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Mechanisms of head stability during gait initiation in young and older women: A neuro-mechanical analysis

A. Maslivec[a,b,*], T.M. Bampouras[b], S. Dewhurst[c], G. Vannozzi[d], A. Macaluso[d], L. Laudani[b,d,e]

[a] Department of Clinical Sciences, Brunel University, United Kingdom
[b] Active Ageing Research Group, Medical and Sports Sciences, University of Cumbria, United Kingdom
[c] Department of Sport and Physical Activity, Bournemouth University, United Kingdom
[d] Department of Movement, Human and Health Sciences, University of Rome Foro Italico, Italy
[e] Cardiff School of Sport and Health Sciences, Cardiff Metropolitan University, United Kingdom

A R T I C L E I N F O

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A B S T R A C T

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 neuro-mechanical mechanisms underpinning head stabilisation in young and older women during gait initiation. Eleven young (23.1 ± 1.1 yrs) and 12 older (73.9 ± 2.4 yrs) women initiated walking at comfortable speed while focussing on a fixed visual target at eye level. A stereophotogrammetric system was used to assess variability of angular displacement and RMS acceleration of the pelvis, trunk and head, and dynamic stability in the anteroposterior and mediolateral directions. Latency of muscle activation in the sternocleidomastoid, 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 of sternocleidomastoid 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 delayed onset of neck muscles activation.

1. 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 challenge 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 and 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 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

* Corresponding author at: Department of Clinical Sciences, Brunel University, United Kingdom.
E-mail address: amy.maslivec@brunel.ac.uk (A. Maslivec).

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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 neuro-mechanical mechanisms underpinning head stabilisation in young and older individuals during gait initiation. In particular, we aimed to examine control of upper body accelerations and muscle activation patterns of the trunk and neck, which represent two of the main neuro-mechanical strategies underpinning head stability. Additionally, we investigated the control of dynamic balance in young and older participants 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 dynamic stability, compared to the younger women.

2. Methods

2.1. 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 dynamic 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.

2.2. 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 us to design 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.

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) preparatory 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 (1 cm disc-electrodes, 2 cm inter-electrode distance) from the: sternocleidomastoid (SCM), 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 (Falla et al., 2004); and for the ES, electrodes were placed 2 cm lateral of the spinal process at T9 and L3.

2.3. Data analysis

2.3.1. 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 preparatory 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:

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

x = 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 (preparatory phase: 1–100%, execution phase: