meter squared
LDE	local divergence exponent
LoS	limits of stability
m	metres
mm	millimetres
mins	minutes
m.s	metres per second
m.s ⁻²	metres per second squared
m.s ⁻³	metres per second cubed
ML	medio-lateral
MoS	margin of stability
ms	milliseconds
NHS	National Health Service
OAs	older adults
OG	over ground
PRISMA	preferred reporting items for systematic reviews and meta-analyses
RoM	range of motion
SABRE	Southall and Brent Revisited
ST	single task
step.min ⁻¹	steps per minute
TM	Treadmill
XCoM	extrapolated centre of mass
YA	young adult

Chapter 1 - Introduction

With the increasing life expectancy in the UK and other developed countries there is increasing presentation of individuals with complex health care needs. This is the result of advancing medical sciences and public health systems. Internationally, falls among older adults (OAs) are recognised as a growing concern, impacting health care services, and quality of life (Rogers and Mille, 2003; Liu and Lockhart, 2009; Kao et al., 2015; Ding and Yang, 2016; Roelofs et al., 2019). In 2008 Anderson found serious falls among the elderly in the UK were costing the National Health Service (NHS) £1.7 billion annually, while also observing an increase in both the overall population and the proportion of adults over 65 years-old. In 2012, the direct medical cost per fall was calculated to be \$11,502 (Carande-Kulis et al., 2015).

Falls can also cause social issues such as loss of independence, isolation and reduced quality of life (Nagarajan et al., 2020). Many OAs who go into hospital following a fall have been shown to reduce their social interactions, which can lead to decreased mobility and depression (Nagarajan *et al.*, 2020). These, in turn, can cause deconditioning leading to a further increased risk of future falls (Petersen, König and Hajek, 2020). Approximately 30% of adults over 65 years-old fall at least once per year, while those over 80 years of age have a 50% risk of falling, further increasing the social and economic burden (Carande-Kulis et al., 2015). Of those fallers, 20-30% sustain a moderate to severe injury which in turn increases the chance of admission to health care services (Enderlin et al., 2015). It has been estimated adequate interventions could reduce moderate to severe falls among OAs by 30%. Furthermore, 15% of all falls cause serious injury, wrists and hips being the most common injury sites; notably, 90% of hip and wrist fractures and 60% of head injuries occur as a result of a fall (Anderson, 2008). Thus far there is a lack of scientific measure which can accurately identify those at risk of falling in any age category.

Neural and cognitive decline

Gait and balance require cognitive capacity, therefore, as neural decline occurs with age, so does an individual's functional mobility (Al-Yahya *et al.*, 2019). Total brain volume is seen to decrease with age, increasing risk of mild cognitive impairments (Erten-Lyons et al., 2013; Vitorio et al., 2017). Cognitive ability and agility is known to deteriorate as age increases, however, groups shown to be more physical active such as master athletes have significantly higher performance on cognitive tasks over the age of 75 years (Schott and Krull, 2019). Key findings were: a routine of healthy meals, good sleeping pattern and regular physical activity reduced the rate of mental decline. Researchers emphasise the impact of neurological factors including cognitive function on individuals' risk of falls (Lord, Delbaere and Gandevia, 2016).

Executive function is defined as a set of cognitive skills required to plan control and execute a complex sequence of actions leading to a single goal (Royall *et al.*, 2002). Dual task

conditions, which increase cognitive demand through the addition of a task such as talking, are widely used to add context to gait and balance assessments by increasing the ecological validity of the assessment (Belghali *et al.*, 2017). Performance deficits associated with the addition of a motor or cognitive task are known as dual task interference (Al-Yahya *et al.*, 2019). A study investigated gait speed of over 700 community-dwelling OAs (mean age: 72.7 (5.9) years) while they performed a number of dual task gait conditions. Results showed the association between gait speed and executive function was task dependent, based on the sensorial adaptation required for the performance of the walking tasks (Coppin *et al.*, 2006). Specifically, gait speed was linked to executive function in the normal walking trial, when picking up an object, stepping over objects and during a weighted 60 m walk. Poor executive function was shown to directly impact the cost associated with the addition of carrying and talking tasks during normal walking (Coppin *et al.*, 2006). These results suggest executive function allows individuals to successfully distribute cognitive capacity to ensure successful task execution. Those with lower levels of executive function have shown to not compensate for the dual task conditions, suggesting an increased risk of task or gait failure, however this has not shown to be a conclusive indicator of fall risk.

Neuromuscular decline

From the 30th year of life there is a gradual, continual decline in muscle, neuromuscular performance and bone strength as well as a reduced physical capacity, and increased adipose tissue (Keller and Engelhardt, 2013; Trombetti *et al.*, 2016). Following the 50th year of life it has been observed humans experience a loss of 1-2% of lean body mass and 1.5-5% of strength per year (Keller and Engelhardt, 2013; Alcazar *et al.*, 2018). Decreased anabolic hormone production is one factor which reduces OAs ability to build and retain muscle strength and muscle mass (Keller and Engelhardt, 2013). Further evidence of decreased muscle strength is shown by the increased energy cost of walking as age increases (Wert *et al.*, 2010). OAs have been observed to walk more slowly to compensate for reduced strength, endurance and proprioceptive capacities (Brodie *et al.*, 2014).

Humans have evolved the ability to walk over almost any terrain regardless of texture and diversity (Santuz *et al.*, 2018). Research has shown a decrease in dynamic stability during gait as a result of destabilising environments (Hak *et al.*, 2012). Further findings show when experiencing medio-lateral (ML) perturbation participants choose to modify spatio-temporal gait parameters over centre of mass (CoM) velocity (Hak *et al.*, 2012). To overcome unpredictable or uneven terrains, humans continually adapt their locomotor patterns, which increases the load on the control system (Santuz *et al.*, 2018). As humans age there is a reduced proprioceptive capacity and increased processing time of afferent information (Bierbaum *et al.*, 2010). This, combined with a balance perturbation, will cause an even greater cognitive load and increase the risk of falls among OAs in comparison to younger adults (YAs). As already discussed, as humans age there are significant decreases in both

muscle mass and strength which, in turn, leads to reduced functional mobility, one area particularly affected is balance (Cruz-Jentoft *et al.*, 2010).

Comorbidities and general health

There are several components to physical deterioration as a result of ageing (Lord, Delbaere and Gandevia, 2016). Lord and colleagues emphasise the importance of other health factors on fall risk such as cognitive function and sight loss. Further research has shown diseases such as dementia and acute macular degeneration along with a history of polio or stroke all have significant negative impacts on individuals risk of falling (Lord, Delbaere and Gandevia, 2016). Increased prevalence of comorbidities of chronic diseases such as acute macular degeneration, stroke, dementia and multiple sclerosis, are likely to be contributing to increasing fall prevalence (Anderson, 2008). There are a number of clinical conditions which attribute to increased difficulty maintaining stability, such as vertigo, visual impairment, arthritis, postural hypertension and cognitive impairment. Furthermore, there are a range of medications which contribute to falls by causing sedation, confusion, unsteadiness or hypotension (Anderson, 2008).

All risk factors for falling can be divided into internal and external categories, internal risk factors being those associated with age and health problems or side effects of medication, while external factors are those of the environment which can be limited with correct resources and knowledge, such as appropriate foot wear and avoiding destabilising or uneven surfaces (Enderlin *et al.*, 2015). It is rare for any of these risk factors to occur in isolation, often falls are caused by the culmination of multiple risk factors.

Gait stability

Gait is a cycle which can be divided into two main stages; the stance and swing phase. The stance phase is further divided into: loading response; mid-stance; terminal stance and pre-swing. The swing phase is divided into: initial swing; mid-swing and terminal swing (Gage, 1990). Human gait typically develops in the first year of life, with gait maturation occurring around 7 years-old. Gait is maintained - unless substantial injury or illness occurs – on average until 60 years (Prince *et al.*, 1997). Many factors contribute to the development of individual gait patterns and deterioration. Much of the deterioration is attributed to reduced mobility, muscle mass and neuromuscular function which occurs as part of the natural ageing process (Siparsky, Kirkendall and Garrett, 2014). These deteriorations have a negative effect on daily functioning and can increase individuals' instability. During gait the CoM remains within the individual's limits of stability (LoS). A loss of balance occurs when the CoM strays outside of the LoS. There are then two possible outcomes; depending on the acceleration, direction and displacement of the CoM the individual may recover by either gaining control of the CoM or moving the base of support (BoS) to bring the CoM back within the LoS (Pai and Patton, 1997). This is known as recovery equilibrium; if not achieved the other outcome is a fall.

OAs most commonly experience falls during motion rather than from static positions. This highlights the need for dynamic measures of stability as this is when individuals are most at risk and stability is most compromised (Hamacher *et al.*, 2011). During gait and other dynamic exercises such as turning, constant adjustments to the BoS and CoM are needed to maintain balance and keep CoM motions efficient (Hamacher *et al.*, 2011). There is a wide variety of methods used to measure dynamic stability including but not limited to: CoM velocity, harmonic ratio, local divergence exponent, Lyapunov exponent, margins of stability and multiscale entropy (Riva, Bisi and Stagni, 2014; McIntosh, Zettel and Vallis, 2017; Rieger *et al.*, 2020).

Balance ability is observed to decrease as age increases. This is shown by a number of measures in different studies. As age increases, CoM sway and velocity also increases (Roman-Liu, 2018). OAs have been seen to walk with reduced harmonic ratios compared to YAs (Lowry, Lokenvitz and Smiley-Oyen, 2012). Lyapunov exponent as a metric measure of balance ability is also seen to reduce as age increases (Liu, Wang and Xiao, 2015). Age reduces postural stability shown by reduced margins of stability (Slobounov, Haibach and Newell, 2006). Local divergence exponents has also been shown to correlate to kinematics and electromyography, showing greater instability among OAs compared to YAs (Kang and Dingwell, 2009). Multiscale entropy as a measure of dynamic stability also agrees in finding OAs are less stable during dynamic tasks (Kosciessa, Kloosterman and Garrett, 2020). As always, depending upon the aim of the research, different methods of measurement are selected, some of which have shown excellent reliability and correlations with fall history (Riva, Bisi and Stagni, 2014).

Importance of medio-lateral stability of walking gait

Previous research has shown the effectiveness of the inverted pendulum model at predicting lower trunk displacement (Zijlstra and Hof, 2003). The inverted pendulum model is widely accepted as an explanation of lower limb and CoM motion during healthy gait (Vlutters, Van Asseldonk and Van Der Kooij, 2016; Bruijn and van Dieen, 2018). However, this model does not consider the frontal plane of motion, only the sagittal plane (González *et al.*, 2007). Research has stated the inverted pendulum model is too simplistic to predict CoM and CoP displacement during gait, therefore cannot be used as a reliable measure of dynamic stability (Vlutters, Van Asseldonk and Van Der Kooij, 2016). Previous research has shown moderate to weak association to other known fall risk factors in sitting and standing tasks (Regterschot *et al.*, 2015). This further supports the need for a multidirectional approach to falls assessment research. There is a general consensus on the need for additional measures outside of spatio-temporal parameters in order to successfully distinguish fallers from non-fallers (Mortaza, Abu Osman and Mehdikhani, 2014). As a result, researchers have adopted dynamic measures of stability which aim to incorporate the different elements of gait required to maintain stability, particularly in the ML direction

(Bierbaum *et al.*, 2010; Bhatt *et al.*, 2011; Chulvi-Medrano, Martínez-Ballester and Masiá-Tortosa, 2015; Ghahramani, 2016).

Studies have shown ML balance impairment measured by centre of pressure (CoP) and spontaneous sway is a predictor of future falls risk (Fallers *et al.*, 2000). Furthermore, ML balance assessments have shown to be an ecologically valid tool for balance assessments in community dwelling OAs showing minor balance impairments (Eduardo Cofré Lizama *et al.*, 2015). This study suggested ML balance assessments are more accurate and able to detect more subtle changes in an individual's stability compared with anterior-posterior (AP) measures. During gait the CoM sways from left to right. This motion needs to be controlled by decelerating and accelerating the CoM with each foot strike and push off, while also propelling the CoM forward (Pai and Patton, 1997). The gluteus muscle group is key for gait production and the maintenance of stability due to the control over the pelvis they provide (Chulvi-Medrano, Martínez-Ballester and Masiá-Tortosa, 2015). The glute muscles also provide a layer of protective tissue over and around the pelvis against impact. Among OAs, those who displayed significantly higher hip abduction strength and power were better able to overcome ML balance perturbations during gait (Addison *et al.*, 2017). This has also been shown to apply to real world settings by a reduction in the rate of falls per year (Inacio, Creath and Rodgers, 2018).

The inverted pendulum model explains lower limb motion in the sagittal plane and highlights some of the gait complexities. However, the model assumes linearism and does not account for the ML aspect of foot placement (Bruijn and van Dieën, 2018). Previous research has also stated the inverted pendulum model is too simplistic to predict CoM motion due to the lack of consideration of active control (Vlutters, Van Asseldonk and Van Der Kooij, 2016).

Perturbation of walking gait

The frequency of falls among over 65 year-olds, with 30% falling at least once a year is not sufficient to collect data through real world observations (Anderson, 2008). For example, one study conducted in a care home recorded and analysed falls experienced in communal areas by 130 participants over a 38 month period (Robinovitch *et al.*, 2013). The aim was to identify the cause of falls in an attempt to reduce the risk to residents. However, over the 102,830 participant days data was collected, only 227 falls were recorded.

To overcome the difficulties associated with observing real world falls many efforts have been made to recreate a perturbation in a laboratory setting such that participant responses can be recorded and analysed more accurately. Some methods have used treadmills and altered the belt speed while the participant is walking (Madehkhaksar *et al.*, 2018; Roeles *et al.*, 2018; Onushko *et al.*, 2019), while others have used a device attached to participants to push and pull the participant as they walk (Martelli, Kang and Agrawal, 2017; Zadravec, Olenšek and Matjačić, 2017; Hof and Duysens, 2018a). Further methods have used floor-

level moveable platforms which participants walk over (Capin et al., 2017; McIntosh, Zettel and Vallis, 2017; Shulman, Spencer and Vallis, 2018).

Recent developments in falls research has shown promise, with one study suggesting participant reactions to perturbation testing has been shown to be one of the best predictors of a future fall (Coelho and Teixeira, 2017). Previous research has shown significant relationships between knee extensor peak torque values and successful recovery during AP treadmill perturbations in a laboratory setting (Ding and Yang, 2016). The same study also reported significance of the same variables compared to knee flexion. There was also a Cohen's d effect size of 0.947 with knee extension and falling and 0.689 with flexion (Ding and Yang, 2016). This shows leg strength and power have a stronger relationship to successful perturbation recovery among OAs in a laboratory setting. However, there is a lack of research regarding the reliability of laboratory perturbations to real world perturbations.

There is currently a lack of consistency regarding perturbation studies, in particular the methods used to perturb participants and the point in the gait cycle the perturbation is delivered (Lesinski *et al.*, 2015). Previous inconsistencies in results suggest perturbation type and timing could have an impact on outcome variables (Rabago, Dingwell and Wilken, 2015; Martelli, Kang and Agrawal, 2017; Huijben *et al.*, 2018; Madehkhaksar *et al.*, 2018; Kongsuk, Brown and Hurt, 2019; Rieger *et al.*, 2020).

Role of neuromuscular function in walking stability

During normal walking, all age groups display the dominant joint moments occurring in the sagittal plane, however, when gait is challenged OAs adopt a dominant joint moment in the frontal plane (Liu and Lockhart, 2009). This suggests that, as an individual's functional level begins to decline, there are compensatory mechanisms used which place a greater load on the ML stabilising muscles of the trunk and pelvis (Chulvi-Medrano, Martínez-Ballester and Masiá-Tortosa, 2015). A further example of the importance of ML stabilisation during normal gait is the condition known as Trendelenburg gait. This is a disorder of the hip abductor muscles either not activating correctly or being too weak to support the torso (Gandbhir, Lam and Rayi, 2022). This weakness causes the pelvis to drop during gait to the contralateral side causing discomfort, increased energy expenditure during gait and increased fall risk (Gandbhir, Lam and Rayi, 2022).

Research by Addison and colleagues has found differences in movements when comparing age groups (Addison et al., 2017). These differences showed significant links to participant's strength in the frontal plane lower limb movement, with stronger participants choosing a more demanding but also more stable stance when perturbed (Addison et al., 2017). Ageing reduces muscle neuromuscular performance, reducing individuals' ability to support themselves in more strenuous recovery positions after experiencing a perturbation (Ding and Yang, 2016).

Hip abduction and adduction strength has shown a significant link to increased successful response to lateral perturbation among OAs (Addison et al., 2017). This was a laboratory study conducted on 40 inactive but healthy over 65 year-olds. The researchers investigated the methods of recovery, dividing participants in to cross-steppers or medial-steppers. Cross steppers were participants who crossed their legs either in front or behind them to regain balance, medial steppers brought their feet together without crossing over in a side-step motion. Medial-steppers experienced a 38% higher rate of falls compared to cross-steppers. Cross-stepping was more common, more stable and cross-steppers had greater high-density lean tissue between the patella and L4. Participants were also tested in their peak hip-abduction torque which was also higher in the cross-steppers.

Addison and co-workers' findings are some of the first of this nature, suggesting that peak torque and power is associated with increased gait stability rather than merely strength. These findings were later supported by an eight-week intervention comparing power and strength training among 18 healthy participants over 70 years-old. These findings not only showed power training to be more effective than strength training in peak torque production, but also that power training of the hip, abduction and adduction muscles was more successful in real life application, resulting in a reduction in median falls per year from three to zero (Inacio, Creath and Rogers, 2018).

