AGE RELATED CHANGES IN THE MECHANISMS CONTRIBUTING TO HEAD STABILISATION, AND WHOLE BODY STABILITY DURING STEADY STATE GAIT AND GAIT INITIATION

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ABSTRACT

Head stabilisation during gait related tasks is thought to be fundamental to whole body stability, but this has received little attention in the older population. There is a need to examine any age related changes in neuromechanical mechanisms underpinning head stabilisation that may challenge the control of head stability, and consequently whole body stability. The present Thesis examined the mechanisms contributing to head stabilisation, and whole body stability during two gait tasks, steady state gait and gait initiation in young and older females, with the overall aim of contributing to negating fall risk. Four studies were designed to examine a) head position and walking speed on gait stability during steady state gait; b) neuromechanical mechanisms underpinning head stabilisation during gait initiation; c) head position on whole body stability during gait initiation; and d) head stabilisation during gait initiation at different speeds. Results showed that a) gait stability, was unaffected by head position and different walking speeds during steady state gait, b) decreased head stability in older individuals during gait initiation can be attributed to a deterioration of the neuromechanical mechanisms relating to head stability, c) free head movement during gait initiation does not affect head stabilisation or whole body stability but it does affect gait parameters, while d) initiating gait at faster than comfortable speeds compromises head stabilisation and reduces whole body stability in older individuals. Collectively, these results demonstrate that older individuals adopt an increased head flexion position when walking, while impaired head stability can be attributed to deterioration of the function of their neuromechanical mechanisms compared to their younger counterparts during gait tasks at comfortable speeds. These findings provide an understanding of the effect head stabilisation can have on older adults' gait and on their fall risk during gait and gait initiation.

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ABBREVIATION LIST

ANOVA	Analysis of Variance
AP	Antereoposterior
APA	Anticipatory Postural Adjustment
AvgSd	Average Standard Deviation
BOS	Base of Support
COM	Centre of Mass
COP	Centre of Pressure
EMG	Electromyography
ES	Erector Spinae
GSR	Gait Stability Ratio
ML	Margin of Stability
MOS	Mediolateral
NE	Neck Extensors
NVT	No Visual Target
PCA	Principal Component Analysis
RMS	Root Mean Sqaure
SCM	Sternocleidomastoid
VOR	Vestibular Occular Reflex
VT	Visual Target

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CHAPTER 1

Introduction

With age, there is a decline in the functional capacity of the neuromuscular system (Prince, Corriveau, Hébert, & Winter, 1997), which has serious implications for the performance of simple tasks, especially the control of posture and balance during gait and gait related tasks. Gait is a habitual activity, requiring transition from a stable to an unstable position, i.e. from double to single leg support. Such movement results in a continuous perturbation in the balance equilibrium, as the centre of mass (COM) alters in relation to the concurrently changing base of support (BOS) (Woollacott & Tang, 1997). This can prove challenging for older individuals (Ihlen et al., 2012; Prince et al., 1997), reflected by the fact that a substantial number of falls occur during walking in older individuals (Rubenstein, 2006).

Along with the challenges of steady state gait, gait initiation can also threaten the control of balance of older individuals. Transitioning from a stationary to a dynamic situation involves interactions between neural and biomechanical factors that challenge the balance system and ultimately may increase fall risk in older adults (Winter, 1995; Polcyn et al., 1998; Crenna, Cuong & Brénière, 2001). During gait initiation, anticipatory adjustments are required to uncouple the centre of pressure (COP) and COM, shifting the COP backwards towards the swing limb in order to move the COM forward towards the stance limb over a changing base of support (Jian, Winter, Ishac, & Gilchirst, 1993). However, presently, there is limited understanding of how the ageing process affects anticipatory control of upper body mechanics and the effect on whole body stability during gait and gait related tasks.

Early gait studies consider the upper body as one rigid segment (Winter, MacKinnon, Ruder, & Wieman, 1993), however as the upper body has many degrees of freedom, the upper body should be considered as separate components. The goal of the human postural system is to control the motion of the upper body in order to stabilise the head during gait by attenuating gait related oscillations of the lower body. Analysis of the head and trunk interaction have identified controlled movement strategies that enhance head stability during walking in young adults, (Cromwell et al., 2004; Cromwell & Wellmon, 2001) therefore it is important to understand if this is reflected in older adults to maintain head stability.

It has been reported that older individuals typically implement head flexion as a behavioural adaptation to help to identify lower limb trajectory to enable footfall vision during gait tasks (Marigold & Patla, 2008). However, it has also been suggested that head flexion may not be a behavioural adaptation, but a physiological adaptation, associated with the age related loss of the number of motor units per muscle (Enoka et al., 2003) and weakness of the neck flexors and neck extensors (Griegel-Morris, Larson, Mueller-Klaus, & Oatis, 1992). The understanding of whether head flexion is a safe behavioural adaptation implemented by older adults or whether it is a physiological adaptation attributed with the ageing process, which could be undermining head stabilisation and subsequently, whole body stability is warranted.

Head stabilisation has been reported to be critical to the control of whole body stability, (Kavanagh et al., 2005; Pozzo et al., 1990). Maintaining head stabilisation is the dynamic process whereby stable equilibrium of the head in space is maintained, which requires movement of the head on trunk to compensate for external perturbations or from the trunk (Cromwell, Newton, & Carlton, 2001). Perhaps surprisingly, only one study has examined the control of head stability during gait initiation which reported older females to have increased head movement variability with a more anterior angular displacement and therefore decreased head stability, compared to young (Laudani, Casabona, Perciavalle, & Macaluso, 2006). However, no studies have investigated the mechanisms underpinning such decreased head stability and the effect on whole body stability.

In addition to the lack of research on the mechanisms underpinning head stability during gait related tasks in older individuals, there are some methodological issues that need to be examined, to ensure correct interpretation of results and advice provided. Firstly, to account for head movement, studies investigating gait typically use a fixed visual target at eye level e.g. (Caderby, Yiou, Peyrot, Begon, & Dalleau, 2014; Hirasaki, Moore, Raphan, & Cohen, 1999; Laudani et al., 2006). If indeed older individuals do adopt an increased flexed head position, then walking whilst focussing on a fixed visual target at eye level does not illustrate a context-specific activity, as experimental controls restrict natural head movement (Hirasaki, Kubo, Nozawa, Matano, & Matsunaga, 1993). To date, no studies have challenged the use of fixed visual implementation during gait tasks. The evaluation of this will help to provide more realistic and ecologically valid results to provide useful information for clinical and research assessments.

Secondly, in addition to using a visual target, gait studies in the older population commonly use one self-selected gait speed - comfortable. Research considers gait speed as an outcome measure of gait ability (Montero-Odasso et al., 2004) but it is rarely used as the subject of investigation, particularly within the older population. However, regardless of age, environmental demands sometimes require different walking speeds to an individuals perceived ‘comfortable’, for example, walking fast when wanting to catch a bus. Given that the majority of falls occur during walking (Winter, 1995), a better understanding of different walking speeds and the effect on head stability and subsequently whole body stability is warranted. One study has examined head stability during different steady state walking speeds and found that young were able to maintain control of head stabilisation regardless of walking speeds, but this was compromised in older females at fast walking (Mazzà, Iosa, Pecoraro, & Cappozzo, 2008). However there was no measure of whole body stability. Further examination of the effect of different speeds on head stabilisation and the effect it has on whole body stability in older adults is needed.

The aim of this research project was to examine age related changes in head stabilisation and the mechanisms underpinning this, and whole body stability during two locomotive tasks, steady gait and gait initiation with the overall aim of contributing to negating fall risk in older females.

CHAPTER 2

Literature Review

This literature review will draw on the key studies involving gait and gait initiation in older adults. The section will commence with the discussion of gait and gait initiation, before going into more detail on anticipatory postural adjustments (APA's) and their contribution to overall balance. A discussion on head position and head stability along with the mechanisms underpinning head stabilisation will be presented and methodological considerations for gait assessment will be highlighted.

Gait

The ageing process can make walking a challenging process, due to reduced neuromuscular performance in several aspects e.g. strength, balance, functional ability (Boyer, Johnson, Banks, Jewell, & Hafer, 2017). Indeed, gait and gait disorders are an important contributor to falls in older adults (Rubenstein, 2006). As a result, several studies have investigated gait mechanics in older adults and the changes that occur due to ageing (Aboutorabi, Arazpour, Bahramizadeh, Hutchins, & Fadayevatan, 2016; Boyer, Johnson, Banks, Jewell, & Hafer, 2017; Prince et al., 1997). Gait patterns of young adults are characterised by phases of instability that allow for efficient forward progression and lateral shifting of the body's COM with each step. With ageing, adaptations in older adults' gait pattern aim to increase stability, however this can decrease the capacity for moving the body forward (Cromwell, Newton, & Forrest, 2002). Older adults tend to adopt a cautious gait strategy such as, shorter step lengths, increased step width, reduced gait speed and increasing double support time (Prince, et al., 1997), that prioritises the maintenance of whole body stability. Despite the adaptation in gait parameters, older adults' cadence is reported to remain similar to that of young adults (Rogers, Cromwell, & Grady, 2008). This means that as older adults shorten their step length, they cover less distance with the same number of steps. By adapting gait in this manner, older adults spend more time in the double limb support phase and less in the single limb support phase, thus creating a more stable walking pattern (Cromwell et al., 2001). This increased stability however, proves less effective for moving the body forward.

To account for the change in velocity and step length altering interdependently, the Gait Stability Ratio (GRS) was developed as a measure that reflects changes in both of those parameters (Cromwell & Newton, 2004). Through 2D analysis, GSR i.e. the ratio of steps to velocity is calculated to provide an indication of walking stability. An increase in GSR indicates that more steps per unit of distance and therefore a greater portion of the gait cycle is spent in the double support phase.

Gait initiation

Gait initiation must precede walking and as a result, it is a common task performed in daily life that requires a complex muscular synergy to generate the decoupling of the COM and COP during standing, which causes the COM to move forward (Jian, Winter, Ishac, & Gilchirst, 1993). The transition phase from a static to a dynamic condition involves two postural stabilisation functions: to prepare for the action and to recover from the perturbation of overcoming inertia, which can be challenging for older adults. During this phase, the motor system is in a state of transient disequilibrium, which means that equilibrium will be restored in the next support phase of periodic movements like steady state walking (Winter, 1995).

Dividing gait initiation into separate phases provides more information related to specific mechanisms present rather than analysing gait initiation as an overall movement (Jian et al., 1993). Consequently, to better understand gait initiation as a whole task, gait initiation is typically divided into two phases: anticipatory and execution. The present Thesis will focus primarily on the anticipatory phase of gait initiation. Figure 1 demonstrates the movement of the COP across the anticipatory phase needed to initiate gait. Initially, COP moves to the swing foot before any detectable foot movement, this phase is called the anticipatory phase where APAs are needed to initiate gait.

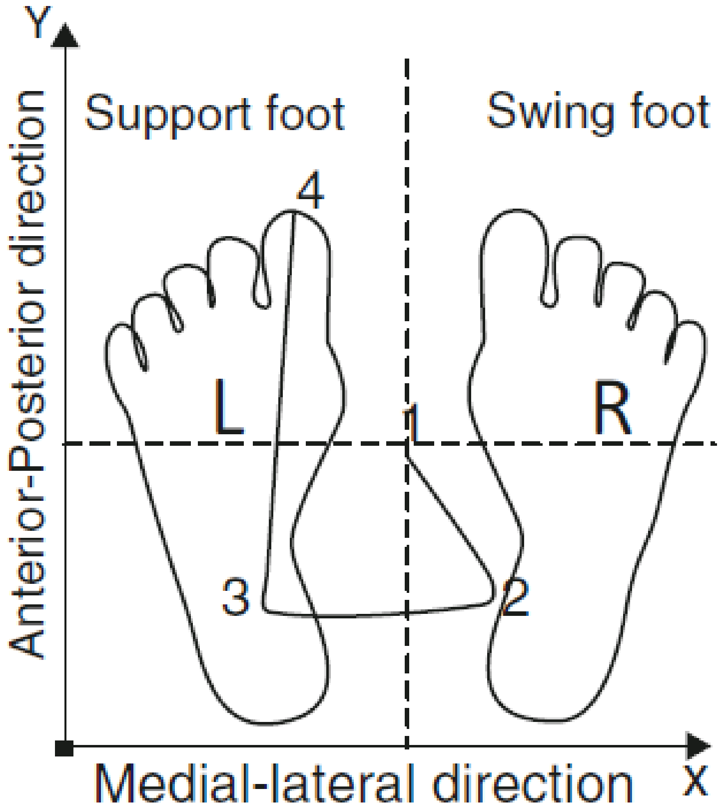

Figure 1. A schematic diagram of a typical COP path during gait initiation. 1–2: phase one -APA phase; 2–3: phase two-Swing foot unloading phase; 3–4: phase three-Support foot unloading phase.

Anticipatory control

APA's are considered to be under prospective control (von Hofsten, 2004) and, as such, they are used to assess the ability to feedforward control motor actions, in other words preparing for a movement. It is likely that dynamic instabilities during step transitions may be more critical factors to reach a steady state walking speed than standing postural control. The transition from standing to steady state gait, therefore, is an ideal paradigm to examine the impact of advancing age on balance (Muir, Rietdyk, & Haddad, 2014) since those changes may not be apparent during quiet support tests. It has been shown that older adults demonstrate a reduced ability to generate APAs in gait initiation compared to their younger counterparts, through a delayed muscle onset time of the leading limb as well as a slower velocity and lesser backward displacement of COP, which consequently negatively affects the performance of the task (Khanmohammadi et al., 2015). APAs of the lower limbs have been extensively researched during gait initiation, (Hyodo et al., 2012; Kubicki, Mourey, & Bonnetblanc, 2015; O'Kane, McGibbon, & Krebs, 2003; Woollacott & Manchester, 1993) however less is known about APAs of the upper body during gait initiation in older adults. There is a need to examine this phenomenon at head level to gain an understanding if any anticipatory adjustments of the head are either beneficial or detrimental to overall balance in older during gait initiation.

A major feedforward mechanism of postural stabilisation is the APAs that represent changes in the activity of postural muscles prior to the initiation of voluntary motor actions (Cordo & Nashner, 1982). The postural control system involves different control mechanisms and behavioural strategies, where compensatory (feedback) and anticipatory (feedforward) postural adjustments are implemented. Compensatory postural adjustments use afferent feedback to control the position of the body when the initial setting is disturbed (Woollacott & Manchester, 1993). In contrast, anticipatory postural control is based on predictive control to prepare the body for movements as disturbances are anticipated using higher-order processing rather than feedback (Malouin & Richards, 2000). Thus, this mechanism involves muscle activation prior to the disturbance. APA's operate when individuals deal with disturbances generated by their own movements, such as the transition of standing to walking during gait initiation (Prince et al., 1994).

In older adults, APA's have been seen not to be used effectively due to the inability of their to use feedforward control, possibly relating to the deleterious motor and sensory modifications associated with ageing (Ene et al., 2003). A reduced ability to produce APAs relates to an increased likelihood of falls in older populations, whereas older adults demonstrating APAs show no difference in stability as compared to young adults during stepping with lateral perturbations (Hyodo et al., 2012). If these anticipatory adjustments are not effective, the disturbance will be greater, and compensatory postural adjustments may not be enough to recover balance during gait initiation.

Head position

It is thought that older individuals typically implement head flexion as a cautious strategy to identify lower limb trajectory to enable footfall vision (Marigold & Patla, 2008) or to identify hazards on the ground (Menant, George, Fitzpatrick, & Lord, 2010). Alternatively, it has been suggested that head flexion may not be a behavioural adaptation, but a physiological adaptation in the older population associated with weakness of the neck flexors and shortening of the neck extensors (Griegel-Morris et al., 1992). Lee, Han, Cheon, Park, & Yong, (2015) found that head flexion reduces the muscle activity of the neck flexors which is problematic, as this may compromise overall balance in older adults. It is important to understand whether head flexion is a safe behavioural adaptation implemented by older adults or whether it is a physiological adaptation attributed with the ageing process which could be undermining head stabilisation and therefore overall balance. Activation of the neck flexors has been reported to be critical in the role of maintaining head stability (Dos Santos, Degani, & Latash, 2007) and therefore overall balance.

Whichever the reason for head flexion in the older, it remains that a considerable segment ~7% of overall body mass (de Leva, 1996) is moved forwards, and is, thus, likely to affect the COM. For example, during quiet standing in older individuals, Buckley, Anand, Scally, & Elliott (2005) reported a destabilising effect of head flexion with an increased AP sway as a result of an increased shift of the AP COM. If stability is reduced in static conditions with head flexion, it can be reasonably hypothesised that a flexed head can exacerbate the already forward shifting of the centre of mass during gait or during the fall forward when initiating gait, which will threaten balance further.

Head stabilisation

The movement of the large mass of the head, (de Leva, 1996), can lead to a larger tendency to head instability and can therefore negatively influence whole body stability. The importance of head stability for dynamic tasks has been studied extensively by Pozzo et al., (1990) in healthy young adults. The head contains two crucial sensory systems: the eyes and the vestibular organs. If the head is moved excessively, these two systems can be disturbed and therefore become less useful for overall balance. Head movement alters visual input and the vestibular by the endolymph fluid in the semicircular canals exerting pressure against the canal's sensory receptor. This can result in a perturbation to the postural system, which may be avoided through stabilisation of the head in space. The Vestibular Occular Reflex (VOR) is important for stabilising gaze on regions of interest during head movement, and is critical for gaining estimates of self-motion based on visual information. Studies have reported age related declines in VOR which has been correlated to fall risk (Baloh et al., 2003). It is possible that since the VOR promotes stabilisation of gaze relative to movements of the head, head flexion, along with the age related deterioration of the VOR, may disrupt this function and may affect components of balance, however, the consequences of this are yet to be discerned.

Given that head stabilisation presumably relies heavily on visual information, the interaction between head stability and postural control on the other hand is an interesting area of research. Although this area has been widely explored in young adults, it is not well researched in older adults. During dynamic situations head stabilisation is said to be achieved through coordinated movements of the head on trunk, to compensate for oscillations from the lower parts of the body (Cromwell et al., 2001). To date, only one study to date has investigated control of head stability in young and older individuals during gait initiation. Laudani et al., (2006) found decreased head stability of older compared to young, however the mechanisms causing such instability is not well examined.

Muscle Activation

There is an age related loss of the number of motor units per muscle, and an increase in the variability of motor unit firing that together reduce the precision with which older adults can generate muscle force (Enoka et al., 2003). From the literature available concerning upper body muscle activation during locomotion, studies have highlighted the importance of the different levels of the erector spinae in the organisation of locomotor tasks (Anders et al., 2007; de Sèze et al., 2008). A 'top down approach', with the aim of attenuating postural perturbations from the lower body has been identified during dynamic tasks. This entails control of the paraspinal muscles, which stabilises the head first, and subsequently inferior parts of the upper body, in young individuals during locomotion (Winter et al., 1993).

Ceccato et al., (2009) reported anticipatory muscular activity during the anticipatory phase of gait initiation in young individuals, propagated from the superior to inferior sections of the trunk (C7 – L3), confirming a feedforward mechanism. It may be suggested that maintenance of head stability primarily relies on a feedforward command, with anticipatory activation of neck muscles at C7 level during the anticipatory phase of gait initiation in healthy young individuals (Ceccato et al., 2009). However, this mechanism remains relatively unexplored in older individuals during gait initiation.

Acceleration Attenuation

A key mechanism in maintaining head stability is the attenuation of accelerations through the trunk. Accelerations are attenuated from the lower part of the body to the head (Kavanagh et al., 2008), therefore the upper body is crucial in maintaining dynamic equilibrium to minimise oscillations in response to the lower limbs (Cromwell et al., 2001). Consequently the ability to attenuate acceleration is considered a strong balance control indicator for children, adults, and elderly individuals during steady gait (Mazzà et al., 2008; Mazzà, Zok, & Cappozzo, 2010). It has been reported that whilst young healthy adults are able to attenuate the accelerations from pelvis to head even when increasing their walking speed (Latt, Menz, Fung, & Lord, 2007), this ability is challenged in older adults (Kavanagh et al., 2003). Older adults, in particular, typically develop axial rigidity, which can impair their ability to attenuate the accelerations that are from the lower limbs during gait to the upper body, impacting on head stability. Further , difficulties in controlling the upper body accelerations have also been reported to be associated with the risk of fall (Marigold & Patla, 2008).

There is inconsistency within the literature regarding the amount of attenuation that each acceleration component undergoes. Menz et al., (2003) found higher ML accelerations at head level in older male/female adults as compared with young adults, despite smaller accelerations at pelvis level. Kavanagh et al., (2006) found significant differences between the older and young males/females only in the AP direction. Walking at faster gait speeds to comfortable reduces the ability to attenuate accelerations at head level (Mazzà, Iosa, Pecoraro, & Cappozzo, 2008), resulting in a reduced ability to maintain head stability. While it is known that head stability is threatened at fast steady state walking, studies fail to address what happens during transitory locomotive tasks such as gait initiation.

While acceleration attenuation of upper body segments has been investigated during steady state waking in both young and older adults, little is known about acceleration attenuation during the task of gait initiation. To date, only one study has examined acceleration attenuation of the upper body during the task of gait initiation, however this was in a pathological population (Buckley, Galna, Rochester, & Mazzà, 2015). The results demonstrated impaired attenuation of accelerations from the pelvis and neck to the head in adults living with Parkinson's disease. How well healthy older adults are able to attenuate such accelerations during gait initiation is yet to be examined.

Visual target implementation

Stationary visual targets are commonly used in gait studies, in both young and older populations. Although the reason for using a visual target during walking is not fully understood, one likely reason of directing gaze at a target during gait is to provide a focal point that will consequently reduce head movement, thus providing better head stability. Given that gait induces linear and angular head perturbations, causing head movements to occur across the gait cycle (Hirasaki et al 1993; Pozzo et al., 1990) and that older adults tend to implement head flexion during gait (Marigold & Patla, 2008), implementing a visual target may restrict what is naturally occurring.

Altering head orientation to focus on a visual target may involve different motor control strategies and alterations of body alignment and consequently may affect whole body stability (Buckley et al., 2005; Ustinova & Perkins, 2011). During static conditions, instructed head flexion in comparison to fixating on a stationary target at eye level has been shown to have destabilising effects evidenced by an increased postural sway in older adults (Buckley et al., 2005). However, during gait, free gaze compared to fixating on a stationary target at eye level has been shown to have no effects on whole body stability in older adults (Thomas et al., 2017). Similarly, Cromwell et al., (2002), found that older adults demonstrated improvements in head stabilisation during walking with a fixed visual target. In response to fixing their gaze on a stationary target, head-on-trunk with respect to trunk movements were similar to that of young adults, however there was no measure of whole body stability. While the effects of focussing on a stationary target has on stability measures is known in static and dynamic conditions in older adults, little is known about fixating on a target has on head movements and whole body stability during the transitory task of gait initiation.

Such information is important to ensure appropriate gait assessment techniques are employed to provide realistic and ecologically valid results. Evaluating this is not only important in contributing to the understanding of gait aspects but also to more accurately understanding and preventing loss of balance in the elderly, which can be costly and debilitating (Brunner, Eshilian-Oates, & Kuo, 2003).

CHAPTER 3

Ethical procedures and considerations

Integral to any process of investigation is the requirement to conduct the research process in an ethical manner. During the experimental procedures, it was ensured that research participants came to no harm when engaging with the research. This included getting consent from participants, respecting anonymity and confidentiality of participant's data and to fully inform the participant on how and where the information that they contribute is likely to be used.

To ensure that the older participants were comfortable with the demands of the study and could execute the tasks required, in addition to the standard ethical procedures, a different approach to the conventional one was followed. Initially, the Active Ageing Research Group approached the Lancaster and Morecambe University of Third Age branch to deliver a talk on topics of U3A's membership interest while promoting the Group's research. During that presentation, the present study was also presented to the attendees (>80) and an expression of interest contact was proposed. Once the participants made contact, the study demands were explained in detail and opportunities for questions were provided. Following that conversation, the Participant Information Sheet and Informed Consent Form (see Appendix 1 and 2) were e-mailed to them, to allow them to read them at their own pace and ask any more questions.

Subsequently, once arrangements were made for the participant's visit to the laboratory, a further verbal explanation / reminder was provided and the experimental set up was explained thoroughly, to ensure any concerns were removed. Prior to starting the experiment, participants were asked to fill out a health screening form to ensure they were medically stable to participate in the study (Appendix 3). Several trials were allowed to gain familiarisation and confidence. Participants were encouraged to visit with a friend (regardless of whether the friend would take part or not) to ensure comfort during the process.

As part of the Thesis, experiments were also conducted at the University of Rome *Foro Italico,* in Italy, a similar approach was followed in order to recruit participants. Local older adults groups were approached to promote the project whilst any individuals that were interested in participating were given the Participant Information Sheet (see Appendix 4) and invited to the Laboratory to carry out the testing. However, the added challenge was the translation of the study's demands to the Italian language, ensuring that all relevant Ethical aspects were covered while the study's purpose remained unaltered. Several discussions with colleagues took place to ensure that the translation of the material was accurate, while meeting the Foro Italico's ethical requirements. The above procedures were deemed to increase comfort of the older adults participating in the study, as it increased the information provided to the participants but also decreased the possible apprehension of the experimental environment. Prior to starting the experiment, participants gave consent to participate in the study (see Appendix 5) and were asked to fill out a health screening form to ensure they were medically stable to participate in the study (see Appendix 6). Ethical approval from each Institution was granted to carry out the each of the experiments.

