Effects of Fear and Attention on Human Balance Control

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By

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Thesis Summary

A fall is one of the main causes of injury-related hospitalisation and injury-related deaths. Besides physical degeneration, fear of falling and attentional focus strategies are related to fall risk and decline of balance performance. The aim of this research was to expose the mechanisms by which fear of falling and attentional focus affect human balance control. We used galvanic vestibular stimulation (GVS) to induce vestibular balance reflexes while participants stood at ground level and on a narrow walkway at 3.85 m height to induce fear of falling. Using questionnaires and skin conductance measurements, a fear of falling at height was confirmed. Full-body kinematics was collected to measure the vestibular balance response. We concluded that fear modifies vestibular balance control and proposed a mechanism in which both the short- and medium-latency reflexes functionally contribute to whole body balance. Furthermore, the literature suggests that fear of falling could impair balance mechanisms in elderly through changes in attentional focus. Therefore, we also investigated the effect of attentional focus (internal vs. external focus and reinvestment) and fall history on walking stability in healthy older adults. Participants' gait was perturbed through randomly occurring unilateral treadmill decelerations to evoke balance recovery movements. Using full body kinematics, coefficients of variation of spatiotemporal gait parameters and local divergence exponents were calculated to assess gait performance of balance recovery responses and unperturbed gait. Fallers showed increased gait variability and decreased gait stability, however no effects of attentional focus were found. The benefits of an external focus of attention on motor performance do not seem to apply to gait in elderly. Continued investigation into attentional focus effects and fear of falling on gait including holistic and partial internal focus and continuous gait perturbations, might further clarify the relations between fear of falling and attentional focus and how they could affect fall risk. Follow-up studies with clinical subgroups could further clarify the relation between fear of falling, attentional focus and balance performance.

Conferences and Publications

Conference presentations

International Society of Posture and Gait Research (Sevilla 2015)

Oral: Full Body Kinematic Analysis of Altered Vestibular Reflexes Caused by Postural Threat. Abstract reference O.2.3.

Poster: Spatiotemporal Gait parameters in Elderly are unaffected by internal or external attention. Abstract reference P3-O-72.

International Society of Biomechanics (Glasgow 2015)

Nominated for "the David Winter Young Investigator Award".

Oral: The Effect Of Fear of Falling and Vestibular Sensory Feedback on Postural Control Kinematics. Abstract reference AS-0246.

Poster: Effects of Internal and External Attention on Gait in Elderly. Abstract reference PO-0147.

European Society for the Cognitive Sciences of Music (Manchester 2015)

Oral: The Effect of Fear on Human Motor Performance. Abstract reference 4A.

Publications

de Melker Worms J L A, Stins J F, Beek P J, Loram I D. The Effect of Fear of Falling on Vestibular Feedback Control of Balance, *Manuscript submitted for publication*.

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Chapter 1

Introduction

1.1 Move-Age programme

The work in this thesis was part of the Move-Age joint doctorate programme, which is funded by the European Commission as part of the Erasmus Mundus programme. This PhD project is a collaboration between Manchester Metropolitan University and Vrije Universiteit Amsterdam on the topic of fall prevention and mobility in elderly. The project was supported by existing expertise, personnel, development of techniques and lines of enquiry in both of the participating research groups (Prof. Dr. I. D. Loram from the School of healthcare science, MMU and Prof. Dr. P. J. Beek and Dr. J. F. Stins from MOVE research Institute Amsterdam, Vrije Universiteit Amsterdam).

1.2 Fall risk factors

The ageing population is confronted with the problem of mobility loss. Approximately one in three older adults will annually lose their balance and experience a fall, and approximately half of these individuals will experience more than one fall per year (Blake *et al.*, 1988; Tinetti *et al.*, 1988; Downton & Andrews, 1991). For older adults (age ≥ 65) falls are one of the main causes of injury-related hospitalisation and injury-related deaths (Rubenstein, 2006). This also results in a significant global economic cost (Stevens et al., 2006).

Consequently, a significant programme of research and body of literature is aimed at finding risk factors for falls. If elderly with a high propensity to fall can be identified, early interventions might be able to reduce the number of falls. Initially (1990 – 2002) this field of research was dominated by a physiological characteristics approach.

Impairment of vision, peripheral sensation, muscle strength, reaction time, and balance were all found to be risk factors for falls (Lord *et al.*, 1994a; Lord *et al.*, 1994b). In a 1-year prospective study with 341 women, discriminant function analysis with these risk factors differentiated elderly with multiple falls from non-multiple fallers within that year with 75% accuracy (Lord *et al.*, 1994b). In later studies the effects of interventions were investigated. These interventions included strength and balance training (Buchner *et al.*, 1997; Wolf *et al.*, 1997; Campbell *et al.*, 1999a; Day *et al.*, 2002), optimising vision (Day *et al.*, 2002), provision of pace makers to prevent drop attacks (Kenny *et al.*, 2001), environmental modifications to increase safety in the home (Cumming *et al.*, 1999; Day *et al.*, 2002; Nikolaus & Bach, 2003) and reduction of hazardous medication use (Campbell *et al.*, 1999b). These intervention studies showed improvement for the most vulnerable and high-risk groups. The type and level of frailty were found to be important factors to determine what interventions are suitable for risk prevention. Interventions that reduced fall occurrence in a broader population included physiotherapy (Campbell *et al.*, 1999a; Robertson *et al.*, 2001), group exercise (Day *et al.*, 2002; Barnett *et al.*, 2003; Lord *et al.*, 2003) and multifactorial interventions (Tinetti *et al.*, 1994a; Close *et al.*, 1999).

This physiological approach has advanced our understanding of fall risk. However, subsequent multifactorial fall research has exposed a broader range of fall risk factors (Delbaere *et al.*, 2010a). Cognitive factors such as executive function (Anstey *et al.*, 2009; Delbaere *et al.*, 2010a) and attentional focus (Wong *et al.*, 2008; Wulf, 2013) were also found to be related to balance performance and falls. Executive function is defined as the ability to independently perform complex, goal-directed, and self-serving behaviours (Delbaere *et al.*, 2010a) and is mediated by processes of selection and reinforcement learning operating through frontal basal ganglia networks (D'Esposito *et al.*, 1995; Houk *et al.*, 2007; Cohen & Frank, 2009).

Other factors such as fear of falling and balance confidence showed a strong relation with balance and falls as well (Hadjistavropoulos *et al.*, 2007; Delbaere *et al.*, 2010b). In particular for fear of falling and attentional focus, the mechanisms subserving the relation with balance control and fall risk are not yet clearly identified. From a cognitive motor control perspective, fear is a response that follows when the central nervous system classifies the environment as requiring a fear response (LeDoux, 1998). Fear of falling is selected following perception of the situation and can be reinforced within a vicious cycle of positive feedback leading to reduced mobility. Alternatively, fear of falling can be progressively diminished leading to increased mobility (Loram, 2015). The ability to accurately assess whether an environmental context is potentially threatening is dependent upon executive function which allows one to adapt rationally to the environment by combining sensory analysis with selective inhibition to diminish unnecessary fearful responses (Loram, 2015).

In this chapter we therefore explore the mechanisms by which perceptual context influences balance. Before we examine these mechanisms we will first elucidate the concept and assessment methods of fear of falling and balance confidence, and their relation with balance control.

1.3 Fear of falling

Following a fall, elderly may lose confidence in their ability to balance, and develop a fear of falling. Fear of falling has been observed in 50% - 60% of reported fallers in multiple community samples (Legters, 2002). Avoidance of physical activity has been acknowledged by 25% - 33% of these fearful individuals (Legters, 2002). This reduction of physical activities may lead to (more) health problems and loss of independence (Vellas *et al.*, 1997). However in many seniors without a history of falls or related injuries, fear of falling has been established as well (Legters, 2002). Furthermore, fear of falling and lowered balance confidence have shown to be predictive of future falls (Cumming *et al.*, 2000; Delbaere *et al.*, 2004; Hadjistavropoulos *et al.*, 2007).

To relate fear of falling and balance confidence to balance performance and fall risk, appropriate measurements and clear conceptualisations are needed. This could lead to the development of new intervention strategies to enhance balance and perhaps reduce risk of falls.

1.3.1 Effect of fear on sensorimotor control

The issue as to how fear could impair balance performance has often been addressed against the backdrop of Bernstein's degrees of freedom problem (Bernstein, 1967; Higuchi *et al.*, 2002). This problem is based on the argument that actors have multiple ways to perform a movement to achieve the same goal, because of the extreme abundance of degrees of freedom in our movement system. In terms of kinematic degrees of freedom, moving body segments can display different trajectories and velocities to achieve the same goal. In terms of degrees of freedom in muscular activation one could identify different muscle activation patterns that produce the same movement output. It could be the case that under stressful situations the burden of concurrently coordinating all degrees of freedom becomes too demanding for our nervous system. As a result, certain degrees of freedom are frozen in stressful situations thereby facilitating control. Therefore, movement becomes more constrained when anxiety increases (Higuchi *et al.*, 2002) and efficient balance performance could be jeopardized.

Arousal accompanied by fear could also lead to aberrant movement patterns (Heckman *et al.*, 2008). Through persistent inward currents in spinal motor neurons, noradrenaline increases the global excitability of the muscles (Heckman *et al.*, 2008). This might enhance levels of co-contraction of antagonistic muscles within the same joint, which in turn increases joint stiffness. Therefore, instead of freezing certain degrees of freedom, fear stimulates our nervous system to generate a general over-excitation of the entire system resulting in stiffening of our joints. Therefore, fear-induced muscle excitation is non-specific. However inhibition of excitation acts through specific localized reciprocal inhibition (Hyngstrom *et al.*, 2008).

1.3.2 Measurement of fear of falling

To assess the presence of fear or anxiety, three components can be distinguished, (1) physiological (e.g., increased autonomic reactivity), (2) behavioural (e.g., cautious and slow gait) and (3) cognitive (subjective estimation of the level of danger and ability to avoid a fall) (Rachman, 1982). Fear responses have shown to be accompanied by increased arousal (Critchley, 2002). Therefore many authors have focussed on physiological arousal to investigate the physiological anxiety component, for example by measuring skin conductance (SC) using two electrodes placed on the hand palm or fingers of a subject (Critchley, 2002; Davis *et al.*, 2009). Additionally, the vocal fundamental frequency has been used to grade the level of anxiety (Weeks *et al.*, 2012). Kinetics (e.g. ground reaction forces) (Carpenter *et al.*, 1999; Carpenter *et al.*, 2001; Laufer *et al.*, 2006; Davis *et al.*, 2009) and kinematics (e.g. 3d motion capture) (Hsu *et al.*, 2007; Park *et al.*, 2012) have been analysed to assess the behavioural aspects of the fear response. With respect to the cognitive component, various self-evaluation questionnaires have been implemented.

With respect to fear of falling the simplest assessments have been limited to 'yes' or 'no', or graded scale answers to the question: "Are you afraid of falling", whereas the Survey of Activities and Fear of Falling in the Elderly (SAFE) assesses fear of falling in elderly and provides an index for activity avoidance due to fear (Jorstad *et al.*, 2005). The two parts of the State-Trait Anxiety Index (STAI) are more general self-evaluation questionnaires of anxiety (Gros *et al.*, 2007). One part measures the time specific anxiety of a subject which fluctuates depending on the subject's current state. The second part aims to measure the more persistent levels of anxiety, related to one's personality profile. The STAI questionnaire only taps into the cognitive component of fear, whereas the SAFE aims to tap into the behaviour component as well. In addition, qualitative research has shown that fear of falling is often related to a fear of institutionalisation (e.g. highly dependent nursing homes) or fear of losing the ability to walk, e.g. having to use a wheel chair (Wright *et al.*, 1990).

Thus, a complete understanding of the fear response requires joint investigation of the physiological, behavioural and cognitive components. With respect to balance control, another important cognitive factor that is also related to fear of falling is balance confidence, as discussed in the next section.

1.3.3 Falls efficacy and fear of falling

Fear of falling is related to the level of confidence in one's own balancing skills. The Activity Balance Confidence scale (ABC), Falls Efficacy Scale (FES) and more recently the FES International (FES-I) have been used to measure balance confidence and falls efficacy in the elderly (Jorstad *et al.*, 2005; Hadjistavropoulos *et al.*, 2007; Delbaere *et al.*, 2010b). Falls efficacy refers to beliefs in balancing skills. The relation between fear and beliefs about one's own ability is now well-established (Barlow, 2008). Correlations as high as 0.86 were found between FES and ABC scores (Hotchkiss *et al.*, 2004). As such, the terms 'balance confidence' and 'falls efficacy' were considered to be interchangeable (Hadjistavropoulos *et al.*, 2011). In a longitudinal study with community dwelling older adults, fear of falling and fall-efficacy were also found to be correlated (Hadjistavropoulos *et al.*, 2007).

Therefore elderly with increased fear of falling are likely to have low balance confidence as well. However, Butki *et al.* (2001) found no association between state anxiety and falls-related self-efficacy. Therefore, fear of falling and balance confidence (falls efficacy) are still argued to be distinct dimensions (Moore & Ellis, 2008; Hadjistavropoulos *et al.*, 2011). Furthermore, the FES-I was found to be a predictor for falls (Delbaere *et al.*, 2010b) and falls efficacy (ABC, FES) was also found to be a better predictor for falls than fear of falling (SAFE) (Hadjistavropoulos *et al.*, 2007). As such, one could argue that falls efficacy mediates the relationship between fear of falling and the occurrence of falls.

1.3.4 How does fear of falling affect balance performance?

Fear of falling and falls efficacy are not only related to fall history (Lachman *et al.*, 1998; Fletcher & Hirdes, 2004), but also to future falls (Cumming *et al.*, 2000; Delbaere *et al.*, 2004; Hadjistavropoulos *et al.*, 2007). However, no consensus is established yet on the mechanisms that cause this relation.

Many authors assume that fear of falling induces activity avoidance, which in turn results in decline of balance performance, and thereby increases fall risk. Even though this mechanism is widely accepted, there is no clear evidence for this mechanism. The association between activity avoidance, and fear of falling and falls efficacy seems to be well established (Tinetti *et al.*, 1994b; Petrella *et al.*, 2000; Li *et al.*, 2003; Jorstad *et al.*, 2005; Delbaere *et al.*, 2009), however determining the direction of causality remains problematic. Additionally, a more recent study did not support this relation as they did not find a reduction in planned exercise for elderly with increased concern about falling (Delbaere *et al.*, 2016).

One might also question whether activity avoidance by itself predicts falls. The relation between activity avoidance and falls is undisputed for high levels of activity avoidance, as the adverse effects on balance performance and mobility are evident. Insufficient exercise could increase muscle atrophy, the risk for obesity, neuropathy and other factors that reduce mobility (Balducci *et al.*, 2006; Seguin *et al.*, 2012).

It may therefore come as a surprise that the literature on the relation between avoidance and falls is inconsistent. A weak relation between falls and avoidance was found by Delbaere *et al.* (2004). However a 6-month prospective study with 492 community-based adults found that activity avoidance did not predict falls, whereas falls efficacy and to a lesser extent fear of falling did predict falls (Hadjistavropoulos *et al.*, 2007). As such, no clear evidence exists that activity avoidance is a necessary component for fear of falling to cause fall risk.

Consequently, two different theories were proposed that did not include activity avoidance (Hadjistavropoulos *et al.*, 2007). First, fear of falling in elderly could be the result of an accurate self-appraisal of balancing abilities and fall risk. In a review on this topic it was concluded that this possibility has not been studied adequately (Hadjistavropoulos *et al.*, 2011). However, this issue was covered by Delbaere *et al.* (2016) and they found no support for the theory that fear of falling represents realistic appraisal of balance performance. For participants with high concerns for falls and good balancing abilities, a high fear of falling was still related to future falls. This association was mediated by other psychological/social factors such as depression, community participation, and physical activity.

Height-induced fear of falling directly impairs balance

Apart from the first possibility that elderly fear of falling constitutes a realistic appraisal of balancing abilities, an alternative theory states that fear of falling might directly impair balance performance. In support of the latter theory, Delbaere *et al.* (2006) found reduced dynamic balance performance in elderly with inappropriate high levels of fear, based on the number of previous and prospective falls. Elderly with inappropriately low fear overestimated their balance capacities.

However, most evidence for the theory that fear directly impairs balance performance was found using height-induced postural threat to elicit fear of falling. A frequently used paradigm involves positioning participants on the edge of an elevated platform at different heights to elicit a height-induced fear of falling (Carpenter *et al.*, 1999; Adkin *et al.*, 2000; Carpenter *et al.*, 2001; Carpenter *et al.*, 2004; Laufer *et al.*, 2006; Davis *et al.*, 2009; Horlings *et al.*, 2009; Huffman *et al.*, 2009). Using ground reaction forces (GRF), centre of pressure (COP) excursion data were analysed to assess balancing behaviour of participants. Carpenter et al. found that postural threat induced a tighter control of upright posture, reflecting a 'stiffening' strategy

(Carpenter *et al.*, 1999; Carpenter *et al.*, 2001; Carpenter *et al.*, 2004). When exposed to postural threat by standing quietly on the edge of an elevated platform, participants had a decreased mean sway amplitude of the COP, calculated as the offset removed root mean square (RMS). In addition, a higher COP mean power frequency (MPF) was found for participants standing at high compared to low elevation. Compared to young healthy adults, elderly were found to show an exaggerated response to postural threat that involved a larger decrease in RMS and larger increase in MPF (Carpenter *et al.*, 2006; Laufer *et al.*, 2006).

In a subsequent study, the effect of postural threat on participants with low vs. high levels of self-reported fear of falling was compared in young healthy adults (Davis *et al.*, 2009). The postural response of the non-fearful group showed the expected postural patterns (decrease in RMS and increase in MPF) with increased elevation. Conversely, the fearful group showed increased RMS and increased MPF compared to the ground condition. This fearful response for the fearful group indicates that postural threat induces a similar effect of increased frequency of corrective movements. However the increased RMS also indicates an increase instead of a decrease in sway amplitude of COP for the fearful group. Therefore, fear of falling is directly related to hampering regulation of postural sway at height. However, the direction of causality is undetermined, as it is unclear whether fear affected balance control, or whether the altered balance control caused the fear. This also relates to the old James-Lang vs. Canon-Bard discussion on the origin of emotion and the entangled physiological reactions (Cannon, 1987). Nevertheless, Davis *et al.* (2009) concluded that fearful subjects adopt a different control strategy than non-fearful subjects. However, no changes in self-reported state-anxiety or physiological arousal (SC) were found between the two groups.

While standing at height the depth of vision is larger than standing at ground level and this has shown to destabilise balance (Simeonov *et al.*, 2005).

Therefore it was studied whether this disparity in visual feedback could be the main cause of impaired balance at height, instead of the knowledge of danger (Tersteeg *et al.*, 2012). In this experiment, participants walked on a narrow high walkway while sheets placed around the walkway at the same height blocked the sight of drop. The risk and knowledge of danger was retained with this setup. Compared to walking on the walkway without the sheets no difference was found in gait progression and double support duration. Compared to ground level walking these gait parameters as well as physiological arousal were significantly altered. Therefore the main cause of altered balance control and arousal by height-induced postural threat is the knowledge of risk and reckoning of danger, rather than the visual feedback needed for balance control.

In summary, it has been established that fear of falling could lead to decreased balance performance and increased fall risk, but this does not have to be mediated by activity avoidance. Balance performance can be acutely impaired by fear of falling and thus potentially increase fall risk.

1.4 Vestibular balancing reflexes

Balance performance is largely dependent on reflexes that are triggered by feedback from the vestibular organs. It is currently debated whether fear of falling could influence balance performance at the level of these vestibular reflexes (Horslen *et al.*, 2015a, b; Reynolds *et al.*, 2015a, b).

1.4.1 Inducing vestibular balance reflexes

To study vestibular reflexes, a frequently used method is binaural bipolar Galvanic Vestibular Stimulation (GVS) (Fitzpatrick *et al.*, 1994; Fitzpatrick & Day, 2004; Osler *et al.*, 2013; Horslen *et al.*, 2014). GVS is applied by placing electrodes behind the ears on the mastoid processes. A current applied to these electrodes stimulates the vestibular nerves changing information sent from the vestibular organs to the brain. This creates artificial vestibular feedback of lateral rotation, causing a reflexive counter leaning movement of the whole body in the opposite lateral direction, see Figure 1.1.

The artificial vestibular feedback induced by GVS has been specified in detail by Fitzpatrick and Day (2004). They found that binaural bipolar GVS evokes an afferent signal of angular velocity and angular acceleration about an axis in the sagittal plane, located between the vestibular organs directed backward and 18.8 degrees upward from Reid's line. Therefore, during normal upright standing when Reid's line is nearly horizontal, an afferent of roll rotation with a small yaw component is evoked.

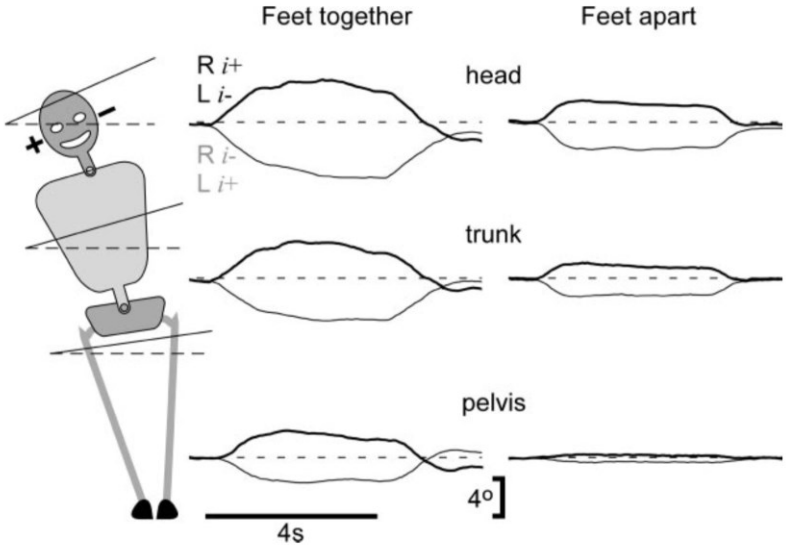

Figure 1.1: GVS induced body sway. The mediolateral GVS sway response is shown for the anode left and cathode right vs. anode right and cathode left configurations. Standing with the feet together increases the sway amplitude. Redrawn from Fitzpatrick and Day (2004).

The induced body sway is directed towards the anode GVS electrode. Therefore the direction of the balancing sway response depends on head orientation. When standing in a normal upright position with the anode electrode attached behind the right ear and the cathode electrode behind the left ear, the stimulation will induce a sway to the right. However with the head rotated 90 degrees to the left, the anode electrode is positioned on the anterior side with respect to the rest of the body. With this configuration, electrical stimulation causes anterior sway and the weight is shifted towards the toes. Therefore the GVS response is considered craniocentric (Lund & Broberg, 1983). Typically, square wave GVS intensities between 0.5 and 2 mA are used to elicit the balancing sway response (Britton *et al.*, 1993; Fitzpatrick *et al.*, 1994; Fitzpatrick & Day, 2004; Osler *et al.*, 2013).

To test how fear of falling could affect the latency and amplitude of the vestibular balance reflex, one needs to know what muscles and joints are involved and at what latency the balance response occurs. As such, GRF and EMG data of the GVS induced vestibular balance reflex has been collected. These measurements have revealed two phases of the GVS response; a short- and medium-latency response (Britton *et al.*, 1993; Fitzpatrick *et al.*, 1994; Fitzpatrick & Day, 2004). The lower limb short-latency EMG responses seem to cause a lateral shear GRF peak towards the cathode electrode side, whereas the medium-latency responses seem to cause an opposite anode directed GRF peak (Figure 1.2). This medium-latency GRF peak towards the anode implies an acceleration of the COM in the same direction. Therefore the medium-latency response is assumed to be responsible for the whole-body sway response to the anode side (Fitzpatrick *et al.*, 1994; Fitzpatrick & Day, 2004). However, the contribution of the short-latency response to balancing movements is still unknown (Fitzpatrick & Day, 2004; Horslen *et al.*, 2014). For EMG responses of shank muscles the onset of the short-latency responses ranged from 42 to 65 ms and for medium-latency from 98 to 120 ms post

GVS onset (Britton et al., 1993; Fitzpatrick et al., 1994; Ali et al., 2003; Fitzpatrick & Day, 2004; Son et al., 2008; Mian et al., 2010; Muise et al., 2012).

A problem with measurement of these GVS induced vestibular reflexes relates to the naturally occurring body sway when standing upright, which is the same order of magnitude as the GVS induced sway response. Therefore averaging over a large number of trials is needed for reliable measurement of the sway response. In addition, both polarity configurations (anode left and cathode right, vs. anode right and cathode left) should be used in randomised order.

A different method to induce vestibular reflexes is stochastic vestibular stimulation (SVS). With this method a large number of trials is not needed, therefore the time required for data collection is significantly shorter. However, no prominent body sway is produced. Instead of uni-directional discreet square wave GVS; continuous sine wave stimulation including both polarities is used with SVS. Coupling between the balance response (GRF and EMG data) and the SVS stimulation signal is determined using correlation measures for different time lags (cumulant density function). With this method, similar short- and medium-latency vestibular reflexes patterns were found in lower limb EMG and GRF data (Figure 1.2D). (Dakin et al., 2007; Dakin et al., 2010; Mian et al., 2010).

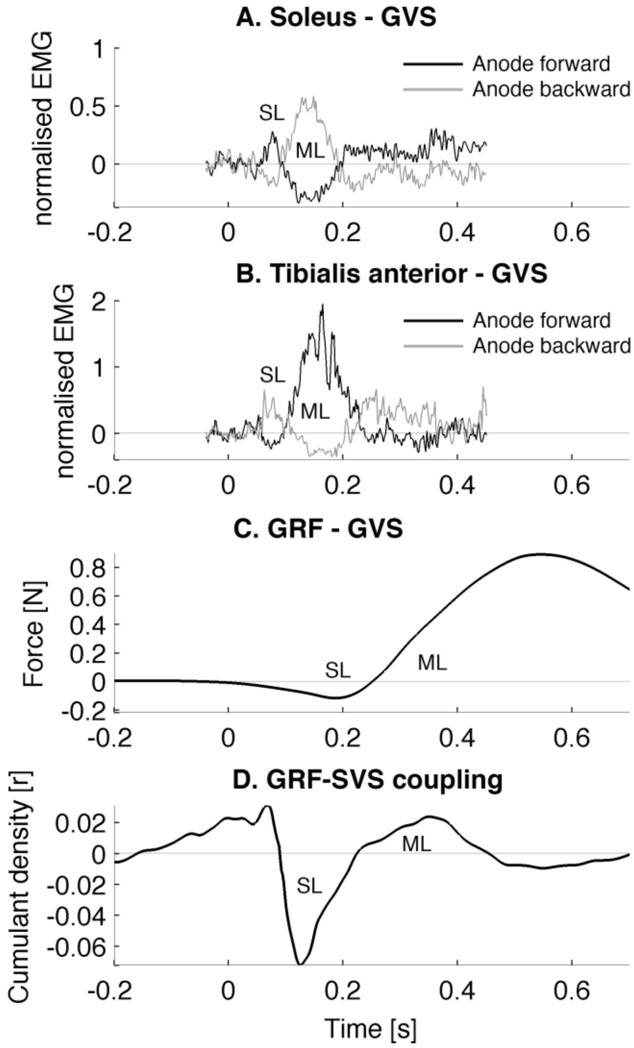

Figure 1.2: Short- and medium-latency vestibular balance response.

A&B, GVS stimulation starts at 0 s. The EMG response of the soleus and tibialis anterior are shown for two anode-cathode configurations of the GVS electrodes attached behind the participants' ears. The head was turned to the left so the anode faced either forward or backward, inducing a forward or backward body sway. In both muscles a reciprocal short- and medium-latency pattern of inhibition and activation was observed, depending on the GVS polarity. Only the medium-latency response would explain the observed whole body sway. Redrawn from Fitzpatrick and Day (2004), original data from Fitzpatrick *et al.* (1994).