The most common method of examining the effect of age on postural stability is currently measuring postural sway area, this one variable does not consider frequency or velocity of CoM motion (Park *et al.*, 2016). Balance is extremely complex and with previous research only measuring area to quantify balance this simplifies a process which is extremely complex and using multiple control systems. Jerk as a third derivative of acceleration is a relatively new method of assessing and monitoring dynamic stability in gait (Brodie *et al.*, 2014). A recent study used jerk as a measure of smoothness during different levels of loaded walking (Krammer, Drew and Brown, 2021). Results showed that, as participants fatigued, there was a significant decrease in the smoothness of motion, showing a decrease in neuromuscular control by an increase in jerk values (Krammer, Drew and Brown, 2021). Further findings suggest that, as age increased, ML jerk increases whereas vertical jerk decreases (Brodie *et al.*, 2014). This further supports the link between increased jerk and decreased neuromuscular control, as vertical jerk increases with gait speed and ML jerk increases with instability. Results showed jerk to be highly reliable in determining a given participant's age, jerk was normally distributed and independent of stature and gender. The conclusion was that, as age increased, the ML jerk was an indicator of changes in dynamic stability, while vertical jerk was an indicator of gait vigour (Brodie *et al.*, 2014). Researchers stated that the smoother jerk profiles of the younger participants was a result of their ability to rely on the automatic control of dynamic stability, while the OAs had to self-regulate to a larger extent (Brodie *et al.*, 2014).

Aims of the thesis

This thesis aims to explore the utility and application of jerk, a relatively novel variable collected using a tri-axial accelerometer, to identify participants with and without a fall history among an older population, as a marker of gait stability. It also aims to identify and summarise the common methods of laboratory gait perturbation used to assess participant response and the measures used. Finally, it aims to apply the methods learned in a laboratory setting to assess participant's response to perturbation during over ground walking within a healthy young and older adult populations.

Chapter 2 – Medio-lateral jerk during unperturbed walking as a discriminant of fall risk in older adults

Introduction

Falls are the leading cause of injury among over 65 year-olds within developed countries, accountable for 30% of all non-fatal injuries (NICE, 2013). As humans age, proprioceptive capacity reduces and processing speed of afferent information slows (Bierbaum *et al.*, 2010). Whilst there is significant evidence indicating that higher levels of physical and mental activity reduces the rate of this neurological deterioration; researchers have yet to establish a successful method of preventing the neurological degeneration associated with ageing (Schott and Krull, 2019). While some healthy OAs have shown similar movement patterns when recovering from a perturbation to YAs, the success of the movement is not comparable (Riva *et al.*, 2019). However, it has been stated that most commonly there is significant variation in the spatio-temporal variables as age increases such as reduced step length, increased double support time and cadence (Riva *et al.*, 2019). The single stance phase of gait accounts for 80% of a healthy gait cycle and is shown to decrease among OA as a result of ageing and increased fear of falling.

When participants aged 65 years and above (mean: 74 (1.3) years) were assessed in their frontal plane lower limb strength, a significant relationship was found between increased strength and successful recovery from ML perturbation (Addison *et al.*, 2017). This has been further supported by previous findings on the importance of lateral stability in reducing the fall risk among OAs during daily tasks (Rogers and Mille, 2003). Ageing causes reduced maintenance of lean muscle mass which allows individuals to support themselves in the recovery positions frequently experienced when subjected to a perturbation (Bierbaum *et al.*, 2010). Those who show significant weakness find it more difficult to control CoM to maintain stability (Puniello, McGibbon and Krebs, 2000). As humans age, response time to stimuli increases; delayed response time results in a larger compensatory motion needed to regain balance. If an individual is not strong or stable enough to elicit a successful response they will fall (Kongsuk, Brown and Hurt, 2019). However, there is yet a threshold to determine when an individual will be at a high fall risk.

The understanding of the importance of lateral stability in OAs is not a recent breakthrough; a 2003 study found that lateral instability was a key factor in OAs falls (Rogers and Mille, 2003). Moreover, there was a suggestion of peak ML hip forces to be targeted by therapists in rehabilitation and prevention interventions. Hip abduction and adduction strength has shown a significant link to increased successful response to lateral perturbation in OAs (Addison *et al.*, 2017). The previous study was laboratory-based and was conducted on 40 inactive but healthy over 65 year-olds. The researchers investigated the methods of recovery by dividing participants in to cross-steppers and medial-steppers. A cross-stepper

was defined as an individual crossing one leg over the other to recover from perturbation, while a medial-stepper was an individual who brought their feet together. Cross-steppers were more prevalent, displayed an increased stability and had greater high-density muscles mass between the patella and the fourth lumbar vertebra (L4). Those who medial-stepped had 38% greater fall risk compared to cross-steppers. Participants were also tested in their peak hip-abduction torque which was also higher in the cross-steppers. The additional findings are some of the first of this nature, suggesting that peak torque and power is associated with increased stability rather than merely strength (Addison *et al.*, 2017). Yet there is currently no intervention process should an individual show weekend hip abductors. Kinematic measures in jerk units (m.s^{-3}) have been used to represent the smoothness of movement as a higher jerk implies a less smooth motion. If a motion is not smooth, it is more difficult to control and coordinate, leading to the interpretation of deteriorating control mechanisms (Puniello, McGibbon and Krebs, 2000). The gait results of 149 healthy non-fallers (mean age: 57.7 (17) years) in the 6th and 7th decade of life began to show gait balance decline. Yet sway frequency and stride length also show linear deterioration from 21 years-old (Park *et al.*, 2016). Sway has been used by previous researchers as a measure of postural motion and has been found to be a reliable measure of dynamic postural stability during walking (Lee *et al.*, 2014). Supporting results from laboratory walking assessments show increased AP jerk, decreased ML jerk and increased ML sway values as age increased (Park *et al.*, 2016). Further studies have supported these findings, showing AP angular displacement at baseline is associated with a 64% increased fall risk when assessed over a 12-month period (Mahoney *et al.*, 2017).

There has been an absence of research into CoM jerk specifically in association with fall risk or fall history. Measures of CoM jerk can be a powerful tool in the assessment of gait stability and neurological control among OAs. Unlike common measures of stability and neuromuscular control of gait -such as local divergence exponents and multiscale entropy - which require measurements taken from hundreds of consecutive strides, jerk allows for the assessment of stability from periods of walking as short as 10 m.

The aim of this study was to use CoM ML jerk during normal walking to discriminate between those with a fall history in the last 12 months and those without, among a large cohort of OAs. It was hypothesised that ML jerk values would be significantly higher among fallers compared to non-fallers. It was also hypothesised that a regression model would be able to distinguish fallers from non-fallers using ML CoM jerk.

Methods

Participants

A total of 867 participants completed the study (44.3% female, 55.7% male). This study was the third stage (V3) of a larger observational cohort and cross-sectional study known as Southall and Brent Revisited (SABRE). The SABRE study involved participants from three

different ethnic groups (European, South Asian and North African/Caribbean). The first stage (V1) occurred 1988-91. For this, 2346 Europeans, 1710 South Asians, and 801 African Caribbean origin from first generation of migrants were recruited. The second stage of study (V2) was a 20-year follow-up and took place 2008-11. SABRE survivors, then aged around 75 years (interquartile range 69-80), were invited for clinical investigation (V3, 2014-17) and/or completed a health and lifestyle questionnaire to allow access to their health-related records. For this stage, researchers requested updated addresses and GP details for surviving members of the SABRE study, using participant's NHS numbers, for those participants who consented to future contact or follow-up. The data used here was collected from an instrumented 10-metre walk test completed by participants in the V3 stage of the study. This study was not investigating ethnicity, so all participants were analysed as a single data group. All processing here was by secondary analysis. Data sets were excluded due to incomplete data sets or use of a mobility aid during the gait assessment.

Procedures and data analysis

During V3, participants performed an instrumented 10 metre walk. Data was collected using a LPMS control inertial measurement unit (IMU) attached to the skin at the 4th Lumbar vertebra with double adhesive tape. The IMU sampled at 100Hz. Using a custom MATLAB program (Mathworks, MA, USA), linear acceleration calculated from an IMU (LP-RESEARCH, Tokyo, Japan) was filtered using a 4th order Butterworth band-pass filter (0.2-15Hz) (Chidean *et al.*, 2018; Shahabpoor and Pavic, 2018).

Heel strike is defined as the first point of heel to ground contact during gait (Wu *et al.*, 2021). Heel strike was defined here using the zero-crossing method where foot contact was defined as the instant where the forward acceleration crosses the zero value and becomes deceleration (Zijlstra and Hof, 2003). ML jerk was calculated as the average of the rate of change in ML acceleration from 200ms prior to peak CoM lateral relative position to the instant of peak CoM lateral relative position for each step, as well as from the instant of peak ML relative position to 200ms after. The average of the values prior to CoM ML peak displacement were labelled the ML jerk deceleration, the values after CoM ML peak displacement were labelled ML jerk acceleration. CoM ML peak acceleration and deceleration were the independent variables, with falls in the last 12 months used as the dependent variable.

Validity of the IMU

The IMU is extremely divers as it is small (39x39x8 mm, 12g) and wireless, connecting to a laptop via Bluetooth. This allows testing to occur in many different environments and wearing of the device does not interfere with an individual's gait. Findings showed the accuracy of the IMU to be critical for the use of a quaternion rotation matrix in combination with integration approach to transform CoM accelerations (Esser *et al.*, 2009). Findings showed IMUs can be used to acquire spatio-temporal measurements in human walking in

clinical settings leading to more accurate analysis of human gait. This method of IMU analysis was validate in 2009 (Esser *et al.*, 2009).

Statistical analysis

Following extraction of the data, mean and standard deviation values were calculated and tested for normality via Q-Q plots. A one way ANCOVA was performed to identify any group effect in the jerk results when grouped by fall history. Partial eta squared (η_p^2) was used to describe the effect size with $\eta_p^2 = 0.01$, $\eta_p^2 = 0.06$ or $\eta_p^2 = 0.14$ representing a small, moderate and large effect sizes respectively (Lakens, 2013). A binomial logistic regression was performed to identify the probability of jerk being able to distinguish between fallers and non-fallers. Height (m), body mass (kg), age (years) and sex were covariates and the acceptance level of significant of $p < 0.05$ was used in all statistical tests.

Results

Following exclusion ($n = 33$) due to incomplete data sets or use of a mobility aid during the gait assessment, 834 participants were included in the analysis (male = 463, female = 371). Of the 834 participants (age: 72.9 (6.6) years, height: 1.65 (0.09) m, body mass: 76.2 (14.0) kg) 114 (13.6%) had a fall in the 12 months prior to data collection, resulting in a classification as a faller (age: 73.8 (6.9) years, height: 1.63 (0.08) m, body mass: 75.1 (13.3) kg), while the remaining 720 had not experienced any falls in the 12 months prior to data collection (age: 72.8 (6.6) years, height: 1.66 (0.09) m, body mass: 76.4 (14.1) kg).

Data was checked for normality via Q-Q plots (see appendix A), on which normality was assumed. The ANCOVA results for accelerating jerk showed there was a significantly lower accelerating jerk among fallers compared with non-fallers ($t(1) = 4.160$, $p = 0.048$) however the effect size was small $\eta_p^2 = 0.005$. The same test was performed for the decelerating jerk, finding fallers had significantly lower jerk values again ($t(1) = 3.806$, $p = 0.049$), effect size $\eta_p^2 = 0.005$. Jerk values by fall history are given in figure 1. A binomial logistic regression showed that accelerating and decelerating jerk did not significantly classify fallers and non-fallers ($X^2(8) = 3.450$, $p = 0.903$). The model explained 3.8% (R^2) variation in fall history.

Discussion

The aim of this study was to identify whether ML CoM jerk during a 10 meter walk could discriminate between fallers and non-fallers using a secondary data set. The binomial logistic regression found no significant results, while the ANCOVA found ML jerk was higher among non-fallers compared to fallers. When only the jerk variables were used as predictors, ML CoM jerk was not able to classify fallers from non-fallers using the binary logistic regression model. However, when age, height and body mass were controlled for, there were significant differences between fallers' and non-fallers' accelerating and decelerating ML CoM jerk among OAs during a 10 meter walk test, which is novel. OAs with reduced gait stability are known to walk more conservatively by reducing gait speed, step length and increasing step width (Huijben *et al.*, 2018; Qiao, Feld and Franz, 2018;

Belvederi Murri *et al.*, 2020). This gait style supports the results found here that fallers show increased active control of CoM in the ML direction compared to non-fallers, resulting in smoother ML CoM motion and lower jerk values during unperturbed over-ground walking.

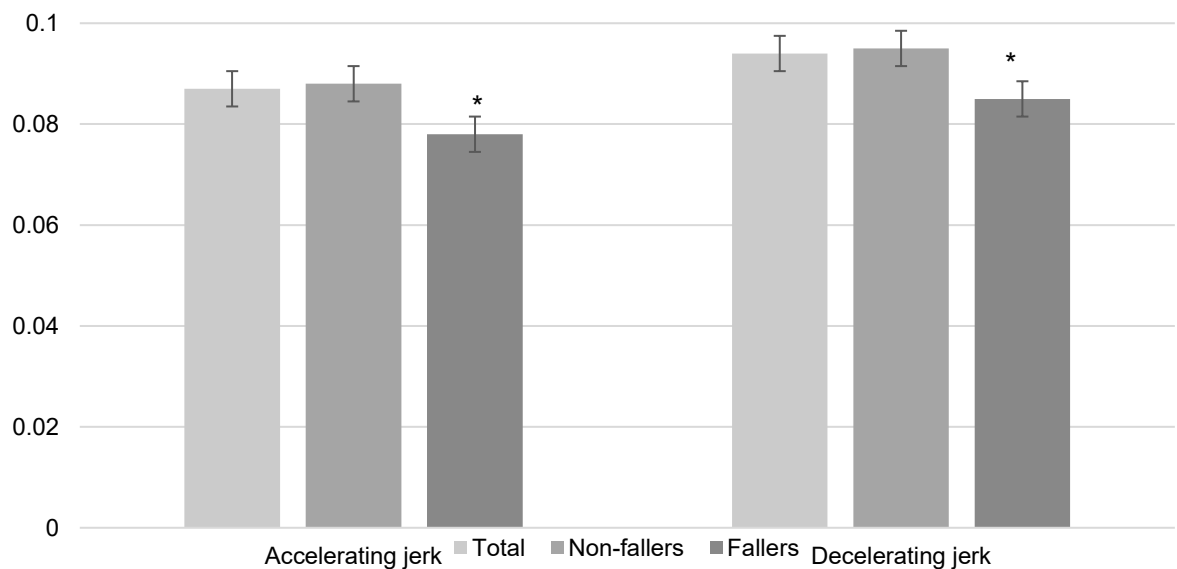


Figure 1 Accelerating and decelerating mean ML CoM jerk values for fallers and non-fallers (*indicates significantly different from non-fallers)

It was concluded that the significance between fallers and non-fallers was due to a large cohort size based on the effect size which may have caused a false positive result. This may be partially due to the age distribution of participants, a mean age of 72.9 years but a standard deviation of only 6.6 years, showing all participants are largely within a single decade of one another. Had there been a younger cohort to compare jerk values against, there may have been a significant effect of age due to the wider range. Sex was controlled for in the statistical tests as there is a higher fall risk among females compared to males (Schulz, Ashton-Miller and Alexander, 2007; Hull, Kneebone and Farquharson, 2013; Kiami, Sky and Goodgold, 2019).

Another explanation for the small effect size seen could be the health and lifestyle of the cohort. All participants were community dwelling and generally in good health. Further research into 427 community dwelling and institutionalised OAs found community dwellers to be significantly less likely to have experienced a fall and showed lower risk of future falls compared institutionalised OAs (Susilowati *et al.*, 2020). Participants in this study were all able to live independently, showing a higher level of functioning, this is particularly important for the older participants. While the sample size was large, the range of data collected was small due to the lack of diversity in health, age and independence of the participants. Had the gait of these participants been challenged by a perturbation during gait, there may have been alternative findings.

The previous explanations above relating to small effect sizes in group differences could also contribute to the inability of the logistic regression model to significantly classify fallers and non-fallers. Additionally, previous research has demonstrated that the causes of falls

are multifactorial and occur mostly during dynamic challenging movements (Enderlin *et al.*, 2015). However, in this study participants were all walking in a single task, unperturbed condition for 10 metres over ground. Additionally, as all participants walked unaided; this was not a gait challenging situation. Therefore, the chosen model in this study only captures a limited amount of information about the balance ability of the participants and does not factor in other external or internal factors which would be independent of the CoM jerk but contribute to the causes of falls. Due to the general good health of this population, a bout of 10 metres of normal walking is not sufficient to challenge the balance of the participants in order to elicit a significant response for this variable, so alternative testing may be needed to allow more reliable identification of fallers.

A 10 metre walk test is shown to be a valid and reliable standard measure of mobility (Peters, Fritz and Krotish, 2013). This is a standard clinical assessment used in a variety of populations to assess functional mobility and vestibular function (Watson, 2002; Chan and Pin, 2019). However, given the relative health status and independence of the two groups in this study, it is likely that a single task, over ground 10 meter walking test in controlled conditions was not challenging enough to elicit group differences in ML stability and thus large effects on the ML jerk. Previous research has been able to obtain a better understanding of functional balance from perturbed gait rather than normal walking (Havens, Mukherjee and Finley, 2018). This is likely due to the large balance adjustment needed to recover from a perturbation during gait which would be detectable by measuring CoM ML jerk. Earlier findings have shown a higher sensitivity during gait to ML perturbations compared to AP perturbations in the same study and highlighted the importance of ML perturbations to challenge gait (Madehkhaksar *et al.*, 2018). Consequently the next chapter will explore perturbation recovery and assessment methods, looking at both YAs and OAs populations.

Conclusion

ANCOVA results showed accelerating and decelerating ML jerk values were significantly higher among non-fallers compared to fallers during a 10 meter over ground walk. However, the binomial logistic regression did not find any effect. Both of the researchers' hypotheses were shown to be incorrect. The small effect size seen in the ANCOVA results between fallers and non-fallers (in their accelerating and decelerating ML jerk during normal walking) indicated the significance is likely due to the large sample size. The population of this study are very similar demographically, and their gait was unperturbed during data collection. This has led researchers to suggest a wider range of participants could produce more meaningful results, and also that gait must be challenged in order to elicit an altered gait response when aiming to determine fallers from non-fallers.