CHAPTER 4

Thesis structure and presentation

To examine the effect of head position and the strategies underpinning head stability, and whole body stability during steady state gait and gait initiation, four studies were carried out. A summary for each is presented below and the studies are presented in this Thesis as experimental standalone papers in the next four chapters.

Study 1

This study aimed to examine A) if head position and gait spatiotemporal parameters were altered when walking with free head movement and with a visual target, and B) how the effect of using a visual target may change head position at different walking speeds. Walking trials were performed on an unobstructed 9m flat walkway under two visual conditions: walking with no visual target, and walking focusing on a fixed visual target set at eye level. All trials were completed at three walking speeds (slow, comfortable and fast).

Head and trunk flexion angles were measured using 2D video analysis, whilst spatiotemporal gait parameters (velocity, step length, double support time, step time and gait stability ratio) were recorded. A mixed design 2 x 2 x 3 ANOVA was performed, with age group as a between subject factor, and visual condition and walking speed as within subject factors.

Study 2

This study aimed to examine neuromechanical strategies underpinning head stabilisation and whole body stability during gait initiation between young and older females. Participants initiated gait in a straight line for at least three steps at their comfortable walking speed. In addition, they were asked to focus on a fixed visual target set at eye level.

Head stabilisation and whole body stability was assessed using 3D analysis and EMG. Variability of angular displacement (AvgSD and principal component analysis (PCA)), acceleration attenuation coefficient and the onset and activation amplitude of the neck (SCM and NE) and trunk (ES) muscles were used as a measure of upper body stability. Whole body stability was quantified using an adapted version of the margin of stability. Independent sample t–tests were used to examine for differences between groups.

Study 3

This study aimed to examine the effect of free head movement on head stability and the neuromechanical strategies underpinning head stability, and whole body stability in young and females during gait initiation. Participants initiated gait in a straight line for at least three steps at their comfortable walking speed under two target conditions: with no visual target, and focussing on a fixed visual target set at eye level.

Angular head displacement, head stabilisation, gait spatiotemporal parameters and whole body stability was assessed using 3D analysis and EMG. Variability of angular displacement (AvgSD), acceleration attenuation coefficient and the onset and activation amplitude of the neck (SCM and NE) and trunk (ES) muscles were used as a measure of upper body stability. Whole body stability was quantified using an adapted version of the margin of stability. Paired comparisons were made to examine for differences between the two target conditions.

Study 4

This study aimed to examine the effect of initiating gait at different speeds on neuromechanical mechanisms underpinning head stability, and whole body stability in young and older individuals. Participants initiated gait in a straight line for at least three steps under two speed conditions: comfortable and fast.

Head stabilisation, gait spatiotemporal parameters and whole body stability was assessed using 3D analysis and EMG. Variability of angular displacement (AvgSD), acceleration attenuation coefficient and the onset and activation amplitude of the neck (SCM and NE) and trunk (ES) muscles were used as a measure of upper body stability. Whole body stability was quantified using an adapted version of the margin of stability. Paired comparisons were made to examine for differences between the two target conditions.

CHAPTER 5

Head flexion and different walking speeds do not affect gait stability in older females

A version of the work from this chapter has been published as:

Maslivec, A., Bampouras, T.M, Dewhurst, S. (2017). Head flexion and different walking speeds do not affect gait stability in older females. Humans Movement Science, *55*, 87-93.

Abstract

Head flexion is destabilising in older individuals during quiet stance, yet the effect head flexion has on gait is not known. The study examined whether head flexion and gait parameters were altered when walking freely and fixed to a visual target, at different walking speeds. 15 young (23 ± 4 years) and 16 older (76 ± 6 years) healthy females walked at three different walking speeds (slow, comfortable, and fast) under two visual conditions (natural and fixed [focusing on a visual target set at eye level]). Head flexion was assessed using 2D video analysis, whilst gait parameters (step length, double support time, step time, and gait stability ratio) were recorded during a 9 m flat walkway. A mixed design ANOVA was performed for each variable, with age as the between-subject factor and, visual condition and walking speed as within-subject factors. When walking freely, older displayed a greater need for head flexion between walking speeds ($p < 0.05$) when compared to young. Walking under fixed condition reduced head flexion at all walking speeds in the older ($p < 0.05$), but had no effect on the young. Walking at different speeds showed no difference in head flexion when walking under either visual condition and had no effect on gait stability for both groups. Despite older displaying differences in head flexion between visual conditions, there was no effect on gait parameters. Walking speed presented trivial difference in head flexion in older females, whilst overall gait stability was unaffected by different walking speeds

Introduction

Walking is a habitual activity, requiring transition from a stable to an unstable position, i.e. from double to single leg support. Such movement results in a continuous perturbation in the balance equilibrium, as the centre of mass (COM) alters in relation to the also changing base of support (BOS) (Woollacott & Tang, 1997). This can prove challenging for older individuals (Ihlen et al., 2012; Prince et al., 1997), reflected by the fact that the majority of falls occur during walking in older individuals (Rubenstein, 2006).

Given the challenge gait poses to older individuals, head flexion is typically implemented to identify lower limb trajectory and enable better footfall vision (Marigold & Patla, 2008b) and to gather more information when walking towards an obstacle (Muir et al., (2015). This increased head flexion, whilst enabling better lower visual vision, may however have a negative impact on postural control, and subsequently, on fall risk (De Groot et al., 2014). During walking, at heel strike, the pelvis moves posteriorly due to ground reaction forces, which consequently causes the upper body to rotate forward over the feet, altering the COM towards the limits of the BOS, thus challenging balance (Winter, 1995). A flexed head, weighing ~7% of overall body mass (de Leva, 1996), may exacerbate this forward shifting of the COM, further threatening stability or making recovering from an unexpected perturbation difficult (De Groot et al., 2014). During quiet standing in older individuals Buckley et al., (2005) reported a destabilising effect of head flexion. Although this destabilising effect has been seen in static conditions, it has not been examined in dynamic conditions and given the previously mentioned problems with falls during walking, it is important to examine the effect head flexion could have to either consider it in future studies and interventions or reject it as a contributor to gait instability.

Head flexion has also been shown to be influenced by gait speed in young individuals. Hirasaki, Moore, Raphan, & Cohen (1999) reported that at speeds >1.2$m \cdot s^{-1}$, there was an increased magnitude of head pitch displacement, such that a greater amount of head flexion was observed. Although gait speed is commonly assessed as an outcome measure of functional capacity and gait ability in the older population (Bongers et al., 2015; Montero-Odasso et al., 2004; Toots et al., 2013; Verghese, Holtzer, Lipton, & Wang, 2009), it has rarely been considered the subject of investigation. In other words the effect of different walking speeds on head flexion has rarely been examined in older adults. During day to day life, however, walking at different speeds is required, for example, when walking faster due to being late for an appointment, or in contrast, walking slower to negotiate a busy shopping centre. If the findings by Hirasaki et al., (1999) in young also hold true for older adults, it is feasible that as walking speed increases, concurrently increasing head flexion, postural control may also be increasingly challenged.

In addition to the postural control issue that head flexion could cause during heel strike, it also raises an important methodological question. Gait studies typically instruct participants to focus on a visual target fixed at eye level to standardise head position during walking (Cromwell et al., 2002; Hirasaki et al., 1999). Such instructions, which constrain head movement, may mask a true effect, as they would reduce the naturally occurring head flexion. In turn, this could impact on gait stability and postural control, most likely underestimating the true challenge walking poses on older individuals and potentially reaching to erroneous results and less specific intervention advice. Therefore, understanding differences between a natural head position and a typical standardised head position, at different walking speeds, is warranted.

The aim of the study was twofold; to examine A) if head flexion and gait parameters were altered when walking without and with a visual target, and B) how the effect of using a visual target may change at different walking speeds. It was hypothesised that the implementation of a visual target would restrict head flexion, which in turn, would alter gait pattern. Females were the focus of the study as it has been reported that females whole body stability declines to a greater extent than males (Wolfson, Whipple, Derby, Amerman, & Nashner, 1994) and tend to fall more often (Schultz, Ashton-Miller, & Alexander, 1997).

Methods

Participants

Sixteen healthy older females (age 75.5 ± 6.2 years, height 1.62 ± 0.04 m, body mass 74 ± 6.8 kg) and 15 healthy young females (age 23±3.5years, height 1.67±0.04m, body mass 63.3 ± 6.0 kg) participated in the study. Older females were recruited from local community groups while young were students at the Institution. All participants had no known neuromuscular disorders, impaired postural alignment such as kyphosis, osteoarthritis or neck related pain, while older participants were community residing, functionally independent, considered medically stable (Greig et al., 1994). All participants were able to perform all conditions without the use of bifocal or multifocal spectacles and had an uncorrected visual acuity ≥20/100 measured on the day of testing.

Ethical approval was obtained from the Institutional Ethics Committee and written informed consent was obtained prior to testing.

Protocol

Walking trials were performed on an unobstructed 9m flat walkway under two visual conditions; walking with no visual target and walking with a visual target. In the no visual target condition, no instructions were given to participants as to where to orient their gaze whereas in the visual target condition, participants were instructed to focus on a stationary target located at eye level, 3 m directly ahead of the end of the walkway. The visual target consisted of a black circle (15 cm diameter) on a white background. The position, size and distance of the visual target were decided following pilot testing, which allowed a target which could be comfortably seen by the participants without excessive eye focusing effort. All participants underwent familiarisation with each visual condition and speed, and confirmed they were able to clearly see the target from the beginning of the walkway without the use of glasses.

Three trials were completed at three walking speeds (slow, comfortable, and fast). Instructions for walking speeds were given by associating the walking speeds to everyday activities (Thomas, De Vito, & Macaluso, 2007). Slow walking speed was described as 'the way you would walk during relaxed window shopping'; comfortable as 'how you would normally walk in a relaxed mood' and fast walking as 'how you would walk when late for an appointment'. Participants completed 18 trials in total (three trials per walking speed in both no visual target and visual target condition). The order of visual condition and walking speed was randomised and the mean of three trials was used for analysis.

Head flexion

To measure head flexion, a marker was placed on the apex of the skull (attached to a headband secured around the participant's head, horizontal to the ground, during standing in the reference body position) and a marker placed on the seventh cervical vertebrae (C7). The angle formed by the vertical axis (passing through the C7 marker) and the straight line between the C7 and the apex of the skull markers, was measured as head flexion angle.

Trunk flexion

To account for any trunk flexion and ensure any differences seen were the result of head flexion alone, trunk angle was also measured. A third marker was placed at the hip joint (firmly attached to a belt securely fastened around the participants' hips, horizontal to the ground, with the participant standing in the reference body position). The angle formed by the vertical axis (passing through the C7 marker) and the straight line between the hip joint and the C7 markers, was measured as trunk flexion. Both segments can be seen in the schematic diagram (Figure 2).

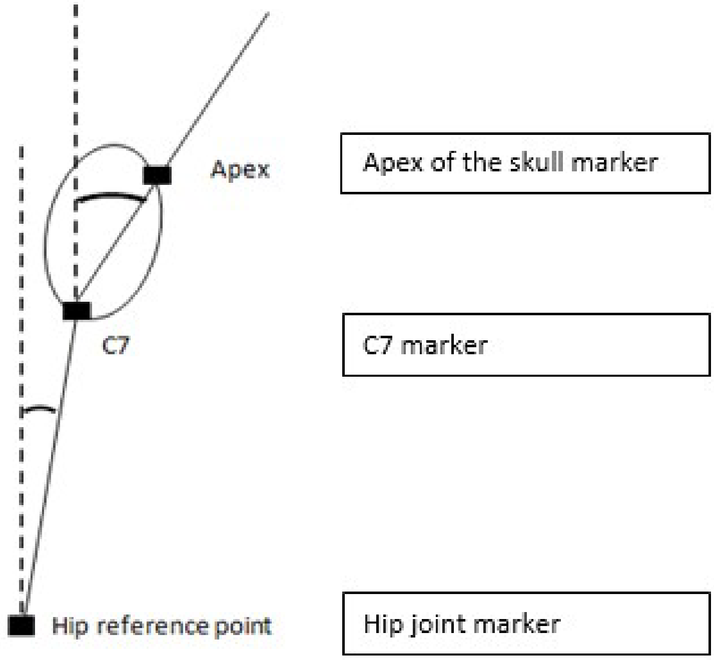

Figure 2. Sagittal plane angles of the head and trunk. Black filled squares indicate markers placed on body landmarks

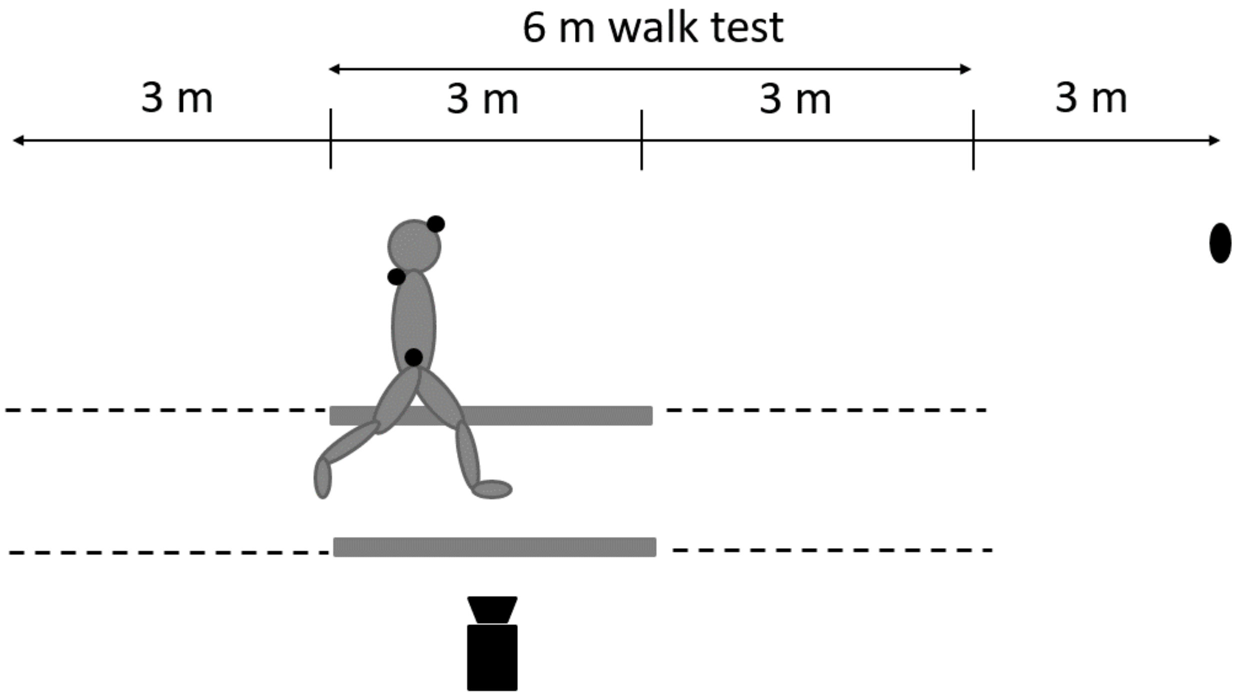

Figure 3. Experimental setup

Head and trunk flexion angles were measured in the sagittal plane using 2D video analysis (Kinovea for Windows, Version 0.8.15, www.kinovea.org) with a sampling frequency of 100 Hz, at the first heel strike of the left foot (first frame the heel made contact with the ground) as soon as the participant crossed the 6 m marker. To obtain a realistic understanding of changes in head and trunk flexion angles by avoiding 'postural adjustments' during standing measurements (Thomas, Bampouras, Donovan, & Dewhurst, 2016) the difference in angle from comfortable to slow walking speed and comfortable to fast walking speed (Δ values) were calculated for both age groups and both visual target conditions. Positive angle values indicated greater head and trunk flexion of the given walking speed in comparison to comfortable walking speed.

Walking velocity

The 6m walk test was used to measure walking velocity at each walking speed as it has been shown to have high reliability for comfortable and fast walking (ICC=.97 and .96, respectively) (Steffen, Hacker, & Mollinger, 2002). Walking velocity was calculated from the time taken to walk between 3m and 9m (6m) of the walkway using wireless timing gates (Brower timing gates, Draper, UT, USA) set at hip height.

Gait parameters

Gait parameters were recorded using the Optojump Next Jump System ® (Microgate SRL, Bolzano, Italy) and included step length, double support time, and step time from the middle 3m of the walkway. These gait variables were selected as they are frequently reported in the literature and been shown to be sensitive measures of changes in gait (Callisaya, Blizzard, Schmidt, McGinley, & Srikanth, 2010). The Optojump Next Jump System ® (Microgate SRL, Bolzano, Italy) is an optical measurement system consisting of two infrared photocell bars that can derive contact time of each foot from the breaking of the transmitted beam. Gait stability ratio (GSR, calculated as cadence / velocity) has been developed from 2D gait analysis of flat walking and was used as a measure of walking stability. A higher GSR indicates a greater proportion of the gait cycle is spent in contact with the floor, thus avoiding the dynamic components of walking (Ronita L Cromwell & Newton, 2004), as one would do when a greater need for stability is required.

Statistical analysis

To assess intrarater reliability of angle and gait measurements, sensitivity (typical error (TE), calculated as standard deviation of the change scores between measurement / $\sqrt{2}$) and intraclass correlation coefficient (ICC, calculated as $1 - TE^2$ / mean between-subject standard deviation between measurements) between the three trials were obtained from a customised spreadsheet (Hopkins, 2000).

Statistical analyses were carried out using IBM SPSS, v19 (IBM Corporation, Armonk, NY, USA). Normality of data was examined using the Shapiro-Wilk test and confirmed for all variables. A mixed design ANOVA was performed for each variable, with age group as a between-subject factor and visual condition and walking speed as within-subject factors. When a main effect existed, between group differences were examined using independent samples t-tests, while within group comparisons were conducted using a repeated measures ANOVA followed by dependent-samples t-tests, if a difference was found, with Bonferroni correction. For comparisons which showed significant differences, effect size (ES) was calculated to provide indication of the magnitude of difference, with 0.2, 0.5, and 0.8 representing a small, medium, and large effect respectively (Fritz, Morris, & Richler, 2012). An alpha level was set at $p < 0.05$. Data are presented as mean ± standard deviation (SD).

Results

Data for gait parameters are presented in Table 1. For clarity, effect sizes for significant differences are reported only if they were below moderate (0.05).

Reliability

Head and trunk flexion ICCs for both age groups in all measurements ranged from 0.89 – 0.90, indicating high reliability, whilst only a small TE (<1.12°) was present. Similarly, ICC for step length ranged from 0.83-0.95, with only fast walking with visual target for the young exhibiting a lower ICC (0.77).

Head flexion

There was a significant main effect of age ($p = 0.001$), walking speed ($p = 0.021$) and visual condition ($p = 0.001$) for head flexion angle. There were significant interactions for age × visual condition ($p = 0.001$) and walking speed × visual condition ($p = 0.011$). Delta values showed head flexion was significantly greater between slow-comfortable walking and comfortable-fast walking during the no visual target condition in older ($p = 0.01$, ES = 0.06). Delta values showed head flexion was significantly lower at slow-comfortable walking compared to comfortable-fast walking during visual target condition in young ($p = 0.032$, ES = 0.09). Changes in head flexion between walking speeds for young and older in each visual condition are depicted in Figure 3.

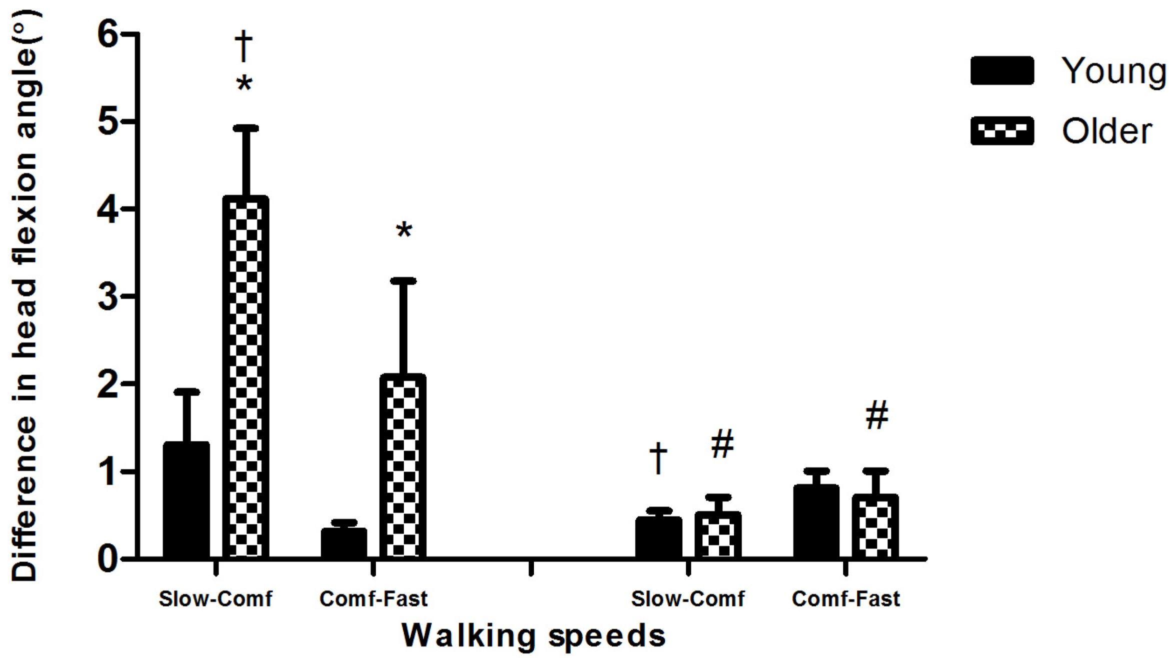

Figure 4. Δ values in head flexion angle between walking speeds for young and older in each visual condition. Data is presented as mean ± SD. * indicates significant differences between age group, # indicates significant difference between no visual target and visual target conditions and † indicates significance between slow-comfortable and comfortable-fast walking speed. Slow-comf = slow to comfortable walking speed. Comf-fast = comfortable to fast walking speed.

Trunk flexion

There was a significant main effect of age ($p = 0.013$) and visual condition ($p = 0.02$) on trunk angle. There were significant interactions for age × visual condition ($p = 0.026$). There was no difference in trunk flexion at any walking speed or visual condition between young and older Delta values showed older displayed a greater increase in trunk flexion angle from the fast walking to comfortable walking ($p = 0.001$, ES = 0.08) in the no visual target compared to the visual target condition, while there were no differences between visual conditions in young.

Walking velocity

There was no difference in gait velocity between visual conditions for either group. Predictably, walking velocity significantly increased with walking speed in both age groups ($p = 0.019$, ES = 0.24-0.49 and $p = 0.038$, ES = 0.28–0.39 for young and older respectively), indicating participants successfully followed walking speed instructions. Young were significantly faster at comfortable and fast walking compared to older ($p = 0.008$), however there was no difference in gait velocity at slow walking speed between groups.

Gait parameters

Gait parameters for both age groups for all conditions are shown in Table 1. There was a significant main effect of age and walking speed for all gait parameters ($p = 0.001$) with the exception of GSR. There was a significant interaction for age × walking speed ($p = 0.002$). When comparing visual conditions, there were no differences in any gait parameters. When comparing between group differences, older had a significantly shorter step length and step time compared to young at all walking speeds ($p = 0.032$), whilst double support time was significantly greater in older adults at all walking speeds ($p = 0.01$). Older displayed higher GSR than young at all walking speeds ($p = 0.028$). With respect to within group differences in walking speeds, step length significantly increased from slow to comfortable to fast walking speeds whereas double support time and step time significantly decreased in duration across the same conditions in both age groups ($p = 0.026$). There was no difference in GSR between walking speeds for either age group.

Table 1. Gait parameters during both visual conditions for each walking speed for young and older. Data is presented as mean ± SD.

		Young		Older	
Variable	Speed	No visual target	Visual target	No visual target	Visual target
Gait Speed [a, b] ($m \cdot s^{-1}$)	*Slow*	1.05 ± 0.15	1.10 ± 0.20	0.94 ± 0.21	1.01 ± 0.26
	Comf	1.58 ± 0.20	1.51 ± 0.10	1.32 ± 0.28	1.42 ± 0.29
	Fast	2.37 ± 0.30	2.23 ± 0.20	1.81 ± 0.20	1.81 ± 0.32
Step Length [a, b] (cm)	*Slow*	64.1 ± 3.94	65.16 ± 4.77	59.3 ± 5.20	65.4 ± 5.90
	Comf	73.8 ± 5.03	74.40 ± 4.38	68.9 ± 4.90	70.6 ± 4.20
	Fast	86.9 ± 4.96	86.18 ± 5.48	76.0 ± 5.10	76.6 ± 5.40
DST [a, c] (s)	*Slow*	0.39 ± 0.05	0.38 ± 0.07	0.45 ± 0.13	0.43 ± 0.11
	Comf	0.27 ± 0.04	0.27 ± 0.03	0.35 ± 0.21	0.31 ± 0.06
	Fast	0.15 ± 0.04	0.16 ± 0.03	0.20 ± 0.03	0.22 ± 0.04
Step Time [a, b] (s)	*Slow*	0.61 ± 0.04	0.57 ± 0.04	0.42 ± 0.07	0.41 ± 0.07
	Comf	0.49 ± 0.03	0.49 ± 0.03	0.29 ± 0.04	0.29 ± 0.02
	Fast	0.38 ± 0.05	0.42 ± 0.05	0.21 ± 0.03	0.22 ±0.030
GSR [a] ($Step \cdot m^{-1}$)	*Slow*	1.62 ± 0.18	1.61 ± 0.16	2.48 ± 0.16	2.41 ± 0.22
	Comf	1.41 ± 0.12	1.39 ± 0.14	2.45 ± 0.70	2.53 ± 0.12
	Fast	1.16 ± 0.07	1.17 ± 0.12	2.92 ± 0.23	2.62 ± 0.26

Significant effects on the variables as follows.

a $p < 0.05$ for significant effects of age

b $p < 0.05$ for significant effect of walking speed

DST, double support time; GSR, gait stability ratio

Discussion

The purpose of the present study was to examine if head flexion and gait parameters were altered when walking without and with a visual target, and how the effect of using a visual target may change at different walking speeds. Findings showed that older individuals adopted greater head flexion at all walking speeds in the no visual target condition compared to young. Head flexion was constrained to that similar of young when walking with a visual target with no changes in gait parameters.