C, GVS stimulation starts at 0 s and shear anteroposterior GRF is shown (in the direction of body sway, from cathode towards anode). After an electromechanical delay a comparable bi-phasic short- and medium-latency response pattern is observed, where again only the medium-latency response would explain the observed whole body sway towards anode. Redrawn form Marsden *et al.* (2005).

D, In this graph the coupling between shear GRF and continuous Stochastic Vestibular Stimulation (SVS) is shown for a range of time lags. The SVS frequency content was 2-25 Hz, excluding prominent body sway. This coupling was quantified using a cumulant density function, which revealed a similar short- and medium-latency response pattern. Participants stood with the head turned 90 degrees to the right. Redrawn from Horslen *et al.* (2014).

1.4.2 Effect of fear on vestibular balance reflexes

The debate as to whether fear of falling could influence balance performance at the level of vestibular reflexes has not yet been resolved (Horslen *et al.*, 2015a, b; Reynolds *et al.*, 2015a, b). These latter studies concerned a so-called cross-talk debate, which took place in the Journal of Physiology. That debate mainly revolved around the opposing conclusions of two studies that used a height-induced postural threat (standing on an elevated surface) to elicit a fear of falling, combined with GVS (Osler *et al.*, 2013; Horslen *et al.*, 2014).

Osler *et al.* (2013) used a narrow walkway elevated 3.85 m above ground level to induce postural threat. Applying GVS caused a lateral whole body sway in the direction of the edge of the walkway. Trunk and head kinematics showed that lateral sway amplitude after 800 ms was significantly and substantially attenuated at height compared to standing at ground level. However no difference was found between ground and height within the first 800 ms. Therefore it was concluded that fear of falling does not influence the faster vestibular balancing reflexes. Hence, fear of falling would not affect early reflexive balance control and would only interfere when volitional motor control influences balance as well.

Conversely, Horslen *et al.* (2014) did find effects of height-induced fear on vestibular reflexes. In that study SVS was used, and shear GRF data was collected instead of kinematics. They found an increased gain of both the short- and medium-latency vestibular balance reflexes at height. As such, a fear of falling would affect this fast reflexive balance control before volitional motor control kicked in.

In the crosstalk debate on this topic the functional implication of these increased short- and medium-latency GRF responses on balance was questioned (Reynolds *et al.*, 2015a). For kinematic data of the trunk and head GVS response, no difference was found within the first 800 ms between ground and height conditions (Osler *et al.*, 2013). Therefore it was argued that the increased short- and medium-latency responses might not be functionally contributing to balancing movements.

1.4.3 Function of short- and medium-latency responses

In the literature on the vestibular balancing reflex, the medium-latency response induced by GVS is assumed to cause the whole body sway. However it is unclear how the short-latency response contributes to balance control.

Cathers *et al.* (2005) proposed that the short-latency response originated from a different part of the vestibular organs than the medium-latency response, namely the otoliths instead of the semi-circular canals. However subsequent research did not support this possibility (Mian *et al.*, 2010).

Multiple studies supported a possible difference between the short- and medium-latency response in their contribution to balance. In two of them GVS was applied to standing participants with the neck flexed 90 degrees, so the head was facing downward (Cathers *et al.*, 2005; Mian *et al.*, 2010). In this posture, the axis of GVS induced illusory rotation is vertical instead of horizontal and the sway response (measured at the pelvis) towards the anode was abolished (Cathers *et al.*, 2005). Lower limb EMG data also showed an abolished (Cathers *et al.*, 2005) or attenuated medium-latency response (Mian *et al.*, 2010), however the short-latency response was unaffected compared to normal upright standing.

Other studies found further disparity between the short- and medium-latency EMG responses, as the short-latency stimulus threshold was higher (Fitzpatrick *et al.*, 1994) and the short-latency response amplitude seemed to reduce with ageing (Welgampola & Colebatch, 2002). The short-latency response was also attenuated for longer GVS onset rise times whereas the medium-latency response was not, and the bandwidths of coherence between SVS and EMG were different (Dakin *et al.*, 2007). Therefore one might argue that both responses have different neural underpinnings. However, both responses are craniocentric (dependent on head angle) and both responses in the legs are abolished when the participant is seated (Britton *et al.*, 1993; Fitzpatrick *et al.*, 1994). Hence the relative contribution of short- and medium-latency responses to balance remains to be determined.

Measurements of full body kinematics might shed more light on this issue. According to Newton's second law of motion the GRF is equal to the mass of the body multiplied by the acceleration of the centre of mass. Therefore the short- and medium-latency GRF responses that were affected by fear of falling should also be found in the acceleration of the centre of mass (COM). As the short-latency response was not found in kinematics data of the head and trunk, this response should be part of the acceleration responses of body parts other than the trunk and head.

Full body kinematics measurements of the effect of fear of falling on vestibular evoked reflexes could uncover the complete movement pattern of the GVS sway response. In addition, kinematic measurement of the short- and medium-latency responses could clarify their interplay and how they contribute to maintaining and restoring balance. This would also provide an answer to the question whether fear of falling modifies vestibular balance reflex movements or not.

1.5 Cognition mediates the effect of fear on motor control

In the field of motor control the effects of psychological state variables on motor performance has been studied extensively, specifically in relation to attentional focus. In normal healthy adults most movements are learned and executed with little attentional effort bypassing explicit volitional control.

1.5.1 Reinvestment

In challenging situations, e.g. when recovering from a fall or in fearful states, individuals may choose to consciously monitor their movements in an effort to enhance motor control (Wong *et al.*, 2008). This conscious control generally involves explicit knowledge or strategies processed in working memory. Explicit knowledge is knowledge that we are aware of and can be verbalized, as opposed to implicit knowledge that we cannot easily verbalize and that we are generally unaware of (Wong *et al.*, 2008). This process of shifting from an implicit and more automated form to a more conscious and explicit form of motor control has been termed reinvestment (Masters, 1992; Masters *et al.*, 1993). Reinvestment often occurs when an individual is fearful, highly motivated, under pressure, or has difficulty to move successfully (Wong *et al.*, 2008). A high predisposition to reinvest has been associated with e.g. disrupted performance under psychological pressure in sports (Masters *et al.*, 1993) and with diseases such as Parkinson's disease (Masters *et al.*, 2007).

To assess the level of reinvestment the Movement Specific Reinvestment Scale (MSRS) has been developed, and is now routinely used in scientific studies and clinical practice. Elderly with a history of falling have shown to score significantly higher on the MSRS than elderly non-fallers (Wong *et al.*, 2008). Therefore, fear of falling in elderly possibly induces reinvestment and thereby disrupts the automaticity of movements, which may in turn impair efficient balance control. Huffman et al. used a state specific version of the MSRS to study the effect of postural threat and fear of falling on reinvestment in young healthy adults (Huffman *et al.*, 2009). Subjects standing at the edge of an elevated surface 3.2 m above ground had a significantly higher fear of falling. Moreover, they scored significantly higher on the MSRS (Huffman *et al.*, 2009), which suggests that fear induced a change in cognitive strategies.

1.5.2 Motor control mechanisms of reinvestment

The reinvestment response could also be described in terms of sensorimotor control. For this model the relation between fear and impaired motor control could be described as part of an overall feedback loop in the central nervous system. This is a feedback loop of perception, selection and motor control as formulated by Loram (2015), see Figure 1.3. Perception requires sensory analysis, integrating all sensory modalities with prior experience. Acting through central pathways such as the basal ganglia loops, responses are selected. Recent evidence suggests selection converges to a serial process with maximum rate of 2-4 selections per second (refractory response planner) (Loram *et al.*, 2014). The motor system translates selected goals, actions, movements and control priorities into coordinated motor output. Within the slow feedback loop restricted to the voluntary bandwidth of control (2 Hz) the motor system generates coordinated motor responses sequentially from each new selection. With the fast loop restricted to a higher bandwidth (>10 Hz) acting through trans-cortical, brain stem and spinal pathways, the motor system uses selected parameters to modulate habitual-reflexive feedback (Loram *et al.*, 2011; van de Kamp *et al.*, 2013; Loram *et al.*, 2014).

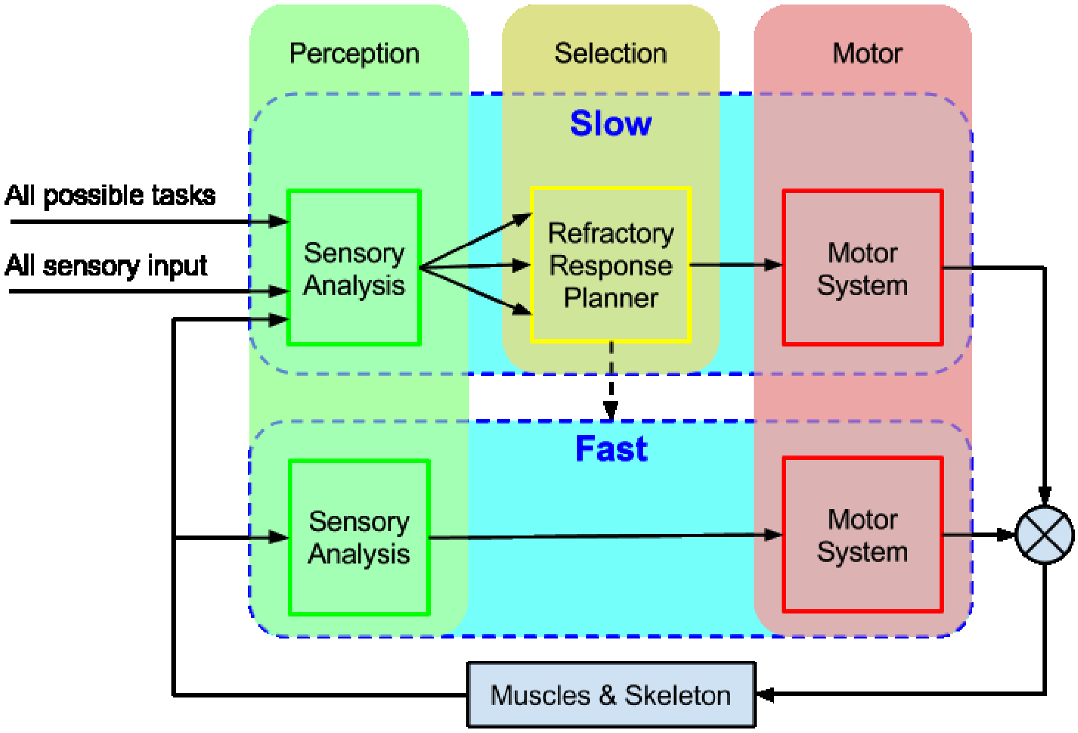

Figure 1.3: Sensorimotor model. Overall scheme of sensory-selection-motor integration. Adapted from Loram (2015).

For healthy adults, most daily life motor tasks are performed through the fast loop with little mental effort. However for elderly with a fear of falling, perception of the task could lead to a decreased confidence in motor control. In addition, it has been proposed that anxiety increases sensitivity to self-motion, through noradrenergic and serotonergic input to the vestibular nuclei (Balaban, 2002). This could increase attentional focus to self-movement. As a result, a strategy could be adopted where motor control is consciously monitored and/or evaluated using mainly the slow volitional loop. In other words, one rethinks the movement from scratch, and reinvestment occurs by shifting to the slow loop. This imposes a heavier load on "sensory analysis" as this area is now analysing the demands of the task and the machinery at its disposal. A resulting maladapted motor response might further undermine perception, creating a vicious cycle.

1.5.3 Internal and external focus of attention

Reinvestment is a possible explanation for the cause of balance impairment and increased fall risk in elderly. For this reason we might ask whether diverting attention away from our own body movements could temporarily enhance balance. The constrained action hypothesis formulated by Wulf and Prinz (2001b) states that an internal focus of attention interferes with automaticity by inducing a more conscious and explicit type of control. Conversely, an external focus of attention promotes a more automatic mode of control that employs more unconscious and implicit control processes. In balance tasks and various sports (e.g. swimming, basketball, golf, darts, volleyball, football and frisbee), enhanced performance was found for an external focus compared to an internal focus. A review by Wulf (2013) explores these beneficial effects of external focus on motor performance and motor learning in more detail.

Internal focus is thus defined as a focus of attention to the movement of one's own body, while external focus is related to the movement effect in the environment. For motor tasks where there is no external object movement to control, only movement of the body itself is involved, e.g. postural control on solid ground. For these tasks external focus instructions were used that direct the focus of attention to a physical surface in the environment on which force is exerted through muscle activity, and which is relevant to successful motor performance, e.g. the ground one is standing on in gymnastics (Lawrence *et al.*, 2011), postural control (Wulf *et al.*, 2007) and golf swing form assessment (An *et al.*, 2013). A limitation that is shared in all research on internal and external focus of attention is that it cannot be measured whether the participant is following the focus instructions or not.

Beneficial effects of external focus for balance

The effects of internal/external focus on postural control were only found for balancing tasks that were more challenging than standing on solid ground (Wulf *et al.*, 2007). These balancing tasks involved standing on an unstable surface, e.g. a stabilometer (balance board with mediolateral instability) (McNevin *et al.*, 2003) or an inflated rubber balance disk (Wulf *et al.*, 2007). With the stabilometer the angle of the balance board was measured and balance performance was measured as either RMS deviation from 0 degrees or as 'time in balance'. This time in balance was calculated as the time in which the balance board was within ±5 degrees deviation from horizontal. Instructions for internal focus were to focus on keeping the feet horizontal. For external focus, instructions were to keep two orange markers horizontal that were attached to the balance board in front of the feet. In both conditions participants were also instructed to look straight ahead, while concentrating on the feet or markers. This extra instruction to look straight ahead was added to keep visual feedback the same in both conditions. However, participants' line of sight was not measured in these studies. For the inflated balance disk, performance was measured with a force platform. RMS amplitude of deviation from the mean centre of pressure position was calculated to quantify balance performance. Internal focus instructions were: "Minimise movements of the feet", and external focus instructions were: "Minimise movements of the balance disk".

External focus has been shown to produce benefits on balance performance through repeated measures of the same participants in both internal and external focus conditions in young healthy adults (Wulf *et al.*, 2004; Wulf *et al.*, 2007). Retention studies with different young healthy adults for each condition also showed improved motor learning for external focus in balance tasks (Wulf *et al.*, 1998; Shea & Wulf, 1999; Wulf *et al.*, 2001; McNevin *et al.*, 2003; Wulf & McNevin, 2003; Chiviacowsky *et al.*, 2010).

A possible explanation of the difference in motor performance between internal and external focus of attention conditions is that the internal focus instructions cause the participant to focus too much on moving the feet, while the control of whole body centre of mass movement is reduced. Movements of all body parts need to be coordinated to keep the balance board or disk horizontal. In addition, the external focus instructions are more closely related to the goal of the task. Therefore one could argue that that external focus is advantageous to an internal focus on a subset of body movements, as the whole body needs to be coordinated in order to successfully accomplish the task.

The benefits of external focus for balance performance in postural control were limited, as they were only found for balancing tasks that were different than normal standing on a solid surface (Wulf *et al.*, 2007). However, some support was found for the claim that external focus on a suprapostural task could also improve postural balance performance for standing on solid ground (McNevin & Wulf, 2002). In that study GRF data were collected for participants who were instructed to stand still while lightly touching a loosely hanging sheet with their fingertips. Instructions varied slightly between conditions. For internal focus they were asked to minimise movements of the finger and for external focus they were instructed to minimise movements of the sheet. No difference in postural sway amplitude was found, but MPF (mean power frequency) was higher for external focus. It was concluded that response frequency and therefore balance responses were improved. One could argue however, that an increase in response frequency without a decrease in postural sway does not necessarily imply that balance responses are improved.

The discussed body of literature on attentional focus supports the beneficial effects of external focus on postural balance performance for young healthy adults. Whether this effect is also present in elderly is insufficiently studied. One study did conclude that external focus causes improved balance learning in healthy elderly. The effect of focus of attention on motor learning in postural control was studied in 32 elderly standing on a stabilometer (Chiviacowsky *et al.*, 2010). On the first day of testing the external focus group had more 'time in balance', however this difference between groups was not significant. Learning effects were assessed with retention tests on the next day without focus instructions. These retention tests did show significantly longer 'time in balance' for the external focus group, however it was not tested whether the increase of 'time in balance' on the second day was larger for external than for the internal focus group. Therefore one could wonder whether this study showed a learning effect. Furthermore, the sample size of 32 participants might be too small for between-subjects comparisons of balance performance. However this study does suggest that the improvement of balance performance by external focus can be extended from the young adults to elderly.

1.5.4 Effects of attentional focus on gait performance

Studies on the effects of attentional focus on balance performance in gait are very scarce and their methodologies have been disputed. Canning (2005) studied gait of Parkinson's disease patients who carried a tray with glasses during two conditions. For internal focus they were instructed to direct the focus of attention towards walking ("Attend to maintaining big steps while walking") and for external focus towards balancing the tray of glasses ("Attend to balancing the tray and glasses"). Increased gait velocity and stride length was found for the internal focus condition. This operationalization of internal and external focus was criticised by Wulf (2013), as these instructions refer to two different motor tasks, as opposed to internal and external focus with regard to the same task. Furthermore, the internal focus instructions did not refer to the body itself.

Shafizadeh *et al.* (2013) compared acute effects of attentional focus on gait as well. They assessed gait of multiple sclerosis patients walking on a treadmill. For internal focus, the patients focussed on foot performance presented on a screen, and for external focus they focussed attention on external markers and auditory information. The authors found increased stride length, step length, step speed and energy expenditure per step for the external focus condition. Based on these findings they concluded that external focus induced improved gait performance. However the different modes of feedback that were used for internal and for external focus might not result in a useful comparison. The difference in gait parameters might just be caused by the extra information that was presented through more sensory channels for the external focus condition. In addition, no dependent variables were tested that were directly related to balance and stability of gait.

In sum, research on the effects of internal and external focus of attention on gait performance in elderly could be improved by using measures of gait performance that have been related to falls in elderly.

1.5.5 Effects of dual-tasks on balance and falls

For most circumstances in daily life, balance control is performed with at least one other concurrent task that requires some degree of mental effort, e.g. thinking and/or talking. Therefore a body of literature on fall research in elderly assessed balance and gait performance while a concurrent cognitive task was performed as well. This experimental design is referred to as the dual-task paradigm. These dual-tasks have qualitatively different effects on postural control than fear, as differences in neuromuscular regulation were found which indicate distinct control processes (Stins *et al.*, 2011).

Dual-task performance has been related to fall risk. In a 5-year prospective study, executive function and dual-task gait variability were predictors for falls (Mirelman *et al.*, 2012). Dual-task intervention studies have also shown to improve balance, gait performance and dual-task gait performance in elderly (Dorfman *et al.*, 2014). In addition, dual-tasks have shown to acutely affect balance performance. Stins and Beek (2012) argued that even though fast reflexive postural adjustments are 'cognitively impenetrable', attention demanding control can be exerted to some extent when needed. Evidence was found that some degree of attention might be needed in postural control for sensory integration and to respond to balance perturbations (Shumway-Cook & Woollacott, 2000; Woollacott, 2000; Redfern *et al.*, 2001; Teasdale & Simoneau, 2001). Therefore some studies found that a concurrent cognitive task impairs balance performance (Maylor & Wing, 1996; Andersson *et al.*, 1998; Shumway-Cook & Woollacott, 2000; Condron & Hill, 2002), however other research suggests that this cognitive-motor dual-task acutely improves balance performance (Dault *et al.*, 2001; Andersson *et al.*, 2002; Brown *et al.*, 2002; Deviterne *et al.*, 2005).

To explain these findings it was proposed in several papers that relatively easy (low effort) cognitive tasks improve concurrent balance performance, whereas more demanding cognitive tasks impair concurrent balance performance (Riley *et al.*, 2003; Vuillerme & Nougier, 2004; Deviterne *et al.*, 2005). This U-shaped relation between balance performance and cognitive dual-task difficulty was supported by Huxhold *et al.* (2006) for both young and older adults.

Lovden *et al.* (2008) tested whether this U-shaped relation between motor performance and concurrent cognitive task difficulty could be extended to gait. For gait performance the relation between variability of stride-to-stride gait parameters and cognitive task difficulty was tested, however no evidence was found for same U-shaped pattern. The results did show increased gait variability for increased cognitive demand for young adults, but not for elderly.

Dual-task, internal and external focus of attention

In line with the theory of reinvestment one could speculate that prevention of internal focus without movement related external focus of attention might also result in improved balance performance. This prevention of internal focus might be achieved through a dual-task. Therefore Wulf and McNevin (2003) investigated the effects of internal and external focus and dual-tasking on balance performance on a stabilometer in a retention study. For the dual-task condition participants were instructed to shadow (i.e. pay attention to) a narrated story played through a speaker system while balancing on the stabilometer. Balance learning occurred in all conditions, however the external focus condition showed increased balance learning compared to the internal focus, dual-task and baseline condition. No significant difference was found between the control, internal and dual-task conditions. It was therefore concluded that simply distracting balance performers is not enough to improve balance performance. However, the number of participants was relatively small for a between-subjects analysis as 14 participants were included for each of the internal, external and control conditions and 13 for the dual-task condition. Furthermore, in addition to the internal and external focus conditions, the focus of attention or cognitive performance in the dual-task condition was not measured or assessed.

1.6 Analysis of kinematics

1.6.1 Gait stability and variability

To study gait, 3d kinematics of the body can be recorded to measure the movement patterns of the entire body. Spatiotemporal gait parameters, e.g. step length, step width, stance time and swing time can be calculated from these kinematic data. Variability of a gait pattern has been quantified with the coefficient of variation (CV) of spatiotemporal parameters. The CV is calculated as the standard deviation divided by the mean of the parameter and multiplied by 100 to express the variability in percentage of the mean. Gait variability is associated to fall risk (Hausdorff *et al.*, 2001) and fall history (Hausdorff *et al.*, 1997; Toebes *et al.*, 2012).

More recently, the Local Divergence Exponent (LDE) has increased in popularity as a measure of gait stability (Rosenstein *et al.*, 1993; Lockhart & Liu, 2008; Bruijn *et al.*, 2010; Bruijn *et al.*, 2012; Toebes *et al.*, 2012; Rispens *et al.*, 2014; Arvin *et al.*, 2015). LDE, also called local dynamic stability and derived from Lyapunov exponents, can be calculated from kinematic data and is a measure of the average logarithmic rate of divergence of a system. Therefore, an increase in LDE represents a decrease in gait stability. A distinction is made between the short term and the long term LDE, where the short term LDE typically refers to the divergence within the time window of 1 step. Short term LDE was also found to be a predictor for fall history (Liu *et al.*, 2008; Lockhart & Liu, 2008; Toebes *et al.*, 2012) and was suggested to be an indicator of future falls (Lockhart & Liu, 2008). A popular method to calculate LDE was published by Rosenstein (Rosenstein *et al.*, 1993).

Most gait research has focussed on steady state gait. However falls could be related to deteriorated responses to gait perturbations. Therefore gait stability has also been assessed through measurement of responses to mechanical perturbations of the gait pattern (Bruijn *et al.*, 2010). Centre of mass velocity time series of these responses provide valuable information on the response amplitude and the time it takes to return to a normal gait pattern.

1.6.2 Statistical Parametric Mapping (SPM)

Balance responses in postural control and gait are measured as time series data. Statistical testing of time series usually involves scalar extraction and qualitative interpretation, e.g. selection of peak times and peak amplitudes.

This is needed for most conventional methods of statistical analysis (e.g. ANOVA and student's t-test) as they cannot handle time series as a whole as input data. However, each point in time is of interest in time series data of balance responses to a perturbation. Therefore a method of statistical analysis is needed that tests the whole time series of a certain variable. SPM is a validated method of statistical analysis where time series can be used as the unit of observation instead of scalar values. This allows for the often-neglected time dependence of the signal to be incorporated in statistical testing. This method is now increasingly used in the field of biomechanics (Pataky, 2012; Robinson *et al.*, 2014; Serrien *et al.*, 2015). SPM for time series is implemented by the open-source toolbox SPM1D (v.M0.1, Todd Pataky 2014, www.spm1d.org,) in Matlab (The MathWorks, Inc., Natick, Massachusetts, United States).

With SPM, traditional scalar tests are repeated for each time sample of the tested signal(s). E.g. with an SPM t-test, one could test at which points in time two groups of signals are significantly different from each other. The output statistic, SPM{t}, contains a trajectory consisting of a t-test value for each time point. The critical threshold of significance is then defined based on the smoothness of the signals (Friston *et al.*, 2007), random field theory expectations (Adler & Taylor, 2007) and the alpha value (typically 5%). The interpretation of significance is similar to a traditional t-test. When the SPM{t} trajectory exceeds the threshold of significance (alpha) at certain time samples, the null hypothesis is rejected for these time samples. The threshold is often exceeded during one or more time windows of the tested signals, due to interdependence of neighbouring points. Therefore these significant time windows are called "supra-threshold clusters". A single p-value is then calculated for each supra-threshold cluster (Adler & Taylor, 2007). See Figure 1.4 for an example.

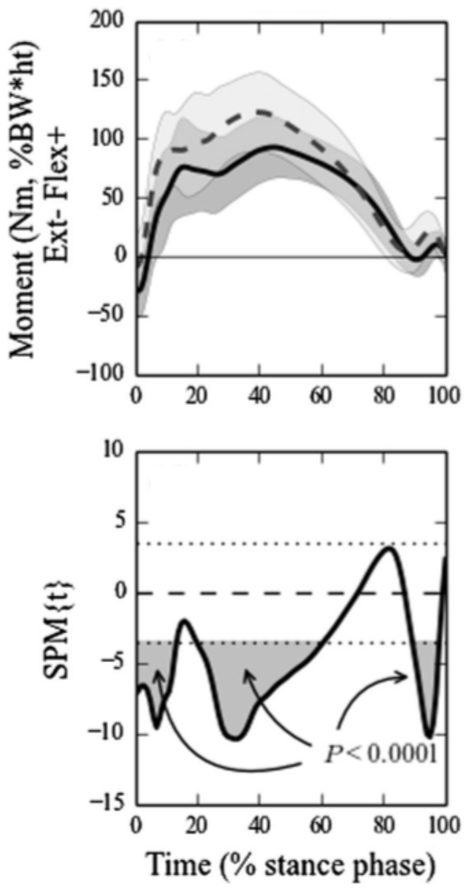

Figure 1.4: Redrawn from Dingenen *et al.* (2015). In the top graph mean hip flexion moments are shown for drop vertical jumps (thick line) and single leg drop vertical jumps (thick dashed line). Shaded areas represent standard deviations. The bottom shows the SPM{t} trajectory of an SPM independent samples t-test. The dotted lines indicate the threshold of significance and the shaded areas are the supra-threshold clusters. The p-value is shown for each supra-threshold cluster.

1.7 Conclusions

The research field of human balance control in relation to falls is rapidly expanding. However, the mechanisms relating cognitive sensorimotor control, focus of attention and fear of falling are not well established. Many authors assume that fear of falling causes activity avoidance, which causes decline of balance performance and thereby increases fall risk. Although, this mechanism is widely accepted, no clear evidence for this theory was found. Multiple studies did support a direct relation between fear of falling and balance impairment, without mediation of activity avoidance. Using GVS and height-induced postural threat to induce fear of falling, vestibular balance responses were found to be amplified by fear of falling. However, it is debated whether these amplified vestibular balance responses affect balance performance. Full body kinematic measurements of the GVS induced sway response, could clarify the interplay of the short- and medium-latency response, its relation to fear of falling and to balance performance.

Furthermore, it was suggested that fear increases sensitivity to self-motion. Indeed, elderly with a history of falls show higher levels of reinvestment. Additionally, improved balance performance was found with external focus when balance is challenged. As such, the effects of attentional focus on balance performance are evident. However, a gap of knowledge exists regarding the effects that internal and external focus of attention could have on stability and balance in gait, specifically in elderly. Therefore, future research on the effect of fear on balance performance and studies on attentional focus strategies using gait performance measures that have been related to falls, might provide new intervention strategies to reduce the number of falls in elderly.