Chapter 3: Gait perturbation methods and outcome measures: A systematic review

Introduction

Balance capacity assessment is often performed with the use of perturbation testing (Bhatt *et al.*, 2011). Assessing balance is key in fall risk assessment, particularly among OAs over the age of 65 years, with 30% of OAs and 50% of those over the age of 80 years experiencing a fall every year (NICE, 2013). As fall risk is linked to balance capacity, in recent years researchers have developed a variety of methods of perturbing participants in order to understand balance and recovery mechanisms (Hof, Vermerris and Gjaltema, 2010; Chien and Hsu, 2018; Zadavec *et al.*, 2020). In the context of this review, a balance perturbation refers to an external action applied to a participant causing disruption to balance control (Rogers and Mille, 2016).

Various approaches have been developed to recreate a balance perturbation and assess balance responses in a safe, controlled environment. This is achieved by controlled mechanical perturbations being applied to participants to aid clinical investigation of risk factors to future fallers (Bhatt *et al.*, 2011). These perturbations have taken many forms, including visual field oscillations, pulls and pushes applied to the waist, and floor level translations (Martelli *et al.*, 2016; Thompson and Franz, 2017; Vervoort *et al.*, 2020). An individual's ability to produce an appropriate mechanical response to these perturbations can reduce fall risk (Zadavec *et al.*, 2020). In addition to perturbation characteristics, this review collected spatio-temporal, kinetic, kinematic and muscle activation outcome variables, all of which were used in reporting participant's responses to balance perturbations.

Previous reviews have focused on the effectiveness of perturbation training in reducing fall frequency, or improving balance perturbation response amongst OAs and clinical populations (McCrum *et al.*, 2017; Mansfield *et al.*, 2018; Olson, Lockhart and Lieberman, 2019). With previous research having identified significant diversity in perturbation methods, there is currently an insufficient discussion of the methods used in the application of the balance perturbations (Lesinski *et al.*, 2015). Therefore, the aim of this review was to identify and describe the current methods and outcome variables used to apply and report participant's responses to balance perturbations during gait across all adult populations.

Methods

Search strategy

This review is registered with PROSPERO, reference number CRD42020211876. Web of Science, CINAHL and PubMed online databases were searched, using the following search terms: (fall* OR perturb* OR slip* OR trip*) AND (adult*) AND (mediolateral OR medio lateral OR medio-lateral OR frontal plane) AND (gait OR walk* OR locomotion). The final search

was conducted on 18th February 2021. Once duplicates were removed, remaining article titles and abstracts were screened against the eligibility criteria. Following exclusions full texts of the remaining titles were retrieved and were also screened against the eligibility criteria. At each stage of screening, all articles were cross-checked by a second reviewer, whereby agreement was $\geq 95\%$ and any discrepancies were resolved between reviewers.

Eligibility criteria

The inclusion criteria included 1) participants aged 18 years or older, 2) participants were subjected to mechanical balance perturbation during gait, 3) outcome variables reported which included kinetic, kinematic, spatio-temporal or muscle activation, 4) perturbation methods were described, 5) peer reviewed articles published during or after 2015 and 6) published in English. 2015 was selected as the cut-off date as the most recent and relevant methods and measures were of interest to the investigation.

Exclusion criteria included 1) all participants under the age of 18 years, 2) where balance perturbations did not occur during continuous gait e.g. standing, at gait initiation, during a single stepping task, reaching or another specific movement scenario; 3) where there was no perturbation reported and 4) observational studies, case studies or literature reviews.

Risk of bias

The Appraisal tool for Cross-sectional Studies (AXIS) risk of bias assessment tool was used to assess potential risks of bias in the methods and outcome variables reported (Downes *et al.*, 2016). All included studies were assessed using 20 questions within the categories of: introduction, methods, results, discussion and other.

Data extraction

Data synthesis was limited to a qualitative evaluation of balance perturbation methods and outcome variables. Participant data extracted included demographic parameters (number, age, sex, height, body mass, body mass index (BMI) and fall history) and any stated comorbidities. Method-specific variables extracted included locomotion type (e.g. treadmill or over ground walking and gait speed), balance perturbation type and details (timing within gait cycle, direction, acceleration, frequency, force, intensity, mode of application) balance perturbation, acclimation procedures, and gait outcome variables.

Results

Search results

Searches of the online databases produced 378 records, with a further 148 records identified from screening records reference lists included within the database. This was reduced to 327 records following duplicate removal. Screening of search results identified 26 studies that met the inclusion criteria. A PRISMA flow diagram of study selection is given in Figure 2.

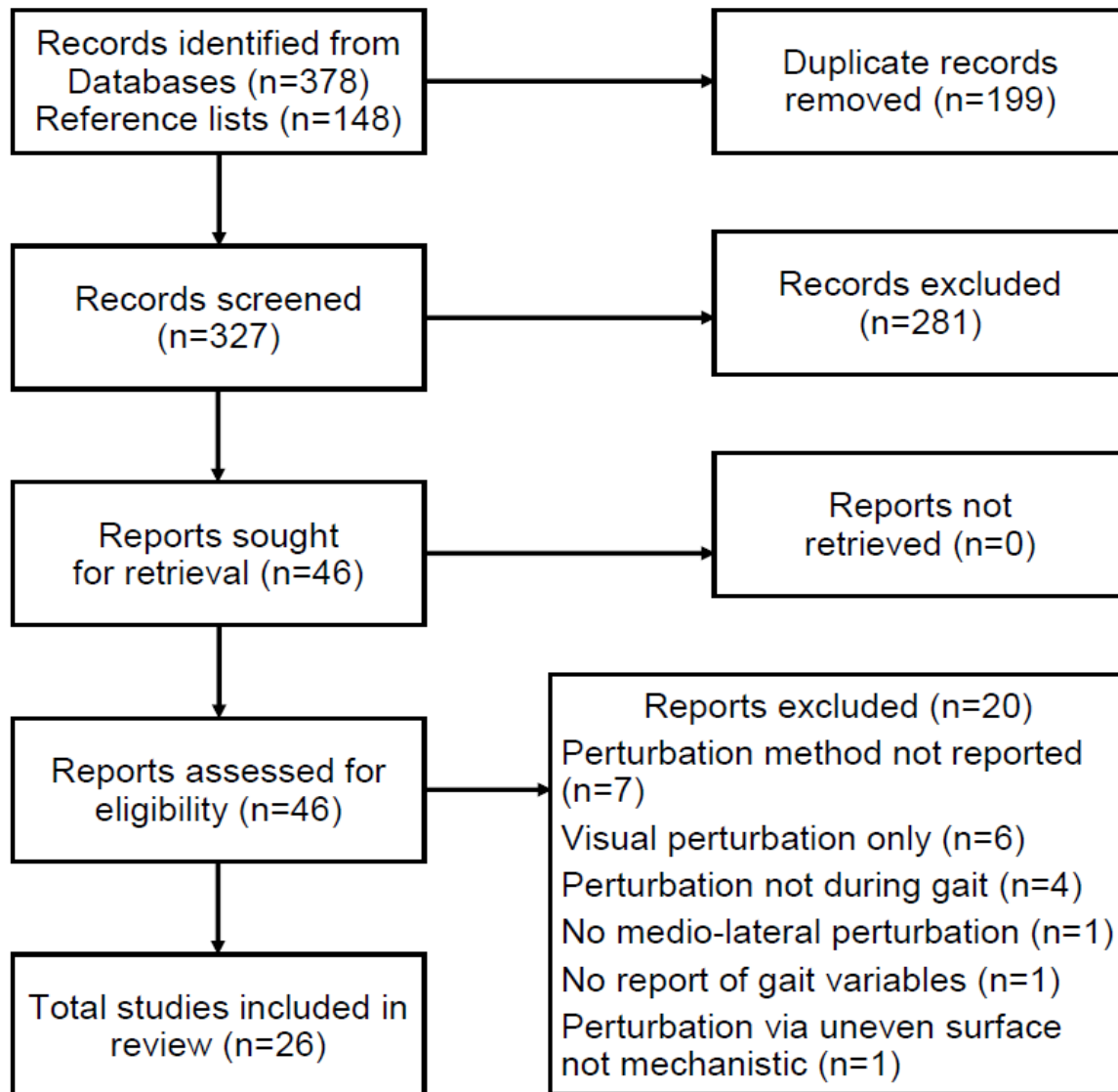


Figure 2 PRISMA flow diagram of search and exclusion criteria (Page et al., 2021).

Risk of bias

An AXIS risk of bias assessment was completed for all included studies and the full results can be found in appendix B. The AXIS tool is divided into five sections: Introduction, Methods, Results, Discussion and Other. Largely there was a low risk of bias across all included studies. None of the included studies stated: if measurements were undertaken to address and categories non-respondents; if the response rate raised concerns about non-response bias or stated information about non-respondents. There was also a lack of detail regarding participant recruitment methods. All studies clearly stated the aim and selected appropriate methods and outcome measures for the aims.

Characteristics of included studies

All included studies were laboratory based, 15 used a treadmill and eleven perturbed participants during over-ground walking. A total of 572 participants (mean age: 45 (22) years, height: 1.63 (0.05) m, body mass: 70.7 (6.8) kg) from 26 studies were included. Full details of participants are given in table 1. Two studies (n = 9, n = 20) did not report

participant age, height or body mass (Hof and Duysens, 2018b; van Hal *et al.*, 2018). One study reported 15 participants with a fall history from the last 12 months (Punt *et al.*, 2017).

Balance perturbation methods and details

Full details of balance perturbation methods of each paper are given in Table 2. 15 studies used a treadmill to perturb participants, of which seven studies used a split-belt treadmill (Aprigliano *et al.*, 2015, 2016, 2019; Punt *et al.*, 2017; van Hal *et al.*, 2018; Rieger *et al.*, 2020; Vervoort *et al.*, 2020), five used a single-belt treadmill (Kurz *et al.*, 2016; Haarman *et al.*, 2017; Madehkhaksar *et al.*, 2018; Roeles *et al.*, 2018; Onushko *et al.*, 2019), and three studies oscillated the treadmill during gait (Eduardo Cofré Lizama *et al.*, 2015; Sheehan *et al.*, 2015; Rosenblum *et al.*, 2020).

Seven studies used participant-worn devices to perturb participants; five were waist or hip worn attachments connected to external structures which pulled or pushed participants (Martelli *et al.*, 2016; Vlutters, Van Asseldonk and Van Der Kooij, 2016; Martelli, Kang and Agrawal, 2017; Zadavec, Olenšek and Matjačić, 2017; Hof and Duysens, 2018b). One device applied vibrations to the hip abductor muscles (Arvin *et al.*, 2018). One study used an oscillating pendulum system worn on participants' backs (Best *et al.*, 2019) and one study applied a narrow stepping task while participants were executing a cognitive task (Francis *et al.*, 2015).

A translational platform here refers to a movable floor-level platform designed to cause perturbation to participants BoS during gait. Three studies perturbed participants during over ground walking, two using a translational platform (McIntosh, Zettel and Vallis, 2017; Shulman, Spencer and Vallis, 2018), and one used a physical therapist who applied the balance perturbations by pushing the participant at waist level (Capin *et al.*, 2017).

The largest perturbation displacement was 18.0 cm, delivered by two studies using ML treadmill translation and ML translation of a platform during over ground gait (Kurz *et al.*, 2016; Shulman, Spencer and Vallis, 2018). The smallest displacement was 1.0 cm by ML treadmill translation (Kurz *et al.*, 2016). The largest acceleration was $16.0 \text{ m}\cdot\text{s}^{-2}$ by ML treadmill translation (Kurz *et al.*, 2016) and the smallest was $0.1613 \text{ m}\cdot\text{s}^{-2}$ with a split belt treadmill (Rieger *et al.*, 2020). Where perturbation number was reported, the average number applied was 14 with all studies reporting at least 10 perturbations per participant.

Timing of balance perturbation onset

14 studies initiated balance perturbations at heel strike (Aprigliano *et al.*, 2015, 2016, 2019; Martelli *et al.*, 2016; Zadavec, Olenšek and Matjačić, 2017; Martelli, Kang and Agrawal, 2017; McIntosh, Zettel and Vallis, 2017; Punt *et al.*, 2017; Roeles *et al.*, 2018; van Hal *et al.*, 2018; Madehkhaksar *et al.*, 2018; Onushko *et al.*, 2019; Rosenblum *et al.*, 2020; Rieger *et al.*, 2020). Four studies applied perturbations continuously during gait (Francis *et al.*, 2015; Sheehan *et al.*, 2015; Best *et al.*, 2019; Vervoort *et al.*, 2020).

Table 1 Summary of included studies' participant details

Reference	Participants (YA/OA)	Age years (SD) YA/OA	Height m (SD) YA/OA	Body mass kg (SD) YA/OA	Health status
(Aprigliano et al., 2015)	5	25.4 (3.1)	1.7 (0.1)	63.2 (11)	Healthy
(Aprigliano et al., 2016)	15	26.1 (1.3)	1.78 (0.06)	68.8 (12.3)	Healthy
(Aprigliano et al., 2019)	6	68.7 (5.2)	1.76 (0.1)	76.9 (7.9)	Healthy
(Arvin et al., 2018)	30 (12/18)	27.3 (1.7) 70.8 (6.8)	1.68 (0.109) 1.703 (0.86)	60.6 (10.5) 80.1 (8.8)	Healthy
(Best et al., 2019)	12	21.8 (1.0)	1.802 (0.092)	72.3 (11.2)	Healthy
(Capin et al., 2017)	40	23.0 (7.0)	1.78 (0.07)	86.0 (11.5)	Post-operative ACL rupture
(Eduardo Cofré Lizama et al., 2015)	19	72 (5)	1.73 (0.09)	76.6 (15.0)	Healthy
(Francis et al., 2015)	23 (12/11)	23.6 (3.9) 71.2 (4.2)	1.69 (0.25) 1.64 (0.06)	70.7 (11.3) 66.9 (9.6)	Healthy
(Haarman et al., 2017)	10	52 (16)	1.75 (0.06)	82.5 (13.6)	Paretic
(Hof and Duysens, 2018b)	9	19-23	Not reported	Not reported	Healthy
(Kurz et al., 2016)	53	80.1 (5.6)	1.583 (0.089)	68.3 (13.9)	Healthy
(Madehkhaksar et al., 2018)	10	26.4 (4.1)	1.7 (0.08)	64.4 (12.5)	Healthy
(Martelli, Kang and Agrawal, 2017)	8	29.9 (4.9)	1.68 (0.08)	67.9 (9.4)	Healthy

Reference	Participants (YA/OA)	Age years (SD) YA/OA	Height m (SD) YA/OA	Body mass kg (SD) YA/OA	Health status
(Martelli et al., 2016)	18	23.9 (4.2)	1.75 (0.08)	70.3 (8.1)	Healthy
(McIntosh, Zettl and Vallis, 2017)	21 (11/10)	23.8 (3.1) 71.1 (3.1)	1.762 (0.009) 1.676 (0.009)	71.2 (12.4) 72.6 (11.1)	Healthy
(Onushko et al., 2019)	15	21.3 (1.4)	1.7 (0.1)	68.8 (10.7)	Stroke survivors
(Punt et al., 2017)	38	60.2 (9.5)	1.715 (0.115)	85 (19.6)	Healthy
(Rieger et al., 2020)	30	70.1 (4.5)	1.72 (0.09)	75.2 (10.2)	Healthy
(Roeles et al., 2018)	18 (9/9)	25.1 (3.4) 70.1 (8.1)	1.76 (0.09) 1.70 (0.11)	76.6 (15.1) 77.9 (10.5)	Healthy
(Rosenblum et al., 2020)	24 (12/12)	26.9 (3.4) 69.5 (5.2)	1.684 (0.073) 1.697 (0.067)	63.67 (10.26) 78.34 (16.22)	Healthy
(Sheehan et al., 2015)	22	27.2 (6.9)	1.76 (0.095)	84.8 (13.9)	Healthy
(Shulman, Spencer and Vallis, 2018)	34 (18/16)	21.7 (2.6) 75.6 (5.3)	1.80 (0.1) 1.70 (0.1)	72.8 (11.0) 72.7 (13.4)	Healthy
(van Hal et al., 2018)	20	Not reported	Not reported	Not reported	Not reported
(Vervoort et al., 2020)	75	48.1 (17.95) (18-79)	1.75 (0.10)	73.81 (10.81)	Healthy
(Vlutters, Van Asseldonk and Van Der Kooij, 2016)	10	25 (2)	1.8 (0.11)	67 (12)	Healthy
(Zadavec, Olenšek and Matjačić, 2017)	7	33.4 (8.5)	1.81 (0.05)	80.1 (11.6)	Healthy