It has been suggested that older individuals implement head flexion to enable better vision to identify potential hazards located at ground level (Marigold & Patla, 2008; Menant et al., 2010). Gait in the present study was over a known flat walkway free of obstacles; yet older still implemented a greater need for head flexion compared to young during the no visual target trials between speeds. Consequently, this raises the question of why older demonstrated greater difference in flexion than young, despite not being exposed to external threats. It could be suggested that head flexion is adopted by older individuals regardless of environment or threat perception. Alternatively, head flexion may be an adaptation which older individuals have become accustomed to due to having to negotiate obstacles on an everyday basis (Keller Chandra et al., 2011). The combination of older individuals, typically having an impaired lower visual field (Freeman, Muñoz, Rubin, & West, 2007) and the need to look two steps (of shorter step length) ahead (Patla & Vickers, 2003) to ensure a clear path, may have caused older to implement greater head flexion. Walking in the way they were used to, possibly increases the subjective feeling of stability regardless of walking environment, contributing to the lack of difference in GSR between visual conditions. Similar to findings by (Hirasaki et al., 1993), greater head movement during double stance was found. Hirasaki et al., (1993), however, reported increased head extension whilst the present study found increased head flexion. The reasons for this can possibly be attributed to a difference in population characteristics between studies. Interestingly, DST was not significantly different between young and older in Hirasaki et al., while the present results showed that older did have significantly longer DST. Thus, results show that differences were not only seen for head position but actually for gait variables, lending to the speculation that differences may be due to differences in population characteristics. Hirasaki et al., proposed that older may have reduced flexibility of the vertebral column preventing flexion of the head, however very little information is given about the older participants in the study. Older participants in the present study were healthy and physically active, therefore flexibility of the vertebral common may not have been a problem, allowing a more unrestricted head movement.

The hypothesis of focusing on a visual target (to reduce head flexion) altering gait parameters, can be rejected as gait parameters remained similar in both visual conditions. Sway has been seen to be affected by head flexion during static conditions (Buckley et al., 2005). Despite differences in head flexion between visual conditions for the older, there was no difference in GSR values, suggesting head flexion whilst walking did not pose any additional fall risk. In the present study, head flexion was measured independent of the trunk. Previous studies have shown that the trunk flexion can influence gait results (Saha, Gard, & Fatone, 2008), however the present study found no trunk flexion, demonstrating that trunk was not responsible for head flexion.

The original hypothesis raised an important methodological question. Gait studies typically instruct participants to focus on a visual target fixed at eye level to standardise head position during walking (e.g. Cromwell et al., 2002; Hirasaki et al., 1999). Such instructions, which constrain head movement, may have masked a true effect, as they would reduce the naturally occurring head flexion, supported by the findings in the present study. It was hypothesised that this in turn, this would impact on gait stability and postural control, most likely underestimating the true balance challenge walking poses on older individuals and potentially reaching to erroneous results and specific intervention advice. From the findings of the present study, however, this does not appear to be so for the population studied.

Hirasaki et al., (1999) reported that at speeds >1.2 $m{\cdot}s^{-1}$, there was an increased magnitude of head pitch displacement and a further increase when walking at speeds of 2 $m{\cdot}s^{-1}$ in young individuals. These results are reflected in the present study as young had greater head flexion between comfortable – fast (2.53 $m{\cdot}s^{-1}$ compared to slow-comfortable (1.51 $m{\cdot}s^{-1}$). The present results support Hirasaki et al., previous reports that head displacement changes with walking speed for young adults, however the opposite effect was found in older as older produced greater head flexion at slow – comfortable walking speed compared to comfortable – fast walking speed (with low effect sizes, however). Despite trivial differences in head flexion at different speeds, overall gait stability was unaffected in both age groups.

GSR values did not change despite the decrease in double support time with the associated increase in walking speed, suggesting the different walking speeds did not pose a perceived increased threat to overall gait stability. Kang & Dingwell (2008) suggested that factors such as loss of strength and flexibility must be taken into account rather than simply walking speed when identifying gait variability and instability in the older population. The present data supports the notion that other measures are contributing to gait instability and that ageing effects on speed are not straight forward, thus a more holistic assessment is warranted.

Older individuals can have a kyphotic posture, an exaggerated anterior curvature which tends to increase with age (Ailon, Shaffrey, Lenke, Harrop, & Smith, 2015; Katzman, Wanek, Shepherd, & Sellmeyer, 2010). This impaired postural alignment affects physical functioning and can have implications on fall risk for the elderly (Ailon et al., 2015; Katzman et al., 2010). The participants in the present study were free from such condition and it would be expected that kyphotic individuals would present different findings to the current participants. Further, to ensure that changes in head flexion angle could be attributed to head movement rather than trunk flexion, the two segments were examined separately. The results showed that trunk flexion did not change in any substantial way (as indicated by the very small effects sizes), suggesting that trunk flexion remained stable when changing between visual target conditions and walking speeds.

A limitation to the study was that although a visual target approach was used, it was not quantified using an eye tracking device to examine whether participants was fixating on the target. However, the use of the target was not aimed to fixate gaze, but rather to adjust the head position by fixing the gaze. This was achieved, even if eyes were not always on the target, as the instruction of keeping the head up was followed.

Conclusion

Older females displayed greater head flexion compared to young when walking without a visual target; however, head flexion was constrained to that similar of young when walking with a visual target. Despite the difference in head flexion between visual conditions in older, there was no effect, either beneficial or detrimental to gait parameters and stability. Walking speed presented trivial difference in head flexion, whilst overall gait stability was unaffected by different walking speeds.

CHAPTER 6

Mechanisms of head stability during gait initiation in young and older women: a neuromechanical analysis

Maslivec A, Bampouras TM, Dewhurst S, Macaluso A, Vanozzi G, Laudani L. Mechanisms of head stability during gait initiation in young and older women: a neuro-mechanical analysis. Journal of Electromyography and Kinesiology. 38 (2018) 103-110.

A version of the work from this chapter was presented at the BASES Biomechanics Interest Group meeting, 3rd March 2016, Liverpool John Moore's University, UK.

A version of the work from this chapter was presented at the European College of Sport Sciences, 6 - 9 July 2016, Vienna, Austria.

Abstract

Decreased head stability has been reported in older women during locomotor transitions such as the initiation of gait. The aim of the study was to investigate the neuromechanical mechanisms underpinning head stabilisation in young and older women during gait initiation. Eleven young (23.1 ± 1.1 years) and 12 older (73.9 ± 2.4 years) women initiated walking at comfortable speed while focussing on a fixed visual target. A stereophotogrammetric system was used to assess variability of angular displacement and RMS acceleration of the pelvis, trunk and head, and whole body stability in the anteroposterior and mediolateral directions. Amplitude and latency of muscle activation in the sternocleidomastoid, neck extensor, and upper and lower trunk muscles were determined by surface electromyography. Older displayed higher variability of head angular displacement, and a decreased ability to attenuate accelerations from trunk to head, compared to young in the anteroposterior but not mediolateral direction. Moreover, older displayed a delayed onset and greater relative amplitude of neck flexors activation than young. In conclusion, the age related decrease in head stability could be attributed to an impaired ability to attenuate accelerations from trunk to head along with altered activation patterns of trunk and neck muscles.

Introduction

Stabilisation of the head in space is fundamental to optimise inputs from the visual, vestibular, and somatosensory systems and, therefore, to maintain whole body balance during locomotion (Kavanagh et al., 2005; Pozzo et al., 1990). Decreased head stability has been reported in older individuals during different types of locomotion, including steady-state walking (Cromwell et al., 2001) and locomotor transitions such as gait initiation (Laudani et al., 2006). Transitory locomotor tasks, in particular, involve complex interactions between neural and mechanical factors which may challenge whole body balance to a greater extent than unconstrained walking (Nagano et al., 2013). This challange may help to explain why the number of falls in older individuals are frequent during locomotor transitions such as gait initiation and termination (Winter, 1995).

In young individuals, head stabilisation is ensured during steady state walking by cyclically controlling the upper body accelerations caused by the lower body movement, through coordinated movements of the trunk (Kavanagh et al., 2006). In older individuals, however, control of acceleration from the lower to the upper body during steady state walking has been shown to be less effective than in young individuals (Mazzà et al., 2008). As walking is initiated from a standing position, steady state velocity is achieved within the first step (Breniere & Do, 1986); due to the transient nature of gait initiation, therefore, higher upper body accelerations are likely to be seen compared to steady-state walking. Subsequently, this could challenge the control of upper body acceleration and therefore head stabilisation in older individuals. To the best of the authors' knowledge, however, there are no studies focusing on the control of upper body accelerations during the transitory task of gait initiation in healthy young and older individuals.

From a neuromuscular point of view, electromyography (EMG) studies have highlighted the importance of trunk paraspinal muscle activation in actively attenuating postural perturbations from the lower body during locomotor tasks (Anders et al., 2007; de Sèze et al., 2008). A 'top down' anticipatory control of erector spinae muscles, which stabilises the upper trunk first and subsequently the lower trunk, has been reported in young individuals during gait (Winter et al., 1993; Prince et al., 1994). In line with that, Ceccato et al., (2009) have reported a metachronal activation of erector spinae muscle occurring during the preparation of the first step for gait initiation. To date, most of the studies on older individuals have revealed characteristic age related changes of muscle recruitment in the lower limb during gait initiation. For instance, older individuals have been shown to initiate walking with greater co-contraction of the lower leg muscles (Khanmohammadi et al., 2015a) and a delayed activation of the tibialis anterior muscle compared to young individuals (Khanmohammadi et al., 2015b). It is not known, however, whether older individuals would effectively recruit the trunk muscles and/or adopt an anticipatory control in order to actively aid stabilisation of the head during the transitory phase of gait initiation.

The aim of the present study, therefore, was to investigate the neuromechanical mechanisms underpinning head stabilisation in young and older individuals during gait initiation. In particular, to examine control of upper body accelerations and muscle activation patterns of the trunk and neck, which represent two of the main neuromechanical strategies underpinning head stability. Additionally, the control of dynamic balance in young and older participants was investigated by evaluating whether the conditions for dynamical stability were met within each age group. It was hypothesised that older women would a) demonstrate reduced ability to attenuate acceleration from lower to upper parts of the upper body, b) have impaired muscle activation pattern of the trunk and neck and c) have reduced whole body stability, compared to the younger women.

Methods

Participants

Eleven healthy young (age: 23.1 ± 1.1 years, height: 1.64 ± 0.71 m, body mass: 57.5 ± 6.7 kg) and 12 healthy older (age: 73.9 ± 2.4 years, height: 1.63 ± 0.45 m, body mass: 66.2 ± 10.2 kg) females volunteered to participate in the study. Women were the focus of the study as it has been reported that their whole body stability declines to a greater extent than males (Wolfson et al., 1994) and tend to fall more often (Schultz et al., 1997). Older participants were considered 'medically stable' to participate in the study, according to exclusion criteria for older people in exercise studies (Greig et al., 1994). No participants had any history of neurological disorders that would affect their balance or gait ability, and were able to complete the task without the use of bifocal or multifocal spectacles. Written informed consent was provided by all participants and ethical approval was given by the institution's ethics committee.

Experimental protocol and equipment

Participants wore their everyday flat shoes. Instructions were to stand as still as possible with their feet in a comfortable position at shoulder width apart, and with the arms alongside the trunk. Participants were verbally instructed to start walking on their own accord from a single force platform (Bertec Corp, Worthington, OH) and to continue to walk forwards in a straight line for at least three steps at their comfortable walking speed. In addition, they were instructed to focus on a fixed visual target, which was set at eye level for each participant and located five metres ahead of the starting position. The position, size and distance of the visual target were decided following pilot testing, which allowed a target which could be comfortably seen by the participants. The right leg was used as the starting (swing) leg for all trials. Starting feet position at shoulder width apart was marked on the force platform and participants repositioned themselves in that position for each trial. In total five trials were completed and analysed.

A seven camera motion analysis system (VICON, Oxford Metrics, London, England) was used to record and reconstruct the 3D position of 35 reflective markers placed on body landmarks, following the Davis protocol (Davis et al., 1991) with a sampling rate of 100 Hz. The VICON whole body plug-in-gait model was used to define a local anatomical reference frame for the pelvis (markers on the left and right anterior and posterior superior iliac spines), trunk (markers located at the clavicle and sternum level as well as at C7 and at T10), and head (four markers, placed on the left and right side of the front and back of the head) and then calculating the relevant kinematic data. The force platform was used to track COP motion with a sampling frequency of 1000 Hz and then resampled.

Temporal aspects of gait initiation were determined relative to COP onset. The onset of COP displacement was automatically estimated as the time point at which the AP component of the ground reaction force overcame the threshold defined as 3 standard deviations of its peak- to peak value during static posture AP force. Gait initiation was performed as a whole movement and divided into two phases*: 1) anticipatory phase*, which lasted from the onset of COP motion to the instant of toe off of the swing limb 2)*: execution phase*, which lasted from toe off of the swing limb to the instant of toe off of the stance leg. Temporal events of gait initiation were obtained from both position and velocity curves derived from markers placed on the calcaneus and fifth metatarsal bones (Mickelborough et al., 2000). These events corresponded to the instants of heel off, toe off and heel contact of the swing limb. Angular displacement and the motion of the upper body segments (pelvis, trunk, and head) were measured in the AP and ML direction. Additionally, whole body COM was recorded as a weighted sum of all body segments using the whole plug-in-gait model in the AP and ML direction.

Muscle activity was determined by surface EMG recordings (BTS Bioengineering, Italy). EMG signals were collected bilaterally using bipolar disposable electrodes (1cm disc-electrodes, 2cm inter-electrode distance) from the: sternocleidomastoid (SCM), neck extensors (NE), and erector spinae (ES) at the level of T9 and L3, with a sampling frequency of 1000 Hz. Electrode sites were prepared by gently abrading the skin to ensure good contact. For the SCM, electrodes were positioned at 1/3 of the distance from the sternal notch to the mastoid process at the distal end overlying the muscle belly; the NE electrodes were positioned over the distal half of the distance between the base of the occiput and the spinous process of the seventh cervical vertebrae (Falla et al., 2004) whilst for the ES, electrodes were placed 2 cm lateral of the spinal process at T9 and L3.

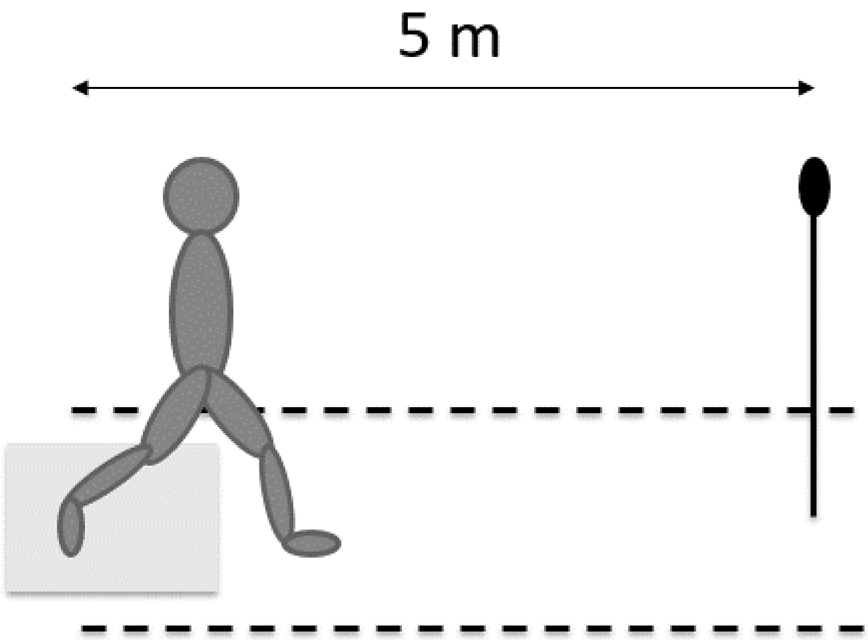

Figure 5. Experimental setup

Figure 6. Marker and electrode placement

Data analysis

Variability of angular displacement

Angular displacement of the pelvis, trunk, and head was filtered using a second-order low-pass Butterworth filter with a cut-off frequency of 5 Hz and re-scaled to the first value of the anticipatory phase. To quantify variability of the pelvis, trunk, and head motion during gait initiation, the average standard deviation (AvgSD) was calculated using the following equation:

$$AvgSD = \sqrt{\frac{\sum \theta^2}{100}}$$

θ = Angular displacement of the segment.

This measure has previously been used to assess the stability of individual body segments, with decreased variability indicating increased segment stability (Laudani et al., 2006). To further quantify the variance of angular displacement waveforms of the pelvis, trunk, and head in the AP and ML direction, principal component analysis (PCA) was applied to each data set (young and older) computed by a customised Matlab 7.5 script (Mathworks, Inc, USA). The objective of using PCA was to transform the waveform data to reduce the number of variables but retain most of the original variability in the data (Kirkwood et al., 2011). The first principal component (PC) accounts for the highest variability in the data, with subsequent PCs accounting for the remaining variability. For this analysis, a 90% trace variability threshold was used to determine the number of PCs required to retain the most common patterns of angular displacement within each age group. Angular displacement traces used for the PCA were time normalised by interpolation into 100 data points for each phase, corresponding to 1% intervals (anticipatory phase: 1-100%, execution phase: 101-200%).

Attenuation of upper body accelerations

Acceleration of the pelvis, trunk and head segments was calculated by double derivative of the 3D position of the origin of each upper body segment reference frame in the AP, ML and cranio-caudal (CC) direction. It was computed by a customised Matlab 7.5 script (Mathworks, Inc, USA) and filtered using a second-order low-pass Butterworth filter with a cut- off frequency of 5Hz. The magnitude of acceleration of each segment was calculated using the root mean square (RMS) in the AP, ML and CC direction. RMS acceleration values are known to be influenced by gait velocity (Kavanagh & Menz, 2008), thus AP and ML RMS acceleration were normalized by CC acceleration RMS as proposed by Iosa et al., (2012). The ability to attenuate accelerations through the upper body segments was quantified using the attenuation coefficient expressed as a percentage. The attenuation coefficient describes the ability to reduce accelerations from inferior to superior segments, with reduced linear acceleration from inferior to superior parts of the upper body used as an indicator of upper body stability (Summa et al., 2016). The attenuation coefficients were calculated using RMS values of each segment as follows (for both AP and ML direction):

$$\mathrm{Cxy} = \left(1 - \frac{\mathrm{RMS}_x}{\mathrm{RMS}_y}\right) * 100$$

x = inferior segment y = superior segment

Each coefficient represents the attenuation from a lower to an upper body level. C_{PH} representing the attenuation from the pelvis to the head, C_{PT} representing the attenuation from the pelvis to the trunk, and C_{TH} representing the attenuation from the trunk to the head. A positive coefficient value indicated a reduced acceleration whilst a negative coefficient value indicated a greater acceleration between the two specified segments.

Activation patterns of the trunk and neck muscles

Raw EMG signals were first high-pass filtered at 20 Hz to remove movement artifacts, then full-wave rectified and filtered using a second-order high-pass Butterworth filter with a cut- off frequency of 50 Hz using a custom Matlab script. Figure 6 shows a typical example of the filtered data split into the baseline (defined as the stance, 500ms prior to COP onset), anticipatory phase and execution phase, as described previously. The onset of muscular activity was visually estimated by the same experimenter for all calculations, which has been shown to be reliable to achieve muscle onset (Micera et al., 2001), and was expressed as a percentage from COP onset to the end of the anticipatory phase. The relative amplitude of muscular activity was calculated from the area under the EMG curve of each muscle using a customised Matlab script and further expressed as a percentage normalised to the EMG activity of the execution phase of gait initiation.

Whole body stability during gait initiation

Margin of stability, using the extrapolated centre of mass (exCOM) introduced by Hof et al., (2005), was used to quantify whole body stability in the AP and ML direction. The exCOM concept extends the classical condition for static equilibrium of an inverted pendulum by adding a linear function of the velocity of the COM to COM position. This method describes how close an inverted pendulum is to falling, given the position and velocity of its COM, and the position of the margins of its base of support (BOS). For the calculation of the margin of stability, the positions of the COM and BOS need to be known. COM was recorded as a weighted sum of all body segments using the whole plug-in-gait model while BOS was calculated from the distance between the position of the swing heel marker at heel-contact and the position of the stance heel marker at toe off represented the step length and width, and was representative AP and ML BOS respectively. MOS was taken at heel contact of the swing limb, as it has previously been shown that foot strike was systematically made with the heel (Caderby et al., 2014).

The position of the $exCOM$ was then calculated as follows

$$exCOM = xCOM + \frac{x'COM}{\sqrt{\frac{g}{l}}}$$

With $xCOM$ and $x'COM$ representing the COM position and velocity respectively, g = 9.81m·s^{-1}, the gravitational acceleration, and l corresponding to the limb length, taken from anthropometric measurements prior to data collection (inverted pendulum eigenfrequency). The MOS corresponded to the difference between the AP and ML boundary of support (BOS) and the AP and ML position of the 'extrapolated COM' ($exCOM$) at heel contact and defined as BOS - $exCOM$. The lower the MOS value, the closer the $exCOM$ is to the BOS, indicating reduced whole body stability.

Statistical analysis

All statistical analyses were carried out using IBM SPSS, v19 (IBM Corporation, Armonk, NY, USA). Normality of data was examined and confirmed for all variables using the Shapiro-Wilk test. A series of independent samples t tests were used to test for difference between young and older groups for the AvgSD of angular displacement of each upper body segment, RMS of acceleration at each upper body segment and attenuation of such acceleration and MOS values, with Bonferroni correction for multiple comparisons applied. Finally, for the onset of muscular activity and relative amplitude of muscle activity of the anticipatory phase. Statistical significance was assessed with an alpha level of 0.05. All data are presented as mean ± SD unless otherwise stated.

Results

Variability of angular displacement

During the anticipatory phase, older had a significantly higher AP AvgSD of angular displacement of the head compared to young (3.7 ± 0.84° and 1.5 ± 0.56°, respectively; $p = 0.004$), with no differences in AP AvgSD of angular displacement of the pelvis and trunk between groups. During the execution phase, there were no differences in AP AvgSD of angular displacement of the pelvis, trunk or head between groups (Figure 4). During both the anticipatory phase and execution phase, there were no differences in ML AvgSD of angular displacement of the pelvis, trunk or head between groups (Figure 4).

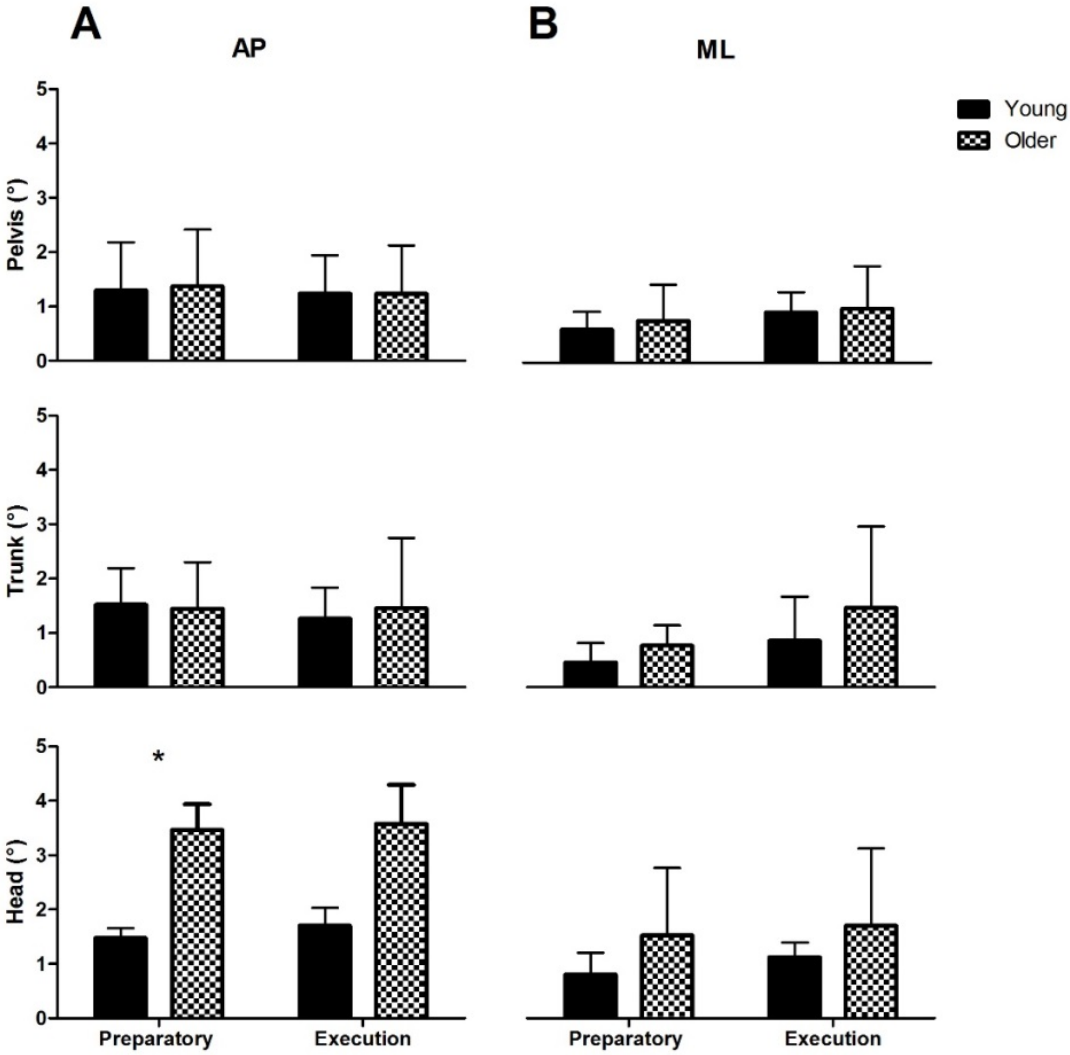

Figure 7. Young and older mean ± SD of variability of the pelvis (top row), trunk (middle row) and head (bottom row) segment angular displacement during anticipatory phase and execution phase in the anterior posterior direction (AP) and mediolateral direction (ML), evaluated by calculation of the average standard deviation (AvgSD). * indicates significance between groups.

