Department of Biomedical Engineering

# The development of a reactive gait assessment

## *Toward identifying risk for falls in older adults*

By

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## ABSTRACT

Falls are the leading cause of injuries in older adults. To prevent falls, early identification of individuals at risk is therefore needed. Although gait impairments are among the main risk factors, currently available clinical balance and gait assessments lack sensitivity in identifying risk for falls. Since most falls occur following trips or slips reduced ability to adequately recover from gait perturbations may be indicative of fall risk. The main aim of my industrial doctorate programme was to investigate whether reactive gait assessment can be used to identify risk for falls in older adults. I used innovative technology from the company Motek and developed a mixed-perturbation protocol to challenge the individual's gait pattern and measure the recovery responses in a safe, standardized and objective manner. The initial protocol and outcome measures were based on a literature review. Three subsequent phases of development and evaluation were used to deliver a standardized reactive gait assessment. Using this assessment I was able to reveal that older adults with a history of falls were more affected by contralateral sway and deceleration perturbation in terms of continuous trunk motion than those without a history of falls, while no differences between fallers and non-fallers were found in clinical and steady state gait measures. This suggests that reactive gait assessment has added value in fall risk identification in older adults. Therefore, I recommend using the contralateral sway and deceleration perturbations of our developed perturbation protocol and assessing responses by means of continuous trunk motion to further evaluate the use of reactive gait assessment for fall risk identification in older adults. Motek as an industrial company can facilitate this by providing customers with the developed reactive gait assessment and encourage clinically driven research to examine the reproducibility and validity of reactive gait assessment using evidence-based, affordable and easy-to-use technologies.

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| | |
|---|---|
| AP | Anterior-posterior |
| BoS | Base of Support |
| CAREN | Computer Assisted Rehabilitation ENvironment |
| CoM | Centre of Mass |
| CWS | Comfortable walking speed |
| D | Dominant |
| FES | Falls Efficacy Scale |
| FPE | Foot Placement Estimator |
| FSR | Feasible Stability Region |
| GRAIL | Gait Real-time Analysis Interactive Lab |
| GSN | Gait Sensitivity Norm |
| HBM | Human Body Model |
| LDS | Local Dynamic Stability |
| ML | Medio-lateral |
| MoS | Margins of Stability |
| ND | Non-dominant |
| OLST | One-legged stance test |
| PAQ | Physical Activity Questionnaire |
| PD | Pre-perturbed dominant step |
| PND | Pre-perturbed non-dominant step |
| SD | Standard Deviation |
| TUG | Timed Up&Go test |
| VT | Vertical |
| XCoM | Extrapolated Centre of Mass |

# CHAPTER 1

# Introduction

## 1.1 Falls in older adults: a health care problem

Falls are the leading cause of injuries in older adults (Ambrose et al., 2013). Over thirty percent of people of 65 years or older fall at least once a year (Rubenstein and Josephson, 2006). This percentage increases with age up to 50% in people older than 80 years of age (Stalenhoef et al., 1997; Stevens et al., 2006). Besides injury, a significant fall often results in a decreased quality of life and in addition may result in fear of falling. Fear of falling, in turn, may lead to a decrease in physical activity, causing muscle strength to deteriorate and hence increasing the risk of falling (Li et al., 2003). In addition to the physical and mental burden associated with falls, costs associated with falls in older adults are high as ten to 20% of these falls result in serious injuries in which people are treated in emergency departments or hospitalized. Annual costs associated with accident and emergency department attendance or hospital admission are estimated at €912 million in the Netherlands (VeiligheidNL, 2016) and £2 billion in the UK (Tian et al., 2013). All in all, falls in older adults are considered a major health problem in our ageing society.

**PROBLEM** Falls in older adults are considered a major health care problem due to the high incidence, severe consequences on quality of life and associated medical costs.

## 1.2 The balance control system: age-related impairments

To prevent falls we have to identify those at risk for falls and provide effective interventions in early stages. Falls are a multifactorial problem and often times there are several risk factors contributing to an increased risk for falls. Risk factors can be either intrinsic, such as age, gender and cognitive impairments, or extrinsic like hazards in the home environment, footwear and bad lighting. Among the main risk factors for falls are balance and gait impairments (Ambrose et al., 2013; Boelens et al., 2013; Rubenstein, 2006). Adequate balance control during standing and walking comprises a complex interaction between the sensory-, motor-and central nervous systems, which all show age-related functional decline. Here, we will describe the effect of ageing on these balance control subsystems.

### 1.2.1 The sensory system

The sensory system provides information on the position and movement of the body with respect to the environment or body segments relative to each other. This system consists of the visual, vestibular, and somatosensory systems. Humans typically rely most on the somatosensory system (70%) followed by the vestibular (20%) and visual (10%) system (Horak, 2006). However, depending on the context, the relative contribution of these systems to balance control may change. For example, the visual system will become less important when entering a dark room. While the ability to re-weight sensory information depending on the context is thus important, age-related functional decline in any of the sensory systems may result in an increased reliance on the other subsystems and hence impaired balance control.

Vision provides information about the world. By changes in the visual field, we perceive the orientation and movement of our body with respect to its environment, which is used to control posture and gait. The importance of vision on balance control can simply be demonstrated by executing balance tasks such standing on one leg or in a tandem stance with eyes open and eyes closed. Eyes closed conditions result in increased body sway which has been associated with falls in older adults (Lord et al., 1996, 1991; Paulus et al., 1984). In addition, vision provides information about potential environmental hazards such as loose tiles and is therefore required to prevent falling. Vision declines with ageing due to physiological changes of the eye leading to misjudgement in depth perception, reduced contrast sensitivity, reduced acuity thereby increasing risk for falls due to impaired obstacle avoidance and increased risk at tripping (Black and Wood, 2005; Klein et al., 2003; Salonen and Kivelä, 2012).

The vestibular system provides information about the orientation of the head. As we age, loss of sensors in the vestibular organs result in decreased corrective reflexes to stabilize the head (Sturnieks et al., 2008; van Dieën and Pijnappels, 2017). However, there is no clear relationship between vestibular function loss due to ageing and falls (Sturnieks et al., 2008).

The somatosensory system includes proprioceptive and exteroceptive information. The proprioceptive system compromises information coming from muscle spindles, Golgi tendon organs and joint mechanoreceptors, which together provide information on joint orientation and movement. Muscle spindles, located between the muscle fibres provide information on the (rate of) change in muscle length. Golgi tendon organs are located between at the muscle-tendon interface and thereby provide information on the tension on the muscle. The joint mechanoreceptors provide information on the position and movement of the joint. Proprioceptive information in the lower limbs is essential to maintain balanced, for example when standing on one leg or ensuring proper foot placement during gait. The number of sensors in the proprioceptive system reduce with ageing, while the sensitivity of the remaining sensors decreases hence reducing joint position sense (Shaffer and Harrison, 2007; van Dieën and Pijnappels, 2017) and therefore impairments in this system have been associated with falls (Lord et al., 1999). In addition, reduced sensitivity in sensory feedback (exteroceptive information) from the soles of the feet result in affected ability to detect changes in centre of pressure and are associated with impaired balance control (Menz et al., 2005; Shaffer and Harrison, 2007).

### 1.2.2 The motor system

The effect of ageing on the motor system is associated with reduced muscle strength and power, lower rate at which muscle power is developed, decreased precision in force generation and reduced force transmission from the muscles to the skeleton (Sturnieks et al., 2008; van Dieën and Pijnappels, 2017). Decrease in muscle strength and power is attributed to loss of muscles fibres as well as denervation due to chronic inflammation of motor neurons and axons

(Manini et al., 2013). Lower extremity muscle weakness is associated with falls, recurrent falls and injurious falls (Moreland et al., 2004). Given that the fast-twitch type II muscle fibres are primarily affected, not only muscle strength but also the ability to rapidly generate force declines with age (Lexell et al., 1988). Subsequently, older adults are at increased risk for falls when fast responses are required such as during obstacle avoidance tasks (Schultz et al., 1997) or when being tripped (Pijnappels et al., 2008). Loss of motor units reduces precision in muscle force generation and could explain falls in older adults (Carville et al., 2007). Finally, tendon stiffness reduces with ageing leading to slower force transmission from the muscle to the skeleton resulting in deficits in dynamic balance control (Karamanidis et al., 2008).

### 1.2.3 The central nervous system

The central nervous system integrates information from the sensory and motor systems and facilitates balance control on three levels: 1) the spinal cord, 2) the cerebellum and 3) the cortex. Spinal reflexes initiate muscle activity around the joint as a result of muscle stretches induced by unexpected perturbations. These responses are fast and occur about 50ms after a balance perturbation (Welch and Ting, 2014). Larger responses are initiated by the cerebellum as a result of deviation in centre of mass (CoM) kinematics. These occur later, about 100 ms after the perturbation, and the effect on the CoM can take up to 200ms due to electromechanical delay (Welch and Ting, 2014). In addition, sensory information is integrated in the cerebellum which is required to determine the orientation of the body in the environment and hence initiate adequate stabilizing motor responses (van Dieën and Pijnappels, 2017). Processing of visual information, anticipatory muscle activation and inhibition of the initially planned movement are all regulated by the cortex. The (pre)frontal cortex plays an important role in movement planning such as stepping over an obstacle and is therefore essential to prevent loss of balance. Ageing is associated with loss of grey and white matter which hampers sensory integration, resulting in reduced balance control and increased risk for falls (Papegaaij et al., 2014; Sullivan et al., 2000).

**FALL RISK FACTORS** Among the main risk factors for falls in older adults are balance and gait impairments as a result of age-related functional decline in the sensory-, motor- and central nervous systems.

### 1.3 Fall risk assessment: the current situation

Given that balance and gait impairments are among the main risk factors for falls, balance and gait assessment is considered an important aspect in fall risk identification. At present, guidelines for clinical practice state that balance and gait assessment should be performed when an individual has a) experienced a fall in the past year, and/or b) has difficulties with gait or balance. Such assessments typically consist of quick and easy-to-perform tests like the Timed Up & Go test (TUG) (Podsiadlo and Richardson, 1991) or Short Physical Performance Battery (Guralnik et al., 1994). However, currently available clinical balance and gait assessments lack sensitivity in identifying risk for falls, especially in active older adults (Barry et al., 2014; Gates et al., 2008; Laessoe et al., 2007). Hence, there is a need to develop objective and more sensitive assessment tools to evaluate potential balance and gait impairments as an indicator of risk for falls in older adults.

**FALL RISK IDENTIFICATION** Currently available balance and gait assessments are insensitive and fail to identify risk for falls in older adults.

Most falls occur while walking, due to trips or slips (Berg et al., 1997; Robinovitch et al., 2013; Talbot et al., 2005). This is not surprising, given that walking is one of the most common and often challenging activities in our everyday life which requires adequate balance control.

Gait stability has been defined as “gait that does not lead to falls in spite of perturbations” (Calandre et al., 2005). This is a broad definition as perturbations can come from various sources. To prevent falling while walking we need to be able to handle: 1) internal perturbations like neuromuscular noise, 2) expected perturbations such as walking on an uneven surface or avoiding an obstacle, and 3) unexpected perturbations like recovering from trips or slips (Bruijn et al., 2013a). The first two aspects require *proactive* adaptations in the gait pattern, whereas the latter requires fast and accurate *reactive* responses. Adequate reactive responses rely more heavily on the balance control system compared to proactive adaptations as there is no room for compensatory strategies. Assessment of the ability to resist and recover from gait perturbations, here referred to as “reactive gait assessment” may thus have an important contribution in identifying risk for falls in older adults.

**MAIN AIM** To investigate whether reactive gait assessment can be used to identify risk for falls in older adults.

## 1.4 Reactive gait assessment: a technical solution?

Human gait has been studied using instrumentation for almost one and a half centuries. The first studies used photographs to analyse gait kinematics and took place around the 1870s. As technology improved, photo cameras were replaced by dedicated motion capture cameras. The first force plates were introduced in the 1970s allowing for kinetic analyses (Whittle, 1996). Over the last two decades technology has improved exponentially resulting in the development of instrumented split-belt treadmills, motion platforms, high performance PCs and virtual reality environments. The introduction of these technologies provide us with experimental setups to challenge the individual’s gait pattern in a safe and standardized manner, as well as to accurately measure the individual’s recovery responses. A reactive gait assessment should consist of a perturbation type that significantly affects gait stability and a sensitive outcome measure to quantify the perturbation response.

Motek (Amsterdam, The Netherlands) provides high-end rehabilitation products that combine abovementioned hardware. Their software package, D-Flow, synchronizes data streams in real-time thereby providing the opportunity to develop a reactive gait assessment (Geijtenbeek et al., 2011).

**TECHNICAL SOLUTION** Innovative technologies allow us to challenge an individual's gait pattern and measure the recovery responses in a safe, standardized and objective manner.

### 1.5 Bridging the gap: toward clinical practice

This thesis was written as part of an industrial doctorate programme performed at Motek and supervised by the University of Strathclyde and the Vrije Universiteit Amsterdam. The experimental studies performed in this project were carried out at the University of Strathclyde, which houses one of Motek's high-end rehabilitation systems: the Computer Assisted Rehabilitation ENvironment (CAREN) Extended, and at Motek using the Gait Real-time Analysis Interactive Lab (GRAIL) systems.

The work presented in this thesis reflects a sequence of technical and experimental developments toward an evidence-based, affordable and easy-to-use protocol for reactive gait assessment to identify risk for falls in older adults and facilitate clinical acceptance (Rowe, 2012). The first step in this endeavour was to perform a literature review (Chapter 2), providing the rationale for the aims and outline of objectives (Chapter 3) of the developmental and experimental steps to follow (Chapters 4-7), leading to contributions and recommendations (Chapter 8).

Given the industrial nature of this thesis, the development of the reactive gait assessment was constrained to the specifications of available Motek equipment. Moreover, we were to warrant clinical feasibility in our decision-making process.

**CHALLENGE** A reactive gait assessment needs to be evidence-based, affordable and easy-to-use in order to be adopted in clinical practice.

# CHAPTER 2

# Reactive gait assessment

## 2.1 Introduction

Over the past few decades, many have studied the response to gait perturbations. Recovery from perturbations requires fast and accurate responses. This differs from steady state gait assessment where one is not challenged but rather moves within ones limits of stability. Quantifying the response to standardized gait perturbations may therefore be of added value when evaluating gait impairments in older adults.

Early gait perturbation studies primarily focussed on reflective responses and muscle activity following the perturbation (Berger et al., 1984; Figura and Felici, 1986; Gollhofer et al., 1986; Nashner, 1980; Schillings et al., 1996; Woollacott and Tang, 1997). A leap forward in understanding the ability to recover from perturbations was made by the group of Grabiner, who were the first to analyse the biomechanical effects of trip perturbations (Brady et al., 2000; Grabiner et al., 1993; Pavol et al., 2001, 1999a). More recently, there has been growing interest in understanding the effect of perturbations on gait stability providing insight in the extent to which the gait pattern is affected and how the ability to regain a stable gait pattern differs among individuals (Bhatt et al., 2005; Hof et al., 2010; Yang et al., 2012).

As the body of work on responses to gait perturbations grows, so does the range of experimental setups that have been used. The majority of perturbations mechanically affect the gait pattern by means of collision with obstacles, platform movement, slippery surfaces, treadmill belt accelerations and decelerations, and waist-pulls and -pushes. While this improves our understanding of the requirements for successful recovery responses, differences in the selected perturbation types and outcome measures make it difficult to directly compare results. Therefore, it is currently unknown which perturbation types should be included in a reactive gait assessment to identify risk for falls in older adults. Developing a perturbation protocol including various perturbation types and evaluating it in older adults is a critical step toward examining the feasibility of reactive gait assessment for identification of risk for falls in the ageing population. To this end, a literature review was performed to provide a rationale for the selection of outcome measures and perturbation types to be included in such a perturbation protocol.

The outline of this section is as follows: first, requirements to prevent falling are discussed. Based on these requirements a selection of potential gait stability measures to quantify the perturbation responses is given. Next, a summary of experimental setups to induce balance loss and the recovery responses to these perturbations is provided. The emphasis of the literature review is on: 1) the biomechanical effects of the perturbations on the ongoing gait pattern, 2) the effect of ageing on the success of recovery responses and 3) differences in the gait pattern associated with (un)successful recovery from perturbations. Finally, we suggest appropriate outcome measures and perturbation types to be included in our perturbation protocol.

## 2.2 Biomechanical requirements to prevent falling

Traditionally, one is considered balanced when the CoM lies within the base of support (BoS) (Winter et al., 1990) (Figure 2.1 – left panel). However, Pai and Patton (1997) recognized the importance of the CoM velocity when evaluating stability. When the CoM is located within the BoS, but exhibits a velocity directed toward the border of the BoS, balance may be compromised (Figure 2.1 – mid panel). Conversely, the CoM position may be located outside the BoS, but with a significant velocity towards the BoS and hence one can be considered stable (Figure 2.1 – right panel).

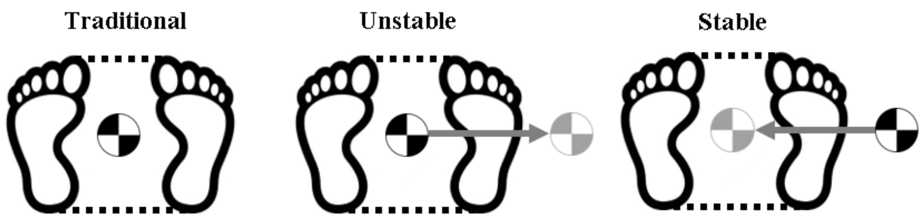



**Figure 2.1** Concept of stability. Traditionally one is considered stable when the centre of mass (CoM) is located within the boundaries of the base of support (BoS) (left panel). However, high outward velocity of the CoM may compromise balance even when the CoM is well within the BoS (mid panel). Contrary, when the CoM is located outside the BoS, one may be considered stable when the CoM is directed inward (right panel).

The CoM velocity thus needs to be taken into account when evaluating stability, especially in dynamic conditions such as walking, where it is likely have a considerable velocity. The extrapolated centre of mass (XCoM) does exactly this by correcting the CoM position for its velocity. During gait, added complexity in defining stability comes from the changing BoS caused by alternation between single- and double-support phases. In fact, we are often unstable in the forward direction (i.e. XCoM exceeding the forward border of BoS) facilitating forward progression. Yet, most of the time we walk around without falling. As such, gait has also been considered a “continuous state of falling and recovering” (Patla, 2003) and hence the requirement for stable gait seems to be the ability to control the XCoM within the changing BoS.

**CONCLUSION** A gait stability measure should capture the relation between the centre of mass motion state and the base of support.

## 2.3 Reactive gait stability measures

This section provides an overview of currently available gait stability measures that capture the aforementioned requirements to prevent falling, i.e. that is the ability to control the XCoM within the changing BoS. We aimed to select a biomechanically sound, yet clinically feasible outcome measure to quantify response to perturbations. A recent review by Bruijn and colleagues (2013a) divided gait stability measures in three categories: 1) measures that quantify the ability to recover from small internal perturbations like neuromuscular variability, 2) measures that quantify the ability to recover from larger external perturbations (i.e. the type of perturbations we intend to include in our perturbation protocol) and 3) measures that quantify the maximum perturbation an individual is able to recover from. Based on this review we discuss the following gait stability measures: the XCoM, the foot placement estimator (FPE) and the gait sensitivity norm (GSN). The XCoM has been used in comparable yet different stability measures: the feasible stability region (FSR) and the margins of stability (MoS) which will both be discussed. In addition, we include stabilizing and destabilizing forces. This measure was categorized by Bruijn and colleagues to describe small internal perturbations, but has since also been used to evaluate perturbed gait (Dubreucq et al., 2017; Ilmane et al., 2015).

### 2.3.1 Extrapolated centre of mass

As described above the XCoM is a spatial measure capturing the CoM motion state by extrapolating the CoM position in the direction of its velocity. It can be calculated as follows:

$$XCoM = CoM_{pos} + \frac{CoM_{vel}}{\sqrt{\frac{g}{l}}}$$ **Equation 2.1**

in which $CoM_{pos}$ is the position of the CoM in one plane and $CoM_{vel}$ the velocity in the corresponding plane, $l$ is the length of the inverted pendulum and $g$ the acceleration of gravity.

The XCoM was introduced by Hof and colleagues (2005), but built on the work of (Townsend, 1985) and Pai & Patton (1997). Based on the XCoM concept, Pai and Hof developed the feasible stability region (FSR) and margins of stability (MoS), respectively.

#### 2.3.1.1 Feasible stability region

The feasible stability region (FSR) defines a range in the anterior-posterior (AP) direction in which an individual is considered stable given a certain combination of COM velocity and foot position. Outside this range a fall is initiated either in the forward or backward direction, depending on which boundary is exceeded (Figure 2.2). The further away from the boundaries of the FSR, the higher the degree of instability.

The boundaries for forward and backward balance loss were initially defined using a modelling approach to evaluate static balance control. The model was later modified to evaluate gait stability following forward slips, using a 7-segment model (left and right feet, shanks and thighs and one segment including the head, arms and trunk) and a set of assumptions (Yang et al., 2008).

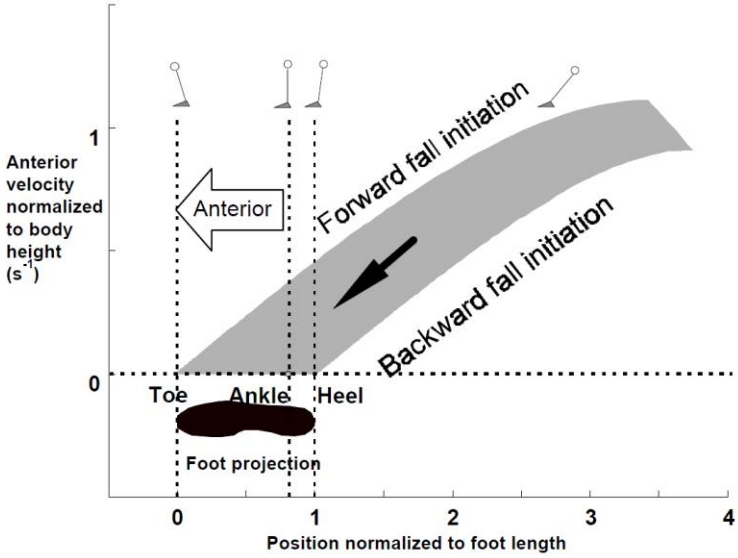



**Figure 2.2** Feasible stability region (FSR) for anterior-posterior stability. When a given combination of foot position (x-axis) and centre of mass velocity (y-axis) exceeds the upper boundary of the shaded area, a forward fall will be initiated. Whereas exceeding the lower boundary initiates a backward fall. Adapted from (Pai and Patton, 1997)

The group of Pai have used the FSR for over a decade to evaluate stability following slip perturbations during sit-to-stance transfers (e.g. Yang et al., 2009), slip perturbations during overground (e.g. Bhatt et al., 2011b) and treadmill walking (e.g. Liu et al., 2015). Their large body of work has for example demonstrated that higher gait speed is more beneficial to resist forward slips but reduced stability at lower speeds can be compensated for by using appropriate stepping strategies (Bhatt et al., 2005; Espy et al., 2010) and that reduced gait stability was associated with unsuccessful recovery of forward slips (Bhatt et al., 2011a). Nonetheless, to our knowledge the FSR has not been adopted by other research groups. Possibly because of the introduction of the margins of stability measure, which is based on the same principles as the FSR but follows more simple biomechanical reasoning (Hof et al., 2005).

#### 2.3.1.2 Margins of stability

The margins of stability (MoS) evaluates the distance between the XCoM and the border of the BoS. Depending on the direction of interest, MoS can either be defined in the forward, backward or lateral direction. If the XCoM lies within the corresponding BoS an individual is considered stable, outside the BoS a fall is initiated (Figure 2.3). In contrast to the FSR, MoS is based on an inverted pendulum model, which makes the model simpler in terms of calculations and also allows for reduced kinematic data to be used. For example, estimating CoM motion based on four pelvic markers as opposed to a full body marker set (Hak et al., 2013b) or centre of pressure data (Hof et al., 2007).

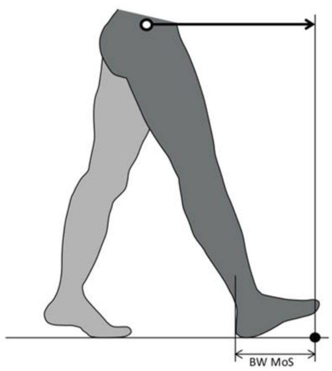



**Figure 2.3** Schematic representation of backward margins of stability (MoS). See text for a detailed explanation. Adapted from (Hak et al., 2013a).

The use of MoS to evaluate gait stability has become very popular. Not only has it been used to evaluate stability in various patient populations such as lower-limb amputees (Curtze et al., 2011, 2010; Gates et al., 2013; Hak et al., 2014, 2013d; Hof et al., 2007), stroke (Kao et al., 2014) or multiple sclerosis (Peebles et al., 2016) patients, but also various walking conditions like overground and treadmill walking (Rosenblatt and Grabiner, 2010), or the effect of cell phone use during walking (Kao et al., 2015; Marone et al., 2014). Moreover, responses to gait perturbations like continuous platform and visual oscillations (Beltran et al., 2014; Hak et al., 2013c; McAndrew Young et al., 2012), AP treadmill belt perturbations (Aprigliano et al., 2015; McCrum et al., 2016; Punt et al., 2017; Süptitz et al., 2013) or waist-pulls and pushes (Hof et al., 2010; Vlutters et al., 2016) have been studied using the MoS.

### 2.3.2 Foot placement estimator

The foot placement estimator (FPE) was introduced in robotics. Its rationale comes from the concept that gait is a state of falling and recovery (Wight et al., 2008). The FPE determines where the foot needs to be placed in order to recover balance using an inverted pendulum model (Figure 2.4c). In short, the total body angular velocity is used to determine at which leg angle the body would come to a standstill (i.e. when the energy of the body after foot placement equals apex potential energy). From this angle, it is calculated where the foot needs to be placed. If the foot is placed posterior to this point, the pendulum will fall forward (Figure 2.4a). When placed anterior to this point the pendulum will fall backward, onto the swing leg (Figure 2.4b).

Only a few studies using the FPE in human gait have been conducted. Two sensitivity analyses were performed to evaluate whether the assumptions for the FPE were violated. Since human gait involves controlled weight acceptance and push-off, the assumption that body angular momentum is conserved did not hold (Millard et al., 2009). The error induced by this violation was smaller than differences found in foot placement at various gait speed, indicating that the FPE is a sensitive measure. Additionally, this study showed that the foot was placed further behind the FPE at higher walking speeds but not during normal and slow walking, allowing individual's to maintain forward progression. In line with this, the effect of different gait speeds on the FPE in children with cerebral palsy (CP) were larger compared to the differences caused by violation of the FPE assumptions (Bruijn et al., 2013b). However, contrary to healthy young adults, changes in leg length rather than angular momentum caused the largest error. Another recent study, used the FPE to investigate the relation between arm posture and step width in children with CP (Meyns et al., 2016). The authors argued that altered arm posture in children with CP is a compensation for their reduced gait stability. Surprisingly, however, the use of FPE to quantify gait stability was not discussed and presented as a golden standard despite the limited experimental studies that have been performed so far.

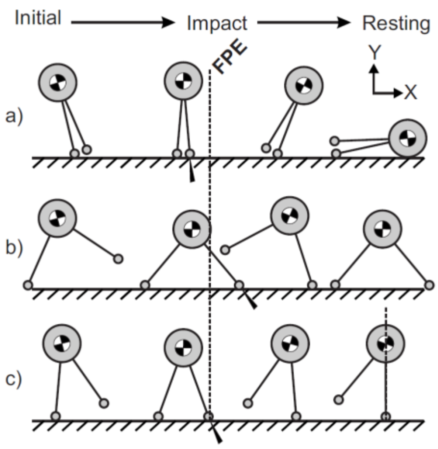



**Figure 2.4** Foot placement estimator (FPE). Stepping relative to the foot placement estimator (FPE). Stepping posterior to the FPE will result in a forward fall (a), stepping anterior to the FPE will result in falling back onto the swing leg (b) and stepping on the FPE will balance the centre of mass above the centre of pressure (c). Adapted from Millard et al. (2009).

While the FPE may be a promising stability measure, the lack of experimental studies using the FPE is a drawback. Additionally, it does not take into account the execution time of a step but perhaps more importantly the FPE evaluates recovery within a single steps whereas recovery from a perturbation is often achieved over multiple consecutive steps. However, the distance between actual foot placement and the estimated foot placement over time may give an indication of the number of steps required to restore balance. Finally, its calculation requires a full body marker set which limits clinical feasibility. Altogether, we conclude that the FPE is not yet a suitable measure for reactive gait assessment.

### 2.3.3 Gait sensitivity norm

The gait sensitivity norm (GSN), introduced in the field of robotics, quantifies how much and how long the gait pattern is affected following a perturbation (Hobbelen and Wisse, 2007). Steps following the perturbation onset are compared to the average steady state step, and normalized to the perturbation intensity. While the GSN has been considered a gait stability measure (Bruijn et al., 2013a), it thus differs in the sense that it captures deviation from steady state gait rather than the relation between the XCoM and the BoS. Any parameter (called gait indicator) can be evaluated and thus previously described gait stability measures can be included. In addition, multiple gait indicators can be evaluated, but a suitable rationale is required for successful use of the GSN (Hobbelen and Wisse, 2007).

The GSN is defined as follows:

$$GSN = \frac{1}{|e_0|}\sqrt{\sum_{i=1}^{q}\sum_{k=0}^{\infty}(g_k(i) - g^*(i))^2}$$ **Equation 2.2**

in which $g_k(i)$ is the value of the $i$th gait indicator at step k following the perturbation, $q$ is the number of gait indicators, $g^*(i)$ the average steady state value of the gait indicator and $e_0$ the perturbation size. Higher values of the GSN indicate larger deviation from steady state gait and thus lower gait stability.

The GSN definition raises a few problems. First, differences in gait indicators units may bias the GSN value. For example, a 100% increase in step width will have a greater effect on the GSN compared to a 100% increase in step length. Although not described in the original paper, including multiple gait parameters thus requires normalization of the data. Second, given Equation 2.2, unperturbed gait (that is $e_0$ equals zero) should lead to a GSN value of zero. However, due to variability in human gait, some deviation from average steady state gait values will always be present. This limits between-individual comparison, as the perturbation response may be confounded by steady state gait variability. Again, normalizing the gait indicator to steady state variability may solve this problem (Bruijn et al., 2013a). Finally, there is no concept on how to determine the perturbation size ($e_0$).

While the GSN was designed for evaluation of robotic gait, it has the potential to be used in human gait. Yet, to our knowledge, there are only two studies that experimentally explored its feasibility (Bruijn et al., 2013a; van den Noort et al., 2017) and one modelling study that evaluated the correlation between the GSN and risk of falling in a human walking model (Thangal et al., 2013). In a pilot study, Bruijn and colleagues used the CoM position as gait indicator to evaluate gait speed when recovering from sideways waist-pull perturbations (Bruijn et al., 2008). Lower GSN values (i.e. higher gait stability) were found during slow walking, which indicates that the GSN is sensitive to the effect of walking speed on gait stability. Recent work by van den Noort and collagues (2017) showed that knee flexion and abduction were sensitive to the intensity of sideways platform perturbations, but only if the perturbation intensity was quantified by means of displacement and not the sum of squared medio-lateral speed (representing the kinetic energy action on the body). Careful selection of gait indicators and measures to quantify the perturbation intensity are thus required.

Although these experimental studies show that the GSN can be used in human gait, a number of aspects remain to be investigated. Primarily, the lack of knowledge on suitable gait indicators and the effect of normalizing the perturbation response to steady state gait variability limits its direct application in clinical practice. However, as of yet, it is the only measure that specifically relates perturbed gait to steady state gait and hence eliminates between-individual differences in steady state gait.

### 2.3.4 Stabilizing and destabilizing forces

Duclos and colleagues (2009) introduced the concept of stabilizing and destabilizing forces. The stabilizing force is defined as “the theoretical force the subject needs to stop himself by preventing the CoP from moving outside the BoS at each instant of the task”. It quantifies how much force is required to maintain stability while taking into account the velocity of the CoM in the direction of the BoS as demonstrated in the following equation:.

$$F_{st} = -\frac{1}{2} * \frac{m * {CoM_{vel}}^2}{d}$$ **Equation 2.3**

in which $m$ is the individual's body mass [kg] and $CoM_{vel}$ is the velocity of the CoM [m/s] and $d$ the minimum distance between the CoP and the border of the BoS. Higher stabilizing force values indicate that more work is required to stop the CoP and the CoM and thus a lower gait stability.

Contrary, the destabilizing force is defined as: "the theoretical force necessary to move the CoP to the limit of the BoS" and can be calculated as follows:

$$F_{dst} = (\frac{F * n}{CoM_{VT}})D_{CP}$$ **Equation 2.4**

in which $F$ is the ground reaction force, $n$ the unitary vector normal to the contact surface, $CoM_{VT}$ the height of the CoM and $D_{CP}$ the horizontal distance from the CoP to the border of the BoS. Somewhat surprisingly, the destabilizing force does not include CoM movement and may therefore be considered too simplistic (Bruijn et al., 2013a). Finally, gait stability is defined by the ratio between the destabilizing and stabilizing force but this parameter has little meaning given that it includes the destabilizing force.

Only a handful of publications evaluated gait by means of stabilizing and destabilizing forces. Duclos and colleagues (2009) demonstrated that stabilizing forces were 2-3 times higher at maximum gait speed compared to comfortable walking speed, while destabilizing forces were not affected by speed. Later work of the same group, revealed that stabilizing forces increased and destabilizing forces decreased with increasing belt acceleration and deceleration perturbations in young adults walking on a treadmill (Ilmane et al., 2015). Additionally, Dubreucq and colleagues (2017) demonstrated that awareness of an upcoming perturbation did not change gait stability in older adults, while stabilizing forces were increased and destabilizing force decreased in young adults. In conclusion, while conceptually the stabilizing force shows potential, the destabilizing force lacks validity without taking into account the CoM motion state.

### 2.3.5 Discussion

We discussed four gait stability measures that quantify the relation between CoM motion and the BoS: feasible stability region, margins of stability, the foot placement estimator, gait sensitivity norm and the stabilizing and destabilizing forces. We concur with Bruijn and colleagues (2013), that all measures apart from the destabilizing forces are biomechanically sound and hence have good construct validity. Technically, the stabilizing force can be used without the destabilizing force and considered a single measure however, experimental evidence for this measure is lacking. Little experimental evidence exists for the foot placement estimator despite its solid construct validity. More importantly, a major drawback of this measure is the need for full body motion capture to calculate total body angular momentum which limits clinical feasibility. Finally, both the feasible stability region and margins of stability are biomechanically sound and have been widely used to discriminate between various (patient) populations and evaluate responses to perturbations. Margins of stability appears favourable because of its simplicity.

The gait sensitivity norm may be a feasible method to explore deviation in the gait pattern as a result of the perturbation. However, knowledge on proper selection of gait indicators is lacking and therefore further work on the feasibility of this concept is required.

To conclude, margins of stability may be considered a simplified version of the feasible stability region. While this may reduce sensitivity it appears a suitable candidate outcome measures to quantify stability in response to gait perturbations.

**CONCLUSION** Margins of stability is a biomechanically sound and clinically feasible gait stability measure that captures the relation between the centre of mass motion state and the base of support.

## 2.4 Gait perturbations

This section provides an overview of the use of perturbations to quantify reactive gait stability. Our objective was to determine which perturbation types should be included in a protocol to assess reactive gait in older adults. Gait stability can either be affected directly by means of mechanical perturbations such as trips or slips, or indirectly by manipulating sensory input using for example visual perturbations. Furthermore, we can discriminate between discrete and continuous perturbations. Discrete perturbations refer to an instant disturbance of the gait pattern usually applied within one step (referred to as perturbation step) followed by one or more recovery steps. In contrast, continuous perturbations are applied over longer periods including multiple subsequent steps. Therefore, discrete perturbations requires fast responses whereas continuous perturbations can be resisted by adapting spatio-temporal parameters (Hak et al., 2013b, 2013c, 2013d). Moreover, discrete perturbations may be more ecologically valid. We aimed to develop a to assess reactive gait stability, hence limited our review to discrete gait perturbations. Moreover, we limited the review to healthy young adults (to understand the biomechanical effect of the perturbation on the gait pattern), older adults (to understand the effect of ageing on the perturbation response) and fallers and non-fallers (to understand which factors may be indicative of risk for falls).