ACL – anterior cruciate ligament, YAs – young adults

Table 2 Details of perturbation methods

Reference	Perturbation cause	Perturbation details
(Aprigliano <i>et al.</i> , 2015)	Split belt TM	Initiation at HS, 5 left belt perturbations 5 right belt
(Aprigliano <i>et al.</i> , 2016)	Split belt TM	Initiation at HS, 5 left belt perturbations 5 right belt
(Aprigliano <i>et al.</i> , 2019)	Split belt TM	Initiation at HS, anterior or lateral translation of right belt
(Arvin <i>et al.</i> , 2018)	Hip abductor vibration during gait	Mean vibration duration 0.41s, 40% stance phase
(Best <i>et al.</i> , 2019)	Pendula mass oscillating during gait	Pendula mass 6.4% of body mass, at 130% and 70% of strides
(Capin <i>et al.</i> , 2017)	OG walking physical therapist perturbations	Lack of detail, progressive perturbation applied by physical therapist
(Eduardo Cofré Lizama <i>et al.</i> , 2015)	Single belt TM oscillations	132-135s at 1.13Hz TM oscillations
(Francis <i>et al.</i> , 2015)	Narrow TM walking, cognitive task	Perturbation sum of sinusoids waves, virtual hallway, amplitude 0.175m
(Haarman <i>et al.</i> , 2017)	Movable TM	Initiation at toe-off, 150ms block pulses at 4,8,12 &16% of body mass
(Hof and Duysens, 2018b)	Waist belt, right side, pneumatics	Push & pulls of 100ms throughout gait, at 2.7-12.4 kg.m.s ⁻¹
(Kurz <i>et al.</i> , 2016)	Movable TM	Perturbation every 20-40s for 14 mins, 0.01-0.18 m, 0.5-16.0 m.s ⁻²
(Madehkhaksar <i>et al.</i> , 2018)	Single belt TM	16 perturbations of 2.5 m.s ⁻¹ , HS within four five min blocks
(Martelli, Kang and Agrawal, 2017)	Waist belt, cable pulleys	Pulsation rise, hold, fall time of 150ms. 10, 15, 20% body mass, HS
(Martelli <i>et al.</i> , 2016)	Waist belt, cable pulleys	Pulsation rise, hold, fall time of 150ms. 10, 15, 20% BM, HS
(McIntosh, Zettel and Vallis, 2017)	OG walking floor translation	0.15m, 0.316 m.s 2.4 m.s ² , HS
(Onushko <i>et al.</i> , 2019)	Single belt TM	ML ± 0.08 , 0.165 & 0.25m oscillations HS
(Punt <i>et al.</i> , 2017)	Split belt TM	16 ML perturbations of 0.045m separated by 15s
(Rieger <i>et al.</i> , 2020)	Split belt TM	0.31s, 0.05m, 0.1613 m.s ² , HS medial translation
(Roeles <i>et al.</i> , 2018)	Single belt TM	0.7s, 0.05m, 2.04m.s ⁻²
(Rosenblum <i>et al.</i> , 2020)	ML TM oscillations	0.15m displacement in 0.92s, HS
(Sheehan <i>et al.</i> , 2015)	ML TM oscillations	Continuous 3mins ± 12 cm displacement from centre
(Shulman, Spencer and Vallis, 2018)	OG walking ML floor translation	6 left, 4 right side prior to toe-off at 0.6m.s, 2m.s ² , 0.18m
(van Hal <i>et al.</i> , 2018)	Split belt TM	Initiation at target leg HS, lateral pull, 0.66 m.s ⁻² of 0.05m
(Vervoort <i>et al.</i> , 2020)	Split belt TM	Randomly, one belt at 0.7m.s ⁻¹ , one at 1.4 m.s ⁻¹
(Vlutters, Van Asseldonk and Van Der Kooij, 2016)	Waist belt, motors	150ms block pulses, forces 4, 8, 12, 16% of body mass
(Zadavec, Olenšek and Matjačič, 2017)	Waist belt	5 repeats of 4 directions. 15% BW, 150ms

BM – body mass, HS – heel strike

Two studies perturbed participants at toe-off (Vlutters, Van Asseldonk and Van Der Kooij, 2016; Haarman *et al.*, 2017). One perturbed participants prior to toe-off (Shulman, Spencer and Vallis, 2018).

One study applied perturbations at different points throughout the gait cycle (Hof and Duysens, 2018a). One study applied perturbations every 20-60 seconds for 14 minutes irrespective of the point of gait cycle (Kurz *et al.*, 2016). One study perturbed participants at 40% of stance phase (Arvin *et al.*, 2018). Two studies did not specify at which point in the gait cycle perturbations was applied (Eduardo Cofré Lizama *et al.*, 2015; Capin *et al.*, 2017).

Preferred or prescribed walking speed

15 studies prescribed a fixed walking speed for participants when subjected to perturbations (Eduardo Cofré Lizama *et al.*, 2015; Sheehan *et al.*, 2015; Francis *et al.*, 2015; Vlutters, Van Asseldonk and Van Der Kooij, 2016; Martelli *et al.*, 2016; Haarman *et al.*, 2017; Martelli, Kang and Agrawal, 2017; McIntosh, Zettel and Vallis, 2017; Punt *et al.*, 2017; Hof and Duysens, 2018b; Madehkhaksar *et al.*, 2018; Best *et al.*, 2019; Rieger *et al.*, 2020; Rosenblum *et al.*, 2020; Vervoort *et al.*, 2020). Nine studies allowed participants to walk at their preferred walking speed when perturbed (Aprigliano *et al.*, 2015, 2016, 2019; Kurz *et al.*, 2016; Arvin *et al.*, 2018; Roeles *et al.*, 2018; Shulman, Spencer and Vallis, 2018; van Hal *et al.*, 2018; Onushko *et al.*, 2019). Two studies did not report if gait speed was prescribed or self-selected (Capin *et al.*, 2017; Zadavec, Olenšek and Matjačić, 2017).

Outcome variables

Details of outcome variables by study are given in Table 3 along with the results.

Kinetics

Two studies reported differing changes in vertical ground reaction force (vGRF) results, one reported an increase (Vlutters, Van Asseldonk and Van Der Kooij, 2016) the other reported no significant effect of split-belt perturbation on participant's vGRF in normal gait (Punt *et al.*, 2017). One study used the vGRF data to produce CoP data (Arvin *et al.*, 2018).

Three studies reported kinetic outcome variables of and moments (Capin *et al.*, 2017), angular momentum (Sheehan *et al.*, 2015). Perturbation exposure decreased angular momentum (Sheehan *et al.*, 2015). Stride-to-stride variability increased, as did range of angular momentum (Sheehan *et al.*, 2015). There was no significant effect of perturbation exposure on angular momentum (Sheehan *et al.*, 2015), nor was there a significant difference in hip, knee and ankle moments (Capin *et al.*, 2017).

Kinemematics

Physical therapist delivered perturbations decreased hip, knee and ankle excursions at loading (Capin *et al.*, 2017). However, there was no significant difference in hip, knee and ankle joint angles (Capin *et al.*, 2017). Two studies reported joint angles alone (Capin *et al.*, 2017; van Hal *et al.*, 2018).

Table 3 Outcome measures and the perturbation effect

Reference	Perturbation	Outcome variables	Change in outcome variables
(Aprigliano et al., 2015)	Split belt TM, AP or ML translation	Hip, knee, ankle joint angles, compensatory time, MoS	Joint angles altered, no detail. Increased compensatory time with perturbation intensity. Decreased MoS
(Aprigliano et al., 2016)	Split belt TM, AP or ML translation	Stride time, stance %, hip, knee, ankle joint angles	Decreased stride time, stance %, hip, knee, thigh, shank, foot RoM. Increased ankle RoM.
(Aprigliano et al., 2019)	Split belt TM, anterior or lateral belt translation	Dynamic stability, step width, single and double support time, hip, knee, ankle RoM, MoS.	Decreased dynamic stability, step width, single and double support time, knee, ankle RoM. Increased frontal plane hip RoM, sagittal and frontal plane MoS. No change hip RoM.
(Arvin et al., 2018)	Hip abductor muscle vibration	CoM position, CoM velocity, step width.	Decreased step width at perturbation, increased step width post perturbation. No change CoM position or velocity
(Best et al., 2019)	Oscillating mass	MoS, CoM amplitude, step width, step length, step time	Increased MoS only in out of phase. Decreased CoM amplitude. No change step width, step length, step time.
(Capin et al., 2017)	Physical therapist in OG gait	Hip, knee and ankle joint angles, excursions, moment	Decreased hip, knee, ankle excursions at loading. No change hip, knee, ankle joint angles, moments
(Eduardo Cofré Lizama et al., 2015)	ML TM oscillations	ML LDE	Decreased LDE on TM than OG. Negative correlation between FPS and TM LDE
(Francis et al., 2015)	Narrow walking task	ML LDE, step width, step length, step width variability, step length variability, gait speed	Decreased LDE on TM than OG. Negative correlation between FPS and TM LDE. No change step width, step length, step width variability, step length variability, gait length, step width variability.
(Haarman et al., 2017)	ML translation	Gluteus medius activation, CoM velocity, step width, support time, foot velocity, eXCoM	Increased swing time, gluteus medius activation CoM velocity variability. Increased lead foot ML distance from CoM, foot velocity increased. No change step width,

Reference	Perturbation	Outcome variables	Change in outcome variables
(Hof and Duijsens, 2018b)	Waist belt pulls and pushes to right side	Tibialis anterior, plantar flexor, soleus activation, CoP	Increased tibialis anterior, plantar flexor, soleus activation. Decreased CoP motion.
(Kurz et al., 2016)	AP or ML TM translation	CoM sway, sway velocity, sway area, foot contact time, voluntary stepping	Decreased CoM AP and ML-sway, sway velocity, sway area. Interaction for foot contact time of the voluntary step execution.
(Madelhkhaksar et al., 2018)	AP or ML TM translation	Step length, step width, cadence, dynamic stability, MoS step length, step width, cadence,	Increase step width, cadence, dynamic stability, step length variability, step width variability, cadence variability, dynamic stability variability, AP and ML MoS. Decrease step length.
(Martelli, Kang and Agrawal, 2017)	AP or ML waist pull or push	Dynamic stability, MoS, BoS, adaptive response, step length, step width	Increased dynamic stability, MoS, BoS, adaptive response, step length. Decreased step width.
(Martelli et al., 2016)	AP or ML waist pull or push	Step length, step width, MoS, BoS.	Increased step length, MoS, step width at first perturbation. Decreased step width over all, AP & ML BoS, AP MoS, at first recovery step. No change at perturbation onset AP BoS, AP & ML MoS.
(McIntosh, Zettel and Vallis, 2017)	OG, AP or ML floor translation	CoM velocity, CoM displacement, step width, step length, recovery step type.	Increase CoM velocity, CoM displacement range, normalised ML trunk CoM displacement, step width. Decrease step length. 85 of 840 recovery steps were cross-over.
(Onushko et al., 2019)	ML TM translation	Step length, step frequency, step width, MoS.	Increased step frequency, step width, ML MoS, ML MoS variability. Decreased step length.
(Punt et al., 2017)	ML TM translation	Step length, step width, step time, MoS, dynamic stability.	Increase step length, step width, ML MoS, step time. Decrease AP MoS, dynamic stability
(Rieger et al., 2020)	Medial TM translation	Step width, step time, swing time, stance time, step length, dynamic stability, LDE, recovery	Increase dynamic stability, step time, swing time, stance time, step length, recovery performance. Decrease LDE, step width, recovery time.

Reference	Perturbation	Outcome variables	Change in outcome variables
(Roeles et al., 2018)	AP or ML TM translation	Step width, step length, step time, MoS.	Increased AP and ML MoS, step width. Decreased step length, step time.
(Rosenblum et al., 2020)	ML TM translation	Step length, step width, recovery time.	Increased step length, step width. Total recovery occurred 4-6s after perturbation
(Sheehan et al., 2015)	ML TM translation	Stride variability, angular momentum, gait phases.	Increased stride variability, range of angular momentum. Decreased angular momentum during second double support.
(Shulman, Spencer and Vallis, 2018)	OG, ML floor translation	BoS, step width, single support time, swing time, step length, gait speed	Increased BoS, step width. Decreased single support time, swing time, step length. No change gait speed.
(van Hal et al., 2018)	Lateral TM translation	vGRF kinetic, kinematic, spatio-temporal	Preliminary trial paper so no results.
(Vervoort et al., 2020)	Split belt TM at two speeds	Cadence, MoS, step length, single and double support time.	Increased cadence, ML MoS of slow leg. Decreased step length, single and double support time
(Vlutters, Van Asseldonk and Van Der Kooij, 2016)	AP or ML waist pull or push	CoM velocity, CoP, GRF, step length, step width single and double support time.	Linear relationship between CoM velocity, CoP and GRF. Increased vGRF, step length, single and double support time. Decreased step width.
(Zadravec, Olenšek and Matjačić, 2017)	AP or ML waist pull or push	Step time, step width, step length.	Increase step time. Decrease step width, step length. No change in 82% of all stepping response parameters between OG and TM perturbations.

OG – over ground, LDE – local divergence exponent, TM - treadmill

Elevation angles, joint angle and joint range of motion (RoM) were reported by two studies (Aprigliano *et al.*, 2015, 2016). First and second double support phase, single support phase and swing phase duration were all reported by one study (Aprigliano *et al.*, 2019). ML foot placement and lower limb RoM were reported by one study (Arvin *et al.*, 2018) with three others reporting combined AP and ML foot placement (Vlutters, Van Asseldonk and Van Der Kooij, 2016; Haarman *et al.*, 2017; Zadavec, Olenšek and Matjačić, 2017). Seven studies reported CoM displacements and CoM velocity (Vlutters, Van Asseldonk and Van Der Kooij, 2016; McIntosh, Zettel and Vallis, 2017; Arvin *et al.*, 2018; Madehkhaksar *et al.*, 2018; Shulman, Spencer and Vallis, 2018; van Hal *et al.*, 2018; Best *et al.*, 2019). CoM sway velocity, sway area and mean sway during gait was reported by one study (Kurz *et al.*, 2016). Five studies used CoP displacement as an outcome variable (Vlutters, Van Asseldonk and Van Der Kooij, 2016; Arvin *et al.*, 2018; Shulman, Spencer and Vallis, 2018; van Hal *et al.*, 2018; Vervoort *et al.*, 2020). Five studies reported measures of dynamic stability, including maximum Lyapunov exponent also known as local divergence exponent (Eduardo Cofré Lizama *et al.*, 2015; Martelli, Kang and Agrawal, 2017; Madehkhaksar *et al.*, 2018; Roeles *et al.*, 2018; Aprigliano *et al.*, 2019). Two studies reported MoS (Martelli *et al.*, 2016; Roeles *et al.*, 2018) and one reported BoS area (Martelli *et al.*, 2016). One study reported participant balance, measuring balance by the range of frequency in target signals described by a gain of linear constant coefficient transfer function between CoM and the target signal (Eduardo Cofré Lizama *et al.*, 2015).

A linear relationship was identified between CoM velocity, CoP and GRF during perturbation recovery (Vlutters, Van Asseldonk and Van Der Kooij, 2016). Conflicting findings were reported for CoM velocity with over ground platform translation increasing CoM velocity (McIntosh, Zettel and Vallis, 2017) while hip abductor vibration showed no effect on CoM velocity (Arvin *et al.*, 2018). When participants were perturbed with a ML oscillating treadmill, ML CoM displacement increased (Eduardo Cofré Lizama *et al.*, 2015). On the contrary, over ground platform translation perturbation showed no effect on either ML CoM displacement or normalised ML displacement (McIntosh, Zettel and Vallis, 2017). AP and ML treadmill translation significantly decreased CoM AP sway (Kurz *et al.*, 2016). 16 AP and ML treadmill belt translations of $2.5 \text{ m} \cdot \text{s}^{-1}$ (Madehkhaksar *et al.*, 2018), AP and ML waist belt pulls and pushes (Martelli, Kang and Agrawal, 2017) and medial belt translation (Rieger *et al.*, 2020) all studies reported increase participants dynamic stability (Martelli, Kang and Agrawal, 2017).

One contradictory study reported AP and ML split-belt treadmill translation resulting in decreased dynamic stability among participants (Aprigliano *et al.*, 2019). MoS results were also inconsistent. ML treadmill belt translation (Onushko *et al.*, 2019), AP and ML waist belt pulls and pushes (Martelli *et al.*, 2016) and the oscillating worn mass perturbation method

increased participants MoS (Best *et al.*, 2019) all significantly increased participants AP and ML MoS. However, single belt AP and ML treadmill translation of 0.05m at 2.04 m·s⁻², decreased participants ML MoS and BoS (Roeles *et al.*, 2018). Finally, one study reported no significant effects of split belt treadmill perturbation on participants' MoS (Punt *et al.*, 2017). Just one study reported age specific results following a split belt treadmill set at two different speeds, OAs displayed reduced adaptive control within ML MoS compared to YAs (Vervoort *et al.*, 2020). BoS was found to increase in all reported cases from AP and ML waist belt pulls and pushes (Martelli *et al.*, 2016) and over-ground floor translations (Shulman, Spencer and Vallis, 2018).

Spatio-temporal

Seven studies report other spatio-temporal outcome variables including duration of gait phases (Aprigliano *et al.*, 2015, 2016, 2019; Vlutters, Van Asseldonk and Van Der Kooij, 2016; Haarman *et al.*, 2017; Punt *et al.*, 2017; Hof and Duysens, 2018b). Total stride time, was reported by two studies (Aprigliano *et al.*, 2015, 2016), stance and swing time was reported by one additional study (Rieger *et al.*, 2020). First and second double support phase, single support phase and swing phase duration were all reported by one study (Aprigliano *et al.*, 2019). 16 studies reported spatio-temporal parameters, nine reported step width and step length (Francis *et al.*, 2015; McIntosh, Zettel and Vallis, 2017; Zadravec, Olenšek and Matjačić, 2017; Madehkhaksar *et al.*, 2018; Shulman, Spencer and Vallis, 2018; van Hal *et al.*, 2018; Best *et al.*, 2019; Onushko *et al.*, 2019; Rosenblum *et al.*, 2020). Two studies reported using GRF results to measure spatio-temporal variables (Martelli, Kang and Agrawal, 2017; Best *et al.*, 2019). With one study reporting step width without step length (Martelli, Kang and Agrawal, 2017). Step width variability and step length variability were reported by one study (Francis *et al.*, 2015). Step time was reported as an outcome variable by two studies (Kurz *et al.*, 2016; Zadravec, Olenšek and Matjačić, 2017). Gait speed was reported by three studies (Francis *et al.*, 2015; van Hal *et al.*, 2018; Rosenblum *et al.*, 2020) and step frequency was reported by four studies (Madehkhaksar *et al.*, 2018; Shulman, Spencer and Vallis, 2018; van Hal *et al.*, 2018; Onushko *et al.*, 2019). Included studies reported contradicting results of participants step width, showing an increase following perturbations via, ML treadmill translation (Onushko *et al.*, 2019), AP and ML treadmill translations (Madehkhaksar *et al.*, 2018), single belt AP and ML treadmill translation of 0.05 m at 2.04 m·s⁻² (Roeles *et al.*, 2018) and hip abductor muscle vibration during gait; with a larger effect among OAs than YAs (Arvin *et al.*, 2018). Contradicting this, anterior and lateral split belt treadmill translations (Aprigliano *et al.*, 2019), post gait vibration of hip abductor muscles (Arvin *et al.*, 2018) and some medial treadmill translations (Rieger *et al.*, 2020), all showed reduced step width. Further opposing results showed no significant effect on step width when participants performed a narrow stepping task (Francis *et al.*,

2015), were subjected to the oscillating mass perturbation (Best *et al.*, 2019), over ground AP and ML floor translations (McIntosh, Zettel and Vallis, 2017), or ML treadmill translations (Punt *et al.*, 2017). Only one study reported step width variability and found no significant effect of a narrow walking task (Francis *et al.*, 2015).