1.8 Aim and outline of the thesis

A gap exists in the literature on the effects of fear of falling on balancing reflexes and the effects of attentional focus on gait performance, especially in

elderly. Therefore, the general aim of this PhD project was to assess the effects of fear of falling and focus of attention on human balance control.
Chapter 2 of this thesis describes a study conducted at the Manchester Metropolitan University on the effect of fear of falling on balance. Young healthy adults were stimulated with GVS to elicit vestibular balancing reflexes. To induce fear of falling they were stimulated while standing on a narrow 3.85 m high walkway. These responses were compared to standing at ground level and measured using full body kinematics. The main aim of this study was to investigate whether fear of falling influences vestibular balancing reflexes or not. In addition we aimed to gain insight into the contribution of the short-latency response to balance and its interplay with the medium-latency response. Knowledge of these fundamental balancing mechanisms will expand our understanding of human balance performance and might advance future fall prevention methodologies.

Chapters 3 and 4 elaborate on the effects of attentional focus and fall history on gait variability and stability in elderly. This study was conducted at the Vrije Universiteit Amsterdam. Full body kinematics of elderly was collected while they walked on a split belt treadmill with a virtual reality environment to induce realistic optic flow. In addition, gait was perturbed by unilateral treadmill decelerations at unexpected time intervals. In Chapter 3 we focussed on the effects of attentional focus and fall history on the gait stability and variability of direct balancing responses to the perturbations. Attentional focus and fall history effects on stability and variability of the unperturbed gait bouts between perturbations are investigated in Chapter 4. If external focus would result in increased gait stability, new tools to advance the field of fall prevention could be developed. In Chapter 5 the collective findings in Chapter 2-4 are reviewed in a general discussion.

Chapter 2

The Effect of Fear of Falling on Vestibular Feedback Control of Balance

Introduction: Vestibular sensation contributes to head stabilisation and fall prevention. To what extent fear of falling influences these different vestibular feedback processes is currently undetermined.

Method: We used galvanic vestibular stimulation (GVS) to induce vestibular reflexes while participants stood on a narrow walkway at 3.85 m height to induce fear of falling and at ground level. Fear was confirmed by questionnaires and elevated skin conductance. Full-body kinematics were collected to measure anode-cathode related vestibular responses (head orientation, whole body balance) and to clarify the debated functional goal of lower extremity short-latency responses. Statistical parametric mapping analysis provided sensitive discrimination of early GVS and height effects.

Results: The GVS response comprised a rapid, anode-directed cervical-head acceleration, a short-latency cathode-directed acceleration of lower extremities and pelvis, an upper thorax anode-directed acceleration and subsequently a medium-latency anode-directed acceleration of all body parts. At height, head and upper thorax early acceleration were unaltered in size and latency, however short-latency lower extremity acceleration was increased. The effect on balance was a decreased duration and increased rate of change of the COM acceleration pattern.

Discussion: Kinematic analysis of the effect of height confirms: (ii) Fear modifies vestibular control of balance, (iii) head-in-space stabilisation is governed by different mechanisms and is unaffected by fear of falling. We propose that both the short- and medium-latency reflexes functionally contribute to whole body balance and are biomechanically coupled as one coordinated response.

Adapted from: de Melker Worms, J. L. A., Stins, J. F., Beek, P. J., Loram, I. D. (2016). The Effect of Fear of Falling on Vestibular Feedback Control of Balance. *Manuscript submitted for publication.*

2.1 INTRODUCTION

To maintain balance, humans rely upon vestibular information. The vestibular system is highly sensitive to large, fast movements of the head and provides fast, strong responses to preserve whole body balance (Forbes *et al.*, 2014). Although less sensitive to small, slow changes, the vestibular system also provides a sense of upright orientation of the head and body (Fitzpatrick & Mccloskey, 1994; Fitzpatrick & Day, 2004).

Fear of falling is known to influence human balance (Stins *et al.*, 2011; Tersteeg, 2012; Osler *et al.*, 2013). When fearful, movements become more cautious and joint stiffness tends to be increased (Adkin *et al.*, 2002; Tersteeg, 2012; Osler *et al.*, 2013; Young & Mark Williams, 2015). In addition, it has been proposed that anxiety increases sensitivity to self-motion, through noradrenergic and serotonergic input to the vestibular nuclei (Balaban, 2002). Studies of fall risk in the elderly have shown associations between cognitive motor measures (e.g. concern about falling and poor executive function) and physiological measures of impaired balance (Delbaere *et al.*, 2010a; Hadjistavropoulos *et al.*, 2011). From a healthy ageing perspective there is a need to understand the mechanisms relating fear of falling to balance and mobility in the elderly. A recent cross talk debate (Horslen *et al.*, 2015b, a; Reynolds *et al.*, 2015b, a; van Dieen *et al.*, 2015) highlighted the range of potential mechanisms related to fast physiological processes, slower processes and processes more traditionally associated with psychology, such as "reinvestment". Here we focus on the fastest vestibular contributions to human balance and the potential interplay with fear of falling.

It is currently controversial whether fear of falling influences the vestibular control of balance. Bipolar binaural Galvanic Vestibular Stimulation (GVS) is a frequently employed method to study vestibular balance reflexes (Fitzpatrick & Day, 2004). Cutaneous electrical stimulation at the mastoid processes stimulates the vestibular nerves and creates a sensation of roll rotation. This elicits a lateral body sway response towards the anode electrode. A paradigm of standing at height on a 22 cm narrow walkway to evoke fear of falling, combined with GVS, has shown that fear of falling might differentially affect the feedforward and feedback components of the vestibular-evoked balance response (Osler *et al.*, 2013). Given sufficient time to integrate proprioception of movement with vestibular sensation, vestibular evoked sway is strongly arrested at height compared to ground. However, kinematic data of head and torso showed that fear had no measureable effect on the initial (0-800 ms) vestibular evoked balance response. In contrast, Horslen *et al.* (2014) have shown increased gain in the initial vestibular reflex response, using a similar height paradigm. However, in their study ground reaction force (GRF) data was used to assess balance responses and a different stimulation paradigm was employed (stochastic vestibular stimulation, SVS) to elicit vestibular balancing reflexes.

Vestibular information is used within a variety of mechanisms related to balance. Vestibular sensory feedback is used to regulate eye movement through the vestibulo-ocular reflex (VOR), to regulate head orientation through the vestibulocolic reflex (VCR), and to regulate balance through responses that control movement of the whole-body COM. It could be the case that fear (with both motivational and perceptual consequences) has differential effects on these three vestibular balancing responses, as these responses have different onset latencies to GVS. This implies distinct neural pathways. As such it cannot be assumed that fear operates equally on all mechanisms related to vestibular responses.

EMG data can be used to reveal the latency of vestibular responses and thereby help to identify the neural pathways that could be involved. For example, the VCR has a latency of approximately 8-10 ms (Watson & Colebatch, 1998; Forbes *et al.*, 2014). When recording lower limb muscles during upright standing, short- and medium-latency vestibular balancing responses were found. The onset of these short-latency responses ranged from 42 to 65 ms and for medium-latency from 98 to 120 ms post GVS onset (Britton *et al.*, 1993; Fitzpatrick *et al.*, 1994; Ali *et al.*, 2003; Fitzpatrick &

Day, 2004; Son *et al.*, 2008; Mian *et al.*, 2010; Muise *et al.*, 2012). In addition, the short- and medium-latency responses cause GRF peaks at approximately 120-200 ms and 290-400 ms latency, respectively, due to an electromechanical delay (Mian & Day, 2009; Dakin *et al.*, 2010; Mian *et al.*, 2010; Horslen *et al.*, 2014; Mian & Day, 2014). These short- and medium-latency responses in EMG and/or GRF data are well established as they were replicated in at least 5 different research institutions.

According to Fitzpatrick *et al.* (1994) the short-latency response can produce small segmental movements, but has no effect on the whole-body sway response. It is assumed that the medium-latency response is responsible for the GVS induced sway response, however the neurophysiological origin of the short-latency response and its contribution to balance are still debated (Cathers *et al.*, 2005; Mian *et al.*, 2010). While the short-latency response occurs only in muscles required for balance, the functional relationship with the medium-latency response is unclear.

In general, the relationship between muscle activity and the resulting body movement is unclear due to insufficient knowledge of how muscle forces combine to produce movement in a non-rigid, multi-segmental body. Therefore, the movement pattern related to vestibular-evoked balancing reflexes, and its mapping to EMG and force plate data is insufficiently understood. Even though EMG analysis has yielded useful insights, measurement of kinematics is required to determine the effects of head stabilisation (VCR) and balancing reflexes on body movements. GRF measurements in isolation are insufficient because GRF reveals the acceleration of the COM, but does not reveal individual joint movements. Markers tracking only head and trunk as in Osler *et al.* (2013) are also inadequate to distinguish head stabilisation from whole body balancing reflexes. To our knowledge, full body kinematics of the GVS response has not been measured before. According to Newton's second law of motion, the GRF pattern is inevitably proportional to the body's COM acceleration. Therefore, we anticipate that full-body kinematic analysis will also reveal a short- and medium-latency movement pattern similar to the GRF short- and medium-latency response pattern.

2.1.1 Aims and approach

In this experiment we investigated how vestibular balance reflexes are influenced by fear of falling. It is unknown whether, and to what extent, this psychological state modulates the vestibular reflex mechanisms involved in balance control. To challenge the balance system we used GVS to evoke substantial mediolateral sway both at ground level and at a height that is known to invoke fear of falling (Osler *et al.*, 2013). We recorded full-body kinematics to measure the balance response to GVS, in order to discriminate the VCR response from regulation of COM (i.e. the balancing response), and to gain insight into the origin of the balance response. Our kinematic data was analysed using statistical parametric mapping (SPM). SPM is a validated method of statistical analysis for time series data, which is now increasingly used for kinematic time series (Pataky, 2012; Robinson *et al.*, 2014; Serrien *et al.*, 2015). We focussed on the short- and medium-latency vestibular responses (0 – 400 ms). In our study we compared our full-body kinematic data to known EMG and GRF responses established in multiple laboratories. Our main research question was: What is the effect of fear of falling on vestibular control of balance and head stabilisation?

2.2 METHODS

2.2.1 Ethical approval

This study was approved by the local ethics committee of the Science & Engineering faculty, Manchester Metropolitan University. Participants were naive to the precise purpose of the experiment and gave written informed consent prior to their participation. The study conformed to the standards set by the latest revision of the Declaration of Helsinki.

2.2.2 Participants

Sixteen young healthy adults with no known neurological, musculo-skeletal, balance or vestibular disorder were recruited as a sample of convenience. Ten men and six women were tested. The averaged participant characteristics were as follows: mean (standard deviation); age: 25.9 (5.1) years, height: 1.74 (0.1) m, weight: 69.5 (13.5) kg, BMI: 22.9 (3.5).

2.2.3 Material

Vestibular-evoked balance responses were studied in two conditions. In one condition participants stood on a 22-cm-wide walkway placed on the laboratory floor. In the other condition, participants stood on a 22-cm-wide walkway elevated 3.85 m from ground level. The high walkway extended from a mezzanine into a larger neighbouring room (Figure 2.1). Access to the walkway was provided by sliding doors opening the laboratory wall (width 3.57 m). Stimulation and data acquisition devices were stationed on the mezzanine.

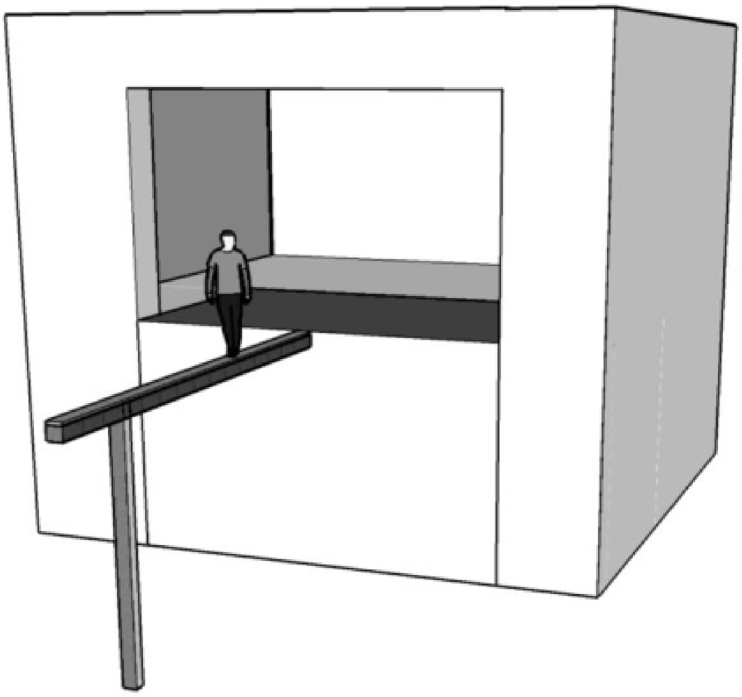

Figure 2.1: Narrow walkway at height.

2.2.4 Safety system

In both the ground and height conditions participants wore a full-body harness attached to a safety system to prevent a possible fall. The safety system consisted of an inertial reel and a dynamic rope system that was belayed by a certified assistant. Both were attached to a trolley-mounted anchor point positioned directly above the participant to allow walking and standing without creating drag on the participant. This was the same safety system as used by Osler *et al.* (2013). As the system was attached to the back of the harness, the ropes ran behind the participant outside their visual field. Participants were fully informed of the safety system. However, during data collection, participants could neither see nor feel the safety ropes. Furthermore, they did not test the system prior to the experiment. Verbal, post-experiment debriefing confirmed that knowledge of the safety system provided little comfort to participants who generally reported the experience to be rather testing.

2.2.5 Data collection

Full-body kinematics were collected by means of a 3D motion capture system operating at a sample frequency of 100 Hz using 52 retroreflective passive markers and 9 infrared cameras (Vicon Motion Systems Ltd., Oxford, UK). The marker placement was as follows: 5 on the head (frontal bone, 2 on left and 2 on right zygomatic bone), 2 on sternum, upper back at C7, lower abdomen, 5 on pelvis (ASIS, PSIS and sacrum), upper lateral thigh (iliotibial band), 5 per knee (femoral and tibial condyles, and tibial tuberosity), lower lateral shanks, medial and lateral ankles, 2 per foot (heel and base of the 3rd metatarsal), shoulders (acromion), upper arms (deltoid insertion), medial and lateral elbows, lateral lower arms (ulna shaft), 2 per wrist (radial and ulnar styloid process), 1 per hand (2nd metacarpal head).

Furthermore, skin conductance (SC) was recorded during all trials as a measure of physiological arousal. SC was measured using two self-adhesive gel electrodes that were placed on the palmar surface of the distal phalanges of the first and third fingers. The electrodes were connected to a GSR Amplifier (ADinstruments Ltd., model ML116, Dunedin, New Zealand).

Kinematics and SC data was collected and synchronized using Vicon Nexus software (1.8.5.61009h, Vicon Motion Systems Ltd., Oxford, UK). GVS impulses with a current of 1 mA and duration of 2 s were delivered using carbon rubber electrodes (46 by 37 mm) placed in a binaural bipolar configuration similar to the method of Osler *et al.* (2013). This type of stimulus has shown to evoke significant body sway responses (Day *et al.*, 2010; Osler *et al.*, 2013).

To assess participants' state of fear, the State-Trait Anxiety Index (STAI) (Rossi & Pourtois, 2012) was used. From the STAI questionnaire only the state anxiety index was measured. Moreover, participants were asked to verbally rate their fear of falling on a 1-10 Likert scale anxiety thermometer at several instances of the experiment. The anxiety thermometer has been shown to have fair validity and reproducibility (Houtman & Bakker, 1989). In a more recent study a one-question 5-point Likert anxiety scale was found to be suitable for anxiety measurement (BinDhim *et al.*, 2013).

2.2.6 Procedure

In a repeated measures design participants were tested during the same series of trials in the high and ground walkway conditions in counter-balanced order. Participants were instructed to stand still but relaxed 1.5 m out on the walkway with their head facing forwards and the feet directed along the anterior-posterior axis of the walkway (Figure 2.1). To maximize lateral sway and rule out effects of vision, participants stood with their feet together and eyes closed. After 10 familiarizing GVS stimuli, thirty GVS impulses (15 anode-left, 15 anode-right, randomly ordered) were applied. It is important to note that the direction of illusory movement evoked by the stimulus was always towards either the right or the left edge of the walkway, depending on GVS polarity (anode left or right). Participants were permitted to open their eyes after each block of 10 trials. These trials were repeated, meaning that all participants completed 3 blocks of 10 trials in both the height and the ground condition. Data acquisition for each trial began 3 s prior to and ended 6 s following GVS onset. After each 6th trial in the 1st block, each 8th trial in the 2nd block and each 3rd trial in the 3rd block of trials participants were asked to verbally rate their level of fear of falling for the anxiety thermometer.

2.2.7 Data processing

Baseline SC was calculated as the mean SC level over 2 seconds of quite standing at ground level. Pre and post GVS onset SC levels were calculated by averaging SC between 3 and 0.5 s before GVS onset, and between 0 and 6 s after GVS onset, respectively. SC signals were normalised by subtracting the baseline signal and dividing by the standard deviation of the pre GVS values in the ground condition.

Using Visual 3D (v5.02.07, C-Motion Inc., Germantown, USA) mediolateral displacement of the following body nodes were calculated: Whole-body COM, head COM, upper thorax (superior end of thorax segment), pelvis COM, and the elbows, wrists, knees and ankles. These locations are collectively referred to as nodes. Additionally foot-in-space and head-in-space segment angles as well as ankle, knee, hip, lower back, neck, shoulder, elbow and wrist joint angles in the frontal plane were calculated. A GVS stimulus causes increased mediolateral body sway to the side on which the anode electrode is placed on the head. For half of the GVS trials the anode of the GVS electrodes was on the right side and for the other half of the trials it was on the left side. Therefore, instead of analysing right and left body nodes and angle variables on their own (e.g. right or left knee), these segments were analysed and named based on the anode-cathode configuration, e.g. 'anode knee' refers to the knee on the anode side of the body (Figure 2.2).

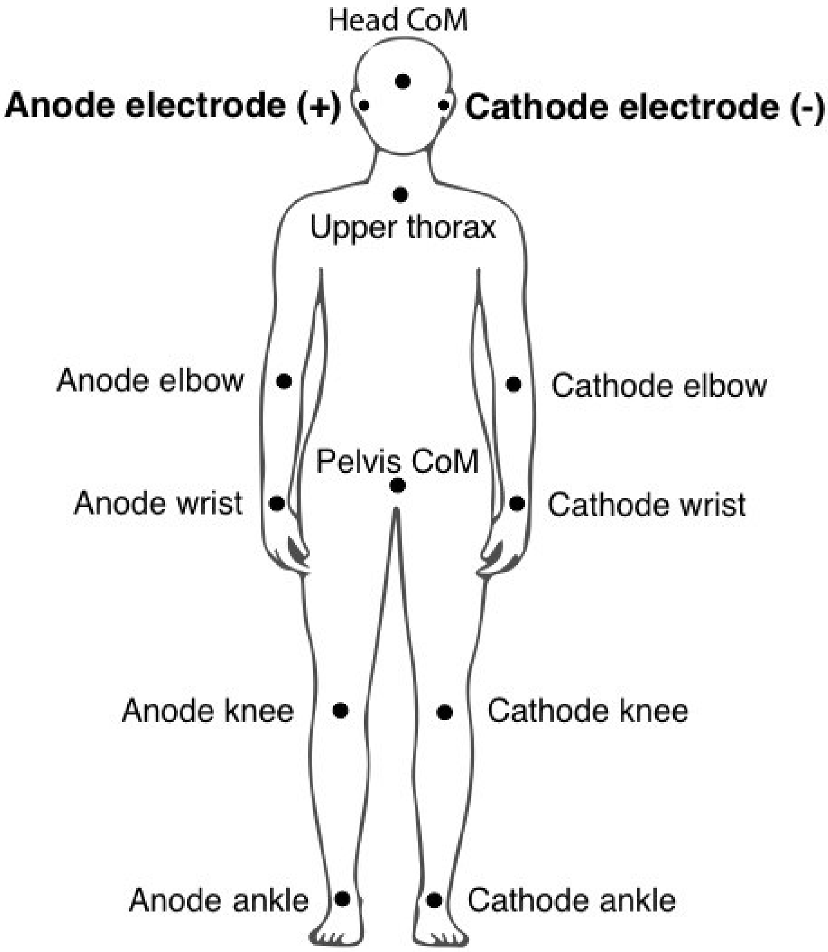

Figure 2.2: Body nodes based on GVS electrode configuration. We focussed on mediolateral linear acceleration of the indicated body nodes. These nodes were analysed based on the anode-cathode configuration as the GVS polarity changed between trials.

Kinematics of these body nodes provides information on how the balancing responses affect the whole body movement. Acceleration of the head results most directly from neck muscle contraction caused by the VCR. Acceleration of the lower body results from activation of muscles relevant to moving the whole body by mechanical interaction with the ground. Acceleration of the upper limbs results passively from trunk motion, from actions of muscles relevant to adjusting balance through inertial action and from possible efforts to protect the body.

For each positional and angular variable, the value at GVS onset of a trial was subtracted from all values of the time series of the trial. Furthermore the sign was corrected based on anode electrode location. Analysis of published data shows that the frequency bandwidth of the short- and medium-latency GRF GVS responses averaged over multiple trials and participants does not exceed 3 Hz (Marsden *et al.*, 2005; Mian & Day, 2014). Therefore we filtered our kinematic data using a 6 Hz low pass Butterworth filter and differentiated twice using a 3rd order Savitsky-Golay filter with a temporal window of 170 ms (Press *et al.*, 1999). As we were interested in the vestibular reflex response we analysed node acceleration and angle acceleration data in the time domain between 0.2 s before and 0.7 seconds after GVS onset.

2.2.8 Statistics

Questionnaire and SC data

Student's paired t-tests were used to test whether STAI state, anxiety thermometer and SC were increased at height compared to ground. Lastly, correlations between all combinations of SC, anxiety thermometer scores and STAI state scores were calculated using Spearman's rho. The statistics toolbox in Matlab was used for statistical testing.

Kinematics: Statistical Parametric Mapping (SPM)

To answer our research question all linear and angular acceleration time samples within the first 400 ms after GVS onset were of interest. Therefore we used a validated method (SPM) to test at what times the signals were statistically different from zero and when they were different between conditions. All SPM analyses were implemented using the open-source toolbox SPM-1D (v.M0.1, Todd Pataky 2014, www.spm1d.org,) in Matlab R2014a. SPM regards the whole time series as the unit of observation and is now increasingly used in the analysis of kinematic time series (Pataky, 2012; Robinson *et al.*, 2014; Serrien *et al.*, 2015). This allows time dependence to be incorporated directly in statistical testing.

In this study SPM statistics were calculated of the averaged trials per participant for each condition. A SPM two-tailed one-sample t-test was used separately for the ground and height condition data to test if linear and angular acceleration of previously mentioned body nodes, joints and segments is different from zero (α=0.05). Additionally a SPM two-tailed paired samples t-test (Robinson *et al.*, 2014) was used for a ground vs. height comparison of the same dependent variables. The scalar output statistic, SPM{t}, was calculated separately at each individual time sample. To test the null hypothesis the critical threshold is calculated at which only α % (5%) of the analysed trajectories would be expected to traverse. This threshold of significance is based upon estimates of trajectory smoothness (Friston *et al.*, 2007) and Random Field Theory expectations (Adler & Taylor, 2007). Conceptually, a SPM t-test is similar to the calculation and interpretation of a scalar t-test; if the SPM{t} trajectory crosses the critical threshold at any time sample, the null hypothesis is rejected. However, a SPM t-test avoids the false positives of multiple scalar t-tests and avoids the false negatives of scalar t-tests with Bonferroni correction (Adler & Taylor, 2007). Typically, due to interdependence of neighbouring points, multiple adjacent points of the SPM{t} curve often exceed the critical threshold. We therefore call these "supra-threshold clusters". SPM then calculates cluster specific p-values which indicate the probability with which supra-threshold clusters could have been produced (Adler & Taylor, 2007).

2.3 RESULTS

2.3.1 Questionnaires and SC confirm increased fear of falling at height

STAI, anxiety thermometer and SC data showed that participants had a higher level of fear of falling and physiological arousal in the high walkway condition than in the ground walkway condition (Table 2.1)

Table 2.1: STAI and anxiety thermometer scores. State anxiety scores (STAI) may range between 20 and 80. Anxiety thermometer scores may range between 1 and 10.

	STAI State		**Anxiety thermometer**	
	Ground	Height	Ground	Height
Mean (SD)	27.4 (5.7)	34.8 (9.3)	2.0 (1.1)	4.7 (3.2)
Min	20	20	1	1
Max	37	48	4	10

SC was increased significantly in the height condition both pre ($t = -2.709$, $df = 15$, $p = 0.016$) and post ($t = -2.743$, $df = 15$, $p = 0.015$) GVS onset. In the height condition, the STAI state scores were positively correlated with SC scores ($n = 15$, $rho = 0.506$, $p < 0.05$). For one participant skin conductance was not recorded due to a technical malfunction. At height, 7 out of 16 participants had an average anxiety thermometer score of 6 or higher. For 6 out of 16 participants it was 7 or higher.

2.3.2 Kinematic analysis of vestibular responses to GVS

Full-body kinematic analysis provides a characterisation of the vestibular response that complements the information provided by GRF and EMG data in the literature (Fitzpatrick & Day, 2004). The whole-body COM shows the integrated effect of all responses, whereas the position of the whole-body COM in relation to the base of support is relevant to balance. Acceleration of the whole-body COM is proportional to the GRF revealed by a force plate.

Representative response of the whole-body COM

Standing at height has a modest effect on the early sway response (before ~400 ms), and a clear effect on the late GVS body sway response after ~400 ms. Figure 2.3 shows example whole-body COM mediolateral displacement and acceleration of a representative participant. At ~200 ms after GVS onset the whole-body COM started to accelerate towards the anode electrode in both the ground and height condition. However at ground level peak acceleration was reached at 490 ms and at height at 300 ms. The amplitudes of this anode-directed (anodal) peak acceleration at ground and height were relatively similar. At ground level whole-body COM started decelerating at 890 ms and height deceleration started at 610 ms. In the height condition this resulted in reduced sway displacement after ~1 s compared to ground.

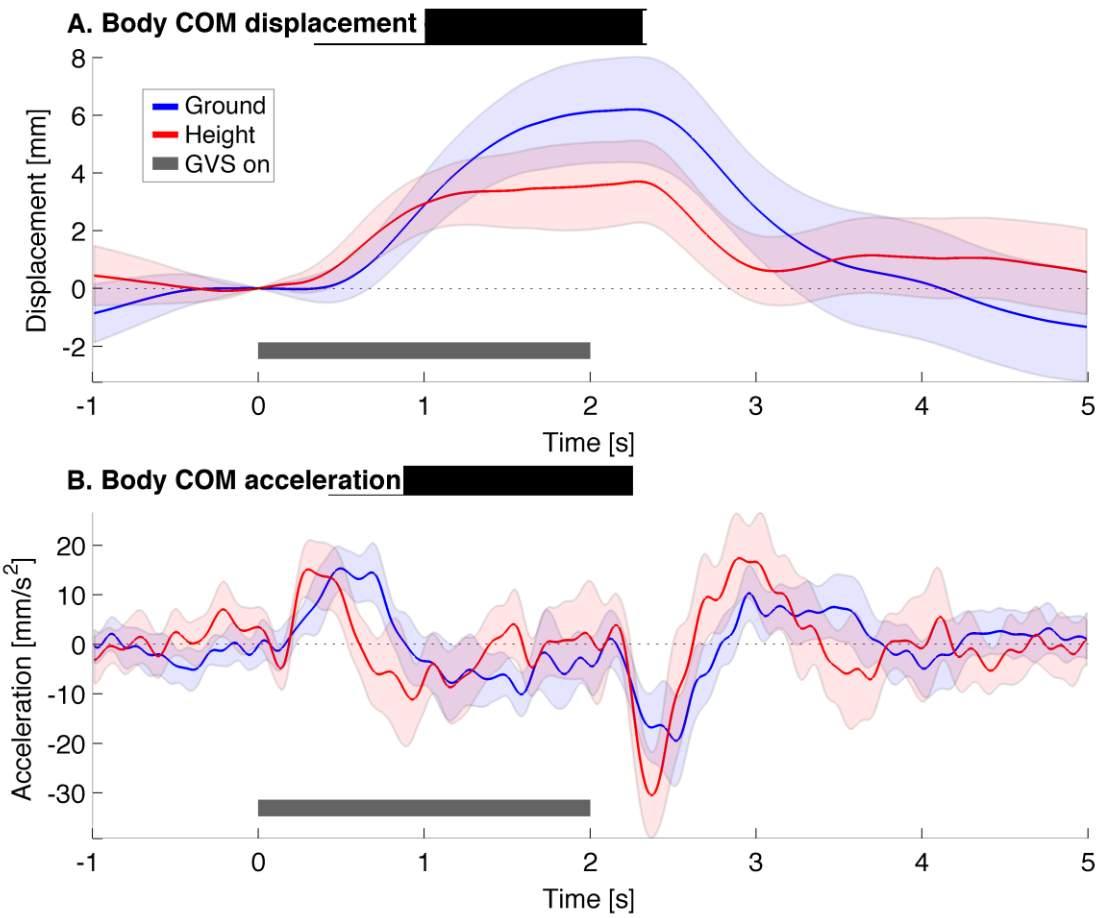

Figure 2.3: Effect of height on body COM response to GVS of representative participant. The mediolateral body COM displacement (***A***) and acceleration (***B***) of one participant are shown. GVS onset occurs at 0 seconds and ends at 2 seconds. Lines represent condition means and shaded areas represent confidence intervals of the trials. The black bar shows the time at which GVS was on. For each trial, COM displacement was scaled to t = 0, i.e., GVS onset.