Within discrete gait perturbation studies, the vast majority of work includes horizontal plane perturbations. Therefore, we first discriminated between perturbations in the sagittal plane, inducing forward (FW) and backward (BW) balance loss, and in the frontal plane, inducing medio-lateral (ML) balance loss. Next, we compared horizontal plane perturbations. Finally, we discussed other perturbation types either directly or indirectly affecting gait stability. Each section provides an overview of the most commonly used experimental setups as well as the biomechanical effects of the perturbation on the gait pattern and the effect of ageing on the ability to recover from the perturbation.

### 2.4.1 Sagittal plane perturbations

Sagittal plane perturbations challenge forward or backward stability. As described previously, forward stability is compromised when the XCoM exceeds the anterior border of the BoS, and likewise backward instability is the result of the XCoM exceeding the posterior border of the BoS. Balance in the sagittal plane can thus be challenged by either imposing an AP perturbation to the BoS or the CoM. The former has been done by perturbing the stance leg or the swing leg. Although, technically the swing leg is not part of the BoS, interrupting the swing leg will affect the to be established BoS. Furthermore, waist pulls and pushes have been used to affect CoM movement. First, we will provide an overview of the experimental setups used to impose AP swing leg, stance leg and waist perturbations. Next, we will discuss how these setups were used to induce forward and backward balance loss, how individuals recover and the effect of ageing on the perturbation responses.

#### 2.4.1.1 Experimental setups to impose a swing leg perturbation

Swing leg perturbations interrupt the forward progression of the swing leg and are referred to as trips or stumbles. Early work on responses to trip perturbations included physical objects on an overground walkway, either at a fixed location while participants' vision was partly restricted (Grabiner et al., 1993; Pijnappels et al., 2001), arising from the floor when manually triggered by the investigator (Pavol et al., 1999b) or arising based on force plate data input (Eng et al., 1994). Later, in the experimental setup of Pijnappels and colleagues (2004), timing and position of the arising obstacle was determined based on kinematic data, allowing for standardized perturbation onsets at a certain percentage of the swing phase (Figure 2.5). In addition to overground trips, trips during treadmill walking have been induced by putting physical objects on the treadmill belt (Schillings et al., 2005, 2000, 1996) or mimicked by a backward pull of the swing leg via a rope around the ankle (Forner Cordero et al., 2003; Karamanidis et al., 2011; McCrum et al., 2016; Smeesters et al., 2001).

#### 2.4.1.2 Experimental setups to impose a stance leg perturbation

A perturbation to the stance leg is referred to as a slip and can be applied in both the forward and backward direction. A slip occurs when the horizontal forces during a step exceed the friction with the support surface. This is usually around initial contact or push-off, when horizontal forces are highest and results in a sudden increase in velocity of the stance foot (Redfern et al., 2001; Woollacott and Tang, 1997). Various materials and equipment have been used to lower the friction with the support surface, for example: paper sheets with different friction coefficients (Heiden et al., 2006), soap (Haynes and Lockhart, 2012; Liu and Lockhart, 2009) and oil (Allin et al., 2016) or rollers mounted on a force plate (Marigold and Patla, 2002).

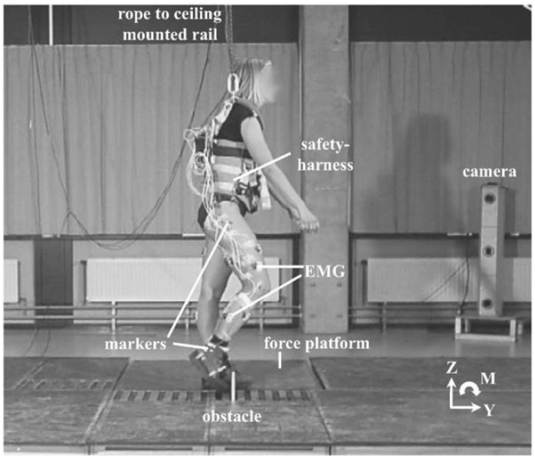



**Figure 2.5** Experimental setup to induce trip perturbations during overground walking Timing of the obstacle appearance was controlled based on real-time kinematic data. Adapted from Pijnappels et al. (2005a).

Prominent in the literature is the work of the group of Pai (Bhatt et al., 2005; Bhatt and Pai, 2009; Liu et al., 2015; Yang et al., 2011; Yang and Pai, 2014). Their experimental setup consists of sliding platforms in an overground walkway (Figure 2.6). Such perturbations lower friction of the support surface rather than the friction between the BoS and the support surface. More recently, Pai and colleagues (Ding and Yang, 2016; Lee et al., 2016) as well as others (Aprigliano et al., 2015; Dubreucq et al., 2017; Ilmane et al., 2015; Kagawa et al., 2011;

Martelli et al., 2017) used sudden changes in belt speed to impose slips during treadmill walking. Alternatively, external forces moving the BoS by means of platform movement have been used (McIntosh et al., 2017; Oliveira et al., 2012b). Both active platform movement as well as treadmill belt perturbations allow for standardization of the perturbation intensity by controlling BoS motion.

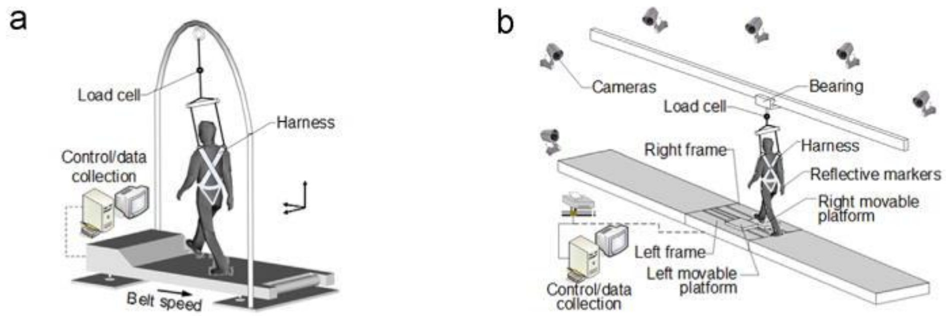



**Figure 2.6** Experimental setups to induce slip perturbations during treadmill (a) and overground (b) walking. Adapted from Lee et al. (2016).

#### 2.4.1.3 Experimental setups to impose a waist perturbations

A number of studies have included waist-pull or -pushes on a treadmill (Bruijn et al., 2010; Martelli et al., 2016; Misiaszek, 2003; Misiaszek and Krauss, 2005; Vlutters et al., 2017, 2016). Such setups typically consist of pneumatic or electrical cylinders, connected via cables or piston rods to a stiff belt around the participant’s waist. Recently, a balance assessment robot was developed to study responses to overground waist perturbations (Olenšek et al., 2016).

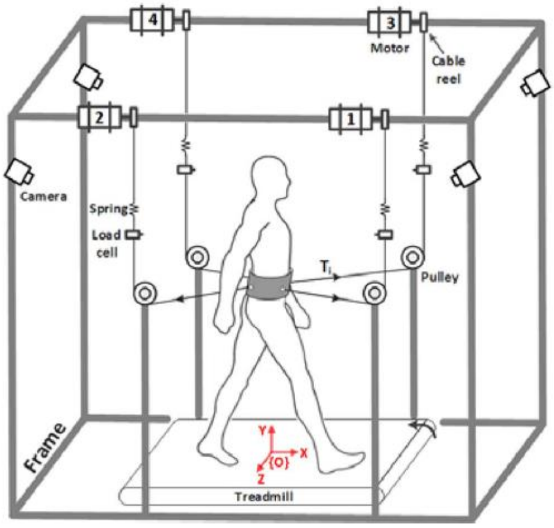
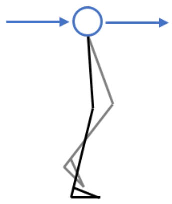



**Figure 2.7** Experimental setup to induce waist-pull perturbations in the anterior-posterior and medio-lateral direction Adapted from Martelli et al. (2016)**.**

#### 2.4.1.4 Recovery from forward balance loss

Forward balance loss can be induced by perturbing the swing leg, stance leg or waist (Table 2.1). Swing leg perturbations, often referred to as trips, increased forward rotation of the body and hence the ability to successfully recover from trips lies in the capacity to reduce this rotation (Grabiner et al., 1993). This is achieved by reducing trunk flexion velocity (Forner Cordero et al., 2003; Pavol et al., 2001) and proper foot placement and push-off (Pijnappels et al., 2005b, 2005c). Two recovery strategies have been distinguished: the elevating and the lowering strategy (Eng et al., 1994). During an elevating strategy, the tripped foot is lifted over the obstacle by increasing flexion in the hip, knee and ankle. This strategy is most often seen in early swing. A lowering strategy is characterized by placement of the tripped foot while the contralateral foot is used to step over the obstacle. This strategy is usually applied when tripped in the late swing phase but has also been found during mid-swing trips (Pijnappels et al., 2005c). Similar responses have been reported for recovery from obstacle-induced trips on a treadmill (Schillings et al., 2005, 2000) and despite the absence of a physical object also when recovering from backward pulls of the swing leg (McCrum et al., 2016).

**Table 2.1** Methods to induce forward balance loss during overground (OG) and treadmill (TM) walking.

| | A | B | C | D |
|---|---|---|---|---|
| |  |  |  |  |
| **Condition** | Obstacle (OG/TM)<br>Ankle pull (TM) | Platform translation (OG) | Belt acceleration (TM) | Waist pull or push (OG/TM) |
| **Type** | Trip | Backward slip | Backward slip | Forward pull or push |

Pavol and colleagues (2001) studied responses to overground trips during the mid-to-late swing phase in 46 older adults. They demonstrated that older adults who were not able to successfully recover, showed increased forward body rotation compared to those who did recover. Additionally, the fallers also showed less proper foot placement (i.e. shorter step length compared to non-fallers). However, the authors argued that improper foot placement following late swing phase trips may be a consequence, rather than the cause, of increased forward body rotation. Steps should have been much faster and longer compared to normal to avoid falling.

The association between the inability to control forward angular momentum and falls in older adults was confirmed in a later study investigating early-swing overground trip perturbation in 12 young and 11 older adults (Pijnappels et al., 2005b). Despite comparable perturbation effects across groups, it appeared that all young participants were able to reduce forward angular momentum, either during the push-off phase or the recovery step. While none of the older adults achieved this, some (4/11) managed to prevent further increase in the forward angular momentum and hence successfully recovered. Those who could not control forward angular momentum during push off (7/11) did not successfully recover at first. However, in later trials, this was compensated for by better foot placement of the recovery step leading to successful recovery. The contribution of proper foot placement thus seems to depend on the timing of the perturbation; when the trip is imposed early in the gait cycle, effective reduction of forward angular momentum allows for time to lengthen the swing phase and properly place the recovery foot which facilitates further reduction of forward angular momentum.

While the dynamics on a treadmill are different because the BoS is moving backward, similar principles for balance loss apply: when the XCoM is posterior to the BoS, backward balance is challenged and when the XCoM is anterior to the BoS forward balance is challenged. Indeed, belt accelerations applied during stance phase (i.e. a backward slip) can be used to move the CoM toward the anterior border of the BoS and increase stabilizing forces, i.e. higher forces are required to stop the CoP at the border of the BoS (Ilmane et al., 2015). Other work including treadmill perturbations evaluated the effect of sudden backward pulls of the swing leg on margins of stability in young-aged (22-30 years), middle-aged (41-59 years) and old-aged (62-75 years) women (Karamanidis et al., 2011; Süptitz et al., 2013). It was shown that the trip initially induced forward instability (negative MoS) in all age groups, but middle-aged and old-aged needed three more recovery steps compared to young-aged adults due to the inability to increase their BoS (step length). Additionally, some older adults needed to hold on to the handrails to prevent falling, which did not seem to be the case in the middle-aged group.

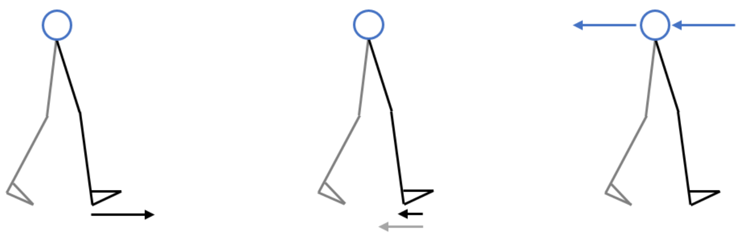
#### 2.4.1.5 Recovery from backward balance loss

Backward balance loss can be induced by a perturbation to the stance leg, mimicking a forward slip (Table 2.2). A series of publications from Lockhart and colleagues, induced forward slips using motor oil on a large circular track. They demonstrated that older adults fell more often compared to young adults. Older adults showed higher horizontal heel contact velocity when stepping in the oil and were less able to increase friction due to slower transition of the CoM (Lockhart, 2008; Lockhart et al., 2005a, 2003). Some older adults did recover but generated less efficient joint moments to regain stability compared to younger adults (Liu and Lockhart, 2009).

A large study, including 119 older adults, demonstrated that gait stability (measured by the shortest distance to the backward boundary of the FSR) in individuals who fell following an overground slip perturbation was lower compared to those who recovered (Bhatt et al., 2011a). The authors argued that lower gait stability may not only increase the likelihood of falling but also reduce the chances of successful recovery.

On a split-belt treadmill, Martelli and colleagues (2017) investigated the effect of forward slips by reversing belt speed on one side, on spatio-temporal parameters and backward MoS in young and older adults. While none of the participants fell, the data demonstrated that older adults had more difficulty in reversing the forward movement of the swing leg to take a compensatory backward step and regain balance compared to young adults. As such, their MoS was more affected compared to young adults.

**Table 2.2** Methods to induce backward balance loss during overground (OG) and treadmill (TM) walking.



| | **A** | **B** | **C** |
|---|---|---|---|
| **Condition** | Slippery surface (OG)<br>Platform translation (OG) | Belt deceleration (TM) | Pull or push (OG/TM) |
| **Type** | Forward slip | Forward slip | Backward pull or push |

Less obvious are responses to deceleration perturbations when the treadmill belt slows down but speed is *not* reversed. If the XCoM is around or over the BoS at toe-off, the swing phase can be continued to lengthen the step. When the XCoM remains posterior to the BoS at toe-off, backward balance loss is induced (Ilmane et al., 2015; Kagawa et al., 2011). This occurs with larger reductions in gait speed and results in a quick foot down of the swing leg posterior to the stance foot in order to regain backward balance. However, the early double support phase hampers forward progression and results in a more backward position on the treadmill. As such, quick recovery steps are required to keep up with the running belt. This has been associated with longer recovery times compared to forward balance loss induced by unilateral belt acceleration perturbations (Ilmane et al., 2015).

#### 2.4.1.6 Discussion

Forward balance loss can be evoked by interrupting the swing leg or backward movement of the stance leg, often referred to as trip, or backward slip, respectively. Alternatively, a forward push or pull of the waist may be used. These perturbation types typically results in reduced margins of stability in the forward direction, increased forward angular momentum and trunk flexion. Older adults who were not able to successfully recover from the experimental perturbations showed reduced ability to arrest forward angular moment and less suitable foot placement during recovery. These studies also showed that the initial effect of the perturbation did not differ between fallers and non-fallers, suggesting that the ability to prevent falling depends on the capacity to recovery from rather than to resist the perturbation. However, this may also be related to these specific perturbations.

Backward balance loss has been initiated by forward movement of the stance leg either during overground walking or by reversing treadmill belt speed or by backward waist pushes or pulls. Additionally, backward balance loss can be induced by treadmill belt decelerations, but only if the perturbation intensity and timing is such that the XCoM remains posterior to the posterior border of the BoS. Older adults who did not successful recovery from backward balance loss were initially more affected as indicated by higher horizontal heel velocity and lower gait stability. During recovery they showed reduced ability in reversing forward movement of the swing leg and translating the CoM to increase friction with the support surface.

**CONCLUSION** Ageing affects the ability to recover from forward and backward balance loss during gait.

### 2.4.2 Frontal plane perturbations

Investigating responses to ML gait perturbations has received less attention. A possible explanation could be that most falls occur during tripping or slipping. However, the importance of lateral balance control to prevent falling has been well established (Bauby and Kuo, 2000; Maki, 1997; Mille et al., 2013; O'Connor and Kuo, 2009; Rogers and Mille, 2003). An alternative explanation could be that ML perturbations requires active perturbations such as sideways platform translations, while AP perturbations can also be induced passively by means of sliding platforms or slippery surfaces.

#### 2.4.2.1 Experimental setups to induce medio-lateral balance loss

A few studies have included sideway translations of large movable platforms (McIntosh et al., 2017, Figure 2.8) or embedded in an overground walkway (Oliveira et al., 2012a, 2012b) but most work applied waist perturbations using setups as described above (section 2.4.1.3). While most AP gait perturbation studies evaluate recovery responses in either the forward or backward direction, research on ML gait perturbations include opposing perturbations (i.e. to the left and right side).

#### 2.4.2.2 Recovery from medio-lateral balance loss

Medio-lateral balance loss can be induced by perturbing the stance leg or waist (Table 2.3). A variety of ways have been used to describe the direction of frontal plane perturbations, like left/right, medial/lateral or inward/outward. However, a comparable perturbation in terms of direction may yield opposite recovery responses depending on timing in the gait cycle (Hof et al., 2010; Vlutters et al., 2016). Therefore, to capture the relation between direction and timing, we propose to distinguish between ipsilateral and contralateral perturbations. Ipsilateral perturbations refer to a perturbation in the direction of the (leading) stance leg. For example, an ipsilateral waist pull perturbation means that the waist is pulled to the left (right) during the left (right) stance phase (Table 2.3 – D). In contrast, a contralateral waist pull perturbation refers to a pull toward the right (left) during the left (right) stance phase (Table 2.3 – B). Note that, as in the AP direction, loss of balance can be induced by applying stance leg or waist perturbations in opposite directions (see Table 2.3 perturbation A vs B and C vs D).

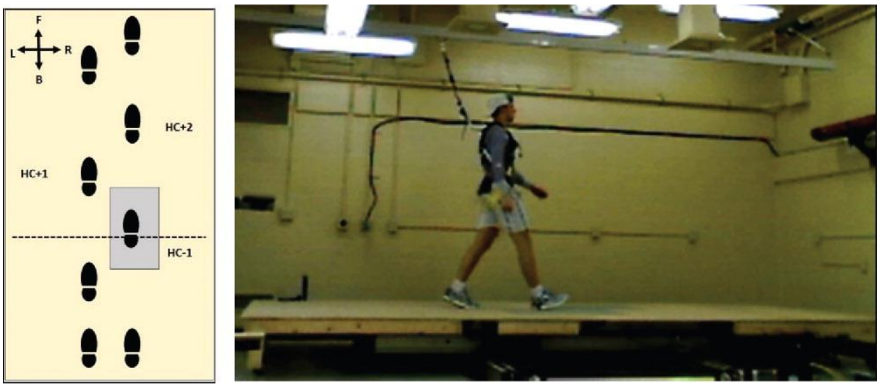
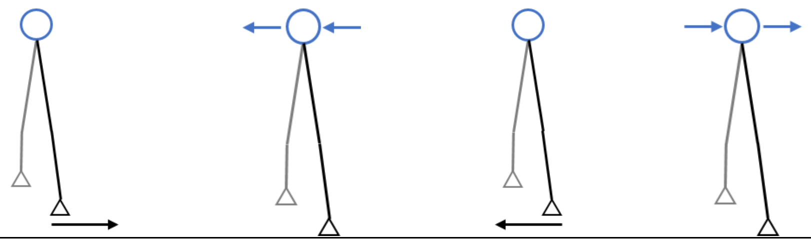



**Figure 2.8** Experimental setup to induce stance leg perturbations in the medio-lateral and anterior-posterior direction. Perturbations were triggered at the right heel strike (HC-1). Therefore, perturbations to the left (right) could be considered contralateral (ipsilateral) sway perturbations. Backward (forward) platform movement induced forward (backward) balance loss. Two steps following the perturbation (HC+1 and HC+2) were evaluated. Adapted from McIntosh et al. (2017).

The majority of ML gait perturbations have been applied to the waist. Responses to ipsilateral waist perturbations (i.e. moving the CoM toward the lateral border of the BoS) are considered more challenging compared to contralateral waist perturbations (i.e. moving the CoM away from the lateral border of the BoS) and result in increased trunk lean, extension and external rotation (Oliveira et al., 2012b). When the XCoM is moved over the lateral border of the BoS, successful recovery is often achieved by taking cross-over steps (Hof et al., 2010; Martelli et al., 2016; Olenšek et al., 2016; Vlutters et al., 2016). While this has been considered a dangerous strategy it also allows for quick foot placement to bring the XCoM back into the newly established BoS (Vlutters et al., 2016). Although duration of the cross-over step is not always faster and may depend on the timing of the perturbation as well as gait speed (Hof et al., 2010). Moving the waist away from the lateral border of the BoS enforces early termination of the swing phase and an increase in step width (Hof et al., 2010; Vlutters et al., 2016).

**Table 2.3** Methods to induce forward balance loss during overground (OG) and treadmill (TM) walking.



| | **A** | **B** | **C** | **D** |
|---|---|---|---|---|
| **Condition** | Platform translation (OG/TM) | Waist pull or push (OG/TM) | Platform translation (OG/TM) | Waist pull or push (OG/TM) |
| **Type** | Ipsilateral perturbation | Contralateral perturbation | Contralateral perturbation | Ipsilateral perturbation |
| **Balance loss** | Medial balance loss | Medial balance loss | Lateral balance loss | Lateral balance loss |

Logically, ML BoS perturbations would yield similar yet opposite responses compared to waist perturbations. Indeed, a study evaluating perturbation responses in young and older adults demonstrated that recovery from an ipsilateral BoS perturbation (i.e. moving the BoS away from the COM) was achieved by taking a wide step while a contralateral BoS perturbation (moving the BoS toward the CoM) resulted in a narrow step or a cross-over step (McIntosh et al., 2017). Only minor differences in recovery responses between young and older differences were found. Older adults showed larger ML CoM displacement following the ipsilateral sway perturbation, but these differences appeared to be confounded by differences in steady state gait. Furthermore, a significant step x age interaction effect, but no direction effect, for step length was found. Meaning that young and older adults recovered differently in terms of step length, but these differences were not affected by the direction of the perturbation. Taking ipsilateral, contralateral, anterior and posterior perturbations together on average step length reduced in both young and older adults, but older adults needed more steps compared to young adults to return to baseline values. This may suggest that recovery capacity is reduced in older adults, but a clear interpretation of these findings or guidance on which perturbation types should be included in a reactive gait assessment to identify risk for falls in older adults was not given.

#### 2.4.2.3 Discussion

Medio-lateral balance loss can be evoked by moving the BoS or the CoM in the frontal plane. When the XCoM exceeds the lateral border of the BoS, a cross-over step is used to regain balance. Such perturbations may be considered more challenging compared to situations where the XCoM is moved toward but not over the BoS or in the opposite direction, when the XCoM moves away from the BoS. The effect of ageing on responses to ML gait perturbations has been scarcely investigated, and given the importance of ML balance control further investigation in age-related decline and the association with falls in older adults is needed.

CONCLUSION Despite the importance of lateral balance control to prevent falling, the effect of ageing on the ability to recover from medio-lateral balance loss during gait has been scarcely investigated**.**

### 2.4.3 Horizontal plane perturbations

A few experiments have included both AP and ML perturbations. Comparison between these perturbations types may enhance our understanding of perturbation responses and contribute to developing a perturbation protocol for reactive gait assessment. As described in the previous chapters, Vlutters and colleagues (2016) examined responses to AP and ML waist perturbations. They showed that single support time was decreased when the CoM was pushed away from or over the BoS. Moreover, using different perturbation intensity the authors demonstrated a relation between ML CoM velocity and ML foot placement. This indicates that targeted ML foot placement is used to regain stability. In contrast, single support time did not decrease following AP perturbations nor was there a relationship between CoM velocity and AP foot placement. Hence the authors argued that different strategies such as ankle torque modulation play a role when recovering from AP perturbations. This ankle strategy may reduce the initial effect of the perturbation and hence there is less need to alter the gait pattern following the perturbation. In line with these findings, McIntosh and colleagues (2017) demonstrated that CoM velocity was more affected following backward platform translation compared to forward platform translation, and following ipsilateral platform translations compared to contralateral perturbations. Moreover, contralateral perturbations increased ML CoM velocity by up to five times compared to steady state velocity whereas backward perturbations only resulted in an increase by a factor of 1.5. Contralateral perturbations thus seem more challenging compared to backward perturbations. However, the key question is whether we can identify fallers based on recovery responses to these perturbation types. Since no differences between young and older adults for these perturbation types were found, further work is needed to determine whether ipsilateral, contralateral, backward, forward perturbations or a combination should be included in a reactive gait assessment aiming to identify risk for falls in older adults.

**CONCLUSION** It is currently unknown which perturbation type has most potential to identify older adults at risk for falls.

## 2.5 Conclusion and recommendations

This literature review provides an overview of the current state-of-the-art for gait stability measures and various perturbation types to quantify reactive gait stability in older adults. While early work primarily quantified responses to perturbations in terms muscle activity, spatio-temporal parameters and joint kinematics, evaluating the effect of perturbations by means of gait stability measures is gaining more attention. Several publications demonstrated age-related decline in gait stability following perturbations. Moreover, differences in gait stability have been used to discriminate between older adults who were able to recover from an experimentally induced perturbation and those who were not. While this shows the potential that perturbations can be used to quantify reactive gait stability, the relation between reactive gait stability and risk for falls in everyday life is limited. Hence, predictive validity of reactive gait assessment for identification of risk for falls in older adults, either by retrospective or prospective analysis, is needed.

Another important consideration is the selection of a suitable gait stability outcome measure. The margins of stability measure appears to be an eligible candidate, because of its simplicity and ease of interpretation. Furthermore, including conventional spatio-temporal gait parameters will improve understanding as to why gait stability is affected and how it is recovered.

Most of the previous studies include perturbation types in single plane (AP or ML), and most often in only one direction. There are a few publications that included both AP and ML perturbations, and to our knowledge, only one of these studies investigated age-related effects (McIntosh et al., 2017). Risk for falls, however, is often a multiple factorial problem and hence evaluating the ability to recover from various perturbation types may yield a more complete reactive gait assessment.

**RECOMMENDATION** The reactive gait assessment should contain a mixed perturbation protocol including perturbations that induce forward, backward and medio-lateral balance loss. Responses to these perturbations should be evaluated by means of Margin of Stability measures.

# CHAPTER 3

# Aims, objectives and research questions

The overall aim of this thesis was to investigate whether reactive gait assessment can be used to assess risk for falls in older adults. This aim required the development and evaluation of a standardized reactive gait assessment.

Based on the literature review (**Chapter 2**), we hypothesized that responses to challenging gait perturbations can be used to identify older adults with increased risk for falls. The discriminative ability, however, was expected to depend on the perturbation type, as well as the sensitivity of the outcome measure.

As part of this industrial doctorate programme, we aimed to translate knowledge gained in this project and provide the company Motek with 1) a perturbation application programmed in D-Flow to standardize future reactive gait assessment studies and make it more accessible for both internal use as well as researchers using Motek products and 2) recommendations and opportunities for including reactive gait assessment as an evidence-based, affordable and easy-to-use protocol in Motek's product portfolio.

The outline of the remainder of this thesis is as follows:

In **Chapter 4**, an overview of the equipment used to develop and evaluate the reactive gait assessment is provided. Then, three subsequent phases of development and evaluation of the reactive gait assessment are described.

The first development phase, described in **Chapter 5,** consisted of the initial development of the reactive gait assessment and the evaluation on young and older adults. Our objectives were to evaluate: 1) whether different perturbation types affected gait in terms of spatio-temporal parameters and discrete gait stability measures and 2) whether we could discriminate between young and older adults based on these measures.

In **Chapter 6**, the second development phase is described. Based on the findings of the first phase, our objectives were to modify the reactive gait assessment and evaluate its sensitivity to discriminate between older adults with and without a history of falls.

The third development phase, as presented in **Chapter 7**, included the implementation of a continuous trunk motion measure and our objective was to evaluate differences between older adults with and without a history of falls based on this continuous measure of trunk motion. In this phase, we also further explored the validity of the reactive gait assessment toward the use in clinical practice to identify older individuals at risk for falls. We evaluated whether older adults with and without a history of falls adapted their pattern in anticipation of repeated perturbations and whether these proactive adaptations confounded the perturbation effect and the ability to discriminate between groups. Furthermore, a descriptive study was performed to gain insight in what the data means for the individual.

Finally, in **Chapter 8**, the use of reactive gait assessment to identify risk for falls in older adults based on the findings of this thesis is discussed and directions for future research and development toward clinical use of the reactive gait assessment, as well as further potential of implementation of the protocol are described.

# CHAPTER 4

# Equipment

## 4.1 Introduction

To evaluate our main aim, whether reactive gait assessment can be used to identify risk for falls in older adults, we developed a research application containing various submaximal perturbation types. This perturbation protocol was first evaluated in young and older adults at the University of Strathclyde (Chapter 5). In a second experimental study, conducted at Motek, we evaluated the differences between older adults with and without a history of falls (Chapter 6.1). The equipment used in these studies and the development of the perturbation protocol will be discussed in this chapter.

## 4.2 Equipment

The Computer Assisted Rehabilitation ENvironment (CAREN) Extended installed at the University of Strathclyde, used in our first experimental study, is a high-end rehabilitation technology system to assess and train balance and gait. It consists an instrumented dual-belt treadmill, motion platform, motion capture system, a virtual reality environment and a surround sound system (Figure 4.1 - Left). The CAREN Extended is controlled using D-Flow. D-Flow is a modular-based programming tool which integrates hardware sources in real-time by receiving input data from sources like motion capture cameras or force plates and controlling hardware such as treadmill speed or platform movement.

**Figure 4.1** The CAREN Extended system (left) and GRAIL system (right).

The second study was conducted on the Gait Real-time Analysis Interactive Lab (GRAIL) at Motek (Figure 4.1 - Right). The main difference between the two systems is the motion platform which contains six degrees of freedom for the CAREN Extended and two degree of freedom (pitch and sway) for the GRAIL. A detailed overview of the similarities and differences in components for the CAREN Extended and GRAIL components is presented in Table 4.1.

**Table 4.1** Overview of the components for the CAREN and the GRAIL.

| Components | CAREN Extended | GRAIL |
|---|---|---|
| Instrumented dual-belt treadmill (Motek, Amsterdam, The Netherlands) | ✓ | ✓ |
| Platform | Treadmill mounted on top of a degrees-of-freedom motion base (Moog, Nieuw-Vennep, The Netherlands) | Treadmill embedded in a two degrees-of-freedom frame (Motek, Amsterdam, The Netherlands) |
| Infra-red Bonita motion capture cameras (Vicon, Oxford, UK) | 12 | 10 |
| Virtual reality environment projected on a semi-cylindrical screen | ✓ | ✓ |
| Surround sound system | ✓ | ✓ |
| Analog 4-output Phidget | ✓ | ✓ |
| D-Flow control software | ✓ | ✓ |
| Human Body Model software | ✓ | ✓ |

## 4.3 Data integration

D-Flow can both retrieve data from a wide variety of hardware sources and control many different hardware components. Here, we focus on the equipment used in this study. Motion capture and force plate data (D-Flow input) were used to capture the participants' steady state gait pattern and responses to the various perturbations. The perturbation protocol was programmed in D-Flow and thus used to control the treadmill, motion platform, virtual environment and surround sound system.

### 4.3.1 Motion capture and force plate data (D-Flow input)

Three-dimensional marker position data was captured by Vicon Bonita cameras (Vicon, Oxford, UK) running at 100 Hz and fed into Nexus (Vicon software). Marker data was labelled in Nexus. For a detailed description on the marker set and use of this data the reader is referred to section 4.3.2.2. Additionally, 1000 Hz analog force plate data coming from the instrumented treadmill (Motek, Amsterdam, The Netherlands) was retrieved and synchronized in Nexus. From there on, labelled marker data and raw force plate data was streamed to D-Flow for real-time use (Figure 4.2).

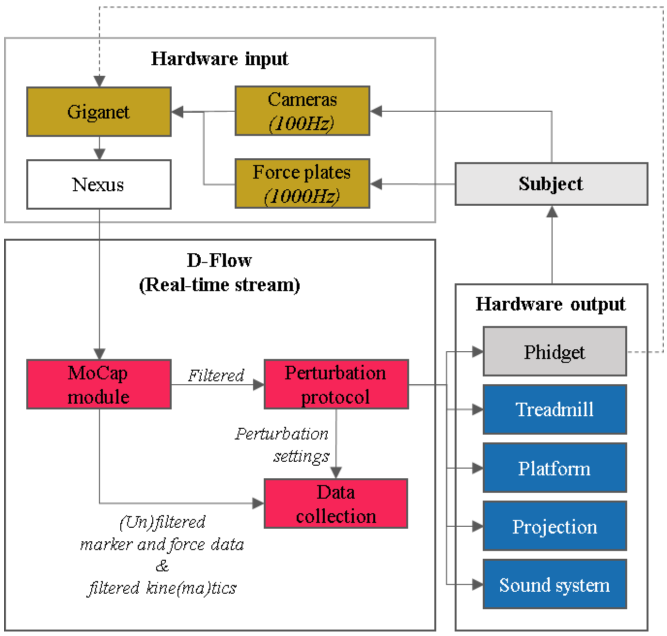



**Figure 4.2** Hardware integration for the CAREN Extended and GRAIL systems.

### 4.3.2 D-Flow

The D-Flow software is the link between the incoming data and the hardware components. The software consists of an editor in which the perturbation application was build (Figure 4.3 – panel A) and a distributed rendering system (DRS) window visualizing the virtual reality environment (Figure 4.3 – panel B). In our protocol, the virtual reality environment was used to provide optical flow synchronized with the treadmill speed. Furthermore, D-Flow contains a user interface called the Runtime Console to control the application (Figure 4.3 – panel C). Finally, the incoming data is processed in the Motion Capture (MoCap) module.

#### 4.3.2.1 MoCap module

As described above, D-Flow is a modular based software package. The Motion Capture (MoCap) module was used for a number of essential processing steps, namely: to receive incoming marker and force plate data, to calculate the CoM position, to detect gait events for timing of the perturbation onset and to record data.

First, the incoming force plate data is down sampled to 100 Hz using a 2$^{nd}$ order low-pass Butterworth filter with a cut-off frequency of 15 Hz, to match the marker data sample frequency. Following this, 100 Hz marker and force plate data are filtered at 6 Hz to calculate the CoM position using Motek's Human Body Model (HBM). HBM is a musculo-skeletal model which can be used in real-time. For a detailed description on HBM the reader is referred to section 4.3.2.2. Additionally, gait events were detected using an AP treadmill-based velocity algorithm (Zeni et al., 2008). Finally, both raw and filtered marker and force plate data were recorded.

#### 4.3.2.2 Human Body Model

HBM is musculo-skeletal model designed by Van den Bogert and colleagues (2013) for real-time analysis of joint kinematics, kinetics, CoM and estimation and visualization muscle forces. The model is based on a global optimization technique, meaning that the entire skeleton is modelled at once instead of each segment individually (Lu and O'Connor, 1999). Therefore, it is fast and less sensitive to marker dropout and skin artefacts. The model was further developed and has been implemented in D-Flow by Motek. The immediate feedback allows for a variety of clinical applications for example reducing the knee adduction moment in osteoarthritis patients (Richards et al., 2017; van den Noort et al., 2014) or to improve hip and knee extension in children with Cerebral Palsy (Booth et al., 2016; van Gelder et al., 2017), but also for offline data analysis in perturbation studies (de Melker Worms et al., 2017; Punt et al., 2017; Sloot et al., 2015).

HBM output can be based on a full body or a lower body marker set. The full body marker set contains 47 markers, whereas the lower body marker contains 29 markers on the trunk and lower limbs only. The latter was developed to reduce preparation and processing time in gait analysis, which is typically limited to examining lower body movement only. In our first study, we used the full body HBM marker set. However, we did not use HBM for real-time purposes but solely to calculate CoM position data. In the second study, we applied the lower body marker set.