PCA of angular displacement is presented in Figure 5 and 6 in the AP and ML direction respectively. In the AP direction, both groups demonstrated a similar amount of variability of pelvis angular displacement as two PCs explained over 90% of the movement pattern variance in both groups. Both groups demonstrated low variability of trunk angular displacement, as only one PC was needed to explain over 90% of the movement pattern variance. Young showed low variability of angular head displacement as only one PC was needed to explain over 90% of variance. Older however, demonstrated high variability in head angular displacement indicated by the requirement of three PCs to explain over 90% of variance (Figure 5).

In the ML direction, young displayed low variability of pelvis angular displacement as one PC was needed to explain over 90% of variance. Older displayed higher variability, requiring two PCs to explain over 90% of variance. Both groups demonstrated similar variability of trunk angular displacement. Both groups displayed high variability of head movement as both required three PCs to explain over 90% of the movement pattern variance (Figure 6).

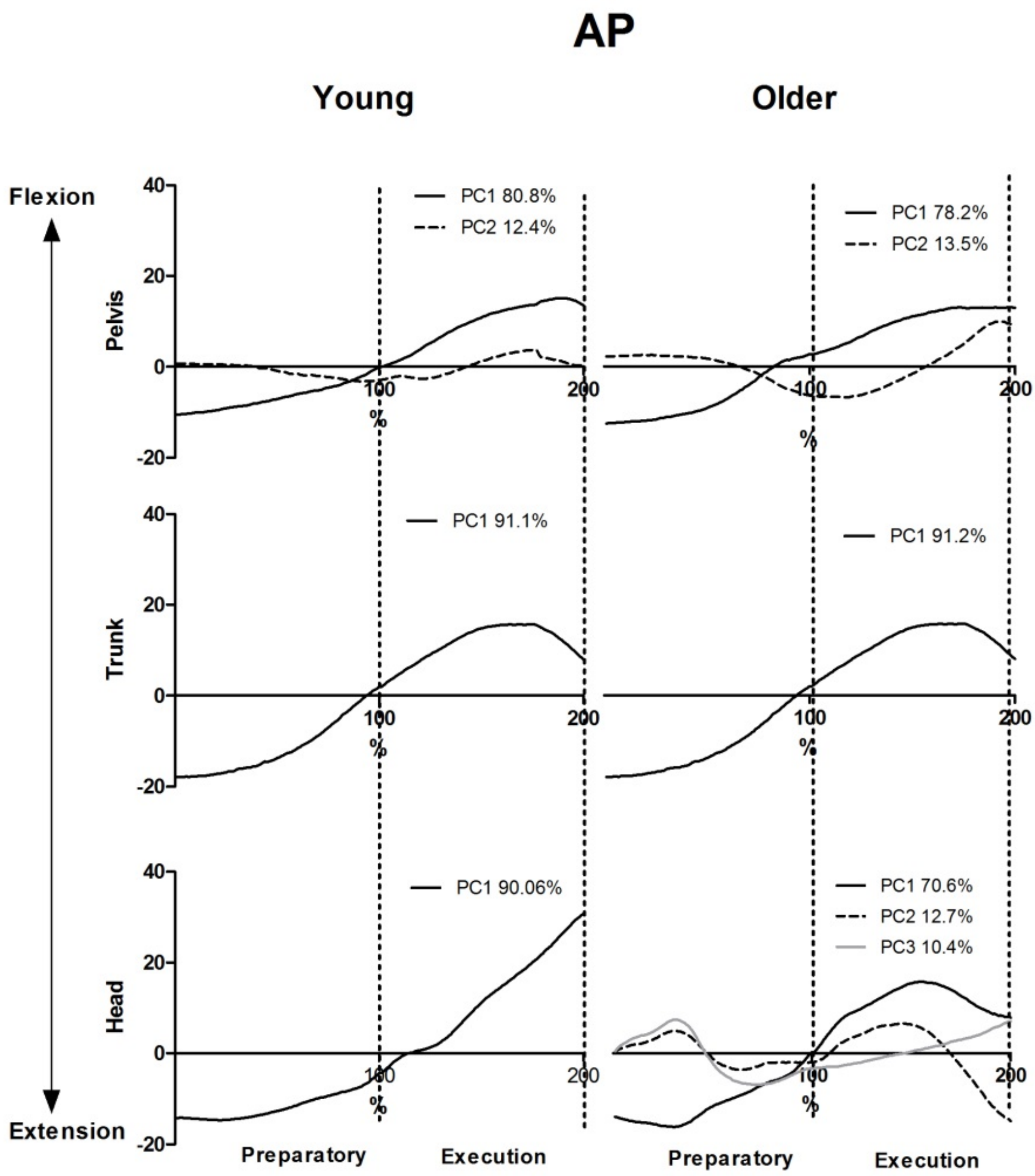

Figure 8. Principal component analysis on the data set of angular displacement of the pelvis, trunk, and head in the anteroposterior (AP) direction during the whole movement of gait initiation. Positive and negative values indicate flexion or extension, respectively (direction is indicated by the arrow to the left of the figures). The axes intersection (0) represents the COP onset, the first perforated line indicates the end of the anticipatory phase while the second perforated line indicates the end of the execution phase. Each line represents one principal component and the percentage of variance accounted for is reported.

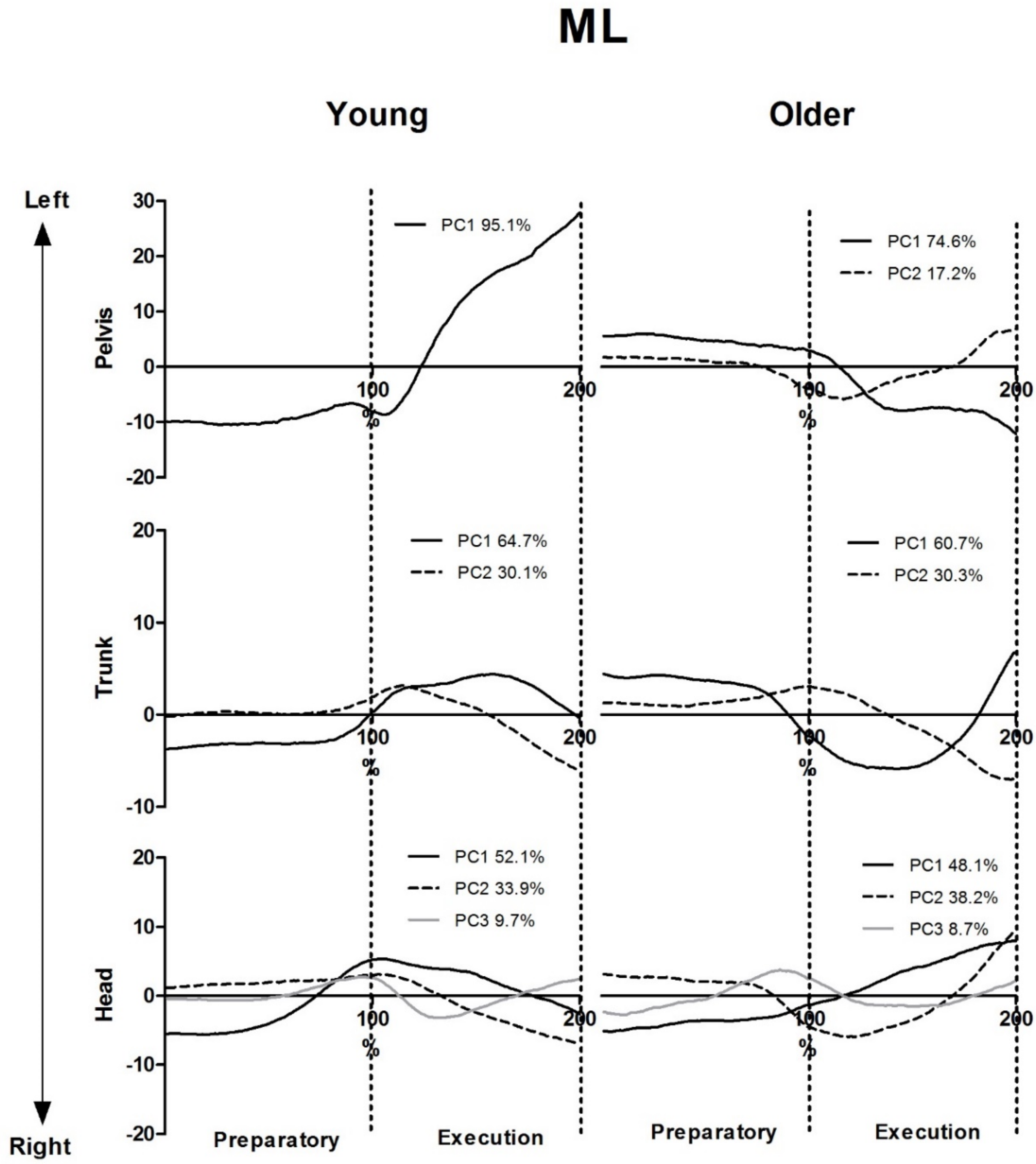

Figure 9. Principal component analysis on the data set of angular displacement of the pelvis, trunk, and head in the mediolateral (ML) direction during the whole movement of gait initiation. Positive and negative values indicate abduction or adduction, respectively (direction is indicated by the arrow to the left of the figures). The axes intersection (0) represents the COP onset, the first perforated line indicates the end of the anticipatory phase while the second perforated line indicates the end of the execution phase. Each line represents one principal component and the percentage of variance accounted for is reported.

Attenuation of upper body accelerations

During the anticipatory and execution phase, young displayed significantly greater AP RMS acceleration for the pelvis, trunk and head compared to older ($p < 0.05$) (Figure 7A and B). During the anticipatory phase, AP C_{TH} was significantly lower in older compared to young (-1.9 ± 20.2% versus 10.1 ± 21.6%, $p = 0.02$, respectively) (Figure 7C). During the execution phase, there were no significant differences in acceleration attenuation between groups (Figure 7D).

During the anticipatory and execution phases, there was no difference in ML RMS acceleration for the pelvis, trunk or head between age groups (Figure 8A and B). During the anticipatory phase, ML accelerations were attenuated for both groups, with the exception of older not able to attenuate C_{PT}, however there were no significant differences between groups (Figure 8C). During the execution phase, both groups did not attenuate ML accelerations, however there were no significant differences between groups (Figure 8D).

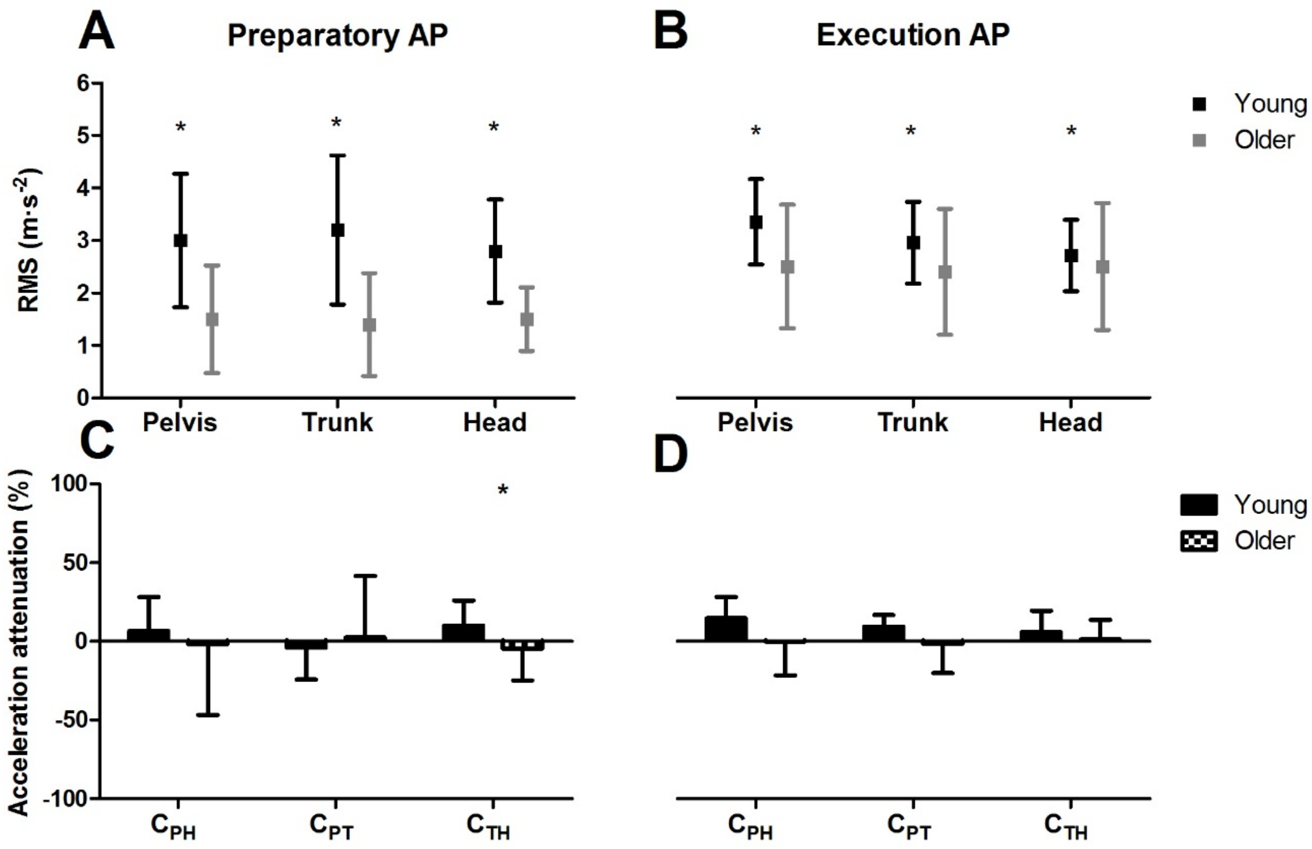

Figure 10. Mean ± SD of the acceleration root mean square (RMS) values at pelvis, trunk and head level (panel A & B) and coefficients of attenuation for pelvis-head (CPH), pelvis-trunk (CPT) and trunk-head (CTH) (panel C & D) for young and older during the anticipatory phase and execution phase in the anteroposterior (AP) direction. * indicates significance between groups

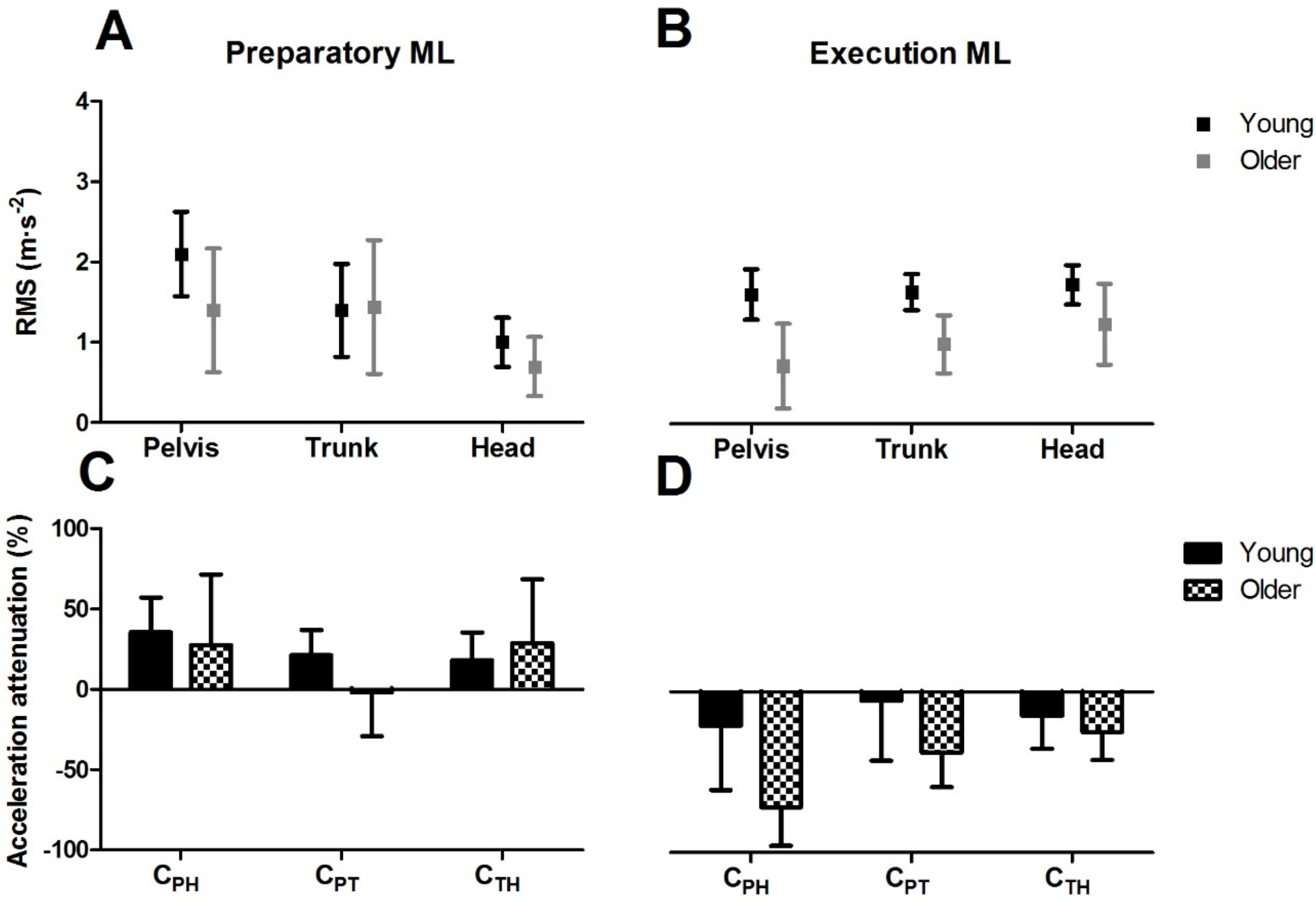

Figure 11. Mean ± SD of the acceleration root mean square (RMS) values at pelvis, trunk and head level (panel A & B) and coefficients of attenuation for pelvis-head (CPH), pelvis-trunk (CPT) and trunk-head (CTH) (panel C & D) for young and older during the anticipatory phase and execution phase in the mediolateral (ML) direction.

Muscle activity

EMG data for young were analysed for all muscles and six out of 12 older was analysed for the neck muscles. It was not possible to gain muscle onset data for the neck extensors for older due to difficulty in identifying an onset point because of noisy signal. Older displayed a significantly delayed muscle activity onset of the SCM compared to young ($p < 0.05$) (Table 2). There were no differences in muscle activity onset time for the ES (T9) or ES (L3) between groups. Both groups activated all muscles to a greater extent in the execution phase compared to the anticipatory phase as shown by less than 100% EMG values. Older had significantly greater relative muscle activity in the anticipatory phase compared to young for the SCM, ES (T9) and ES (L3) ($p < 0.05$) (Table 2).

Whole body stability

There was no difference between groups for AP MOS, however older displayed a significantly lower ML MOS compared to young ($p = 0.035$).

Table 2. The time of the onset of muscle activity given as a percentage of total duration of the anticipatory phase of gait initiation. Amplitude is given as a percentage, normalised to the execution phase of gait initiation. P value ($p < 0.05$) indicates significance between groups.

	Young (n =11)	Older (n = 6)	P-value
SCM			
Onset (%)	20.5 ± 13.2	50.5 ± 15.4	**0.028**
Amplitude (%)	49.3 ± 20.7	88.2 ± 19.8	**0.002**
Extensor			
Onset (%)	54.8 ± 22.1	N/A	N/A
Amplitude (%)	32.3 ± 16.2	59.6 ± 16.1	0.32
Upper spine (T9)			
Onset (%)	42.2 ± 20.5	63.3 ± 24.7	0.182
Amplitude (%)	28.3 ± 22.6	52.9 ± 22.1	**0.005**
Lower spine (L3)			
Onset (%)	53.1 ± 25.6	60.7 ± 22.5	0.192
Amplitude (%)	23.8 ± 11.5	41.5 ± 15.4	**0.025**

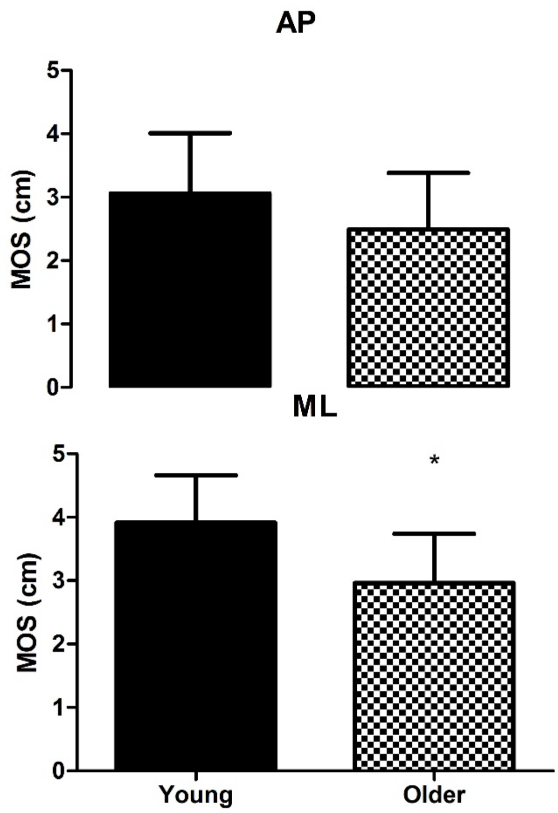

Figure 12. Margin of stability (MOS) at swing heel contact in the anteroposterior (AP) and mediolateral (ML) direction. * indicates significance between young and older.

Discussion

The purpose of the study was to examine any age related change in the neuromechanical strategies underpinning head stabilisation and whole body stability during gait initiation. Older displayed lower AP acceleration of the upper body segments compared to younger and were less able to attenuate AP accelerations between trunk and head compared to young. Older revealed a greater relative magnitude and delayed anticipatory activation of the neck muscles compared to young. Finally, older demonstrated reduced ML whole body stability, while there was no difference between age groups for AP whole body stability. Older participants showed greater variability of head angular displacement in AP direction compared to young participants during both the anticipatory and execution phase of gait initiation, which is in agreement with a previous study by Laudani et al., (2006).

In the present study, young displayed greater AP RMS acceleration at each upper body segment compared to older, indicating older may adopt a more cautious strategy in order to move from a standing posture to forward walking (Menz, Lord, & Fitzpatrick, 2003). No difference between groups existed for ML acceleration attenuation, and similar to previous studies (Kavanagh et al., 2005; Mazzà et al., 2008), both groups found it difficult to attenuate ML accelerations during the execution phase.

Our data are in accordance with previous gait studies demonstrating higher AP RMS of upper body segments in young compared to older during walking (Mazzà et al., 2008). Despite young producing higher AP RMS acceleration of each upper body segment, young were able to attenuate such accelerations from the lower to the upper parts of the upper body segments to a greater extent compared to older. In particular, whilst young were able to attenuate accelerations from trunk to head, aiding protection of the head, older could not, suggesting acceleration did not decrease from the trunk to the head. The inefficiency in attenuating these accelerations may be attributed to deleterious age related changes to passive structures of the spinal column or to sequential activation of the axial musculature (Doherty, 2003) .

From a passive point of view, the age related reduction in acceleration attenuation can be associated with the so called "*en bloc*" movement, related to the documented rigidity of the head-trunk system during gait initiation (Laudani et al., 2006). From a neuromuscular point of view, head stabilisation during dynamic tasks has been thought to be planned early in the central nervous system (CNS), aiming to attenuate postural perturbations of the lower limbs (Pozzo et al., 1990). For example, Ceccato et al.,(2009) observed a 'top down' approach to anticipatory control of the paraspinal muscles (C7 – L3), stabilising the head first, and subsequently lower parts of the upper body during gait initiation. In line with that, the present study reports that the SCM was activated earlier than the trunk muscles in both young and older individuals, suggesting mechanisms of head stabilisation may rely on feed-forward commands from the CNS, a likely mechanism employed to maintain stability of the visual field and offer protection to the head. Nonetheless, the larger relative EMG amplitude demonstrated by older during the anticipatory phase of gait initiation compared to the young is indicative of a greater effort required to stabilise the neck during the controlled fall preceding the first step execution (Winter, 1995) and may explicate the inability to attenuate accelerations from the trunk to the head.