Further contradicting results were seen in step length results, with AP and ML waist pulls and pushes (Martelli, Kang and Agrawal, 2017) and medial treadmill belt translations (Rieger *et al.*, 2020) increasing step length. Opposing this was findings that treadmill perturbations of 0.05 m at $2.04 \text{ m}\cdot\text{s}^{-2}$ decreased step length (Roeles *et al.*, 2018) as did ML treadmill translations (Onushko *et al.*, 2019) and AP and ML treadmill translations of $2.5 \text{ m}\cdot\text{s}^{-1}$ (Madehkhaksar *et al.*, 2018). Just one study described an effect of increased age also reducing step length in response to different split belt speeds (Vervoort *et al.*, 2020). Finally there was no significant difference in step length following over ground AP and ML floor translations (McIntosh, Zettel and Vallis, 2017), ML treadmill translations (Punt *et al.*, 2017) or a narrow stepping task (Francis *et al.*, 2015). Only one study reported step length variability and found no significant effect of a narrow walking task (Francis *et al.*, 2015).

Both studies reporting step frequency agreed to a significant increase following AP and ML treadmill translation of 0.05 m at $2.04 \text{ m}\cdot\text{s}^{-2}$ (Roeles *et al.*, 2018) and 16 AP and ML treadmill translations at $2.5 \text{ m}\cdot\text{s}^{-1}$ applied at heel strike (Madehkhaksar *et al.*, 2018). Just one study described an age effect stating increased age also increased cadence in response to different split belt speeds (Vervoort *et al.*, 2020).

Additional findings showed AP perturbations reducing stride duration significantly more compared with ML, or diagonal perturbations (Aprigliano *et al.*, 2015, 2016). Further dispute is reported where ML treadmill translation caused a decrease in participants step time (Onushko *et al.*, 2019), while the same perturbation method by a different study showed no effect of the perturbation on step time (Punt *et al.*, 2017). Inconsistent results were also found regarding single support time, ML treadmill oscillations (Haarman *et al.*, 2017) and medial treadmill translations (Rieger *et al.*, 2020) increased single support time. However, anterior and lateral split belt treadmill translations (Aprigliano *et al.*, 2019), hip abductor muscle vibration (Arvin *et al.*, 2018) and differing split belt speeds (Vervoort *et al.*, 2020) all showed decreased single support time. Additionally increased age was found to further decrease single support time (Vervoort *et al.*, 2020). Conversely, regarding double support duration, all studies reported a decreased duration following perturbations of anterior and lateral split belt treadmill translations (Aprigliano *et al.*, 2019), hip abductor muscle vibration (Arvin *et al.*, 2018) and differing split belt speeds (Vervoort *et al.*, 2020). AP and ML treadmill translations were shown to increase participants foot contact time, however this was not specific to single, double or both support durations (Kurz *et al.*, 2016). There was report of medial treadmill translations increasing participants swing time (Rieger *et al.*, 2020). Yet

over ground floor translations showed a decrease in swing time, significantly more so for OAs compared to YAs post perturbation (Shulman, Spencer and Vallis, 2018).

No study reported a significant effect of perturbation on gait speed. One study discussed the time taken for each outcome variable to return to baseline following perturbation, there was no age effect on recovery time with step length recovered in 4.7 - 5.1 seconds and step width in 4.9 - 6.0 seconds (Rosenblum *et al.*, 2020). One paper compared differences of over ground and treadmill walking when the same perturbations were applied, finding no significant difference between conditions for step length, step width, step time and foot placement (Zadravec, Olenšek and Matjačić, 2017).

Muscle activations

Two studies reported muscle activations (Haarman *et al.*, 2017; Hof and Duysens, 2018a). Gluteus Medius activation increased significantly as ML distance between the CoM and stance foot increased a result of perturbation application (Haarman *et al.*, 2017). All muscles measured - tibialis anterior, peroneus longus and soleus - showed significant increase in electromyography response following gait perturbation (Hof and Duysens, 2018a).

Discussion

The purpose of this review was: identify and describe recent methods and outcome variables of balance perturbation used during gait, from the available up-to-date literature, with specific attention to the outcome variables used. The most frequent method of balance perturbation was found to be ML treadmill belt translation, initiated at heel strike. All included studies reported the use of spatio-temporal outcome variables, most frequently step width. Some studies reported the collection of only spatio-temporal parameters, while other studies suggested the addition of kinetic data in the outcome variables, this added depth to results and potentially provided an alternative interpretation of the results.

Studies which perturbed participants via treadmill belt translation or floor translation during over ground gait relocate participants BoS and CoP. Among these studies there was agreement regarding an increase in BoS area, muscle activation, dynamic stability and decreased CoM sway. MoS and the relevant interactions could be used to establish any links in participant's spatio-temporal parameters of gait to dynamic stability (Madehkhaksar *et al.*, 2018; Roeles *et al.*, 2018; Aprigliano *et al.*, 2019; Rieger *et al.*, 2020). There was disagreement on the results regarding MoS, joint angle RoM, step width and length, with some studies reporting increases and others reporting decreases. This could be due to the differing participant ages as increased age leads to decreased joint RoM, step width and step length (Zhao and Wang, 2016; Kluft, van Dieën and Pijnappels, 2017). Equally, the physical condition of participants was not stated; therefore, while participants may have been under 65 years - depending on physical activity levels - this could also have affected responses due differing strength, physical and balance capacities (Sparling *et al.*, 2015).

The most frequent method of perturbation was by treadmill belt translation, however the results from such perturbations were inconsistent as previously discussed. The inconsistencies in results are likely due to the variety of methods and lack of consistency between studies. While each study will have applied appropriate controls and remained consistent throughout the experiment, these controls may not have been present in other studies due to differing protocols or aims.

All perturbations applied to CoM were via the waist, most commonly by a waist belt attached to an external device. These belts were attached to cable pulleys (Martelli, Kang and Agrawal, 2017; Martelli *et al.*, 2019) a pneumatic device (Hof and Duysens, 2018b) or a motor driven device (Vlutters, Van Asseldonk and Van Der Kooij, 2016; Zadavec, Olenšek and Matjačić, 2017).

Within the results there was agreement on increased muscle activation, dynamic stability, MoS, BoS peak vGRF, step time and decreased CoP motion. There were conflicting results regarding step width and length with some studies reporting increases and other decreases. While there was higher levels of agreement across studies that perturbed the CoM there were few cases of CoM perturbation compared with BoS perturbations. This increased agreement is likely due to response to a CoM perturbation eliciting a response of the BoS. Previous research has shown that the muscle synergies activated to maintain CoM control during perturbation are indicative of the desired direction of the CoM, rather than the direction of the perturbation (Chvatal *et al.*, 2011). This suggests that recovery response is based on the individual's intentions rather than the perturbation they experience. Previous literature has stated the most prevalent cause of 227 falls among OA participants residing in a long-term care facility was incorrect weight shifting, accounting for 41% of all recorded falls (Robinovitch *et al.*, 2013). This was followed by trips and stumbles, causing 21% of falls, while just 3% of falls were a result of slipping. However, among YAs slipping is reported to be the most prevalent cause of falls (Nazifi, Beschorner and Hur, 2019). A slip can be recreated in a laboratory setting by perturbation of the BoS. By comparison, the fall mechanisms typically experienced by OAs are more indicative of perturbation to the CoM. Suggesting investigation of OAs perturbation should be to the CoM, with YAs perturbed at the BoS.

The timing of the perturbation within the gait cycle can also affect results. The most popular point in the gait cycle for perturbation initiation was a heel strike. Some studies did report perturbing participants at specific time intervals regardless of gait cycle, and reported decreased CoP motion (Hof and Duysens, 2018b), decreased CoM sway (Kurz *et al.*, 2016) and decreased MoS and dynamic stability (Punt *et al.*, 2017). These results contradict previous literature as well as other studies in the review; while inconsistency of methods are abundant, the timing of perturbation onset should not be discounted to have had an

influence on results (Rabago, Dingwell and Wilken, 2015; Martelli, Kang and Agrawal, 2017; Huijben *et al.*, 2018; Madehkhaksar *et al.*, 2018; Kongsuk, Brown and Hurt, 2019; Rieger *et al.*, 2020). Further to this, perturbation at heel strike is performed using split belt and single belt treadmills, providing contradicting motivation and intention of the perturbation. If the researcher wanted to perturb both feet, this would be the optimal time to do so. However, use of a split belt treadmill will only perturb one foot. More detailed explanation is needed in studies as to the reasoning for the chosen point of perturbation in order to allow further understanding and testing of the implications.

One study compared participant response to the same balance perturbation applied during over ground and treadmill gait, and found no significant difference between walking conditions for step length, step width, step time or foot placement (Zadravec, Olenšek and Matjačić, 2017). However, previous research has shown treadmill gait to be significantly less complex than over ground gait, meaning the effect of reduced complexity is unlikely to be detectable with spatio-temporal variables alone (Hollman *et al.*, 2016).

Limitations

Many studies were excluded in the screening process due to a lack of detail in the description of the methods of balance perturbation. The research question was specific to recent perturbation methods, although limiting the search to publication during or after 2015 may have excluded studies which did justify the chosen perturbation methods. However, the lack of discussion of this among included studies suggests otherwise. Due to the broad nature of the literature search and the lack of exclusion criteria based on participant parameters, this may have predisposed the search results to an increased variety of perturbation methods due to differing populations presenting a variety of ages, physical conditions and fall history. The variety of the perturbation methods identified by this review is not extensive and more methods appear to be emerging as the field progresses. The importance of implication of perturbation type on results is key to improve accuracy and reliability of results.

Clinical application

One study aimed to determine the difference between fallers and non-fallers from their response to perturbations. However, the results confirmed they were unable to do so (Punt *et al.*, 2017). This may have been due to the outcome variables selected or the balance perturbation methodology applied. There is currently a lack of standardisation regarding perturbation testing and training methods to assess and improve balance recovery; this extends to the outcome variables used by researchers. For standardised testing to be developed, a set of outcome variables need to be identified which most accurately assess fall risk. To date, there are a number of risk factors identified which indicate that an individual has a higher fall risk such as decreased preferred gait speed during daily life but very poor

correlation was found from daily life speed to preferred gait speed in a laboratory setting (Van Ancum *et al.*, 2019). The disparity between daily life gait speed and laboratory gait speed explains how some studies have found the 'Timed Up and Go' test to have a 56% predictive capacity of fall risk when participants were perturbed, yet other studies have found the test to have limited predictive capacity of falls during daily life (Bhatt *et al.*, 2011; Van Ancum *et al.*, 2019). Furthermore, falls specific perturbation method research is needed to enable identification of fallers prior to the first fall based on laboratory data collection (Pai *et al.*, 2006; Rogers and Mille, 2016; Degani, Leonard and Danna-dos-Santos, 2017).

Conclusion

This review summarises perturbation methodologies and provides insight into the outcome variables used in assessing these methods among participants. Whilst there are novel technological advances in this field of research, there is a lack of consideration regarding the selection of perturbation type and outcome variables selected, and the impact these decisions have on the results of a study. This potentially limits the progress in this area of research, therefore resulting in limited and slow clinical translation. The most ecological method of perturbing participants during walking is during over ground walking, despite the draw backs of the participant's knowledge of the location of the perturbation site. The results here also show the importance of using a variety of variables and type of variables. Findings here have been used to inform the methods used to perturb participants in chapter 4. Floor level perturbations during over ground walking have shown to be the most representative of exposure to slips and trips in normal walking conditions.

Chapter 4 – Ageing effects on medio-lateral jerk during perturbed and unperturbed walking

Introduction

Perturbation testing has been used to investigate a number of different populations in balance control and fall risk. One study into healthy community-dwelling OAs found reaction to perturbation to be the most effective measure of fall prediction from an unannounced perturbation during gait compared to both Berg Balance Scale and Timed Up and Go assessments (Bhatt *et al.*, 2011). The study focused on assessing fall outcome rather than fall risk itself, stating the importance of good levels of bone mineral density and lower extremity extension strength for successful recovery from perturbations. This is supported by a different study also investigating perturbation response among OAs, finding multiple recovery steps are linked to increased fall risk (Borrelli *et al.*, 2019). The authors also stated the importance of both hip abduction strength and power in successful perturbation recovery, but, were unable to give a definitive threshold for individuals at a falls risk and those who were not. Conversely, a further study stated stepping behaviours contributed little to balance control when continuous ML perturbations were applied (Best *et al.*, 2019). The mixed findings in recent years are supported by the review in Chapter 3 showing the extreme variety of perturbation methods and outcome variables used to assess participant responses.

OAs are seen to have a quicker stepping response as a result of a lower threshold for perturbation tolerance, therefore their reaction is to step sooner to reduce the risk of being unable to recover (Meyer and Ayalon, 2006). Response time to a perturbation varies depending upon the current activity; during quiet standing young participants (average age 23.5 years) showed faster muscular response to perturbations compared to when perturbation occurred during normal gait (Murnaghan and Robinovitch, 2013). This could suggest that there is a delay in the neuromuscular response as the new information needs to interrupt the current stimuli in order to produce a physical response. Further studies have shown reaction time has minimal influence upon the success of the balance perturbation response (de Freitas, Knight and Barela, 2010). Recent research has found a significant relationship between single stepping response and delayed recovery onset (Borrelli *et al.*, 2019). This shows reaction time has little to no impact on quality, and therefore success of recovery mechanisms, further supporting the finding that the important factors when determining participants' stability are strength, power, and joint RoM, rather than reaction time.

Baseline angular displacement of the CoM has been shown to be associated with a 64% increased falls risk, supporting the need for development of a dynamic measure of stability

to be used in falls assessments (Mahoney *et al.*, 2017). The measure of postural stability is known to be influenced by hydration, height, body mass, peripheral neuropathy and lower limb injury, all of which are also known to affect fall risk (Johansson *et al.*, 2019).

It has been demonstrated by previous research that jerk can represent the stability of gait (Park *et al.*, 2016). Furthermore, in Chapter 2 it was demonstrated that ML jerk was significantly different between OAs with a fall history and those without. However, effect sizes were small between these two groups of otherwise healthy, community-dwelling OAs. The conclusion was made that perturbations are required to challenge gait in order to elicit a response enabling the assessment of the balance system to maintain stability. Therefore, further insight is required to understand the utility of ML jerk as a measure of stability.

Many studies have reported the negative impact of dual task condition on OAs, more so than on YAs, which is reported to be the effect of decaying of the prefrontal cortical circuitry, loss of prefrontal brain mass and the associated deterioration of executive function (Beurskens and Bock, 2012). Therefore, a dual-task condition was also included here to assess any differences elicited by a greater stability challenge by perturbing gait during a mental task.

Considering the results of chapter 2 and 3, the lack of consistency regarding methods of perturbation and the small effect size seen between fallers and non-fallers jerk results, the conclusion was made to assess healthy adults during perturbed walks. It was determined the method most similar to slips in the real world was to perturb participants at floor level during over ground walking. While there is not an abundance of research into this particular method of perturbation consistency of methods is essential in further developing this area of study.

The aim of this study was to identify any differences in ML jerk values in healthy participants aged 18-90 years during over ground normal walking, perturbed walking and dual-task perturbed walking. It was hypothesised that jerk would increase as participant age increased and during dual task walking compared to single task walking. OAs were hypothesised to have slower walking speed, smaller step length and increase cadence in all conditions compared to YAs. It was hypothesised that the perturbation and dual-task trials would affect OAs gait more so than YAs, shown by increased jerk, decreased step length, decreased gait speed and increased cadence.

Methods

Participants

32 community-dwelling adults expressed an interest in participating in the study. One was excluded due to a frequent fall history (more than one fall in the last 12 months), leaving 31 participants (19 males, 12 females) for analysis. Ethical approval was obtained from the Oxford Brookes University Research Ethics Committee UREC (Ref: 201359). Inclusion

criteria required participants to be able to walk unaided for at least 20 metres, and to not be suffering from diabetes, neurological impairment, cardiovascular disease or lower limb injury. 17 YAs aged 21 - 44 years (12 male, 5 female, average age: 32.8 (6.8) years, height: 1.77 (0.10) m, body mass: 84.6 (16.2) kg, BMI: 27.0 (4.7) kg.m⁻²), and 15 OAs aged between 67 - 87 years (7 male, 8 female, age: 74.9 (6.4) years, height: 1.70 (0.10) m, body mass: 70.4 (19.2) kg, BMI: 24.8 (4.8) kg.m⁻²) provided informed consent to participate.

Data collection

Figure 3 shows the experimental set up. Participants were asked to refrain from alcohol and rigorous exercise for 24 hours prior to participation. Height and body mass while shod were recorded to allow for accurate calculations from the overhead harness. Participants performed a minimum of 40 10-metre walks. All walks were over a custom build perturbation plate at five meters which would only move in the ML plane of motion when a participant walked over it. The plate could also be locked in place to prevent any movement when walked over, participants were always made aware when the plate could move and when it was fixed in place. When the plate was not fixed, participants were instructed to continue their walk to the end of the walkway even if perturbed. There were four walking conditions, all participants completed at least ten walks in each condition and were asked to walk at their preferred speed throughout: 1) Walking without the safety harness over the fixed plate. 2) Walking with the safety harness attached to the overhead rail system over the fixed plate. 3) Walking with the safety harness attached to the overhead rail system over the movable plate. 4) Walking with the safety harness attached to the overhead rail system over the movable plate while performing the mental task of counting backwards in sevens from a number given by the researcher. Condition four provided a dual task condition and participants were scored on their performance with one point awarded for each correct number said out loud during the 10 metre walk. An average from all participants' trials was taken. If a participant was unable to recover from a perturbation, the harness prevented contact with the ground, preventing injury. All participants completed the testing and overcame all perturbations.