Group results

The group results reported below show: (i) How the kinematic response relates to published VCR, short- and medium-latency responses, and (ii) the effect of height on the vestibular-evoked acceleration response.

Response of the whole-body COM

GVS evoked whole-body COM sway towards the anode (positive) was conventional in that it plateaued at ~1 s, and was preceded by a small cathode-directed (cathodal) peak (negative) at ~250 ms (Figure 2.4A).

(i) The whole-body COM showed a small initial cathodal acceleration and a main anodal acceleration of ~ 20 mm s-2. The timing of cathodal and anodal acceleration responses showing peaks at ~150 ms, and at ~400 ms was comparable to short- and medium-latency vestibular reflex responses found previously in GRF data (Figure 2.4D and Figure 2.11).

(ii) The main effect of height was an increased magnitude of the early cathodal acceleration and a decreased latency of both cathodal and anodal acceleration phases (Figure 2.4D, G). At height, cathodal acceleration was significantly different from zero at 120-140 ms ($p = 0.027$) followed by significant anodal acceleration at 230-470 ms ($p < 0.001$). In the ground condition no significant cathodal acceleration was found, however anodal acceleration was significant at 230-670 ms ($p < 0.001$). At 550-650 ms the ground-height difference was significant ($p < 0.001$) for body COM acceleration. The ground-height time difference between anodal acceleration peaks was 110 ms and the body COM sway terminated more promptly by ~ 300 ms at height (Figure 2.4A). To summarise, at height the response of the body COM to GVS had a shorter latency and cathodal acceleration was larger than at ground.

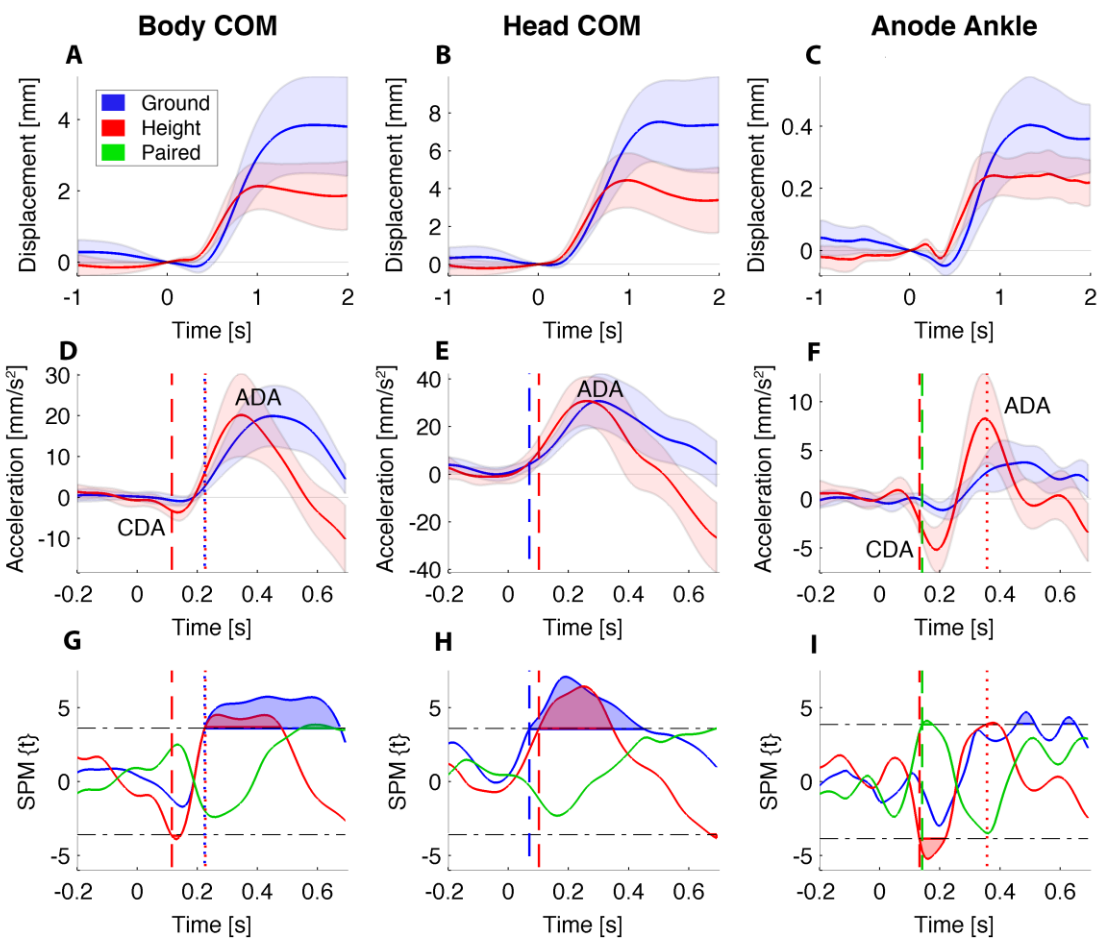

Figure 2.4: GVS effects and ground-height difference effects found on acceleration within 0.2 s after GVS. The *left, middle, right* columns show movement of nodes for: whole-body COM, head COM and anode ankle, respectively. ***A-C,*** Upper row, shows mediolateral position. ***D-F,*** Middle row, shows mediolateral acceleration. Lines represent condition means and shaded areas represent 95% confidence intervals of the ground and height conditions. Anode- and cathode-directed acceleration peaks are indicated by ADA and CDA, respectively. ***G-I,*** Bottom row shows statistical parametric maps. Ground, height, and ground-height difference are in blue, green and red, respectively. Lines represent SPM{t} time series of the separate one-sample t-tests for ground and height data and paired t-tests for the ground-height difference. Horizontal dash-dot lines are the thresholds of significance. Shaded areas are supra-threshold clusters that indicate the time domains with significant effects. GVS onset occurs at 0 s. Vertical dashed and dotted lines represent the onset of significant short- and medium-latency acceleration, respectively. These vertical dashed and dotted lines are shown for significant effects in the ground and height conditions, as well as for the ground-height difference.

Response of the Head COM & upper thorax nodes

Following GVS, the head swayed consistently to the anode before plateauing at ~ 1s (Figure 2.4B).

(i) Initial acceleration of the head COM and upper thorax node was anodal (Figure 2.4E and Figure 2.5). Head COM acceleration was significant from 70 ms ($p < 0.001$, Figure 2.4H), and larger (30 mm/s2) than whole-body COM acceleration, consistent with the VCR. Upper thorax acceleration was significant from 160 ms (Figure 2.5).

(ii) The anodal acceleration of the head and upper thorax nodes were unaffected by height. No significant ground-height difference was found for head COM or upper thorax within the first 0.4 s (Figure 2.4H, 4). This lack of difference between height and ground replicates the head and trunk kinematics collected by Osler et al. (2013).

Response of the lower extremities: pelvis, knee and ankle nodes

Initial cathodal acceleration was observed in the pelvis and lower limbs. This response occurred at short-latency and was followed by anodal acceleration at medium-latency (Figure 2.3 and Figure 2.4).

(i) For the pelvis, both knees and ankles, cathodal acceleration was significant from 100-150 ms (Figure 2.4F and Figure 2.5). These short-latency cathodal acceleration clusters were followed by significant medium-latency anodal acceleration clusters (pelvis and knees), which started between 270 and 370 ms (Figure 2.5).

(ii) The effect of height was to increase substantially, the size of the initial cathodal acceleration in the lower limbs. Inspection of Figure 2.4F and Figure 2.5 shows the increase in size was dramatic for the knee and ankle nodes, as confirmed by the significant ground-height difference in the initial cathodal acceleration. Cathodal acceleration was also observed earlier at height (Figure 2.4F and Figure 2.5).

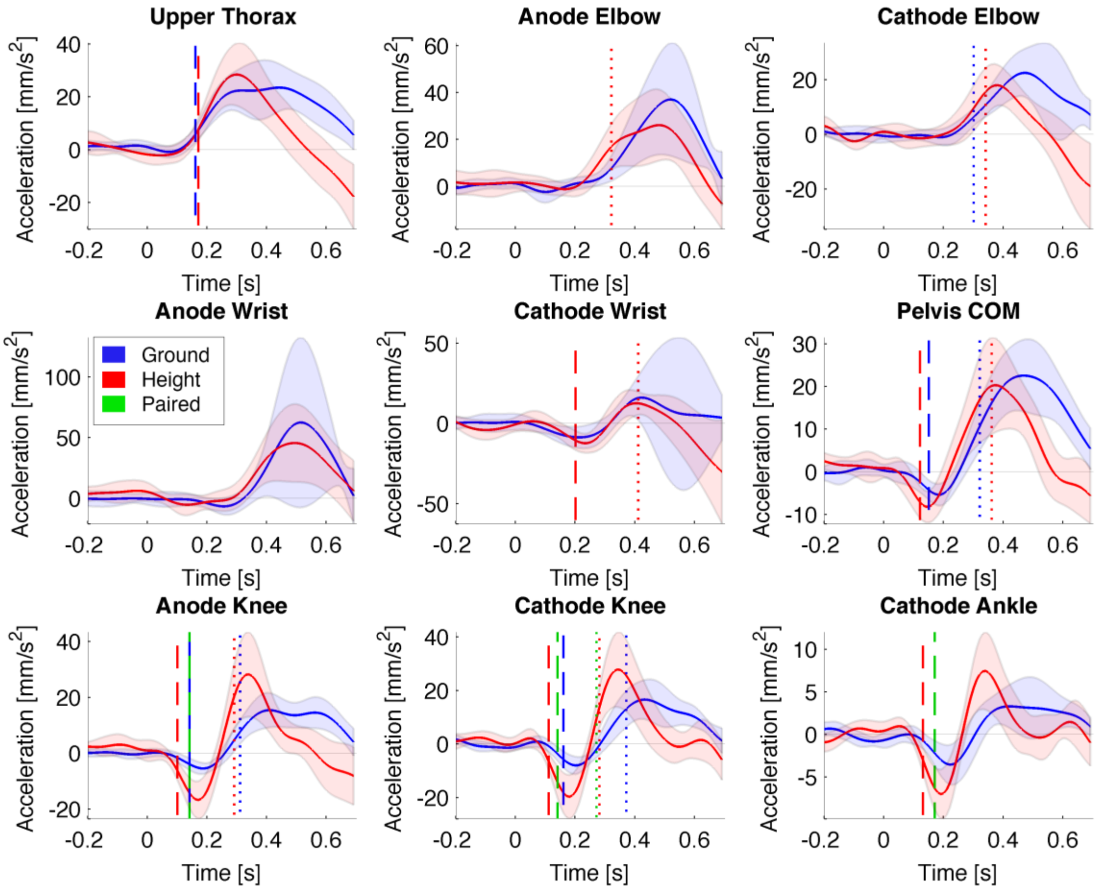

Figure 2.5: Cathodal acceleration around ~0.2 s in pelvis and lower extremities only. Data is shown of all nodes that are not included in Figure 2.4. Nodes are ordered from superior to inferior. Lines represent condition means and shaded areas represent 95% confidence intervals of the ground and height conditions. Positive values are mediolateral anodal acceleration and negative values are mediolateral cathodal acceleration. Vertical dashed and dotted lines represent the onset of significant short- and medium-latency acceleration, respectively. These vertical dashed and dotted lines are shown for significant effects in the ground and height conditions, as well as for the ground-height difference.

Response of the upper limbs: elbow and wrist nodes

(i) The upper limbs showed a clear anodal acceleration at medium-latency with the same size and timing as whole body anodal acceleration (Figure 2.5). The upper limbs were notable for their absence of response at short-latency timescales. Only the cathode wrist showed a significant cathodal response at short-latency. The amplitude was similar to the pelvis COM, therefore the pelvis acceleration could have been transferred mechanically to the cathode wrist.

(ii) The effect of height was to decrease the latency of the reduction in anodal acceleration (Figure 2.5).

Summary of GVS response revealed by node movements

Figure 2.6 provides a sequential overview of the GVS response and the effect of height for all body nodes. The GVS response comprises an early anodal acceleration of the head and upper thorax, a short-latency cathodal acceleration of the pelvis and lower limbs and a medium-latency anodal acceleration of the whole-body COM resulting in sustained anodal sway of the whole body. Cathodal acceleration had a short-latency origin and was restricted to the pelvis and lower limbs. The effect of height-induced fear of falling on vestibular reflexes was only significant in acceleration of lower extremity nodes.

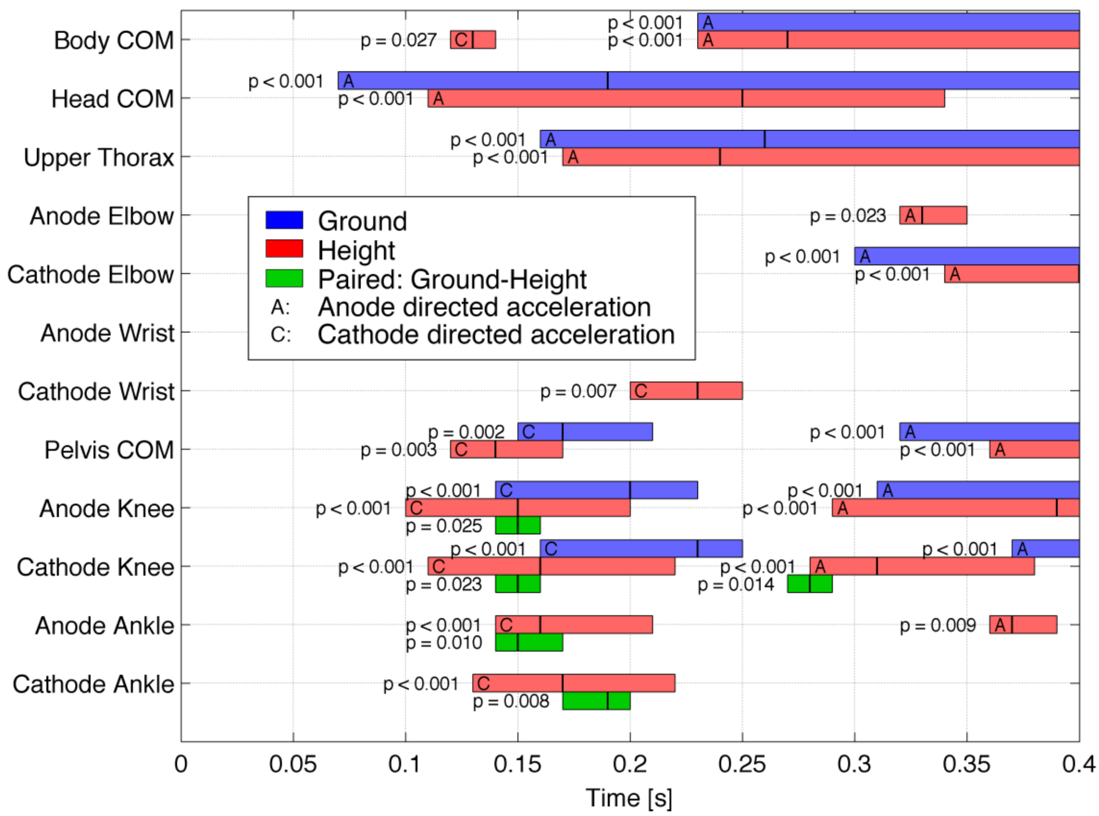

Figure 2.6: Body node acceleration: Significant time domains at ground vs. height. The bars show significant time domains of the SPM one-sample t-tests for ground and height, and the SPM paired t-tests on the ground-height difference. Vertical lines within each supra-threshold cluster bar indicate the time of maximum significance. The p-value of each cluster is shown left of each bar. Significant short-latency ground-height differences within 0.14 – 0.2 s was found in acceleration of lower extremity nodes only. A significant medium-latency ground-height difference was found for cathode knee only from 0.27 s to 0.29 s.

Figure 2.7 shows the mean displacement and acceleration at key time points. A video of the GVS response showing movement of stick figures comparable to Figure 2.7 can be found in Supplementary Material. At 170 ms, comparable with the GRF short-latency response, the cathodal acceleration and increased magnitude at height is evident at the ankle, knee and pelvis nodes. At 330 ms, comparable with the GRF medium-latency response, it seems that the acceleration and displacement of the whole body towards the anode is associated with cathodal buckling of the lower limbs centred at the knee, and that this effect was increased at height.

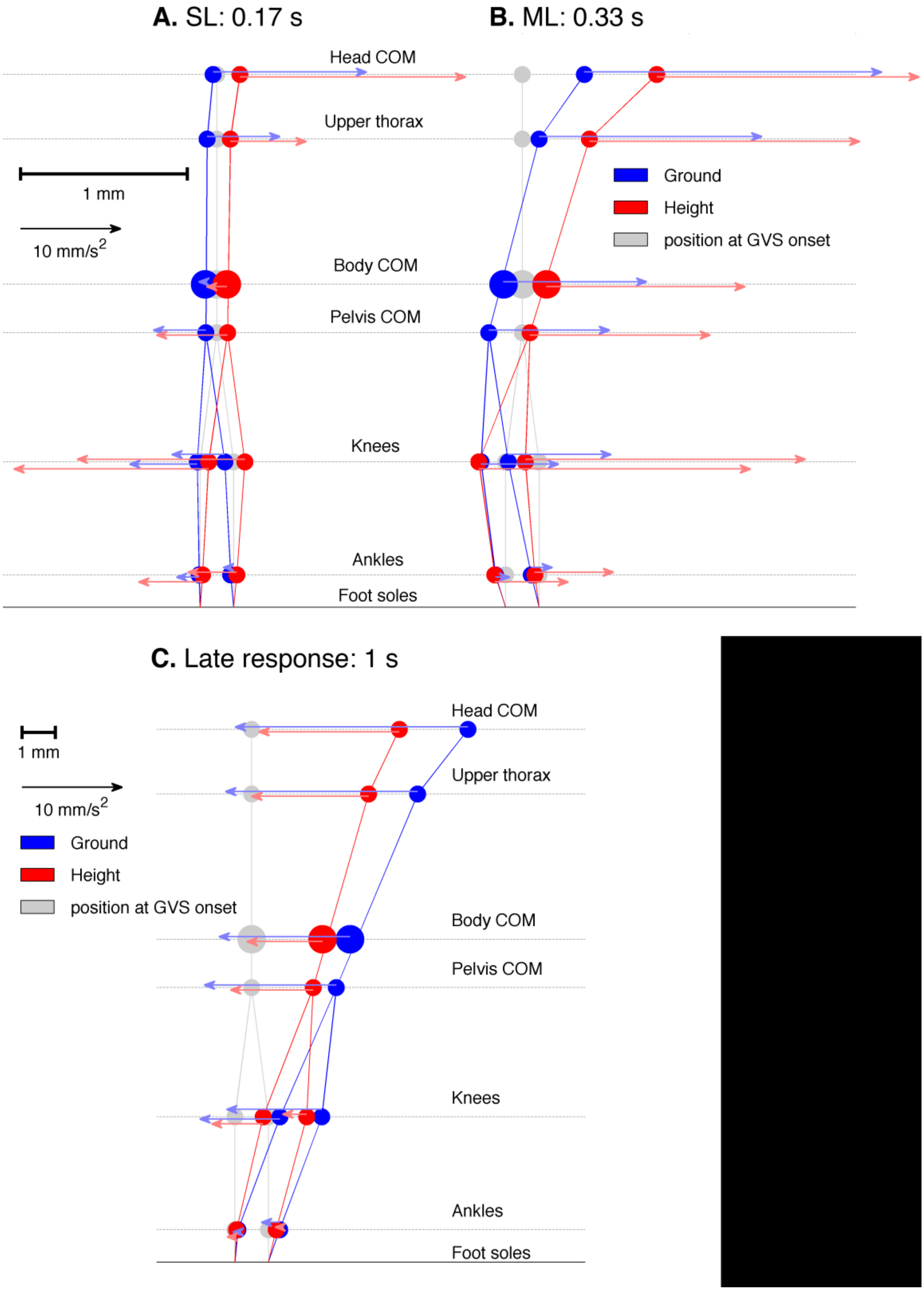

Figure 2.7: Nodes at different times after GVS onset. Dots and stick figures show mediolateral displacement of the head, trunk and lower extremity body nodes with respect to the position at GVS onset. This displacement is shown for 3 different points in time. For each stick figure the left side represents the cathode side and the right side represents the anode side. Arrows represent mediolateral acceleration. At the 3 time points, short-latency (***A***), medium-latency (***B***) and late (***C***) acceleration responses are shown. Mediolateral

displacement and acceleration scales are shown in the legend. Note that the node position scale for the lower stick figures (***C***) is 5 times smaller than the scale for the top stick figures (A-B). Inter-node distances are not scaled.

Analysis of joint and segment angle acceleration

Node movements result from a combination of joint rotations. For example head movement summarises the cumulative rotation of joints from the ankles to the neck. The following results are presented to remove ambiguity regarding the source of the node accelerations.

Anode and cathode flexion

Instead of anodal and cathodal, the direction of angular acceleration for joint and segment angles is indicated by anode and cathode flexion or roll acceleration. Anode or cathode flexion means that the segments on either side of the joint have moved towards folding together on the anode or cathode side of the joint. This terminology is comparable to anterior or posterior neck flexion, which indicates a folding together of the head and thorax on the anterior or posterior side of the neck.

Neck, lower back and head-in-space rotations

Linear anodal acceleration of the head and upper thorax were confirmed as arising from rotations at the neck and lower back (Figure 2.8 and Figure 2.9).

(i) GVS induced fast, consistent vestibular reflexes in the neck and lower back (Figure 2.8). In both conditions the VCR was faster than the vestibular reflex in any of the other joints (Figure 2.10).

(ii) Height had no significant effect on the magnitude of these reflexes (Figure 2.8G, H) which were remarkably consistent in magnitude and timing at ground and height (Figure 2.8A, B). However, these reflexes were more variable at height as shown by reduced significance of the GVS response (Figure 2.8G, H).

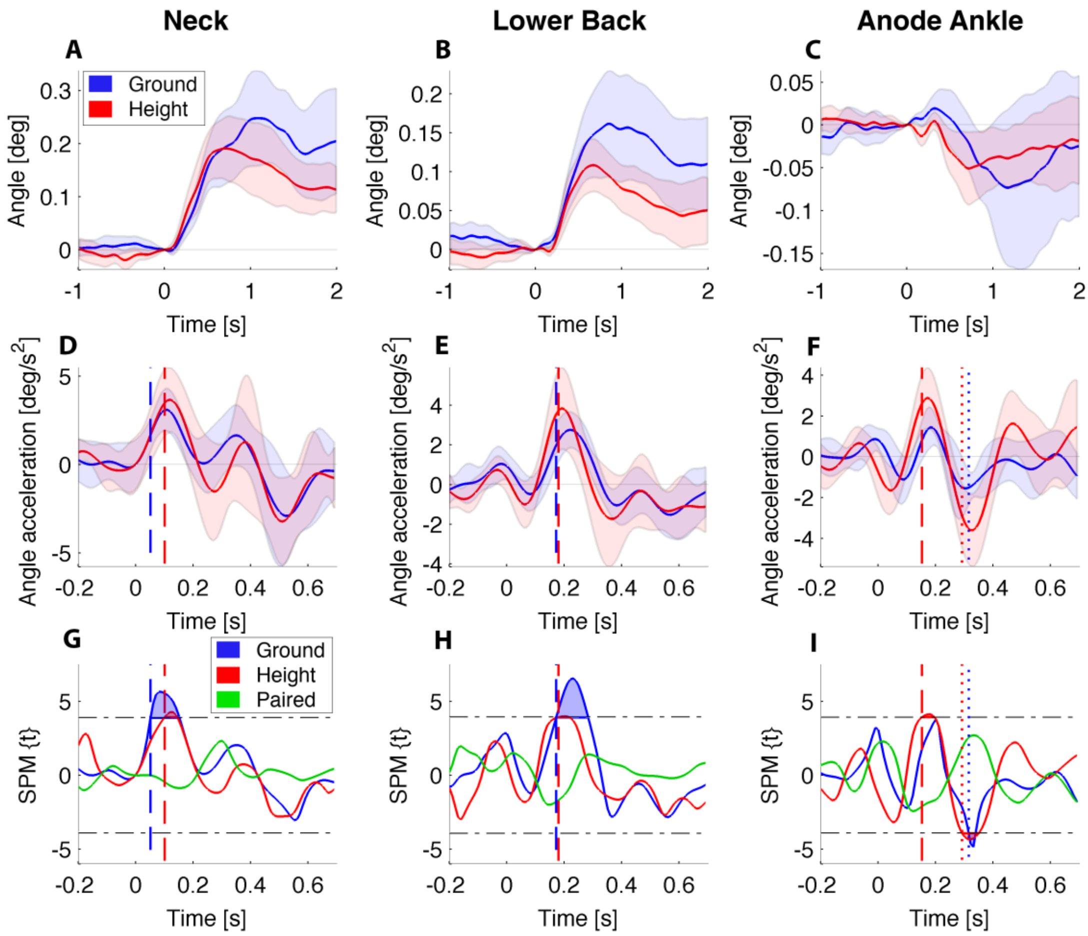

Figure 2.8: GVS effects in both conditions for angle accelerations, no ground-height difference effects. The *left, middle and right* columns of graphs represent neck lateral flexion, lower back lateral flexion and anode ankle lateral flexion, respectively. ***A-C***, The first row, shows lateral flexion angles. **D-F**, the second row shows angle acceleration. Lines represent condition means, and shaded areas represent 95% confidence intervals of the conditions (ground and height). Positive values are lateral flexion towards anode and negative values are lateral flexion towards cathode. ***G-I***, The bottom row, shows statistical parametric maps. Lines represent SPM{t} time series of the separate one-sample t-tests for ground and height data and paired t-tests for the ground-height difference. Horizontal dash-dot lines are the thresholds of significance and shaded areas are supra-threshold clusters that indicate the time domains with significant effects. GVS onset occurs at 0 s. Vertical dashed and dotted lines represent the onset of significant short- and medium-latency acceleration, respectively. These vertical dashed and dotted lines are shown for significant effects in the ground and height conditions. No significant ground-height difference effect was found in any of the measured angles.