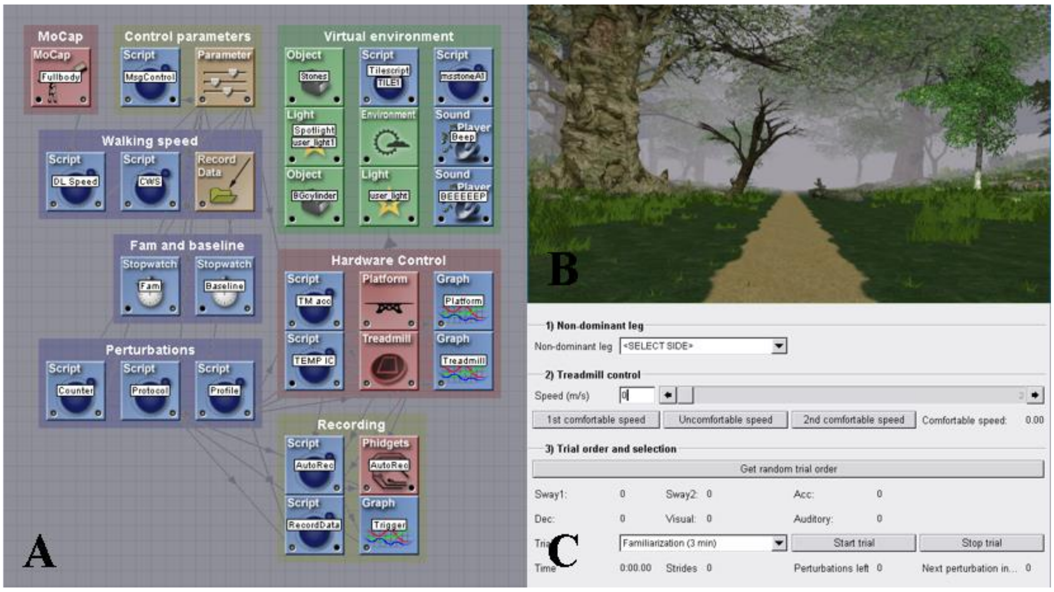


**Figure 4.3** D-Flow editor (A), the virtual environment (B) and Runtime Console (C).

### 4.3.3 Hardware control (D-Flow output)

D-Flow was set up to control the treadmill speed, platform motion, virtual reality environment and the audio system. The virtual reality environment provided optical flow synchronized to treadmill speed.

### 4.3.4 Data collection and processing

The following data were collected:

1. Marker and force plate data
2. CoM position data
3. Treadmill speed
4. Platform movement
5. Perturbation settings (type, intensity and timing)

Marker and force plate data were collected in both Nexus and D-Flow. While all data is available in D-Flow, raw data in Nexus was recorded to allow for post-processing of marker data. The quality of real-time labelled marker data is acceptable, but can be improved during post-processing gap filling of missing markers and labelling unlabelled trajectories.

CoM positions were recorded in D-Flow but reprocessed from cleaned Nexus marker data if required.

Treadmill speed was recorded in D-Flow using the encoder counters of the treadmill. Using a Analog 4-output Phidget (Phidgets Inc, Calgary, Canada), the treadmill speed was sent to Nexus to be recorded with the raw marker and force plate data. This way, all data were synchronized when post-processing was required.

Platform movement was captured by putting three markers on the platform and therefore recorded in both Nexus and D-Flow.

Perturbation settings were recorded in D-Flow. Additionally, the Analog 4-output Phidget was used to send a pulse to Nexus so the perturbation onset could be recorded with the other data.

After post-processing in Nexus, data were restreamed to D-Flow to calculate CoM position using HBM.

# CHAPTER 5

# Reactive gait assessment development

# Phase 1

## 5.1 Development

### 5.1.1 Perturbation types

Based on the literature review we aimed to induce forward, backward, medial and lateral balance loss. Moreover, two exploratory perturbation types were included aiming to evaluate the effect of low light conditions and loud noises on the gait pattern. If these sensory perturbation types could affect the gait pattern they may provide a simple way to assess reactive gait. This resulted in the following six perturbations (Figure 5.1):

1. **Ipsilateral sway (medial balance loss)** platform translation consisting of an outward translation of the BoS. In other words, the platform movement to the left (right) at left (right) initial contact.
2. **Contralateral sway (lateral balance loss)** platform translation consisting of an inward translation of the BoS. In other words, the platform movement to the left (right) at right (left) initial contact.
3. **Acceleration (forward balance loss)** of the unilateral belt, meaning that at left (right) initial contact the left (right) belt speed increases.
4. **Deceleration (backward balance loss)** of the unilateral belt, meaning that at left (right) initial contact the left (right) belt speed decreases. Note that speed did not reverse.
5. **Visual** by rapidly darkening the room.
6. **Auditory** in the form of a loud air horn.

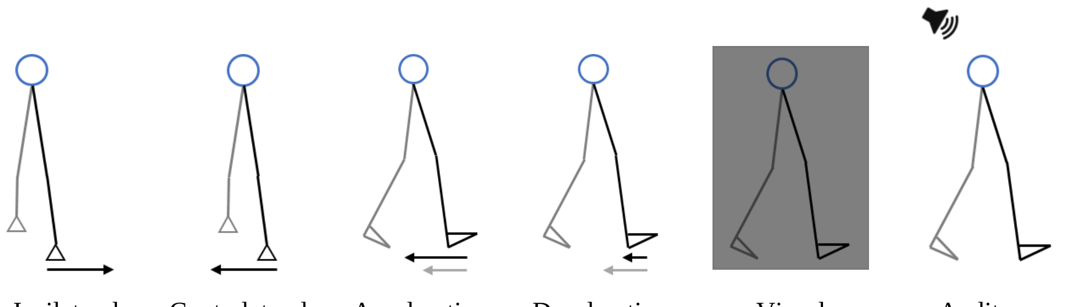



**Figure 5.1** Perturbation types included in the initial perturbation protocol.

### 5.1.2 Perturbation profiles and intensities

The perturbation profiles were based on a previously developed perturbation application by Motek (MM Gait 2.1 - Perturbations) and contained three phases: 1) an incremental phase, 2) a delay phase and 3) a decremental phase. During the incremental phase, the perturbation value ($v$) increased exponentially until a pre-set amplitude was reached. Next, during the delay phase, $v$ remained constant for a given time period. During the decremental phase, $v$ exponentially returned to its initial value (Figure 5.2). All perturbations followed the same perturbation profile regardless of type, with exception of the sway perturbations which solely consisted of the incremental phase (Appendix VI).

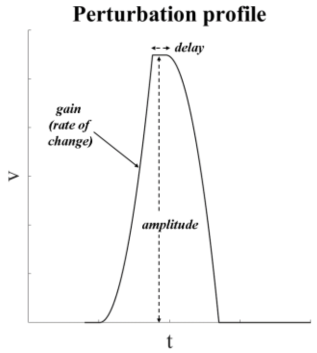



**Figure 5.2** The perturbation profile.

Hence, the incremental phase was characterized by a change in position for the sway perturbations, an increase and decrease in speed of one belt for the acceleration and deceleration perturbations, respectively, a decrease and recovery of illumination for the visual perturbation and a loud sound for the auditory perturbation.

Pilot testing with a physical therapist was performed to explore which perturbation intensities would be appropriate. Previous studies included both maximal (i.e. inducing a fall) and submaximal perturbation intensities (section 0). Although we can impose maximal perturbations in research in a safe manner using body weight support systems or full body safety harnesses to minimize the impact of a fall, one can imagine that this is not feasible in clinical practice given the costs, size and complexity of such systems. In addition, even when a fall may not induce physical injury, it can put unnecessary psychological stress on the individual and therefore reduce the chances of clinical acceptance (Chen and Bode, 2011). Therefore, for clinical use, assessing reactive gait using submaximal perturbations is warranted and hence we aimed to choose the perturbation intensities such that they would challenge the gait pattern yet without inducing an actual fall. For each perturbation type, we defined three intensity levels (Table 5.1).

**Table 5.1** Settings for low, medium and high intensity perturbations per type.

| Type | Intensity | Amplitude | Gain | Delay [s] |
|---|---|---|---|---|
| Sway | Low | 0.02 m | 0.5 | - |
| | Med | 0.04 m | 0.5 | - |
| | High | 0.05 m | 0.5 | - |
| Acc/Dec | Low | 40 % | 0.1 | 0.1 |
| | Med | 50 % | 0.1 | 0.1 |
| | High | 60 % | 0.1 | 0.1 |
| Visual | Low | 1* | 1 | 2 |
| | Med | 1* | 1 | 3.5 |
| | High | 1* | 1 | 5 |
| Auditory | Low | 20 | 0.1 | 0.1 |
| | Med | 40 | 0.1 | 0.1 |
| | High | 60 | 0.1 | 0.1 |

* 1 = virtual reality environment off.

Pilot data of a participant walking at 1.3 m/s were used to calculate the duration, displacement, velocity and acceleration, and the timing at which the maximum velocity and acceleration occurred. Average values are presented in Figure 5.3, Figure 5.4 and Table 5.2.

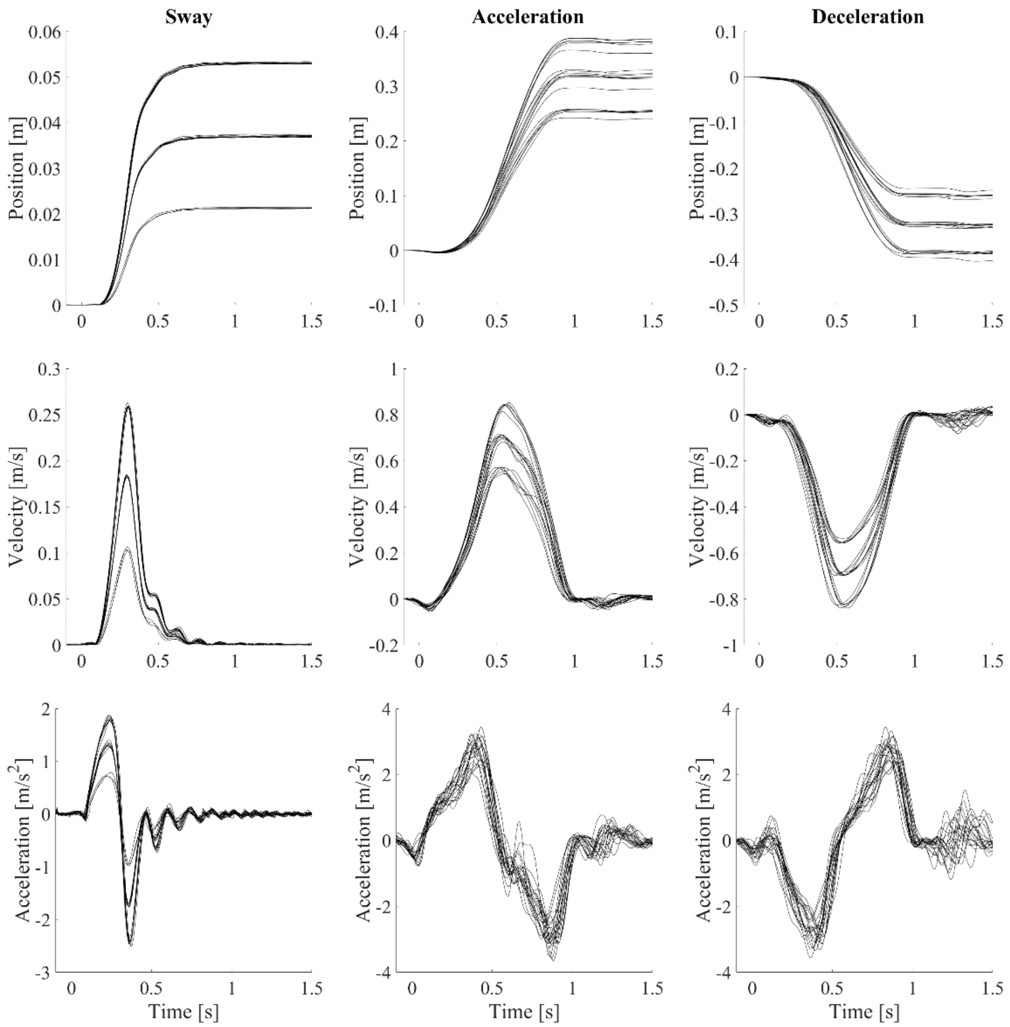



**Figure 5.3** Position, velocity and acceleration of the mechanical perturbations at low, medium and high intensity.

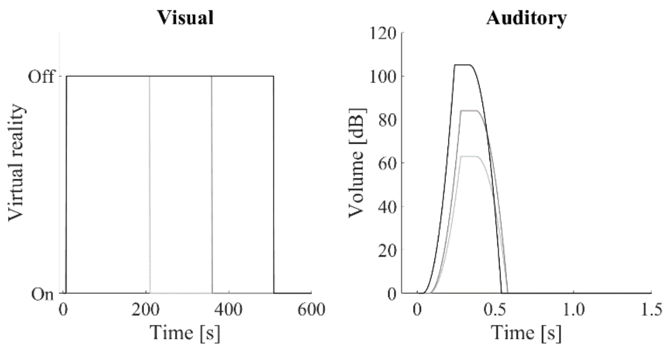



**Figure 5.4** Perturbation profiles for the visual and auditory perturbations at low medium and high intensity.

**Table 5.2** Characteristics for the mechanical perturbations. Mean and SD over six perturbations for sway, acceleration (Acc) and deceleration (Dec) perturbations per intensity level. Duration is the total perturbation duration. Note, the change in speed for the belt perturbations is expressed relative to the baseline speed which was set to 1.3 m/s.

| Type | | Duration [s] | Displacement [m] | Max Δvel [m/s] | Max vel time [s] | Max acc [m/s$^2$] | Max acc time [s] |
|---|---|---|---|---|---|---|---|
| Sway | Low | 0.233±0.006 | 0.021±0.000 | 0.104±0.002 | 0.130±0.000 | 0.965±0.022 | 0.190±0.000 |
| | Med | 0.341±0.004 | 0.037±0.000 | 0.184±0.001 | 0.151±0.006 | 1.727±0.030 | 0.210±0.008 |
| | High | 0.380±0.000 | 0.053±0.000 | 0.259±0.002 | 0.166±0.005 | 2.435±0.045 | 0.227±0.005 |
| Acc | Low | 0.830±0.009 | 0.257±0.006 | 0.564±0.011 | 0.407±0.031 | 2.413±0.196 | 0.268±0.026 |
| | Med | 0.840±0.015 | 0.323±0.009 | 0.703±0.011 | 0.390±0.008 | 2.987±0.118 | 0.266±0.027 |
| | High | 0.860±0.014 | 0.385±0.007 | 0.839±0.013 | 0.417±0.012 | 3.242±0.111 | 0.273±0.020 |
| Dec | Low | 0.848±0.029 | 0.252±0.006 | -0.552±0.008 | 0.385±0.034 | -2.545±0.129 | 0.243±0.031 |
| | Med | 0.890±0.035 | 0.319±0.004 | -0.693±0.005 | 0.405±0.028 | -2.944±0.169 | 0.270±0.039 |
| | High | 0.930±0.026 | 0.383±0.006 | -0.830±0.008 | 0.425±0.020 | -3.297±0.183 | 0.262±0.019 |

### 5.1.3 Research protocol

The perturbation protocol was performed at comfortable walking speed (CWS) determined as described by Hak and colleagues (2012). First, the treadmill speed was gradually increased until the participant reported to be walking at a comfortable speed. The treadmill speed was then increased further until an uncomfortable speed was reached. Thereafter, the treadmill speed was gradually decreased until the participant reported to be walking at a comfortable speed again. The first and second comfortable speed values were then averaged and used throughout the entire protocol.

Once comfortable walking speed was determined, participants completed a familiarization trial of three minutes. This was intended as a warm-up and to familiarize the participant with both the setup and treadmill walking. Thereafter, a two-minute baseline trial was completed to evaluate potential differences in steady state gait between young and older adults, as well as to examine whether the perturbations affected the gait pattern.

Following the steady state gait trial, six perturbation trials were completed, one for each perturbation type, in randomized order. A perturbation trial consisted of ten perturbations. An initial perturbation which was of medium intensity followed by three repetitions per intensity level (Figure 5.5). The later nine perturbations in a set of ten were randomized and included in the data analyses. The initial perturbation was not included to avoid overestimation of the perturbation response as a result of the "first-trial effect" (Owings et al., 2001; Sessoms et al., 2014). While evaluation of the initial perturbation by itself may be of interest, we did not explore this effect as it has been demonstrated that the "first trial effect" can be retained over a period up to 12 months (Liu et al., 2017). If reactive gait assessment will be adopted in clinical practice, it is likely that individuals will undergo the assessment multiple times within this time frame, for example during yearly fall risk screening or to evaluate balance after fall prevention training. Hence, one could argue that there is only one "true" first trial.

A 10-15 stride interval between subsequent perturbations was applied to ensure participants had fully recovered before the next perturbation would be applied. Each perturbation trial lasted about 2.5 minutes. Altogether, the perturbation protocol thus contained 60 perturbations applied over a period of approximately 15 minutes of walking. If needed, participants were allowed to rest in between the trials.

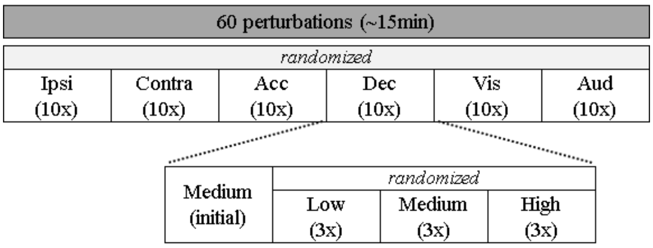



**Figure 5.5** Schematic overview of the perturbation protocol.

Finally, the participant's experience was evaluated after the walking trials. Questions regarding the overall experience (i.e. project aims, provided instructions, wearing tight clothes) and impact of the experiment (i.e. fear of falling, duration, exhaustion) as well as questions specifically evaluating perturbed walking (i.e. difficulty) were included (Appendix II).

## 5.2 Evaluation

The perturbation protocol as described in the previous section was evaluated in nine young and nine older adults. This is presented in journal paper format as an experimental report entitled "*Gait stability in response to platform, belt and sensory perturbations in young and older adults"* which was submitted to Medical & Biological Engineering & Computing on June 27$^{th}$, 2017 (section 5.2.1). In addition, in this section we will present the participants' experience based on the user evaluation questionnaire (section 5.2.2). Matlab codes for data analyses can be found in Appendix VI.

### 5.2.1 Experimental study 1 – Gait stability in response to platform, belt and sensory perturbations in young and older adults

**Abstract**

Reactive gait assessment has been used to quantify gait stability in older adults. However, knowledge on which perturbation type is most suitable to identify poor gait stability is lacking.

We evaluated the effects of ipsi- and contralateral sway, belt acceleration and deceleration, and visual and auditory perturbations on ML and BW margins of stability (MoS) in young and older adults. We aimed to evaluate 1) which perturbation type disturbed the gait pattern substantially, 2) how participants recovered and 3) whether recovery responses could discriminate between young and older adults. Nine young (25.1±3.4y) and nine older (70.1±7.6y) adults walked on the CAREN Extended (Motek Medical, The Netherlands). The perturbation effect was quantified by deviation in MoS over six post-perturbation steps compared to baseline walking. Contralateral sway and deceleration perturbations resulted in the largest ML (1.9-4 times larger than other types) and BW (1.6-5.6 times larger than other types) perturbation effects, respectively. After both perturbations types, participants increased MoS by taking wider, shorter and faster steps. No differences between young and older adults were found. We suggest to evaluate the potential of using contralateral sway and deceleration perturbations for identification of risk for falls by including both healthy and frail older adults.

## 1. Introduction

Gait impairments are among the main risk factors for falls in older adults (Ambrose et al., 2013). Since walking is one of the most common activities in our everyday life, it is not surprising that most falls occur while walking, due to trips or slips (Berg et al., 1997; Talbot et al., 2005). Gait stability assessment to identify individuals at risk for falls is therefore of great importance (Ambrose et al., 2013). Gait stability has been defined as “gait that does not lead to falls in spite of perturbations” and requires fast and accurate responses. However, the ability to respond adequately declines with age due to changes in the central nervous system and muscle properties (van Dieën and Pijnappels, 2017). Despite this knowledge, conventional balance and gait assessments solely evaluate self-initiated tasks (e.g. sit-to-stance transfers or turning). Such tasks allow for safe and controlled movement execution within ones limits of stability. Recovering from gait perturbations, on the other hand, targets fundamentally different stability components. Therefore, it has emerged over the last few decades as a method to quantify gait stability in research, but not yet in clinical practice (Bhatt et al., 2011a; Grabiner et al., 1993; Merrill et al., 2017; Pijnappels et al., 2005b; Süptitz et al., 2013).

The majority of gait perturbation studies have included AP perturbations using either moveable platforms (Oliveira et al., 2012b; Yang and Pai, 2014), obstacles (Crenshaw et al., 2013; Pavol et al., 2001; Pijnappels et al., 2010) in an overground walkway, slippery surfaces (Lockhart et al., 2005b), break-and-release systems (McCrum et al., 2016; Süptitz et al., 2012) or sudden treadmill belt accelerations and decelerations (Ilmane et al., 2015; Liu et al., 2015; Sessoms et al., 2014; Sloot et al., 2015). Additionally, ML perturbations have been applied by means of sideways platform movement (Hak et al., 2013b; McAndrew Young et al., 2012; Sturdy et al., 2014) or waist-pulls (Bruijn et al., 2010; Hof et al., 2010; Martelli et al., 2016; Toebes et al., 2014; Vlutters et al., 2016). Of less focus have been sensory perturbations, such as visual oscillations (Beurskens et al., 2014; O'Connor and Kuo, 2009) or low light conditions and distracting sounds (Rogers et al., 2008; Smith et al., 2010; Thies et al., 2005). Despite the growing body of work on the use of perturbations to evaluate one's ability to resist or recover from a perturbation, it remains difficult to compare the wide range of applied methodologies and determine which perturbation type is appropriate for reactive gait assessment.

The effect of perturbations on the gait pattern can be quantified by the ability to control the centre of mass (CoM) movement relative to the base of support (BoS) using measures like stabilizing and destabilizing forces, feasible-stability-region, and margins of stability (MoS) (Bruijn et al., 2013a). The latter is defined as the difference between the extrapolated centre of mass (XCoM; i.e. CoM position corrected for its velocity) relative to the border of the BoS. When the XCoM lies within the BoS one can be considered stable. In contrast, when the XCoM exceeds the border of the BoS, a corrective step needs to be taken to regain balance and avoid a fall, hence one can be considered unstable (Hof et al., 2005). In line with previous work, we quantified ML and backward (BW) MoS using the lateral and backward border of the BoS, respectively (Hak et al., 2013b). As such, taking wider steps (i.e. stepping more lateral to the XCoM) results in larger ML MoS while faster and shorter steps (i.e. stepping more behind the XCoM) results in larger BW MoS (Hak et al., 2013b). Stepping responses to successfully recover from gait perturbations may provide valuable input for the development of tailored fall prevention training programs.

We developed a gait perturbation protocol, including six different perturbation types: two ML platform perturbations, two AP uni-lateral belt perturbations and two sensory (visual and auditory) perturbations, and tested it on healthy young and older adults. Our first aim was to evaluate which types of external perturbations affect the gait pattern the most in terms of ML and BW MoS, and as such, would be most suitable for perturbation-based gait stability assessment. Secondly, we identified how spatio-temporal adjustments were used to recover ML and BW gait stability. Finally, we evaluated whether these perturbation responses were sensitive to discriminate between young and older adults. Resulting knowledge can contribute to the design of an optimal experimental protocol that would have the best predictive value in identifying older adults at risk for falls.

## 2. Methods

### 2.1. Participants

Nine young adults (3 women, age: 25.1±3.4 years, height: 1.76±0.09 m, weight: 76.6±15.1 kg) and nine healthy older adults (7 women, age: 70.1±8.1 years, height: 1.70±0.11 m, weight: 77.9±10.5 kg) participated in this study. Inclusion criteria were: normal lower limb function and being able to walk for 20 minutes. Exclusion criteria were: neuromuscular deficits or weighing more than 135 kg. The Biomedical Engineering departmental ethics committee at the University of Strathclyde approved the protocol before measurements were performed. All participants gave informed consent prior to the measurement.

### 2.2. Equipment

Participants walked on the CAREN (Computer Assisted Rehabilitation Environment) Extended (Motek Medical, Amsterdam, The Netherlands) at the University of Strathclyde, which consists of a six degree-of-freedom motion base with an instrumented dual-belt treadmill mounted on top, 12 infra-red Vicon Bonita cameras (Vicon, Oxford, United Kingdom) operating at 100 Hz and a virtual reality environment projected on a semi-cylindrical screen and a surround sound system. D-Flow software (version 3.20.0) was used to control all hardware components and to visualize the virtual environment (Geijtenbeek et al., 2011). The Human Body Model (Motek, Amsterdam, The Netherlands) containing 47 markers was used to calculate the body CoM (van den Bogert et al., 2013). Participants wore a safety harness to arrest potential falls.

### 2.3. Protocol

First the participant's dominant leg (preferred leg for kicking, climbing a stair and recovery from a push) was determined. Subsequently, comfortable walking speed (CWS) was assessed by first gradually increasing treadmill speed until the participant had reached a comfortable speed. Speed was then further increased until participants reported to be uncomfortable. Thereafter, speed was gradually decreased until a comfortable speed was reached again. The treadmill speed was fixed to the average of the two reported comfortable speeds (Hak et al., 2013a) after which a 3-minute familiarization and a 2-minute baseline trial were completed.

The perturbation protocol contained six perturbation types all triggered at non-dominant initial contact (Zeni et al., 2008): 1) ipsilateral sway consisting of a 5 cm platform translation in approximately 0.7s (maximum acceleration of 2.04 m/s$^2$) to the non-dominant side, 2) contralateral sway which was identical to the ipsilateral sway perturbation but to the dominant side; 3) unilateral belt acceleration of the non-dominant side to 160% CWS in approximately 0.4s (maximum acceleration of 2.43 to 5.13 m/s$^2$); 4) unilateral belt deceleration which was identical to the acceleration perturbation but with a minimum speed of 40% CWS; 5) a visual perturbation by rapidly darkening the room for 5 seconds to <1 lux and 6) an auditory perturbation in the form of a 0.5 second lasting air horn at 82 dB (Figure 5.6). The protocol consisted of six trials, each consisting of one perturbation type which was repeated four times. The six trials were presented in random order. Ipsilateral and contralateral sway trials always started with an ipsilateral and contralateral sway perturbation, respectively. The remaining perturbations were paired and presented in a pseudo-random order. This was necessary because the maximum platform excursion was 15 cm to each side.

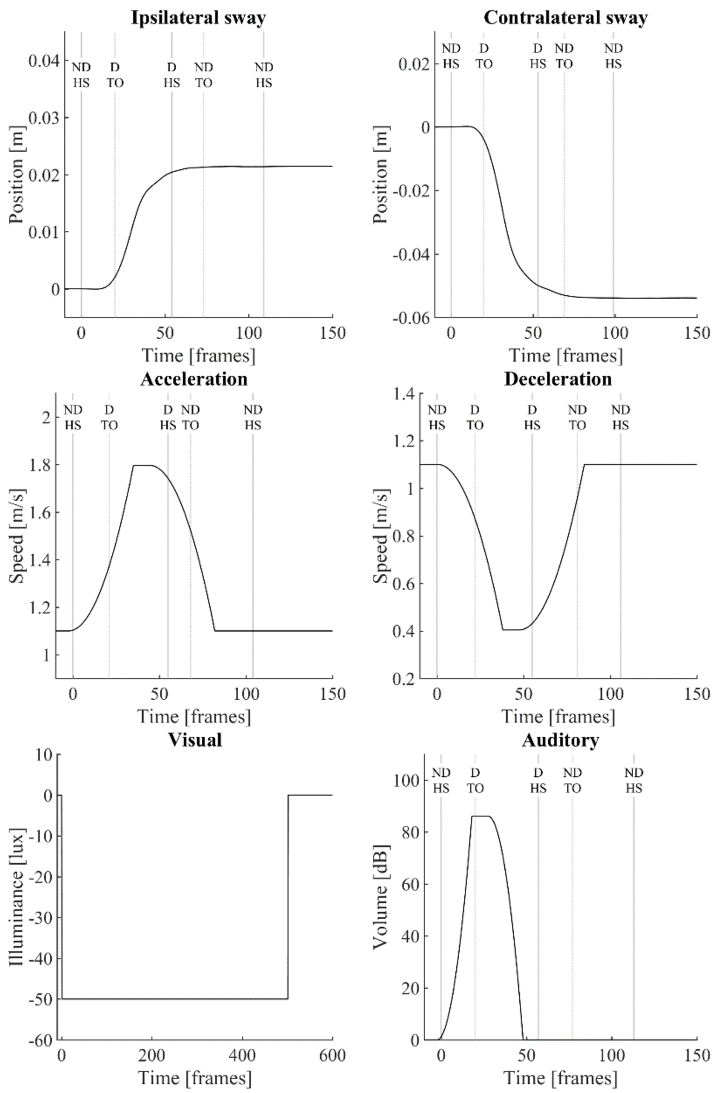



**Figure 5.6** Perturbation profiles over time (recorded at 100 Hz). Dominant (D) and non-dominant (ND) initial contact (HS) and toe-off (TO) events are indicated by the vertical lines. All perturbations were triggered at the ND HS at Time = 0

## 2.4. Data analyses

All data were analysed using custom-written Matlab scripts (version 2015a; The Mathworks, Natick, MA, USA). First, marker data was filtered using a 6 Hz second-order bidirectional Butterworth filter. Initial contact events were determined using the local maxima in the AP position of the heel marker relative to the pelvis (Zeni et al., 2008).

Three spatio-temporal gait parameters were calculated: step time, step length and step width. Step time, step length and step time were defined as the elapsed time, AP distance and ML distance between two consecutive initial contacts, respectively.

Gait stability was quantified by the MoS, as determined by the minimum distance between the border of the BoS and the XCoM. The XCoM was estimated by the CoM position plus its velocity divided by $\sqrt{g/l}$ in which $g$ is the acceleration of gravity and $l$ the average greater trochanter markers' height times 1.34 (Hof et al., 2005). The ML lateral malleolus marker position of the leading foot quantified the ML border of the BoS whereas the AP heel marker position was used to define the BW border (Figure 5.7). Thereby, negative ML and BW MoS values indicated *instability* in the lateral and backward direction, respectively.

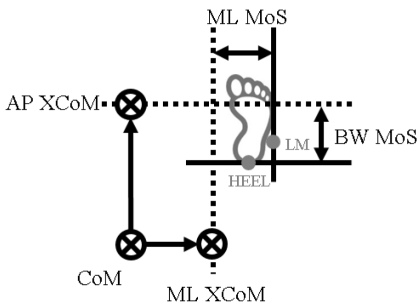



**Figure 5.7** Schematic representation of margins of stability (MoS) for the right side in the medio-lateral (ML) and backward (BW) direction

Baseline values for spatio-temporal parameters and MoS were calculated and averaged over 100 consecutive dominant (BD) and 100 consecutive non-dominant (BND) steps.

Additionally, local dynamic stability (LDS) of ML, AP and vertical (VT) trunk velocity over the same 100 strides was calculated as described in Bruijn et al. (2009) and used to evaluate unperturbed gait stability. LDS reflects the ability to cope with small internal perturbations (e.g. variability in neuromuscular control) rather than external perturbations and has been used to detect age-related decline in steady state gait stability (Buzzi et al., 2003; Kang and Dingwell, 2009; Terrier and Reynard, 2015). Lower LDS values imply more stable gait.

For the perturbation trials, spatio-temporal parameters and MoS were calculated for six steps pre- and six post-perturbation steps. To quantify which perturbation type affected the gait pattern the most, the difference of the six post-perturbation steps (1D, 2ND, 3D, 4ND, 5D, 6ND) with respect to BD and BND steps was calculated as:

$$6S = \sum_{i=1}^{2} \sum_{j=1}^{3} \sqrt{\mathrm{B}(i) - P(i + (j-1) * 2)^2} \qquad \textbf{Equation 5.1}$$

where *6S* is the deviation from baseline walking, *B* is baseline step for the $i^{th}$ side (with $i$=1 representing the dominant side and $i$=2 representing the non-dominant side) and *P* is the post-perturbation step for $j^{th}$ stride. We hereby captured the overall deviation from steady state walking while ignoring differences in recovery over subsequent steps. For example, a large initial deviation in step width but quick recovery to baseline values may result in similar 6S values as compared to a small initial deviation but slow return to baseline values.

Gait stability and stepping strategies in response to the perturbations were analysed by comparing average dominant pre-perturbation steps (PD) to dominant post-perturbation steps (i.e. 1D, 3D, 5D) and average non-dominant pre-perturbation steps (NPD) to non-dominant post-perturbation steps (i.e. 2ND, 4ND, 6ND). All perturbation measures were averaged over the last three perturbations of each perturbation trial.

### 2.5. Statistical analyses

All statistical analyses were performed using SPSS version 23 (SPSS Inc, Chicago, IL, US). Gait parameters were tested for normality using a Shapiro-Wilk test. Differences between young and older adults in steady state gait stability (i.e. LDS) were analysing using independent *t*-tests. To evaluate whether dominant and non-dominant gait parameters differed at baseline, a mixed-model analysis of variances (ANOVA) was used (within factors: two sides; between factor: group). To examine which types of gait perturbations affect the gait pattern the most in terms of ML and BW MoS, mixed-model ANOVAs (within: six perturbation types; between: group) were applied for the total perturbation response (i.e. 6S). Post-hoc pairwise comparisons were then used to find the perturbation types that affected gait stability the most for ML and BW directions. Subsequently, recovery from the perturbation was evaluated by analysing individual post-perturbation steps (i.e. 1D-6ND). The data dictated that participants pro-actively adapted their gait in anticipation of subsequent perturbations. Therefore, we first examined how participants adapted their gait by comparing baseline walking to pre-perturbation steps for all gait parameters using mixed-model ANOVAs (within: baseline and pre-perturbation step; between: group). Thereafter, to examine how participants recovered from the perturbations in terms of spatio-temporal parameters and ML and BW MoS, mixed-model ANOVAs (within: two pre-perturbation and six post-perturbation steps; between: group) for the individual steps were used. A Greenhouse-Geisser correction was used when the assumption of sphericity was violated. Post-hoc paired-samples *t*-tests with a Bonferroni correction for each perturbation type were used to investigate whether post-perturbation steps differed from pre-perturbation steps. The level of significance was set at 0.05.

## 3. Results

All participants completed the protocol without falling in the harness. Mean CWS (Y: 1.26±0.17 m/s, O: 1.17±0.23 m/s) did not significantly (*t*=0.888, *p*=0.388) differ between young and older adults.

### 3.1. Baseline walking

Except for a larger dominant than non-dominant ML MoS (*t*=5.702, *p*<0.001) in both younger and older participants, the mixed-model ANOVA did not reveal any main or interaction effects when comparing dominant and non-dominant steps. LDS was not significantly different between young and older adults in any direction (ML: *p*=0.835; BW: *p*=0.164; VT: *p*=0.516. Table 5.3).

**Table 5.3** Mean±SD of baseline gait parameter for the dominant (BD) and non-dominant (BND) steps in young and older adults. Significant effects at p<0.05 are printed in bold.

| Parameter | | Young adults | Older adults | Main effect (Sides) | | Between Subjects effect (Group) | | Interaction effect (Steps x Group) | |
|---|---|---|---|---|---|---|---|---|---|
| | | Mean±SD | Mean±SD | F | p | F | p | F | p |
| ML MoS [m] | D | 0.065±0.012 | 0.061±0.013 | **31.339** | **<0.001** | 1.218 | 0.286 | 0.388 | 0.542 |
| | ND | 0.053±0.006 | 0.047±0.013 | | | | | | |
| BW MoS [m] | D | 0.169±0.047 | 0.142±0.052 | 2.888 | 0.109 | 1.729 | 0.207 | 1.009 | 0.330 |
| | ND | 0.175±0.044 | 0.143±0.045 | | | | | | |
| Step time [s] | D | 0.544±0.046 | 0.555±0.033 | 0.015 | 0.905 | 0.285 | 0.600 | 0.113 | 0.741 |
| | ND | 0.545±0.043 | 0.554±0.032 | | | | | | |
| Step length [m] | D | 0.680±0.066 | 0.656±0.125 | 2.488 | 0.134 | 0.430 | 0.521 | 2.113 | 0.165 |
| | ND | 0.692±0.066 | 0.656±0.113 | | | | | | |
| Step width [m] | D | 0.126±0.032 | 0.113±0.056 | 1.539 | 0.233 | 0.356 | 0.559 | 0.293 | 0.596 |
| | ND | 0.126±0.032 | 0.113±0.056 | | | | | | |
| LDS ML | - | 1.829±0.294 | 1.798±0.334 | - | - | - | - | - | - |
| LDS AP | - | 1.585±0.263 | 1.404±0.262 | - | - | - | - | - | - |
| LDS VT | - | 1.826±0.428 | 1.701±0.366 | - | - | - | - | - | - |

### 3.2. Which perturbation type affected the gait pattern the most?

The gait pattern was differently affected by the different perturbation types, without group or interaction effects (Main effects of perturbation for 6S ML MoS $F$=76.023, $p$<0.001, and for 6S BW MoS $F$=85.281, $p$<0.001). Post-hoc pairwise comparisons revealed that 6S ML MoS in response to the contralateral sway perturbation was significantly larger compared to all other perturbation types meaning that ML MoS deviated most from baseline waking after the contralateral sway perturbation (mean difference: 0.103-0.159 m; all at p<0.001) (Figure 5.8a). Similarly, 6S BW MoS was significantly larger for the deceleration perturbation compared to all other perturbation types (mean difference: 0.287-0.430 m; all at p < 0.001) (Figure 5.8b). Based on the significant effects of the contralateral sway and deceleration perturbation on 6S ML and BW MoS respectively, these perturbation types were further investigated.