Instability during walking in older populations is commonly considered in the ML plane, while loss of ML stability can have a profound effect on walking function (Maki, 1997b). Interestingly, differences in upper body stabilisation between young and older were only observed in the AP direction during the present thesis. Even though differences in upper body stabilisation were apparent between age groups, there were no differences in AP MOS between groups. A possible explanation is that upper body differences were not considerable enough to alter AP whole body stability. AP MOS has previously been described as similar between young and older females during steady state walking (McCrum et al., 2016). Despite no differences between groups in the ML direction of upper body variability or attenuation of acceleration, older demonstrated significantly reduced MOS, indicating reduced ML whole body stability. This may have implications for fall risk as whole body stability is an indicator of fall risk (Lockhart & Liu, 2008; Toebes, Hoozemans, Furrer, Dekker, & van Dieën, 2012). Caderby et al., (2014) observed that young were able to maintain ML whole body stability during gait initiation, while ML whole body stability in older during gait initiation warrants further research to generate an understanding of why ML whole body stability declines during gait initiation in older females.

Conclusion

This study demonstrated that the ability to stabilise head movements in the AP direction during gait initiation is compromised in older women. Decreased head stability in older women was attributed to an impaired ability to attenuate accelerations from the trunk to the head along with altered muscular activation of the trunk and neck muscles. On the other hand, there was a discrepancy between head stabilisation and whole body stability in the AP and ML direction, meriting further investigation.

CHAPTER 7

Effects of free head movement on head stability and whole body stability during gait initiation in young and older women

A version of the work from this chapter has been submitted to Gait and Posture (Appendix 10)

Abstract

Head flexion is typically implemented by older adults, however, how this may affect head stability and whole body stability during gait initiation is unknown. The present study examined the effect of free head movement during gait initiation on upper body, lower body and whole body stability of older females. Eleven young (23 ± 1 years) and 12 older (74 ± 2 years) healthy females initiated gait under two conditions, free head movement gait initiation(no visual target (NVT)) and an instructed gait initiation (head focusing on a visual target) (VT). A stereophotogrammetric system was used to assess head stability, whole body stability and gait spatiotemporal parameters in the anteroposterior and mediolateral directions. Amplitude and latency of muscle activation in the sternocleidomastoid, neck extensor, and upper and lower trunk muscles were determined by surface electromyography. Young demonstrated no differences in head movement, head stability or whole body stability between conditions, but had greater AP step velocity and step length in VT. Older demonstrated greater AP angular displacement during NVT, but no differences in head stability or whole body stability between conditions. Older implemented anticipatory muscle activation of an earlier onset and greater amplitude of neck muscles in order to achieve head stability in VT. Older produced greater ML step velocity and step width in NVT compared to VT. Results show that older implement greater head flexion during gait initiation, however it does not affect head stability or whole body stability allowing for a naturally adopted head position during gait assessments.

Introduction

The ageing process can make walking a challenging process, due to reduced neuromuscular performance in several aspects e.g. strength, balance, functional ability (Boyer et al., 2017). Indeed, gait and gait disorders are an important contributor to falls in older adults (Rubenstein, 2006). As a result, several studies have investigated gait mechanics in older adults and the changes that occur due to ageing (Boyer et al., 2017; Aboutorabi et al., 2016, Prince et al., 1997). The vast majority of these studies, however, focus on lower limbs and the effects of ageing on e.g. step width, step length, or walking speed, with little attention given to upper body movements.

Head flexion, for example, is a movement frequently implemented by older adults during gait to identify lower limb trajectory to enable better footfall vision or to gather more information when walking towards an obstacle (Muir, Haddad, Heijnen, & Rietdyk, 2015). Walking results in a continuous perturbation in the balance equilibrium, as the centre of mass (COM) alters in relation to the also changing base of support (BOS) (Woollacott & Tang, 1997), this head flexion may exacerbate the forward shifting of the COM closer to the BOS, further threatening stability.

The effect head flexion has on postural balance has been examined in a few studies, with varied results. During quiet standing, older adults were instructed to flex the head while sway was recorded. Head flexion, in comparison to head in upright position (achieved by fixating on a stationary target at eye level) was shown to have destabilising effects as evidenced by an increased postural sway (Buckley et al., 2005). On the contrary, during steady state gait, walking with the head free to move or the head in an upright position, focused on a visual target, did not show any differences, suggesting that head movement in a dynamic situation did not have an effect (Thomas et al., 2017). Head stability is thought to be critical in the control of whole body stability (Pozzo et al., 1990), however, how head movement relates to the mechanisms underpinning head stability warrants further understanding. While the effects of head movement and focusing on a stationary target on stability measures have received some attention in static and dynamic conditions in older adults, little is known about the effect of head movement and head stability has on whole body stability during the transition from a standing position to walking (gait initiation).

Therefore, the aim of the present study was to examine the effect of free head movement during gait initiation on upper body, lower body and whole body stability parameters of older females. To achieve this, there was two conditions; gait initiation with free head movement and an instructed gait initiation (head focusing on a visual target). It was hypothesised that participants would have better head stabilisation with a visual target (VT) compared the free head movement condition with no visual target (NVT). Females were the focus of the study as it has been reported that their whole body stability declines to a greater extent than males (Wolfson et al., 1994) and they tend to fall more often (Schultz et al., 1997).

Methods

Participants

Eleven healthy young (age: 23.1 ± 1.1 years, height: 1.64 ± 0.71 m, body mass: 57.5 ± 6.7 kg) and 12 healthy older (age: 73.9 ± 2.4 years, height: 1.63 ± 0.45 m, body mass: 66.2 ± 10.2 kg) females volunteered to participate in the study. Older adults were considered ‘medically stable’ to participate in the study, according to exclusion criteria for older people in exercise studies of Greig et al., (1994). In addition to these criteria, no participants had any history of neurological disorders that would affect their balance or gait ability, and were able to complete the task without the use of bifocal or multifocal spectacles. Written informed consent was provided by all participants and ethical approval was given by the institution's ethics committee.

Experimental Protocol

Participants wore their everyday flat shoes. Instructions were to stand as still as possible with their feet in a comfortable position at shoulder width apart, and with the arms alongside the trunk. Participants were verbally instructed to start walking on their own accord from a single force platform (Bertec Corp, Worthington, OH) and to continue to walk forwards in a straight line for at least three steps at their comfortable walking speed. The right leg was used as the starting (swing) leg for all trials. Starting feet position at shoulder width apart was marked on the force platform and participants repositioned themselves in that position for each trial.

Trials were performed under a visual target (VT) and free head movement (NVT) condition. In the VT condition, participants were instructed to focus on a fixed target at eye level, 5 m ahead of the starting position. The position, size and distance of the visual target were decided following pilot testing, which allowed a target which could be comfortably seen by all participants. No participant wore glasses, but they all confirmed they were able to clearly see and focus on the visual target. In the NVT condition, no instructions were given as to where to focus gaze and the visual target was removed. Trial order was randomised and counterbalanced for each participant whilst familiarisation was given for each target condition. In total, 10 trials were completed and analysed (five trials per target condition). Testing took place on a single occasion.

A seven camera motion analysis system (VICON, Oxford Metrics, London, England) was used to record and reconstruct the 3D position of 35 reflective markers placed on body landmarks, following the Davis protocol (Davis et al., 1991) with a sampling rate of 100 Hz. The VICON whole body plug-in-gait model was used to define a local anatomical reference frame for the pelvis (markers on the left and right anterior and posterior superior iliac spines), trunk (markers located at the clavicle and sternum level as well as at C7 and at T10), and head (four markers, placed on the left and right side of the front and back of the head) and then calculating the relevant kinematic data. The force platform was used to track Centre of Pressure (COP) motion with a sampling frequency of 1000 Hz and then resampled.

Temporal aspects of gait initiation were determined relative to COP onset from the force platform. The onset of COP displacement was automatically estimated as the time point at which the AP component of the ground reaction force overcame the threshold defined as 3 standard deviations of its peak-to-peak value during static posture AP force. Gait initiation was performed as a whole movement but divided into two phases for analysis*:* *1) anticipatory phase*, which lasted from the onset of COP motion to the instant of toe off of the swing limb 2)*: execution phase*, which lasted from the instant of toe off of the swing limb to the instant of toe off of the stance leg. Temporal events of gait initiation were obtained from both position and velocity curves derived from markers placed on the calcaneus and fifth metatarsal bones (Mickelborough et al., 2000). Angular displacement of the upper body segments (pelvis, trunk, and head) were measured in the AP and ML direction. Additionally, whole body COM in the AP and ML direction was estimated using the mentioned plug-in-gait model.

Muscle activity was determined by surface EMG recordings (BTS Bioengineering, Italy). EMG signals were collected bilaterally using bipolar disposable electrodes (1 cm disc-electrodes, 2 cm inter-electrode distance) from the sternocleidomastoid (SCM), neck extensors (NE), and erector spinae (ES) at the level of T9 and L3, with a sampling frequency of 1000 Hz. Electrode sites were prepared by gently abrading the skin to ensure good contact. For the SCM, electrodes were positioned at 1/3 of the distance from the sternal notch to the mastoid process at the distal end overlying the muscle belly; for the NE, electrodes were positioned over the distal half of the distance between the base of the occiput and the spinous process of the seventh cervical vertebrae (Falla et al., 2004) whilst for the ES, electrodes were placed 2cm lateral of the spinal process at T9 and L3.

Gait spatiotemporal parameters

Gait initiation velocity was calculated as single derivative of the above mentioned COM which was recorded as a weighted sum of all body segments using the whole plug-in-gait model. Step length and step width was calculated from the distance between the position of the swing heel marker at heel-contact and the position of the stance heel marker at toe off in the AP and ML direction respectively.

Angular displacement and variability

Absolute angular displacement of the pelvis, trunk, and head was filtered using a second-order low-pass Butterworth filter with a cut-off frequency of 5 Hz and re-scaled to the first value of the anticipatory phase. Absolute angular displacement was measured across the whole movement of gait initiation. To quantify the variability of the pelvis, trunk, and head angular displacement of the anticipatory and execution phase, the average standard deviation (AvgSD) was calculated using the following equation:

$$AvgSD = \sqrt{\frac{\sum \theta^2}{100}}$$

θ = Angular displacement of the segment.

This measure has previously been used to assess the stability of individual body segments, with decreased variability indicating increased segment stability (Laudani et al., 2006).

Attenuation of upper body accelerations

Acceleration of the pelvis, trunk and head segments was calculated by double derivative of the 3D position of the origin of each upper body segment reference frame in the AP, ML and cranio-caudal (CC) direction. It was computed by a customised Matlab 7.5 script (Mathworks, Inc, USA) and filtered using a second-order low-pass Butterworth filter with a cut- off frequency of 5Hz. The magnitude of acceleration of each segment was calculated using the root mean square (RMS) in the AP, ML and CC direction. RMS acceleration values are known to be influenced by gait velocity (Kavanagh & Menz, 2008), thus AP and ML RMS acceleration were normalized by CC acceleration RMS as proposed by Iosa et al., (2012) (Iosa et al., 2012). The ability to attenuate accelerations through the upper body segments was quantified using the attenuation coefficient, expressed as a percentage, describing the ability to reduce accelerations from inferior to superior segments. A positive value of this index was shown as an indicator of upper body stability (Summa et al., 2016). The attenuation coefficients were calculated using RMS values of each segment as follows (for both AP and ML direction):

$$\mathrm{Cxy} \quad = \left(1 - \frac{\mathrm{RMS}_x}{\mathrm{RMS}_y}\right) * 100$$

x = inferior segment y = superior segment

Each coefficient represents the attenuation from a lower to an upper body level. So that, C_{PH} represents the attenuation from the pelvis to the head, C_{PT} the attenuation from the pelvis to the trunk, and C_{TH} the attenuation from the trunk to the head. A positive coefficient value indicated an acceleration reduction whilst a negative coefficient value indicated an amplification of the acceleration between the two specified segments.

Muscle activity

Raw EMG signals were first high-pass filtered at 20 Hz to remove movement artifacts, then full-wave rectified and filtered using a second-order high-pass Butterworth filter with a cut- off frequency of 50 Hz using a custom Matlab script. The onset of muscular activity was visually estimated from COP onset by the same experimenter for all calculations, which has been shown to be reliable to achieve muscle onset in clinical applications (Micera et al., 2001). Amplitude of muscular activity was calculated from the area under the EMG curve of each muscle using a customised Matlab script.

Whole body stability

Margin of stability, using the extrapolated centre of mass (exCOM) introduced by Hof et al (2005) (A L Hof et al., 2005), was used to quantify whole body stability in the AP and ML direction. The exCOM concept extends the classical condition for static equilibrium of an inverted pendulum by adding a linear function of the velocity of the COM to COM position. This method describes how close an inverted pendulum is to falling, given the position and velocity of its COM, and the position of the margins of its base of support (BOS). For the calculation of the margin of stability, the position of the BOS needs to be known. This was calculated as the distance between the position of the swing heel marker at heel-contact and the position of the stance heel marker at toe off represented the step length and width, and was representative AP and ML BOS respectively. MOS was taken at heel contact of the swing limb, as it has previously been shown that foot strike was systematically made with the heel (Caderby et al., 2014).

The position of the $exCOM$ was then calculated as follows:

$$exCOM = xCOM + \frac{x'COM}{\sqrt{\frac{g}{l}}}$$

With $xCOM$ and $x'COM$ representing the COM position and velocity respectively, g = 9.81m.s^{-1}, the gravitational acceleration, and l corresponding to the limb length, taken from anthropometric measurements prior to data collection. The MOS corresponded to the difference between the AP and ML boundary of support (BOS) and the AP and ML position of the 'extrapolated COM' ($exCOM$) at heel contact and defined as BOS - $exCOM$. The lower the MOS value, the closer the $exCOM$ is to the BOS, indicating reduced whole body stability.

Statistical analysis

Statistical analyses were conducted using IBM SPSS, v22.0 (IBM Corporation, Armonk, NY, USA). Normality of data distribution was examined and confirmed for all variables using the Shapiro-Wilk test ($p > 0.05$). In order to examine for any changes due to the imposed condition, paired samples *t*-tests were conducted on each variable, for both age groups. Independent samples t-tests were used to examine between–group comparisons. Statistical significance was set at an alpha level of 0.05. All data is presented as mean ± SD.

Results

Gait spatiotemporal parameters

Results from gait parameters are displayed in Table 3. Young displayed higher AP step velocity and step length during VT compared to NVT ($p < 0.05$) whereas older showed no difference in gait velocity or step length between target conditions. Older displayed lower ML step velocity and step width during VT compared to NVT ($p < 0.05$).

Table 3. Gait spatiotemporal parameters, angular displacement and margin of stability. * indicates significant differences between target conditions, ≠ Indicates significant differences between groups

	Young		Older	
	Visual target	No visual target	Visual target	No visual target
AP COM velocity ($m \cdot s^{-1}$)	1.01 ± 0.18*	1.13 ± 0.12	0.89 ± 0.05	0.86 ± 0.24
ML COM velocity ($m \cdot s^{-1}$)	0.18 ± 0.05*	0.11 ± 0.03	0.08 ± 0.02*	0.10 ±0.03
Step length (m)	0.68 ± 0.11*	0.78 ± 0.16	0.58 ± 0.17	0.59 ± 0.22
Step width (m)	0.21 ± 0.03*	0.16 ± 0.03	0.19 ± 0.06*	0.23 ± 0.08
AP Angular head displacement (°)	4.8 ± 2.01	5.6 ± 1.52≠	4.9 ± 3.39*	9.73 ± 3.21
ML Angular head displacement (°)	3.4 ± 1.2	2.3 ± 0.091	2.8 ± 1.2	3.43 ± 1.9
AP Angular trunk displacement	9.43 ± 1.44	11.19 ± 3.54	6.27 ± 3.77	5.97 ± 2.31
ML Angular head displacement (°)	4.23 ± 2.96	4.78 ± 2.11	5.12 ± 2.66	4.54 ± 1.98
AP Margin of stability (cm)	2.98 ± 1.1	3.06 ± 0.95	2.49 ± 0.89	2.13 ± 0.54
ML Margin of stability (cm)	3.75 ± 0.54≠	3.91 ± 0.75≠	2.96 ± 0 .48	3.01 ± 0.96

Angular displacement and variability

Results from angular displacement are displayed in Table 1. Older produced lower AP head displacement during VT compared to NVT ($p < 0.05$), whilst young showed no difference in AP angular head displacement between target conditions. Neither group displayed any difference in ML head displacement between target conditions. Young produced higher AP angular trunk displacement during VT compared to NVT whilst older showed no differences in AP angular trunk displacement ($p < 0.05$). Older produced greater AP head angular displacement in VT compared to young ($p < 0.05$) but no difference was shown during NVT between groups Neither group displayed any difference in variability of angular head displacement evidenced by no difference in the AvgSD of AP or ML angular displacement of upper body segments during the anticipatory phase or the execution phase of gait initiation between VT and NVT (Figure 11).

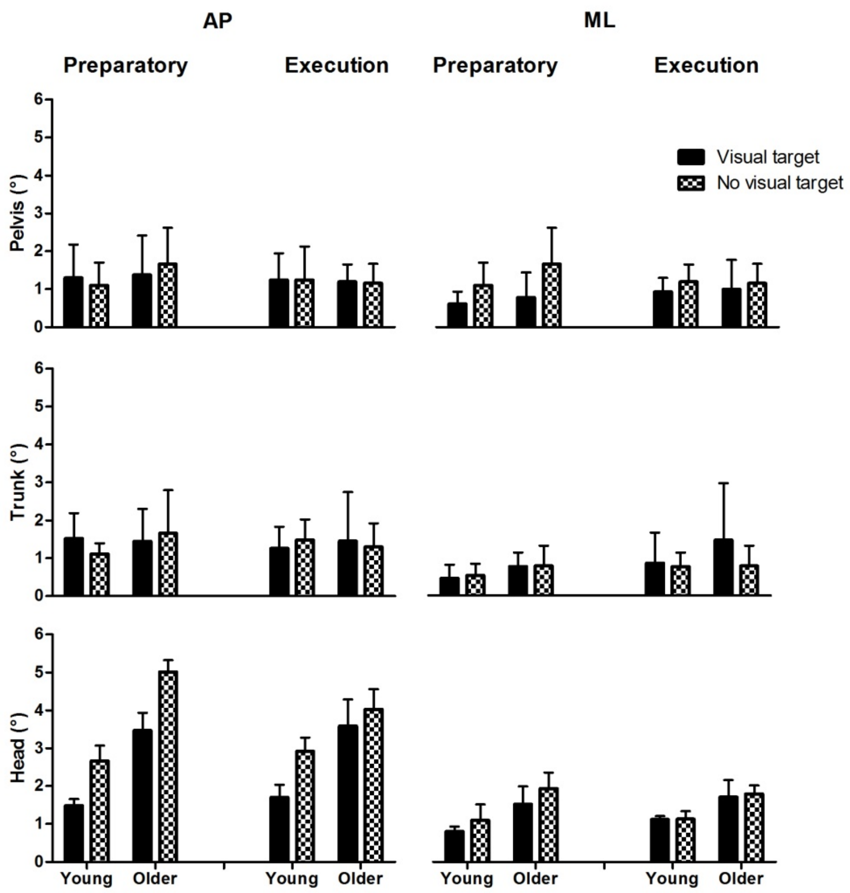

Figure 13. Young and older mean ± SD of variability of the pelvis (top row), trunk (middle row) and head (bottom row) segment angular displacement during anticipatory phase and execution phase during the visual target and no visual target condition in the anterior posterior direction (AP) and mediolateral direction (ML), evaluated by calculation of the average standard deviation (AvgSD).

Attenuation of upper body accelerations

Neither young nor older demonstrated any differences in acceleration attenuation of the upper body segments between VT and NVT, for either phase (Figure 12).

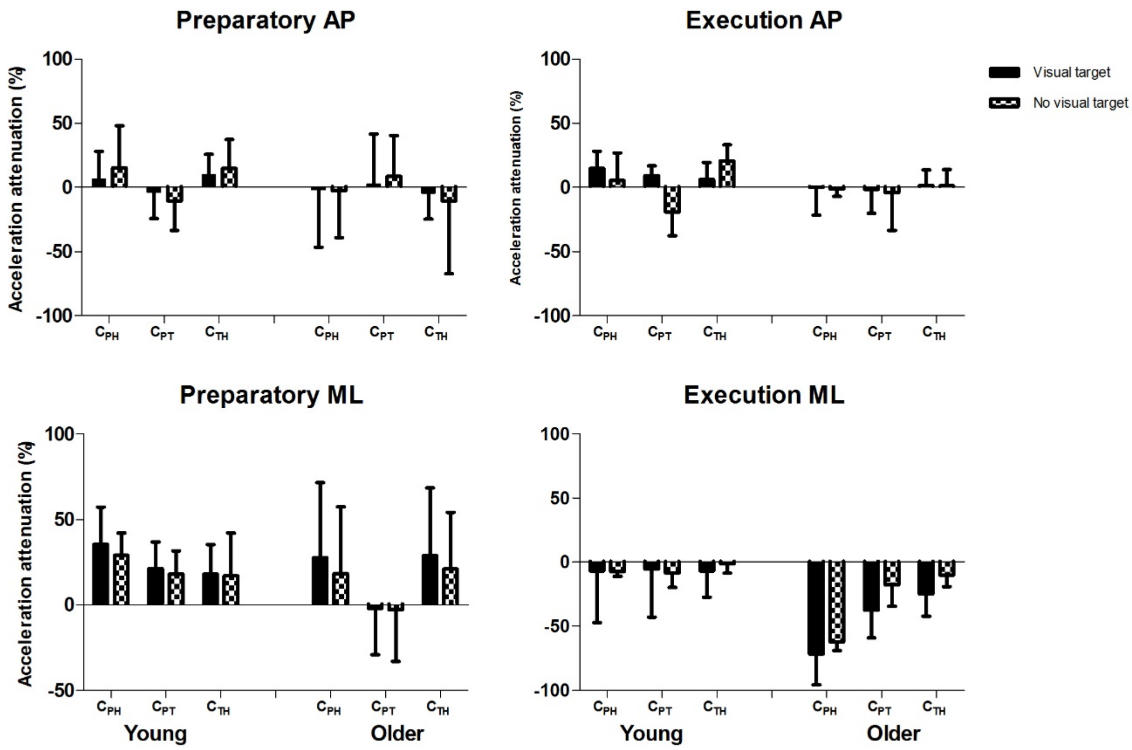

Figure 14. Mean ± SD of the coefficients of attenuation for pelvis-head (CPH), pelvis-trunk (CPT) and trunk-head (CTH) (C&D) for young and older during the anticipatory phase and execution phase in the anteroposterior (AP) and mediolateral (ML) direction. * indicates significance between target conditions

Muscle activity

EMG data for young were analysed for all muscles and six out of 12 older were analysed for the neck muscles. It was not possible to gain muscle onset data for the neck extensors for older due to difficulty in identifying an onset point because of noisy signal (Table 4 and 5). Young displayed no effect of visual target on muscle activation latency or amplitude of the SCM and, while the NE exhibited an earlier muscle onset time during VT compared to NVT ($p < 0.05$). Older displayed an earlier onset of the SCM, and lower amplitude of the SCM and NE during the anticipatory phase during VT compared to NVT ($p < 0.05$). Older displayed no differences between VT and NVT in muscle onset time or amplitude of the upper and lower spine.

Table 4. Mean ± SD for the time of the onset of muscle activity given as raw values of muscle onset time and amplitude of the anticipatory and execution phase for older respectively. * indicates significant differences between target conditions.

Young	Visual target		No visual target	
	Preparatory	Execution	Preparatory	Execution
Flexor				
Onset (ms)	118.77 ± 76.9	-	184.5 ± 66.9	-
Amplitude	7.6 ± 4.9	15.6 ± 14.1	10.1 ± 16.1	18.9 ± 20.1
Extensor				
Onset (ms)	317.5 ± 145.8*	-	369.1 ± 165.7	-
Amplitude	3.9 ± 2.9	12.07 ± 7.2	5.9 ± 7.9	15.1 ± 15.2
Upper Spine (T9)				
Onset (ms)	244.5 ± 99.5	-	233.2 ± 116.5	-
Amplitude	7.6 ± 6.1	26.8 ± 20.1	9.1 ± 8.2	38.2 ± 15.3
Lower Spine (L3)				
Onset (ms)	137.9 ± 76.3	-	210.0 ± 98.4	-
Amplitude	7.4 ± 5.9	31.1 ± 20.9	9.5 ± 7.1	30.1 ± 11.6

Table 5. Mean ± SD for the time of the onset of muscle activity given as raw values of muscle onset time and amplitude of the anticipatory and execution phase for older respectively. * indicates significant differences between target conditions.

Older	Visual target		No visual target	
	Preparatory	Execution	Preparatory	Execution
Flexor				
Onset (ms)	233.9 ± 78.3*	-	291.4 ± 112.7	-
Amplitude	5.3 ± 3.6 *	6.02 ± 3.2	4.2 ± 3.4	5.8 ± 3.0
Extensor				
Onset (ms)	-	-	-	-
Amplitude	6.1 ± 5.3 *	10.1 ± 5.8	3.8 ± 2.9	8.9 ± 4.3
Upper Spine (T9)				
Onset (ms)	293.1 ± 100.2	-	183.7 ± 87.9	-
Amplitude	7.9 ± 4.1	14.9 ± 9.2	10.0 ± 4.7	16.7 ± 8.1
Lower Spine (L3)				
Onset (ms)	281.1 ± 98.6	-	252.4 ± 124.5	-
Amplitude	7.3 ± 5.0	17.6 ± 10.3	6.3 ± 3.8	13.5 ± 7.2

Whole body stability

Neither group displayed any difference in AP or ML MOS between VT and NVT.