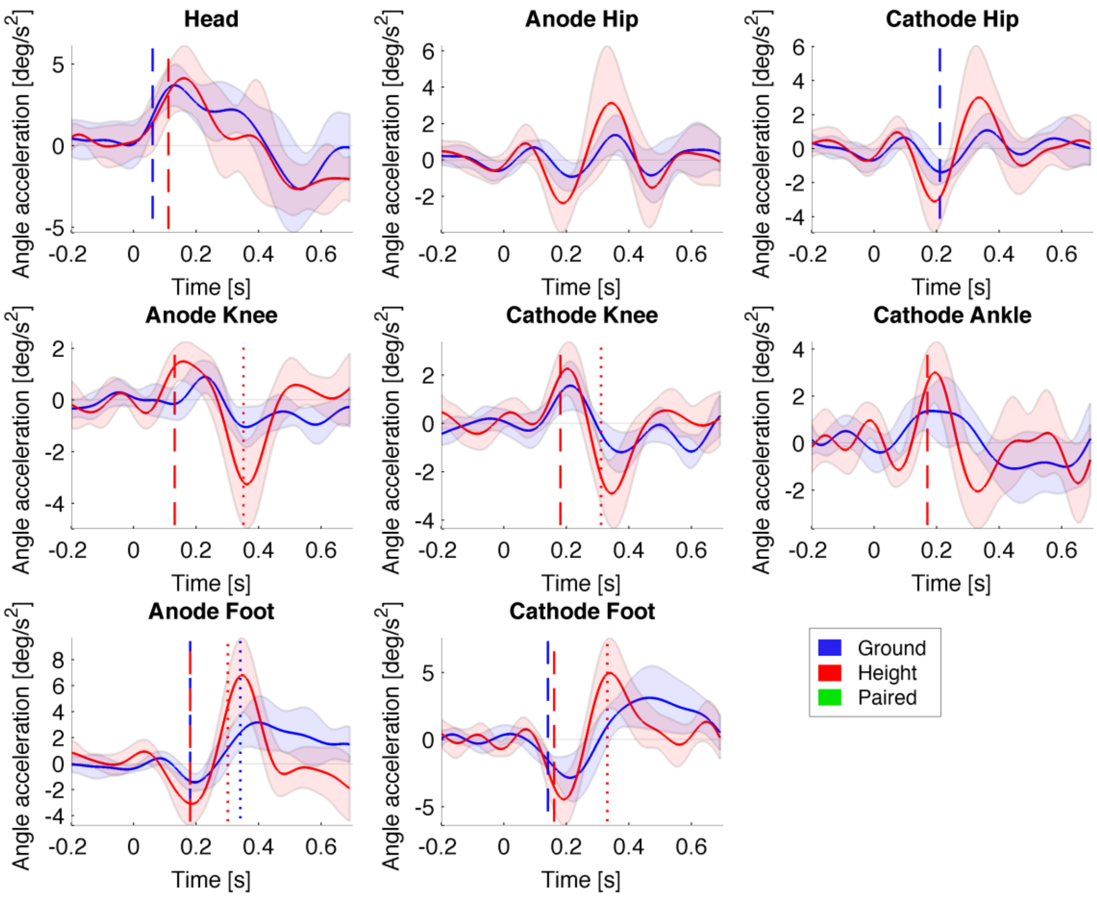

Figure 2.9: Opposite angle acceleration peaks at ~0.2 s and ~0.35 s in lower extremity angle accelerations. Angle acceleration is shown of all frontal plane angle variables except shoulder, elbow and wrist angles. Angle variables already shown in Figure 2.8 are also excluded. Lines represent condition means and shaded areas represent 95% confidence intervals of the ground and height conditions. Positive values are anode flexion/roll acceleration and negative values are cathode flexion/roll acceleration. Vertical dashed and dotted lines represent the onset of significant short- and medium-latency acceleration, respectively. These vertical dashed and dotted lines are shown for significant effects in the ground and height conditions. No significant ground-height difference effects were found for any of the measured angles.

Lower extremities rotations

Mediolateral linear acceleration of the ankle, knee and pelvis nodes was confirmed as arising from foot roll, and rotation at the ankle, knee and hip.

(i) GVS induced angular acceleration of the foot-in-space and in the hip and knee at short- and medium-latency. (ii) Height increased the statistical significance and size of angular accelerations at the knee, ankle and foot at short- and medium-latency (Figure 2.8 and Figure 2.9).

Upper limb rotations

Linear acceleration of the upper limbs was confirmed as arising from acceleration of the trunk. While some individuals showed upper limb joint rotations as a consistent group effect, GVS induced no significant acceleration in any of the shoulder, elbow and wrist joint angles.

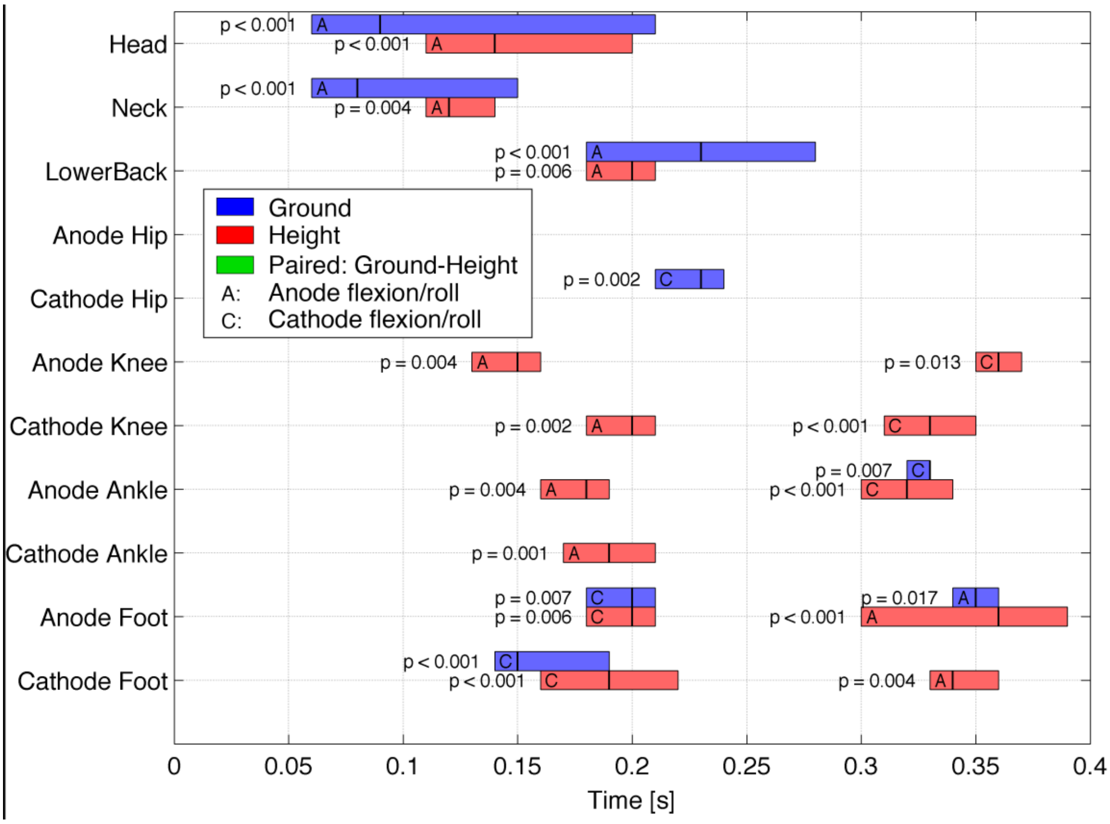

Figure 2.10: Angle acceleration: Significant time domains at ground vs. height. The bars show significant time domains of the SPM one-sample t-tests for ground and height, and the SPM paired t-tests on the ground-height difference. Vertical lines within each bar indicate the time of maximum significance per supra-threshold cluster. The p-value of each cluster is shown left of each bar. No significant ground-height difference was found in any of the measured angles.

2.4 DISCUSSION

The goal of this study was to investigate the effects of fear of falling on vestibular control of whole body balance. We used GVS and full-body kinematics to study this mechanism and below we elaborate on our main findings.

2.4.1 Short- and medium-latency vestibular reflexes are reflected in full-body kinematics

Our results show a unidirectional, anodal acceleration of the head COM and upper thorax in response to GVS. This is consistent with previous findings (Osler *et al.*, 2013). Our novel findings in the body COM, pelvis and lower limbs show a pattern of opposing cathodal and anodal acceleration (Figure 2.6 and Figure 2.10) that is consistent with the well-established short- and medium-latency GRF and EMG responses to vestibular stimulation (Britton *et al.*, 1993; Fitzpatrick *et al.*, 1994; Ali *et al.*, 2003; Fitzpatrick & Day, 2004; Son *et al.*, 2008; Mian & Day, 2009; Dakin *et al.*, 2010; Mian *et al.*, 2010; Muise *et al.*, 2012; Horslen *et al.*, 2014; Mian & Day, 2014), and is also consistent with a small cathodal sway preceding the larger anodal sway of the pelvis shown previously by Cathers *et al.* (2005) in their Figure 2. For reference, Figure 2.11 shows published GRF records of the short- and medium-latency responses and confirms that the timing of short- and medium-latency responses is consistent with our acceleration data. The short-latency cathodal acceleration is part of a lateral, buckling movement pattern of the lower limbs (Figure 2.7) supporting the idea that the source of force generation moving the whole body towards the anode is to be found in the lower limbs.

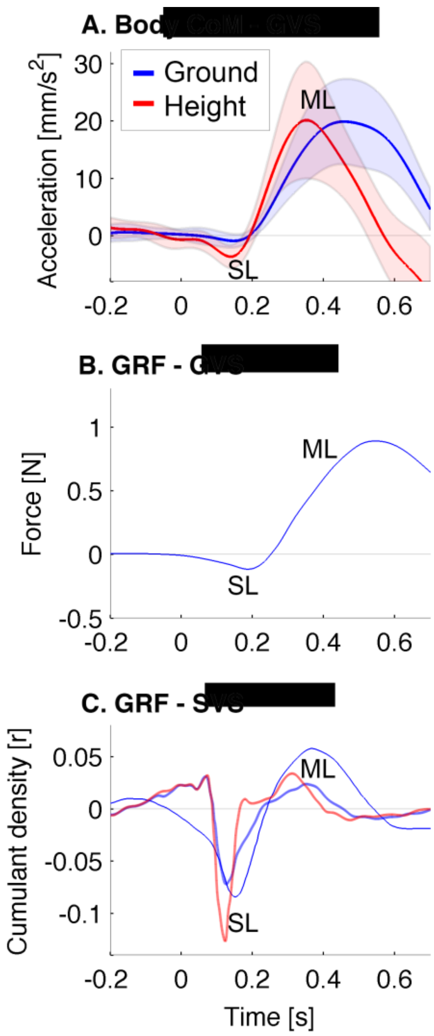

Figure 2.11: Short- (SL) and medium-latency (ML) responses in different publications. ***A,*** Mediolateral acceleration of body COM from this study at ground and height is shown. Acceleration towards the anode GVS electrode (ADA) is positive and cathode directed acceleration (CDA) is negative. 1mA GVS stimulation starts at 0 s with 2 seconds in duration. ***B,*** This graph is redrawn from Marsden *et al.* (2005). A 1 mA GVS of 3 seconds duration starts at 0s and the shear GRF is plotted. GRF towards anode is positive and towards cathode is negative. Participants stood at ground level. ***C,*** SVS-GRF coupling (cumulant density) is shown as a function of the SVS-GRF time lag. GRF-SVS (2-25 Hz) cumulant density of participants standing at low and at high altitude is shown by the thick lines. This data is redrawn from Horslen *et al.* (2014) so that positive values indicate coupling of vestibular stimulation (SVS) with shear GRF towards anode and negative values indicate coupling of SVS with shear GRF towards cathode. The thin line shows GRF-SVS (1-20 Hz) cumulant density data at ground level redrawn from (Mian *et al.*, 2010). The short- and medium-latency (SL and ML) responses follow a pattern that is comparable to the short- (CDA) and medium-latency (ADA) responses found in the body COM and lower body nodes with GVS in this study.

2.4.2 The short-latency response contributes to balance control

The contribution of the short-latency response to balance control and its neural underpinnings has been debated in the literature. Despite questions regarding the sensory origin, the principle of craniocentricity holds for the short-latency as well as the medium-latency response (Mian *et al.*, 2010). The principle of craniocentricity states that the direction of the balancing sway response to GVS is determined by head orientation. The short-latency response is known to be part of this craniocentric balancing response, however its contribution to balance has remained unclear (Fitzpatrick & Day, 2004).

Consistent with a semi-circular canal origin, we propose that the short- and medium-latency responses are coupled into a combined balance reflex. The short-latency response is the first stage of the combined balancing response, which generates the whole-body sway towards the anode electrode. Expression of the balancing response through EMG, GRF and kinematic acceleration depends upon the configuration of the body, the direction of illusory rotation and the intensity of the GVS stimulus.

In both ground and height conditions, GVS caused a rapid generation of lateral anodal body sway to counter the illusory rotation. Biomechanically, acceleration of the whole body COM requires rapid, active generation of an internal muscular moment on the trunk relative to the ground. Generation of a moment on the trunk relative to the ground occurs via muscles distributed across ankles, knees, hips and lower back. Acceleration of the linked segments is inversely proportional to their inertia, therefore the lightest segments and nodes show largest acceleration and the cathodal 'buckling' is most visible at the knee (Figure 2.7). Hence, the short-latency cathodal acceleration of pelvis and lower extremities seems to be part of the attempt to rapidly generate anodal movement of the whole body COM. The associated expression in GRF and EMG data likely reflects the same mechanism.

Furthermore, the generation of whole body acceleration depends upon the difference between internal muscular and external gravitational moments. In addition to generation of muscular moments, the standing configuration allowed the possibility to alter the gravitational moment by changing the base of support. The observed roll of the foot and cathodal displacement of the ankle, knee and hip (caused by the short-latency response) increases the gravitational moment relative to the ground. This gravitational moment induces acceleration of the whole body in the opposite (anodal) direction, as shown by the medium-latency acceleration. To illustrate, one could compare this method by balancing an upright stick on the palm of your hand by moving the hand in the horizontal plane. In this case the hand moves the base of support and changes the gravitational moment on the stick.

Mian *et al.* (2010) proposed that the short- and medium-latency responses may in fact be independent responses. As well as standing upright, Mian *et al.* (2010) applied SVS also to participants standing with their neck flexed anteriorly, so they were looking down. Because of the craniocentric response to vestibular stimulation, the axis of illusory rotation induced by SVS was pointed up and down instead of anterior/posterior in the head-upright posture. The short- and medium-latency responses were measured with gastrocnemius medialis EMG and GRF data. Their results showed that the medium-latency SVS-EMG coupling response in the head-faced-down posture was attenuated, whereas the short-latency SVS-EMG coupling response seemed unaffected (compared to head-upright). Therefore one might assume that the short- and medium-latency responses are independent.

The authors assumed that the SVS-induced sensation of yaw rotation about the earth vertical axis in the head-faced-down posture does not contribute to postural balance control. However, for this interpretation to be correct, this vertical axis of rotation should intersect the whole body COM. In reality, in the head-faced-down position the vertically oriented vector of illusory rotation does not intersect with the participant's COM and passes in front of it. In this head down configuration one would expect an initial lateral acceleration response of the whole body COM, along the circumference of the arc around the vertical axis. The direction of this COM acceleration in the head down position should therefore be the same as for the head upright position. COM acceleration is proportional to GRF data. This explains why the observed short-latency GRF response pattern as found by Mian *et al.* (2010) was similar in both head up and head down conditions.

These insights update the preceding observation that the short-latency response has no effect on the GVS-induced whole body movements (Fitzpatrick & Day, 2004). In our study we measured vestibular reflexes in one postural configuration. We predict that for different body configurations, the role of the short-latency response in postural balance will be the same. However, the expression in EMG, GRF and movement will reflect a distinct pattern that is needed to balance the whole-body COM in that configuration.

2.4.3 Fear of falling influences vestibular balancing reflexes, but not the VCR

Whether and how fear of falling influences vestibular reflexes is currently debated (Horslen *et al.*, 2015b, a; Reynolds *et al.*, 2015b, a). Here we consider the early part of the response attributed to vestibular mechanisms only.

Our results show that fear of falling had *no* effect on the size or latency of the early acceleration of the head and upper thorax. Neck-generated acceleration of the head, as part of the VCR, was one of the most consistent responses. Only for the lower limbs early GVS-induced acceleration was significantly increased by fear of falling. Statistical significance was detected in movements that were remarkably small (Figure 2.4C). This confirms the sensitivity of our experiment and underscores that early acceleration of the head and upper thorax arising from angular acceleration of the neck and lower back were not influenced by fear. Fear thus increased the lower extremity acceleration at short- and medium-latency related to generation of sway of the whole body and hence regulation of balance.

Our findings are consistent with those of Horslen *et al.* (2014) who found an increased gain of the GRF-SVS response at short- and medium-latency as a result of postural threat. Our findings are also consistent with the seemingly opposing results of Osler *et al.* (2013) who found no effect of postural threat on early acceleration of the head and upper trunk. As they only collected kinematics of head and trunk but not of the lower limbs, they concluded that fear of falling does not affect the vestibular balance reflex. Our study shows that fear of falling *does* affect the vestibular balance reflex, as the reflex gain of short- and medium-latency responses found in lower limb kinematics was increased at height.

2.4.4 Axial head-in-space stabilisation is task-independent

The distinct effects of fear of falling indicate that short- and medium-latency lower extremity responses are governed by different mechanisms than thoracolumbar and neck muscle responses.

Vestibular afferents are used in different feedback pathways for different functional purposes. Regulating visual gaze, regulating the head-in-space to stabilise gaze and regulating the whole-body COM to maintain balance can be distinguished as separate goals with different underlying mechanisms (Day *et al.*, 1997). These goals are related hierarchically in the sense that balance of the whole body depends upon integration of vestibular with proprioceptive information, which depends upon vestibular regulation of the eyes (VOR) and of the head (VCR). Forbes *et al.* (2015) made a distinction between vestibular mechanisms that govern axial and appendicular reflexes. In this paper, muscles moving joints of the spine including the neck are referred to as axial. Muscles moving joints of the legs and arms are referred to as appendicular.

Neck muscles play a crucial role in the regulation of the position and orientation of the head-in-space. The VCR regulates head orientation through neck muscle contractions that counteract perceived head movement (Suzuki & Cohen, 1964). Vestibulocollic neural pathways innervating neck muscles mostly comprise three-neuron-arcs. They primarily originate from medial vestibular nuclei and response latencies of these pathways are short (~8-10 ms) (Watson & Colebatch, 1998; Forbes *et al.*, 2014). Additionally, the VCR latency response of the sternocleidomastoid (SCM) muscle response was found to be unaltered by manipulation of vision, external support, stance width and posture (Watson & Colebatch, 1998; Welgampola & Colebatch, 2001). Forbes *et al.* (2014) tested the effect of fixating the trunk and head position on the VCR with the idea that this fixation rendered the neck muscles irrelevant to head posture. The VCR was still present in the fixed condition and was therefore concluded to be task independent.

Furthermore, the thoracolumbar vestibular reflexes have not been studied as extensively as the VCR. However, Forbes *et al.* (2013) found erector spinae muscles to only respond to low frequency vestibular stimuli. Therefore they concluded that the contribution of these muscles to standing balance might be limited compared to lower extremity and neck muscles.

2.4.5 Appendicular whole body stabilisation is task-dependent

The whole body sway response is task-dependent and more flexible than the VCR. Day *et al.* (1997) studied the effects of changes in posture on the GVS response and concluded that the vestibular response is organised to stabilise the body rather than the head in space. Appendicular muscles are innervated through vestibulospinal tracts originating from the lateral vestibular nuclei. Direct and indirect connections via spinal interneurons to motor neurons of extremities have been found in animal studies (Lund & Pompeiano, 1968; Shinoda *et al.*, 1986). In humans, response latencies of ~50–60 ms were found for appendicular vestibular reflexes (Britton *et al.*, 1993; Fitzpatrick *et al.*, 1994; Day *et al.*, 1997; Ali *et al.*, 2003; Son *et al.*, 2008). These latencies are longer than expected for the presence of direct vestibulospinal connections. Therefore, Forbes *et al.* (2013) argued that modification and low-pass filtering occurs via spinal interneurons to improve control of the mechanical system.

2.4.6 Fear of falling affects appendicular balance reflexes, not head-in-space stabilisation

It is currently controversial whether and how fear influences vestibular reflexes of whole body movement (Horslen *et al.*, 2015b, a; Reynolds *et al.*, 2015b, a). This study provides evidence confirming that vestibular reflex gain of whole-body stabilisation is increased by height-induced fear of falling. Since standing at height influences the short- and medium-latency balance response, we conclude that fear of falling increases the gain of vestibular balance reflexes. We observed a decreased duration and increased rate of change of the COM acceleration pattern. The functional effect of fear of falling is an earlier arrest of anodal sway, halving the distance moved by the whole-body COM towards the dangerous edge (Figure 2.3).

Our results also provide evidence that regulation of head orientation through the VCR is not influenced by fear of falling. The effect of fear of falling seems to lie not in the immediate vestibular processing serving the VCR, but in the effect of vestibular sensation, which impacts the control of balance. The vestibular balance reflex acts only through muscles engaged in balance. As such, our results update the view of Day *et al.* (1997) as we found that the GVS response contains separate components related to head regulation uninfluenced by fear, and components related to balance which are influenced by fear.

As discussed by Fitzpatrick and Day (2004), between immediate vestibular processing and regulation of balance there is a process of coordinate transformation from head-in-space to body-in-space and a process of gating or selection of biomechanically appropriate muscles. This chain of events might be influenced by fear. Possible targets for modulation include the vestibular cortex, the lateral vestibular nuclei, vestibulospinal tracts and subsequent spinal processing (Fitzpatrick & Day, 2004; Forbes *et al.*, 2015).

In our study, the GVS response was dependent on the randomly ordered polarity directions (anode left or right) and effects of height-induced fear of falling were not found in axial vestibular reflexes. Therefore, a general over-excitation of the motor neuron pool as a result of increased fear of falling, would not explain our findings. Fear-inducing stimuli are associated with activity of the amygdala. Consequently, two pathways between the amygdala and vestibular nuclei could be involved, one via the parabrachial nucleus and one via the vestibular cortex (Lang *et al.*, 2000; Balaban & Thayer, 2001; Balaban, 2002; Staab *et al.*, 2013).

To summarise, axial and appendicular GVS reflexes are distinguished by several features. These include invariance of latency and magnitude of the response to fear of falling, and absence of cathodal acceleration at short-latency. These different properties may reflect differences in innervation (medial vs. lateral vestibulospinal tracts) and different functional goals (head stabilisation vs. whole body balance).

2.4.7 Implications for fear of falling

Clinically, an important question is how and by what mechanisms balance responses are influenced by fear (van Dieen *et al.*, 2015). Our findings show that fear influences vestibular balancing reflexes. The efficacy of balancing reflexes is central to the risk of falling. However, it is important to note, that while fear of falling increases the gain of this primitive balance reflex, it remains undetermined whether this leads to an increase or decrease in the risk of falling in the general population and also in elderly persons with a fear of falling. Efficient balance control enables mobility. Hence, future studies could investigate whether the effect fear of falling on vestibular reflexes, increases or decreases mobility in the general population and in the elderly population. Additionally, the asymmetric decline of sensory and vestibular function with ageing may leave individuals vulnerable to the influence of fear on vestibular processing (Horak *et al.*, 1989; Baloh *et al.*, 1993; Kristinsdottir *et al.*, 2000). Patient-specific identification of the origin of balance performance decline is required and follow-up studies with elderly persons and clinical subgroups could clarify mechanisms relating fear of falling to balance and mobility.

2.4.8 Conclusion

In this study galvanic vestibular stimulation was used to evoke vestibular body sway reflexes, while participants stood at height to induce fear of falling and at ground level. The fast vestibular axial reflex acceleration for the head and thorax was unaltered at height. However, reflex-induced acceleration of lower extremities was increased at height. These results illustrate how balancing vestibular reflexes are influenced by fear of falling, whereas head stabilisation seems to be governed by different mechanisms that are unaffected by fear of falling. The findings in this study offer a novel interpretation of the short- and medium-latency responses of vestibular balancing reflexes. Traditionally the kinematic GVS sway response is described only as anodal roll of the pelvis, trunk and head segments. However, cathodal acceleration in the non-rigid lower extremities, observed at short-latency, was shown to be part of the appendicular mechanism generating anodal whole body acceleration from lower extremity muscles. In the literature only the medium-latency response has been assumed to cause the GVS induced body sway. However, we propose that both the short- and medium-latency reflexes functionally contribute to whole body balance and are biomechanically coupled as one coordinated response.

Chapter 3

Effects of attention on balancing responses to perturbations during walking in elderly

Introduction: A fall is one of the main causes of injury-related hospitalisation and injury-related deaths. We investigated the effect of internal vs. external attention and fall history on perturbed walking stability in healthy older adults.

Method: Participants' gait was perturbed through randomly occurring unilateral decelerations on a split-belt treadmill to evoke balance recovery movements. The internal focus of attention instruction was: "Concentrate on the movement of your legs", while the external focus of attention instruction was: "Concentrate on the movement of the treadmill". In both conditions participants' were asked to look ahead at a screen. Outcome measures were coefficient of variation of step length and step width, and the centre of mass velocity time series as analysed using statistical parametric mapping.

Results: After each perturbation participants took two to three strides to regain a normal gait pattern, based on the centre of mass velocity response. No significant difference was found between the effects of internal and external focus of attention instructions on walking stability parameters of perturbation responses based on any of the outcome measures.

Discussion: We conclude that, compared to an internal focus of attention instruction, an external focus of attention to the walking surface does not lead to improved balance recovery responses to gait perturbations in the elderly.

Adapted from: de Melker Worms, J. L. A., Stins, J. F., van Wegen, E. E. H., Verschueren, S. M. P., Beek, P. J., Loram, I. D. (2016). Effects of attention and fall history on perturbed walking stability in elderly. *Manuscript submitted for publication.*

3.1 Introduction

A fall is one of the main causes of injury-related hospitalisation and injury-related deaths in the elderly (Rubenstein, 2006). It has been suggested that fall risk and decline of balance performance in the elderly are not solely related to physical degeneration; psychological factors such as attentional focus strategies may be involved as well. Some studies suggest that individuals with increased fall risk have heightened conscious attention to their own movements, which otherwise would be more automated and require less attentional control (Wong *et al.*, 2008; Wulf, 2013; Young *et al.*, 2015).

In the motor control and learning literature, a distinction is made between an external and an internal focus of attention, which purportedly have differential effects on motor performance. Wulf & Prinz (Wulf & Prinz, 2001b) described an internal focus of attention as directing the performers' attention to movement of their own body, e.g. towards movements of their feet while standing on an unstable balance board (McNevin *et al.*, 2003; Chiviacowsky *et al.*, 2010; McNevin *et al.*, 2013). In contrast, an external focus of attention was described as directing attention to the effect of the movement in the environment, e.g. the trajectory of a golf ball relative to the hole (Bell & Hardy, 2009) or the movement of a balance board or platform one is standing on (McNevin *et al.*, 2003; Chiviacowsky *et al.*, 2010; McNevin *et al.*, 2013). In some tasks the goal is not to move or act upon an external object, but to control movement of the body itself. In that case external attention comprises directing attention to the surface on which force is exerted by the human performer and which is relevant to successful motor performance, e.g. the ground one is standing on in gymnastics (Lawrence *et al.*, 2011; An *et al.*, 2013; Wulf, 2013).

According to the constrained action hypothesis (McNevin *et al.*, 2003), an external focus of attention facilitates performance on challenging motor tasks, as it allows more 'automatic' or 'efficient' control mechanisms to come into play, compared to an internal focus of attention. Furthermore, an internal focus of attention is thought to place a constraint on previously internalised ‘automatic’ movement by consciously controlling (part) of the movement, which reduces performance quality (Wulf & Prinz, 2001b; McNevin *et al.*, 2003; Landers *et al.*, 2005; Wulf *et al.*, 2009; Freudenheim *et al.*, 2010; Lohse *et al.*, 2010b; Wulf *et al.*, 2010).

Additionally, when older adults attempted to learn a new balance task, balance performance increased faster with an external focus of attention compared to an internal focus of attention (Chiviacowsky *et al.*, 2010). One could speculate that elderly with a fall history might also adopt a more internally directed focus of attention as a protective strategy; especially when walking stability is challenged. Furthermore, physical therapists have been found to employ more internal than external focus of attention instructions and feedback in gait re-education, which might attenuate motor learning (Johnson *et al.*, 2013).