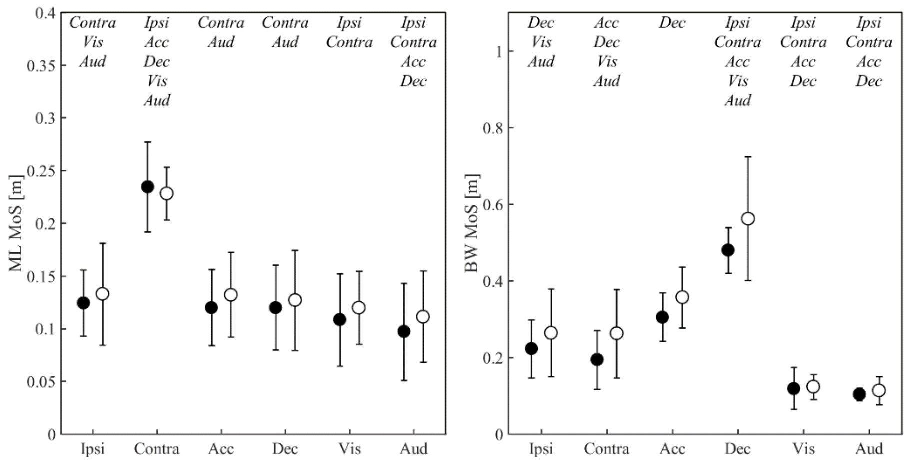



**Figure 5.8** The overall perturbation effect 6S (see text for details). Mean and standard deviations for medio-lateral (ML) and backward (BW) margins of stability (MoS) after the ipsilateral sway (Ipsi), contralateral sway (Contra), acceleration (Acc), deceleration (Dec), visual (Vis) and auditory (Aud) perturbations. Black dots represent values for young adults whereas white dots represent older adults. Significantly different pairwise comparisons are indicated at the top per perturbation type in italic

### 3.3. How did participants adapt their gait in between perturbations?

Step width significantly increased prior to the contralateral sway perturbation compared to baseline walking for the dominant side (BD: 0.120±0.045, PD: 0.127±0.053m, $F$=4.830, *p*=0.043) and a trend toward a significant increase was found for the non-dominant side (BD: 0.120±0.045, PD: 0.127±0.053m, *F*=4.150, *p*=0.059).

BW MoS prior to the deceleration perturbation was significantly larger (i.e. more stable in the backward direction) compared to baseline walking for the non-dominant side (BD: 0.166±0.043, PD: 0.181±0.041m, *F*=11.709, *p*=0.004) and near significant for the dominant side (BD: 0.162±0.047, PD: 0.171±0.044m, *F*=4.231, *p*=0.059). Step width was significantly increased prior to the perturbation compared to baseline walking for both the dominant (BD: 0.128±0.038, PD: 0.141±0.044m, *F*=12.492, *p*=0.003) and non-dominant (BD: 0.129±0.038, PD: 0.143±0.043m, *F*=10.119, *p*=0.007) side.

### 3.4. How were spatio-temporal adjustments used to recover ML and BW gait stability?

Mixed-model ANOVAs for the contralateral sway perturbation revealed significant main effects of Steps on all gait parameters while no significant Group or Group x Steps interaction effects were found. Post-hoc analyses showed that step width and ML MoS were reduced at step 1D (Figure 5.9 and Table A-III.1). Step width and ML MoS increased during step 2ND though ML MoS values remained smaller than at baseline. Thereafter, both parameters increased and remained larger compared to baseline walking. In other words, ML stability was initially compromised by the contralateral sway perturbation but was restored to values greater compared to baseline walking during the subsequent recovery steps. Step length (1D to 6ND) and step time (2ND to 5D) decreased, while BW MoS (2ND to 6D) increased meaning that participants became more stable in the backward direction. Figure 5.10 shows the relation between gait stability and spatio-temporal parameters for a typical response contralateral sway perturbation response.

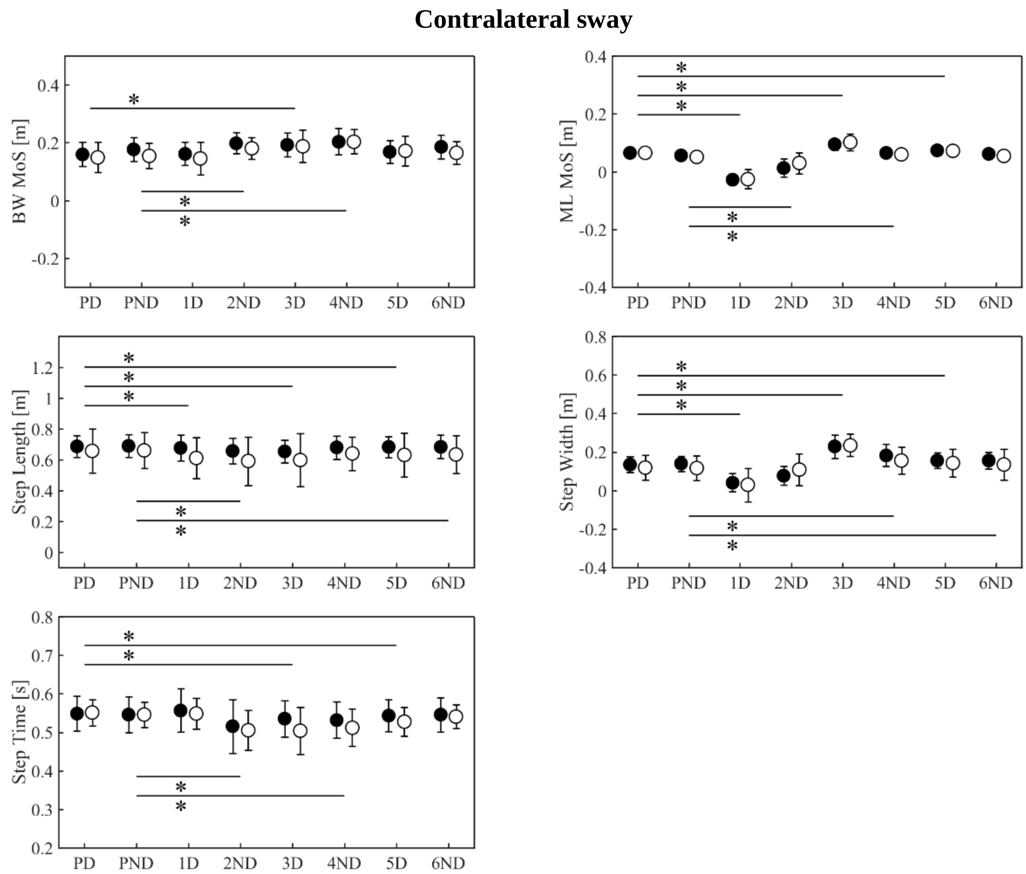



**Figure 5.9** Responses to the contralateral sway perturbations. Mean and standard deviations of backward (BW) and medio-lateral (ML) margins of stability (MoS), step length, -width and -time for pre-perturbation (PD and PND) and post-perturbation (1D to 6ND) steps. Black dots represent for values young adults whereas white dots represent older adults. Significant differences between pre- and post-perturbation steps are indicated with *

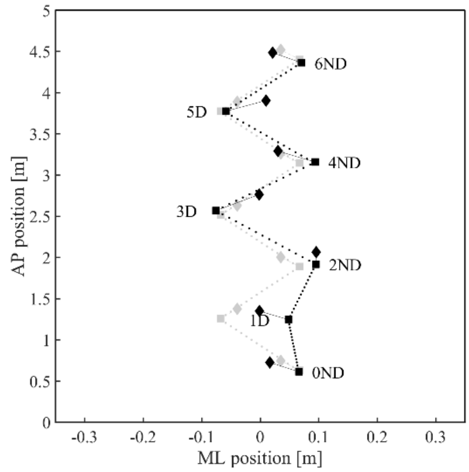



**Figure 5.10** A schematic representation of a typical response (black) to a contralateral sway perturbation as compared to baseline walking (grey). Squares represent ML (step width) and AP (step length) foot placement whereas the line density is an indication of the time elapsed (step time) between consecutive steps. Margins of stability in the ML and BW direction are indicated by the diamonds. The perturbation was triggered at step 0ND

Mixed-model ANOVAs for the deceleration perturbation revealed significant main effects for all Steps on all gait parameters, while no significant Group or Group x Steps interaction effects were found. Post-hoc analyses revealed a reduction to negative BW MoS (i.e. instability in the backward direction) at step 2ND (Figure 5.11 and Table A-III.1). During step 3D to 5D, BW MoS was increased to values larger than those at baseline. A significant reduction in step time was found during step 3D and 4ND, whereas step lengths reduced during step 1D, 2ND, 4D and 6ND. Moreover, both ML MoS (step 3D and 4ND) and step width (1D, 3D and 4ND) increased. Figure 5.12 shows the relation between gait stability and spatio-temporal parameters for a typical deceleration perturbation response. Results on the analyses of the other perturbation types can be found in Appendix III (Figure A-III.1 and Table A-III.1).

## 4. Discussion

We developed a gait perturbation protocol containing two platform, two belt and two sensory perturbations. Our main aim was to evaluate which perturbation type affected stability the most in young and older adults. We found very little differences in our groups of participants. However, the results showed that all mechanical perturbations effectively altered the gait pattern in both young and older adults while the sensory perturbations did not affect the gait pattern. The contralateral sway and deceleration perturbation appeared most challenging. Visual and auditory perturbations did not affect the gait pattern. This is in line with previous work, which showed that low light conditions did not affect spatio-temporal parameters (Rogers et al., 2008; Thies et al., 2005). To our knowledge auditory perturbations by means of acoustic startles have not been investigated previously.

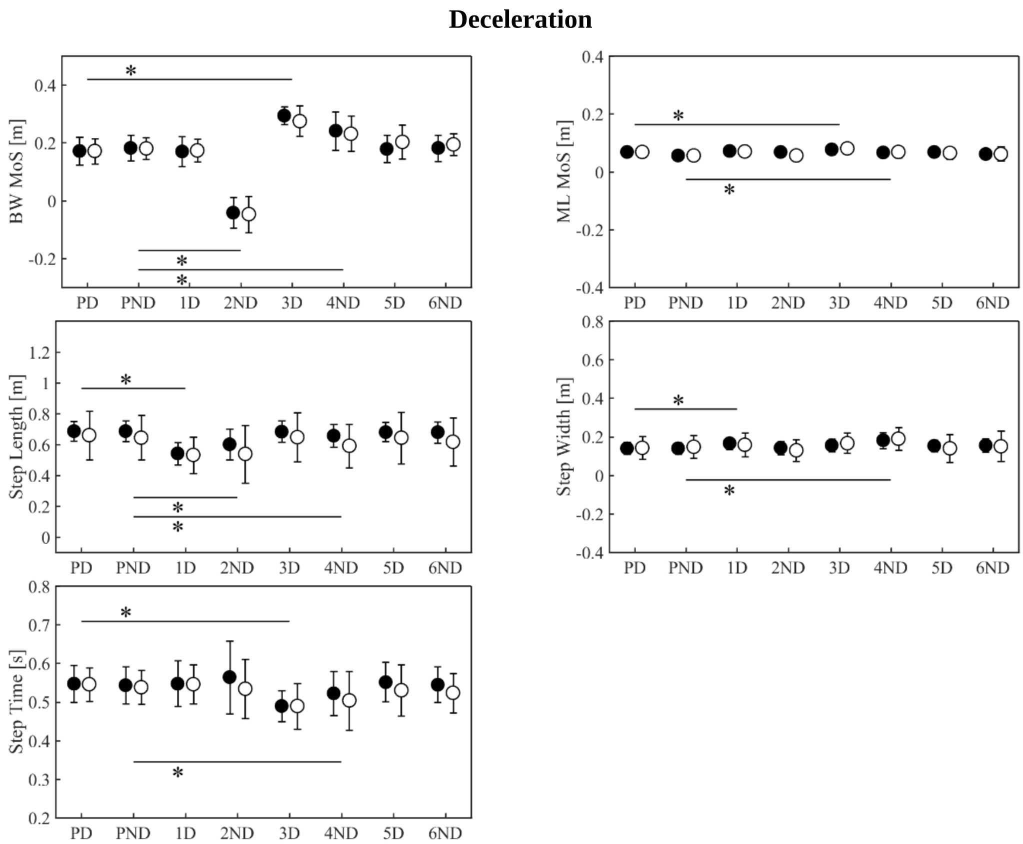



**Figure 5.11** Responses to the deceleration perturbations . Mean and standard deviations of backward (BW) and medio-lateral (ML) margins of stability (MoS), step length, -width and -time for pre-perturbation (PD and PND) and post-perturbation (1D to 6ND) steps. Black dots represent for values young adults whereas white dots represent older adults. Significant differences between pre- and post-perturbation steps are indicated with *

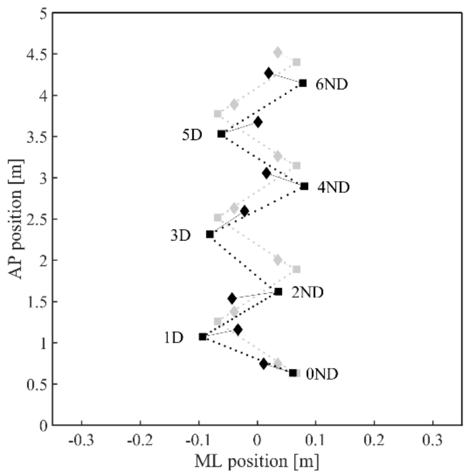



**Figure 5.12** A schematic representation of a typical response (black) to a deceleration perturbation as compared to baseline walking (grey). Squares represent ML (step width) and AP (step length) foot placement whereas the line density is an indication of the time elapsed (step time) between consecutive steps. Margins of stability in the ML and BW direction are indicated by the diamonds. The perturbation was triggered at step 0ND

The contralateral sway perturbation (i.e. platform movement to the right at left initial contact or the left at right initial contact) induced BoS movement towards the XCoM and thus ML MoS decreased. Consequently, the majority of the participants were required to take a cross-step to prevent falling. Following the initial perturbation response, ML MoS was recovered by taking faster, shorter and wider steps. Due to the adaptations in step length and step time, BW MoS increased as well (Hak et al., 2013b). Likewise, the deceleration perturbation reduced the distance between the border of the BoS and the XCoM in the backward direction and thus BW MoS initially decreased. Again, stability was recovered by taking faster, shorter and wider steps. Previous work from Hof and colleagues (2010) reported comparable perturbation responses after ML waist-pushes. By definition, a BoS perturbation is expected to have a similar effect on ML MoS as a CoM perturbation in the opposite direction. Indeed, Hof and colleagues' (2010) waist-pushes to the left at left initial contact were more challenging as compared to left pushes at right initial contact. The fact that acceleration perturbations appeared less challenging as compared to decelerations has been demonstrated previously in younger adults but to our knowledge not in older adults (Ilmane et al., 2015). Ilmane and colleagues (2015) showed that the initial effect of the acceleration perturbations was larger compared to the deceleration, but recovery from the deceleration perturbations took much longer (up to four steps compared to one for the acceleration perturbation). The reduction in XCoM induced by the deceleration perturbation is extra challenging as one needs to maintain forward velocity to keep up with the treadmill speed.

While contralateral sway and deceleration perturbations evoked the largest responses, this does not necessarily mean that ipsilateral sway and acceleration perturbations should not be included in perturbation-based gait assessment. However, by applying more challenging perturbations, the (in)ability to adequately recover may be more profound and hence the perturbation response may be more sensitive to discriminate between fallers and non-fallers. The question whether the contralateral sway or deceleration perturbation is most challenging is more difficult to answer. Deviation from baseline (6S) was more than twice as large for the deceleration (0.52±0.12 m) compared to the contralateral sway perturbation (0.22±0.03 m). However, the fact that an ML change in *position* was induced by the sway perturbation as opposed to an AP change in *velocity* by the deceleration perturbation limits direct comparison. In a recent study, McIntosh et al. (2016) used ML and AP overground platform perturbations in young and older adults and found that contralateral sway perturbations were most challenging, followed by ipsilateral sway and then forward-backward perturbations. However, they quantified perturbation response by CoM displacement and velocity, while ignoring its relation to the BoS. Hence it is unknown to what extent stability was affected. Additionally, whether ML or AP perturbations are more challenging may be patient-specific as a result of individual risk factors for falls such as decline in muscle strength or ineffective stepping strategies (Ambrose et al., 2013). Therefore, including both contralateral sway and deceleration perturbations in gait stability assessment might give a more complete representation of one's ability to resist or recover from a gait perturbation.

Successful recovery from a perturbation is determined by the combination of stability prior and in response to the perturbation. By pro-actively increasing gait stability, one might reduce the effect of the perturbation and minimize risk of falling (Bhatt et al., 2006; McIntosh et al., 2017; Yang et al., 2016). Although we did not aim to evaluate such adaptations, the data revealed that participants pro-actively adapted their gait pattern. Gait adaptations were perturbation type specific but did not differ between age groups. Of interest would be to investigate whether more frail older adults show similar pro-active gait adaptations and whether these adaptions are indicative of risk for falls.

In contrast to our expectations based on previous studies (McIntosh et al., 2017; Pijnappels et al., 2005b; Senden et al., 2014; Süptitz et al., 2013), we did not find any differences in perturbation effects and recovery responses between young and older adults. This may be explained by the fact that the majority of our older adults were recruited through fitness classes and therefore very fit and healthy. This potential selection bias was confirmed by the non-significant differences in steady state local dynamic stability during baseline walking, which is known to decrease with age (Buzzi et al., 2003; Kang and Dingwell, 2009; Terrier and Reynard, 2015). Furthermore, the perturbation intensities may have been too low to provoke responses close to the individuals' boundaries. For example, McIntosh and colleagues (2016) used 15 cm ML platform excursions to discriminate between young and older adults as opposed to 5 cm in this study. Decelerations of 8 $m/s^2$ (as opposed to our 2.43-5.13 $m/s^2$) were used to distinguish fallers from non-fallers (Ding and Yang, 2016). The perturbation intensities in this study were chosen such that a fall would not be induced, which we believe is preferable in clinical practice, but higher intensities might be required to reveal subtle group differences. Additionally, within this fit group more sensitive outcome measures may have been required to discriminate between young and older adults. For example, evaluation of trunk kinematics may have been of added value (Crenshaw et al., 2012).

## 5. Conclusion

No differences between young and older adults were found in the recovery response to ML platform, AP belt and sensory perturbations. However, our results revealed that contralateral sway and deceleration perturbations show most potential in disturbing the gait pattern in young and healthy older adults. Therefore, including these specific perturbation types in perturbation-based gait assessment may be preferred over ipsilateral sway, acceleration, visual or auditory perturbations. Further investigation including comparison between older adults with and without a history of falls and possibly at higher intensities is required to see if and how perturbation-based gait assessment can be used to identify risk for falls in the older adults.

### 5.2.2 User evaluation

Participant experience was evaluated using a five-point Likert scale (Appendix II). Both young and older adults found it useful to evaluate risk for falls by means of gait assessment and understood why we explored the use of gait perturbations to assess risk for falls. Additionally, they indicated that the instructions provided were sufficient. Despite the unfamiliar experimental setup and the fact that they were aware that their gait pattern would be challenged, they experienced limited fear of falling. Furthermore, the experiment was not too long, nor too tiring. Some older adults felt uncomfortable wearing tight shorts and a lycra top, which were required to avoid movement artefacts of the markers (Figure 5.13).

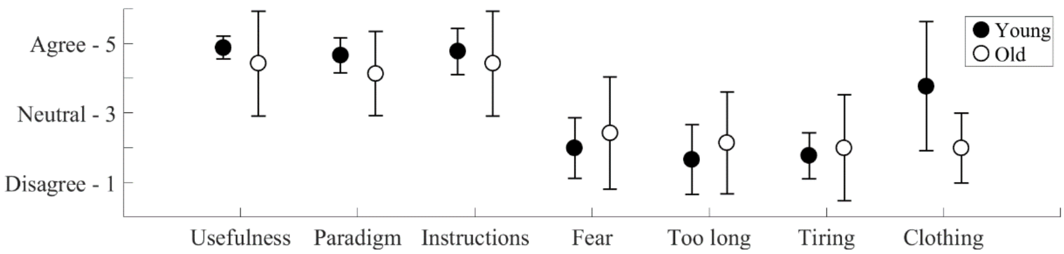


**Figure 5.13** Participant feedback.

### 5.2.3 Discussion

The present study explored the use of gait perturbations to assess reactive gait stability in young and older adults. We aimed to evaluate both the discriminative capacity of our perturbation protocol as well as the feasibility of using such an assessment in older adults. While the mechanical perturbations challenged the gait pattern of both young and older adults, our perturbation protocol did not discriminate between the two groups. The older adults recruited in our study, however, understood the potential contribution of reactive gait assessment for fall risk identification in older adults and were willing to undergo the perturbation protocol but felt more reluctant toward wearing tight clothes compared to young adults. Therefore, our next iteration of the protocol will:

- Focus on improving discriminative capacity of our perturbation protocol, which could potentially be achieved by
    - Including (more) challenging perturbations
    - Including (more) sensitive outcome measures
- Take a step toward clinical implication by evaluating the accuracy of reduced marker sets to minimize preparation time and complexity in data collection
- Examine perturbation responses of older fallers to evaluate whether we can discriminate between older fallers and non-fallers

# CHAPTER 6

# Reactive gait assessment development

# Phase 2

## 6.1 Development

No spatio-temporal differences in responses to perturbations were found previously between young and older adults. Towards the overall aim of developing a protocol that can discriminate fallers from non-fallers, we further explored a modified version of the perturbation protocol aiming to improve the discriminate power by: 1) more challenging perturbation with higher intensities, 2) decrease expectancy to the gait perturbations and minimizing a subsequent proactive increase in gait stability and 3) including potentially more sensitive outcome measures. Furthermore, we added various clinical assessments to describe our population of older adults. The current section provides an overview of the modifications in the study design.

### 6.1.1 Perturbation protocol modifications

The following modification in the perturbation protocol to that reported previously were made (Appendix VI):

1. **Perturbation types:** the visual and auditory perturbations were removed from the protocol as they did not challenge the gait pattern in young nor in older adults.
2. **Increase in perturbation intensities:** the intensity levels of the "High" perturbations were further increased. The change in speed of the belt perturbations was increased from 60% gait speed to 100% gait speed. Further, belt accelerations and decelerations were increased from ±3 m/s$^2$ to almost 10 m/s$^2$. The actual deceleration was slightly lower (almost 9 m/s$^2$). For the sway perturbations, the velocity was increased from 0.26 to 0.31 m/s, the amplitude remained similar (i.e. 0.05 m), due to the limited maximum excursion of the platform. The comparative data is presented in Table 6.1 and the resulting perturbation profiles in Figure 6.1.
3. **Perturbation side:** perturbations were applied to the left and right side rather than the non-dominant side only, to minimize anticipation and hence proactive gait adaptations.

4. **Perturbation randomization:** to further minimize proactive gait adaptations, the different perturbations types were randomized. A randomized block design was used to allow for breaks in the protocol if required, for example in case the participant needed an extra break or a marker fell off. A block consisted of eight perturbations, four perturbations types applied to each side. Five blocks were applied, totalling 40 perturbations (Figure 6.2).
5. **Fixed gait speed:** we aimed to have participants walk at a 1 m/s to standardize the perturbation intensity and its relative timing in the gait phase.

Furthermore, for comparison between the studies performed in first and second phase, the highest intensity perturbations from the previous study were also included (Figure 6.1). Finally, the multiple head and arm markers were removed from the marker set to reduce preparation time as they contributed little to the estimate of the CoM position (Appendix I).

**Table 6.1** Perturbation characteristics for a baseline gait speed of 1 m/s.

| Type | Phase | Duration [s] | Displacement [m] | Max Δvel [m/s] | Max vel time [s] | Max acc [m/s$^2$] | Max acc time [s] |
|---|---|---|---|---|---|---|---|
| Sway | 1 | 0.380 | 0.053 | 0.260 | 0.170 | 2.421 | 0.230 |
| | 2 | 0.250 | 0.053 | 0.301 | 0.130 | 3.256 | 0.230 |
| Acc | 1 | 0.830 | 0.295 | 0.636 | 0.380 | 2.999 | 0.280 |
| | 2 | 0.290 | 0.138 | 0.950 | 0.140 | 9.727 | 0.080 |
| Dec | 1 | 0.940 | 0.293 | -0.626 | 0.510 | -2.925 | 0.370 |
| | 2 | 0.370 | 0.133 | -0.882 | 0.180 | -8.737 | 0.110 |

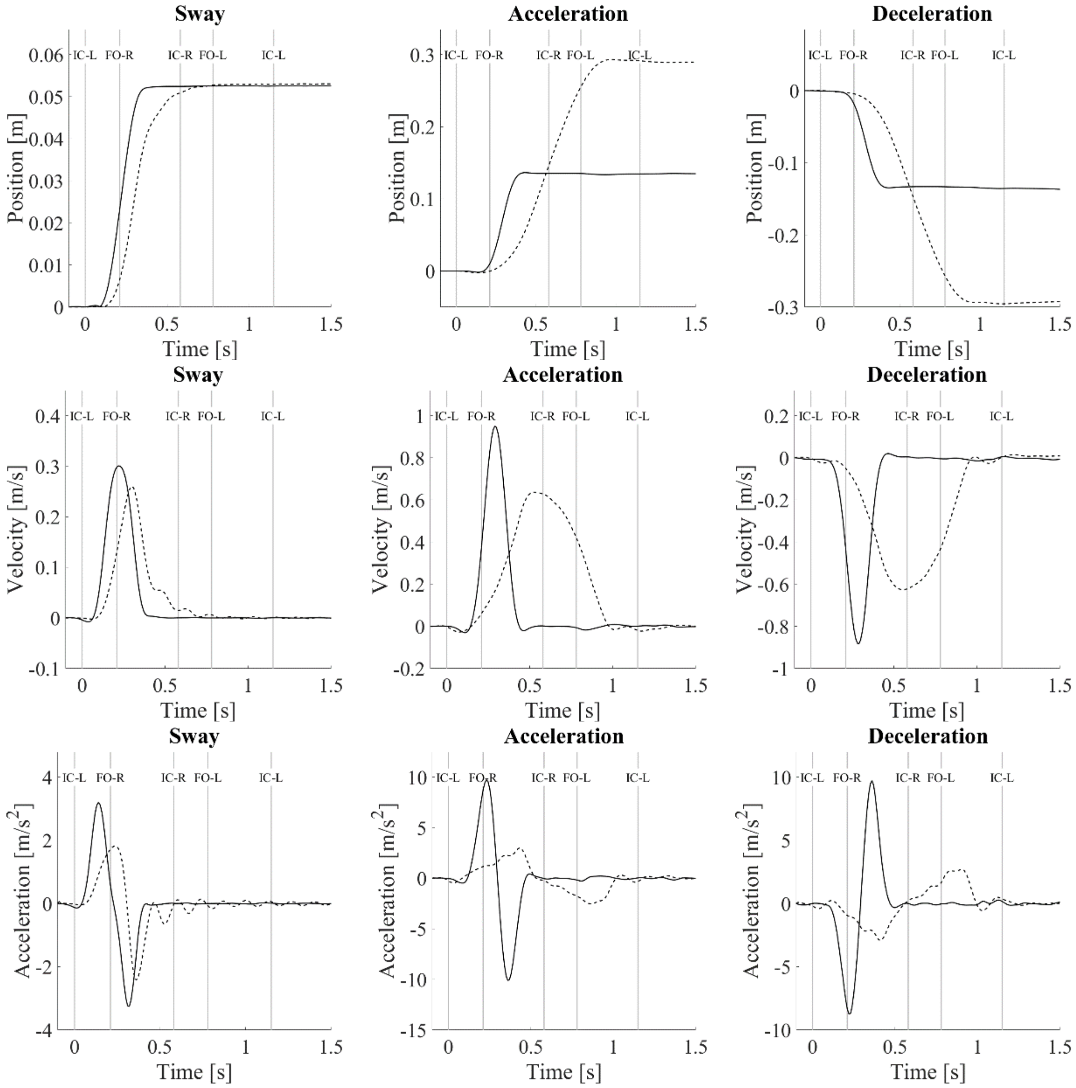



**Figure 6.1** Comparative examples of perturbations profiles at intensities used in the study comparing young and older adults (dashed lines) and the study comparing fallers and non-fallers (solid lines) study. Left (L) and right (R) initial contact (IC) and foot off (FO) events are provided to give an indication of the relative timing in the gait cycle. In this example, the perturbations are triggered at IC-L.

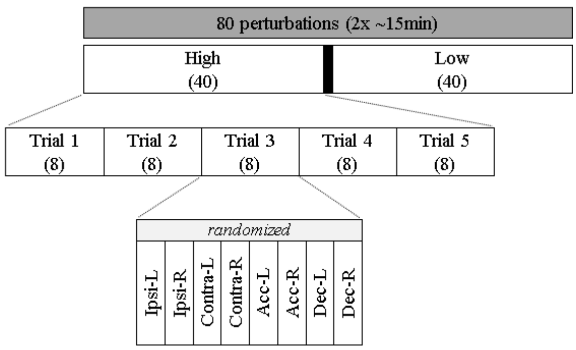



**Figure 6.2** Schematic overview of perturbation protocol.

### 6.1.2 Outcome measures

In preparation for data collection, we noticed considerable trunk movement in response to the amended gait perturbations, particularly in response to the deceleration perturbations. The trunk alone contains nearly 50% of the total body weight, and two thirds when including the head and arms (MacKinnon and Winter, 1993). In combination with its significant height above the support surface, trunk movement has a considerable effect on the CoM position. While the ability to control the trunk when recovering from gait perturbations has been addressed previously (Pavol et al., 2001; Pijnappels et al., 2005b), gait stability measures (section 2.3) indirectly include trunk movement by evaluating overall CoM movement. Nonetheless, Crenshaw and colleagues (2012) brought to the attention that trunk kinematics (i.e. trunk flexion angle and velocity) and stability measures (time-to-boundary, reflecting the time it would take for the CoM to reach the border of the BoS given the current velocity and/or acceleration) may reflect different aspects of recovery responses. In a study evaluating recovery from large postural perturbations, they demonstrated that both trunk kinematics and stability measures could be used to accurately classify falling and successful recovery following a large postural perturbation (92.3% and 80.8% accuracy, respectively). However, low correlations ($0.20<r<0.52$) between trunk kinematics and stability measures were found indicating that they quantify different components of recovery capacity. Therefore, we added trunk velocity in three directions (AP, VT and ML) and three planes (frontal, transversal and sagittal) to our previously selected spatio-temporal parameters and gait stability measures (Appendix VI).

### 6.1.3 Clinical assessment

Self-reported history of falls in the past 12 months was used to classify older adults. Individuals who experienced at least one fall were classified as fallers. In case of a fall history, further questions were asked regarding circumstances, cause, frequency and fall-related injuries.

We included the Timed Up and Go test (TUG) which is a widely used clinical assessment tool to assess physical performance and risk for falls in older adults and various patient populations (Podsiadlo and Richardson, 1991). The patient is asked to rise from a chair, walk three meters in a straight line at a self-selected speed, turn around, walk back to the chair and sit down again. The time it takes to complete the task gives an indication of functional performance, with a cut-off time greater than 13.5 s indicating a risk for falls (Shumway-Cook et al., 2000). Despite being among the most commonly accepted clinical fall risk assessments, it has repeatedly been demonstrated that the TUG fails to identify individuals at risk in older adults, especially in the active population (Barry et al., 2014; Laessoe et al., 2007). As such, we did not expect that the TUG would discriminate between those with and without a history of falls, but we included this assessment to describe our population.

The one-legged stance test (OLST) is a simple, commonly used assessment of postural stability. It evaluates the time one can stand on a single leg. A review showed that some found reduced stance time in fallers whereas other did not find any differences either prospectively or retrospectively in fallers (Persad et al., 2010). A possible explanation could be the lack of standardization in for example the selected stance leg, eyes open or closed or maximum measurement time. Our participants were instructed to stand on their preferred leg, with their eyes open, for as long as possible but with a maximum of 30 seconds.

The Falls Efficacy Scale (FES) assesses the concern of falling during a selection activities of daily living. The Dutch version, including ten activities, with a scale of 10-40 was used (Bosscher et al., 2005).

Physical activity was assessed by means of the Physical Activity Questionnaire (PAQ), a short questionnaire of three items evaluating how often one participates in very, moderately and mildly vigorous activities (Innerd et al., 2015). While this assessment has been designed for the elderly population (>85 years of age), it allows for quick examination of an individual’s activity level.

## 6.2 Evaluation

The modified reactive gait assessment as described in the previous section was evaluated in 10 older adults with and 39 older adults without a history of falls. This is presented in journal paper format as an experimental report entitled "*Can responses to different gait perturbations discriminate between older adults with and without a history of falls*".

### 6.2.1 Experimental study 2 – Can responses to different gait perturbations discriminate between older adults with and without a history of falls?

**Abstract**

Gait impairments are among the main risk factors for falls in older adults. In addition to steady state gait, reactive gait assessment (i.e. quantifying the ability to recovery from gait perturbations) may be relevant to identify risk for falls. We developed a mixed perturbation protocol, including submaximal perturbation intensities, and investigated its ability to discriminate between older adults with and without a history of falls, based on recovery responses.

Forty-nine older adults walked on the Gait Real-time Analysis Interactive Lab (GRAIL, Motek, The Netherlands), while being exposed to four types of perturbations: ipsilateral sway, contralateral sway, belt acceleration and deceleration perturbations. Responses to the perturbations were quantified by means of spatio-temporal parameters, peak trunk velocity and gait stability measures. In addition, clinical measures and steady state gait stability were assessed.

Ten older adults experienced a fall in the past 12 months and were classified as fallers. No statistically significant differences between fallers and non-fallers were found in the clinical or steady state gait measures, indicating that our fallers appeared relatively fit and healthy. All

perturbation types had a significant effect on all outcome measures for the group as a whole, but minimal differences were found between fallers and non-fallers. Any differences between the two groups seemed to manifest in a delay in peak trunk velocity measures for the fallers following the deceleration perturbation. We recommend further exploring the use of continuous trunk measures following deceleration perturbations to reveal subtle differences in older adults at risk of falling.

## 1. Introduction

Since most falls in older adults occur due to unsuccessful recovery from external perturbations such as trips, slips and sudden weight transfers (Robinovitch et al., 2013; Talbot et al., 2005), reactive gait assessment could be relevant to identify fallers. Successful recovery from perturbations is achieved by (regaining) control of the centre of mass (CoM) motion state (i.e. position and velocity) in relation to the base of support (BoS) (Hof et al., 2005). An important factor to achieve this is regulation of the trunk because of its mass and considerable height above the support surface and hence impact on the CoM. Additionally, proper foot placement is needed to establish the BoS. Over the last few decades, quantifying perturbation responses by means of spatio-temporal parameters and gait stability measures have been used in an attempt to understand the recovery of gait stability in older adults and demonstrate age-related decline in reactive gait ability (Bhatt et al., 2005; Crenshaw et al., 2012; Grabiner et al., 1993; Karamanidis et al., 2011; Lockhart et al., 2005a; McIntosh et al., 2017; Pijnappels et al., 2005b; Wang et al., 2017).