Discussion

Older adults implemented greater head flexion, evidenced by greater AP displacement with free head movements in the no visual target condition (NVT) compared to focussing on a visual target (VT). This increased head flexion did not impair head stability or whole body stability during gait initiation, however, the use of a fixed visual target did reduce head flexion and alter gait spatiotemporal parameters in older.

The increased head flexion implemented by older during NVT condition confirms previous studies that older adults implement head flexion during steady state gait with free head movements (Marigold & Patla, 2008; Muir et al., 2015). Increased head movement may lead to head instability and head instability during gait initiation in older has been attributed to increased movement variability (Laudani et al., 2006), reportedly due to the inability to attenuate accelerations from the trunk to the head (Mazzà et al., 2008) along with altered muscular activation of the trunk and neck muscles. However, it was found that such head flexion did not affect head stability evidenced by no difference in AvgSD values.

Although there were no differences in the ability to attenuate acceleration between conditions, older did alter anticipatory muscular activation in order to achieve head stability during VT. This was seen through an earlier onset of SCM and increased activation amplitude of SCM and NE during VT. These findings suggest that mechanisms of head stability may rely on feedforward commands from the CNS (Prince, Winter, Stergiou, & Walt, 1994) and a larger EMG amplitude may be indicative of a greater effort required to stabilise the neck during the controlled fall preceding the first step execution (Winter, 1995). The fact that older implemented head flexion when there was no instructions on where to look, the need for anticipatory adjustments was less, and therefore older produced less muscle activation in anticipation of head flexion to take place. However, when a target was introduced, that became unnatural for older and therefore to prepare for the movement, feedforward mechanisms was implemented to ensure the head was upright, achieved with earlier and higher activation of the neck muscles.

Similar to the findings for head stabilisation, neither group demonstrated any differences in whole body stability evidenced by no differences in MOS values between conditions. These results oppose findings in static conditions reporting that head flexion induces increased postural sway and in turn decreases stability (Buckley et al., 2005). On the contrary, the present study agrees with previous reports that during steady state gait, walking with the head free to move compared to focussing on a visual target, does not show any differences in stability whole body stability measures (Thomas et al., 2017). These results, taken together suggest that the influence of target fixation on stability may be task-dependent, that free head movements during static conditions may be destabilising, but not during transitory or steady state gait tasks.

Interestingly, despite no difference in whole body stability between target conditions, gait spatiotemporal parameters did change. During VT, young produced higher AP velocity and step length when compared to NVT, supporting previous research which observed walking with a target increased gait velocity (Cole, Riccio, & Balcetis, 2014). Older displayed greater step width, a commonly accepted measure of gait stability

(Bruijn et al., 2013), during NVT. Fixating on a visual target is known to decrease postural sway during static conditions compared to no visual target (Neil Marshall Thomas et al., 2016). Accepting that participants in the present study fixated on the target during VT, and postural sway was greater during NVT during the stance period before the anticipatory phase, this may explain the greater step width employed by both groups during NVT. This may have been a compensatory mechanism to control for instability during the static phase and lack of anticipatory adjustments which is commonly seen to be less effective in older (Khanmohammadi et al., 2015), however this is speculative and requires further research. Moreover, differences in step length and step width, would be expected to result in differences in MOS values, as two of the variables used to calculate MOS (please see Methods). However, as step width increased (increased stability), older increased ML COM velocity (decreased stability), which likely negated the effect of the former.

The fact that gait parameters did alter between conditions, raises methodological concerns about using a visual target when studying gait as typically gait and gait initiation studies use a visual target (Caderby et al., 2013; Hirasaki et al., 1993; Laudani et al., 2006). Gait parameters in gait initiation may not be ecologically transferable to conditions where a visual target is not utilised, and consequently may underestimate the true challenge walking poses on older individuals, potentially reaching to erroneous results and less specific intervention advice. Therefore, future gait studies should consider using a natural walking approach, without a visual target fixation, to achieve a more realistic understanding of the effect of a given intervention.

Conclusion

The present study demonstrated that older adults implement greater head flexion in free head movement conditions with an altered muscle activation pattern to maintain head stabilisation when focussing on a visual target. Whole body stability not alter between target conditions, therefore these results may inform future gait studies for older adults to allow for a naturally adopted head position opposed to implementing a visual target approach during gait initiation. In addition, future studies assessing gait ability should exert caution when interpreting results using a visual target as gait parameters may be underestimated.

CHAPTER 8

Head stability and whole body stability are reduced when initiating gait at fast speeds in older females

A version of the work from this chapter is under review for publication in The Journal of Aging and Physical Activity (Appendix 11)

Abstract

The study investigated the mechanisms underpinning head stability in young and older individuals when initiating gait at different speeds. Eleven young and 12 older healthy females initiated gait at two self-selected walking speeds. A stereophotogrammetric system assessed variability of angular displacement, acceleration attenuation of the upper body segments and whole body stability. Amplitude and onset latency of muscle activation of trunk and neck muscles was assessed using surface electromyography. Older demonstrated higher head angular displacement variability, decreased ability to attenuate accelerations from the trunk to the head, and decreased whole body stability during fast initiation ($p < 0.05$). Both age groups exhibited similar EMG patterns with an earlier onset and greater amplitude of muscle activation during fast initiation speed compared to comfortable ($p < 0.05$). It is recommended that a structured approach to walking as a physical activity is employed, to ensure that older adults are able to perform brisk walking without an increased fall risk.

Introduction

The decline of gait speed with advancing age is a common feature of the ageing process (Prince et al., 1997). Gait speed is commonly used as an outcome measure of gait ability (Montero-Odasso et al., 2004) and functional status (van Kan et al., 2009) in older adults, therefore gait studies typically investigate comfortable gait speed only. However, older adults also walk faster than their comfortable speed during everyday life, for example walking faster due to being late for an appointment, highlighting the need to investigate different walking speeds.

Stabilisation of the head is fundamental to optimise inputs from the visual, vestibular, and somatosensory systems and, therefore, to maintain whole body balance during locomotion (Kavanagh, et al., 1990), pertinent to reduce fall risk in the older population. There are certain challenges associated with steady state gait at speeds faster than comfortable, as fast walking amplifies accelerations of the upper body segments, resulting in increased demands for control of the neck segment to stabilize the head (Kavanagh et al., 2006). Age related deterioration of motor control and muscle function, however, makes this additional control demand difficult for older adults (Doherty, 2003). Consequently, attenuating accelerations at head level becomes challenging (Mazzà et al., 2008), resulting in a reduced ability to maintain head stability.

While it is known that head stability is threatened at fast steady state walking, such studies fail to address what happens during transitory locomotive tasks such as gait initiation. Gait initiation is challenging due to the transient nature from static to dynamic situation, overcoming inertia, and consequently resulting in the requirement to attenuate high accelerations of the body. Indeed, decreased head stability has been reported in older individuals when initiating gait at comfortable speed (Laudani et al., 2006). As accelerations are likely to be exacerbated when initiating gait at fast speeds, in turn further challenging head stability, it is important to examine any changes in the mechanisms contributing to head stability and their effect on whole body stability of older adults. With the majority of falls happening during gait and gait related tasks in older adults (Rubenstein, 2006), examining the effect initiating gait at fast speeds may have on head stability and overall whole body stability, and whether it could contribute to increased fall risk, is warranted.

The present study investigated whether the mechanisms underpinning head stability and whole body stability in young and older individuals alter when initiating gait at different speeds. It was hypothesised that older participants would have difficulty in attenuating associated higher accelerations when initiating gait at faster speeds, reducing head stability and consequently affecting whole body stability. Females were the focus of the study as it has been reported that their whole body stability declines more than in men (Wolfson et al., 1994) and they tend to fall more often (Schultz et al., 1997).

Methods

Participants

Eleven healthy young (age: 23.1 ± 1.1 years, height: 1.64 ± 0.71 m, body mass: 57.5 ± 6.7 kg) and 12 healthy older (age: 73.9 ± 2.4 years, height: 1.63 ± 0.45 m, body mass: 66.2 ± 10.2 kg) females volunteered to participate in the study. Older participants were considered 'medically stable' to participate in the study, according to exclusion criteria for older people in exercise studies (Greig et al., 1994). In addition to these criteria, no participants had any history of neurological disorders that would affect their balance or gait ability, and were able to complete the task without the use of bifocal or multifocal spectacles. Written informed consent was provided by all participants and ethical approval was given by the institution's ethics committee.

Experimental protocol and equipment

Participants wore their everyday flat shoes. Instructions were to stand as still as possible with their feet in a comfortable position at shoulder width apart, and with the arms alongside the trunk. Participants were verbally instructed to start walking on their own accord from a single force platform (Bertec Corp, Worthington, OH) and to continue to walk forwards in a straight line for at least three steps. The right leg was used as the starting (swing) leg for all trials. Starting feet position at shoulder width apart was marked on the force platform and participants repositioned themselves in that position for each trial.

All trials were performed under two initiation speed conditions: comfortable and fast. Speed conditions were related to everyday activities as follows: comfortable was described as 'think to start a normal walk' and fast as 'think to start walking as fast as you can without running'. The order of trials was randomised for each participant while familiarisation was given for each speed. In total, 10 trials were completed and analysed (five trials per speed condition).

A seven camera motion analysis system (VICON, Oxford Metrics, London, England) was used to record and reconstruct the 3D position of 35 reflective markers placed on body landmarks, following the Davis protocol (Davis et al., 1991) with a sampling rate of 100 Hz. The VICON whole body plug-in-gait model was used to define a local anatomical reference frame for the pelvis (markers on the left and right anterior and posterior superior iliac spines), trunk (markers located at the clavicle and sternum level as well as at C7 and at T10), and head (four markers, placed on the left and right side of the front and back of the head) and then calculating the relevant kinematic data. The force platform was used to track COP motion with a sampling frequency of 1000 Hz and then resampled.

Temporal aspects of gait initiation were determined relative to COP onset. The onset of COP displacement was automatically estimated as the time point at which the AP component of the ground reaction force overcame the threshold defined as 3 standard deviations of its peak- to peak value during static posture AP force. Gait initiation was performed as a whole movement and divided into two phases*: 1) anticipatory phase*, which lasted from the onset of COP motion to the instant of toe off of the swing limb 2)*: execution phase*, which lasted from toe off of the swing limb to the instant of toe off of the stance leg. Temporal events of gait initiation were obtained from both position and velocity curves derived from markers placed on the calcaneus and fifth metatarsal bones (Mickelborough et al., 2000). These events corresponded to the instants of heel off, toe off and heel contact of the swing limb. Angular displacement of the upper body segments (pelvis, trunk, and head) and whole body COM was estimated using the whole plug-in-gait model in the AP and ML direction.

Muscle activity was determined by surface EMG recordings (BTS Bioengineering, Italy). EMG signals were collected bilaterally using bipolar disposable electrodes (1 cm disc-electrodes, 2 cm inter-electrode distance) from the: sternocleidomastoid (SCM), neck extensors (NE), and erector spinae (ES) at the level of T9 and L3, with a sampling frequency of 1000 Hz. Electrode sites were prepared by gently abrading the skin to ensure good contact. For the SCM, electrodes were positioned at 1/3 of the distance from the sternal notch to the mastoid process at the distal end overlying the muscle belly; the NE electrodes were positioned over the distal half of the distance between the base of the occiput and the spinous process of the seventh cervical vertebrae (Falla et al., 2004) whilst for the ES, electrodes were placed 2cm lateral of the spinal process at T9 and L3.

Gait spatiotemporal parameters

Gait initiation speed was calculated from AP COM velocity, which was estimated using whole body plug-in-gait model. Step length and step width were calculated from the AP and ML distance, respectively, between the position of the swing heel marker at heel-contact and the position of the stance heel marker at toe-off.

Variability of angular displacement

Angular displacement of the pelvis, trunk, and head was filtered using a second-order low-pass Butterworth filter with a cut-off frequency of 5 Hz. To quantify variability of the pelvis, trunk, and head motion during gait initiation, the average standard deviation (AvgSD) was calculated using the following equation:

$$AvgSD = \sqrt{\frac{\sum \theta^2}{100}}$$

θ = Angular displacement of the segment.

This measure has previously been used to assess the stability of individual body segments, with decreased variability indicating increased segment stability and vice versa (Laudani et al., 2006).

Attenuation of upper body accelerations

Acceleration of the pelvis, trunk and head segments was calculated by double derivative of the position of the origin of each upper body segment reference frame in the AP, ML. The magnitude of acceleration of each segment was calculated using the root mean square (RMS) of each acceleration component. RMS acceleration values are known to be `influenced by gait velocity (Kavanagh & Menz, 2008), thus AP and ML RMS acceleration were divided by CC acceleration RMS as proposed by (Iosa et al., 2012) to remove this effect. The ability to attenuate accelerations through the upper body segments was quantified using the attenuation coefficient expressed as a percentage. The attenuation coefficient describes the ability to reduce accelerations from inferior to superior segments, with reduced linear acceleration from inferior to superior parts of the upper body used as an indicator of upper body stability (Summa et al., 2016). The attenuation coefficients were calculated using RMS values of each segment as follows (for both AP and ML direction):

$$\text{Cxy} \quad = \left(1 - \frac{\text{RMS}_x}{\text{RMS}_y}\right) * 100$$

x = inferior segment y = superior segment

Each coefficient represents the attenuation from a lower to an upper body level. C_{PH} representing the attenuation from the pelvis to the head, C_{PT} representing the attenuation from the pelvis to the trunk, and C_{TH} representing the attenuation from the trunk to the head. A positive coefficient value indicated reduced acceleration of the upper segment with respect to the lower segment whilst a negative coefficient value indicated greater acceleration of the upper segment with respect to the lower segment.

Activation patterns of the trunk and neck muscles

Raw EMG signals were first high-pass filtered at 20 Hz to remove movement artifacts, then full-wave rectified and filtered using a second-order high-pass Butterworth filter with a cut- off frequency of 50 Hz using a custom Matlab script. The onset of muscular activity was visually estimated from COP onset by the same experimenter for all calculations, which has been shown to be reliable in clinical applications (Micera et al., 2001). Amplitude of muscular activity was calculated from the area under the EMG curve of each muscle using a customised Matlab script.

Whole body stability during gait initiation

Margin of stability (MOS), using the extrapolated centre of mass (*exCOM*) introduced by Hof et al., (2005), was used to quantify whole body stability in the AP and ML direction. The exCOM concept extends the classical condition for static equilibrium of an inverted pendulum by adding a linear function of the velocity of the COM to COM position. This method describes how close an inverted pendulum is to falling, given the position and velocity of its COM, and the position of the margins of its base of support (BOS). For the calculation of the margin of stability, the positions of the COM and BOS need to be known. BOS was calculated from the distance between the position of the swing heel marker at heel-contact and the position of the stance heel marker at toe off represented the step length and width, and was representative AP and ML BOS respectively. MOS was taken at heel contact of the swing limb, as it has previously been shown that foot strike was systematically made with the heel (Caderby et al., 2014).

The position of the $exCOM$ was then calculated as follows:

$$exCOM = xCOM + \frac{x'COM}{\sqrt{\frac{g}{l}}}$$

With $xCOM$ and $x'COM$ representing the COM position and velocity respectively, g the gravitational acceleration, and l corresponding to the limb length, taken from anthropometric measurements prior to data collection. The MOS corresponded to the difference between the AP and ML BOS and the AP and ML position of the $exCOM$ at heel contact and defined as BOS - $exCOM.$ The lower the MOS value, the closer the $exCOM$ is to the BOS, indicating reduced whole body stability.

Statistical analysis

Statistical analyses were carried out using IBM SPSS, v19 (IBM Corporation, Armonk, NY, USA). The level of significance α was set 0.05.Normality of data was examined and confirmed for all variables using the Shapiro-Wilk test. In order to identify each group's potential change due to the imposed conditions, a paired samples t-test was conducted separately on variables for both age groups. The young group was included as a reference for the condition, therefore between subject comparisons was not implemented. All data is presented as mean ± SD.

Results

Gait initiation speed

Gait initiation speed increased significantly from comfortable to fast for both young (comfortable: 1.13 ± 0.12 $m \cdot s^{-1}$, fast: 1.61 ± 0.23 $m \cdot s^{-1}$ $p < 0.05$) and older (comfortable: 0.86 ± 0.19 $m \cdot s^{-1}$, fast: 1.17 ± 0.32 $m \cdot s^{-1}$ [$p < 0.05$]) groups.

Variability of angular displacement

In the AP direction during both anticipatory and execution phases, older demonstrated greater AvgSD of angular displacement for the head at fast initiation speed compared to comfortable initiation speed ($p < 0.05$). In the ML direction during the anticipatory phase, older demonstrated greater AvgSD of angular displacement for the head at fast initiation speed compared to comfortable initiation speed ($p < 0.05$) (Figure 13) There was, however, no effect of initiation speed on AvgSD values of angular displacement of the pelvis or trunk for older. Figure 13 shows AvSD of angular displacement of the head only, for clarity, in both groups.

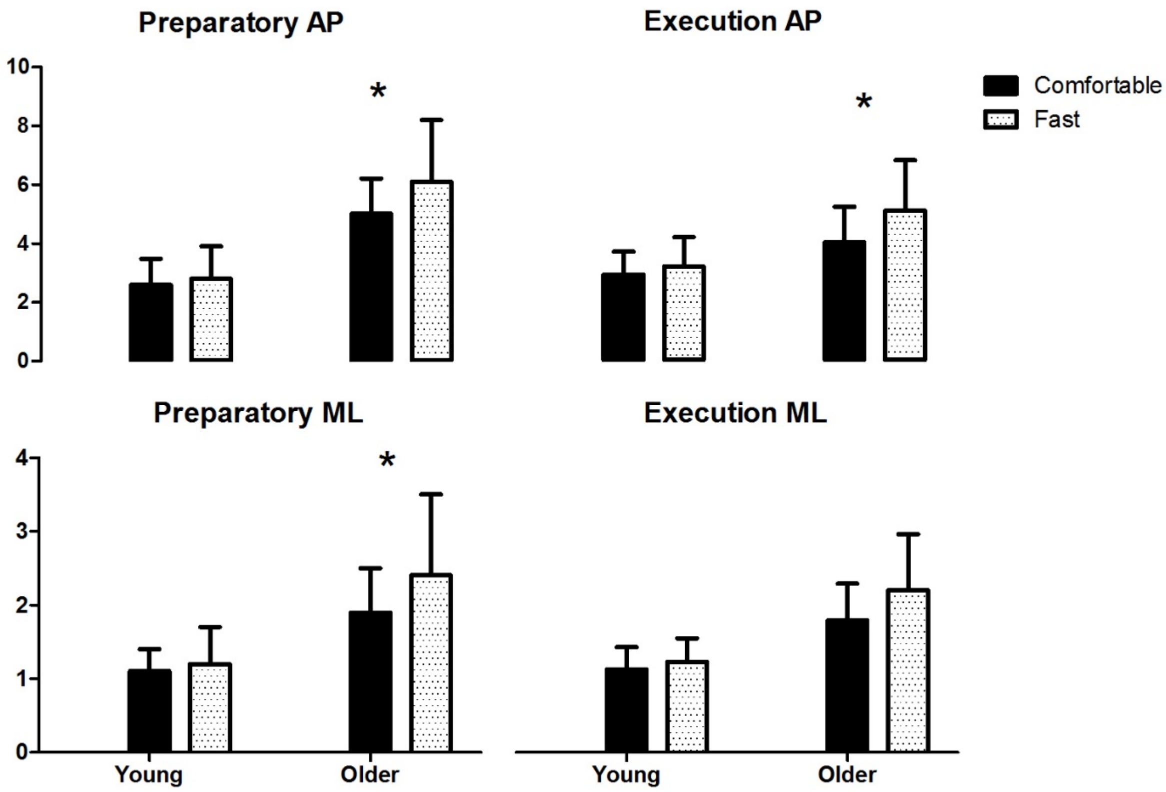

Figure 15. Young and older mean ± SD of variability of the head angular displacement during anticipatory phase and execution phase during comfortable and fast initiation speeds in the anteroposterior direction (AP) and mediolateral direction (ML), evaluated by calculation of the average standard deviation (AvgSD). * indicates condition effect

Attenuation of upper body accelerations

In the AP direction during the anticipatory and execution phase, older demonstrated a reduced ability to attenuate acceleration between the trunk and head at fast initiation speed compared to comfortable ($p < 0.05$). In the ML direction, only during the anticipatory phase, older displayed reduced C_{TH} ($p < 0.05$). There was no effect of initiation speed on the ability to attenuate acceleration between the pelvis and the upper body segments. Young demonstrated no differences of acceleration attenuation between any upper body segments between initiating speeds.

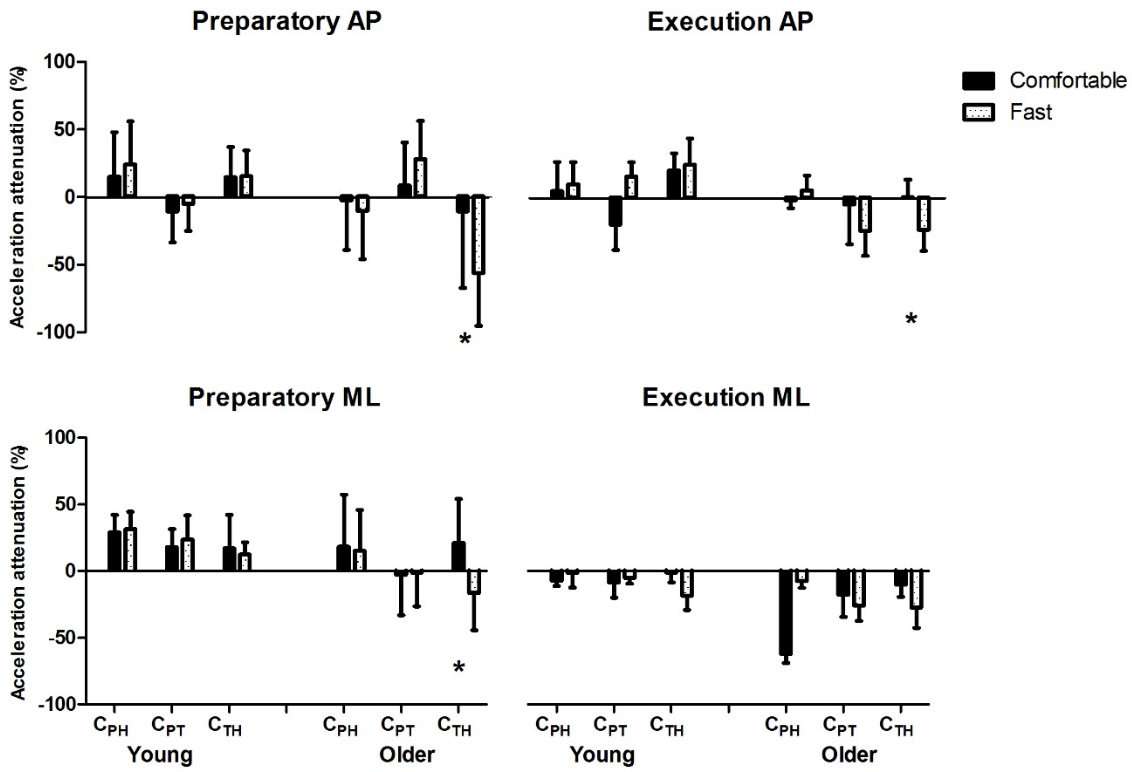

Figure 16. Mean ± SD of the coefficients of attenuation for pelvis-head (C_{PH}), pelvis-trunk (C_{PT}) and trunk-head (C_{TH}) for young and older during the anticipatory phase and execution phase at comfortable and fast initiation speeds in the anteroposterior (AP) and mediolateral (ML) direction. * indicates condition effect

Activation patterns of the trunk and neck muscles

EMG data for young were analysed for all muscles and six out of 12 older was analysed for the neck muscles. Young demonstrated an earlier onset of SCD and NE muscle activity during fast initiation speed compared to comfortable ($p < 0.05$) (Table 6), however amplitude did not change with initiation speed. Older demonstrated an earlier onset muscle activity of all muscle sites (SCD, NE, T9 and L3) during fast initiation speed compared to comfortable. In addition, older had a greater amplitude of the SCD, NE, T9 and L3 during fast initiation speed compared to comfortable ($p < 0.05$) (Table 7).

Table 6. Mean ± SD for the time of the onset of muscle activity given as raw values of muscle onset time and amplitude of the anticipatory and execution phase for young at comfortable and fast initiation speeds for young. * indicates condition effect

Young	Comfortable		Fast	
	Preparatory	Execution	Preparatory	Execution
Flexor				
Onset (ms)	184.5 ± 66.9	-	172.4± 78.1*	-
Amplitude	10.1 ± 16.1	18.9 ± 20.1	12.8 ± 11.1	21.1 ± 16.7
Extensor				
Onset (ms)	369.1 ±	-	298.1 ± 101.2*	-
Amplitude	165.75.9 ± 7.9	15.1 ± 15.2	6.2 ± 4.2	17.6 ± 12.1
Upper Spine (T9)				
Onset (ms)	233.2 ± 116.5	-	202.6 ± 200.1*	-
Amplitude	9.1 ± 8.2	38.2 ± 15.3	10.8 ± 7.0	38.9 ± 21.9*
Lower Spine (L3)				
Onset (ms)	210.0 ± 98.4	-	197.1 ± 88.5	-
Amplitude	9.5 ± 7.1	30.1 ± 11.6	10.2 ± 9.8	35.6 ± 19.7

Table 7. Mean ± SD for the time of the onset of muscle activity given as raw values of muscle onset time and amplitude of the anticipatory and execution phase for older at comfortable and fast initiation speeds for older. * indicates condition effect

Older	Comfortable		Fast	
	Preparatory	Execution	Preparatory	Execution
Flexor				
Onset (ms)	291.4 ± 112.7	-	234.1 ± 104.1*	-
Amplitude	4.2 ± 3.4	5.8 ± 3.0	5.6 ± 7.2*	6.9 ± 8.5*
Extensor				
Onset (ms)	-	-	-	-
Amplitude	3.8 ± 2.9	8.9 ± 4.3	4.7 ± 8.1*	11.4 ± 7.2*
Upper Spine (T9)				
Onset (ms)	183.7 ± 87.9	-	152.2 ± 97.5*	-
Amplitude	10.0 ± 4.7	16.7 ± 8.1	15.1 ± 9.8*	20.6 ± 10.1*
Lower Spine (L3)				
Onset (ms)	252.4 ± 124.5	-	201.9 ± 101.8*	-
Amplitude	6.3 ± 3.8	13.5 ± 7.2	12.7 ± 7.6*	17.9 ± 10.9*

Whole body stability

There was no effect of initiation speed on MOS values for young. There was a significant effect of the initiation speed on MOS values for older ($p < 0.05$). While there was no difference in the AP direction, in the ML direction, older demonstrated reduced whole body stability, indicated by lower MOS values, at fast initiation speed compared to comfortable ($p < 0.05$).