However, to our knowledge it has never been investigated whether attentional instructions alone can alter gait stability in the elderly, and whether this effect is modulated by fall history. In this study we investigated the combined effects of fall history and attentional focus on gait performance in healthy elderly. To test gait stability we applied mechanical perturbations during treadmill walking (Bruijn *et al.*, 2010; Granacher *et al.*, 2010).

3.1.1 Aims and hypotheses

Our main hypothesis is that an external focus of attention temporarily leads to a more stable perturbed walking pattern compared to an internal focus of attention. To challenge gait stability we applied randomly occurring unilateral mechanical perturbations on a split-belt treadmill, and recorded the ensuing biomechanical process of balance recovery. Such perturbations are experienced as a forward slip of the foot, e.g., when walking on a slippery surface. Fall history and decreased gait stability are associated with increased variability of gait (Toebes *et al.*, 2012). We therefore hypothesised that compared to an internal focus attention, an external focus of attention during walking would lead to (1) decreased variability of perturbed step length and step width and (2) faster recovery to a stable gait pattern based on changed centre of mass (COM) velocity profiles. In addition, we examined whether the effect of attentional focus on gait stability is dependent on the fall history of the participants.

3.2 Method

3.2.1 Participants

Twenty-eight healthy older adults (8 males, 20 females) aged 65 or above, who were able to walk independently for at least 10 minutes, were recruited. The average participant age was 69.3 ± 3.7 years (Mean ± standard deviation; range: 65-78). A Dutch version of the Mini-Mental State Examination (MMSE) was used to determine the cognitive status of participants. Participants with a MMSE score below 25/30, any history of rheumatoid arthritis in lower extremities, cerebral vascular disease, Parkinson’s disease, peripheral neuropathy, cardiac arrest, bypass treatment or any other neurological or cardiovascular impairment were excluded. The study received approval from the local ethical committee and participants gave written informed consent prior to their participation.

3.2.2 Material

Participants walked on the Gait Real-time Analysis Interactive Lab (GRAIL) system (Motekforce Link b.v., Amsterdam, The Netherlands). The GRAIL system consists of an instrumented split-belt treadmill in combination with a Virtual Environment (VE) projected on a 180° semi-cylindrical screen (Figure 3.1).

As stated, temporary unilateral treadmill decelerations were used as gait perturbations in the experiment. The VE in this experiment was a virtual straight road, surrounded by a forest and mountains to create realistic optical flow while walking. Motekforce Link's D-flow software was used to control the system. Ten high-resolution infra-red cameras (Vicon, Oxford, UK) and the Human Body Model (HBM, Motekforce Link) full body marker set were used to capture kinematic data at 100 Hz using 47 passive retroreflective markers (van den Bogert *et al.*, 2013). A safety harness system suspended overhead prevented the subjects from falling; however no weight support was provided.

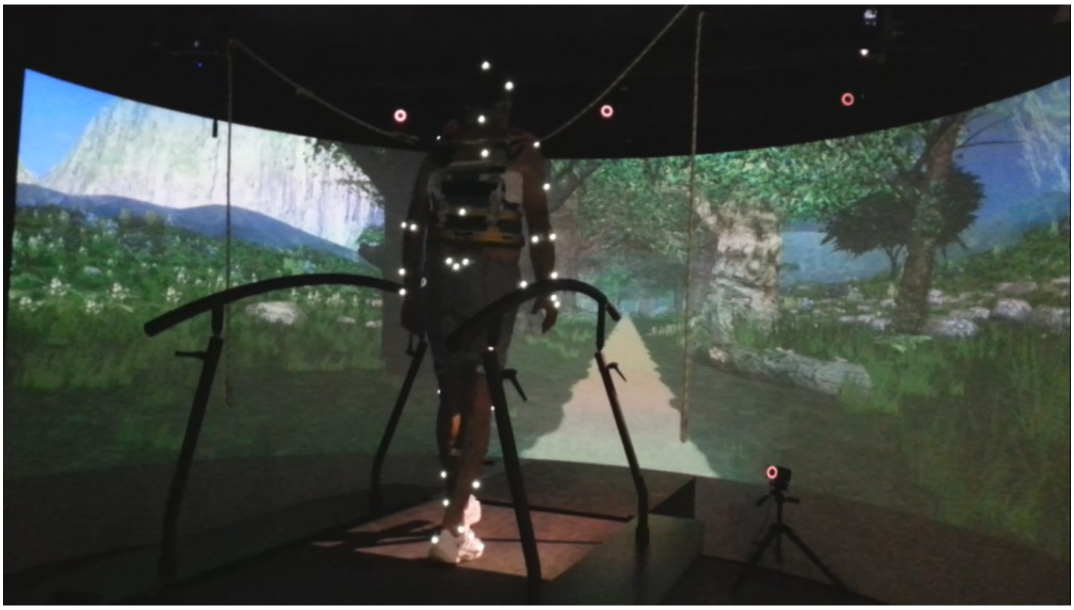

Figure 3.1: Virtual environment.

3.2.3 Fall history

Prior to the experiment participants filled out details about their fall history. A fall was defined as an event in which a person unintentionally comes to rest on the ground or other lower levels (de Zwart *et al.*, 2015). Participants who had experienced one or more falls within 12 months before the experiment were labelled as fallers, while the other subjects were labelled as non-fallers. Falls that resulted from loss of consciousness or acute paralysis caused by stroke, epileptic attacks or violence were excluded.

3.2.4 Procedure

Participants were instructed to always look ahead at the screen and were familiarised with treadmill walking at a speed of 1 m/s including gait perturbations. In all trials this fixed speed was used. Perturbations consisted of short unexpected unilateral decelerations of the split-belt treadmill on the participant's dominant leg side only, which occurred at random intervals between 10 and 20 seconds. Unilateral treadmill decelerations initiated at toe off of the dominant foot. At the following heel strike of the same foot the belt was decelerated to 0 m/s. This resulted in a motor response resembling a forward slip of the foot. At the next heel strike of the same foot, the belt had regained the original velocity of 1 m/s.

The experiment comprised two perturbed gait trials of five minutes per participant, one for the internal focus of attention condition and one for the external focus of attention condition in counter-balanced order. For each condition 20 perturbations were given. In the internal focus of attention condition participants received the following verbal instruction: "Look ahead at the screen and concentrate on the movement of your legs", while in the external focus of attention condition they received this instruction: "Look ahead at the screen and concentrate on the movement of the treadmill". Instructions were repeated every 30 seconds during the trials using a speaker system. As this experiment was part of a multi-experiment protocol, participants had already walked 1 m/s for 20 minutes at the start of this particular experiment.

Data analysis: Step length & step width

The mean step length and step width of the first recovery step following each perturbed heel strike was determined based on heel and toe marker positions. Furthermore the coefficients of variation (*CV*) of step length and step width was calculated for each participant as a percentage of the mean, see equation (1).

$$CV(\%) = 100 \times \frac{\text{standard deviation}}{\text{mean}}, \qquad (1)$$

Step length and step width data was analysed using Matlab (version R2014a, The MathWorks, Inc., Natick, MA, USA).

3.2.5 Data analysis: Normalised Euclidean distance (D)

The normalised Euclidean distance was calculated as a measure of the amount of deviation from a participant's normal gait pattern. From the internal and external focus of attention walking episodes, participants' body COM was calculated using Visual 3D (v5.02.07, C-Motion Inc., Germantown, USA). The velocity of the X-, Y- and Z-time series of the COM was calculated through differentiation using a 4rd order Savitsky-Golay filter with a temporal window of 90 ms (Press *et al.*, 1999). These time series were then normalised using spline interpolation, so that every stride consisted of 100 samples. The COM velocity data between 4 s after each perturbation up until the next perturbation were classified as unperturbed walking (UW) bouts. For each subject and condition (internal vs. external focus of attention) the UW bouts of these time series were combined to create an average limit-cycle for each subject and condition. This limit-cycle represents the average COM behaviour at each percentage of an unperturbed stride in that condition. Furthermore, for each percentage in this limit cycle, the standard deviation in unperturbed walking (v_{UW}) was calculated for each dimension. Walking bouts ranging from the first stride before each perturbation until the fourth stride after the perturbation were classified as perturbed walking (PW) bouts. The normalised Euclidean distances (*D*) of the COM velocity time series between PW bouts and the average limit cycle (UW) were then calculated as described by Bruijn *et al.* (2010), see equation (2).

$$D(k \times 100 + i)_{\substack{k=0:n-1 \\ i=1:100}} = \sqrt{\sum_{d=1}^{3}((\mathrm{UW}(i)_d - \mathrm{PW}(\mathrm{k} \times 100 + i)_d) \,/\, v_{\mathrm{UW}}(i)_d)^2}\,, (2)$$

$D(k \times 100 + i)$ is the normalised distance (in standard deviations) for *i* % of stride *k*+1 (with *n* representing the maximum number of strides in PW); *d* is the dimension number, UW is the limit cycle, PW is the state of the perturbed walking trial, and v_{UW} is the variability of the limit cycle. The COM data was analysed using Matlab.

3.2.6 Step length and step width statistics

A 2×2 mixed ANOVA including effect sizes (partial eta squared) and Bayes factors were calculated to test whether participant means of step length and step width was significantly different between the internal and external focus of attention conditions, between fallers and non-fallers and whether fall history interacts with gait under the two attention conditions. The step width *CV* and step length *CV* data did not pass the Shapiro-Wilk test for normality. Therefore Wilcoxon signed-rank tests were used to compare differences between internal and external attentional focus. Fallers and non-fallers were compared with Mann-Whitney U tests. For fall history effects within attentional focus conditions, subsequent Mann-Whitney U tests with Bonferroni correction were used. For attentional focus condition effects within fallers and non-fallers, subsequent Wilcoxon signed-rank tests with Bonferroni corrections were used. For all tests on *CV* data, effects sizes (r) and Bayes factors were calculated as well. Statistics of means and *CV's* of step width and step length were calculated with IBM SPSS Statistics 20.0, except for the Bayes factors that were calculated with the BayesFactor v0.9.12-2 package for R (bayesfactorpcl.r-forge.r-project.org; R-project.org).

3.2.7 Statistical Parametric Mapping (SPM)

As our second hypothesis pertained to *D* at each percentage of the post perturbation strides, we used a validated method of time series analysis (i.e. SPM) to test whether the *D* time series are statistically different between conditions. All SPM analyses were implemented using the open-source toolbox SPM-1D (v.M0.1, Todd Pataky 2014, www.spm1d.org,) in Matlab R2014a. SPM regards the whole time series as the unit of observation and is now increasingly used in the analysis of kinematic time series (Pataky, 2012; Robinson *et al.*, 2014; Serrien *et al.*, 2015). This allows time dependence to be incorporated directly in statistical testing.

In this study a SPM two-tailed one-sample t-test was used separately for the internal and external focus of attention condition data to test whether *D* is different from the relaxation distance (α = 0.05). Additionally a SPM two-tailed paired samples t-test (Robinson *et al.*, 2014) was used for an internal vs. external focus of attention comparison of *D*. The scalar output statistic, SPM{t}, was calculated separately at each individual time sample. To test the null hypothesis, the critical threshold was calculated at which only α % (5 %) of the analysed trajectories would be expected to traverse. This threshold is based upon estimates of trajectory smoothness (Friston *et al.*, 2007) and Random Field Theory expectations (Adler & Taylor, 2007). Conceptually, a SPM t-test is similar to the calculation and interpretation of a scalar t-test; if the SPM{t} trajectory crosses the critical threshold at any time sample, the null hypothesis is rejected. However, a SPM t-test avoids the false positives of a scalar t-test and avoids the false negatives of a scalar t-test with Bonferroni correction (Adler & Taylor, 2007).

3.3 Results

3.3.1 Mean and CV of step width & step length

The mean and *CV* of step length and step width of the first recovery step following the perturbed heel strikes is shown in Figure 3.2. Inspection of the data revealed that three participants adopted a different recovery step strategy than the remaining participants. For these three participants the first response to the perturbation involved an initial abrupt back stepping movement, after which a normal stepping pattern was resumed. Calculation of step length for these participants would result in negative values; therefore these three participants (one faller, two non-fallers) were excluded from the step length and step width analysis. The scatter plot in Figure 3.2 shows the data for the remaining 25 participants.

No significant difference was found for any of the spatiotemporal parameters between the internal and external focus of attention or between fallers and non-fallers. The interaction effect between fall history and attentional condition was also not significant. Furthermore for the main effect of attention, the Bayes factors for the *CV's* of step width and step length were smaller than 0.33. Therefore the odds for the null hypothesis (no difference) vs. the alternative hypothesis are higher than 3 to 1 for the *CV* variables, see Table 3.1.

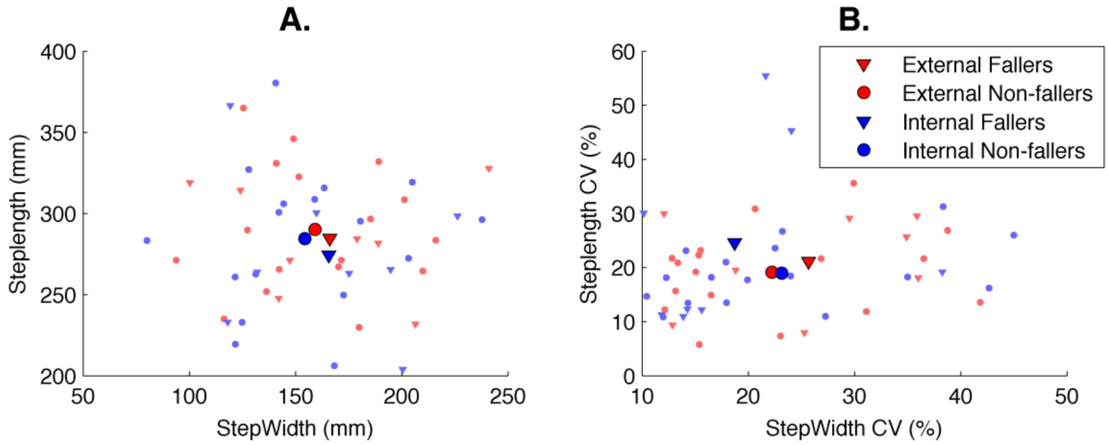

Figure 3.2: Means and coefficients of variation for step length and step width. The first step of each perturbed heel strike was included for this graph. The big dots represent the means per condition while the small dots represent the means for each participant in each condition. Panel ***A*** shows the average step length and step width and panel ***B*** shows the *CV's*. For both the means and *CV's* no significant difference was found between the internal and external focus of attention instructions or between fallers and non-fallers.

Table 3.1: Step width and step length statistics. For all *F* values df_1 = 1 and df_{error} = 23. The Bayes factor (BF_{10}) indicates the odds for the alternative hypothesis vs. the null hypothesis to be true. For the effect of attentional focus these odds are less than 1 to 3 for the *CV* variables. It has been recommended to label these Bayes factor values as moderate evidence for the null hypothesis (Lee & Wagenmakers, 2014).

	Mean (SD)		test stat	*p*-value	effect size	Bayes factor (BF_{10})
Attentional focus	*Internal*	*External*				
Step length (mm)	281 (44.5)	288 (36.9)	$F = 1.21$	0.28	$\eta^2 = 0.05$	0.35
Step width (mm)	158 (38.5)	161 (38.0)	$F = 1.03$	0.32	$\eta^2 = 0.04$	0.51
CV Step length	20.7 (10.8)	19.8 (8.1)	$Z = 0.65$	0.43	$r = 0.03$	**0.24**
CV Step width	21.7 (10.4)	23.3 (9.9)	$Z = 2.11$	0.16	$r = 0.05$	**0.28**
Fall history	*Fallers*	*Non-fallers*				
Step length (mm)	279 (36.5)	287 (31.5)	$F = 0.23$	0.63	$\eta^2 = 0.01$	0.42
Step width (mm)	166 (43.0)	157 (37.0)	$F = 0.31$	0.58	$\eta^2 = 0.01$	0.43
CV Step length	22.8 (12.0)	19.0 (5.3)	$U = 1.15$	0.30	$r = 0.05$	0.58
CV Step width	22.2 (9.0)	22.7 (11.3)	$U = 0.02$	0.90	$r = 0.00$	0.39

3.3.2 Euclidean distances

The averaged inferior/superior (up and down) COM position time series during perturbed and unperturbed walking is shown for a representative participant in Figure 3.3. It shows how the perturbation causes the time series to diverge for both the internal and external focus of attention.

The normalised Euclidean distances (*D*) and the corresponding SPM analysis are shown in Figure 3.4. After perturbation the distance to the unperturbed walking pattern quickly increased and then gradually moved back to the relaxation distance. This relaxation distance resulted from the natural variability of unperturbed gait, i.e. UW bouts. For both conditions the perturbations caused a COM velocity response that was significantly different from unperturbed walking for more than one stride after perturbation onset.

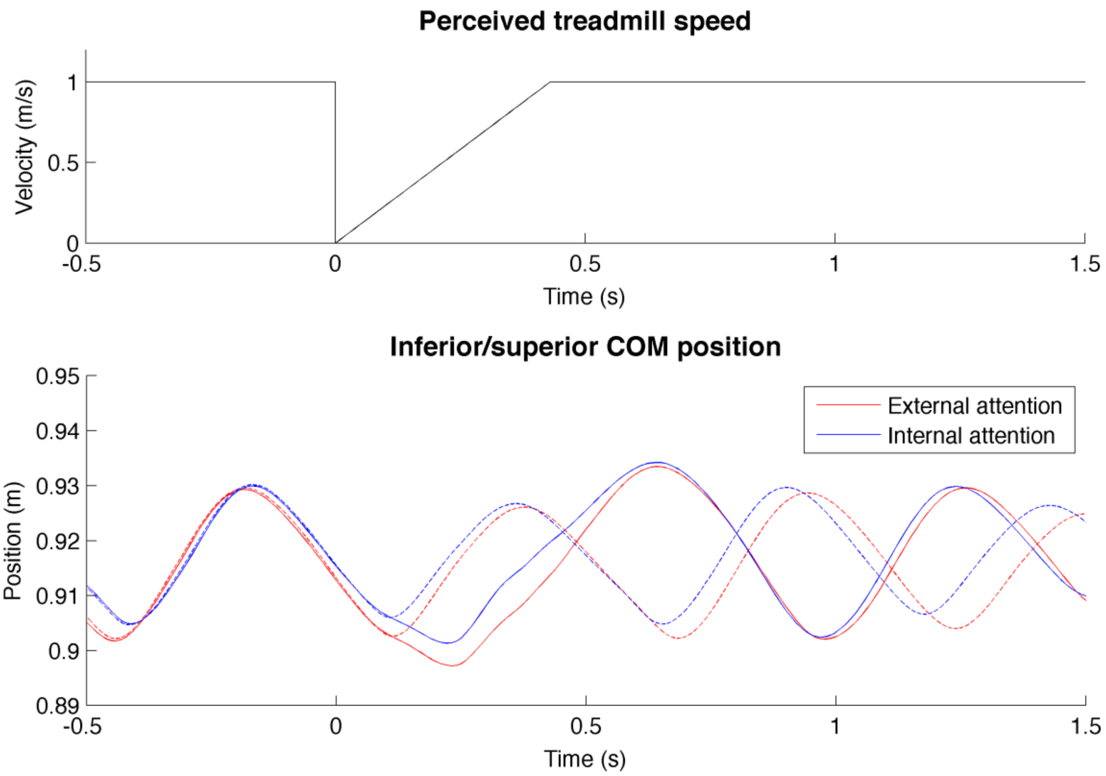

Figure 3.3: Example data of representative participant. The walking perturbations consist of a unilateral treadmill deceleration of the split-belt treadmill on the participant's dominant leg side. For each perturbation the treadmill deceleration starts at toe off when there is no more contact with the dominant leg side of the treadmill. At the next heel strike the treadmill velocity on that side is 0 m/s and starts accelerating again. The top panel shows the perceived speed of the perturbed side of the treadmill. The perturbed heel strikes occur at 0 seconds. The bottom panel shows the inferior/superior position of the participant's COM. The red and blue lines show the mean responses of the participant to the perturbations in the external and internal attention conditions, respectively. The red and blue dashed lines show the unperturbed COM movement where unperturbed heel strikes also occur at 0 seconds.

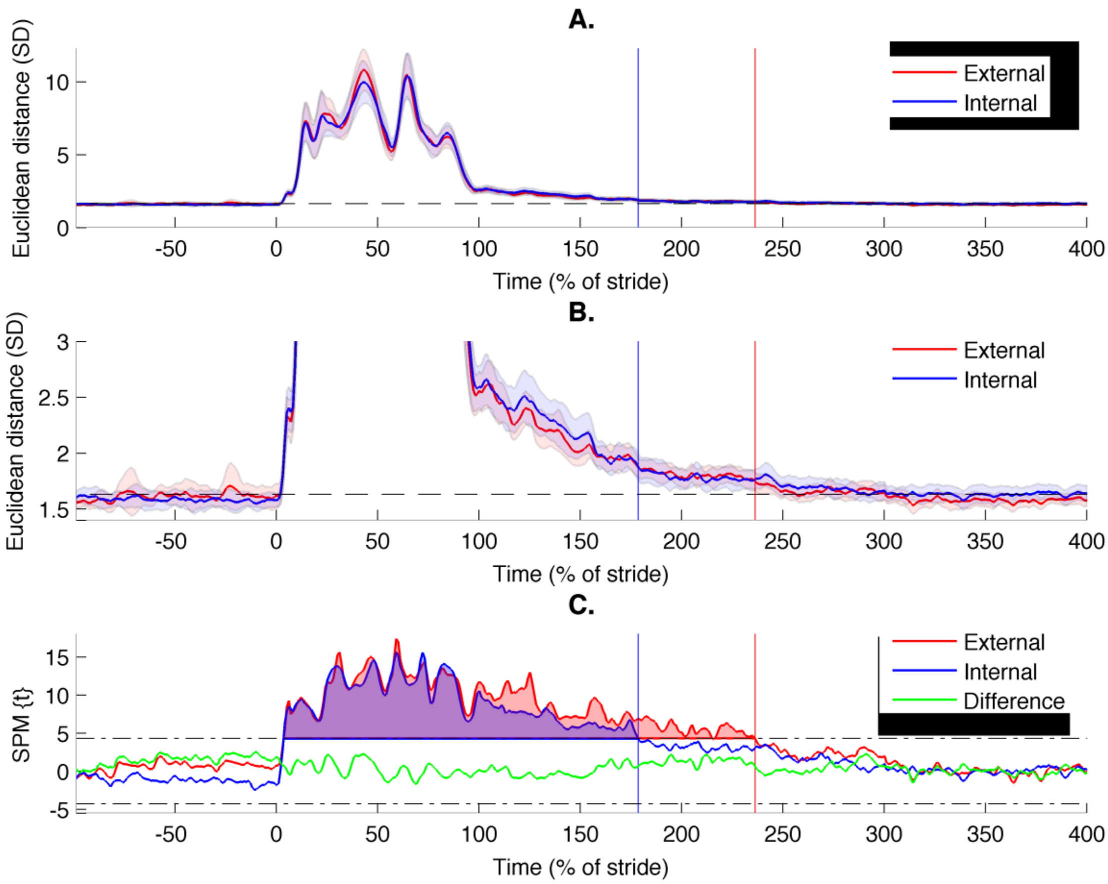

Figure 3.4: COM velocity analysis.

A shows the Euclidean distance of the perturbed response COM velocity to the average unperturbed gait COM velocity. Data was normalised to stride percentage with 100 samples per stride. Each stride started at heel strike of the dominant leg, perturbed heel strikes occur at 0%. Shaded areas indicate 95% confidence intervals. The horizontal dashed line indicates the relaxation distance of unperturbed gait.

B is a vertically zoomed-in version of panel A to visualise the late response after 100%.

C shows SPM graphs of internal, external and the difference between internal and external attention in red, blue and green respectively. Lines represent SPM{t} trajectories of the separate one-sample t-tests for external and internal data and paired t-tests for the external-internal difference. The SPM one-sample t-tests tested whether the internal and external time series from panel A were different from the relaxation distance. Horizontal dash-dot lines are the thresholds of significance. Shaded areas are supra-threshold clusters that indicate the time domains with significant effects. The vertical red and blue lines indicate the stride percentage at which COM velocity ceased to be significantly different from the relaxation distance of unperturbed walking. Even though these stride percentages are 58% apart for internal and external attention, no significant difference between internal and external attention was found for the Euclidean distances.

For internal attention the difference from unperturbed walking was significant from 4% of the first stride until 78% of the second stride (178%) after perturbation onset ($p<0.01$). For external attention the difference was significant from 4% to 236% ($p<0.01$). As the confidence intervals for the external focus of attention are slightly smaller than for internal attention between 178% and 236%, the internal attention SPM graph falls below the threshold of significance in that time window, whereas the external focus of attention SPM graph stays above this threshold. This difference is not caused by a difference of the mean responses between conditions. This is confirmed by the lack of a significant difference between conditions as indicated by the difference (green) graph. The origin of the difference in this time window lies in the slightly smaller between subjects variability in the external focus of attention condition compared to the internal focus of attention condition, as shown by the confidence intervals. So even though the stride percentages at which these effects cease to be significant for the internal and external focus of attention are 58% apart, no significant difference between conditions was found as shown by the SPM paired t-test graph (Figure 3.4).

3.4 Discussion

In this study we investigated the effect of attentional focus and fall history on walking stability as assessed by means of transient mechanical perturbations. No significant difference between internal and external focus of attention and between fallers and non-fallers was found for means and *CV's* of step length and step width of the first step following perturbation. This disconfirms our first hypothesis that an external focus of attention during walking leads to decreased variability of perturbed step length and step width compared to an internal focus of attention in elderly. Moreover, no significant effect of attentional focus was found in the COM velocity during the first four strides following each perturbation. This disconfirms our second hypothesis that an external focus of attention leads to faster recovery to a stable gait pattern in elderly than an internal focus of attention. In addition, fall history does not seem to affect the balancing responses following the walking perturbations.

In sum, the beneficial effects of an external vs. an internal focus of attention on motor performance do not seem to apply to balance control during walking.

3.4.1 Possible reasons for the absence of attentional effects

When the task-goal is to move and act upon an external object, directing one's attention to that object and the corresponding movement effect has shown to produce better performance than directing attention to one's own body movements (Wulf, 2013). Apparently, an external focus of attention provides information that is more useful to the planning and execution of goal-directed instrumental actions than an internal focus of attention. In the present experiment the participants' goal was not to achieve an environmental effect but to maintain an upright walking pattern. To this end, they had to control the movement and location of their own body and external focus of attention instructions could not be given in relation to achieving a particular environmental effect. Visual information about the surroundings aids to determine one's location. Therefore the instruction to look ahead at the screen could have been more useful to provide information about the participant's own location and movement than concentrating on the movement of the legs or the treadmill belt. In other studies where the participants' task was to produce a specific movement of their own body, performance benefits of an external focus of attention was found for the golf swing form (An *et al.*, 2013), but not for gymnastics (Lawrence *et al.*, 2011). In stroke patients an opposite effect was suggested as paretic leg movement performance was increased for an internal rather than an external focus of attention (Kal *et al.*, 2015).

The prevailing notion that an internal self-focus of attention always results in poorer motor performance was recently disputed by Carson and Collins (2015). They argued that a 'holistic' self-focus of attention aids the motor learning process as opposed to a partial self-focus on one of the movement components. In most studies investigating the effects of internal and external focus of attention on motor performance, including the present study, the partial form of self-focus was used for the internal focus of attention condition (Wulf, 2013). Therefore, in future studies a comparison of the effects of holistic and partial internal focus of attention instructions on gait performance could provide more insight the effects of focus of attention.

3.4.2 Conclusion

In the balance recovery response to a walking perturbation no significant difference was found between internal and external focus of attention conditions on walking stability parameters based on step length *CV*, step width *CV* and the COM velocity response. This might be caused by the absence of an external object to move or act upon. We therefore conclude that for elderly gait, external attention to the walking surface does not lead to improved balance recovery responses to gait perturbations.

Chapter 4

Influence of Attentional Focus, Reinvestment and Fall History on Elderly Gait Stability

Introduction: Falls represent a substantial risk in the elderly population. Previous studies have found that focussing attention on the outcome/effect of the movement (external focus of attention) leads to improved balance performance, whereas focussing on the movement execution itself (internal focus of attention) impairs balance performance in the elderly. A shift towards more conscious, explicit forms of motor control occurs when existing declarative knowledge is recruited in motor control, a phenomenon called reinvestment. We investigated the effects of attentional focus and reinvestment on gait stability in elderly fallers and non-fallers.