Most research on responses to gait perturbations in older adults induced either forward or backward balance loss by means of trips or forward slips, respectively. For example, it has been demonstrated that older adults show greater difficulty in arresting forward angular momentum (Pijnappels et al., 2005b), have larger forward trunk rotation (Pavol et al., 2001) and reduced gait stability (Karamanidis et al., 2011; Süptitz et al., 2013) after a trip perturbation compared to young adults. This, in turn, results in insufficient increment of the

BoS and hence longer recovery time (i.e. multiple stepping) or even unsuccessful recovery. Following slip perturbations, unsuccessful recovery was associated with higher horizontal heel velocity (Lockhart et al., 2005a), reduced backward stability (Bhatt et al., 2011a) and insufficient foot placement of the recovery step (Wang et al., 2017). While perturbations in the frontal plane have been used to examine the effect of ageing on reactive responses (McIntosh et al., 2017; Toebes et al., 2014, Roeles et al. submitted), studies discriminating fallers from non-fallers based on these responses are lacking. Ineffective recovery of lateral perturbations while standing, however, has been associated with future falls (Hilliard et al., 2008; Mille et al., 2013). According to McIntosh and colleagues (2017) ML perturbations were more difficult to recover from compared to AP perturbations. It may thus be worthwhile to explore whether responses to ML and AP perturbations can be used to distinguish older fallers and non-fallers, and if so which perturbation would discriminate best.

The abovementioned findings suggest that perturbations can be used to assess reactive gait stability in older adults. However, in these experimental studies, categorization of fallers and non-fallers was generally based on the ability to recover from the experimentally applied perturbations. First and foremost, this means that predictive validity related to real-life falls, either by retrospective or prospective analysis, is still lacking. Second, such an approach requires high intensity perturbations leading to falls during assessment. While this can be undertaken in a safe and controlled manner, it may limit the chances of clinical acceptance of such protocols, due to the psychological and physical stress on the individual. Finally, the ability to recover from experimentally induced perturbations may be direction depended and therefore not transferable to everyday life. To this end, we developed a perturbation protocol using submaximal perturbation intensities (i.e. intended to challenge the gait pattern without inducing a fall). We included both AP (unilateral belt acceleration and deceleration) and ML (ipsilateral and contralateral platform translation) perturbations to evaluate which perturbation type can best discriminate older adults with from those without a history of falls.

The main aim of our study was to evaluate whether we can discriminate between older adults with and without a history of falls based on spatio-temporal parameters, trunk kinematics and stability measures using a mixed perturbation protocol. We hypothesized that both fallers and non-fallers would be more affected by the contralateral sway and deceleration perturbations compared to the ipsilateral sway and acceleration perturbations, respectively. Therefore, we hypothesized that, fallers would have greater difficulty in recovering from these contralateral sway and deceleration perturbations compared to non-fallers. Based on previous work we expected that fallers were less capable in prolonging the swing phase to establish a proper base of support and regain stability resulting in shorter, smaller and quicker steps, and hence reduced backward and ML MoS. In addition, we hypothesized that fallers would be less able to arrest trunk movement following the perturbation and therefore expected a larger increase in trunk velocity in the direction of the perturbation compared to non-fallers.

## 2. Methods

### 2.1. Participants

Forty-nine older adults (≥65 years of age, 26 females, height: 1.71±0.08 m, weight: 74.9±11.6 kg) participated in this study. None of the participants had any self-reported neuromuscular deficits or cognitive impairments. The scientific and ethical review committee of the Faculty of Behavioural and Movement Sciences of the Vrije Universiteit Amsterdam approved the protocol (#2016-133). All participants gave informed consent prior to the measurement.

### 2.2. Equipment

Participants walked on the GRAIL (Gait Analysis Interactive Lab, Motekforce Link BV, Amsterdam, The Netherlands), which consists of an instrumented dual-belt treadmill with sway functionality, ten infra-red Vicon Bonita cameras (Vicon, Oxford, United Kingdom) operating at 100 Hz and a virtual reality environment projected on a semi-cylindrical screen. D-Flow software (version 3.28.0) was used to control all hardware components (Geijtenbeek et al., 2011). The Human Body Model (HBM version 2, Motek, Amsterdam, The Netherlands), a biomechanical model, based on 29 markers applied to the trunk and lower limbs was used to measure kinematics (van den Bogert et al., 2013). Participants wore a safety harness to arrest potential falls.

### 2.3. Protocol

Participants' daily life concern of falling and physical activity level were assessed using the modified Dutch version of the Falls Efficacy Scale (FES) (Bosscher et al., 2005) and the Physical Activity Questionnaire (PAQ) (Innerd et al., 2015), respectively. Participants were asked whether they had any falls in the last 12 months. A fall was defined as 'any unanticipated event that results in a participant coming to the ground, floor or lower level' (Gibson, 1987). Participants were classified as faller when reporting at least one fall over the past 12 months. Physical activity and balance performance were assessed by the Timed Up & Go test (TUG) (Podsiadlo and Richardson, 1991) and the One-Legged Stance Test (OLST) (Borowicz et al., 2016) on the preferred leg with a maximum score of 30 seconds.

Next, the steady state gait and perturbation assessments were conducted. The default treadmill speed was set at 1 m/s, except when participants indicated that this speed was too low or high, in which case the speed was gradually adjusted to a comfortable speed (but not higher than 1.3 m/s or lower than 0.6 m/s). After three minutes of familiarization, a two-minute steady state gait trial was recorded. Subsequently, the perturbation protocol was applied, including four perturbation types: 1) ipsilateral sway perturbations, consisting of platform translation (displacement: 0.05 m; velocity: 0.30 m/s; acceleration: 3.26 m/s$^2$) to the right at right initial contact or to the left at left initial contact 2) contralateral sway perturbations. i.e. platform translations to the right at left initial contact or to the left at right initial contact; 3) unilateral belt acceleration perturbations (max velocity: 1.55-2.43 m/s, max acceleration: 8.07-11.07 m/s$^2$) and 4) unilateral belt deceleration perturbations (min velocity: 0.09-0.18 m/s; max deceleration: 6.85-10.8 m/s$^2$). Perturbations were triggered at initial contact (Zeni et al., 2008) and applied with a 10-15 stride interval. A randomized block design was used to allow for breaks in the protocol if required. One block contained eight perturbations (four perturbation types applied to both sides) and was repeated five times totalling in 40 perturbations (ten repetition per perturbation type). The total perturbation protocol lasted approximately 15 minutes.

### 2.4. Data analyses

All data were analysed using custom-written Matlab scripts (version 2015a; The Mathworks, Natick, MA, USA). Kinematic data were filtered using a bidirectional 6 Hz second-order Butterworth filter. Gait events were detected based on an AP foot velocity-based treadmill algorithm (Zeni et al., 2008).

Three spatio-temporal parameters were calculated: step time, step length and step width, defined as the time elapsed, AP and ML distance between two consecutive foot contacts, respectively.

Linear trunk velocity in the ML, VT and AP direction were calculated by taking the derivative of the trunk position (i.e. average of the C7, T10, xiphoid process and jugular notch marker positions). Angular trunk velocities in the frontal, transversal and sagittal plane were estimated by taking the derivative of trunk angles as calculated by HBM (De Leva, 1996). Subsequently, peak linear and angular trunk velocities were determined for each step.

Gait stability was quantified as the margins of stability (MoS), as determined by the distance between the border of the BoS and the extrapolated centre of mass (XCoM) (Hof et al., 2005). The XCoM was estimated by the CoM position plus its velocity divided by $\sqrt{g/l}$ in which $g$ is the acceleration of gravity and $l$ the average pelvic height (based on left and right anterior and posterior superior iliac spine markers) during stance. The ML lateral malleolus marker position of the leading foot quantified the ML border of the BoS whereas the AP heel marker position was used to define the AP border (Figure 5.7). Thereby, negative ML and AP MoS values indicated *instability* in the lateral and backward direction, respectively.

We first analysed whether steady state gait differed between fallers and non-fallers by averaging spatio-temporal parameters, peak trunk velocities and MoS over 75 left and 75 right steps (per participant). Moreover, local dynamic stability (LDS) of trunk velocity was calculated as described by Bruijn et al. (2009). Briefly, 75 strides were time-normalized to 7500 samples (i.e. approximately 100 samples per stride) from which 5D state spaces were reconstructed with a time delay of 10 samples. Euclidean distances between initially neighbouring trajectories were calculated and averaged over time. The slope of average logarithmic rate of divergence for sample 0-50 was used to estimate LDS. Lower LDS values imply more stable gait.

All 11 measures (three spatio-temporal parameters, six peak trunk velocities and two minimum MoS values) were calculated for six steps after each perturbation. Averages per perturbation type, excluding the initial perturbation, were used for further analyses.

### 2.5. Statistical analyses

All statistical analyses were performed using SPSS version 23 (SPSS Inc, Chicago, IL, US). We first tested demographics, clinical measures (FES, PAQ, TUG, OLST), gait speed and steady state gait parameters for normality using a Shapiro-Wilk test. In case of normal distribution independent *t*-tests were used to assess differences between fallers and non-fallers. When not normally distributed, Mann-Whitney U-tests were used.

To test our main aim, that is, whether fallers and non-fallers recovered differently from the perturbations, mixed model analyses of variance (ANOVAs) were used with the average steady state step and six steps following the perturbation onset included as within factor (Steps) and fallers and non-fallers (Group) as between factor. A Greenhouse-Geisser correction was used when the assumption of sphericity was violated. In case of significant main effects for

Steps, post-hoc Bonferroni corrected simple contrasts were used to evaluate which perturbed steps differed from the average steady state step. When significant Steps x Group interaction effects were found, repeated contrasts for the subsequent recovery steps were used to examine if fallers recovered differently from one step to the next compared to non-fallers.

## 3. Results

Ten participants (20%) were classified as fallers based on their self-reported fall history in the past 12 months. One faller withdrew after 25 (out of 40) perturbations due to anxiety about falling. These perturbations (excluding the first one) were included in the analyses. Based on visual inspection, none of the participants appeared to have fallen into the safety harness.

### 3.1. Demographics, clinical measures and steady state gait

Demographics and clinical outcome measures did not significantly differ between fallers and non-fallers, except for FES scores (Table 6.2). Fallers scored higher on the FES, indicating that they had more concern of falling compared to the non-fallers, although the values were still at the lower boundary for both groups. On average fallers walked at 1.00±0.22 m/s and non-fallers at 1.01±0.12 m/s, which was not significantly different ($p$=0.874). In addition, no differences in steady state gait parameters between fallers and non-fallers were found (Table A-V.1).

### 3.2. Perturbed gait

The ANOVAs revealed significant main effects of Steps for all 11 outcome parameters following all four perturbation types ($p$<0.05), except for peak trunk velocity in the frontal plane following the acceleration perturbation ($p$=0.236), indicating that the perturbations significantly affected the gait pattern (Table 6.3, Table 6.4 and Figure 6.3-Figure 6.6). No main effects of Group were found for any of the outcome parameters. However, significant Steps x Group interaction effects were found for backward MoS following the contralateral sway perturbation and peak trunk velocity in the VT and AP direction following the deceleration perturbation.

**Table 6.2** Mean±SD demographics and clinical measures for (non-)fallers. Significant differences between fallers and non-fallers as evaluated by Mann-Whitney U-tests are printed in bold.

| | Non-fallers (n=39) | | | Fallers (n=10) | | | p |
|---|---|---|---|---|---|---|---|
| Gender [male/female] | 19/20 | | | 4/6 | | | 0.687 |
| Age [years] | 71.5 | ± | 5.6 | 72.8 | ± | 5.8 | 0.599 |
| Height [m] | 1.72 | ± | 0.09 | 1.68 | ± | 0.05 | 0.297 |
| Weight [kg] | 75.3 | ± | 12.5 | 72.5 | ± | 7.4 | 0.455 |
| FES [score] | 11.2 | ± | 1.9 | 12.0 | ± | 1.3 | **0.037** |
| PAQ [score] | 13.6 | ± | 3.7 | 14.5 | ± | 3.8 | 0.345 |
| OLST [s] | 20.7 | ± | 11.3 | 14.4 | ± | 8.7 | 0.100 |
| TUG [s] | 8.1 | ± | 1.1 | 8.9 | ± | 1.9 | 0.320 |

FES: falls efficacy scale (range score: 10-40); PAQ: physical activity questionnaire (range score: 0-18); OLST: one-legged stance test (max score: 30); TUG: Timed Up and Go test.

To evaluate whether fallers and non-fallers recovered differently from one step to the next, post-hoc repeated contrasts were used to further explore the significant interaction effects (Table A-V.2 and Table A-V.3). In response to the contralateral sway perturbation, fallers showed a larger increase in backward MoS from the first to the second step following the perturbation (F=4.857, $p$=0.032). A number of differences in VT and AP trunk velocity were found in response to the deceleration perturbation. During the perturbation step, VT trunk velocity increased in the fallers group whereas a slight decrease in the non-fallers group was found as indicated by the significant difference between the average steady state step and the perturbed step (F=5.803, $p$=0.020). Subsequently (from the first to second step), AP trunk velocity reduced in the fallers group while the non-fallers showed an increase in trunk velocity, albeit with large variability in both groups (F=8.565, $p$=0.005). Next (from the second to third step), fallers reduced VT trunk velocity whereas non-fallers showed an increase in VT trunk velocity (F=7.860, $p$=0.007). Following (from the third to the fourth step), larger increase in

VT trunk velocity was found for the fallers compared to the non-fallers (F=5.894, *p*=0.019). Finally (from the fourth to fifth step), AP trunk velocity was increased in the fallers whereas the non-fallers showed a decrease in velocity (F=13.068, *p*=0.001).

**Table 6.3** Mixed model ANOVA outcomes for the sway perturbations (within factor: average steady state step and six post-perturbation steps; between factor: group). Significant effects are printed in bold.

| | | Ipsilateral sway | | | Contralateral sway | | |
|---|---|---|---|---|---|---|---|
| | | df | F | *p* | df | F | *p* |
| Step Time | Steps | **1.879** | **61.248** | **<0.001** | **2.080** | **25.064** | **<0.001** |
| | Group | 1 | 0.056 | 0.813 | 1 | 0.060 | 0.808 |
| | Steps*Group | 1.879 | 1.634 | 0.202 | 2.080 | 2.921 | 0.056 |
| Step Length | Steps | **2.311** | **68.936** | **<0.001** | **2.094** | **13.554** | **<0.001** |
| | Group | 1 | 0.231 | 0.633 | 1 | 0.568 | 0.455 |
| | Steps*Group | 2.311 | 1.447 | 0.238 | 2.094 | 1.354 | 0.263 |
| Step Width | Steps | **1.947** | **35.621** | **<0.001** | **2.065** | **69.436** | **<0.001** |
| | Group | 1 | 0.374 | 0.544 | 1 | 0.177 | 0.676 |
| | Steps*Group | 1.947 | 1.182 | 0.310 | 2.065 | 0.608 | 0.551 |
| MoS ML | Steps | **2.264** | **28.380** | **<0.001** | **2.209** | **185.329** | **<0.001** |
| | Group | 1 | 0.306 | 0.583 | 1 | 1.258 | 0.268 |
| | Steps*Group | 2.264 | 1.747 | 0.174 | 2.209 | 1.031 | 0.366 |
| MoS BW | Steps | **2.418** | **56.265** | **<0.001** | **3.109** | **41.693** | **<0.001** |
| | Group | 1 | 0.862 | 0.358 | 1 | 0.015 | 0.904 |
| | Steps*Group | 2.418 | 0.494 | 0.646 | **3.109** | **2.922** | **0.034** |
| Trunk Vel ML | Steps | **2.480** | **28.743** | **<0.001** | **2.772** | **102.948** | **<0.001** |
| | Group | 1.000 | 0.003 | 0.957 | 1.000 | 0.039 | 0.844 |
| | Steps*Group | 2.480 | 0.832 | 0.459 | 2.772 | 0.652 | 0.571 |
| Trunk Vel VT | Steps | **3.327** | **5.174** | **0.001** | **3.512** | **16.598** | **<0.001** |
| | Group | 1 | 0.070 | 0.792 | 1 | 0.085 | 0.772 |
| | Steps*Group | 3.327 | 2.827 | 0.035 | 3.512 | 1.230 | 0.300 |
| Trunk Vel AP | Steps | **2.541** | **5.770** | **0.001** | **2.704** | **11.131** | **<0.001** |
| | Group | 1 | 0.704 | 0.406 | 1 | 0.670 | 0.417 |
| | Steps*Group | 2.541 | 2.461 | 0.075 | 2.704 | 1.787 | 0.158 |
| Trunk Vel F | Steps | **3.461** | **4.890** | **0.001** | **2.972** | **7.609** | **<0.001** |
| | Group | 1 | 0.298 | 0.588 | 1 | 0.054 | 0.817 |
| | Steps*Group | 3.461 | 0.407 | 0.775 | 2.972 | 0.046 | 0.986 |
| Trunk Vel T | Steps | **3.467** | **4.785** | **0.001** | **1.566** | **60.707** | **<0.001** |
| | Group | 1 | 1.893 | 0.175 | 1 | 0.758 | 0.388 |
| | Steps*Group | 3.467 | 1.259 | 0.289 | 1.566 | 0.508 | 0.559 |
| Trunk Vel S | Steps | **2.594** | **13.656** | **<0.001** | **2.670** | **30.415** | **<0.001** |
| | Group | 1 | 1.727 | 0.195 | 1 | 0.038 | 0.847 |
| | Steps*Group | 2.594 | 2.347 | 0.085 | 2.670 | 2.602 | 0.061 |

**Table 6.4** Mixed model ANOVA outcomes for the belt perturbations (within factor: average steady state step and six post-perturbation steps; between factor: group). Significant effects are printed in bold.

| | | Acceleration | | | Deceleration | | |
|---|---|---|---|---|---|---|---|
| | | df | F | p | df | F | p |
| Step Time | Steps | **2.257** | **82.554** | **<0.001** | **2.336** | **57.698** | **<0.001** |
| | Group | 1 | 0.131 | 0.719 | 1 | 0.253 | 0.617 |
| | Steps*Group | 2.257 | 1.042 | 0.362 | 2.336 | 1.599 | 0.202 |
| Step Length | Steps | **1.913** | **16.019** | **<0.001** | **2.476** | **163.396** | **<0.001** |
| | Group | 1 | 0.221 | 0.641 | 1 | 0.922 | 0.342 |
| | Steps*Group | 1.913 | 0.564 | 0.563 | 2.476 | 1.860 | 0.150 |
| Step Width | Steps | **3.018** | **39.978** | **<0.001** | **3.508** | **33.725** | **<0.001** |
| | Group | 1 | 0.437 | 0.512 | 1 | 1.025 | 0.316 |
| | Steps*Group | 3.018 | 1.191 | 0.315 | 3.508 | 0.324 | 0.838 |
| MoS ML | Steps | **2.943** | **8.943** | **<0.001** | **3.712** | **30.022** | **<0.001** |
| | Group | 1 | 1.328 | 0.255 | 1 | 0.536 | 0.468 |
| | Steps*Group | 2.943 | 1.117 | 0.343 | 3.712 | 2.346 | 0.061 |
| MoS BW | Steps | **2.273** | **32.944** | **<0.001** | **3.168** | **353.197** | **<0.001** |
| | Group | 1 | 0.589 | 0.446 | 1 | 0.133 | 0.717 |
| | Steps*Group | 2.273 | 0.809 | 0.461 | 3.168 | 1.652 | 0.177 |
| Trunk Vel ML | Steps | **3.022** | **33.593** | **<0.001** | **2.734** | **43.303** | **<0.001** |
| | Group | 1.000 | 0.013 | 0.909 | 1.000 | 0.042 | 0.839 |
| | Steps*Group | 3.022 | 0.521 | 0.669 | 2.734 | 1.134 | 0.335 |
| Trunk Vel VT | Steps | **2.419** | **12.611** | **<0.001** | **2.686** | **7.896** | **<0.001** |
| | Group | 1 | 0.008 | 0.929 | 1 | 0.002 | 0.967 |
| | Steps*Group | 2.419 | 0.758 | 0.494 | **2.686** | **5.115** | **0.003** |
| Trunk Vel AP | Steps | **3.197** | **47.183** | **<0.001** | **2.489** | **3.323** | **0.029** |
| | Group | 1 | 0.705 | 0.405 | 1 | 0.050 | 0.823 |
| | Steps*Group | 3.197 | 0.147 | 0.940 | **2.489** | **6.137** | **0.001** |
| Trunk Vel F | Steps | 2.690 | 1.440 | 0.236 | **3.296** | **23.154** | **<0.001** |
| | Group | 1 | 0.066 | 0.798 | 1 | 0.082 | 0.775 |
| | Steps*Group | 2.690 | 0.776 | 0.496 | 3.296 | 0.460 | 0.728 |
| Trunk Vel T | Steps | **2.005** | **48.128** | **<0.001** | **3.694** | **3.067** | **0.020** |
| | Group | 1 | 0.628 | 0.432 | 1 | 2.101 | 0.154 |
| | Steps*Group | 2.005 | 0.308 | 0.736 | 3.694 | 0.862 | 0.480 |
| Trunk Vel S | Steps | **2.457** | **41.405** | **<0.001** | **1.673** | **35.426** | **<0.001** |
| | Group | 1 | 0.098 | 0.755 | 1 | 0.091 | 0.764 |
| | Steps*Group | 2.457 | 0.671 | 0.542 | 1.673 | 1.084 | 0.333 |

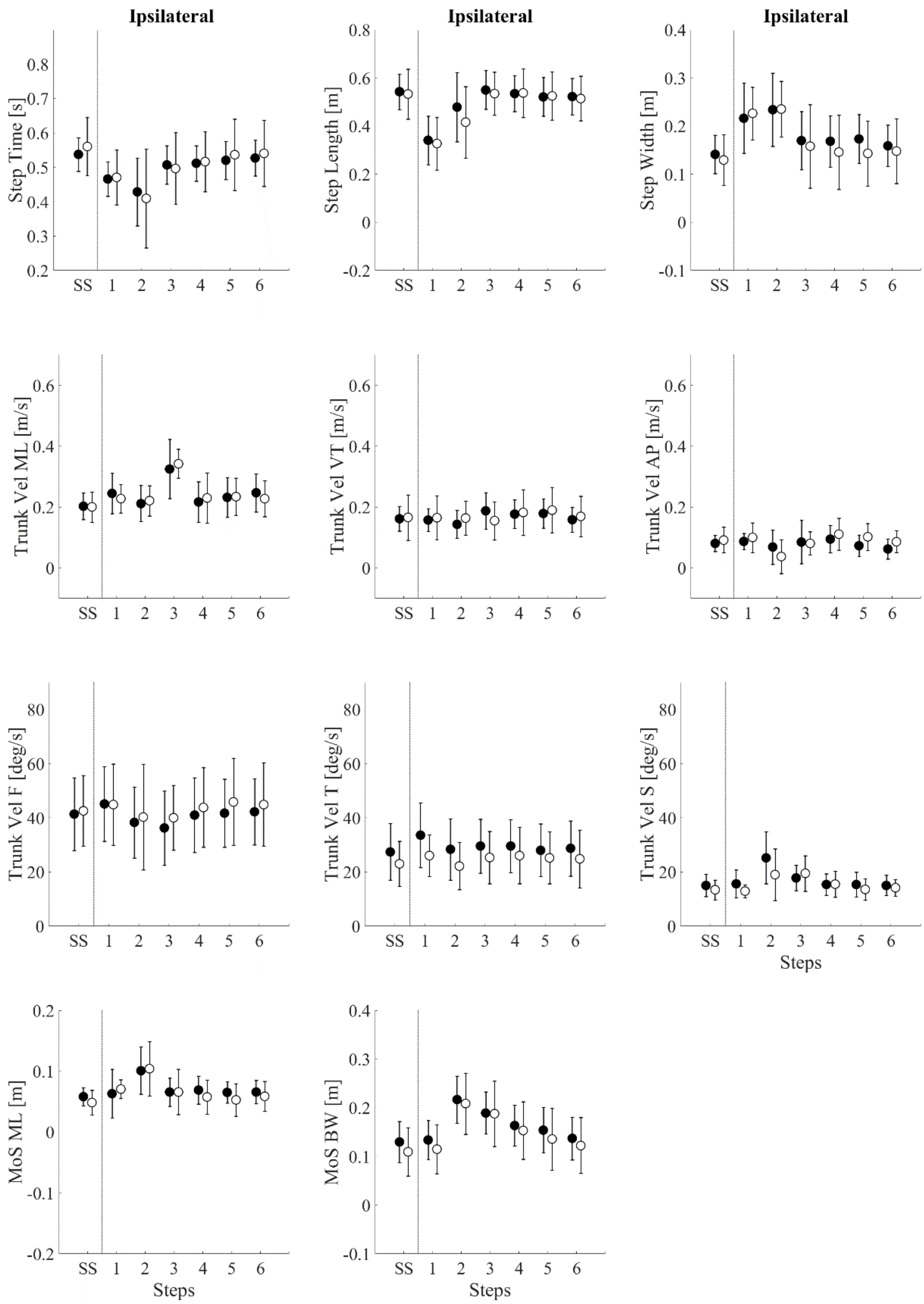



**Figure 6.3** Responses to ipsilateral sway perturbations: Mean values (± standard deviations) of spatio-temporal parameters, linear trunk velocity in the medio-lateral (ML), vertical (VT) and anterior-posterior (AP) direction, angular trunk velocity in the frontal (F), transversal (T) and sagittal (S) planes, and margins of stability (MoS) in the ML and backward (BW) directions after the ipsilateral sway perturbation. The average values for steady state (SS) and six steps after the perturbation onset are shown for non-fallers (black dots) and fallers (white dots).

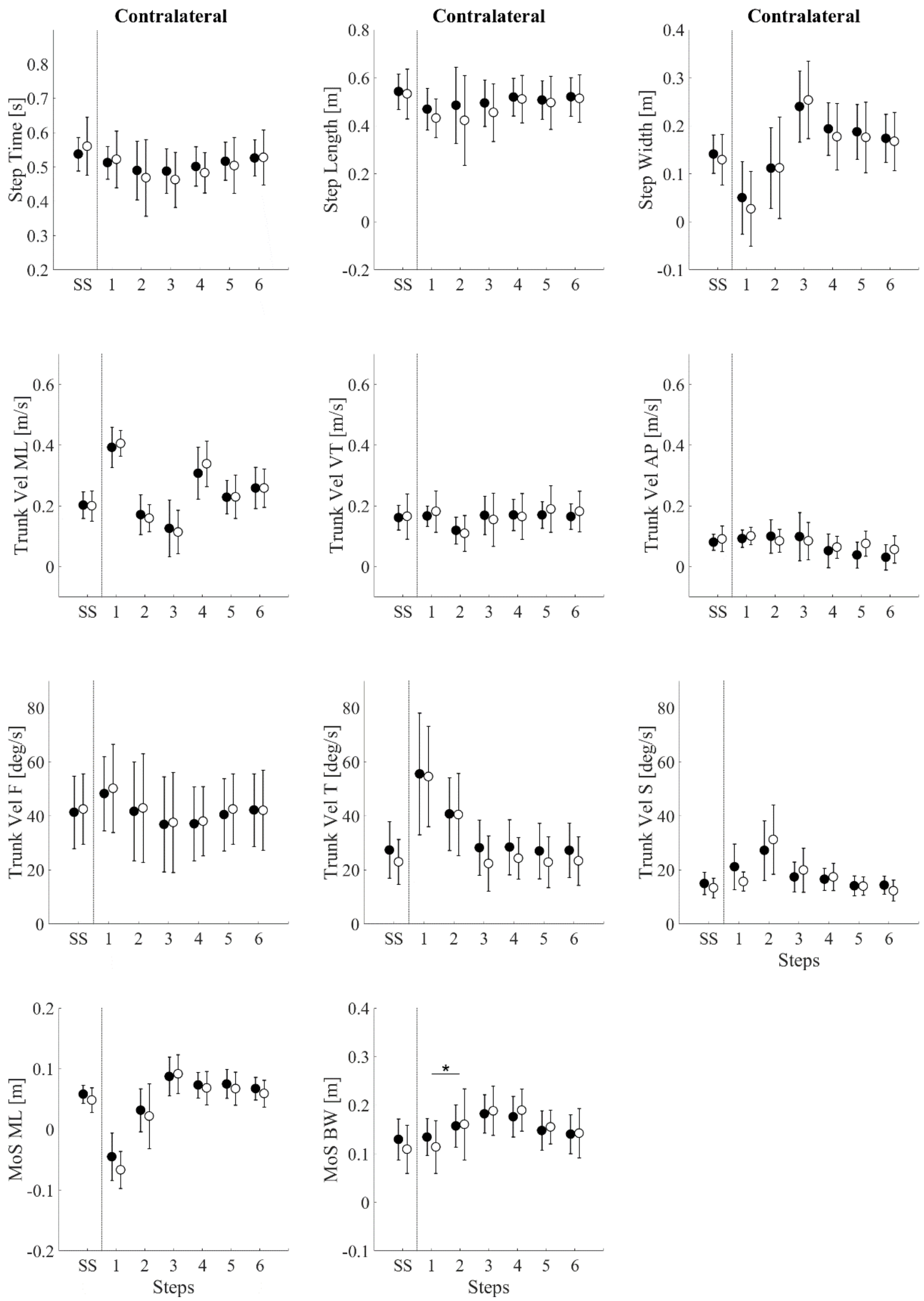



**Figure 6.4** Responses to contralateral sway perturbations: Mean values (± standard deviations) of spatio-temporal parameters, linear trunk velocity in the medio-lateral (ML), vertical (VT) and anterior-posterior (AP) direction, angular trunk velocity in the frontal (F), transversal (T) and sagittal (S) planes, and margins of stability (MoS) in the ML and backward (BW) directions after the contralateral sway perturbation. The average values for steady state (SS) and six steps after the perturbation onset are shown for non-fallers (black dots) and fallers (white dots).

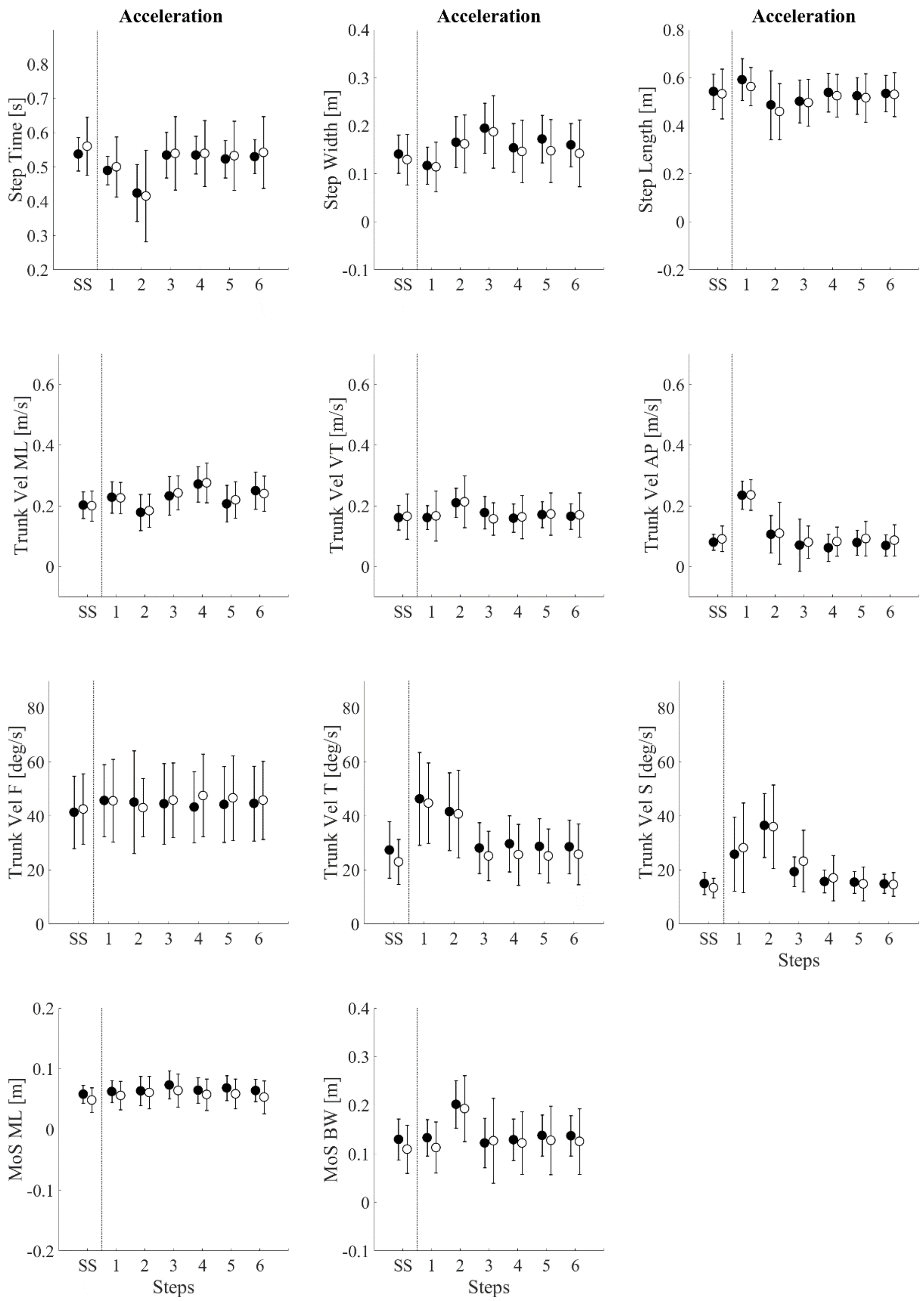



**Figure 6.5** Responses to acceleration perturbations: Mean values (± standard deviations) of spatio-temporal parameters, linear trunk velocity in the medio-lateral (ML), vertical (VT) and anterior-posterior (AP) direction, angular trunk velocity in the frontal (F), transversal (T) and sagittal (S) planes, and margins of stability (MoS) in the ML and backward (BW) directions after the acceleration perturbation. The average values for steady state (SS) and six steps after the perturbation onset are shown for non-fallers (black dots) and fallers (white dots).

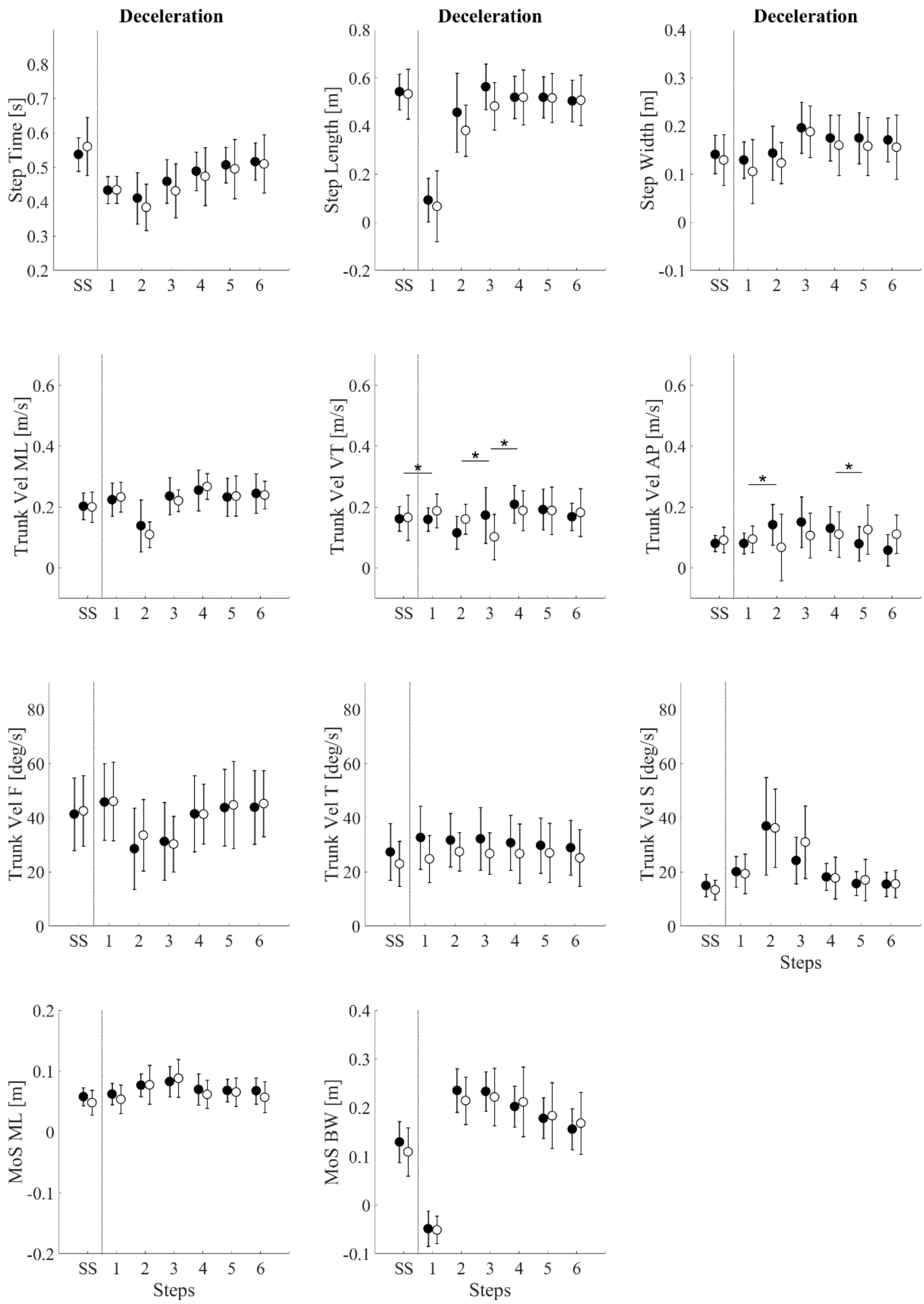



**Figure 6.6** Responses to deceleration perturbations: Mean values (± standard deviations) of spatio-temporal parameters, linear trunk velocity in the medio-lateral (ML), vertical (VT) and anterior-posterior (AP) direction, angular trunk velocity in the frontal (F), transversal (T) and sagittal (S) planes, and margins of stability (MoS) in the ML and backward (BW) directions after the deceleration perturbation. The average values for steady state (SS) and six steps after the perturbation onset are shown for non-fallers (black dots) and fallers (white dots).