Data was collected using the LPMS-B2 (Life Performance Research, Japan) series 9-axis IMU placed on participants' L4 vertebra. A second IMU was attached to the bottom left corner of the perturbation plate to record perturbation motion. The IMU sampled at 100Hz and both devices were time-synched on the initial contact on the trip plate.

Data analysis

Data was divided in a number of methods; firstly, in the four walk conditions: 1) Normal walking 2) Harness walking 3) Single task perturbed and 4) Dual task perturbed walking. Within the two perturbed conditions (3 and 4) data was further separated into step type: A) Approach walk to the perturbation plate B) Perturbed step C) 1st recovery step and D) 2nd

recovery step. The approach walk data was collected from gait initiation until the foot contacted the perturbation plate. The perturbed step was defined from when the foot contacted the perturbation plate, until the next heel strike. The 1st recovery step was from that heel strike until the following heel strike. The 2nd recovery step was from that heel strike to the next.

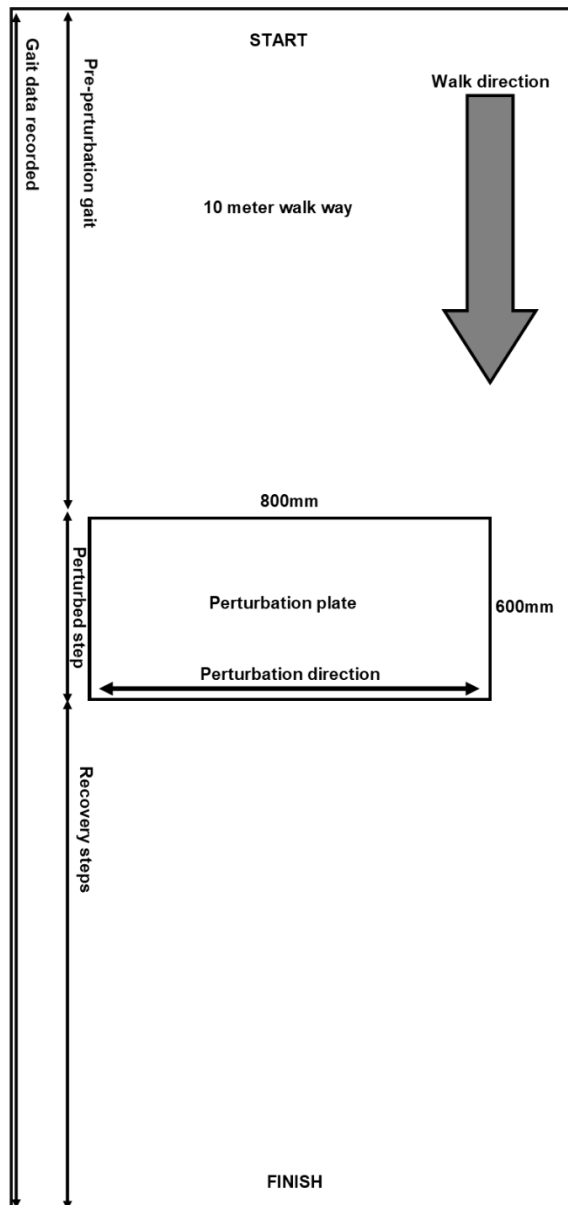


Figure 3 Diagram of the experimental set up of the walk way

From IMU data heel strike was defined using the zero-crossing method where foot contact was defined as the instant where the forward acceleration crosses the zero value and becomes deceleration (Zijlstra and Hof, 2003). Using a custom MATLAB program (Mathworks, MA, USA), linear acceleration calculated from an IMU unit (LP-RESEARCH, Tokyo, Japan) was filtered using a 4th order Butterworth band-pass filter (0.2-15Hz) (Chidean *et al.*, 2018; Shahabpoor and Pavic, 2018). The vertical acceleration signal was rotated from the local reference frame to the global frame using quaternions (Esser *et al.*,

2009) and double integrated to vertical position from where step length, cadence and gait speed were derived according to the inverted pendulum model (Zijlstra and Hof, 2003; Esser *et al.*, 2011). Similar to the data analysis in Chapter 2, CoM ML peak acceleration and deceleration were the independent variables, with falls in the last 12 months used as the dependent variable.

Double integration of ML linear acceleration was used to calculate ML relative position (Esser *et al.*, 2009). Using the same method as explained in chapter 2, ML jerk was calculated as the average of the rate of change in ML acceleration from 200 ms prior to peak lateral relative position to the instant of peak lateral relative position for each step, as well as from the instant of peak ML relative position to 200 ms after. The before values were the ML jerk decelerating, the after were the ML jerk acceleration.

A balance perturbation was defined as a plate displacement larger than 6% of the participant's leg length. An individual's BoS is theorised to be the 12% of the leg length in width; this encompasses the far right and the far left limit of the base of support (Donelan, Kram and Kuo, 2001). Therefore, if the plate displacement exceeded 6% of participants' leg length on either side it was classified as a perturbation. This is further supported by the average leg length of the cohort of 0.90 m, leaving the 6% threshold at 0.054 m. Analysis has been performed on an individual basis according to 6% of the individual's leg length any perturbation beyond the 6% of leg length is a perturbation, and is included in further analysis.

Statistical analysis

Following extraction of the data, mean and standard deviation values were calculated and tested for normality via Q-Q plots. An independent samples t-test was performed to identify any age group differences in age, height, body mass, BMI and dual task score. Age group and walking condition main effects and interactions were determined with a linear mixed-effects model in a repeated measures two by two arrangement. Where age group (YAs and OAs), condition (normal, harness, single task and dual task) and step type (approach walk, perturbed step, 1st recovery step and 2nd recovery step) were set as fixed factors and height and body mass as covariates (SPSS v22, IBM, Armonk, NY). Post-hoc calculations used Sidak correction for multiple comparisons. 95% confidence interval was calculated for significant results. Significance level was set at $p < 0.05$ for all statistical tests.

Results

Q-Q plots showed within age groups that age, height, body mass, BMI and dual task score were all normally distributed. An independent samples t-test showed OAs were significantly older ($t(122) = -36.151$, $p < 0.001$), shorter ($t(122) = 6.063$, $p < 0.001$), lighter ($t(122) = 4.505$, $p < 0.001$) and had a lower BMI ($t(122) = 2.626$, $p = 0.010$) than the YAs group.

There was no significant difference in dual task score between age groups ($t(29) = 0.428$, $p = 0.672$). Mean and standard deviation values are given in Table 4.

Table 4 Participant demographics by age group

	YAs (SD)	OAs (SD)
Age (years)	32.8 (6.8)*	74.9 (6.4)
Height (m)	1.73 (0.10)*	1.68 (0.09)
Body mass (kg)	84.6 (16.2)*	70.4 (19.2)
BMI (kg.m ⁻²)	27.02 (4.71)*	24.84 (4.78)
Dual task score	2.6 (1.1)	2.4 (1.2)

*indicates significant differences to OAs age group.

Age group and condition effect

Table 5 shows the results from the gait analysis for accelerating and decelerating jerk, step length, cadence and gait speed per condition by age group. Data for the single task perturbed and dual task perturbed conditions respectively are for the approach walk to the perturbation plate only.

The linear mixed model showed there was a significant effect of age group for accelerating jerk ($F(1,101) = 5.883$, $p = 0.017$, 95% confidence interval (CI) [-0.029, -0.003]), decelerating jerk ($F(1,101) = 5.139$, $p = 0.026$, 95% CI [0.001, 0.020]), cadence ($F(1,104) = 6.303$, $p = 0.014$, 95% CI [0.787, 6.702]), step length ($F(1,104) = 8.802$, $p = 0.004$, 95% CI [0.776, 0.882]) and gait speed ($F(1,104) = 4.819$, $p = 0.030$, 95% CI [1.450, 1.646]).

Table 5 Gait demographics from the four walking conditions.

	Age group	AJ (SD) (m.s ⁻³)	DJ (SD) (m.s ⁻³)	Step length (SD) (m)	Cadence (SD) (step.min ⁻¹)	Gait speed (SD) (m.s ⁻¹)
Normal	YAs	0.099 (0.041)	0.094 (0.026)	0.921 (0.147)	110.3 (7.35)	1.69 (0.25)
	OAs	0.073 (0.016)	0.092 (0.021)	0.839 (0.211)	111.4 (7.19)	1.55 (0.38)
Harness	YAs	0.092 (0.034)	0.086 (0.016)	0.904 (0.109)	110.0 (7.27)	1.66 (0.18)
	OAs	0.079 (0.023)	0.094 (0.028)	0.851 (0.195)	112.0 (9.08)	1.59 (0.37)
ST	YAs	0.085 (0.051)	0.070 (0.020)	0.960 (0.217)	107.8 (6.16)	1.73 (0.41)
	OAs	0.071 (0.022)	0.092 (0.029)	0.827 (0.197)	114.0 (8.59)	1.57 (0.39)
DT	YAs	0.081 (0.041)	0.080 (0.028)	0.971 (0.250)	106.5 (7.74)	1.73 (0.38)
	OAs	0.071 (0.02)	0.095 (0.024)	0.798 (0.204)	112.3 (9.39)	1.48 (0.46)

(AJ = accelerating jerk, DJ = decelerating jerk, ST = single task, DT = dual task)

For the normal and harness walking conditions and the perturbation plate approach gait for the single and dual task conditions there was no significant effect of condition in accelerating jerk ($F(3,101) = 0.578$, $p = 0.631$), decelerating jerk ($F(3,101) = 1.211$, $p = 0.310$), cadence ($F(3,104) = 0.255$, $p = 0.857$), step length ($F(3,104) = 0.037$, $p = 0.990$) or gait speed ($F(3,104) = 0.065$, $p = 0.978$). There was also no age group condition interaction for

accelerating jerk ($F(3,101) = 0.296, p = 0.828$), decelerating jerk ($F(3,101) = 1.192, p = 0.317$), cadence ($F(3,104) = 0.769, p = 0.514$), step length ($F(3,104) = 0.512, p = 0.675$) or gait speed ($F(3,104) = 0.288, p = 0.834$).

Perturbation plate

Of the 485 perturbations trials recorded, 198 (40.8%) met the threshold for classification as a perturbation trial; 110 in the single task condition and 88 in the dual task condition. For the trials that were excluded, the mean values for the plate motion in the single task condition were displacement 0.020 (0.017) m, velocity: 0.141 (0.098) m.s⁻¹, acceleration: 4.327 (3.425) m.s⁻². In the excluded trials in the dual task condition plate motions were displacement: 0.022 (0.015) m, velocity: 0.152 (0.099) m.s⁻¹, acceleration: 4.447 (3.016) m.s⁻². For the trials classified as perturbation trials, the mean plate motions in the single task condition were displacement: 0.102 (0.032) m, velocity: 0.425 (0.116) m.s⁻¹, acceleration: 14.189 (5.264) m.s⁻². For the dual task condition displacement: 0.107 (0.037) m, velocity: 0.439 (0.127) m.s⁻¹, acceleration: 14.540 (5.861) m.s⁻². Appendix C shows the significant difference between plate motions that classified as a perturbation and those that did not.

Perturbations

Table 6 shows values for accelerating jerk, decelerating jerk, step length and step speed for single and dual task perturbed conditions for the perturbed step, 1st recovery step and second recovery step. Appendix D has the distribution of perturbation displacement during the single task perturbation condition and the same in the dual task condition.

Accelerating jerk

There was no main effect of age group ($F(1,40.00) = 0.49, p = 0.489$) or step type ($F(3,33.62) = 1.85, p = 0.157$) for accelerating jerk. There was however a significant main effect of walk condition ($F(1,42.80) = 7.83, p = 0.008$), with accelerating jerk significantly higher in the single task condition compared to dual task.

There were no age group x step type ($F(3,33.6) = 1.49, p = 0.235$), age group x walk condition ($F(1,42.8) = 2.93, p = 0.094$), step type x walk condition ($F(3,33.06) = 1.06, p = 0.380$) or age group x step type x walk condition ($F(3,33.06) = 0.22, p = 0.883$) interactions for accelerating jerk.

Table 6 Gait variable mean and SD values for perturbed, first recovery and second recovery step by age group and condition

Step	Task	Age group	AJ (SD) (m.s ⁻³)	DJ (SD) (m.s ⁻³)	Step length (SD) (m)	Step speed (SD) (m.s ⁻¹)
Perturbed	ST	YAs	0.104 (0.145)	0.146 (0.063)	0.99 (0.29)	1.79 (0.58)
		OAs	0.119 (0.071)	0.110 (0.041)	0.91 (0.34)	1.69 (0.72)
	DT	YAs	0.065 (0.034)	0.092 (0.049)	1.04 (0.34)	1.89 (0.73)
		OAs	0.105 (0.043)	0.101 (0.046)	0.96 (0.35)	1.78 (0.62)
1 st recovery	ST	YAs	0.125 (0.088)	0.130 (0.063)	0.98 (0.363)	1.96 (0.72)
		OAs	0.101 (0.054)	0.119 (0.054)	1.03 (0.492)	2.06 (0.97)
	DT	YAs	0.071 (0.053)	0.093 (0.039)	1.14 (0.374)	2.28 (0.77)
		OAs	0.086 (0.040)	0.075 (0.047)	1.00 (0.441)	2.02 (0.77)
2 nd recovery	ST	YAs	0.111 (0.060)	0.157 (0.136)	0.98 (0.364)	1.79 (0.56)
		OAs	0.104 (0.072)	0.123 (0.069)	0.91 (0.280)	1.76 (0.58)
	DT	YAs	0.090 (0.062)	0.136 (0.056)	1.07 (0.293)	1.90 (0.54)
		OAs	0.108 (0.059)	0.135 (0.053)	0.95 (0.312)	1.82 (0.61)

(AJ = accelerating jerk, DJ = decelerating jerk, ST = single task, DT = dual task)

Decelerating jerk

There was no main effect of age group for decelerating jerk ($F(1,26.90) = 1.09$, $p = 0.306$). However, there was a significant effect of step type ($F(3,25.20) = 11.58$, $p < 0.001$) and walk condition ($F(1,18.20) = 8.88$, $p = 0.008$) on decelerating jerk. Decelerating jerk was larger in single task compared to dual task walking. During normal walking decelerating jerk was significantly smaller than the perturbed step ($p = 0.048$, CI [-0.051, 0.000]) and second recovery step ($p < 0.001$, CI [-0.047, -0.003]) and decelerating jerk was smaller in the first recovery step than the second recovery step ($p < 0.001$, CI [-0.056, -0.012]).

There were no age group x step type ($F(3,25.20) = 1.83$, $p = 0.168$), age group x walk condition ($F(1,18.20) = 2.17$, $p = 0.158$), step type x walk condition ($F(3,28.70) = 2.94$, $p = 0.050$) or age group x step type x walk condition ($F(3,28.70) = 1.11$, $p = 0.361$) interactions for decelerating jerk.

Step length

There was no main effect of age group ($F(1,123.90) = 1.01$, $p = 0.318$) or walk condition ($F(1,126.70) = 0.66$, $p = 0.418$) for step length. However, there was a significant effect of step type ($F(3,100.50) = 186.85$, $p < 0.001$). Normal walking step length was significantly smaller than the perturbed step ($p < 0.001$, CI [-1.487, -0.959]), first recovery step ($p < 0.001$, CI [-1.827, -1.200]) and second recovery step length ($p < 0.001$, CI [-1.466, -1.035]). However, there was no difference in step length between any of the perturbed and recovery steps.

There were no age group x step type ($F(3,100.50) = 0.02, p = 0.998$), age group x walk condition ($F(1,126.70) = 0.42, p = 0.518$), step type x walk condition ($F(3,101.40) = 0.23, p = 0.876$) interactions for step length.

Walk speed

There was no main effect of age group ($F(1,27.70) = 1.27, p = 0.269$), walk condition ($F(1,109.50) = 1.13, p = 0.290$) or step type ($F(3,38.40) = 1.47, p = 0.239$) for walk speed.

There was also no age group x step type ($F(3,38.4) = 0.07, p = 0.976$), age group x walk condition ($F(1,109.50) = 0.35, p = 0.553$), step type x walk condition ($F(3,58.0) = 0.68, p = 0.566$) or age group x step type x walk condition ($F(3,54.9) = 0.45, p = 0.752$) interactions for walk speed.

Discussion

The aim of this study was to identify differences in ML jerk values in healthy adults during over ground normal walking, perturbed walking and dual-task perturbed walking. There was seen to be a significant effect of age group on the normal walking condition, but not the perturbed steps or recovery steps. Results described here found increased age significantly decreased accelerating jerk, decelerating jerk, step length, walking speed and increased cadence during normal walking. There was also significantly higher accelerating jerk in the single task condition compared to the dual task. Decelerating jerk was significantly higher in the perturbed and second recovery step than in normal walking. Decelerating jerk was also significantly smaller in the first recovery step compared to the second recovery step. Step length was significantly longer for the perturbed step, first and second recovery step compared to normal walking.

The lack of age group effects in the single and dual task condition suggests there was a comparable level of executive functioning across all participants in this study, whereas it would be expected that the OAs would have a lower level of functional capacity (Coppin *et al.*, 2006). This is further supported by other studies also finding an age effect in perturbation response (Mansfield and Maki, 2009). A similar population to that investigated here showed that OAs responded significantly slower to the perturbation compared to YAs (Martelli *et al.*, 2017). However, during the approach walks, OAs walked slower with shorter and more frequent steps when compared to the YAs. From previous literature, this is an expected difference as OAs are known to walk slower, with wider step width, higher cadence and shorter step length (Huijben *et al.*, 2018; Qiao, Feld and Franz, 2018; Belvederi Murri *et al.*, 2020). The lack of main effect of age and interaction effects seen here suggests that both age groups responded similarly to the perturbations for the outcome variables used. An explanation for the lack of age effect observed here is likely due to self-selection bias (Heckman, 2013). All participants were living independently within the community, while part of the selection criteria specified participants had to be able to walk unaided for 20

metres. Therefore, among the current population of healthy, community-dwelling OAs, the selected dual task was not sufficiently challenging to elicit age group differences.