Method: Full body kinematics was collected of 28 healthy older adults walking on a treadmill while focus of attention was manipulated through instruction. Participants also filled out the Movement Specific Reinvestment Scale (MSRS) and the Falls Efficacy Scale International (FES-I), and provided details about their fall history. Coefficients of Variation (CV) of spatiotemporal gait parameters and Local Divergence Exponents (LDE) were calculated as measures of gait variability and gait stability, respectively.

Results: No significant effect of attentional focus was found for any of the gait parameters, and no significant relation between MSRS score (reinvestment) and fall history was found. Larger stance time CV and LDE (decreased gait stability) were found for fallers compared to non-fallers. Higher step width CV and FES-I scores for fallers than non-fallers were borderline significant.

Discussion: We conclude that external attention to the walking surface does not lead to improved gait stability in elderly. Potential benefits of an external focus of attention might not apply to gait, because walking movements are not geared towards achieving a distinct environmental effect.

Adapted from: de Melker Worms, J. L. A., Stins, J. F., van Wegen, E. E. H., Loram, I. D, Beek, P. J. (2016). Influence of Focus of Attention, Reinvestment and Fall History on Elderly Gait Stability. *In press.*

4.1 Introduction

In the elderly population, falls represent a substantial risk. Approximately two thirds of unintentional injury related deaths in older adults are caused by falls (Baker & Harvey, 1985). Falls represent the leading cause of bone fractures (Schwartz *et al.*, 2005) and one third of community-dwelling elderly over the age of 65 suffer at least one fall each year. Consequently, it also imposes a substantial global economic burden (Stevens *et al.*, 2006).

There is considerable interest in psychological / cognitive factors that determine gait performance, and hence fall risk. In pertinent literature it has been suggested that fall risk is larger for individuals with a higher level of conscious attention to their own movements than the general population (Wong *et al.*, 2008; Chiviacowsky *et al.*, 2010; Wulf, 2013; Young *et al.*, 2015). It has further been suggested that the fall risk of such individuals might be reduced if their movements would be more automated and require less attentional control (Chiviacowsky et al., 2010; Wulf, 2013; Young et al., 2015). Conversely, shifts towards more conscious, explicit forms of motor control occur when existing declarative knowledge is recruited in the planning and execution of movements. Masters (1992) dubbed this phenomenon reinvestment (i.e. of said knowledge structures). Reinvestment is thought to be manifested when an individual is highly motivated or under pressure, or has difficulty to move successfully (Wong *et al.*, 2008). Using the Movement Specific Reinvestment Scale (MSRS), Wong et al. found that elderly with a history of falling had a higher predisposition to reinvest compared to elderly non-fallers (Wong *et al.*, 2008).

Akin to the theory of reinvestment is the 'constrained action hypothesis' (Wulf & Prinz, 2001a), which emphasizes the crucial role of attentional processes in motor performance. By now, there is ample evidence that an attentional focus on the outcome/effect of the movement ('external focus of attention') leads to improved motor performance and learning, whereas a focus on the movement execution itself ('internal focus of attention') hampers motor performance and learning. These effects have been found for a wide range of sports and balancing tasks. Wulf (2013). Chiviacowsky *et al.* (2010) showed that this effect generalises to motor learning of balance control in the elderly population, using an unstable balance board to assess balance performance. In a recent study linking the concept of attentional focus to that of reinvestment, higher reinvestment was found to be suggestive of a preference for an internally directed attentional focus (Kal *et al.*, 2015). According to the constrained action hypothesis (Wulf & Prinz, 2001a), an internal focus of attention induces a conscious control of movement that impairs automaticity. Moreover, this theory states that an external focus of attention enhances automaticity and allows for more efficient, implicit control mechanisms to come into play. In subsequent papers this claim of enhanced automaticity has received empirical support in the form of reduced muscular activity (Zachry *et al.*, 2005; Lohse *et al.*, 2010a), and more fluent and more regular movement (Kal *et al.*, 2013).

In some tasks, the goal is not so much to achieve a particular environmental effect, as in goal-directed instrumental actions, but rather to control the movements of the body itself. In such instances, an external focus of attention might be induced by directing attention to physical surface(s) in the environment on which force is exerted through muscle activity, such as the ground one is standing on in a gymnastics floor routine (Lawrence *et al.*, 2011). Critical for the proper use of the term external focus of attention in such situations is not only that reference is made to physical properties of the environment, but also that this reference is relevant for the successful performance of the task (Lawrence *et al.*, 2011; An *et al.*, 2013).

Even though benefits of an external focus of attention have been found for postural balance control, such benefits have to date not been established for elderly balance in gait. In the present study we therefore investigated the effects of attentional focus (a state variable) and reinvestment (a trait variable) on gait stability and variability in elderly fallers and non-fallers.

Gait performance can be assessed by measurement of either steady state gait or perturbed walking. Investigation of perturbed walking involves analysis of the manner in which the actor attempts to regain stability after a perturbation (Bruijn *et al.*, 2010; Granacher *et al.*, 2010). In the present study we adopted a paradigm involving transient mechanical perturbations. The perturbations consisted of unilateral decelerations of a split-belt treadmill, which led to a forward slip of the foot, as when walking on a slippery surface. The perturbations in question were applied at unexpected moments in time and participants were motivated to preserve stable locomotion between perturbations. We here focus on steady gait performance in between the stabilising responses to the perturbations. The direct stabilising responses within the first 4 s after each perturbation is reported in Chapter 3 of this thesis, because each of the two modes of gait assessment brings along specific theoretical and methodological issues.

In order to examine how attentional focus and reinvestment scores affect gait stability, we collected full body kinematics and analysed participants' steady gait bouts between the balance recovery responses to the perturbations. The literature on the relation between elderly falls and gait performance shows that gait variability is increased in elderly fallers compared to non-fallers (Hausdorff *et al.*, 1997; Toebes *et al.*, 2012). Furthermore, prospective research showed an increased fall risk for elderly with increased stride-to-stride gait variability (Hausdorff *et al.*, 2001). A common measure to quantify this variability is coefficient of variation (CV) of spatiotemporal gait parameters (Hausdorff *et al.*, 2001). An alternative approach to assess gait performance is through gait stability, which has been quantified using Local Divergence Exponents (LDE) of kinematic data that approximated body COM movement. (Rosenstein *et al.*, 1993; Liu *et al.*, 2008; Lockhart & Liu, 2008; Toebes *et al.*, 2012). As all balancing movements are related to manipulation of body COM position, this is an important variable for assessment of gait stability. The gait of elderly fallers has been shown to be less stable than non-fallers in terms of such LDE values (Liu *et al.*, 2008; Lockhart & Liu, 2008; Toebes *et al.*, 2012).

The main aim of the present study was to examine whether an external focus of attention leads to a more stable walking pattern and reduced gait variability compared to an internal focus of attention. We further investigated how fall history, balance confidence and reinvestment interact with the gait stability parameters, and whether fall history affects balance confidence, reinvestment or gait stability. To this end, we calculated coefficients of variation (CVs) of step length, step width, stance time and swing time, as measures of gait variability. In addition, we calculated LDE values for the Centre of Mass (COM) velocity time series (Rosenstein *et al.*, 1993), as a measure of gait stability. We expect increased gait stability and reduced gait variability for the external focus condition compared to internal focus.

4.2 Method

4.2.1 Participants

Twenty-eight healthy older adults (8 males, 20 females, age: 65+ years) were recruited with an average participant age of 69.3 ± 3.7 years (mean ± standard deviation; range: 65-78). A Dutch version of the Mini-Mental State Examination (MMSE) was used to determine the cognitive status of participants and they had to be able to walk independently for 10 minutes without a walking aid. Participants with a MMSE score below 25/30, any history of rheumatoid arthritis in lower extremities, cerebral vascular disease, Parkinson's disease, peripheral neuropathy, cardiac arrest, bypass treatment or any other neurological or cardiovascular impairment were excluded from the study. The study received approval from the local ethical committee and participants gave written informed consent prior to their participation.

4.2.2 Material

Participants walked on a split-belt treadmill with a fixed speed of 1 m/s with a 180 degrees semi-circular screen in front of them. A realistic optical flow pattern, based on the treadmill velocity, was projected on the screen and showed a straight forest road with mountains (Figure 4.1). The participants' gait was occasionally perturbed through transient unilateral treadmill decelerations that were initiated right after toe off of the dominant leg. At the following heel strike the velocity of this half of the treadmill was reduced to 0 m/s, causing a gait perturbation. At the next heel strike of the dominant leg the treadmill belt had regained its original velocity of 1 m/s. The perturbations were experienced as a forward slip of the foot. The system was controlled using D-Flow software from Motekforce Link b.v., Amsterdam, The Netherlands. Full body kinematics was collected using 47 passive retroreflective markers (using the Human Body Model from Motekforce Link (van den Bogert *et al.*, 2013)) and 10 high-resolution infrared cameras (Vicon, Oxford, UK).

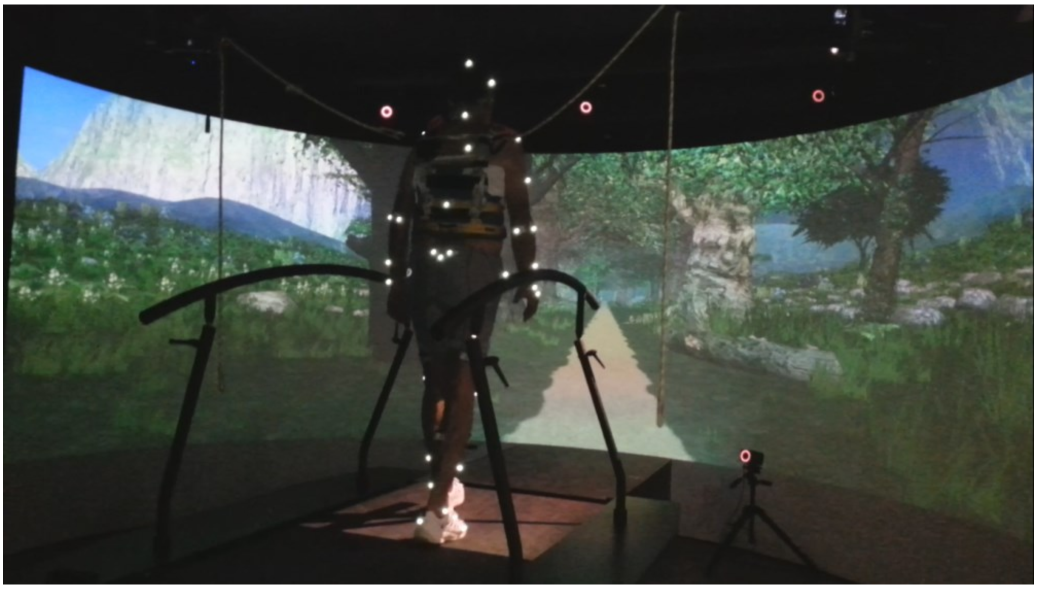

Figure 4.1: Virtual walking environment

4.2.3 Questionnaires

Before the experiment, reinvestment propensity was assessed with the Movement Specific Reinvestment Scale (MSRS) (Kleynen *et al.*, 2013), a Dutch version of the Falls Efficacy Scale International (FES-I) (Kempen *et al.*, 2007) was filled out and fall history details were collected. We defined a fall as follows: "An event in which a person unintentionally comes to rest on the ground or other lower levels" (Gibson *et al.*, 1987; de Zwart *et al.*, 2015). Falls that resulted from loss of consciousness or acute paralysis caused by stroke, epileptic attacks or violence were not included. When a fall had occurred within 12 months prior to the experiment, participants were labelled as fallers. The others were labelled as non-fallers.

The FES-I is a measure quantifying an individual's concern about falling, during various tasks (Morgan *et al.*, 2013; Visschedijk *et al.*, 2015), yielding a score between 16 (low concern about falling) and 64 (high concern about falling). The MSRS is a measure of an individual's propensity for reinvestment and consists of two subscales, pertaining to conscious motor processing (CMP) and movement self-consciousness (MSC), respectively. The first subscale is related to the amount of conscious monitoring of the own movement, whereas the latter is related to the amount of concern, as related to movement (Wong *et al.*, 2008).

4.2.4 Procedure

A fixed walking speed of 1 m/s was used throughout the experiment, gait perturbations excluded. Participants were first familiarised with 5 minutes of treadmill walking including gait perturbations. This was followed by two walking bouts of 5 minutes, one with an internal focus of intention instruction and one with an external focus of attention instruction, conducted in counter balanced order. In the internal focus of attention condition, participants were instructed to look ahead at the screen and concentrate on the movement of their legs. In the external focus of attention condition, they were instructed to look ahead at the screen and concentrate on the movement of the treadmill belt. The instructions were repeated through a speaker system every 30 seconds. For each condition 20 perturbations were given at heel strike, at random time intervals varying between 10 and 20 seconds. As this experiment was part of a protocol involving multiple experiments, participants had already walked on the treadmill for 20 minutes at the start of the experiment.

4.2.5 Spatiotemporal gait parameters

From the focus of attention trials the sections of unperturbed gait between 4 s after each perturbation up until the next perturbation were analysed. From these gait bouts (ranging from 6 to 16 s in duration) we calculated the participants' means and CV of the following spatiotemporal gait parameters for the dominant leg: step length, step width, stance time and swing time.

Step length was calculated as the distance in the anterior-posterior direction between the toe marker of the non-dominant leg and heel marker of the dominant leg, at each heel strike of the dominant leg. Step width was calculated as the distance between the toe marker of the dominant leg and the toe marker of the non-dominant leg in the mediolateral direction, at each heel strike of the dominant leg. Stance time was defined as the time interval between heel strike and toe off, while swing time was defined as the time interval between toe off and heel strike. Per participant the CV of these spatiotemporal gait parameters was calculated according to Equation 1.

$$CV(\%) = 100 \times \frac{\text{standard deviation}}{\text{mean}}, \quad (1)$$

Local divergence exponents (LDE)

Lower LDE values correspond with increased gait stability (Bruijn et al., 2012). LDE was calculated for the 3 dimensions of the COM velocity signals. A state space reconstruction in 9 dimensions was used, including two time delayed copies of the three COM velocity dimensions, one with 10 samples

(0.1 s) and one with 20 samples (0.2 s) time delay (van Schooten et al., 2013). Rosenstein's algorithm was employed to track the average logarithmic divergence between neighbouring trajectories in the reconstructed state space (Rosenstein et al., 1993). LDE was quantified as the slope of the first 60 samples (0.6 s) of the divergence curve, which roughly corresponded to one step, and was calculated over equal-length time series of 7 seconds. All calculations were implemented in Matlab (version R2014a, The MathWorks, Inc., Natick, MA, USA).

4.2.6 Statistical analysis

All dependent variables were tested for normality using the Shapiro-Wilk test. For the variables that did not pass the test we used non-parametric tests.

To cross-validate the questionnaire data with the occurrence of a fall in the past 12 months, FES-I, CMP and MSC scores of fallers were compared to non-fallers using Mann-Whitney U tests, effect size (r) and Bayes factors. Additionally, correlations between all gait parameters (mean and CV of spatiotemporal gait parameters and LDE) vs. the questionnaires (FES-I, MSRS CMP and MSRS MSC) were calculated using Spearman's Rho.

A 2×2 mixed ANOVA (within and between subjects) was used to test whether participant means of the normally distributed gait parameters (step length, step width, stance time, swing time and LDE) were significantly different between the focus of attention conditions, between fallers and non-fallers, and whether interaction effects were present between fall history and attention. The CVs of the spatiotemporal gait parameters did not pass the Shapiro-Wilk test for normality. Effects of fall history on these variables were calculated using Mann-Whitney U tests. Bonferroni corrections were used for subsequent Mann-Whitney U tests for fall history effects within attention conditions. Wilcoxon signed-rank tests were used to calculate effects of internal vs. external attention. Bonferroni corrections were used for subsequent Wilcoxon signed-rank tests for attention condition effects within fallers and non-fallers.

In addition to the above tests for significance, we calculated effect sizes and Bayes factor. The Bayes factor (BF_{10}) represents the likelihood of the alternative hypothesis vs. the null hypothesis. They can also be used to accept the null hypothesis, which is impossible on the basis of just *p*-values. It has been recommended to label BF_{10} values lower than 0.3 as moderate evidence in favour of the null hypothesis, and higher than 3 as moderate evidence in favour of the alternative hypothesis (Lee & Wagenmakers, 2014). All statistical analysis was calculated using IBM SPSS Statistics 20.0, except for the Bayes factors which were calculated with the BayesFactor v0.9.12-2 package for R (bayesfactorpcl.r-forge.r-project.org; R-project.org).

4.3 Results

4.3.1 Fall history, balance confidence and reinvestment

Nine out of twenty-eight participants had experienced a fall within the last 12 months and were labelled as fallers, while the remaining participants were labelled as non-fallers. The higher FES-I score for fallers than for non-fallers was borderline significant. The CMP and MSC scores on the MSRS were not significantly different between fallers and non-fallers (Table 4.1). Furthermore, no significant correlation was found between any of the gait parameters vs. any of the questionnaires (FES-I, MSRS CMP and MSRS MSC).

Table 4.1: Fallers and non-fallers are compared. Means (standard deviation), *p*-values, effect size and Bayes factors (BF_{10}) are shown for the tested gait parameters. Only for the CV, FES-I and MSRS variables medians (inter quartile range) are given.

	Fallers	**Non-fallers**	**p-value**	**Effect size**	**Bayes factor**
Mean step length (mm)	508 (70)	552 (50)	0.07	$\eta^2 = 0.12$	1.33
Mean step width (mm)	147 (35)	134 (29)	0.30	$\eta^2 = 0.04$	0.55
Mean stance time (s)	0.69 (0.09)	0.73 (0.06)	0.20	$\eta^2 = 0.06$	0.68
Mean swing time (s)	0.38 (0.03)	0.41 (0.03)	0.09	$\eta^2 = 0.11$	1.12
CV step length (%)	4.50 (1.21)	4.24 (1.44)	0.29	$r = 0.20$	0.40
CV step width (%)	15.61 (5.96)	18.59 (5.67)	0.07	$r = 0.34$	0.67
CV stance time (%)	3.50 (0.56)	3.01 (0.75)	**0.02***	$r = 0.46$	1.05
CV swing time (%)	4.94 (1.50)	4.41 (1.18)	0.32	$r = 0.22$	0.53
LDE	0.97 (0.12)	0.88 (0.08)	**0.03***	$\eta^2 = 0.16$	2.20
FES-I	20 (6)	17 (3)	0.06	$r = 0.37$	1.39
MSRS - CMP	8 (8)	12 (12.5)	0.64	$r = 0.09$	0.43
MSRS - MSC	5 (5)	6 (6)	0.47	$r = 0.14$	0.42

LDE = local divergence exponent (gait stability), *CV* = coefficient of variation

4.3.2 Gait parameters

No significant difference was found between the internal focus of attention condition and the external focus of attention condition for any of the gait parameters (Table 4.2). Furthermore, no significant interaction effects were found. For the non-fallers, CV of step width was only significantly larger for internal attention compared to external attention without Bonferroni corrections ($Z = -2.17$, $p = 0.03$, $r = 0.5$), see Figure 4.2. After exclusion of an outlier with the highest step width CV, the p-value for this effect without correction for multiple comparisons also increased above 0.05.

For fallers, the stance time CV and LDE were significantly larger than for non-fallers, however Bayes factor analysis did not provide evidence for this difference. The larger FES-I score and smaller step width CV for fallers compared to non-fallers was borderline significant (Figure 4.2 & Table 4.1).

Table 4.2. The internal and external attention conditions are compared. Means (standard deviation), *p*-values, effect size and Bayes factor are shown for the tested gait parameters. Only for the CV variables medians (inter quartile range) are given. The Bayes factor (BF_{10}) indicates the odds for the alternative hypothesis vs. the null hypothesis to be true. For the Bayes factors in bold these odds are less than 1/3.

	Internal focus	**External focus**	***p*-value**	**Effect size**	**Bayes Factor**
Mean step length (mm)	536 (58)	540 (62)	0.32	η^2 = 0.04	0.33
Mean step width (mm)	136 (32)	140 (31)	0.14	η^2 = 0.08	0.87
Mean stance time (s)	0.71 (0.07)	0.72 (0.07)	0.11	η^2 = 0.10	0.67
Mean swing time (s)	0.40 (0.03)	0.40 (0.03)	0.91	η^2 = 0.00	0.27
CV step length (%)	4.23 (1.18)	4.42 (1.62)	0.35	*r* = 0.18	0.39
CV step width (%)	18.51 (7.29)	16.75 (5.71)	0.09	*r* = 0.32	0.59
CV stance time (%)	3.17 (0.63)	3.16 (0.99)	0.84	*r* = 0.04	0.20
CV swing time (%)	4.57 (1.24)	4.60 (1.55)	0.91	*r* = 0.02	0.20
LDE	0.92 (0.12)	0.90 (0.09)	0.21	η^2 = 0.06	0.35

LDE = local divergence exponent (gait stability), *CV* = coefficient of variation

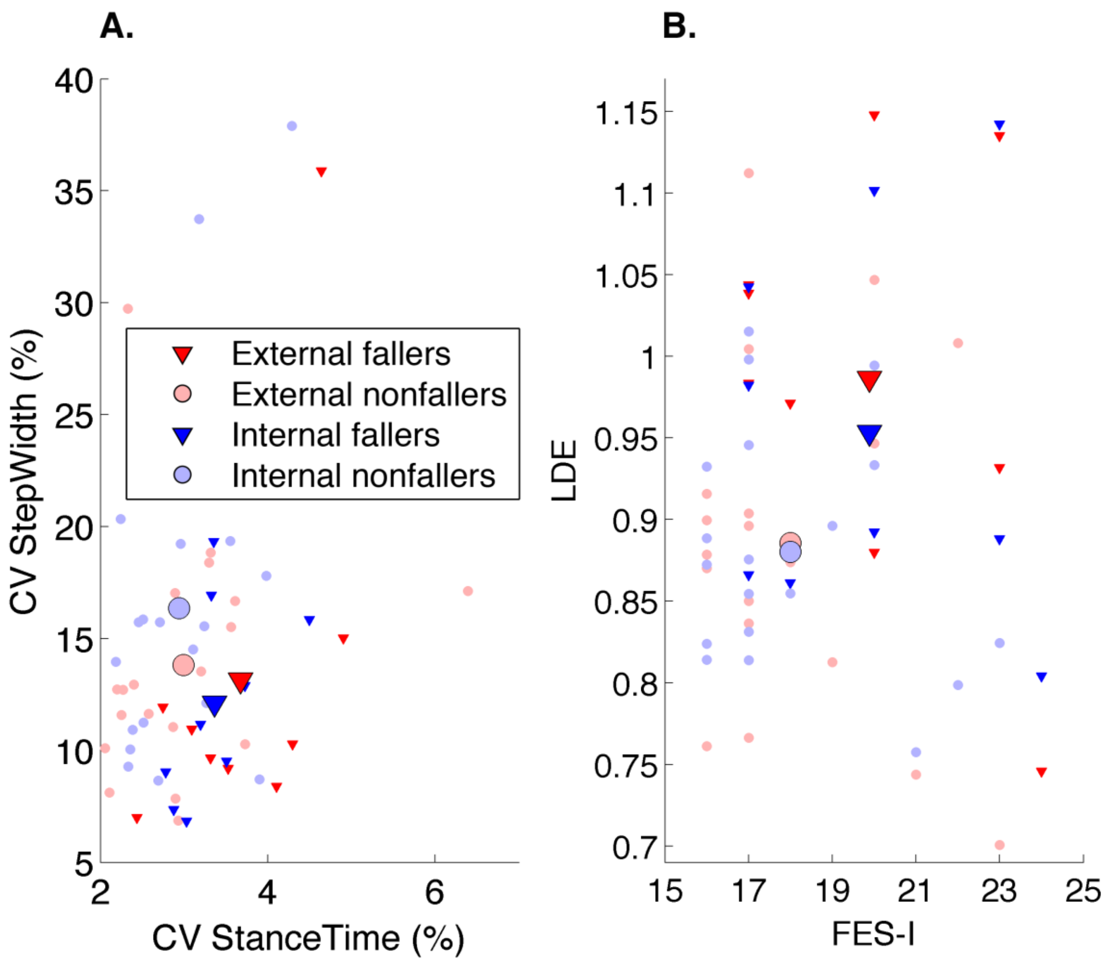

Figure 4.2: Step width vs. stance time and LDE vs. FES-I. (A.) Stance time and step width variability (CV) are shown for each participant in both attention conditions. Fallers had significantly higher stance time CV and the lower step width CV than non-fallers was borderline significant. No significant difference was found between internal or external attention for any of the gait parameters. (B.) Fallers had significantly higher LDE values (lower gait stability) than non-fallers. The higher FES-I score for fallers than non-fallers was borderline significant. Between internal and external attention no significant difference was found for FES-I or LDE.

4.4 Discussion

In the present study we investigated whether an external focus of attention temporarily increases gait stability and/or decreases gait variability compared to an internal focus of attention. No significant effect of attentional focus was found for any of the gait parameters. Furthermore, Bayes factor analysis provided moderate evidence for the null hypothesis that attentional focus does not affect gait variability, based on the CVs of stance time and swing time.

In addition, the effects of fall history, balance confidence and reinvestment on gait stability were examined. The higher LDE and stance time CV indicated significantly lower gait stability and increased gait variability for fallers compared to non-fallers. This supports previous suggestions that gait stability (Liu *et al.*, 2008; Lockhart & Liu, 2008; Toebes *et al.*, 2012) and gait variability (Hausdorff *et al.*, 1997; Toebes *et al.*, 2012) are associated with fall history. The results further suggested higher falls efficacy for fallers compared to non-fallers, as the effect of fall history on FES-I score was borderline significant. However, no significant effect of fall history on the MSRS reinvestment scores was found for either the CMP or MSC subscales. Thus, having experienced falls was not associated with increased reinvestment, which seems to be in contrast to findings from Wong et al. (Wong *et al.*, 2008). On the other hand, the Bayes factors also did not provide evidence to accept the null hypothesis that fall history does not affect reinvestment.

4.4.1 Evaluating effects of attentional focus and reinvestment on gait stability

In the literature on attentional focus, most studies involved a task in which actors were instructed to achieve a specific environmental effect. In that case, an external focus of attention could provide information that facilitates smooth planning and execution of the instrumental actions required to achieve that effect. However, the task considered in the present experiment was to control movement of the body itself (i.e., locomotion), in the absence of a distinct environmental goal. In other studies where the task was to control body movement without such a goal, results have been inconsistent. For the golf swing form, performance benefits were found for an external focus of attention (An *et al.*, 2013). However, these benefits were not found for gymnastics (Lawrence *et al.*, 2011). Surprisingly, an opposite effect was found for stroke patients, where beneficial effects of an internal focus of attention were found for movement performance of the paretic limb (Kal *et al.*, 2015). Furthermore the visual information of participants' surroundings in the present study could have provided more useful information about their body movements than the treadmill belt. In addition, the results might suggest that benefits of an external focus of attention are only present when the instructions imply a movement task originated by the performer, i.e. the direct effect of the movement.

According to the theory of reinvestment a reduced falls efficacy or increased fear of falling could lead to increased conscious attention to movement of the body. This could interfere with the automaticity of motor control and revert the actor back to an earlier declarative stage of learning. Analogous to the theory of reinvestment, an internal focus of attention might trigger the same adverse process. This might explain reduced performance with an internal focus of attention compared to an external focus of attention in ontogenic skills (learned in later life), e.g. with postural control on a stabilometer (Chiviacowsky *et al.*, 2010) and with various sports (Freudenheim *et al.*, 2010; Lohse *et al.*, 2010a; Wulf *et al.*, 2010).

However, because walking and normal postural control on solid ground are phylogenic skills (learned in early childhood, without declarative knowledge) it is unlikely that an internal focus of attention could lead to such a reversal (Young & Williams, 2015). This was supported by findings in postural balance control while standing on solid ground, where no benefits of an external focus of attention over an internal focus of attention were found (Wulf *et al.*, 2007).