## 4. Discussion

Our main aim was to evaluate whether older adults with and without a history of falls showed differences in their recovery response to a mixed protocol of submaximal ML and AP gait perturbations. We demonstrated that all perturbation types included in our perturbation protocol significantly affected the gait pattern of both fallers and non-fallers in terms of spatio-temporal parameters, peak trunk velocity and gait stability measures. However, in contrast to our hypothesis, these responses did not discriminate between those with and without a history of falls when taken collectively across six recovery steps.

One explanation for the absence of these gross differences between groups in overall performance may have been the nature of our population. We aimed to recruit a representative population sample and therefore expected that 30% of our older adults would have experienced at least one fall in the past year (Rubenstein and Josephson, 2006). However, our group of fallers was considerably smaller in size (20%) and also fit and healthy as indicated by the clinical measures and steady state gait parameters. Even though the fallers scored higher on the FES compared to non-fallers (NF: 11.2±1.9, F: 12.0±1.3), these differences were small and scores for the fallers were still low given the score ranged from 10 to 40. Moreover, although local dynamic stability has been associated with falls in older adults (e.g. Lockhart and Liu, 2008; Rispens et al., 2015; Toebes et al., 2012), steady state gait measures did not differ between our groups. It can therefore be questioned whether our older adults with a fall history can be considered actual fallers.

Another possible explanation is that opposing recovery responses masked step-by-step differences between fallers and non-fallers which may have been reflected in the Group x Steps interaction effects following the contralateral sway (in backward MoS) and deceleration (in VT and AP trunk velocity) perturbations. The step-by-step analyses following the contralateral sway perturbation revealed that backward stability in the fallers group was increased during the second recovery step. The fact that a ML (contralateral) perturbation resulted in significant effects in AP stability may seem somewhat counterintuitive. However, a recent publication also demonstrated larger reductions in forward stability (and thus increased backward stability) following ipsilateral perturbations in a group of fall prone stroke patients compared to non-fall prone patients (Punt et al., 2017). Hence in both studies the perturbation direction was perpendicular to the saving reaction. It would appear therefore that a ML balance perturbation is associated with AP changes in foot position in the steps following the perturbation. As such evaluating the perturbation response using spatio-temporal parameters and discrete parameters alone are not sufficient.

Following the deceleration perturbation, a number of Step x Group interaction effects were found in VT and AP trunk velocity. A belt deceleration perturbation typically interrupts forward progression which needs to be compensated for during the recovery steps, especially on a treadmill when speed must be continued (Ilmane et al., 2015; Kagawa et al., 2011). Although treadmill gait imposes a time constraint, similar stepping responses and forward trunk motion have been reported following overground trips (Grabiner et al., 2008). In the present study, the significant Step x Group interaction effects for VT trunk velocity were likely caused by this forward trunk motion. Somewhat unexpected however, these interaction effects did not show up significantly different for peak angular trunk velocity in the sagittal plane. Grabiner and colleagues (2008) have shown increased reaction and recovery time following a trip resulted in increased trunk flexion angle and velocity, which was associated with (age-related) lower muscle strength and power. Moreover, trunk flexion angle and velocity had greater accuracy in classifying falls and recoveries after an AP postural perturbation compared to margins of stability and time to stability measures (Crenshaw et al., 2012). However, our analysis was based on a discrete step-by-step approach but methods using continuous evaluation of trunk velocity may be more readily able to discriminate between fallers and non-fallers. Nonetheless, we did find significant Step x Group interaction effects, with a rather small number of mildly affected fallers and hence reactive gait assessment may indeed be capable of identifying early on those at risk for falls. Additionally, it should be noted that none of the clinical outcome measures used were able to detect differences between fallers and non-fallers, except for the FES but these differences did not seem to be clinically relevant.

## 5. Conclusion

To conclude, our mixed perturbation protocol significantly challenged the gait pattern of healthy older adults with and without a history of falls, and the relatively small group that was classified as fallers showed some differences compared to non-fallers in responses to gait perturbations, particularly in a delay of trunk velocity over steps after treadmill decelerations. In order to reveal subtle differences in reactive gait ability of older adults to diagnose those at risk of falling early, we would encourage others to apply contralateral sway and deceleration perturbations to older adults and consider to analyse the recovery response using continuous rather than discrete measures

# CHAPTER 7

# Reactive gait assessment development

# Phase 3

## 7.1 Development

The previous chapter described the recovery from ipsilateral and contralateral sway, acceleration and deceleration perturbations in older adults with and without a history of falls. We found only subtle differences between fallers and non-fallers. No differences were found in clinical measures or steady state gait. For perturbed gait, significant Step x Group interaction effects were found in backward MoS values following the contralateral sway perturbation and in VT and AP peak trunk velocity values following the deceleration perturbation.

While these findings might be meaningful, the lack of more pronounced differences between those with and those without a history of falls would lead to poor discriminatory ability and some uncertainty in the classification of participants as fallers or non-fallers. There are two likely explanations for this uncertainty: 1) the selected perturbation types are not suitable to assess reactive gait stability and/or 2) the selected outcome measures (i.e. spatio-temporal parameters, margins of stability and peak trunk velocity) may not be sensitive enough to differentiate between fallers and non-fallers. The former appears unlikely as we previously showed that all perturbation types significantly affected the gait pattern (Chapter 6.1). The latter, however, requires more attention, especially in a fit cohort like ours, where between-group and even more so between-individual differences could be subtle. To better understand our findings, we chose to apply a set of less common, more complex, but potentially more sensitive outcome measures. The analyses were performed on the data collected in Chapter 6.1.

## 7.2 Evaluation

In this section, three studies are presented based on the data collected in Chapter 6. In the first analysis study (section 7.2.1), we evaluated whether we can discriminate between older adults with and without a history of falls using a continuous measure of trunk motion to quantify the perturbation response. Next, we evaluated the construct validity of our reactive gait assessment by examining to what extent proactive adaptations in the gait pattern affected the perturbation response (section 7.2.2). Finally, we performed a descriptive study to analyse the effect on an individual level (section 7.2.3).

### 7.2.1 Analysis study 1 – Responses to different gait perturbations using continuous measures of trunk motion to discriminate between older adults with and without a history of falls

#### 7.2.1.1 Introduction

Bruijn and colleagues (2010) introduced a comprehensive method to evaluate continuous trunk motion following a waist-pull perturbation. In short, deviation in the perturbed gait pattern is compared to the average steady state gait pattern. This is done using time-normalized linear (in three directions) and angular (in three planes) trunk velocities from which a number of parameters can be obtained. Bruijn and colleagues discriminate between two phases: the initial phase, in which the maximum deviation from the steady state gait pattern and its duration are quantified, and the recovery phase, in which the rate of return to steady state values is determined. We hypothesize that evaluation of continuous trunk velocities in six dimensions rather than our previously selected peak trunk velocity measure in a single dimension may enhance sensitivity of the outcome measure. Furthermore, by capturing overall deviation in the gait pattern, this trunk motion measure may be less sensitive to different recover strategies and therefore the trunk motion measure of Bruijn and colleagues may be more revealing.

Theoretically, any parameter and any task can be evaluated using Bruijn's method. For instance, the effect of lifting boxes during ML and tilt perturbations (Mavor and Graham, 2015) or muscle fatigue on the responses to ML waist-pulls (Toebes et al., 2014) on trunk motion, or the effect of attentional focus (de Melker Worms et al., 2017) and ageing (Krasovsky et al., 2012) on recovery of CoM motion following a trip have been investigated. The latter study is most comparable to our work and showed that older adults were affected for longer by the perturbation and showed slower recovery of CoM motion. We choose to evaluate trunk motion, given our previous findings of Step x Group interaction effects in peak trunk velocity between older adults with and without a history of falls.

The main aim of this analysis study was to investigate whether we could discriminate between fallers and non-fallers by means of the initial perturbation effect, the duration of the initial response and the rate of recovery from the perturbation using the continuous measure of trunk motion. Since our previous study revealed Step x Group interaction effects for peak trunk velocity over multiple recovery steps following deceleration perturbations, we hypothesized to find differences between fallers and non-fallers following deceleration perturbations. More specifically, we expected that older adults with a history of falls would be more affected by the perturbation and that the initial response would be longer compared to non-fallers. Furthermore, we expected slower recovery in the fallers compared to the non-fallers.

#### 7.2.1.2 Methods

A detailed description of the recruited participants, the equipment and the protocol has been described previously (section 6.2.1). Briefly, 39 elderly non-fallers and ten elderly fallers participated in this study. Participants walked on the GRAIL (Gait Analysis Interactive Lab, Motekforce Link BV, Amsterdam, The Netherlands) while lower limb and trunk kinematics were recorded. A two-minute steady state gait trial was completed followed by four types of perturbations; ipsilateral and contralateral sway, acceleration and deceleration perturbations. Perturbation types were repeated ten times in a randomized order.

**Data analyses**

All data were analysed using custom-written Matlab scripts (version 2015a; The Mathworks, Natick, MA, USA). Marker position data were filtered using a bidirectional 6 Hz second-order Butterworth filter. Gait events were detected based on an AP foot velocity-based treadmill algorithm (Zeni et al., 2008).

We quantified by how much perturbed trunk motion was outside an individual's own steady state range. In order to do this, trunk velocity was calculated in six dimensions: the ML, VT and AP direction and the frontal, transversal and sagittal plane. Time-series were normalized to 101 samples per stride for the steady state and perturbed gait trials. For each of the six velocity time series, averages over 100 strides and variability values for each percentage of the gait cycle were calculated for steady state gait and perturbed gait (Figure 7.1).

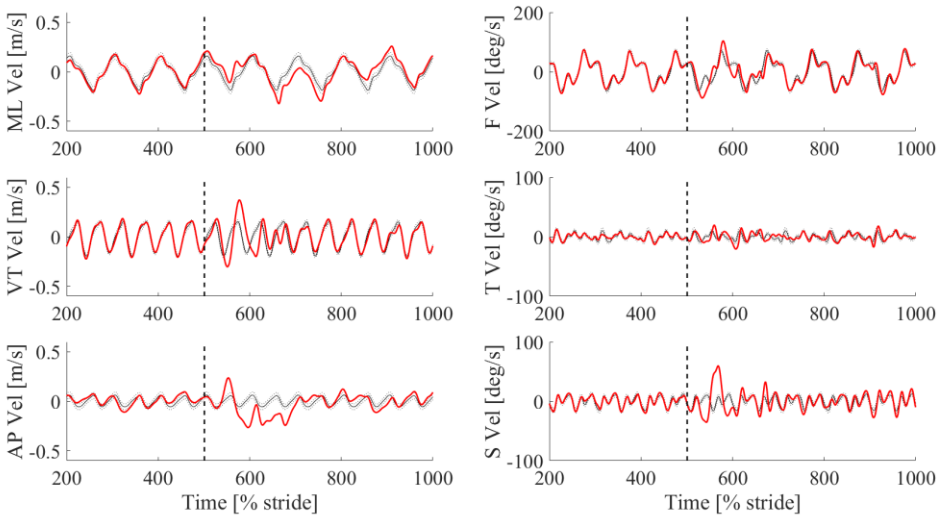



**Figure 7.1** A typical example of normalized trunk velocity time series (red lines) before and after a deceleration perturbation in the medio-lateral (ML), vertical (VT) and anterior-posterior (AP) direction, and the frontal (F), transversal (T) and sagittal (S) plane. The black lines represent the average steady state gait cycle (solid line) and its variability (dotted line). The vertical dashed line represents the perturbation onset.

Thereafter, the deviation from steady state gait during perturbed gait was calculated for each of the six dimensions, corrected for variability in the steady state gait pattern and combined to a comprehensive trunk motion measure as follows (Bruijn et al., 2010):

$$D(kx100+i)_{\substack{k=0:n-1\\ i=1:100}} = \sqrt{\sum_{d=1}^{6}((PG(i)_d - SSG(kx100+i)_d)/SSG_{var}(i)_d)^2} \qquad \textbf{Equation 7.1}$$

where $D(kx100 + i)$ is the normalized distance for $i$% of stride k+1, with $n$ being the number of strides in the perturbed walking trial, $d$ is the dimension (i.e. ML, VT, AP, F, S or T velocity), $SSG$ is the average steady state gait cycle for $d$, $PG$ is the perturbed gait trial for $d$ and $SSG_{var}$ is the variability of in steady state gait cycle for $d$. To clarify this, an example of the deviation in trunk velocity for one dimension (i.e. AP velocity) over one stride is shown in Figure 7.2. Note that larger differences between perturbed (upper panel – red line) and steady state (upper panel – black line) trunk velocity result in larger deviation values (lower panel), regardless of an increase or decrease in trunk velocity. This was applied to all six dimensions for all strides in the perturbed gait trial resulted in a set of time-series for $D$ as illustrated in Figure 7.3.

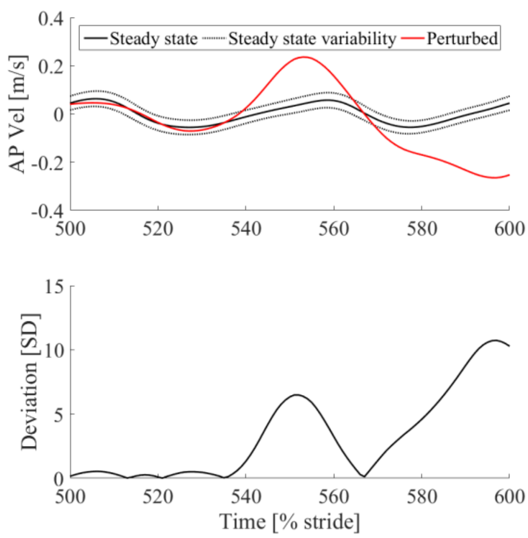



**Figure 7.2** Example of deviation in perturbed trunk velocity from steady state trunk velocity in the anterior-posterior direction (AP). Upper panel: AP trunk velocity for one perturbed cycle (red line), the average steady state cycle (black line) and 1SD (dotted black line). Lower panel: deviation in AP perturbed trunk velocity from steady state. See text for details on calculation.

Three parameters were extracted from $D$. First, to quantify the initial perturbation effect ($B_{pre}$), the maximum value of $D$ in the first stride (i.e. 100 samples) following the perturbation onset was determined. This value was expressed relative to pre-perturbed trunk motion ($C_{pre}$), to allow for comparison between individuals. $C_{pre}$ was calculated as the average value of $D$

over three strides (i.e. 300 samples) prior to the first perturbation (Figure 7.3). Next, the duration of the initial perturbation effect ($\tau$) was detected from the perturbation onset to $B_{pre}$. Finally, we evaluated the perturbation recovery ($\beta$) by quantifying the rate of return to steady state gait using the following equation:

$$D(i) = A + (B - A) * e^{(-\beta(i-\tau))}$$ **Equation 7.2**

in which $A$ refers to the average value of $D$ over the fourth stride (i.e. $i$=300 to $i$=400) following the perturbation. A least squares fitting technique was used to calculate $\beta$. Higher values of $\beta$ indicate faster recovery. In line with previous analyses, outcome measures ($B_{pre}$, $\tau$ and $\beta$) were calculated for each perturbation and averaged per perturbation type excluding the first perturbation.

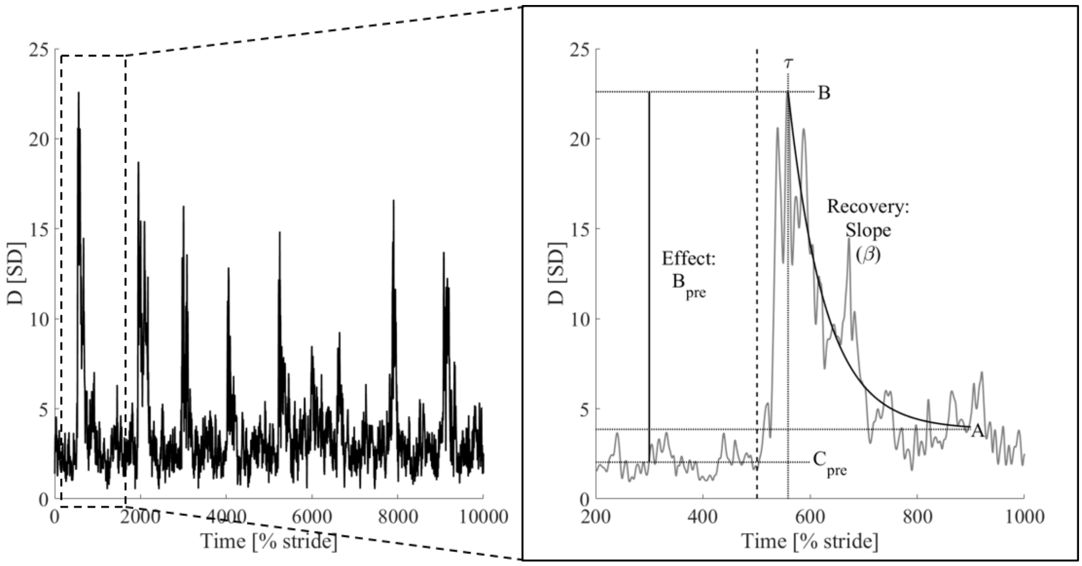



**Figure 7.3** Trunk motion during perturbed gait expressed as deviation from steady state gait in linear and angular velocity. Left panel: a typical example of trunk motion during perturbed walking. Right panel: zoomed in section of the first perturbation in the left panel. See text for detailed explanation of the various parameters.

Furthermore, an exploratory analysis was performed to evaluate to what extent the perturbation parameter ($D$) depended on $SSG_{var}$. Normalizing trunk motion to steady state variability is required to evaluate if the perturbed gait pattern was outside an individual's own steady state range. However, the magnitude of steady state variability may have confounded the initial perturbation effect if different between fallers and non-fallers. Therefore, we checked for the presence of differences between fallers and non-fallers in terms of steady state gait variability using the following equation:

$$MD_{SSG} = \frac{1}{100} \sum_{i=1}^{i=100} \sqrt{\sum_{d=1}^{6} (SSG_{var}(i)_d)^2} \quad \textbf{Equation 7.3}$$

**Statistical analyses**

All statistical analyses were performed using SPSS version 23 (SPSS Inc, Chicago, IL, US). Outcome measures ($B_{pre}, \tau$ and $\beta$ for each perturbation type, and $MD_{SSG}$) were tested for normality using a Shapiro-Wilk test. To evaluate our main aim, whether we could discriminate between fallers and non-fallers based on the initial perturbation effect, the duration of the initial perturbation effect and the perturbation recovery, mixed model analyses of variances (ANOVA) were used. We included perturbation type (ipsilateral sway, contralateral sway, acceleration and deceleration) as within factor and group (fallers and non-fallers) as between factor. When significant main effects for Group or Type x Group interaction effects were found, independent samples *t*-tests were used to explore differences between fallers and non-fallers. In case of significant main effects for Type, pairwise comparisons with Bonferroni corrections were used to evaluate which perturbation type effected trunk motion the most. Finally, between-group differences in steady state gait ($MD_{SSG}$) were tested using an independent samples *t*-test. The level of significance was set to 0.05.

### 7.2.1.3 Results

**The initial perturbation effect**

We found a borderline significant main effect for Group (F(1, 1)=3.543, *p*=0.066) and a significant Type x Group interaction effect for $B_{pre}$ (F(1, 3)=5.283, *p*=0.002). Bonferroni-corrected post-hoc independent samples *t*-tests revealed significantly larger deviations in trunk motion for fallers after the contralateral sway (t=-2.107, df=45, $p$=0.041) and deceleration perturbation (t=-2.646, df=45, $p$=0.011) compared to non-fallers. No differences between fallers and non-fallers were found for the ipsilateral sway (t=-0.576, df=11.157, $p$=0.576) and acceleration (t=-1.080, df=45, $p$=0.286) perturbation (see Figure 7.4 – Left panel). Hence, the overall effect for Group was borderline.

The deceleration perturbation had the largest effect on $B_{pre}$, followed by the contralateral sway perturbation, the acceleration perturbation and the ipsilateral perturbation. All Bonferroni-corrected pairwise comparisons were significantly different ($p$<0.05), except for ipsilateral vs acceleration ($p$=0.121) (see Figure 7.4 – Left panel).

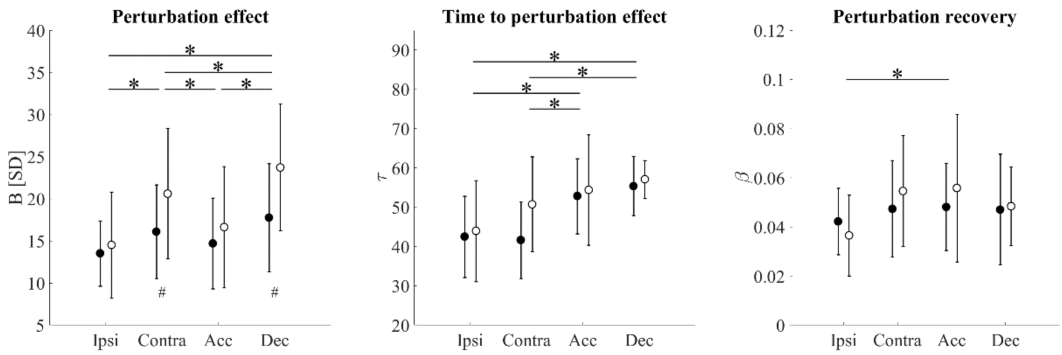



**Figure 7.4** Average perturbation parameters for non-fallers (black) and fallers (white). Left panel: average perturbation effect for ipsilateral sway (Ipsi), contralateral sway (Contra), acceleration (Acc) and deceleration (Dec) perturbation types. Mid panel: average perturbation recovery for the four perturbation types. Right panel: mean deviation from steady state gait. *Indicate significant differences between perturbation types. #Indicate significant differences between fallers and non-fallers. The level of significance was set to 0.05.

### The duration of the initial perturbation effect

No significant main effects for Group (F(1, 1)=1.866, $p$=0.179) or Type x Group interaction effects (F(1, 3)=1.843, $p$=0.142) were found, indicating that the time to reach the maximum effect of the perturbation did not differ between fallers and non-fallers.

The ANOVA did reveal significant main effects for Type (F(1, 3)=19.047, $p$<0.001). The maximum perturbation effect was reached faster for the ML perturbations compared to AP perturbations, but timing did not differ between the ML perturbations or the AP perturbations (see Figure 7.4 – Mid panel).

**The perturbation recovery**

The ANOVA did not reveal significant main effects for Group (F(1, 1)=0.380, *p*=0.540) or Type x Group interaction effects (F(1, 3)=1.008, *p*=0.391) for the perturbation recovery ($\beta$). Significant main effects were found for Type (F(1, 3)=3.343, *p*=0.021). Bonferroni-corrected pairwise comparisons showed that participants recovered faster from the ipsilateral perturbation compared to the acceleration perturbations. None of the other pairings were significantly different. (see Figure 7.4 – Right panel).

**Steady state gait**

Mean deviation in steady state trunk motion (t=0.513, df=45, *p*=0.610) did not differ between fallers (1.68±0.39) and non-fallers (1.61±0.36), indicating that abovementioned differences between fallers and non-fallers in the initial effect of the perturbation were not confounded by steady state gait variability.

#### 7.2.1.4 Discussion

The purpose of this analysis study was to evaluate whether we could discriminate between older adults with and without a history of falls based on the initial perturbation effect, the duration of the initial effect and the perturbation recovery. In line with our hypothesis, the data showed that older adults with a history of falls were significantly more affected by the deceleration perturbation as compared to those without. Using the continuous trunk motion measure we thus confirmed that our previously found Step x Group interaction effects for peak trunk velocity following the deceleration perturbation were indeed the result of differences between older adults with and without a history of falls. Furthermore, for the contralateral sway perturbation, a hint of differences between groups in backward MoS, but not peak trunk velocity, was previously found. Nonetheless, the current data revealed that, in terms of continuous trunk motion, fallers were indeed initially more affected by the contralateral sway perturbation compared to the non-fallers. Hence, continuous trunk motion is more revealing.

We did not find any significant differences between groups in terms of the initial duration of the perturbation effect or the rate of recovery. This was in contrast to our hypothesis, and earlier mentioned work of Krasovsky and colleauges (2012), who found that CoM in older adults was initially disturbed longer and recovery slower. However, direct comparison between our studies to explain these results is difficult. Krasovsky and colleauges applied a 250 ms backward pull of the ankle at 20-30% of the swing phase whereas we applied support surface perturbation starting at initial contact with a duration of 250 ms and 370 ms for the contralateral sway and deceleration perturbations, respectively. In addition, the premise of distinguishing between an initial and recovery phase is theoretical and may overlap. We therefore believe that the method as proposed by Bruijn and colleagues provides a good starting point to evaluate perturbed gait, but the parameter selection to describe differences between groups or conditions should be determined in combination with the protocol at hand and the population of interest.

Although we discriminated fallers from non-fallers based on the initial perturbation effect, a few cautions in the use of this measure should be noted. First, as addressed earlier, trunk motion was corrected for steady state variability. While this is needed to evaluate deviation in trunk motion within an individual’s range, high steady state variability in trunk velocity may result in an underestimation of the perturbation effect. However, mean deviation in steady state gait variability did not differ between fallers and non-fallers and hence there is no evidence of greater variability in trunk motion prior to the perturbations in those with a history of falls and so the measure is likely to be directly comparable. Second, perturbed trunk motion was expressed relative to steady state trunk motion. Previous research has demonstrated that people adapt their gait in anticipation of perturbations (Bhatt et al., 2013; McCrum et al., 2016; Sturdy et al., 2014; Yang et al., 2016). As such, new steady state gait patterns may have been adopted in between the subsequent perturbations. If this was the case, the perturbation effect was overestimated and possibly confounded by differences in an individual's ability to adapt gait. Further analysis is required to investigate whether inter-perturbation gait adaptations should be taken into account when evaluating the effect of perturbations on the gait pattern. Of interest would be to explore whether fallers and non-fallers adapt differently.

#### 7.2.1.5 Conclusion

Gait perturbations can be used to discriminate between older adults with and without a history of falls, but careful selection of outcome measures and perturbation types are required. The present work showed that the initial effect of contralateral sway and deceleration perturbations using a continuous measure of trunk velocity was sensitive enough to distinguish those with a history of falls from those without a history of falls as a group. We recommend it should therefore be used in further research seeking to screen the population for those at risk for falling. Directions for future work may also include exploring the sensitivity of easy-to-use methods to measure trunk motion like a single cluster of markers or an accelerometer on the trunk.

### 7.2.2 Analysis study 2 – Inter-perturbation adaptations in trunk motion to repeated gait perturbations in older adults with and without a history of falls

#### 7.2.2.1 Introduction

In everyday life, environmental perturbations are often unexpected and hence recovery requires *reactive* gait adaptations. Mimicking this in an experimental setup is not possible, which has been well-established by *proactive* gait adaptations in the gait pattern following repeated perturbations (McCrum et al., 2016; Pai et al., 2014b; Pavol et al., 2002; Sturdy et al., 2014; Wang et al., 2012). Such adaptations reduce the perturbation effect and may confound the ability to quantify reactive gait stability. Therefore, the first perturbation response can be considered most representative for an individual's "true" reactive gait stability. However, there is evidence that even the awareness of upcoming perturbations increases proactive gait stability (Heiden et al., 2006; Pater et al., 2015; Yang et al., 2016). In addition, analysing the first perturbation response only, is not possible when including various perturbation types like the present study due to the fact that adaptations occur as soon as after the first one or two perturbations (Pijnappels et al., 2005b; Wang et al., 2012). Thus, since we cannot make experimentally induced perturbations *unexpected*, we should aim to minimize anticipatory gait adjustments by making reactive gait assessment *unpredictable*.

Our mixed perturbation protocol may reduce the ability to effectively increase proactive gait stability in a number of ways. First, by including various perturbation types. Previous work has shown that adaptations may transfer in opposing perturbation directions (trips vs slips) (Bhatt et al., 2013) or from one task to another (recovery from backward slips when rising from a chair vs walking) (Wang et al., 2011) but not necessarily across planes (ML vs AP slips) (Martelli et al., 2016). Second, using treadmill-based perturbations allows for unpredictable timing. Overground perturbation studies have included catch trials (i.e. unperturbed trials) to minimize anticipation to the perturbations, but it is still obvious where the perturbation will occur. Finally, with a fixed treadmill speed gait stability cannot be increased by reducing gait speed (Wang et al., 2012). Nonetheless, generic adaptations to increase gait stability are still to be expected (Hak et al., 2013b), although the ability to do so may be reduced in older fallers compared to non-fallers (Galna et al., 2009; Weerdesteyn et al., 2005).

If gait adaptations are reflected in our trunk motion measure, this would mean that trunk motion values in between the perturbations would not return to pre-perturbed values (inter-perturbation adaptations). This would also mean that the perturbation effect was then overestimated, but more importantly between group differences in the perturbation effect could thus have been confounded by inter-perturbation adaptations. Therefore, the aim of this analysis study was to evaluate whether older adults with and without a history of falls showed proactive adaptations in terms of trunk motion to repeated but different types of perturbations and if so, whether these inter-perturbation adaptations differed between groups. We hypothesized that both fallers and non-fallers would adapt their gait in anticipation of the perturbations, and that these adaptations would be reflected in higher inter-perturbation trunk motion values compared to pre-perturbation trunk motion. Moreover, we hypothesized that the ability to adapt gait would be reduced in older adults with a history of falls and therefore deviation in inter-perturbation trunk motion would be lower compared to older adults without a history of falls. To evaluate our second aim, whether previously found differences between fallers and non-fallers following contralateral sway and deceleration perturbations but not ipsilateral sway and acceleration perturbations would still hold when gait was proactively adapted, we expressed trunk motion following the perturbation onset relative to trunk motion prior to the perturbation onset. This way the perturbation effect was corrected for inter-perturbation adaptations. We hypothesized that, if fallers indeed show less adaptive capacity, we would still be able to distinguish fallers from non-fallers based on the contralateral sway and deceleration perturbation effects. In fact, differences may be even more pronounced.

#### 7.2.2.2 Methods

A detailed description of the recruited participants, the equipment and the protocol has been described previously (section 6.2.1). Briefly, 39 elderly non-fallers and ten elderly fallers participated in this study. Participants walked on the GRAIL (Gait Analysis Interactive Lab, Motekforce Link BV, Amsterdam, The Netherlands) while lower limb and trunk kinematics were recorded. A two-minute steady state gait trial was completed followed by four types of perturbations; ipsilateral and contralateral sway, acceleration and deceleration perturbations. Perturbation types were repeated ten times in a randomized order.

**Data analyses**

To evaluate adaptations in trunk motion, we introduced a new parameter: $C_{inter}$. In line with $C_{pre}$, this is the average trunk motion value over three strides prior to each perturbation other than the initial perturbation. In the previous analysis (section 7.2.1), each perturbation effect was expressed relative to $C_{pre}$. If $C_{inter}$ increases then the use of $C_{pre}$ in the calculations may overestimate the perturbation effect. To determine whether trunk motion ($D$, as defined in Equation 7.1) adapted in anticipation of the perturbations, unperturbed trunk motion in between the consecutive perturbation was averaged and referred to as $C_{inter}$. To assure individuals had fully recovered from the previous perturbation, $C_{inter}$ was calculated over three strides (*i*=-300 to *i*=0) prior to the following perturbation onset. As previously described, average pre-perturbed trunk motion ($C_{pre}$) was calculated over three strides prior to the first perturbation onset (Figure 7.5).

To examine whether we could discriminate between fallers and non-fallers based on the perturbation effect ($B$) when taking into account inter-perturbation adaptations, the perturbation effect was expressed relative to its preceding $C_{inter}$ value and referred to as $B_{inter}$.

Since one participant did not complete the entire perturbation protocol, only the first 24 perturbations were included in the data analyses. Due to our randomized block design, this meant that six perturbations of each type had been applied (section 6.1.1).

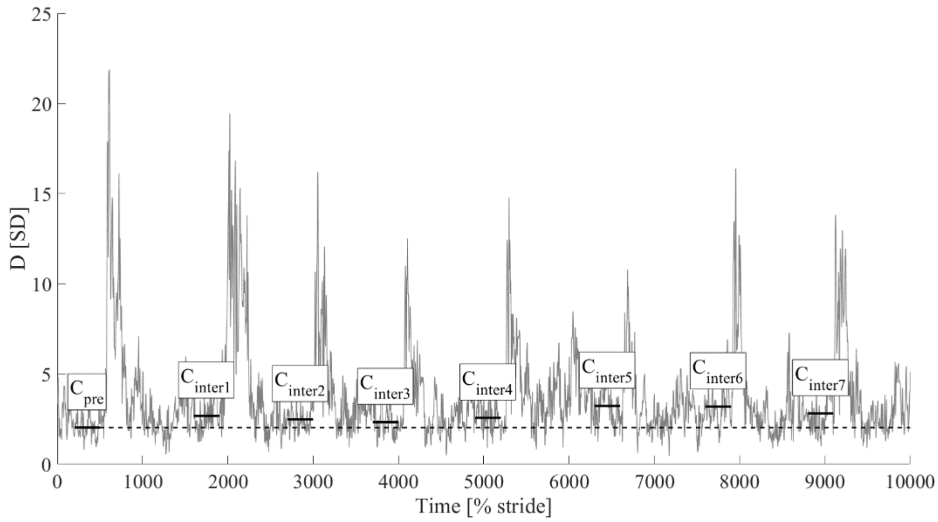



**Figure 7.5** Typical example of trunk motion (grey line) during a perturbation trial, containing eight perturbations. Gait adaptations in anticipation of the perturbation were found, indicated by higher $\boldsymbol{C_{inter}}$ values (second to eight black lines) compared to $\boldsymbol{C_{pre}}$ (first black line and dashed black line for reference purposes).

**Statistical analyses**

To evaluate whether fallers adapted differently compared to non-fallers, a mixed-model ANOVA was used with the perturbation number (perturbation 1 to 24) as within factor and group as between factor. In case of significant main effects for Perturbation number, post-hoc simple contrast were applied to examine whether inter-perturbed trunk motion differed from pre-perturbed trunk motion.

A mixed-model ANOVA with perturbation type (ipsilateral, contralateral, acceleration and deceleration perturbation) as within factor and group (fallers and non-fallers) as between factor was performed to evaluate differences in the perturbation effect. In case of significant main effects for Type, pairwise comparisons with Bonferroni corrections were used to evaluate which perturbation type affected trunk motion the most. When significant main effects for

Group or Type x Group interaction effects, independent samples $t$-tests were used to explore differences between fallers and non-fallers. The level of significance was set at 0.05.

#### 7.2.2.3 Results

**Inter-perturbation adaptations in trunk motion**

The ANOVA revealed a significant main effect for Perturbation number (F(1, 9.710)=3.043, $p$=0.001). Post-hoc simple contrasts showed that inter-perturbed trunk motion was significantly larger compared to pre-perturbed trunk motion, for all 23 perturbations ($p$<0.01). No significant main effect for Group (F(1, 1)=0.015, $p$=0.902) or Perturbation number x Group interaction effect (F(1, 9.710)=0.388, $p$=0.949) on $C_{inter}$ were found (Figure 7.6).