Participants were instructed to walk at their preferred walking speed for the duration of the trials. Results showed no effect of condition on gait speed, but OAs walked significantly slower for the duration of the trials. Research shows preferred or self-paced walking speed is the most reliable means of data collection across populations and should be used to unify protocols as a means of gathering normative data (Plotnik *et al.*, 2015). No effect of walking speed suggests both groups approached the perturbation task similarly and did not alter the overall speed of walking in response to the perturbation. This also shows both groups were able to maintain walking speed with the compensations made for the perturbation. This may have been due to the size of perturbation and functional ability of the participants not needing to slow to overcome the perturbation.

The increase in step length during the perturbed step showed both groups compensated for the perturbation by taking a longer step compared to normal gait. Step length then remained longer than normal walking for both recovery steps, supporting the idea that increased step length was a method of maintain balance. This can be linked back to the normal walking results, showing YAs walking with significantly longer step than OAs which is supported by previous research (Zhao and Wang, 2016; Kluft, van Dieën and Pijnappels, 2017). Increased step length is associated with increased dynamic control and improved balance (Huijben *et al.*, 2018; Qiao, Feld and Franz, 2018; Belvederi Murri *et al.*, 2020). This suggests that OAs here adopted the same coping mechanism as the YAs; resulting in a more stable gait when perturbed.

ML CoM jerk values were highest during the perturbed step; this is expected due to the perturbation being in the ML direction to the BoS, while disruption to the BoS is well documented to disrupt the CoM stability and smoothness of motion (Hamacher *et al.*, 2011). Gait smoothness is frequently measured by CoM jerk and can be used to determine the stability of the gait pattern (Krammer, Drew and Brown, 2021). Therefore, it would be expected that the only step the participants took that had a destabilising effect had the least smooth gait and highest jerk values. The differences in recovery steps seen shows that gait was still less stable than normal walking for two steps after the perturbed step. This could suggest that more than two steps need to be monitored to allow for full recovery response to be collected. The lack of effect of step type on accelerating jerk suggests participant's response was to decrease lateral CoM momentum prior to peak ML displacement by controlling the deceleration phase of gait only in order to better overcome perturbations. The push off away from the peak ML displacement was unaffected.

Decelerating jerk was significantly higher for the perturbed step and first recovery step in the single task condition compared to the dual task condition, but was unchanged between

conditions for the second recovery step. This shows that in the perturbed step and first recovery step for the dual task condition, participants were walking with a smoother gait, likely an attempt to overcome any perturbation more successfully by increasing the dynamic control of their gait. While this will have increased the cognitive load of the dual task condition participants only had ten walks in this condition, as well as breaks between walks should they require them. The dual task was likely not challenging enough for this particular cohort to elicit a response. This links back to previous research where reduced gait speed was observed among participants with a lower level of executive function (Coppin *et al.*, 2006). The participants with lower levels of executive function were also more affected by increasingly difficult walking tasks. This suggests that all participants here were at the higher levels of executive functioning as they altered their gait in order to manage the counting and perturbation tasks successfully within the dual task condition showing minimal effect of the perturbation. The lack of age effect in this study is also supported by the results of the dual task paradigm; participants were given one point for every number they correctly calculated. There was no age group effect here, showing the OAs and YAs found the task equally challenging. This may also explain the lack of effect of age group in the other perturbation trials. Clearly, the OAs were relatively high functioning with good levels of executive function, perhaps compared to the wider OAs population (Schott and Krull, 2019). This, however, is only speculation as there was no measure of executive function taken.

The cut off of 200 ms before and after peak CoM ML displacement was designed to collect half of the step duration during normal walking as the CoM is decelerating to peak ML displacement and then accelerating away to the next step. Previous research has shown CoM motion is symmetrical with healthy adults peak vertical ground reaction force at 46% of the gait cycle (McAndrew Young, Wilken and Dingwell, 2012). Further support from gait segmentation has shown push off occurs at 52.5% of the gait cycle (Afschrift *et al.*, 2018). In this study, data collection began 200 ms before peak ML CoM displacement. The average step duration was 543 ms, so data collection began on average at 63.2 % of the gait cycle. Accelerating jerk data collection ended 200 ms after peak ML CoM displacement at 36.8% of gait cycle.

A number of researchers have found strong evidence for the first trial effect; the measurable rapid adaptive behaviour response change to a single postural perturbation (McIlroy and Maki, 1995; Adkin *et al.*, 2000; Marigold and Patla, 2002). This has been used by intervention studies as an attempt to reduce participant anxiety by only subjecting the individuals to a single perturbation per visit and has found some success with one study reporting participants fall incidence had been halved from single perturbation exposure (Pai *et al.*, 2010). These changes are thought to come about as a result of rapid changes to

control of postural stability and limits of stability (Wang *et al.*, 2011). However, there is less evidence of this among YAs (Lee *et al.*, 2018).

Limitations

As previously discussed, one of the key limitations of this study is the health of the OAs, potentially due to self-selection bias (Heckman, 2013). All participants volunteered for the study from within the local community. Residential homes were contacted but no volunteers came forward that were not independently living. Participants in this study were high functioning and without a fall history, suggesting higher levels of executive function (Coppin *et al.*, 2006). This high level of functioning for an older population in combination with an exposure of only 20 perturbations of displacement: 0.104 m, velocity: 0.431 m.s⁻¹ and acceleration: 14.345 m.s⁻² may not have been sufficient to challenge the participants' balance control to elicit a different response to the YAs. However, previous research has reported ML perturbations of 0.16m displacement, 0.316 m.s⁻¹ velocity and 1.3 m.s⁻² acceleration to be challenging yet safe levels of perturbation (McIntosh, Zettel and Vallis, 2017). The combination of the size of perturbation and health of the participants is likely one of the reasons there was not a significant age effect during perturbation response.

Further to this a consideration must be made for the fact that the perturbation plate did not always move enough to classify as a perturbation. The design of the plate was such that its motion was completely controlled by the participant. Therefore, if there was insufficient ML motion during the participants step onto and off the plate, it would not move. This could be addressed in future by implementing a motorised system into plate design such that the plate would always move a specific amount once participants had stepped onto the plate.

Conclusion

In this study, there was seen to be a significant effect of age group on the normal walking condition, but not the perturbed steps or recovery steps. The differences in normal walking were expected, however it was hypothesised that there would also be an age effect for perturbation response. The lack of age group effect during the perturbed trials suggests further investigation is needed into less healthy adult populations. These results suggest the OAs sampled were highly functioning as all were living independently and responded very similarly to YAs to the same perturbations. A sample of institutionalised OAs, or OAs with a fall history may provide a clearer insight into the ability of CoM ML jerk to identify fallers from non-fallers.

Chapter 5 – General discussion and conclusion

The aim of this thesis was to establish if CoM ML jerk during gait could be used to identify adults over the age of 65 years at risk of falling or with a history of falling. As highlighted in Chapter 1, there is an ever-growing ageing population in the UK, with more people living longer than ever before. This brings inherent challenges in the need for healthy ageing through maintenance of physical and mental health. One of the key issues facing OAs is gait stability and falls. Additionally, as previously discussed in Chapter 1, a fall can have serious physical consequences resulting in major injury or death (Anderson, 2008). Experiencing a fall can be extremely psychologically damaging by inducing or increasing an individual's fear of falling, causing reduced mobility and socialising (Nagarajan et al., 2020). Falls occur as a result of being unable to maintain balance often resulting from a perturbation (Hamacher *et al.*, 2011). Deconditioning as a result of ageing reduces individuals' ability to maintain balance in all situations. However, there is yet to be a complete and reliable or accurate measure of the level of deconditioning an individual has experienced and therefore their risk of falls.

Summary

This thesis was developed from three studies; the first, detailed in Chapter 2, an investigation of 834 participants from a larger observational cohort and cross-sectional study. Participant data was collected from an IMU located on the L4 and jerk analysis was performed. The aim was to identify whether CoM ML jerk could be used to distinguish between OAs with and without a fall history. Chapter 2 demonstrated that fallers did have significantly lower accelerating and decelerating jerk values compared to non-fallers during a 10 metre un-perturbed walk in a laboratory setting. However, there were very small effect sizes. The conclusion was drawn that participants may need to have their gait challenged in order for CoM ML jerk to be used to identify fallers within smaller study populations. It has also been stated in previous studies that perturbations are necessary to understand the balance control of individuals and are necessary to understand a person's stability and recovery capability (Madehkhaksar *et al.*, 2018).

As a measure of global gait balance and control, ML jerk may better elucidate group differences in challenging walking tasks such as perturbed conditions and dual task effects. These findings lead to Chapter 3, a systematic review of recent laboratory studies which perturbed participants during gait. The aim was to identify common methods of perturbing participants during gait and identify the key outcome variables used to assess their response. This review was conducted to inform the design of the subsequent experimental study in Chapter 4, finding that the most ecologically valid method of perturbation was during over ground walking, rather than during treadmill walking and the importance of appropriate selection of outcome measure to allow results to be fully represented. Chapter 4 combined

methods from the previous two studies and applied perturbations during over ground walking to healthy YAs and OAs in a laboratory setting across four conditions. The aim was to distinguish OAs from YAs based on the CoM ML jerk, and identify individuals potentially at risk of future falls. There were significant differences between age groups during normal walking conditions, however there was a lack of age group effect in the perturbation response across all conditions. This may have been due to the lack of fall history among the older participants, the health of the population, the size of the perturbation delivered or the difficulty level of the dual task.

Limitations

Due to the coronavirus pandemic this thesis took a very different path to that intended and which had been laid out in the initial study plan. This led to a number of challenges, mainly due to the lack of laboratory access in order to perform testing. There was also extreme reluctance from participants within the local community to participate in the project due to concerns of catching the virus. Had the gait assessment been portable or able to occur outside, this may have eased participant's fears and increased participation. However, due to the safety harness and the nature of the perturbation platform, transportation of the experimental set up was impossible. Only a small pilot study was able to be performed prior to data collection. While this was still beneficial, had the pilot study had a larger cohort there would likely have been other issues highlighted prior to data collection that were not identified. These included the finding that not all perturbation trials produced a large enough plate motion to produce a perturbation, as well as participants' ability to quickly learn methods to improve their dual task score.

A further limitation was the populations tested. The SABRE cohort was inherently healthy and independent in their living circumstances. This, despite their age, left the population with little range in functional ability. In spite of this, there were still some participants with a fall history and they did have significantly lower jerk than their non-falling counterparts. However, this limitation was also apparent among the participants of the laboratory perturbation study, all participants lived independently and did not have a fall history. This is likely significant in the lack of age group differences in the perturbation trials. The similar response between age groups to the perturbations is something needing further investigation as this was not hypothesised. As with the SABRE participants (Chapter 2), although OAs, they still maintained functional capacity and while it was not directly measured, results implied good levels of executive function. Had participants volunteered who were at a lower functional level, there may have been different results. Further to this, due to the limited laboratory access during the pandemic, a final group of participants with fall history was not able to be collected. This cohort had been intended to identify if CoM ML jerk could be used to distinguish fallers from non-fallers during perturbed gait.

Implications

Irrespective of the issues with the progress of the thesis, some interesting and promising results were obtained. Chapter 2 and 4 show CoM ML jerk can be used as an indicator of fall history and participant age, further suggesting jerk could be used to determine fall risk for the future. While jerk has been identified as an effective measure of smoothness in the past, this collection of studies suggests its utility as a measure of fall risk among OAs (Brodie *et al.*, 2014; Krammer, Drew and Brown, 2021). This collection of studies also supports previous research stating the importance of dynamic measures of stability during gait to assess fall risk and gait stability (Hamacher *et al.*, 2011). Wider application of dynamic gait assessments such as jerk in a perturbation setting could allow for early identification of fallers prior to the first fall. This could lead to the development of more effective exercise interventions to increase gait stability and successful balance perturbation response leading to a decreased fall risk and decreasing the impact of any future falls on physical condition. While jerk has not been shown here to clearly define fallers from non-fallers, further development of the methods and application of this measure could lead to better application as a fall risk identification methods.

Conclusion

This thesis aimed to explore the utility and application of jerk as a measure of gait stability, and a method of identification of future fall risk. Initial results agreed with previous research stating dynamic measures of stability were able to distinguish between fallers and non-fallers (Bierbaum *et al.*, 2010; Bhatt *et al.*, 2011; Hamacher *et al.*, 2011; Hak *et al.*, 2012; Chulvi-Medrano, Martínez-Ballester and Masiá-Tortosa, 2015; Ghahramani, 2016). Prior research suggested the introduction of perturbation during gait can help to distinguish fallers from non-fallers when using dynamic stability as a measure (Havens, Mukherjee and Finley, 2018). The most ecological method of perturbing participants was found to be at floor level during over ground walking, these perturbations best represent slips and trips experienced in daily life (Van Ancum *et al.*, 2019). However, application of this perturbation method did not agree with previous findings of this thesis or previous research. There is support for the perturbation size being insufficient to cause a significant response difference between YAs and OAs (McIntosh, Zettel and Vallis, 2017). This lead to the final conclusion of the thesis being the methods used here should be applied to a population of fallers to establish the true sensitivity of CoM ML jerk in fall risk detection. Additionally, a longitudinal follow up study would allow participants' jerk to be monitored over time and any falls to be recorded identifying more clearly if CoM ML jerk could be used to detect fall risk among OAs.

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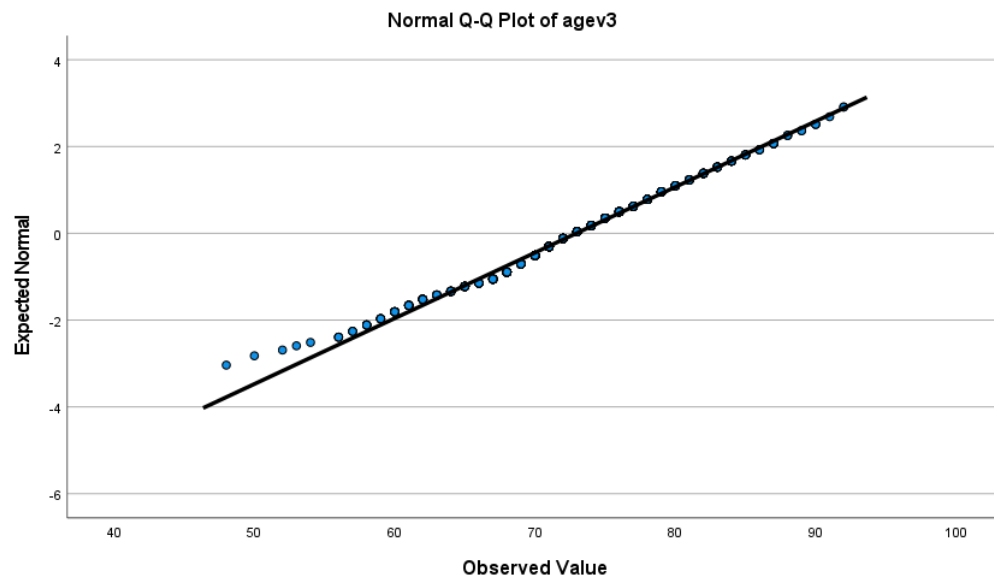
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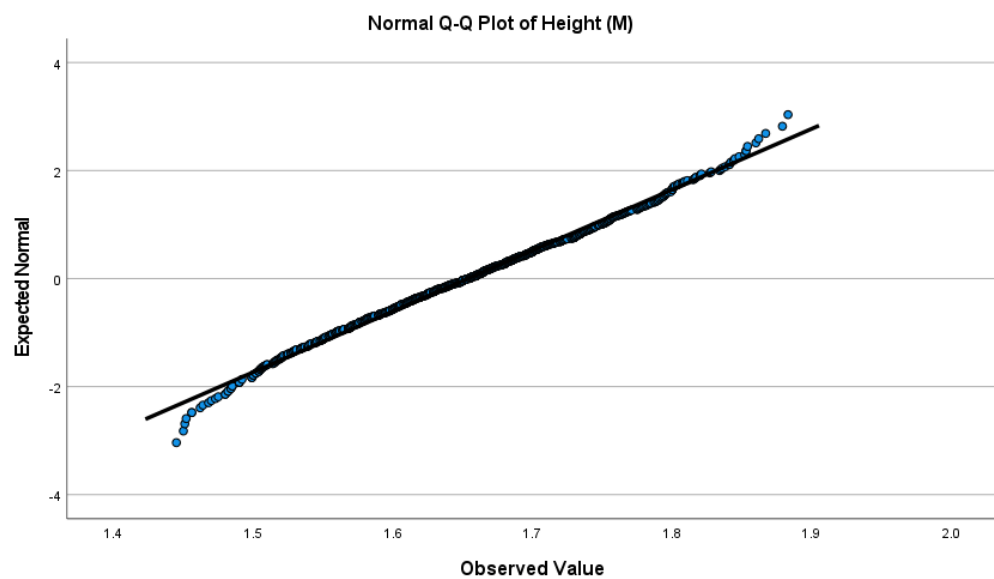
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Appendices

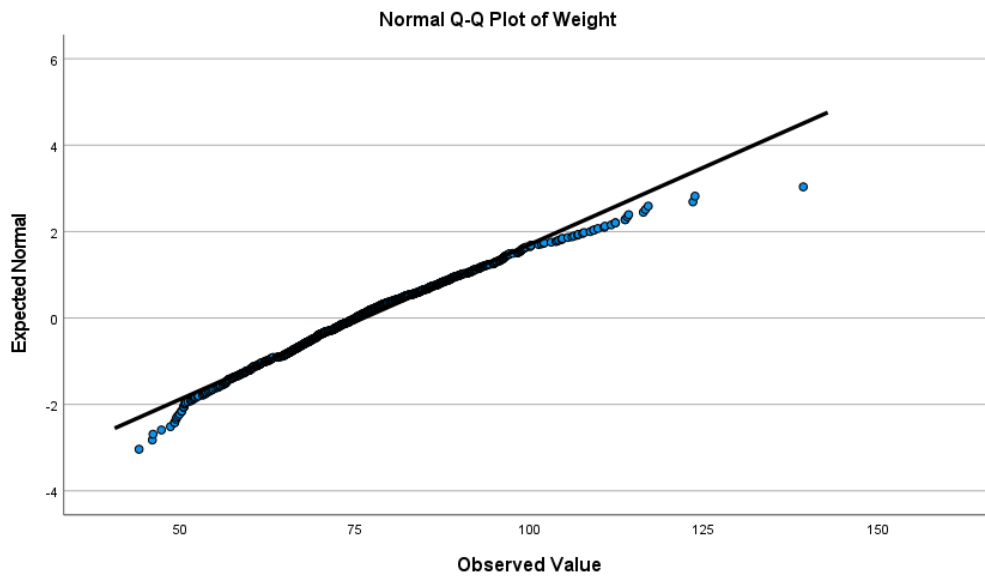
Appendix A – Q-Q plots for participant age, height, body mass, accelerating medio-lateral jerk and decelerating medio-lateral jerk.