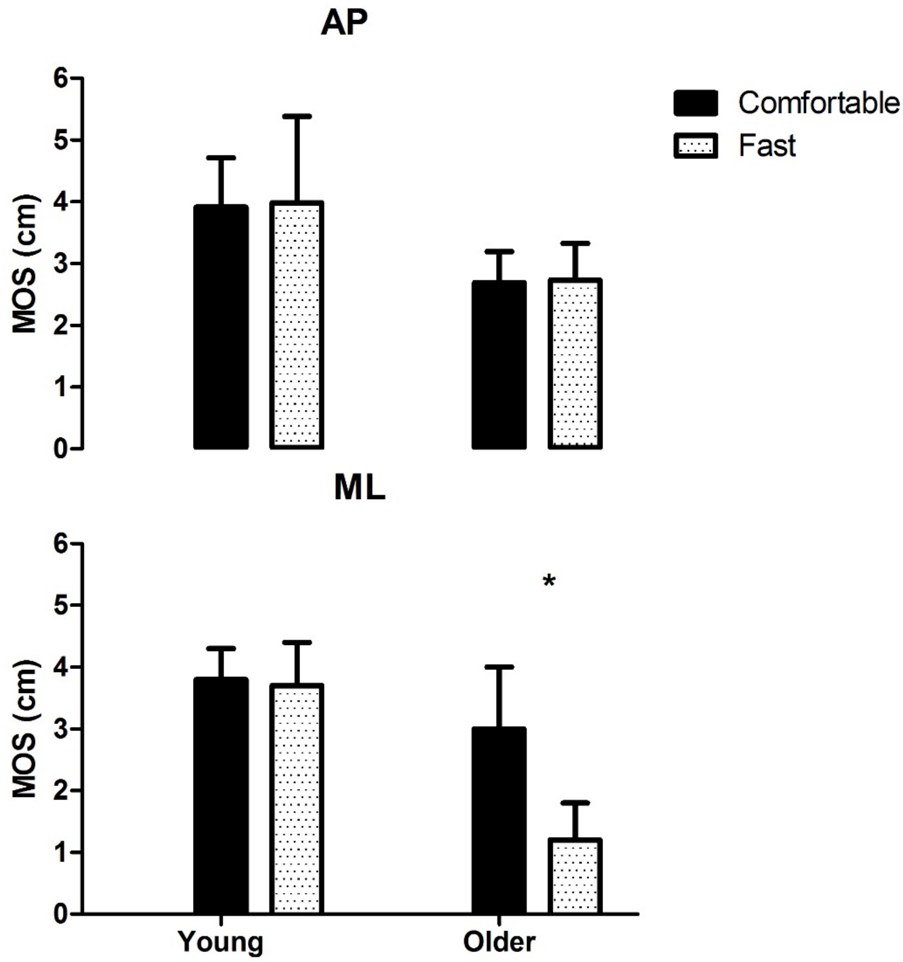

Figure 15. Mean ± SD of Margin of stability (MOS) at swing heel contact in the anteroposterior (AP) and mediolateral (ML) direction for young and older at comfortable and fast walking. * indicates condition effect

Table 8. Mean ± SD of variables contributing to whole body stability for young. BOS = Base of support, xCOM = Centre of mass displacement, x'COM = Centre of mass speed, exCOM = Extrapolated centre of mass, MOS = Margin of stability. * indicates condition effect

Young	AP		ML	
	Comfortable	Fast	Comfortable	Fast
BOS (cm)	78.1 ± 16.0	85.2 ± 21.8*	16.1 ± 3.0	17.2 ± 5.1*
$xCOM$ (cm)	38.7 ± 12.6	30.91 ± 15.6	8.8 ± 2.1	6.15 ± 2.9
$x'COM$ ($cm \cdot s^{-1}$)	113 ± 12.0	161.1± 20.1	11.0 ± 3.0	9.8 ± 3.4
$exCOM$ (cm)	71.28 ± 19.5	81.22 ± 23.1	12.3 ± 3.5	13.5 ± 3.0
MOS (cm)	3.91 ± 0.8	3.98 ± 1.4	3.8 ± 0.5	3.7 ±0.7

Table 9. Mean ± SD of variables contributing to whole body stability for older. BOS = Base of support, xCOM = Centre of mass displacement, x'COM = Centre of mass speed, exCOM = Extrapolated centre of mass, MOS = Margin of stability. * indicates condition effect

Older	AP		ML	
	Comfortable	Fast	Comfortable	Fast
BOS (cm)	59.1 ± 22.0	69.3 ± 24.6	23.0 ± 8.0	17.5 ± 8.2*
$xCOM$ (cm)	29.2 ± 10.1	32.5 ± 9.8	16.9 ± 4.8	11.8 ± 3.9
$x'COM$ ($cm \cdot s^{-1}$)	86.1 ± 24.0	109.2 ± 26.8	10.1 ± 3.0	12.1 ± 2.1*
$exCOM$ (cm)	56.0 ± 19.4	66.6 ± 20.0	20.0 ± 3.9	15.6 ± 2.9
MOS (cm)	2.69 ± 0.5	2.73 ± 0.6	3.0 ± 1.0	1.19 ± 0.6*

Discussion

Older were less able to attenuate accelerations between trunk and head and revealed an earlier anticipatory and greater amplitude of muscle activation when initiating at fast speeds compared to comfortable. Finally older showed greater variability of angular head displacement and reduced ML whole body stability when initiating at fast speeds compared to comfortable and reduced ML whole body stability. As expected, both young and older increased initiating speed from comfortable to fast similar to that of gait initiation of young (Caderby et al., 2014) and steady state gait of (Mazzà et al., 2008), as no studies have studied gait initiation speed at fast of older.

In agreement with the hypothesis, older had a reduced head stabilisation as evidenced by higher AvgSD values and the inability to attenuate accelerations from the trunk to the head when initiating gait at fast speeds. In both age groups, the neck muscles were activated earlier at fast compared to comfortable, suggesting mechanisms of head stabilisation may rely more on feed-forward commands from the CNS when initiating gait at fast speeds, a likely mechanism employed to maintain stability of the visual field and offer protection to the head. Nonetheless, the larger EMG amplitude of the neck and trunk muscles demonstrated by older at fast compared to comfortable is indicative of an inability of using the cervical hinge as an active structure for the attenuation of acceleration. The larger EMG amplitude by older also shows a greater effort required to stabilise the neck during the controlled fall preceding the first step execution (Winter, 1995) which may explicate the inability to attenuate accelerations from the trunk to the head, thus reduced head stability.

Chapter 5 demonstrated reduced head stability compared to young, attributed to a diminished ability to attenuate accelerations from the trunk to the head. Similarly in the present study, older demonstrated a reduced ability to attenuate acceleration from the trunk to the head when initiating gait at fast speeds. Further, previous studies examining steady state gait, also reported that older females have difficulty in attenuating upper body accelerations during fast gait initiation speed when compared to comfortable gait speed (Mazzà et al., 2008). Taken together, the present findings along with the previous studies, suggest older adults have difficulty in attenuating upper body acceleration throughout the whole movement of initiating walking to continuous walking.

Interestingly, older individuals did not follow the same pattern as young, and displayed reduced whole body stability, evidenced by lower MOS values when initiating gait at fast compared to comfortable. Older produced a narrower step width and increased ML COM speed at fast walking compared to comfortable, unlike young. It can be suggested that as older were trying to move forward at a faster speed, more effort was placed in the anterior progression and therefore increased step length was more favoured over a reduced step width to ensure a more efficient and quicker way to move forward. As a consequence, initiating gait at fast speeds poses a higher fall risk threat when compared to comfortable speed in older individuals, which has implications for older as a decline in the control ML stability is known to be a major contributor in the increased risk of falls in older individuals (Maki, 1997).

Our findings on whole body stability are also in agreement with previous data on steady state gait (Hof et al., 2007; Rosenblatt & Grabiner, 2010) and gait initiation (Caderby et al., 2014), which showed that whole body stability was not affected by gait speed in young healthy adults due to an accurate regulation of ML foot placement. In contrast, previous studies have found attenuating accelerations at head level challenging (Mazzà et al., 2008) causing difficulty in controlling displacements of the COM relative to their limits of stability when walking faster (Maki & Mcilroy, 1999). This is the first study to show that head stability and whole body stability are threatened when initiating gait at fast speeds in older, and taken together, results suggest that overall faster speeds can be problematic for older adults during initiating gait and steady state gait.

Health promotion advice encourages older adults to participate in regular moderate intensity activity for example, brisk walking or walking at a faster pace (Pate et al., 1995), while brisk walking has been used as a falls prevention intervention in older adults (Danilovich, Conroy, & Hornby, 2017; Okubo et al., 2016). However, such schemes and interventions that use brisk walking as a fall prevention technique may actually be inducing a fall risk when older adults begin walking, as there was impaired head stabilisation and reduced ML stability (an indicator of increased fall risk) when initiating gait at fast speed. Based on the findings of this study, it is recommended that a more structured approach to walking as a physical activity is employed, to ensure that older adults are able to perform brisk walking without an increased fall risk.

Conclusion

The present study demonstrated that young individuals were able to modulate upper body stabilisation and whole body stability with different gait speeds during gait initiation. In contrast, older adults executed reduced head stabilisation demonstrated by an increased head variability (i.e. decreased stability) and inability to attenuate acceleration from trunk to head when initiating gait at fast speeds. Initiating gait at fast speed has detrimental effects on ML balance in older adults but not young. The present study looked at comfortable and fast gait speeds as separate tasks, future work could be targeted at the change in speed during the same task to replicate everyday activities to gain an understanding of how well older individuals are able to deal with the transition between different speeds during walking.

CHAPTER 9

General Discussion

The aim of the present Thesis was to examine age related changes in head position, head stabilisation and whole body stability during two gait tasks, steady state gait and gait initiation, addressing the methodological issues of focussing on a visual target and different gait speeds. The results showed that a) gait stability was unaffected by head position and different walking speeds during steady state gait, b) decreased head stability in older individuals compared to young during gait initiation can be attributed to a deterioration of the neuromechanical mechanisms relating to head stability, c) free head movement during initiation does not affect head stabilisation or whole body stability but it does affect gait parameters, while d) initiating gait at faster speeds than comfortable compromises the neuromechanical mechanisms underpinning head stabilisation and reduces whole body stability in older individuals.

Effects of ageing on head position and stability

It has previously been reported that older adults implement increased head flexion during steady state gait for various reasons such as an attempt to track lower limb trajectory (Marigold & Patla, 2008), to compensate for impaired lower visual field (Freeman et al., 2007), or to look for obstacles at floor level (Muir et al., 2015) to increase perceived stability. The findings of the present Thesis point to a similar direction, as older adults implemented greater head flexion compared to young adults during steady state gait (Chapter 5). To add, older also implemented increased head flexion during gait initiation in the free head movement condition (Chapter 6) compared to focussing on a visual target, whereas there was no difference in head position during gait initiation with free head movements and focussing on a visual target in young adults (Chapter 6). In the present study there were no obstacles in the walkway and the participants were familiarised with the walkway, yet older still implemented head flexion. Therefore, it can be suggested that this is a habitual effect from adopting this pattern of gait over a period of time as when instructed to look at a visual target, older had the ability to use the neck muscles to restrict head flexion to look at the target during gait and gait initiation, without the change affecting any overall stability measures (Chapter 4 and Chapter 6).

Further to this, it was unknown whether increased head flexion associated with the ageing process is indicative of head instability. The results of Chapter 7 refute such associative links, as evidenced by the EMG data. During gait initiation with or without a visual target to look at, younger showed no difference in head flexion, while older showed higher head flexion in the no visual target condition. Older were able to produce different muscular activation patterns such as earlier onset of SCM and increased relative activation amplitude of SCM and NE when walking focussing on a visual target. These findings suggest that mechanisms of head stability may rely on feedforward commands from the CNS (Prince et al., 1994) as older had to implement anticipatory postural adjustments of the neck muscles in anticipation of adjusting head position to focus on the visual target. This was seen through an earlier onset of SCM and increased activation amplitude of SCM and NE (relative to the anticipatory phase) when walking focussing on a visual target. The earlier muscle activation timing suggests older can alter their stabilisation strategy within the CNS to prepare for movement whilst a greater relative amplitude demonstrates a ‘greater effort’ in order to activate the muscles, thus stability of the head. It was deemed that head flexion is habitual and a greater relative

EMG is needed to restrict head flexion to focus on a visual target. The movement of head flexion with reduced SCM activation in free head movement condition may act as resting the muscle whilst an increase in concurrent activation of the SCM and NE when focussing on a visual target, may lock the head in position. Regardless of how head stability was achieved, there was no difference in head stability between conditions, and an increased head flexion did not affect overall head stability. Together, it could be suggested older may consciously prefer to flex their head during gait and gait initiation without threatening head stability.

Despite older producing greater head flexion and it not being related to head instability, Chapter 6 found older to have decreased head stability and impaired mechanisms underpinning head stability in comparison to young, such as reduced ability to attenuate acceleration between the trunk and head during gait initiation. These findings are in line with previous results of decreased head stability of older individuals during different types of locomotion, including steady state walking (Cromwell et al., 2001) and locomotor transitions such as gait initiation (Laudani et al., 2006). Both studies showed decreased head stability through increased head velocities and variability in space and head-on-trunk, suggesting that head instability is a characteristic of older adults both at gait initiation as well as setady gait too. However, those studies did not examine the anticipatory control of meachanisms underpinning this decreased head stability. The present study adds to the understanding of the mechanisms of head flexion in older adults by examining the neck flexor and extensor muscles activation patterns as well as linking the head flexion with head stability .

Anticipatory postural adjustments are important in order for the body to prepare for movements (Malouin & Richards, 2000) and to deal with a perturbation such as gait initiation. The present Thesis found similar results to previous studies of a decreased head stability during gait initiation in older compared to young evidenced by increased variability using the AvgSD and Principal Component Analysis (Chapter 6). Older adults muscles have been previously shown to be less effective when preparing for gait initiation due to a lower relative amplitude of the tibialis anterior muscle activation compared to young (Khanmohammadi et al., 2015). This reduced effectiveness of the anticipatory postural adjustments of the lower limbs was reflected in the neck muscles of the present Thesis as there was a lower relative amplitude of the SCM in the anticipatory phase of gait initiation. The present Thesis is the first to examine anticipatory postural adjustments of the mechanisms underpinning head stability during gait initiation in older and young adults. This age related decreased head stability was attributed to the age related deterioration of the anticipatory adjustments of the mechanisms underpinning head stability (Chapter 6).

A key mechanism in maintaining head stability is the attenuation of accelerations through the trunk. Older adults tend to develop axial rigidity, which may impair their ability to attenuate the accelerations that are applied to the lower limbs during gait from impacting on head stability. The measurement of attenuation of accelerations through the upper body has previously been investigated as an indicator of head stability for adults, older individuals and adults with pathologies such as Parkinson's (Buckley et al., 2015; Mazzàet al., 2008, Menz et al., 2003). In the present Thesis, whilst young were able to attenuate accelerations from trunk to head, aiding stability of the head, older did not, suggesting acceleration did not decrease from the trunk to the head. The inefficiency in attenuating these accelerations may be attributed to deleterious age related changes to passive structures of the spinal column or to sequential activation of the axial musculature. For the former, it has been previously reported that ageing alters the structural density of vertebra and the vertebra discs, resulting in a more rigid spinal column with increased axial stiffness, which reduces the ability to absorb impact from performed activities allowing greater locomotor impact to travel through the body to the head (Doherty, 2003, Lafortune et al., 1996). For the latter, young participants had a consistent muscular activation strategy while older produced more variable muscular activation strategies, reflecting the previously reported variable activation patterns of the muscles of the lower limbs during gait initiation in older (Mickelborough et al., 2004). Nonetheless, the larger relative EMG amplitude change demonstrated by older during the anticipatory phase of gait initiation compared to the young, is indicative of a greater effort required in an attempt to stabilise the neck. This greater effort by the older may be a result of the inability to attenuate accelerations efficiently, thus greater muscle activation is needed in an attempt to overcome the difficulty in attenuating accelerations from the trunk to the head in an effort to achieve head stability. Chapter 6 highlighted that anticipatory postural adjustments of the upper body have been seen to be less effective during gait initiation in older individuals compared to young, reflecting the age related deterioration of APA's of the lower limbs (Khanmohammadi et al., 2015) due to a reduced ability to attenuate accelerations from the trunk to the head and a delayed onset of neck muscles in order to stabilise the head.

Effects of head movement and head stability on whole body stability

As the present Thesis found age related differences in head position and head stability during gait and gait initiation (Chapter 5 and 7), there was a need to examine the effects head position and head stability had on whole body stability and in turn, make inferences on fall risk during gait and gait initiation. During walking there is an alternation of high and low impact phases which require compensations between movements of the upper body, in particular the trunk and head, in order to attenuate the cyclic perturbations deriving from the lower limbs (Mulavara et al., 2002). Using the extrapolated centre of mass (Hof, Gazendam, & Sinke, 2005) as a measure of whole body provides an indication of the stability of the whole 'system' and it allows to see whether a segment movement, such as the head flexing, can threaten the overall stability and subsequently, the fall risk of the individual. Given that head flexion has been found to be destabilising during static conditions in young (Buckley et al., 2005), it was hypothesised that head flexion may exacerbate the forward shifting of the COM closer to the BOS, further threatening stability. The present Thesis examined whether the increased head flexion implemented by older could also be destabilising during gait, as a large portion of falls happen during walking (Winter, 1995, Rubenstein, 2006). The destabilising effects of head flexion, shown in static conditions in young in a different study (Buckley et al, 2005) were not reflected during steady state gait in the present Thesis. While there was an expected difference in gait stability ratio values between young and older during gait (Cromwell et al., 2001), there was no difference in gait stability ratio values between free head movements (increased head flexion in older) and focussing on a fixed visual target in older (Chapter 5). As the gait stability ratio reflects gait adaptations for higher stability, these findings suggests that maintaining the head in an upright position by means of focusing on a fixed target, did not alter the need for stability. Overall, head position did not affect whole body stability and in turn, may not increase fall risk during gait.

As well as the challenges associated with steady state gait, gait initiation can also challenge stability due to the complex interactions between neural and biomechanical factors and produce greater accelerations experienced as one moves form stationary to moving condition. In order to move from a stationary to dynamic situation, the lower limbs need to rotate the body over the feet (Crenna & Frigo, 1991), therefore creating trunk flexion (Assaiante, Woollacott, & Amblard, 2000). With the addition of an increased head flexion, it was hypothesised this would cause the centre of mass to be shifted considerably closer to the base of support, reasonably provoking instability. It has previously been reported that older adults initiate gait with a concurrent flexion of the trunk and neck system indicating that the upper body behaves rigidly as a single segment (Laudani et al., 2006), with authors hypothesising that this may decrease balance and increase fall risk. Similar to findings of Chapter 5, Chapter 7 found no destabilising effects of head flexion during gait initiation as MOS values remained unchanged between free head movement condition compared to focussing on a visual target. Collectively these findings show that head flexion does not impact on whole body stability during steady state gait or gait initiation. Head flexion did not create a conundrum of prioritising increased information from lower visual field vs increased perceived stability. Instead, findings confirm that this better footfall vision facilitating movement does not impact on whole body stability, probably optimising locomotion in older adults, and therefore is unlikely to be responsible for any falls that occur during gait or initiating gait.

Head stabilisation is reported to be critical in the control of whole body stability (Pozzo et al., 1990). One study examined head stability during gait initiation in the AP direction only (Laudani et al., 2006)., and lacked key measurements related to head stability, such as the ability to attenuate acceleration from trunk to head and the measurement of EMG activation key neck muscles relating to head stability, such as the SCM (Danna-Dos-Santos, Degani, & Latash, 2007) with no measure of whole body stability. The addition of measurements such as acceleration attenuation across the upper body and muscle activity of the paraspinal muscles of the present study enabled a better understanding of the mechanisms underpinning head stability as muscle activation patterns assist in explaining any differences. With a more holistic measure of head stability and the mechanisms underpinning this, an increased variability of movement, inability to attenuate acceleration and more variable patterns of muscle activation, Chapter 6 confirmed that older did experience higher head instability in the AP direction compared to young.

Even though differences in head stability were apparent between age groups, there were no differences in AP MOS between groups. A possible explanation is that upper body differences were not considerable enough to alter AP whole body stability. Despite no differences between groups in the ML direction of upper body variability or attenuation of acceleration, older demonstrated significantly reduced MOS, indicating reduced ML whole body stability. Together, the results of the present Thesis suggest that head movement and head instability may not be sufficient enough to affect whole body stability during gait or gait initiation.

Effects of gait speed on head movement and head stability

Chapter 5 and Chapter 8 investigated the effect of gait speed on steady state gait and gait initiation respectively. During steady gait, Hirasaki et al., (1999) reported that at speeds >1·2 $m \cdot s^{-1}$, there was an increased magnitude of head pitch displacement and a further increase was seen when walking at speeds of 2 m·s^{-1} in young individuals. These results support findings from Chapter 5, as young had greater head flexion change when they moved from comfortable to fast speed (2.37 m·s^{-1} compared to head flexion change from slow to comfortable speed (1.51 m·s^{-1}). The present results agree with Hirasaki et al., (1999) reports that head displacement increases with increasing walking speed for young adults. However, the opposite was found in older, as they produced greater head flexion change when moving from slow to comfortable walking speed compared to the change from comfortable to fast walking speed. These differences, however, were not considerable, as demonstrated by the low effect sizes. Indeed, when the GSR was compared between the difference speeds, there were no differences, supporting the lack of a meaningful impact of head flexion on stability between speeds.

Gait initiation speeds in chapter 6, 7 and 8 were lower than gait speeds in chapter 5 as they were two separate tasks of gait initiation and continuous gait. It has been shown that walking velocity is reached after two steps (Winter, 1995), meaning that during gait initiation (1 step forward), walking velocity has not been reached, explaining why gait initiation speed had lower values than gait speed. Interestingly, older individuals did not follow the same pattern as young, and displayed reduced dynamic stability when initiating gait at fast compared to comfortable speeds, evidenced by lower MOS values. Older produced a narrower step width and increased ML COM speed at fast walking compared to comfortable walking, unlike young who displayed no differences in ML components of stability. These differences are likely to be explained by the demands gait initiation places on the older adults. As older were trying to move forward at a faster speed, more effort was placed in the anterior progression and therefore increased step length was favoured over a reduced step width to ensure a more efficient and quicker way to move forward. The consequence of this reduction on step width, however, could be problematic for older adults, as reduced step width and control of ML stability has been reported to be a major contributor in the increased risk of falling (Maki, 1997). These findings not only further the understanding of the possible effect of gait initiation on fall risk, but also has implications for making older adults more aware of the impact gait initiation can have, thus enabling an adopted approach that would reduce that risk As a consequence, initiating gait at fast speeds poses a higher fall risk threat when compared to comfortable speed in older individuals.

The findings on dynamic stability are also in agreement with previous data from young healthy adults on steady state gait (Hof, van Bockel, Schoppen, & Postema, 2007; Rosenblatt & Grabiner, 2010) and gait initiation (Caderby, Yiou, Peyrot, Begon, & Dalleau, 2014), which showed that dynamic stability was not affected by gait speed. Previous studies have found attenuating accelerations at head level to be challenging in older adults during fast steady state gait (Mazzà, Iosa, Pecoraro, & Cappozzo, 2008) and difficulty in controlling displacements of the COM relative to the limits of stability when walking faster (Maki & Mcilroy, 1999). This is the first study to show that head stability and dynamic stability are threatened when initiating gait at fast speeds in older. Taken together, these results suggest that initiating gait at faster speeds can be problematic for older adults.

Recommendations for researchers, physical activity specialists and older adults

The combination of the findings from the four experimental chapters demonstrates that older adults naturally implement head flexion, however this head movement does not impact head stability nor does it affect whole body stability during gait or gait initiation.

Examining the effect of focusing on a visual target during steady state gait or gait initiation proved neither beneficial nor detrimental effects on whole body stability, however focussing on a visual target during gait initiation did cause gait parameters to alter in both young and older adults during gait initiation. Finally, findings demonstrated that initiating gait at fast speeds did impair head stability and whole body stability.

For researchers

These findings highlighted methodological issues; subsequently, consideration must be given to these aspects to ensure a more accurate assessment of gait and stability in older individuals. The following recommendations must be take into account the findings of the present Thesis and offer guidance on gait assessments in the older population.