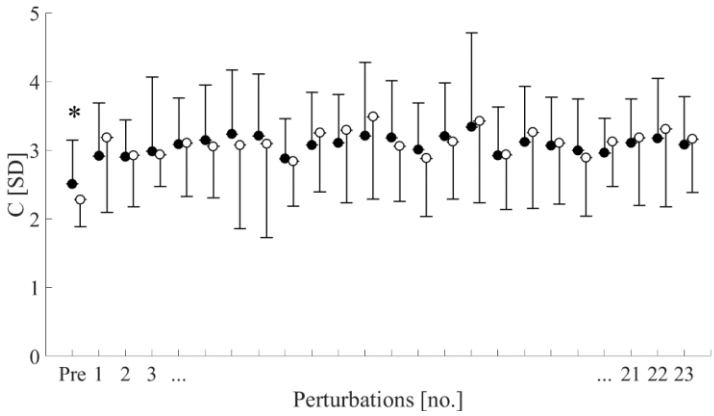



**Figure 7.6** Pre- and inter-perturbation trunk motion (C) for non-fallers (black dots) and fallers (white dots). Error bars represent the group standard deviation. All inter-perturbed trunk motion values were significantly different from pre-perturbed trunk motion (*). No significant main effects for Perturbation number or Group were found.

**The initial perturbation effect corrected for inter-perturbation adaptations**

Results of the ANOVA for $B_{inter}$ were comparable to $B_{pre}$; a near significant main effect for Group (F(1, 1)=3.213, $p$=0.080) and significant Type x Group interaction effect (F(1, 3)=5.088, $p$=0.002) were found. Moreover, post-hoc independent samples $t$-tests revealed significantly larger perturbation effects for fallers after the contralateral sway (t=-2.093, df=45, $p$=0.042) and deceleration perturbation (t=-2.511, df=45, $p$=0.016) compared to non-fallers. No differences between fallers and non-fallers were found in responses to the ipsilateral sway (t=-0.479, df=10.913, *p*=0.641) and acceleration (t=-0.932, df=45, *p*=0.356) perturbation. In addition, a significant main effect for Type (F(1, 3)=34.077, *p*<0.001) was found. Bonferroni-corrected *t*-tests showed that the deceleration perturbation had the largest effect, followed by the contralateral sway perturbation, the acceleration perturbation and the ipsilateral perturbation. All pairings were significantly different (*p*<0.05), except for ipsilateral versus acceleration (*p*=0.114). Between-perturbation and between-group differences for $B_{pre}$ and $B_{inter}$ are shown in Figure 7.7.

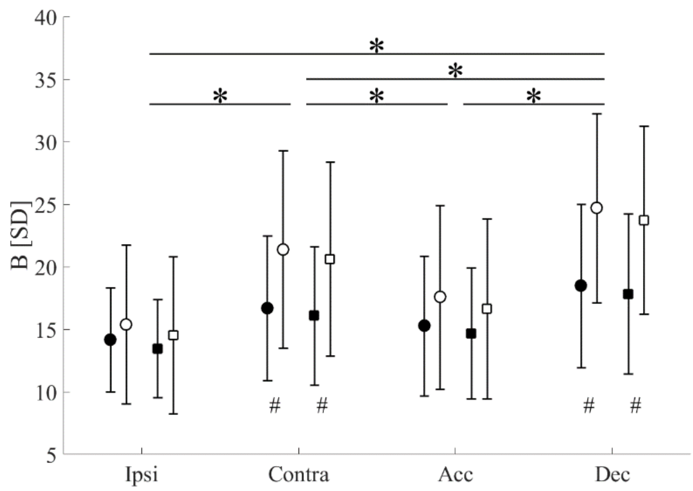



**Figure 7.7** Average perturbation effect for non-fallers (black) and fallers (white)relative to pre-perturbed (dots) and inter-perturbed (squares) trunk motion in response to ipsilateral sway (Ipsi), contralateral sway (Contra), acceleration (Acc) and deceleration (Dec) perturbations. Error bars represent the group standard deviation. *Indicate significant differences between perturbation types for both pre-perturbed and inter-perturbed trunk motion. #Indicate significant differences between fallers and non-fallers.

### 7.2.2.4 Discussion

This analysis study aimed to evaluate if and how inter-perturbed trunk motion changed from pre-perturbed trunk motion and whether older adults with and without a history of falls adapted differently. As hypothesized, we showed that deviation in inter-perturbed trunk motion was higher compared to pre-perturbed trunk motion for both fallers and non-fallers, and hence the perturbation effect was consistently overestimated when not accounting for these adaptations.

However and in contrast to our hypothesis, no differences in these adaptations were found between fallers and non-fallers. Therefore, it did not change the previously found between-types and between-group differences. In other words, deceleration perturbations were most challenging, followed by contralateral sway perturbations and fallers had more difficulty in recovery from both of these perturbation types

Adaptations were present immediately after the first perturbation in both fallers and non-fallers. Furthermore, a plateau response seemed to be reached quickly in both groups. A possible explanation for the lack of differences between adaptations in fallers and non-fallers could be the results of our mixed perturbation protocol. Given that it was unlikely that participants knew what the next perturbation type would be and when it would occur effective gait adaptations may have been limited in both groups. The fact that inter-perturbation adaptations are reflected in the trunk motion measure allows to further investigate the relation between proactive and reactive adaptations and evaluate whether a mixed perturbation protocol can indeed minimize predictability and hence would be preferable over single plane or single direction perturbation types.

A limitation of the trunk motion measure is that it solely captures deviation in the gait pattern and therefore it is unknown how inter-perturbed gait differed from pre-perturbed gait. Insight in adaptations in the individual trunk velocity profiles as well as the relation to spatio-temporal adjustments and gait stability measures could shed light on individual impairments and provide directions for gait training.

Based on this analysis, it can be concluded that older adults with and without a history of falls show statistically significant adaptations in trunk motion, immediately after the first perturbation, but these adaptations did not differ between groups or over time. Taking into account inter-perturbation adaptations did not change the ability to discriminate between fallers and non-fallers following the contralateral sway and deceleration perturbations on group level. However, we recommend to correct for these adaptations as the ability to adapt the gait pattern may vary from individual to individual.

### 7.2.3 Descriptive study – Individual analysis of trunk motion the responses to different gait perturbations

In this section we worked towards our main aim, to investigate whether we can use reactive gait assessment to identify risk for falls in older adults, by evaluating the perturbation response on an individual level. Understanding the individual data is required to support the clinician in interpretation of the data and guide the decision-making process on whether preventive actions to reduce the risk for (future) falls need to be taken.

This descriptive analysis was based on data collected in Chapter 6. The perturbation effect was quantified using the continuous trunk motion measure (section 7.2.1) and corrected for inter-perturbation adaptations (section 7.2.2).

#### 7.2.3.1 The perturbation effect on an individual level

The perturbation effect was quantified as in Equation 7.1. In short, we evaluated by how much perturbed trunk motion was outside an individual's own steady state range. Trunk velocity (in three directions and three planes) following the perturbation onset was compared to steady state trunk velocity and corrected for variability in steady state trunk velocity. The perturbation effect was defined as the maximum deviation in trunk motion in the first stride following the perturbation onset relative to average trunk motion over three strides prior to the perturbation onset.

On group level, fallers were more affected by the contralateral sway and deceleration perturbations compared to the non-fallers (Figure 7.8 – Left), on an individual level these differences were less clear as only a few fallers seemed to be more affected compared to the other fallers and non-fallers (Figure 7.8 – Right).

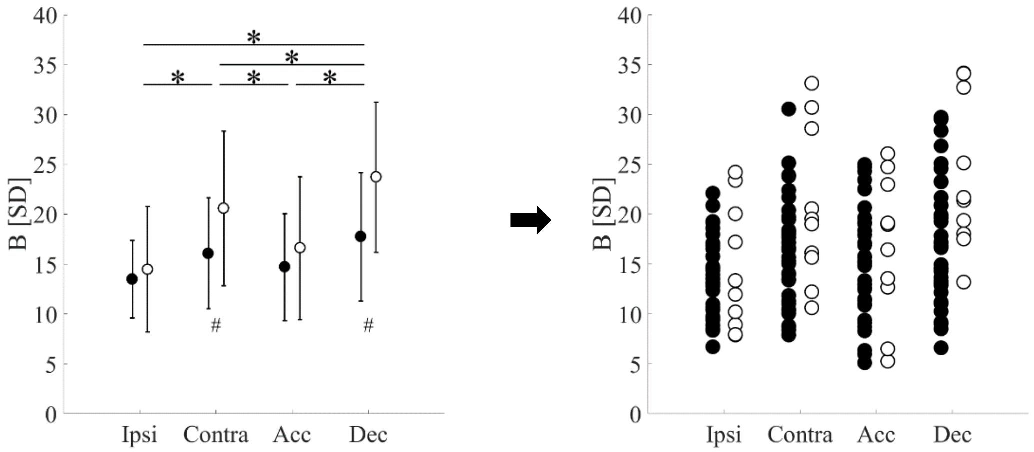



**Figure 7.8** Deviation in trunk motion on group (left) and individual (right) level for non-fallers (black dots) and fallers (white dots) following the ipsilateral sway (Ipsi), contralateral sway (Contra), acceleration (Acc) and deceleration (Dec) perturbations.

To further understand how individuals were affected by the different perturbations we tested the association between the effect of the various perturbation types on trunk motion by means of Pearson's correlation coefficients. The analysis revealed that the effect of all perturbation types were strongly correlated (Table 7.1), indicating that individuals who had difficulties with resisting and recovering from one perturbation type were likely to be more affected by the other perturbation types.

**Table 7.1** Pearson's correlation coefficients between the various perturbation effects. All significant at $p<0.001$.

| | Ipsilateral sway | Contralateral sway | Acceleration | Deceleration |
|---|---|---|---|---|
| Ipsilateral sway | 1 | 0.836 | 0.749 | 0.746 |
| Contralateral sway | | 1 | 0.804 | 0.810 |
| Acceleration | | | 1 | 0.717 |
| Deceleration | | | | 1 |

Next, it was evaluated how individuals performed relative to the group. Given the discriminative ability on group level, we limited this analysis to the contralateral sway and deceleration perturbations. The scatter plot presented in Figure 7.9 shows the relation in the perturbation effect following these two perturbations, for both fallers (white dots) and non-fallers (black dots). The variability (±1SD) in the perturbation effect following the contralateral sway and deceleration perturbations for fallers (light grey) and non-fallers (dark grey) is visualized by the filled ellipses. The centre of the ellipse represent the group average. Variability following perturbation effects were comparable (as indicated by the almost circular ellipses), but appeared larger for the fallers compared to non-fallers. This could have been the result of our heterogenous group of fallers, but may also be explained by the fact that the group of fallers was much smaller.

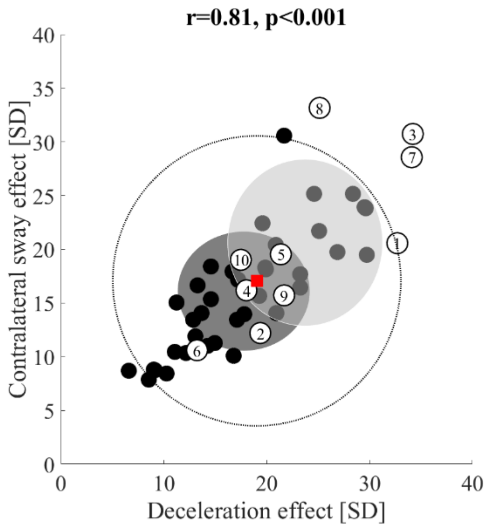



**Figure 7.9** Scatter plot of the contralateral sway and deceleration perturbation effects for the non-fallers (black dots) and fallers (white dots). Participant IDs of the fallers correspond to fall characteristics presented in Table 7.2. The dark grey (non-fallers) and grey (fallers) ellipses represent variability (1SD) for each group. Individuals outside the white ellipse (2*pooled SD) can be considered outliers.

In an attempt to compare the individual to the group as a whole, we searched for outliers. An individual was considered an outlier when deviating more than twice the pooled standard deviation (dashed ellipse) from the mean (red square). The pooled standard deviation was defined as: $\sqrt{SD_F^2 + SD_{NF}^2/2}$. From Figure 7.9 it can be seen that over half (6/10) of the fallers were affected by the contralateral sway and deceleration perturbation in a similar manner compared to the non-fallers. However, all individuals, except for one, who were more affected by the perturbations (i.e. data points in the upper right corner, outside the dashed ellipse) were older adults with a history of falls.

#### 7.2.3.2 The relation between fall circumstances and the perturbation effect

Older adults with a history of falls completed a questionnaire regarding circumstances, cause, frequency and fall-related injuries (Appendix IV). The data is presented in Table 7.2. Participant IDs in the table correspond to the IDs in Figure 7.9. Comparison between fall circumstances and the perturbation effect does not provide an evident relationship. The fallers group, for example, included a recurrent faller (participant 1) but also an individual who fell during running (participant 8).

**Table 7.2** Circumstances surrounding falls in older adults with a history of falls .

| ID | No. of falls | Cause | Activity | Environment | Injury |
|---|---|---|---|---|---|
| 1 | 3 | Loss of balance | Walking | Home, outside | - |
| 2 | 3 | Tripping | Walking | Away, familiair | Ankle sprain |
| 3 | 1 | Tripping/Unknown | Walking | Away, familiair | Minor bruises |
| 4 | 1 | Misplaced step | Walking | Away, familiair | Minor bruises |
| 5 | 1 | Trip | Tennis | Away, familiair | - |
| 6 | 4 | Trip<br>Loss of balance<br>Getting up | Walking<br>Getting out of the car | Everywhere | Bruised ribs |
| 7 | 1 | Slipping | Walking | Away, familiair | - |
| 8 | 1 | Slipping | Running | Away, familiair | Gluteal muscle strain, minor bruises |
| 9 | 1 | Tripping | Running | Away, unfamiliar | - |
| 10 | 2 | Tripping | Walking<br>Working in the shed | Away, familiair<br>Home, inside | - |

#### 7.2.3.3 Discussion

Individual analysis of our population revealed that differences between older adults with and without a history of falls are not clear. Some of the older adults showed larger deviations in trunk motion following the perturbations compared to the group as a whole, but whether or not they were actual fallers remains difficult to conclude. Future work to determine whether reactive gait assessment can be used to identify individuals at risk for falls should include: 1) comparison between perturbation responses of occasional fallers to those of frailer recurrent fallers to determine cut-off values for the perturbation effect, 2) evaluation of added value of reactive gait assessment to clinical and steady state gait measures and 3) prospective analysis.

# CHAPTER 8

# Discussion and conclusion

Early fall risk identification to prevent falls in older adults is highly important. Since most falls result from unsuccessful recovery of trips, slips or pushes, we aimed to evaluate whether we can use reactive gait assessment to identify risk for falls in older adults. Based on the literature (Chapter 2) and the technical equipment available (Chapter 4), we developed a reactive gait assessment including different types of perturbations and outcome measures. The assessment was evaluated in young and older adults (Chapter 5), and older adults with and without a history of falls (Chapter 6.1 and 7). These studies demonstrated that older adults did not differ in their responses from young in terms of spatio-temporal parameters and discrete gait stability measures whereas older adults with a history of falls were more affected by contralateral sway and deceleration perturbations in terms of continuous trunk motion than those without a history of falls (Chapter 7). However, a descriptive analysis (Chapter 7) revealed that on an individual level, these differences were less pronounced. Finally, we explored the validity of the protocol (Chapter 7) and demonstrated that adaptations in anticipation of the repeated perturbations did not differ between older adults with and without a history of falls, indicating that the differences between the groups were the result of reactive rather than proactive gait adaptations.

In this general discussion, we address how this thesis made important **contributions** in exploring the use of reactive gait assessment for fall risk identification in older adults. Furthermore, based on our findings and on limitations in our studies, we provide **recommendations** for the future scientific research and development, as well as the clinical implications of an evidence-based, affordable and easy-to-use reactive gait assessment for identification of risk for falls in older adults. Finally, our findings also provide **opportunities** towards other applications of reactive gait assessment. A schematic overview of this general discussion is presented in Figure 8.1.

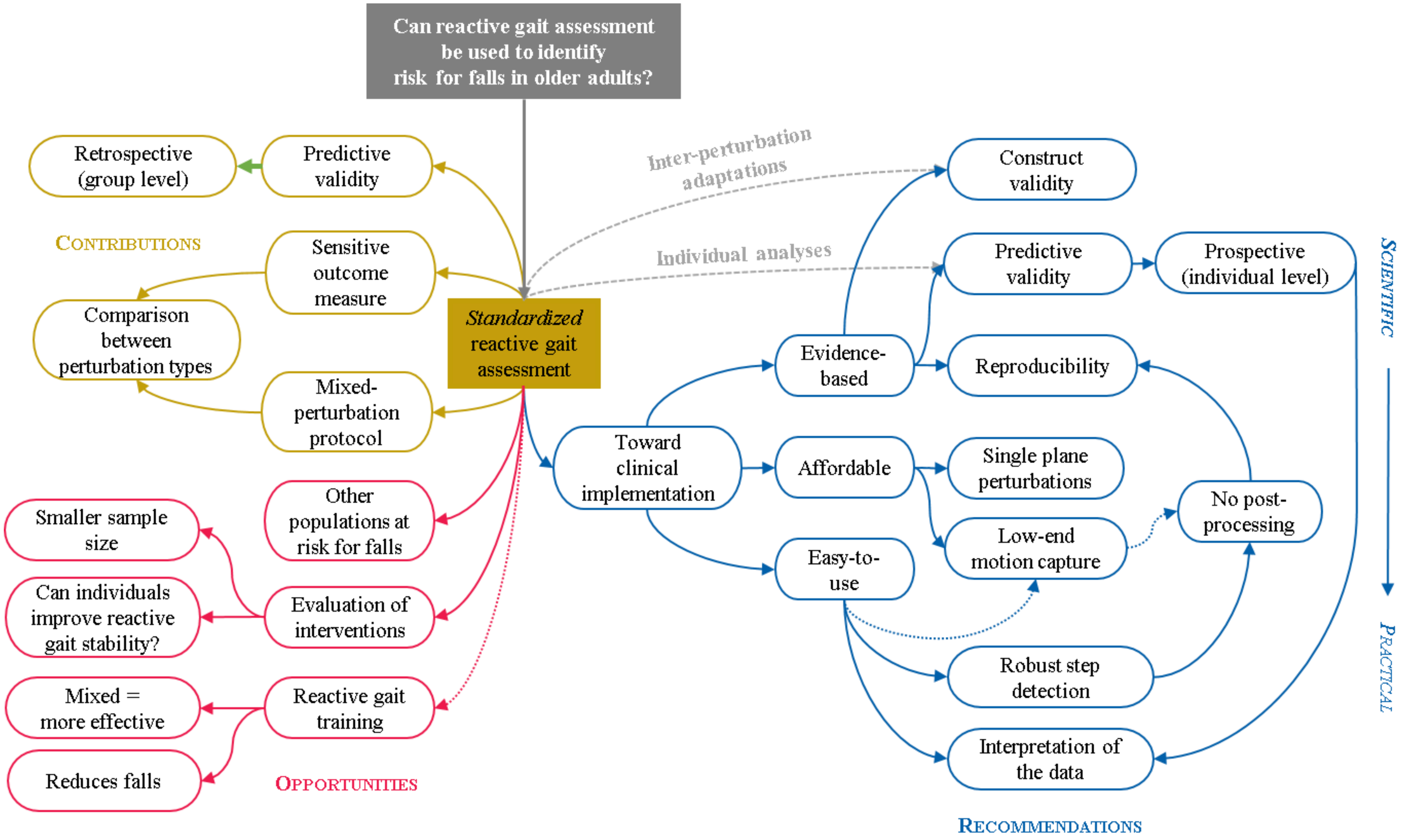



**Figure 8.1** Schematic overview of the contributions of this thesis, recommendations for future work and opportunities towards other implementations.

## 8.1 Contributions in exploring the use of reactive gait assessment for fall risk identification in older adults

A reactive gait assessment requires perturbations that effectively challenge the gait pattern and a sensitive outcome measure to quantify the perturbation response. Selection of a suitable perturbation type for a reactive gait assessment has not been given much attention. With the mixed-perturbation protocol we repeatedly demonstrated that the gait pattern was more affected by contralateral sway and deceleration perturbations compared to ipsilateral sway and acceleration perturbations, respectively (Chapter 5-7.1). Using discrete gait stability measures, our first studies showed that gait stability was more affected in the lateral direction following the contralateral sway perturbation compared to the ipsilateral sway, and in the backward direction following the deceleration perturbation compared to the acceleration perturbation (Chapter 5 & 6.1). These results are not surprising given the perturbation direction but would have been opposite had we have chosen to evaluate balance in the medial and anterior direction, respectively. This discrepancy has been seen in the literature, where the AP perturbation response has both been described in terms of forward stability (Ilmane et al., 2015; Punt et al., 2017) and backward stability (Kagawa et al., 2011; Martelli et al., 2017) irrespective of the perturbation direction, which makes between-studies comparison difficult. Our selected trunk motion measure provides a solution to this problem as it captures overall deviation in the gait pattern and thereby allows for direct comparison of perturbation effects within and across planes. Using this continuous measure we confirmed that the gait pattern was more affected by contralateral sway and deceleration perturbations compared to ipsilateral and acceleration perturbations, respectively, which may suggest that these perturbation types are most suitable for a reactive gait assessment (Chapter 7.1). An additional advantage of comparison between various perturbation types is that it can reveal direction-dependent impairments and guide the clinicians in developing an individualize fall prevention programmes. For example, increasing knee and hip extensor power may be primarily advised when difficulties arise following AP perturbations (Pijnappels et al., 2005c), whereas hip abductor muscle strength and power training may be primarily advised when gait is more affected following ML perturbations (Hilliard et al., 2008). It should be noted that the trunk motion measure does not capture gait stability as previously defined (i.e. the relation between the centre of mass motion state and the base of support) but solely captures deviation in trunk motion and was therefore initially not selected as a potential outcome measure.

**CONTRIBUTION** We developed a standardized reactive gait assessment including a mixed perturbation protocol and a sensitive outcome measures that allows for comparison of responses to different perturbation types and directions.

Using the reactive gait assessment we successfully discriminated between older adults with and without a history of falls. Previous studies have been successful in distinguishing fallers from non-fallers based on the perturbation response (Bhatt et al., 2011a; Pavol et al., 2001; Pijnappels et al., 2005b). However, these studies classified fallers and non-fallers based on the ability to recover from the experimental perturbation. While this provides insight in requirements to prevent falling, this approach has a few limitations: 1) unsuccessful recovery is not necessarily related to increased risk for falls in everyday life, 2) recovery responses are related to a specific perturbation, namely the one induced in the experiment, and different perturbation types may yield different classification, and 3) an actual fall needs to be induced which may not be preferable in clinical practice. Our work contributed to the literature by classifying older adults based on retrospective falls and thereby we took an important first step toward determining the predictive validity of reactive gait assessment for fall risk identification in older adults.

**CONTRIBUTION** We demonstrated that reactive gait assessment has potential to discriminate between (retrospective) fallers and non-fallers and therefore assess fall risk.

## 8.2 Toward clinical implementation – Recommendations for future research and development

To warrant clinical implementation, a standardized reactive gait assessment needs to be evidence-based, affordable and easy-to-use (Rowe, 2012). With respect to **evidence-based**, multiple steps need to be taken to further investigate the potential of reactive gait assessment for fall risk identification in older adults. Given that our work was highly explorative, we first need to reproduce these experiments to evaluate whether our findings are similar in a different population. Our reactive gait assessment provides a standardized manner to collect such data.

Next, the predictive validity of the reactive gait assessment needs to be further evaluated. We were able to discriminate older adults with a history of falls from those without, on a group level. As a next step, we analysed the data on an individual level, which revealed that the differences between fallers and non-fallers were less clear (Chapter 7). Our group of fallers was heterogeneous and also included fit and healthy individuals. This is inherent to the multifactorial nature of falls, which often results from a complex interaction of risk factors such as physical capacity, fear of falling, activity level, risk-taking behaviour or medication (Teasell et al., 2002). Better understanding of what these deviations in trunk motion following the contralateral sway and deceleration perturbation mean for the individual is required to determine risk for falls status and whether or not referral to a fall prevention programme is needed. However, we did find significant differences between fallers and non-fallers in the perturbation response, while no significant differences between the groups in clinical and steady state gait measures were found. Clinical measures show ceiling effects (Barry et al., 2014; Gates et al., 2008; Laessoe et al., 2007), but steady state gait measures have repeatedly discriminated between fallers and non-fallers (e.g. Rispens et al., 2015; Toebes et al., 2012; van Schooten et al., 2016, 2015). The lack of discriminative ability based on these steady state measures may thus suggest that reactive gait assessment is more sensitive in distinguishing healthy and fit older fallers from non-fallers but the added value to these other clinical and steady state measures needs further investigation.

Another step in building evidence for the use of reactive gait assessment in fall risk identification is evaluating the construct validity. In other words, does the assessment indeed quantify reactive gait stability? An analysis study was performed to evaluate whether the perturbation effect was confounded by inter-perturbation adaptations in trunk motion. We demonstrated that gait was immediately adapted after the first perturbation, but these adaptations did not differ between groups. This may suggest that differences between fallers and non-fallers, in the perturbation effect following the contralateral sway and deceleration perturbations, were the result of reduced reactive gait stability rather than smaller proactive gait adaptations. However, we could not evaluate whether these proactive adaptations were adequate and reduced the effect of the different perturbation types. Given the immediate adaptations such analysis would yield comparison with the first perturbation. However, the perturbation type differed between participants due to randomization. Further insight in the relation between proactive adaptations and the perturbation effect could shed light on whether a mixed perturbation protocol can indeed minimize predictability and hence the construct validity of our reactive gait assessment can be established.

**RECOMMENDATION** Due to the exploratory nature of our studies, the added value of reactive gait assessment should be further investigated to determine its reproducibility and validity in identifying the risk for falls on an *individual* level.

The present reactive gait assessment requires high-end technology. Future work is warranted to simplify the perturbation protocol and provide an **affordable** solution for use in clinical practice. Evidently, the hardware requirements need to be minimized to reduce the costs. The first obvious improvement would be to reduce complexity by simplifying the perturbation protocol and including single-plane perturbations only. Therefore, knowledge on which perturbation type discriminates best is required. It has been suggested that controlling ML gait stability is more important (Bauby and Kuo, 2000) and that ML perturbations are more challenging (McIntosh et al., 2017; Vlutters et al., 2016) compared to the AP direction. Our results indicated the opposite, as trunk motion deviated more following the deceleration perturbation compared to the contralateral sway perturbation. However, the former were of much higher intensity (deceleration: 0.13 m, 0.88 m/s, 8.74 m/s$^2$; contralateral sway: 0.05 m, 0.30 m/s, 3.26 m/s$^2$ when walking at 1 m/s) and therefore we cannot conclude which perturbation type was most challenging. Furthermore, the mixed-perturbation protocol was aimed to minimize perturbation-specific adaptations, which may have enhanced the discriminative power. However, future work should further explore whether we can still discriminate between fallers and non-fallers using single plane perturbations.

**RECOMMENDATION** Future research should evaluate the most suitable direction and discriminative ability of a single-plane perturbation protocol to minimize hardware requirements and make a reactive gait assessment affordable for use in clinical practice.

Another way to minimize hardware requirements is the use of low-end motion capture systems to quantify deviation in trunk motion. In the present thesis trunk kinematics were calculated using a musculoskeletal model based on 29 lower limb and trunk markers. Accurate marker tracking requires a number of high-end motion capture cameras (generally at least ten) and advanced software to track and label the markers in real-time, and process the data offline.

Simplifying this procedure will therefore not only reduce costs but also improve **ease-of-use**. An alternative would be a cluster of markers on a rigid plate attached to the trunk. This would cut down the number of cameras required as well as the need for marker labelling and post-processing. An additional cluster on the pelvis may be required to evaluate trunk movement relative to the pelvis (as in the musculoskeletal model) rather the global reference frame. Exploring these options can be done using a selection of the trunk and pelvis markers of the data collected for this thesis. Finally, taking it a step further, evaluating the use of accelerometers may be worthwhile as this would eliminate the need for a motion capture system.

> **RECOMMENDATION** To make the reactive gait assessment more affordable and easier-to-use, evaluation of the accuracy of low-end motion capture devices in quantifying trunk motion following perturbations is needed.

Furthermore, automatic gait event detection appeared to be challenging for perturbed gait, especially following the deceleration perturbations due to highly variable responses. We observed various strategies which were in line with previously reported responses to overground backward slips (Moyer et al., 2009); either the participant was able to elongate the swing phase and lengthen the step or the swing phase was shortened by early foot contact. When the swing phase was aborted quickly, the foot (most often only the forefoot) temporarily contacted the ground while weight remained above the perturbed stance foot allowing the trailing leg to go back into swing phase again (i.e. double step). This way, the normal gait pattern could be recovered relatively quickly. A less efficient recovery strategy seemed to be a slow but preliminary abortion of the swing phase. In this case, the foot entirely contacted the ground and weight was shifted to the trailing leg resulting in a preliminary and longer double support phase. We thus observed variable foot position relative to the pelvis, variable foot orientation at foot contact, inconsistent weight shifts and/or an inconsistent sequence of gait events when recovering from the deceleration perturbation and therefore visually checked the data for missing or incorrectly detected gait events. This is time-consuming, requires expertise and can be error-prone. Therefore, developing a robust step detection algorithm is necessary for both clinical practice and standardization of data analysis. An alternative solution would be to eliminate the need for time-normalization of the data to strides and rather compare perturbed gait to steady state gait over a fixed period of time. However, time shifts in the gait cycle due to changes in temporal parameters, would result in comparing different gait phases and overestimation in deviation from steady state gait. In addition, when gait events are available simple discrete spatio-temporal gait parameters can be used to facilitate understanding the perturbation response.

**RECOMMENDATION** Development of a robust step detection algorithm that consolidates various perturbation recovery strategies to reduce post-processing time, minimize expert knowledge and standardize data analysis.

Finally, our trunk motion measure quantifies the deviation in trunk velocity following the perturbation relative to steady state gait in a 6D state space, which presumably is a rather abstract concept for most clinicians. If we can relate the perturbation response to future falls we can guide the clinician in interpretation of the data and the clinical decision-making process, by for example providing cut-off points for individuals at risk for falling.

**RECOMMENDATION** Future establishment of the predictive validity of reactive gait assessment to *identify* risk for falls may guide the clinicians in the decision-making process to *prevent* falls.

In conclusion, we recommend future work to be clinically driven and focus on simplifying the reactive gait assessment. Therefore, we urge the company Motek to: 1) facilitate standardized data collection by providing current and future customers with the developed reactive gait assessment and 2) collaborate with clinical and scientific research groups to fill the gaps in understanding required to develop of a clinically feasible reactive gait assessment.

## 8.3 Further opportunities for implementation of the reactive gait assessment

Reactive gait assessment for fall risk identification may be applicable for other populations at risk for falls as well. For example, responses to gait perturbations have recently been studied in for example stroke survivors (Honeycutt et al., 2016; Punt et al., 2017), lower limb amputees (Crenshaw et al., 2013; Sessoms et al., 2014; Shirota et al., 2015) or children with cerebral palsy (Sloot et al., 2017). In contrast to idiopathic falls in older adults, these patient populations often experience specific gait impairments, and therefore our findings may not be directly transferable. Nonetheless, the developed reactive gait assessment offers standardized data collection and systematically investigation if and how reactive gait assessment can be used to identify risk for falls in specific patient populations.

Another opportunity for the use of our reactive gait assessment lies in the evaluation of intervention programmes. Currently there is no standardized assessment to examine the effectiveness of fall prevention interventions (Salzman, 2010). With a sensitive assessment we can (at least) evaluate whether older adults engaged in fall prevention programmes can indeed improve reactive gait stability. In addition, sensitive outcome measures allow for smaller sample sizes and hence faster and cheaper randomized controlled trials to investigate the effectiveness of fall prevention interventions. It should be noted, that responses to perturbations can improve with repeated exposure and a training effect in the assessment may occur (Mansfield et al., 2015; McCrum et al., 2017; Pai et al., 2014a). Following this line of reasoning, the perturbation protocol may be a good starting point for a training intervention to improve reactive gait. The various perturbation types included offer variability and minimum predictability, which are believed to be important for reactive gait training (McCrum et al., 2017). In addition, with minimal adjustments in the D-Flow application, the perturbation intensities can easily be tweaked to tailor the training to the individual's ability.

**OPPORTUNITIES** Potential implementations of the developed perturbation protocol include screening other populations at risk for falls, evaluation of fall prevention training and reactive gait training.

## 8.4 Conclusion

In this thesis we demonstrated that responses to gait perturbations can be used to discriminate between older adults with and without a history of falls provided a suitable selection of perturbation type and outcome measures. We were able to discriminate between fallers and non-fallers using our reactive gait assessment, while no differences between fallers and non-fallers were found in clinical and steady state gait measures. This suggests that reactive gait assessment has added value in fall risk identification in older adults. We recommend using the contralateral sway and deceleration perturbations of our developed perturbation protocol and assessing responses by means of trunk motion to further evaluate the use of reactive gait assessment for fall risk identification in older adults. Motek as an industrial company can facilitate this by providing customers with the developed reactive gait assessment and encourage clinically driven research to examine the reproducibility and validity of reactive gait assessment using evidence-based, affordable and easy-to-use technologies.

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## APPENDIX I

### Human Body Model marker sets

**Table A-I.1**Marker placement for the first and second Human Body Model templates.

| Label | Position | Placement remarks | HBM1 | HBM2 |
|---|---|---|---|---|
| | | Head and trunk | | |
| L(R)HEAD | Left (right) head | Just above the ear, in the middle. | ✓ | |
| THEAD | Top head | On top of the head, in line with LHEAD and RHEAD. | ✓ | |
| FHEAD | Forehead | Between line LHEAD/RHEAD and THEAD a bit right from the center. | ✓ | |
| C7 | C7 | On the 7th cervical vertebrae. | ✓ | ✓ |
| T10 | T10 | On the 10th thoracic vertebrae. | ✓ | ✓ |
| SACR | Sacrum bone | On the sacral bone. | ✓ | |
| NAVE | Navel | On the navel. | ✓ | |
| XYPH | Xiphoid process | Xiphiod procces of the sternum. | ✓ | ✓ |
| STRN | Sternum | On the jugular notch of the sternum. | ✓ | ✓ |
| BBAC | Scapula | On the inferior angle of the right scapula | ✓ | ✓ |
| | | Upper limb | | |
| L(R)SHO | Left (right) shoulder | Left (right) acromion | ✓ | ✓ |
| L(R)DELT | Left (right) deltoid muscle | apex of deltoid muscle | ✓ | |
| L(R)LEE | Left (right) lateral elbow | Left (right) lateral epicondyle of the elbow. Upper one in T-pose. | ✓ | |
| L(R)MEE | Left (right) medial elbow | Left (right) medial epicondyle of the elbow. Lower one in T-pose | ✓ | |
| L(R)FRM | Left (right) forearm | On 2/3 on the line between the LLEE and LMW. | ✓ | |
| L(R)MW | Left (right) medical wrist | On styloid process radius, thumb side | ✓ | |
| L(R)LW | Left (right) lateral wrist | On styloid process ulna, pinky side | ✓ | |
| L(R)FIN | Left (right) fingers | Center of the hand. Caput metatarsal 3 | ✓ | |

| Label | Position | Placement remarks | HBM1 | HBM2 |
|---|---|---|---|---|
| | | Pelvis and lower limb | | |
| L(R)ASIS | Pelvic bone left (right) front | Left (right) anterior superior iliac spine | ✓ | ✓ |
| L(R)PSIS | Pelvic bone left (right) back | Left (right) posterior superior iliac spine | ✓ | ✓ |
| L(R)GTRO | Left (right) greater trochanter of the femur | On the center of the left (right) greater trochanter | ✓ | |
| FL(R)THI | Left (right) thigh | On 1/3 on the line between the LGTRO and LLEK. | ✓ | ✓ |
| L(R)LEK | Left (right) lateral epicondyl of the knee | On the lateral side of the joint axis | ✓ | ✓ |
| L(R)MEK | Left (right) medial epicondyl of the knee | On the medial side of the joint axis | | ✓ |
| L(R)ATI | Left (right) anterior of the tibia | On 2/3 on the line between the LLEK and LLM. | ✓ | ✓ |
| L(R)LM | Left (right) lateral malleolus of the ankle | The center of left (right) lateral malleolus | ✓ | ✓ |
| L(R)MM | Left (right) medial malleolus of the ankle | The center of left (right) medial malleolus | | ✓ |
| L(R)HEE | Left (right) heel | Center of the heel at the same height as the toe | ✓ | ✓ |
| L(R)TOE | Left (right) toe | Tip of big toe | ✓ | |
| L(R)MT2 | Left (right) 2th meta tarsal | Caput of the 2$^{nd}$ meta tarsal bone, on joint line midfoot/toes | | ✓ |
| L(R)MT5 | Left (right) 5th meta tarsal | Caput of the 5$^{th}$ meta tarsal bone, on joint line midfoot/toes | ✓ | |
| Total number of markers | | | 47 | 29 |

## APPENDIX II

**User experience evaluation**

**1. I think it is useful to measure ones stability during walking**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**2. I understand why perturbations (i.e. unexpected movements of the platform, noises or changes in room light which have challenged or intended to challenge your balance) were used**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**3. The instructions provided beforehand were sufficient**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**4. I experienced fear of falling throughout the experiment**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**5. The total duration (introduction, preparation, walking trials) of the experiment was too long**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**6. The experiment was tiring**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**7. I did not mind wearing tight clothes**

| 1 | 2 | 3 | 4 | 5 |
|---|---|---|---|---|

Strongly disagree Neutral Strongly agree

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**8. Rank the various perturbation (i.e. unexpected movements of the platform, noises or changes in room light which have challenged or intended to challenge your balance) types from easiest (1) to most difficult (5)?**

- Platform movement to the side; .........
- Trip (i.e. stop of the treadmill belt); .........
- Slip (i.e. acceleration of the treadmill belt): .........
- Sudden darkness: .........
- Loud noise: .........