In should be noted that two previous studies did find an effect of attentional focus on gait performance (Canning, 2005; Shafizadeh *et al.*, 2013). Canning (2005) found improved gait performance for an internal focus of attention instead of an external focus of attention in Parkinson's disease patients. Gait performance was assessed while participants carried a tray with glasses. Attention was either directed to walking (internal focus) or to balancing the tray with glasses (external focus). However, one could argue that in this experiment a focus on two different aspects of the task was compared, while performance of only one of those aspects was assessed (Wulf, 2013). Therefore the inferred benefit of an internal focus of attention might be challenged.

Shafizadeh *et al.* (2013) found an effect of improved gait performance for an external focus of attention in multiple sclerosis patients compared to an internal focus of attention. However, in their experimental conditions, different modes of gait performance feedback were used to focus attention. In the internal focus of attention condition, different information of gait parameters was presented on a screen than in the external focus of attention condition, where auditory feedback was added as well. Therefore, in this study, the observed effect on gait performance could be caused by the inequality of information that was given, as opposed to a cause of attentional focus.

The present study adds to the growing body of literature on the effects of reinvestment and attentional focus on gait stability in elderly and the interaction with fall history. We found that these psychological/cognitive factors had little effect on gait performance. A general limitation with studies manipulating attentional focus using verbal instruction is that it is not possible to independently assess whether participants complied with the instructions. We tried to remedy this by repeating the instructions every 30 seconds, but this yielded no guarantee that attentional focus was successfully manipulated.

In previous studies on balance control, the effects of attentional focus were only found when balance was challenged, e.g., when using an unstable standing surface, but not for normal standing. Perhaps the effects of attentional focus could also emerge for walking when the task to maintain a walking pattern would be more challenging, e.g., through continuous gait perturbations. In addition, it might be possible that there are motor learning effects of attentional focus on walking performance, but no acute effects. In that case, the addition of retention tests might also reveal a relation between gait performance and attentional focus. Further investigation of this topic could also clarify whether external attention instructions remain problematic in tasks where one does not move or manipulate an external object.

4.4.2 Conclusions

The results of this study provide further support for the interrelations between gait variability, gait stability and falls in the elderly, based on increased LDE and stance time CV in elderly fallers compared to non-fallers. No significant difference in MSRS scores was found between fallers and non-fallers, therefore the relationship between reinvestment and fall history wars not supported. Directing attention to the walking surface did not lead to improved gait stability in elderly, compared to internal attention on leg movement. Therefore the possible benefits of external attention for balancing tasks might not be present in elderly gait.

Chapter 5

Epilogue

5.1 Introduction

The general aim of the present project was to assess the effects of fear of falling and attention on human balance control. Knowledge of the neurophysiological and psychological mechanisms that have an adverse effect on balance may ultimately help to design interventions to counteract mobility loss in the elderly, anchored in scientific theory and based on empirical evidence. To achieve this goal two experiments were conducted, one at the Manchester Metropolitan University and one at the Vrije Universiteit Amsterdam. The first experiment investigated the effects of fear of falling on vestibular balance reflexes. Full body kinematics was collected from young healthy adults standing at ground level and at height to induce a fear of falling. Participants were stimulated with GVS to induce vestibular balance reflexes. In the second experiment the influence of focus of attention and fall history on gait performance was studied by applying random mechanical perturbations to gait, in a sample of elderly participants. In this Epilogue the main findings of these studies are summarised and discussed in light of the extant literature. Furthermore, the scientific implications of this work and recommendations for future research are discussed.

5.2 How fear affects balance control

As falls pose a significant threat to the elderly population, a large body of research is dedicated to identifying risk factors for falls, in particular factors that reduce balancing capabilities. In addition to physiological risk factors, psychological/cognitive constructs such as fear of falling and attentional focus have also been found to be important in relation to the occurrence of falls in the elderly.

5.2.1 Vestibular balance control

The literature shows that fear of falling can directly affect fall-risk through impairment of balance control. However, the mechanism behind this relation has not yet been clarified. For example, it is unknown whether fear of falling can influence balance at the level of fast vestibular reflexes. An often-used paradigm to elicit these balance reflexes is by applying GVS. With GVS the vestibular nerves are electrically stimulated, which induces a sensation of lateral rotation of the body. This illusory rotation elicits a whole body sway response towards the side of the anode electrode on the head. Therefore the sway response (i.e. triggered by the vestibular balance reflex) depends on the orientation of the head, see Figure 1.1. Osler *et al.* (2013) collected head and trunk kinematics of the GVS response of participants standing at height to induce fear of falling by means of a postural threat. Each participant was also tested while standing at ground level (no postural threat). They found that height-induced fear of falling did not affect the sway response. Thus, it was concluded that fear of falling does not affect the vestibular-evoked balance response.

Other studies have collected GRF (Mian & Day, 2009; Dakin *et al.*, 2010; Mian *et al.*, 2010; Horslen *et al.*, 2014; Mian & Day, 2014) and lower extremity EMG data (Britton et al., 1993; Fitzpatrick et al., 1994; Ali et al., 2003; Fitzpatrick & Day, 2004; Son et al., 2008; Mian et al., 2010; Muise et al., 2012) in order to characterize the same (GVS induced) vestibular balance reflex. They consistently found a bi-phasic response pattern consisting of a short- and a medium-latency response (see Figure 1.2). Importantly, GRF data from Horslen *et al.* (2014) showed that height-induced fear of falling increases the gain of this bi-phasic vestibular balance reflex, which seems to be in contrast to the kinematic data collected by Osler *et al.* (2013). As such, it was subsequently debated whether the fear-induced increase of the bi-phasic GRF response functionally contributes to balancing movements (Horslen *et al.*, 2015a, b; Reynolds *et al.*, 2015a, b).

To investigate whether fear of falling affects vestibular balance reflexes, we reasoned that a more detailed characterisation was needed of the kinematic pattern constituting this vestibular balance reflex. As opposed to head and trunk kinematics, full body kinematics of the GVS response could clarify how the balancing movements relate to the bi-phasic GVS response found in EMG and GRF data. Our findings are presented in the following paragraphs; we first discuss how the short- and medium-latency response are coupled (5.2.2) and next the effects of fear on this reflex pattern (5.2.3).

5.2.2 Short- and medium-latency response of vestibular balance reflexes

As described in Chapter 2, participants were stimulated with GVS to elicit vestibular balance reflexes. Full body kinematics was collected to characterise the balancing response. In the literature the GVS response has mainly been described with lower extremity EMG and shear GRF data; both types of data showed evidence of a short- and medium-latency response (Marsden *et al.*, 2005; Mian & Day, 2009; Day *et al.*, 2010; Mian *et al.*, 2010; Horslen *et al.*, 2014; Mian & Day, 2014). Interestingly, the short-latency response seemed to 'mirror' the medium-latency response.

More specifically, tibialis anterior, soleus and gastrocnemius muscles showed a pattern of short-latency activation that was followed by medium-latency inhibition (or vice versa, dependent on the anode/cathode configuration and head orientation).

With respect to GRF data, the literature revealed that short-latency cathode directed shear force was typically followed by medium-latency anode directed (i.e., opposite direction) shear force. Head and trunk kinematics data of this response showed a unilateral whole body sway response towards the anode side of the GVS electrodes that was consistent with the medium-latency EMG and GRF response data (Day *et al.*, 1997; Osler *et al.*, 2013). However, the contribution of the short-latency response to balance control was not yet clarified in relation to the kinematic data. Various hypotheses have been tested that might explain the origin of the short-latency response, but they all have been refuted (Britton *et al.*, 1993; Cathers *et al.*, 2005; Mian *et al.*, 2010).

From Newton’s second law of motion it follows that the GRF pattern is proportional to the body COM acceleration. Therefore one could expect to find similar short- and medium-latency responses in the acceleration pattern of the entire body. However this was not reflected in the limited kinematic data that have been presented in the literature (Day *et al.*, 1997; Day *et al.*, 2010; Osler *et al.*, 2013). Therefore in the study described in Chapter 2, we aimed to characterise the vestibular balancing response in more detail using full body kinematics.

In our study we did find the short- and medium-latency acceleration responses, which were directed towards the cathode and anode electrode, respectively. We found this bi-phasic response pattern only in body COM, pelvis and lower extremities acceleration, but not in the head and trunk acceleration. This finding could explain why short- and medium-latency responses were not found in the kinematic data obtained in previous studies, as these were collected from the trunk and head, but not the lower extremities. These findings update the traditional model (Figure 1.1) of the GVS induced sway response. See a link to a video of the GVS sway and acceleration response in the supplementary materials section.

In addition, we proposed a mechanism that includes a functional contribution of the short-latency response to balancing movements. To be specific, we proposed that both the short- and medium-latency reflexes are biomechanically coupled as one coordinated response to guarantee whole body postural stability. The medium-latency sway response could be facilitated by a short-latency response that moves the centre of pressure towards the cathode, whereas the COM does not move to the same extent. This would allow the pull of gravity to aid in swaying the body towards the anode electrode. This balancing mechanism could be compared to balancing an upright stick on the palm of your hand. To move the stick (COM) to the right, you move your hand (COP) to the left. Thus, the hand moves the base of support (short-latency response) in a lateral direction, which then changes the gravitational moment on the stick, facilitating the medium-latency response. The study was not designed to test this theory, as vestibular balance reflexes were tested in one postural configuration and the short latency sway responses in the lower extremities were very small. On the other hand, this theory is consistent with GRF data from other studies where vestibular balance reflexes were tested in multiple configurations (Mian *et al.*, 2010; Horslen *et al.*, 2014).

5.2.3 Effects of fear of falling on vestibular balance reflexes

In the literature, height-induced fear of falling was found to increase the gain of short- and medium-latency vestibular balance reflexes. However, no consensus has been reached whether these changes functionally contribute to balance control. Opposing publications (Osler *et al.*, 2013; Horslen *et al.*, 2014) on this topic were discussed in a recent cross-talk debate (Horslen *et al.*, 2015a, b; Reynolds *et al.*, 2015a, b). We investigated how vestibular balance reflexes are influenced by fear of falling. The GVS induced vestibular reflexes were studied for participants standing at ground level but also while standing on a 3.85 m high narrow walkway to induce a fear of falling. Participants' physiological arousal (skin conductance) and self-evaluated levels of fear of falling were increased while standing at height, indicating that we could successfully induce fear. More importantly, analysis of whole body kinematics showed that the lower extremity short- and medium-latency acceleration responses were altered at height. Our main finding was that the response amplitude was increased, while the time interval during which the responses were executed was decreased, indicating that fear of falling induced stronger and 'brisker' balancing reflexes. However, fear of falling had no effect on the early (0-400 ms) GVS induced torso and head acceleration.

Our findings are consistent with the findings from Horslen *et al.* (2014) who found an increased gain of GRF-SVS short- and medium-latency vestibular balance responses with a height-induced fear of falling. Based on our full body kinematic data we concluded that the gain of the appendicular short-and medium-latency vestibular balancing reflexes increases with fear of falling. However, the fast axial neck and thoracolumbar muscle responses are governed by different neuromuscular mechanisms that seem to be unaffected by fear of falling. Our findings are also consistent with the seemingly opposing findings from Osler *et al.* (2013). They found no effect of height-induced fear of falling on the vestibular balance reflex, as measured with kinematic recordings of the head and trunk only. The distinct dependencies of axial and appendicular vestibular reflexes may reflect different functional goals (head stabilisation vs. whole body balance) and differential innervation (medial vs. lateral vestibulospinal tracts) (Forbes *et al.*, 2015).

In conclusion, Chapter 2 showed that height-induced fear of falling increases the gain of vestibular balance reflexes. Full body kinematic data suggest that both the short- and medium-latency appendicular vestibular balance reflexes functionally contribute to whole body balance and are biomechanically coupled into one coordinated response. Furthermore, axial vestibular reflexes were found to be unaffected by fear of falling and the goal of these reflexes may be more closely related to stabilise the head in space than to whole body balance.

5.3 Attentional focus

A different psychological/cognitive factor that is related to fear of falling and falls in the elderly is focus of attention. Individuals who experience fear of falling or who have low balance confidence may choose to consciously monitor their body movements in an effort to improve motor control (Wong *et al.*, 2008). This change from an implicit, more automated form of motor control to an explicit, more conscious form of motor control has been termed reinvestment, and seems to constitute a cognitive (adaptive) mechanism. Furthermore, a relation was found between reinvestment scores and fall history in elderly (Wong *et al.*, 2008). A separate but related body of literature on attentional focus is based on the 'constrained action hypothesis' (Wulf *et al.*, 2001; Wulf, 2013). This hypothesis asserts that an attentional focus on the movement outcome in the environment ('external focus') results in improved motor performance and motor learning, whereas a focus on movement execution itself ('internal focus') hampers motor performance and motor learning. Beneficial effects of an external focus were found for various sports and balancing tasks, and were reviewed by Wulf (2013). In the following paragraphs the main findings of Chapter 3 (5.3.1) and Chapter 4 (5.3.2) are discussed, followed by a critical evaluation of the attentional focus paradigm (5.3.3). In 5.3.4 we discuss the relations between attentional focus, fear of falling and gait.

5.3.1 Attentional focus and perturbed gait responses

The potential benefits of an external attentional focus on motor performance has not been demonstrated for gait in healthy elderly. The literature suggests that when the task is relatively easy, an external attentional focus yields no additional motor performance benefits. For example, benefits with respect to balance control were only found in more challenging balancing tasks, e.g. standing on an unstable balancing surface, and not for standing on solid ground (Wulf *et al.*, 2007).

As such, steady gait might not be challenging enough for the effect of attentional focus to occur. To tackle this issue, we introduced mechanical gait perturbations to make the walking task more challenging. An experiment was conducted at the Vrije Universiteit Amsterdam and was covered in Chapters 3 and 4 of this thesis. The main aim of this study was to investigate whether an external focus of attention could temporarily enhance gait performance in elderly. If so, this could open up possibilities for cognitive intervention programmes in elderly with fear of falling. Elderly participants walked on a split belt treadmill that was used to apply mechanical gait perturbations at random time intervals to challenge gait stability. A virtual reality environment of a forest road with mountains was projected on a semi-circular screen in front of the treadmill to create a realistic optic flow while walking. Using full body kinematics the effects of internal vs. external attention instructions on the balancing responses to gait perturbations were tested, as described in Chapter 3. Gait performance is associated with gait variability. As a measure of gait variability, *CV* of step length and step width of the first step after gait perturbations was analysed. As such, we expected reduced variability (*CV* values) for external focus compared to internal focus. In addition, velocity of the body COM in three dimensions was used to calculate the orthogonal distance from unperturbed gait, based on the method from Bruijn *et al.* (2010). We used a novel technique (SPM) for statistical analysis of the resultant time series as a whole, in which the temporal dependency within the time series data were taken into account. The first four post-perturbation strides between internal and external attention were tested. Contrary to our expectations, no significant effect of focus of attention was found in any of these dependent variables. We therefore concluded that, relative to an internal focus, an external focus on the walking surface does not benefit balancing responses to gait perturbations.

5.3.2 Attentional focus and continuous gait

In Chapter 4 the effects of attentional focus on the gait bouts of continuous walking were described. By analysing the gait bouts between the perturbations, we measured the unperturbed gait pattern, which might be more sensitive to cognitive influence than abrupt reflexive responses following a perturbation.

Gait variability was assessed with *CV's* of step length, step width, stance time and swing time of the unperturbed gait bouts between the perturbations, while gait stability was calculated with LDE. For reasons outlined above, we expected to find reduced gait variability and increased stability for external attention compared to internal attention. However, also for these variables no effect of attentional focus was found. Hence, we concluded that external attention to the walking surface does not affect gait stability or variability in unperturbed elderly gait compared to internal attention.

In a review on the effects of internal and external focus of attention on motor performance (Wulf, 2013), several other studies were evaluated where null effects of attentional focus were found as well. For some of these studies participants were presented with information on a screen about their movements or the effects of their movements in the environment (De Bruin *et al.*, 2009; Shafizadeh *et al.*, 2013). For example, a moving dot representing the centre of gravity relative to a target (De Bruin *et al.*, 2009). Wulf (2013) argued that null effects in these studies were caused by powerful visual feedback, which presumably obfuscated attentional focus effects.

In the experiments described in Chapter 3 and 4, participants were presented with realistic and gait-specific optic flow. One might therefore also attribute our null-effect to the presence of powerful visual feedback: It might well be that the presented optic flow overruled the effects of the instructions to concentrate on the movements of the treadmill or legs.

However, there is reason to believe that the effects of attentional focus can still manifest themselves in the presence of powerful visual feedback. It is well established that visual information of the surroundings aids to determine one's location in space and bodily orientation. This visual feedback is powerful, e.g. as balancing on an unstable surface (e.g. stabilometer or balance disk) with the eyes closed is much more challenging than with eyes open. For the balancing experiments described earlier in this chapter, effects of attentional focus were found (Wulf *et al.*, 1998; Shea & Wulf, 1999; Wulf *et al.*, 2001; McNevin *et al.*, 2003; Wulf & McNevin, 2003; Wulf *et al.*, 2004; Wulf *et al.*, 2007; Chiviacowsky *et al.*, 2010). These attentional focus effects occurred while participants had their eyes open and were highly dependent on the visual information to regulate their balance. Therefore, the powerful visual feedback did not obfuscate attentional focus effects in these studies. As such, it also seems unlikely that the optic flow one perceives with gait obfuscates attentional focus effects on gait performance.

5.3.3 Limitations of the internal/external focus paradigm

The absence of attentional focus effect on walking performance in general might also be related to the nature of the walking task. During gait, the goal is to maintain an upright walking pattern and to walk in a particular direction. To achieve this goal one does not have to control or manipulate an external object. The aim is to control the movement of the body itself with respect to the environment. In other studies where the task was to control body movement without an external object to manipulate, effects of attentional focus have been inconsistent. E.g., improved swimming performance was found for an external compared to internal focus of attention (Freudenheim *et al.*, 2010; Stoate & Wulf, 2011). However, Lawrence *et al.* (2011) compared the effects of internal and external focus on motor learning for a gymnastics floor routine, and they found no effect of attentional focus on motor learning. Additionally, Kal *et al.* (2015) even suggested an opposite effect, whereby external focus in fact reduced performance of paretic leg movement of stroke patients.

As such, some authors argued that benefits of an external focus of attention do not apply to motor tasks where performance only depends on the movement form or movement pattern of the body itself, and where movement effects on the environment are not of main importance (Lawrence *et al.*, 2011; Peh *et al.*, 2011). Subsequently, Wulf (2013) criticised this view by arguing that the instructions adopted in their gymnastics study (Lawrence *et al.*, 2011) were not relevant for performance of the gymnastics task. Furthermore, multiple other studies did show improvements in movement form (kinematics) with an external focus of attention, e.g. for golf swing (An *et al.*, 2013), darts (Lohse *et al.*, 2010a), rowing (Parr & Button, 2009) and throwing (Southard, 2011). However, for all of these studies manipulation of an external object was involved and the effect of the movement in the environment was crucially important.

In addition, the constrained action hypothesis as a whole has been criticised as well. Previous studies have found performance benefits for an internal focus in long jumping (Mullen & Hardy, 2010), a weight lifting case study (Carson *et al.*, 2014) and a small sample javelin throwing study (MacPherson *et al.*, 2008). Carson and Collins (2015) proposed that the reason for the adverse effects of internal focus on motor performance and motor learning in other studies is due to the partial self-focus of attention. They argued that a more holistic form of internal focus also yields performance benefits. Most internal focus instructions have only referred to movement of a specific part of the body. However, in nearly every movement task the whole body needs to be coordinated. Especially when a new movement pattern needs to be learned, internal focus is often inevitable when one cannot refer to (the effect of) a previously learned movement pattern.

Furthermore, a possible limitation is the relatively low sample size of participants that experienced a fall (nine) compared to the number of non-fallers (seventeen).

5.3.4 Relations between attentional focus, fall history and gait

In Chapter 4 the effect of fall history on gait stability and gait variability of unperturbed gait was studied as well. One of our findings was that participants who had experienced a fall in the 12 months preceding the experiment had significantly higher stance time *CV* and higher LDE (reduced gait stability). This supports the established findings that elderly fallers have reduced gait stability (Liu *et al.*, 2008; Lockhart & Liu, 2008; Toebes *et al.*, 2012) and increased gait variability (Hausdorff *et al.*, 1997; Toebes *et al.*, 2012). However, no effect of fall history was found on the balance recovery response to gait perturbations, based on the variability of spatiotemporal gait parameters and COM velocity data. This shows that fallers and non-fallers had a similar movement pattern of the balancing responses to the perturbations, to recover to a steady gait pattern. However, as the sample size for this between-subjects comparison (8 fallers vs. 17 non-fallers) was relatively small, a larger sample size might be needed to find an effect of fall history for these gait variables.

Furthermore, no significant interaction effect between attentional focus and fall history was found for any of the gait variables. Additionally our data did not support the relation between reinvestment and fall history as previously found by Wong *et al.* (2008), as no significant differences were found between fallers and non-fallers. The process of reinvestment entails a more conscious monitoring of the movement, where one switches back to an earlier and more explicit stage of learning that involves less automated motor control. It might be possible that reinvestment does not occur in phylogenic (learned in early life without declarative knowledge) motor skills as normal postural control and steady gait. For these skills, earlier stages of learning involved implicit learning and probably did not involve more conscious explicit learning (Young & Mark Williams, 2015).

5.4 Implications

This PhD project was part of the Move-Age joint doctorate programme that aims to improve mobility in the elderly population. Falls and mobility problems in elderly are critical issues worldwide. The literature on the factors that might contribute to fall risk shows that fear of falling and attention are important psychological factors. However the mechanisms by which these factors could affect fall risk are unclear. Investigation of the interaction between fear of falling, attention and balance in postural control and gait is needed to gain more insight into fall prevention.

The findings of Chapter 2 provide evidence that fear of falling increases the gain of vestibular balance reflexes. This supports an emergent theme that fear of falling increases sensitisation to balance relevant information (Balaban & Thayer, 2001; Horslen *et al.*, 2014). However, head-in-space stabilization reflexes were unaffected by fear of falling and seemed to be governed by different mechanisms. A direct relation between vestibular balance reflexes and fall risk in the general population or the elderly has not yet been determined. However, increased gain and faster execution may negatively affect the speed accuracy trade-off involved in the required balancing responses. Furthermore, ageing involved with deterioration of sensory and vestibular function could be vulnerable to added effects of fear on balance reflexes (Horak *et al.*, 1989; Baloh *et al.*, 1993; Kristinsdottir *et al.*, 2000).

In addition, our more detailed characterisation of the GVS induced vestibular balancing movements expands our understanding of the manner in which humans regulate their balance. Future studies using full body kinematic measurement of vestibular balance reflexes while standing with different head orientations could provide evidence for our suggested coupling between short- and medium-latency responses. Various authors have used Stochastic Vestibular Stimulation (SVS) instead of GVS. It has been shown that SVS also elicits short- and medium-latency vestibular balance reflexes that can be measured with EMG and GRF (Dakin *et al.*, 2007). Full body kinematic measurement of SVS responses could confirm whether these responses induce the same short- and medium-latency acceleration pattern throughout the body. Additionally, with the SVS method the vestibular stimulation durations needed are much shorter than with conventional GVS, and therefore more experimental conditions could be tested for each participant. As such, in future studies the effect of fear of falling on full body kinematic data of the vestibular balance reflex could be compared between elderly fallers and non-fallers. This might improve the tools we have for fall risk assessment.

The assumed benefits of an external focus of attention to the walking surface do not seem to apply to gait, as the effect of the movement on the environment is less relevant for this task. Continued investigation into attentional focus effects and fear of falling on gait including holistic and partial internal focus might further clarify the relations between fear of falling and attentional focus and how they could affect fall risk.

Impairment of motor control automaticity is a central theme for both the constrained action hypothesis and the reinvestment theory. For fall prevention in elderly it has been recommended to include gait automaticity training in conjunction with dual-tasks (Gschwind *et al.*, 2010). Movement regularity and movement fluency have been used as measures of automaticity and have shown to be affected by attentional focus (Kal *et al.*, 2013). As such, further studies could investigate whether reinvestment could affect gait automaticity in dual-task settings, and how this relates to falls in elderly. More specifically, falls and the degree of reinvestment in elderly might reveal differences in the trade-off between cognitive performance and gait automaticity in dual-tasks. A prediction would be that individuals with high reinvestment scores have greater difficulty in coordinating gait performance and cognitive (secondary) task performance.

Furthermore, brain imaging techniques could provide insight into the neurophysiological basis of how attentional focus and fear of falling might affect motor performance. Research in this area is scarce, although a functional magnetic resonance imaging (fMRI) study from Zentgraf *et al.* (2009) did find increased activity of the primary somatosensory and motor cortex for external focus compared to internal focus using finger movements. In addition, electroencephalography (EEG) studies measured the level of coherence between right hemispheric motor planning regions and left hemispheric verbal-analytical brain areas (Zhu *et al.*, 2011a; Zhu *et al.*, 2011b). These authors found increased coherence between these brain regions with more explicit conscious control of movement compared to more automated and implicit motor control. This was determined using the reinvestment scale (MSRS) and implicit vs. explicit motor learning paradigms. For future research it would be interesting to investigate how these attentional effects relate to internal and external focus conditions. This might reveal whether the same neural substrates and neural pathways are involved with the concepts of the constrained action hypothesis, reinvestment and the effects of implicit and explicit motor learning.

In addition, it has been proposed that fear of falling (Wong *et al.*, 2008) or 'choking' (Wulf, 2013) could instigate reinvestment and an internal focus of attention. Our results have confirmed that height-induced fear of falling affects the vestibular balance reflex. Possible neural targets for modulation of fear include the vestibular cortex, lateral vestibular nuclei, vestibulospinal tracts and subsequent spinal processing (Fitzpatrick & Day, 2004; Forbes *et al.*, 2015). In addition, excitation of the amygdala is associated with fear inducing stimuli. Two pathways connecting the amygdala and vestibular nuclei could be involved with this process, one via the parabrachial nucleus and one via the vestibular cortex (Lang *et al.*, 2000; Balaban & Thayer, 2001; Balaban, 2002; Staab *et al.*, 2013). It would be interesting to test how fall history and fear of falling affects brain activity in these regions. In addition, clinical studies involving patients with brain damage in these areas could provide more insight into the relation between fear and vestibular motor control. For example, it has been shown that amygdala deterioration prevents fear conditioning (Maren & Fanselow, 1996). It might therefore be interesting to assess whether the absence of a fear response in these patients is also reflected in balancing reflexes.

Our results corroborate converging evidence in the motor control literature that fear of falling increases sensitivity to self-motion. Future research on the effect of fear of falling and attentional focus on gait perturbation responses might provide more insight into fall prevention. There are many mechanisms from sensory integration to balancing motor execution to feedback of execution that could be impaired through ageing. Follow-up studies with clinical subgroups could further clarify the relation between fear of falling, attention and balance performance.

5.5 Main conclusions

- Fear of falling increases the gain of vestibular balance reflexes.
- Full body kinematic data suggest that both the short- and medium-latency reflexes functionally contribute to whole body balance and are biomechanically coupled into one coordinated response.
- Head-in-space stabilization reflexes is unaffected by fear of falling and seems to be governed by different mechanisms.
- External focus to a walking surface does not provide benefits for balancing responses to mechanical perturbations in gait of healthy elderly compared to internal focus.
- External focus to a walking surface does not reduce gait variability or increase gait stability in elderly compared to internal focus.
- Elderly fallers have increased gait variability and decreased gait stability compared to elderly non-fallers.

Supplementary materials

A video of the averaged GVS response in the ground and height conditions described in Chapter 2 can be found by scanning the QR-code below or with the following link: *https://mmutube.mmu.ac.uk/media/t/1_ck75hmkd*

As in Figure 2.7, the mediolateral movement of the body nodes is shown. Dots and stick figures show mediolateral displacement of the head, trunk and lower extremity body nodes with respect to the position at GVS onset. The left side represents the cathode side and the right side represents the anode side. Arrows represent mediolateral acceleration. Inter-node distances are not scaled.

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