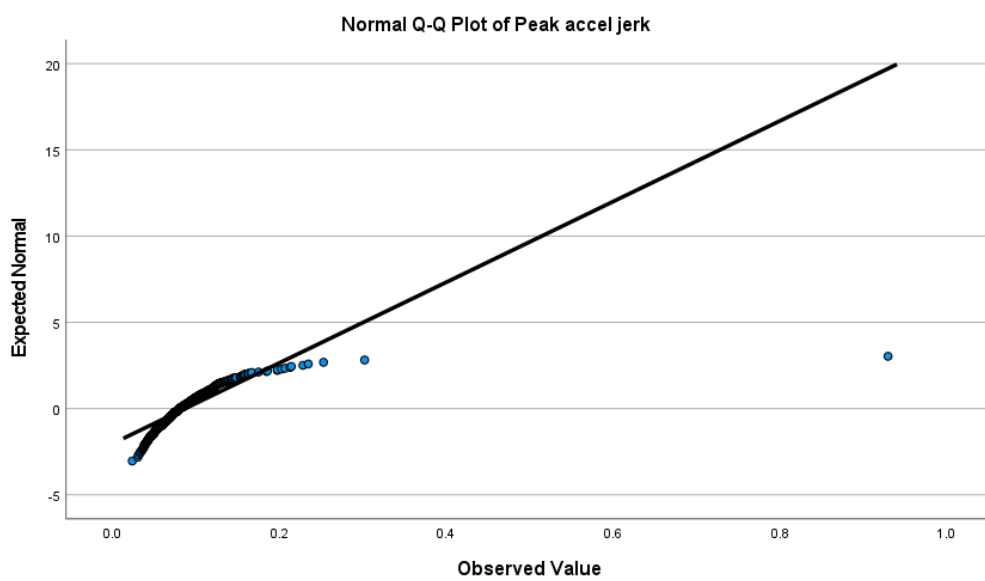
Q-Q plot of age (years).



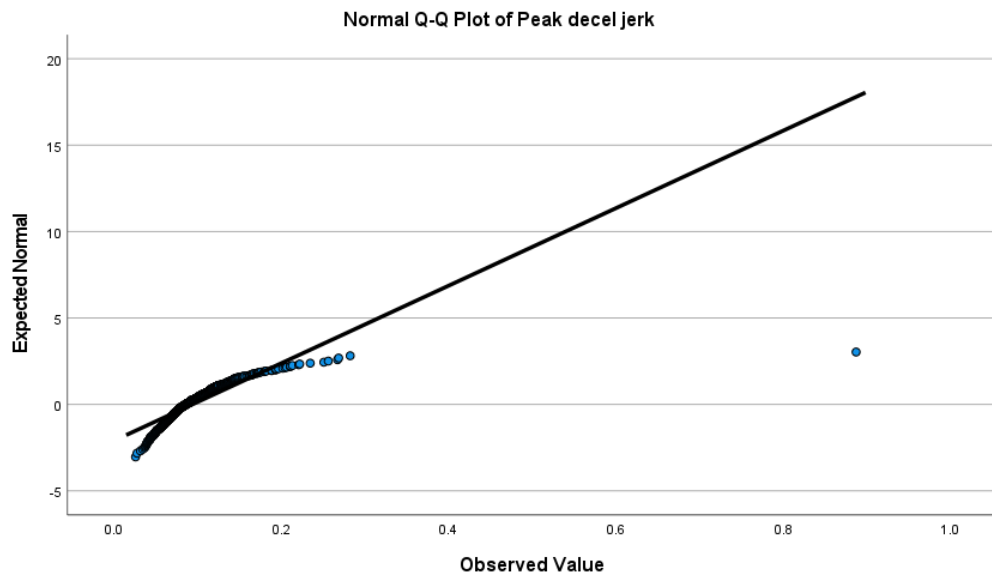
Q-Q plot of height (m).



Q-Q plot of body mass (kg).



Q-Q plot of accelerating jerk (m.s^{-3})



Q-Q plot of decelerating jerk (m.s^{-3})

Appendix B – AXIS risk of bias assessment results. Y – yes, N – no, U – unspecified, (Vervoort et al., 2020) is a preliminary trial paper.

Reference	1	2	3	4	5	6	7	8	9	10
(Aprigliano et al., 2015)	Y	Y	N	N	U	U	U	Y	Y	N
(Aprigliano et al., 2016)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Aprigliano et al., 2019)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Arvin et al., 2018)	Y	Y	N	Y	U	Y	U	Y	Y	Y
(Best et al., 2019)	Y	Y	N	N	U	U	U	Y	Y	Y
(Capin et al., 2017)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Eduardo Cofré Lizama et al., 2015)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Francis et al., 2015)	Y	Y	N	N	U	U	U	Y	Y	Y
(Haarman et al., 2017)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Hof and Duysens, 2018b)	Y	Y	N	N	U	U	U	Y	Y	Y
(Kurz et al., 2016)	Y	Y	N	Y	Y	Y	U	Y	Y	Y
(Madehkhaksar et al., 2018)	Y	Y	Y	N	U	U	U	Y	Y	Y
(Martelli, Kang and Agrawal, 2017)	Y	Y	N	N	U	U	U	Y	Y	Y
(Martelli et al., 2016)	Y	Y	N	N	U	U	U	Y	Y	Y
(McIntosh, Zettel and Vallis, 2017)	Y	Y	N	N	U	U	U	Y	Y	Y
(Onushko et al., 2019)	Y	Y	N	N	U	U	U	Y	Y	Y
(Punt et al., 2017)	Y	Y	N	Y	Y	Y	U	Y	Y	Y
(Rieger et al., 2020)	Y	Y	N	N	Y	Y	U	Y	Y	Y
(Roeles et al., 2018)	Y	Y	N	Y	U	Y	U	Y	Y	Y
(Rosenblum et al., 2020)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Sheehan et al., 2015)	Y	Y	N	N	U	U	U	Y	Y	Y
(Shulman, Spencer and Vallis, 2018)	Y	Y	N	N	U	U	U	Y	Y	Y
(van Hal et al., 2018)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Vervoort et al., 2020)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Vlutters, Van Asseldonk and Van Der Kooij, 2016)	Y	Y	N	Y	U	U	U	Y	Y	Y
(Zadravec, Olenšek and Matjačić, 2017)	Y	Y	N	N	U	U	U	Y	Y	Y

Reference	11	12	13	14	15	16	17	18	19	20
(Aprigliano et al., 2015)	N	Y	U	U	Y	Y	Y	Y	U	Y
(Aprigliano et al., 2016)	Y	Y	U	U	Y	Y	Y	N	Y	Y
(Aprigliano et al., 2019)	Y	Y	U	U	Y	Y	Y	N	Y	Y
(Arvin et al., 2018)	Y	Y	U	U	Y	Y	Y	Y	Y	Y
(Best et al., 2019)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(Capin et al., 2017)	Y	Y	U	U	Y	Y	Y	Y	U	Y
(Eduardo Cofré Lizama et al., 2015)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(Francis et al., 2015)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(Haarman et al., 2017)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(Hof and Duysens, 2018b)	Y	Y	U	U	Y	Y	Y	N	N	Y
(Kurz et al., 2016)	Y	Y	U	U	Y	Y	Y	Y	Y	Y
(Madehkhaksar et al., 2018)	Y	Y	U	U	Y	Y	Y	Y	U	Y
(Martelli, Kang and Agrawal, 2017)	Y	Y	U	U	Y	Y	Y	Y	U	Y
(Martelli et al., 2016)	Y	Y	U	U	Y	Y	Y	Y	U	Y
(McIntosh, Zettel and Vallis, 2017)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(Onushko et al., 2019)	Y	Y	U	U	Y	Y	Y	Y	U	Y
(Punt et al., 2017)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(Rieger et al., 2020)	Y	Y	U	U	Y	Y	Y	Y	Y	Y
(Roeles et al., 2018)	Y	Y	U	U	Y	Y	Y	Y	Y	Y
(Rosenblum et al., 2020)	Y	Y	U	U	Y	Y	Y	Y	U	Y
(Sheehan et al., 2015)	Y	Y	U	U	Y	Y	Y	N	U	Y
(Shulman, Spencer and Vallis, 2018)	Y	Y	U	U	Y	Y	Y	Y	N	Y
(van Hal et al., 2018)	Y	Y	U	U	Y	Y	Y	N	N	Y
(Vervoort <i>et al.</i> , 2020)	Y	U	U	U	U	U	U	U	N	Y
(Vlutters, Van Asseldonk and Van Der Kooij, 2016)	Y	Y	U	U	Y	Y	Y	N	N	Y
(Zadravec, Olenšek and Matjačić, 2017)	Y	Y	U	U	Y	Y	Y	Y	N	Y

Appendix C – The significant difference between plate motions that classified as a perturbation and those that did not for displacement, velocity and acceleration.

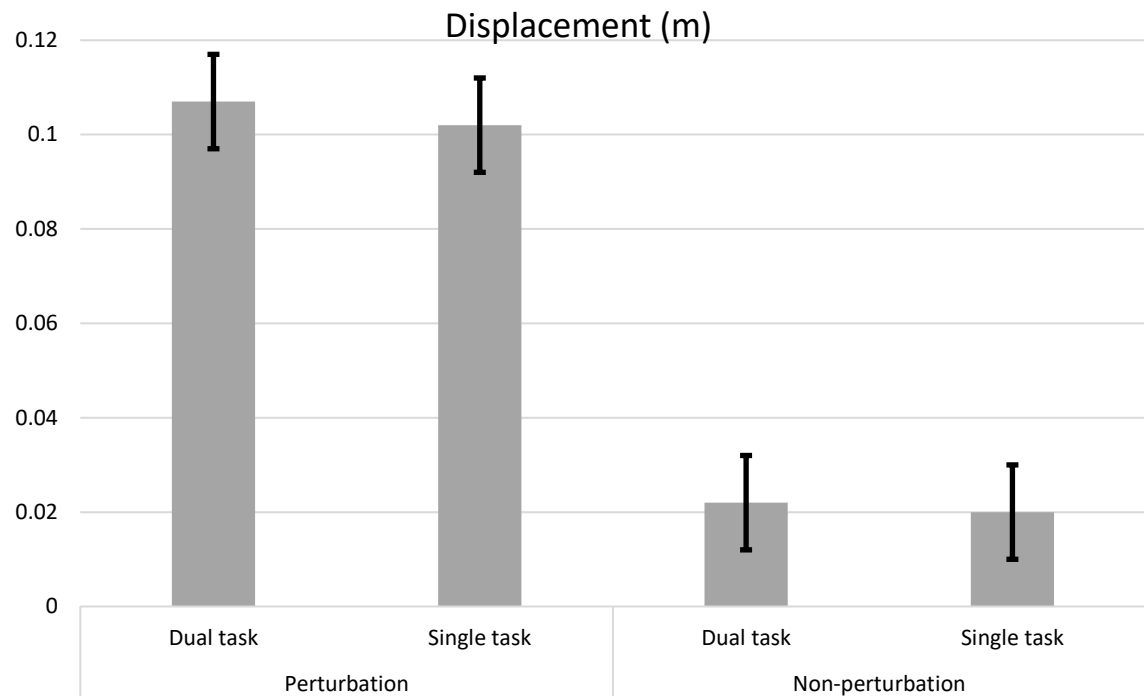


Plate displacement mean values in the dual and single task conditions. Non-perturbation is the trials that did not meet the 6% displacement threshold.

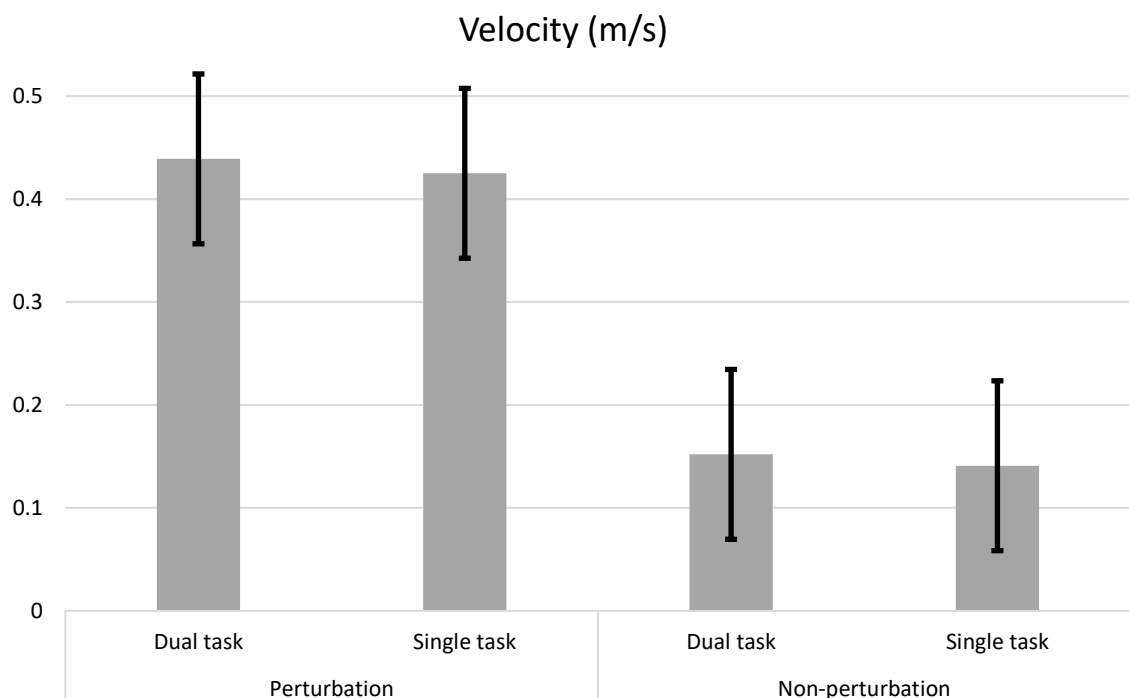


Plate velocity mean values in the dual and single task conditions. Non-perturbation is the trials that did not meet the 6% displacement threshold.

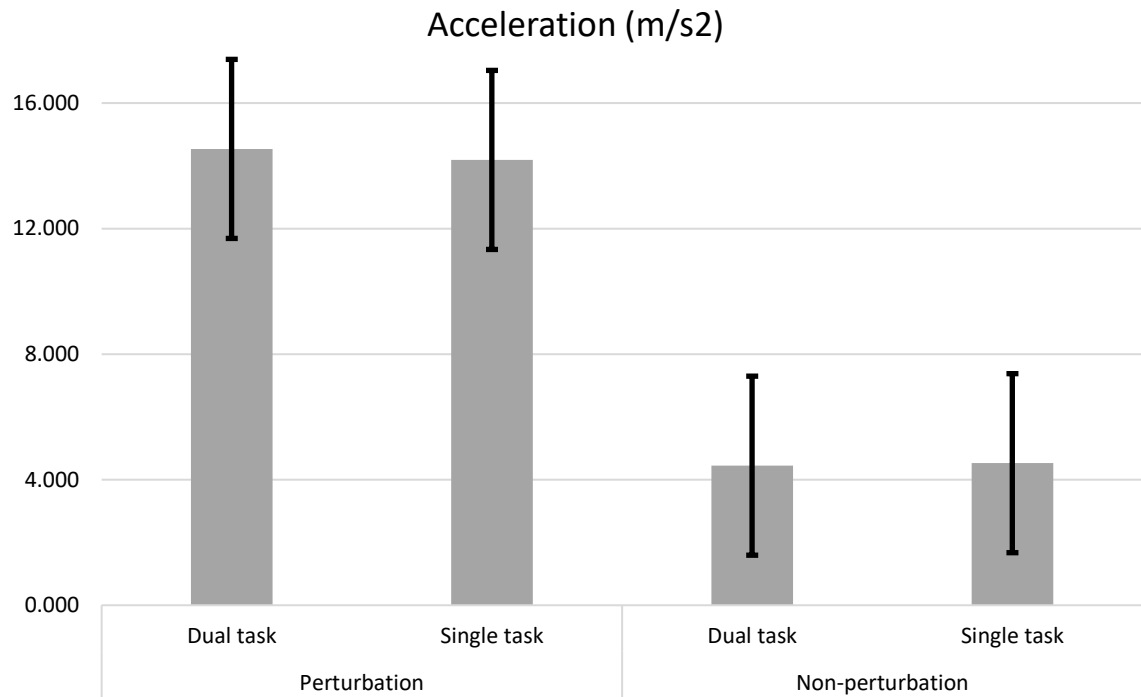


Plate acceleration mean values in the dual and single task conditions. Non-perturbation is the trials that did not meet the 6% displacement threshold.

Appendix D – All plate displacement values for single a dual task conditions.