9. **Did you feel like any of the perturbations were too difficult to resist?** Yes / No

**If yes, please check the concerned perturbations**

- ☐ Platform movement to the side
- ☐ Trip (i.e. stop of the treadmill belt)
- ☐ Slip (i.e. acceleration of the treadmill belt)
- ☐ Sudden darkness
- ☐ Loud noise

10. **Do you have any other comments, remarks, suggestions, etc?**

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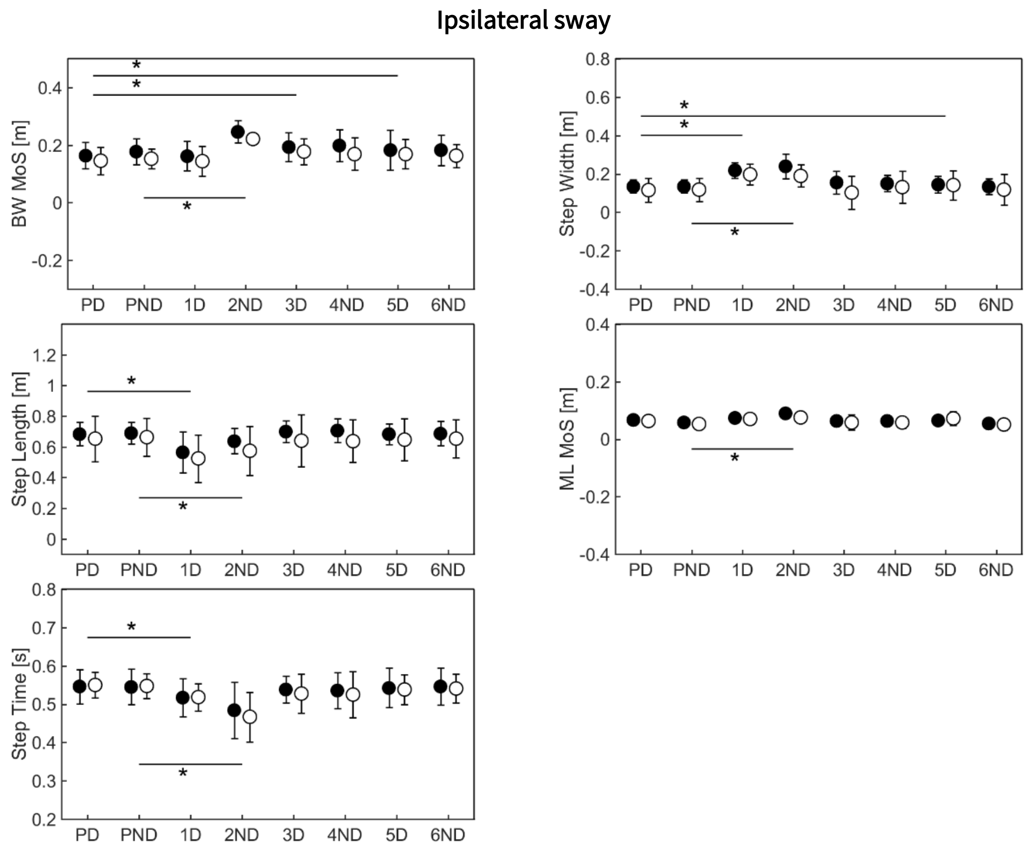
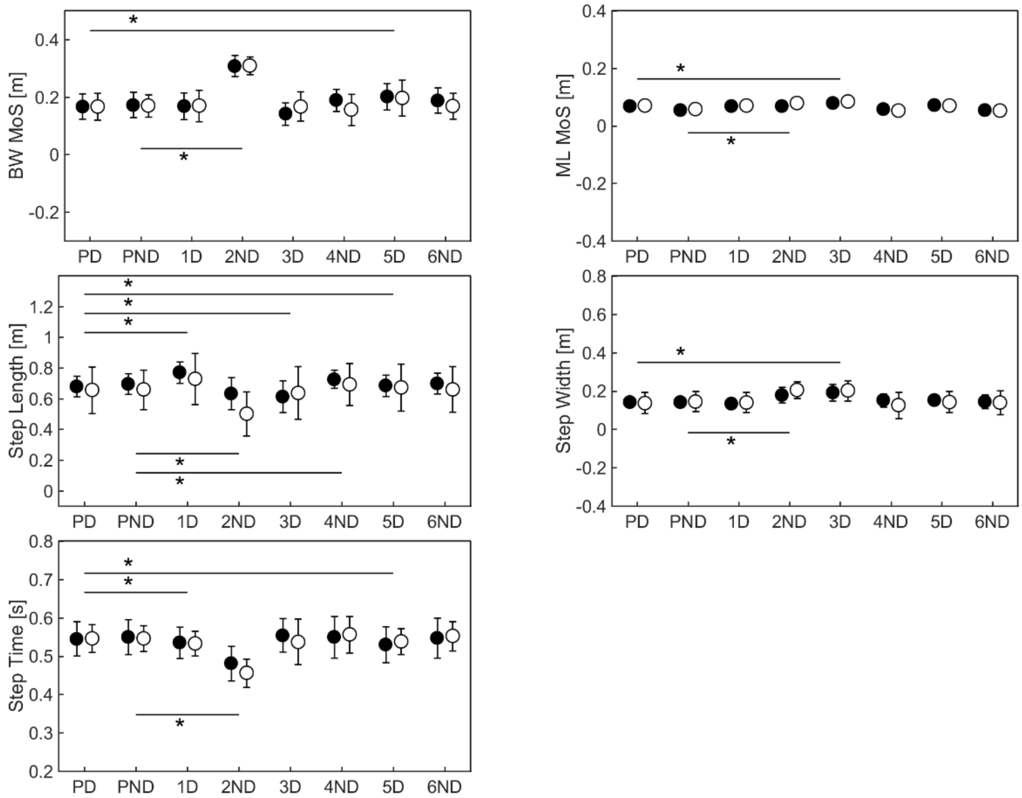
# APPENDIX III

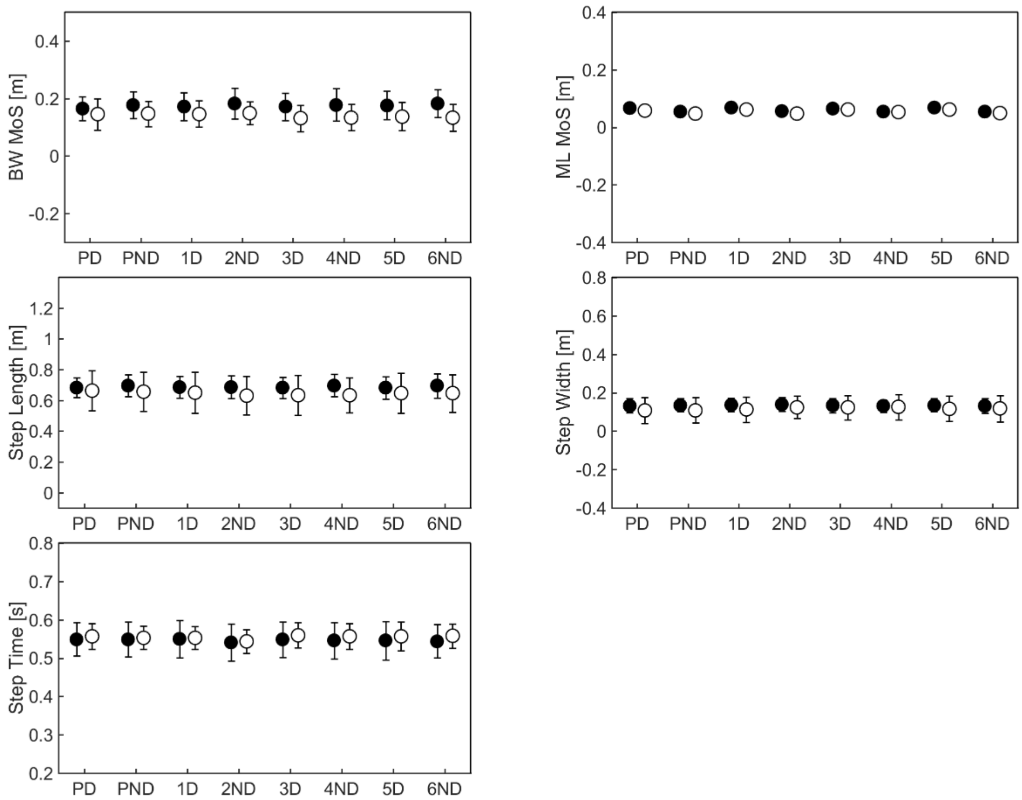
**Supplementary material: Experimental study 1 – Gait stability in response to platform, belt and sensory perturbations in young and older adults**



## Acceleration




## Visual



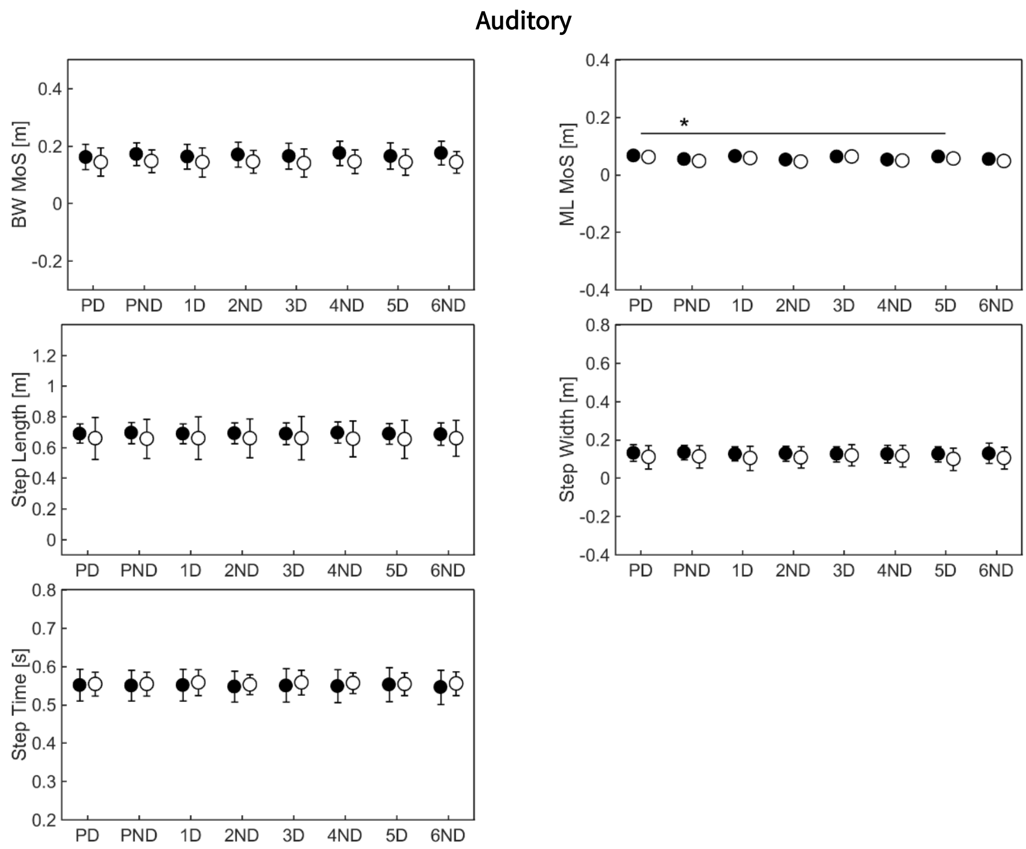



**Figure A-III.1**Responses to ipsilateral sway, acceleration, visual and auditory perturbations. Mean and standard deviations of backward (BW) and medio-lateral (ML) margins of stability (MoS), step length, width and time for steps for the ipsilateral sway, acceleration, visual and auditory perturbations. Black dots represent for young adults whereas white dots represent older adults. Significant differences between pre- and post-perturbations steps are indicated with *.

**Table A-III.1** Pre- versus post-perturbation steps. Paired-samples *t*-tests were used to compare dominant post-perturbation steps (1D, 3D and 5D) with the average dominant pre-perturbation step, and non-dominant post-perturbation steps (2ND, 4ND, 6ND) with the average non-dominant pre-perturbation step. Significantly different effects at $p$<0.05 are printed in bold.

| | ML MoS | | BW MoS | | Step time | | Step length | | Step width | |
|---|---|---|---|---|---|---|---|---|---|---|
| Step | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* |
| Ipsilateral sway | | | | | | | | | | |
| 1D | -2.753 | 0.082 | 0.577 | 1.000 | **7.429** | **<0.001** | **9.667** | **<0.001** | **-13.321** | **<0.001** |
| 2ND | **-7.873** | **<0.001** | **-9.942** | **<0.001** | **6.287** | **<0.001** | **4.815** | **0.001** | **-7.928** | **<0.001** |
| 3D | 1.120 | 1.000 | **-5.464** | **<0.001** | 2.521 | 0.132 | -0.342 | 1.000 | -0.378 | 1.000 |
| 4ND | -2.237 | 0.234 | -2.950 | 0.054 | 2.379 | 0.176 | 0.745 | 1.000 | -2.610 | 0.110 |
| 5D | -1.669 | 0.680 | **-4.315** | **0.003** | 2.473 | 0.146 | 0.684 | 1.000 | **-3.289** | **0.026** |
| 6ND | 1.169 | 1.000 | -1.815 | 0.523 | 0.868 | 1.000 | 1.511 | 0.895 | 0.163 | 1.000 |
| Contratleral sway | | | | | | | | | | |
| 1D | **26.264** | **<0.001** | 0.533 | 1.000 | -0.862 | 1.000 | **3.309** | **0.025** | **14.084** | **<0.001** |
| 2ND | **4.202** | **0.004** | **-4.698** | **0.001** | **4.912** | **0.001** | **4.133** | **0.004** | 2.636 | 0.104 |
| 3D | **-10.245** | **<0.001** | **-5.351** | **<0.001** | **3.357** | **0.022** | **4.411** | **0.002** | **-11.539** | **<0.001** |
| 4ND | **-3.624** | **0.013** | **-6.005** | **<0.001** | **4.419** | **0.002** | 2.528 | 0.130 | **-6.240** | **<0.001** |
| 5D | **-3.631** | **0.012** | -2.731 | 0.085 | **3.666** | **0.011** | **3.152** | **0.035** | **-3.687** | **0.011** |
| 6ND | -2.158 | 0.273 | -2.919 | 0.057 | 1.584 | 0.790 | **3.591** | **0.014** | **-3.121** | **0.037** |
| Acceleration | | | | | | | | | | |
| 1D | -0.312 | 1.000 | -0.846 | 1.000 | **4.715** | **0.002** | **-11.268** | **<0.001** | 0.884 | 1.000 |
| 2ND | **-7.006** | **<0.001** | **-21.256** | **<0.001** | **9.069** | **<0.001** | **4.303** | **0.004** | **-6.164** | **<0.001** |
| 3D | **-5.249** | **0.001** | 1.581 | 0.808 | -0.175 | 1.000 | **3.462** | **0.021** | **-7.935** | **<0.001** |
| 4ND | 0.070 | 1.000 | -0.386 | 1.000 | -0.885 | 1.000 | **-6.823** | **<0.001** | 0.434 | 1.000 |
| 5D | -1.235 | 1.000 | **-5.774** | **<0.001** | **3.717** | **0.012** | **-3.516** | **0.019** | -1.954 | 0.417 |
| 6ND | 1.021 | 1.000 | -1.579 | 0.811 | -0.413 | 1.000 | -0.269 | 1.000 | 0.098 | 1.000 |
| Deceleration | | | | | | | | | | |
| 1D | -1.555 | 0.844 | 0.053 | 1.000 | -0.120 | 1.000 | **12.099** | **<0.001** | **-3.570** | **0.017** |
| 2ND | -1.728 | 0.627 | **19.769** | **<0.001** | -0.804 | 1.000 | **7.342** | **<0.001** | 1.058 | 1.000 |
| 3D | **-4.827** | **0.001** | **-13.325** | **<0.001** | **8.554** | **<0.001** | 0.606 | 1.000 | -2.638 | 0.112 |
| 4ND | **-4.602** | **0.002** | **-5.446** | **<0.001** | **3.548** | **0.018** | **4.712** | **0.002** | **-5.573** | **<0.001** |
| 5D | 1.177 | 1.000 | -2.503 | 0.146 | 0.810 | 1.000 | 1.686 | 0.675 | -1.363 | 1.000 |
| 6ND | -1.760 | 0.593 | -1.294 | 1.000 | 1.519 | 0.897 | 2.204 | 0.261 | -1.582 | 0.807 |
| Visual | | | | | | | | | | |
| 1D | -1.509 | 0.898 | -1.785 | 0.553 | - | - | - | - | - | - |
| 2ND | -0.536 | 1.000 | -1.694 | 0.651 | - | - | - | - | - | - |
| 3D | -0.446 | 1.000 | 0.817 | 1.000 | - | - | - | - | - | - |
| 4ND | -0.922 | 1.000 | 1.602 | 0.765 | - | - | - | - | - | - |
| 5D | -1.281 | 1.000 | -0.389 | 1.000 | - | - | - | - | - | - |
| 6ND | -0.474 | 1.000 | 0.942 | 1.000 | - | - | - | - | - | - |
| Auditory | | | | | | | | | | |
| 1D | 1.454 | 0.985 | - | - | - | - | - | - | - | - |
| 2ND | 1.212 | 1.000 | - | - | - | - | - | - | - | - |
| 3D | -0.170 | 1.000 | - | - | - | - | - | - | - | - |
| 4ND | -0.210 | 1.000 | - | - | - | - | - | - | - | - |

| | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| 5D | **3.218** | **0.030** | **-** | **-** | **-** | **-** | **-** | **-** | **-** | **-** |
| 6ND | 0.243 | 1.000 | - | - | - | - | - | - | - | - |

# APPENDIX IV

## Questionnaires

### 1. Fall history

| | |
|---|---|
| 1. **Did you experience a fall in the past 12 months?** | **Yes / No** |
| If yes: | |
| 2. How often did you fall? | ......... times |
| 3. What caused the fall? | ☐ Tripping<br>☐ Slipping<br>☐ A misplaced step<br>☐ Loosing balance<br>☐ A push<br>☐ I don’t know<br>☐ Other, namely…… |
| 4. What were you doing at the time of the fall? | |
| 5. Where were you when you fell? | ☐ At home, inside<br>☐ At home, outside<br>☐ Away from home, in a familiair environment<br>☐ Away from home, in an unfamiliair environment |
| 6. Did the fall cause any injury? | Yes / No<br><br>If yes, |

## 2. Falls efficacy scale

| How concerned are you about the possibility of falling when... | Not at all concerned | Somewhat concerned | Fairly concerned | Very concerned |
|---|---|---|---|---|
| ...cleaning the house (e.g. sweep, vacuum, dust) | | | | |
| ... getting dressed or undressed? | | | | |
| ...preparing simple meals? | | | | |
| ...taking a bath or shower? | | | | |
| ...going tot he shop? | | | | |
| ... getting in or out of a chair? | | | | |
| ... going up or down stairs? | | | | |
| ...walking around in the neighborhood? | | | | |
| ...reaching for something above your head or on the ground? | | | | |
| ...going to answer the telephone before it stops ringing? | | | | |

## 3. Physical activity score

| How often do you take part in activities which are… | Hardly ever or never | 1-3x a month | 1-2x a week | >3x a week |
|---|---|---|---|---|
| ... very energetic e.g. swimming, cycling or running? | | | | |
| ... moderately energetic e.g. cleaning the car, cleaning windows, scrubbing floors, walking or dancing? | | | | |
| ... mildly energetic e.g. vacuuming, makings beds or mopping the floor? | | | | |

# APPENDIX V

## Supplementary material: Experimental study 2 – Can responses to different gait perturbations discriminate between older adults with and without a history of falls?

**Table A-V.1** Steady state gait parameters for fallers and non-fallers. When normally distributed, independent t-tests were used to assess differences between groups. Non-normally distributed parameters were evaluated using Mann-Whitney U-tests ($^{\#}$).

| | Non-fallers | | | Fallers | | | |
|---|---|---|---|---|---|---|---|
| | Mean | ± | SD | Mean | ± | SD | *p* |
| **Mean** | | | | | | | |
| Step Time [s] | 0.538 | ± | 0.049 | 0.560 | ± | 0.085 | 0.582$^{\#}$ |
| Step Length [m] | 0.543 | ± | 0.073 | 0.533 | ± | 0.103 | 0.753 |
| Step Width [m] | 0.141 | ± | 0.040 | 0.130 | ± | 0.053 | 0.538 |
| MoS – ML [m] | 0.058 | ± | 0.015 | 0.049 | ± | 0.020 | 0.087 |
| MoS – BW [m] | 0.130 | ± | 0.042 | 0.110 | ± | 0.050 | 0.189 |
| Trunk Vel – ML [m/s] | 0.203 | ± | 0.044 | 0.201 | ± | 0.050 | 0.874$^{\#}$ |
| Trunk Vel – VT [m/s] | 0.162 | ± | 0.041 | 0.166 | ± | 0.075 | 0.971$^{\#}$ |
| Trunk Vel – AP [m/s] | 0.081 | ± | 0.026 | 0.092 | ± | 0.042 | 0.423 |
| Trunk Vel – F [deg/s] | 41.381 | ± | 13.385 | 42.566 | ± | 13.027 | 0.779$^{\#}$ |
| Trunk Vel – T [deg/s] | 27.433 | ± | 10.482 | 23.076 | ± | 8.320 | 0.257 |
| Trunk Vel – S [deg/s] | 15.032 | ± | 4.152 | 13.408 | ± | 3.650 | 0.205$^{\#}$ |
| **Variability** | | | | | | | |
| Step Time [s] | 0.017 | ± | 0.006 | 0.020 | ± | 0.012 | 0.760$^{\#}$ |
| Step Length [m] | 0.033 | ± | 0.012 | 0.030 | ± | 0.011 | 0.440$^{\#}$ |
| Step Width [m] | 0.022 | ± | 0.007 | 0.023 | ± | 0.006 | 0.517$^{\#}$ |
| MoS – ML [m] | 0.016 | ± | 0.006 | 0.017 | ± | 0.005 | 0.800 |
| MoS – BW [m] | 0.021 | ± | 0.006 | 0.023 | ± | 0.006 | 0.364 |
| Trunk Vel – ML [m/s] | 0.031 | ± | 0.007 | 0.034 | ± | 0.007 | 0.979 |
| Trunk Vel – VT [m/s] | 0.027 | ± | 0.011 | 0.023 | ± | 0.004 | 0.412$^{\#}$ |
| Trunk Vel – AP [m/s] | 0.045 | ± | 0.017 | 0.043 | ± | 0.012 | 0.932$^{\#}$ |
| Trunk Vel – F [deg/s] | 7.824 | ± | 3.076 | 7.395 | ± | 1.801 | 0.951$^{\#}$ |
| Trunk Vel – T [deg/s] | 4.758 | ± | 1.657 | 3.909 | ± | 1.464 | 0.100$^{\#}$ |
| Trunk Vel – S [deg/s] | 4.740 | ± | 2.097 | 4.713 | ± | 1.963 | 1.000$^{\#}$ |
| **Local dynamic stability** | | | | | | | |
| Trunk Vel – ML | 3.121 | ± | 0.170 | 2.853 | ± | 0.515 | 0.057$^{\#}$ |
| Trunk Vel – VT | 2.486 | ± | 0.212 | 2.402 | ± | 0.159 | 0.264 |
| Trunk Vel – AP | 3.167 | ± | 0.182 | 2.934 | ± | 0.388 | 0.089$^{\#}$ |
| Trunk Vel – F | 1.965 | ± | 0.194 | 1.865 | ± | 0.224 | 0.152 |
| Trunk Vel – T | 1.683 | ± | 0.263 | 1.674 | ± | 0.241 | 0.892 |
| Trunk Vel – S | 1.782 | ± | 0.237 | 1.723 | ± | 0.124 | 0.261 |

**Table A-V.2** Steady state gait versus recovery steps following sway perturbations. Significant differences as indicated by the paired samples *t*-tests are printed in bold.

| | Step 1 | | Step 2 | | Step 3 | | Step 4 | | Step 5 | | Step 6 | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* |
| Ipsilateral sway | | | | | | | | | | | | |
| Step Time | **13.113** | **<0.001** | **9.348** | **<0.001** | **4.988** | **<0.001** | **5.598** | **<0.001** | **3.528** | **0.004** | 2.590 | 0.051 |
| Step Length | **15.791** | **<0.001** | **4.706** | **<0.001** | -0.927 | 1.000 | 1.103 | 1.000 | **3.959** | **<0.001** | **4.842** | **<0.001** |
| Step Width | **-9.440** | **<0.001** | **-11.192** | **<0.001** | **-4.392** | **<0.001** | **-4.792** | **<0.001** | **-5.795** | **<0.001** | **-4.636** | **<0.001** |
| MoS ML | -1.966 | 0.220 | **-9.914** | **<0.001** | **-3.791** | **0.002** | **-5.245** | **<0.001** | **-4.533** | **<0.001** | **-5.484** | **<0.001** |
| MoS BW | -1.337 | 0.750 | **-11.431** | **<0.001** | **-9.102** | **<0.001** | **-7.176** | **<0.001** | **-4.783** | **<0.001** | -1.719 | 0.368 |
| Trunk Vel ML | **-5.542** | **<0.001** | -1.532 | 0.529 | **-11.745** | **<0.001** | -2.584 | 0.051 | **-5.513** | **<0.001** | **-7.794** | **<0.001** |
| Trunk Vel VT | -1.532 | 1.000 | 1.992 | 0.208 | **-2.647** | **0.044** | **-3.062** | **0.014** | **-4.102** | **<0.001** | 0.263 | 1.000 |
| Trunk Vel AP | 1.992 | 0.313 | 2.462 | 0.070 | -0.109 | 1.000 | -2.561 | 0.055 | 1.092 | 1.000 | **3.133** | **0.012** |
| Trunk Vel F | **2.462** | **0.002** | 1.764 | 0.336 | **2.942** | **0.020** | -0.089 | 1.000 | -1.006 | 1.000 | -1.380 | 0.696 |
| Trunk Vel T | **1.764** | **<0.001** | -0.363 | 1.000 | -2.423 | 0.077 | **-3.327** | **0.007** | -1.474 | 0.589 | -1.950 | 0.228 |
| Trunk Vel S | -0.363 | 1.000 | -6.625 | <0.001 | **-4.228** | **<0.001** | -1.259 | 0.856 | -0.518 | 1.000 | -0.530 | 1.000 |
| Contralateral sway | | | | | | | | | | | | |
| Step Time | **5.766** | **<0.001** | **4.979** | **<0.001** | **6.286** | **<0.001** | **6.260** | **<0.001** | **4.763** | **<0.001** | **3.208** | **0.010** |
| Step Length | **9.935** | **<0.001** | **3.629** | **0.003** | **6.006** | **<0.001** | **4.265** | **<0.001** | **6.802** | **<0.001** | **5.946** | **<0.001** |
| Step Width | **9.538** | **<0.001** | 2.415 | 0.078 | **-12.360** | **<0.001** | **-10.718** | **<0.001** | **-8.649** | **<0.001** | **-8.747** | **<0.001** |
| MoS ML | **21.300** | **<0.001** | **5.451** | **<0.001** | **-9.363** | **<0.001** | **-8.860** | **<0.001** | **-8.424** | **<0.001** | **-6.784** | **<0.001** |
| MoS BW | -1.608 | 0.457 | **-6.028** | **<0.001** | **-8.777** | **<0.001** | **-8.654** | **<0.001** | **-4.652** | **<0.001** | **-3.248** | **0.009** |
| Trunk Vel ML | **-22.202** | **<0.001** | **4.283** | **<0.001** | **6.597** | **<0.001** | **-11.979** | **<0.001** | **-5.776** | **<0.001** | **-9.570** | **<0.001** |
| Trunk Vel VT | -1.816 | 0.302 | **5.685** | **<0.001** | -0.669 | 1.000 | -1.428 | 0.639 | -2.186 | 0.135 | -1.260 | 0.855 |
| Trunk Vel AP | **-3.162** | **0.011** | -1.811 | 0.306 | -1.533 | 0.528 | **4.364** | **<0.001** | **7.409** | **<0.001** | **9.901** | **<0.001** |
| Trunk Vel F | **-5.686** | **<0.001** | -0.117 | 1.000 | 2.254 | 0.115 | **3.310** | **0.007** | 0.678 | 1.000 | -0.767 | 1.000 |
| Trunk Vel T | **-10.310** | **<0.001** | **-8.276** | **<0.001** | -0.795 | 1.000 | -1.269 | 0.843 | 0.553 | 1.000 | 0.158 | 1.000 |
| Trunk Vel S | **-4.678** | **<0.001** | **-8.491** | **<0.001** | **-4.004** | **<0.001** | **-2.964** | **0.019** | 1.259 | 0.856 | 1.705 | 0.379 |

**Table A-V.3** Steady state gait versus recovery steps following belt perturbations. Significant differences as indicated by the paired samples *t*-tests are printed in bold.

| | Step 1 | | Step 2 | | Step 3 | | Step 4 | | Step 5 | | Step 6 | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* | *t* | *p* |
| **Acceleration** | | | | | | | | | | | | |
| Step Time | **12.238** | **<0.001** | **11.605** | **<0.001** | 0.515 | 1.000 | 0.922 | 1.000 | 3.355 | 0.006 | 2.092 | 0.167 |
| Step Length | **-6.217** | **<0.001** | **3.460** | **0.005** | **5.128** | **<0.001** | 0.661 | 1.000 | **4.303** | **<0.001** | 1.739 | 0.354 |
| Step Width | **5.937** | **<0.001** | **-4.290** | **<0.001** | **-11.462** | **<0.001** | **-2.836** | **0.027** | **-6.464** | **<0.001** | **-5.147** | **<0.001** |
| MoS ML | **-4.468** | **<0.001** | -2.470 | 0.068 | **-7.244** | **<0.001** | **-3.825** | **0.002** | **-6.267** | **<0.001** | **-4.533** | **<0.001** |
| MoS BW | -1.110 | 1.000 | **-11.343** | **<0.001** | 0.628 | 1.000 | -0.094 | 1.000 | -1.828 | 0.295 | -2.083 | 0.171 |
| Trunk Vel ML | **-8.757** | **<0.001** | **3.360** | **0.006** | **-5.136** | **<0.001** | **-15.402** | **<0.001** | -1.463 | 0.600 | **-8.731** | **<0.001** |
| Trunk Vel VT | 0.046 | 1.000 | **-5.651** | **<0.001** | -2.137 | 0.151 | 0.569 | 1.000 | -2.558 | 0.055 | -1.123 | 1.000 |
| Trunk Vel AP | **-17.952** | **<0.001** | **-2.699** | **0.038** | 0.729 | 1.000 | **2.958** | **0.019** | 0.024 | 1.000 | 2.038 | 0.188 |
| Trunk Vel F | **-4.690** | **<0.001** | -1.351 | 0.732 | -2.248 | 0.117 | **-2.744** | **0.034** | **-3.511** | **0.004** | **-3.789** | **0.002** |
| Trunk Vel T | **-9.641** | **<0.001** | **-8.465** | **<0.001** | -1.276 | 0.832 | **-3.017** | **0.016** | -2.202 | 0.130 | -2.039 | 0.188 |
| Trunk Vel S | **-5.851** | **<0.001** | **-12.858** | **<0.001** | **-5.527** | **<0.001** | -1.892 | 0.258 | -1.445 | 0.620 | -0.065 | 1.000 |
| **Deceleration** | | | | | | | | | | | | |
| Step Time | **13.701** | **<0.001** | **10.702** | **<0.001** | **9.246** | **<0.001** | **8.238** | **<0.001** | **6.260** | **<0.001** | **4.798** | **<0.001** |
| Step Length | **25.497** | **<0.001** | **5.114** | **<0.001** | -0.713 | 1.000 | **3.253** | **0.008** | **3.716** | **0.002** | **6.908** | **<0.001** |
| Step Width | **3.043** | **0.015** | -0.081 | 1.000 | **-11.245** | **<0.001** | **-7.373** | **<0.001** | **-6.606** | **<0.001** | **-7.693** | **<0.001** |
| MoS ML | **-4.083** | **<0.001** | **-8.109** | **<0.001** | **-10.083** | **<0.001** | **-4.794** | **<0.001** | **-6.625** | **<0.001** | **-5.582** | **<0.001** |
| MoS BW | **32.933** | **<0.001** | **-15.478** | **<0.001** | **-17.559** | **<0.001** | **-13.372** | **<0.001** | **-9.847** | **<0.001** | **-6.260** | **<0.001** |
| Trunk Vel ML | **-7.250** | **<0.001** | **6.638** | **<0.001** | **-4.397** | **<0.001** | **-9.121** | **<0.001** | **-5.777** | **<0.001** | **-7.059** | **<0.001** |
| Trunk Vel VT | -0.773 | 1.000 | **3.722** | **0.002** | 0.223 | 1.000 | **-5.564** | **<0.001** | **-3.768** | **0.002** | -1.885 | 0.262 |
| Trunk Vel AP | -0.224 | 1.000 | **-3.146** | **0.011** | **-4.649** | **<0.001** | **-4.826** | **<0.001** | -0.945 | 1.000 | 2.068 | 0.176 |
| Trunk Vel F | **-4.555** | **<0.001** | **5.789** | **<0.001** | **6.414** | **<0.001** | -0.016 | 1.000 | -1.930 | 0.238 | **-2.606** | **0.049** |
| Trunk Vel T | **-3.510** | **0.004** | **-3.282** | **0.008** | **-4.153** | **<0.001** | **-3.888** | **0.001** | **-3.640** | **0.003** | -2.426 | 0.076 |
| Trunk Vel S | **-7.907** | **<0.001** | **-8.704** | **<0.001** | **-8.203** | **<0.001** | **-5.331** | **<0.001** | -1.830 | 0.294 | -1.483 | 0.579 |

# APPENDIX VI

## Index of electronic appendices

| Filename | Description |
|---|---|
| trigger1.lua | Lua code to random triggering of the perturbation onset |
| perturbation1.lua | Lua code for the perturbation profile |
| trigger2.lua | Lua code to random triggering of the perturbation onset |
| perturbation2.lua | Lua code for random selection of the perturbation type and its corresponding perturbation profile |
| detect_gait_events.m | Matlab code for intial contact and foot off detection |
| calc_spat_temp.m | Matlab code for calculation of spatio-temporal parameters |
| normalize_per_cycle.m | Matlab code to normalize time series to 101 samples per gait cycle |
| get_kinematics.m | Matlab code to calculate peak trunk velocity |
| calc_mos.m | Matlab code to calculate margins of stability |
| calc_baseline_discr.m | Matlab code to calculate average discrete baseline parameters |
| calc_baseline_cont.m | Matlab code to calculate average continuous baseline parameters |
| handle_lds.m | Matlab code to calculte local dynamic stability using normalize_per_trial.m, create_state_space.m, find_nearest_neighbor.m and calc_divergence.m |
| normalize_per_trial.m | Matlab code to normalize time series to 100xn strides |
| create_state_space.m | Matlab code to create state space |
| find_nearest_neighbor.m | Matlab code to find nearest neighbor trajectories |
| calc_divergence.m | Matlab code to calculate divergence between nearest neigbor trajectories |
| calc_pert_discr.m | Matlab code to calculate discrete parameters prior to and following the perturbation onsets |
| calc_pert_cont.m | Matlab code to calculate continuous parameters prior to and following the perturbation onsets |