1. Allow for a naturally adopted head positon during gait and gait initiation assessment to achieve a more realistic understanding of the effect of a given intervention.

2. If using a visual target during gait initiation, take into account that this may underestimate gait parameters, consequently results interpretation which may not be correctly applicable to natural gait.

For public health specialists

These findings also have practical implications for physical activity specialists and interventions, and recommendations made for physical activity in older adults. Walking is the most recommended form of physical activity for older adults, while brisk walking is widely recommended as a way to reduce fall risk (Paillard, Lafont, Costes-Salon, Rivière, & Dupui, 2004). However, the findings of the present Thesis support some caution on these recommendations. The population used in the present study was of healthy, older adults with good levels of physical activity. Despite that, they still demonstrated decreased ML stability during gait initiation. Assuming less physically active adults adopt the same strategy, this reduction in ML stability could increase their risk of falling. Therefore, it is recommended that

3. A more structured approach to walking as a physical activity is employed, to ensure that older adults are able to perform brisk walking without an increased fall risk.

4. Awareness of older adults that starting walking with the intention of moving fast increases instability and should be at least avoided unless necessary.

For older adults

Commonly in research, there is an unequal power relationship between academic researchers and the research participants which in turn, limits the application of the findings themselves. A recent report by Age UK entitled "Engaging with older people evidence review", published in 2011 called to break down the barrier of this 'expert' knowledge and 'lay' knowledge. It is important that results are not just used amongst researchers and health professionals, but that results of a given study are disseminated directly back to the participants who are ultimately the intended recipients of the research. Therefore recommendations to older adults are:

5. The orientation of your head, whether it be your head pointing downwards or having your head up looking forwards, does not affect your balance when starting to walk or when walking, therefore walk with your head in a position that is comfortable for you.

6. From a standing position, do not try to start walking at your fast speed. Start walking at your comfortable, and then increase walking speed once you reach steady walking after two steps.

Limitations

The present Thesis did present some limitations. An inherent limitation throughout all experimental chapters was recruitment bias. It was found that the older female volunteers were able, independent older adults, which indeed, is not representative of all older adults therefore, findings may not translate directly to a wider population, which has an established low physical activity participation and a higher frailty risk (Fried et al., 2001). However, differences were found in the mechanisms underpinning head stability, even in healthy older adults who had no prior history of falls that might otherwise indicate functional decline. Therefore, the observed changes in the mechanisms may be a precursor to future, greater, functional declines.

'Older adult' is a term frequently used within literature, however there can be a degree of uncertainty around the terminology of an older adult. Older adults are usually defined as over the age of 65 years, however thereafter, age categories are not well standardised. To be able to interpret results of research studies when comparing old to young, it is necessary to clarify what old age *per se* is. Different age categories of 'old' have been defined as young-old: 65-74 years, old: 75-84 years, old –old: 85-99 years and oldest-old: 100+ years (Spirduso et al, 2005). Often, in studies concerning older adults, age ranges of over 30 years are used, combining the younger cohort of older age with the older age, while no attention is given to potential differences (Whaley, 2014).

In the present Thesis, low sample numbers did not allow for bracketing into sub age categories as suggested by Whaley (2014), resulting in an age range of 65 to 89 years in the older age group. Therefore it was not possible to account for variability across different age ranges within older individuals. The generalisability of these age ranges in older adults can result in differences within and between older adults to be missed. Therefore it is important to recognise the age ranges used and also to identify age categories in future work. Overall, further work is needed to investigate whether there are differences between different subgroups of older adults. In addition to the low sample number, inter-subject variability could also impact in another way. It is possible that the smaller sample size, along with the increased standard deviation may have impacted negatively on the studies' power. Therefore, the conclusions must be interpreted with some caution.

The use of EMG presented some limitations, most notably the inability to identify the muscle activation onset of the SCM and NE of the older individuals, as there was data for only six of the 11 older individuals. Although great effort was taken with area preparation of the electrode placement, there were some challenges to the fixation of the electrodes to the participants' skin that included: 1) difficulty in locating landmarks for EMG electrode placement because of loose skin and excess adipose tissue, and 2) electrode adherence to the skin during movement some resulting in EMG traces to be noisy, making it impossible to identify the point of onset. There are known challenges. of using EMG in older adults and is frequently a problem seen in studies using EMG with older adults and gait (Hanada et al., 2008).

Throughout the 4 experimental chapters, there were no measures of eye movement to confirm that participants were actually looking at the visual target. Previous studies have examined how eye movements looking at a visual target can affect postural control in static and dynamic conditions in older adults (Thomas et al., 2017; Thomas, Bampouras, Donovan, & Dewhurst, 2016). In static conditions it was shown that focusing on a visual target compared to smooth pursuits and saccadic eye movements reduced postural sway whilst focussing on a fixed visual target compared to smooth pursuit's increased dynamic stability. However, those studies focused on gaze, and thus no head movement was examined. Indeed it may have been useful to understand the relationship between head flexion and eye movement, however this was not the aim of the research project. The fact that the orientation of the head did change when instructed to look at a visual target was sufficient to assume that participants were focusing on the visual target.

Future Studies

The present Thesis was set to examine the mechanisms contributing to head stabilisation, and whole body stability during two gait tasks, steady state gait and gait. Although the findings of the experimental chapters contribute towards understanding of gait initiation and the mechanisms and mechanics of it in older adults, as well as its potential impact, the experiments also provided areas for future research needed to gain a better and more holistic understanding. The suggested future areas are outlined below:

Study 1

As the present study examined the effects of head stability and whole body stability during gait and a transitory task of gait initiation, this lends future research to focus on the study area of the transitory task of gait termination. Examining head stability and whole body stability measures during gait termination will help gain an understanding of the whole movement from initiating gait, to cyclic walking of gait and finally terminating gait. It would also be useful to conduct correlation analyses between head stability and whole body stability in order to strengthen the associative link between them.

Study 2

Given that the present Thesis found instability measures during planned movement of gait initiation in older individuals, it would be interesting to investigate unplanned gait termination, which is reflective of a real life situation. Investigating unplanned gait termination will allow researchers to identity of which of the three components (gait initiation, steady state gait and gait termination) is of highest instability and therefore fall risk in older individuals. This in turn will provide information in order to tailor specific physical activity interventions to develop exercises to improve/control upper body stabilisation and whole body stability during the ‘high risk’ stages of the gait cycle.

Study 3

As mentioned, a limitation of the present study was that there were no measure of eye movements. It must be considered that head orientation is not indicative of gaze vector location, therefore understanding the link between eye movements and head position/ stability during such locomotive tasks would be useful. Previous studies have examined how eye movements looking at a visual target can affect postural control in static and dynamic conditions in older adults (Thomas et al., 2017; Thomas, Bampouras, Donovan, & Dewhurst, 2016), however instructions were to keep the head as still as possible. Little is known about how eye movements may link with head stability when the first recommendation point from the section above is considered in such locomotive tasks in older individuals.

Chapter 10

Conclusion and contribution to knowledge

In summary, the present Thesis showed that there are age related changes in head movement and stability during gait and gait initiation, however this is not reflective of whole body stability when walking and initiating gait at comfortable speed. In contrast, older adults found modulating head stability and whole body stability difficult when initiating gait at fast speeds. Additionally, visual target implementation had little effect on whole body stability in both young and older. These findings suggest that head instability in older individuals may not be as detrimental as has been previously stated, therefore allowing a more natural approach allowing for head movement in order to give more realistic results. Findings may inform interventions to exert caution when aiming to increase initiating gait speed as it may be in fact be fall provoking for older adults to exceed initiating speed to what is naturally perceived as comfortable.

Several of the findings from the present Thesis agree with previous literature, offering further support into establishing these results as ‘normative’ for the e.g. differences between young and old gait characteristics, muscle activation patterns etc. However, the examination of the effect head movement on gait and gait initiation, has not been explored before, allowing for several aspect of the Thesis to be the first to demonstrate the effect of head movement on gait and gait initiation, as well as its application on older adult’s locomotion, thus contributing and furthering the knowledge in the field.

More specifically, the findings from Chapter 5 further the understanding of the role of head movement in locomotion of older adults, providing evidence that head flexion during walking in older adults is a habitual activity, enforced by the need for better footfall vison. This movement, however, is not increasing the risk of becoming unstable in older adults.

Gait initiation is a surprisingly little investigated area in older adults, despite walking having received a large amount of attention. Walking, however, is not possible without gait initiation while the change in the state of motion is substantially higher. The present Thesis has provided an initial motivation in the investigation of this particular aspect, by examining the effect of head stability on gait initiation. This was done through investigation of the mechanisms regulating head stability and how they are affected by different conditions, such as increased speed of movement and visual target fixation. These studies contribute to knowledge by providing useful information to researchers regarding the mechanisms for head stability, as well as showing for the first time how head control is achieved during gait initiation. These findings can help researchers to expand on the area and obtain a more holistic understanding of head movement during gait initiation, compare to gait termination and thus contribute to a complete understanding of human locomotion (from start to stopping) in elderly, the importance of which in fall risk has already been highlighted in the Thesis. Further, the EMG results, in particular, offer a baseline to clinicians and neurophysiologists using EMG on what a typical behaviour of healthy older adults would be, offering a possible comparison for assessment of e.g. neuromuscular control deterioration, gait problems, or stroke rehabilitation.

The aim of the Thesis was to examine the effect of head movement on dynamic stability, using dynamic stability and gait related measures. Findings in the latter suggest that head movement can affect gait patterns, altering step length and step width. The changes of these two parameters, although not affecting overall stability (please see chapter 7 for specific results), have implications for gait assessment. With more calls for ecologically valid testing (e.g. Stellman et al., 2015), the novel findings of the Thesis support the use of a naturally moving head to allow for a more accurate assessment of the 'true' gait, thus avoiding conclusions with little transferability to 'real life'.

The final, and perhaps wider–reaching, aspect of the Thesis contributing to knowledge is the application of these findings to large cohort initiatives for increasing physical activity levels, such as brisk walking. Although a common recommendation for older adults (Ebrahim,et al., 1997), some reservations exist about the increased risk this activity can pose. The present results support those calls by showing for the first time that gait initiation at a faster speed can increase the fall risk of older adults. This knowledge should be imperative for professionals working with older adults on physical activity interventions, in order to ensure that recommendations for brisk walking are avoided in less active older adults, as the reduction of ML step width (in indicator of fall risk in older adults) shown in the present Thesis was demonstrated in the gait initiation of older, physically active adults.

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Appendix 1

The effect of gait velocity on posture and dynamic stability in older adults

Participant Information Sheet

About the study
This research will aim to provide an insight into age-related changes in walking patterns. It will focus on the speed of walking and how it affects walking ability, posture and dynamic stability. Being able to change walking speed is important in everyday life, for example, when crossing the road or rushing to get a bus. This is a concern that needs to be addressed as to whether dynamic stability and posture is affected by a change in walking speed.

Some questions you may have about the research project:

Why have you asked me to take part?

Fall risk is an important factor within older adults and can have a detrimental effect on independent living. Falls are commonly seen during walking. As there is an ageing population it is crucial to examine factors that may increase this fall risk. You have been asked to take part in this study as you fall within the inclusion criteria for participation.

What will I be asked to do?

You will be one of many older females (aged 65+ years) who will take part in the study. Before any testing is undertaken, you will be reminded of all procedures and will sign a consent form. Testing will be carried out within an informal atmosphere at the location of the human performance laboratory of the University of Cumbria, Lancaster.

The researcher will talk through the procedure and you will be asked to complete a short health questionnaire and sign a consent form.

You will be asked to complete the following:

Gait Analysis: You will be asked to walk along a 10m flat walkway at 3 different self-selected speeds (slow, natural and fast). You will then be asked to repeat these trials but with a fixed gaze forward. All individual trials will be repeated twice.

Posture Analysis: While walking the 10m walkway you will have some joints marked with a marker or a tape and a video camera will record your walking.

How often will I have to take part and for how long?

You will be asked to participate in one session which will last approximately 45 mins.

When will I have the opportunity to discuss my participation?

You can discuss your participation in the study with the researcher at any time during your participation. Alternatively, you use the contact deatails provided at the bottom of the form.

How long will data be kept and where?

All data that is collected will be stored on a password protected computer in the Sports Centre at the University of Cumbria and will only be accessed by the named investigators. Data in hard copy form will be stored in a locked filing cabinet, only able to be accessed by the named investigators. To ensure participant anonymity, each participant will be assigned an identification code. A list of identification codes and corresponding participant names will be kept separate from the data

What will happen to the information when this study is over?

The information will be analysed and written up as a research paper. All data will be stored at the University of Cumbria with signed consent forms stored separately and according to University policies.

Will anyone be able to connect me with what is recorded and reported?

The information collected in connection with this project will remain confidential forthe duration of the study and after its completion. All records will be stored at the University of Cumbria with signed consent forms stored separately and according to University policies. The publication of the results will not result in you being identified with particular responses.

How can I find out about the results of the study?

You can be provided with your own data from the test immediately if requested. You may also receive a copy of the submitted article on request. If you would like to find out about the results you can contact the researcher on the contact details provided at the bottom of the form.

What if I do not wish to take part?
Your participation in the study is entirely voluntary.

What if I change my mind during the study?
You are free to withdraw from the study at any time without having to provide a reason for doing so.

Will I need to sign any documentation?
You will be asked to sign a consent form before participating in the study.

Whom should I contact if I have any further questions?
Please contact the researcher directly (details below).

Complaints
All complaints from the paritipants are in the first instance to be directed to the Director of Research Office and Head of the Graduate School, University of Cumbria, Bowerham Road, Lancaster, LA1 3JD

Researcher Contact Information:
Amy Maslivec

University of Cumbria
Bowerham Road
Lancaster, LA1 3JD

Tel: 01524 590869
Mobile: 07702228092
Email: amy.maslivec@cumbria.ac.uk

Appendix 2

Participant Consent Form

Please answer the following questions by circling your responses:

Have you read and understood the information sheet about this study? YES NO

Have you been able to ask questions about this study? YES NO

Have you received enough information about this study? YES NO

Do you understand that you are free to withdraw from this study at any time, and without having to give a reason for withdrawal? YES NO

Your responses will be anonymised before they are analysed.

Do you give permission for members of the research team to have access to your anonymised responses? YES NO

Do you agree to take part in this study? YES NO

Your signature will certify that you have voluntarily decided to take part in this research study having read and understood the information in the sheet for participants. It will also certify that you have had adequate opportunity to discuss the study with an investigator and that all questions have been answered to your satisfaction.

Signature of participant:.. **Date:**................

Name (block letters):...

Signature of investigator:.. **Date:**................

Appendix 3

PLEASE READ THE FOLLOWING CAREFULLY AND COMPLETE AS REQUIRED

NAME ______________________________ AGE ____ Date of Birth _______________

- Are you in good health? YES / NO
  If no, please specify below:
  ____

- Have you suffered from a serious illness or accident? YES / NO
  If yes, please specify below:
  ____

- Are you currently taking any medication either supplied on prescription or purchased over the counter? YES / NO
  If yes, please specify below:
  ____

- Are you currently attending your G.P. for any condition? YES / NO
  If yes, please give particulars:
  ____

- Are you presently taking part in any other laboratory experiment? YES / NO
  If yes, please give particulars:
  ____

- Are there any other personal reasons that you would like to discuss relating to your participation in this experiment/assessment? YES / NO

- Persons may be considered unfit to do the assessment if they: Have an infectious disease, have a fever, suffer from fainting spells or dizziness or have a known history of medical disorders such as high blood pressure, sweating disorder, heart or lung disease.
  If any are applicable to you, please give particulars:
  ____

- Please give information on family history in terms of heart disease, fainting or dizzy spells, particularly in relation to exercise
  ____

Signature of volunteer____________________________________

Date ____________________________________

Appendix 4

UNIVERSITÀ DEGLI STUDI DI ROMA "FORO ITALICO"

FOGLIO ILLUSTRATIVO

Titolo dello studio

Controllo motorio della stabilità della testa durante l'avvio e l'arresto del cammino in donne giovani ed anziane

Stato dell'arte e scopo della ricerca

E' noto che gran parte delle cadute avviene a causa di una perdita dell'equilibrio durante camminate di breve distanza e/o durante compiti locomotori di transizione come l'avvio e l'arresto del cammino. Se il ruolo della parte inferiore del corpo è quello di favorire la propulsione del corpo in avanti o l'arresto dello stesso, la parte superiore del corpo invece gioca un ruolo fondamentale per il mantenimento dell'equilibrio corporeo. La stabilizzazione della testa, in particolare, fornisce una piattaforma di riferimento per l'ottimizzazione dello sguardo e l'integrazione delle informazioni visive, vestibolari e somato-sensoriali.

L'obiettivo di questa ricerca è quello di indagare i meccanismi di controllo neuromuscolare alla base della stabilità della testa e dell'equilibrio corporeo durante l'avvio e l'arresto della cammino in donne giovani e anziane.

Rilevanza del progetto

I risultati di questo studio permetteranno una conoscenza più approfondita delle differenze tra giovani ed anziani per favorire la progettazione di nuove e più accurate proposte di intervento con l'obiettivo di migliorare l'equilibrio e ridurre il rischio di cadute nell'anziano.

Cosa comporta la sua partecipazione allo studio

Nel caso in cui decidesse di partecipare allo studio, la Sua presenza sarà richiesta nel laboratorio di Bioingegneria dell'Apparato Locomotore nell'Università degli Studi di Roma "Foro Italico" in una singola occasione della durata di 2 ore.

Inizialmente saranno effettuate una serie di misurazioni antropometriche preliminari quali l'età, l'altezza, il peso e la lunghezza degli arti. Le saranno posizionati sulla pelle o sul vestiario 35 piccoli marcatori riflettenti (piccole palline del diametro di 1 centimetro con una base adesiva) in diversi punti del corpo. Per rilevare l'attività dei muscoli del collo, della schiena e delle gambe, inoltre, saranno posizionate delle piccole piastrine adesive sulla superficie corporea per registrare il cosidetto segnale elettromiografico di superficie.

Le sarà quindi richiesto di effettuare separatamente prove di avvio e arresto del cammino a tre diverse andature: lenta, confortevole, veloce. Durante l'intera seduta sperimentale, dovrà indossare un abbigliamento il più possibile aderente per consentire la fissazione dei marcatori e la loro visibilità (per esempio un top sportivo o una canottiera aderente e un paio di pantaloni corti e, se possibile, aderenti, fino al ginocchio). L'uso di scarpe basse non da ginnastica è raccomandato per le prove.

Per le prove di *avvio del cammino,* Le verrà chiesto di rimanere in posizione eretta su una pedana di forza il più immobile possibile con il busto verticale, le braccia lungo i fianchi e i piedi in una posizione confortevole. Successivamente, dopo il segnale dello sperimentatore, dovrà iniziare a camminare compiendo non meno di tre passi in avanti. Saranno effettuate 5 prove di avvio del cammino per ognuna delle tre andature (15 prove in totale).

Per le prove di *arresto del cammino,* dovrà camminare lungo un percorso lineare di 5 metri e poi fermarsi e mantenere la stazione eretta per 3 secondi. Cinque prove saranno effettuate per ognuna delle tre andature (15 prove in totale).

Entrambe le tipologie di prove (avvio e arresto del cammino) saranno effettuate con o senza la presenza di un obiettivo visivo posizionato a una distanza massima di 5 metri da Lei.

Partecipazione volontaria

Si ricordi che Lei è libero/a di ritirarsi in qualsiasi momento dalla partecipazione allo studio senza l'obbligo di dover fornire la motivazione. Inoltre può discutere i dettagli della sua partecipazione con il ricercatore in qualunque punto della sessione sperimentale.

Riservatezza dei dati personali

Tutti i dati riguardanti la Sua persona saranno trattati con massima riservatezza e nel rispetto della Sua privacy, ai sensi dell'art. 10 della legge n. 675 del 31.12.1996. Per questo motivo, nessun dettaglio personale che potrebbe permettere la Sua identificazione sarà incluso; i Suoi dati saranno identificati con un numero e discussi solamente dai ricercatori per scopi scientifici. I risultati dello studio cui partecipa potranno essere oggetto di pubblicazione in ambito scientifico, ma la Sua identità rimarrà sempre segreta.

Possibili rischi/disagi

Non sono presenti rischi associabili a questo esperimento poiché Lei effettuerà un'attività ad un'intensità tipica della vita di tutti i giorni: il cammino. Nonostante il posizionamento e il mantenimento degli elettrodi e dei marcatori sia completamente indolore, essi devono essere posizionati sulla cute o in alcuni casi sui vestiti, completamente visibili. La Sua privacy sarà comunque sempre rispettata e potrà indossare una canottiera aderente nel caso in cui ritenesse l'uso del top disagiante.

Ulteriori informazioni

Per ulteriori informazioni e comunicazioni, La preghiamo di contattare i seguenti sperimentatori:

Luca Laudani (luca.laudani@uniroma4.it)
Palazzo IUSM – tel: 06 36733560
Andrea Macaluso (andrea.macaluso@uniroma4.it)
Palazzo IUSM – tel: 06 36733242

Appendix 5

GARANTE PER LA PROTEZIONE DEI DATI PERSONALI

Dichiarazione di Consenso Informato

Studio: ***Controllo motorio della stabiltà della testa durante l'avvio e l'arresto del cammino in donne giovani ed anziane***

Responsabile delle sedute sperimentali: Amy Maslivec

Il/la sottoscritto/a…………... dichiara:

Di aver avuto a disposizione tempo sufficiente per poter leggere e comprendere quanto contenuto nel Documento Informativo.

Di aver ricevuto esaurienti spiegazioni in merito alla richiesta di partecipazione allo studio sperimentale in oggetto, secondo quanto riportato nel Documento Informativo.

Di aver potuto porre tutte le domande che ho ritenuto necessarie. Di

aver ricevuto risposte soddisfacenti.

Di essere stato informato del mio diritto di ritirarmi dallo studio in qualsiasi momento, senza dover dare spiegazioni.

Di essere consapevole di poter richiedere sempre informazioni sull'andamento dello studio per quanto mi riguarda.

Di autorizzare i responsabili della sperimentazione ad accedere ai dati relativi alla mia partecipazione allo studio, che verranno comunque trattati come strettamente confidenziali.

Di aver avuto tempo e modi idonei per prendere una decisione consapevole.

Di accettare liberamente di partecipare allo studio, avendo capito completamente il significato della richiesta ed avendo compreso i rischi ed i benefici che esso implica.

Data ____

Firma del volontario ____

……………………………………………………………………………………………………

Il sottoscritto Amy Maslivec attesta che il partecipante………………………………., dopo aver letto il Documento Informativo, ha sottoscritto il presente consenso a partecipare allo studio in oggetto, avendo compreso le informazioni ricevute ed essendo in grado di formulare una scelta pienamente consapevole.

Data____________________________________ Firma del responsabile _
____

Appendix 6

UNIVERSITÀ DEGLI STUDI DI ROMA "FORO ITALICO"
Piazza Lauro De Bosis 15, 00194 Roma - +39.0636733.501/599 - www.iusm.it

Dipartimento di Scienze del Movimento Umano e dello Sport

Questionario sullo stato di salute

COGNOME NOME

DATA DI NASCITA TELEFONO

Nome e indirizzo del medico generico:

Nome e numero di telefono di uno stretto conoscente da contattare in caso di emergenza:

1) E' stato mai ricoverato o curato per gravi malattie? □ NO □ SI: quali?

2) Assume continuativamente medicinali? □ NO □ SI: quali?

3) Ha mai avuto infortuni o problemi ortopedici? □ NO □ SI: quali?

Quanto tempo fa?

4) Attualmente ha problemi di salute? □ NO □ SI: quali?

5) Ha mai avuto svenimenti o malori durante un'attività fisica? □ NO □ SI: quando?

6) E' stato giudicato non idoneo in visite medico-sportive? □ NO □ SI: quando?

7) Ha effettuato un esame elettrocardiografico di recente? □ NO □ SI: quando?

8) Nei familiari più intimi (genitori, fratelli, sorelle, nonni e zii) ci sono stati casi di morte improvvisa o diabete, pressione alta, infarto, malattie della circolazione o ereditarie?

□ NO □ SI: quali decessi o malattie?

9) Hai mai sofferto delle seguenti patologie?

Infarto cardiaco: □ NO □ SI: quando?

Sofferenza cardiaca (sintomi di stenosi aortica, pericardite acuta, miocardite acuta, aneurisma, angina, patologie valvolari, aritmie, claudicazione): □ NO □ SI: quando?

Tromboflebite o embolia polmonare: □ NO □ SI: quando?

Disturbi cerebrovascolari: □ NO □ SI: quando?

Febbre alta (negli ultimi tre mesi): □ NO □ SI: quando?

Disturbi metabolici (diabete, disturbi della tiroide): □ NO □ SI: quando?

Patologie sistemiche (tumori, artrite reumatoide): □ NO □ SI: quando?

Disturbi psichici (esaurimento o depressione): □ NO □ SI: quando?

Artrite agli arti inferiori (incapacità di compiere vigorose contrazioni degli arti inferiori senza
dolore): □ NO □ SI: quando?

Fratture agli arti inferiori □ NO □ SI: quando?

11) Ha mai effettuato un'artroscopia chirurgica agli arti inferiori? □ NO □ SI: quando?

12) Ha mai perso la capacità di muoversi per più di una settimana? □ NO □ SI: quando?

13) E' in grado di camminare da solo, senza alcun supporto? □ NO □ SI

14) Avverte delle difficoltà quando cammina, sale le scale o corre? □ NO □ SI: quali?

15) Avverte difficoltà a respirare quando fa attività fisica? □ NO □ SI

Data Firma