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 neuro- mechanical mechanisms underpinning head stabilisation in young and older women during gait initiation. Eleven young (23.1±1.1yrs) and 12 older (73.9±2.4yrs) 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 of 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.

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 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 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.

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 dynamic stability declines to a greater extent than males (Wolfson et al, 1994) and tend to fall more often (Schultz, Ashton-Miller, & Alexander, 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 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.

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 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: 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 and 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):

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

x = inferior segment   y = superior segment

each coefficient representing 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 artefacts, 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 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 preparatory phase.

Dynamic stability during gait initiation

Margin of stability, using the extrapolated centre of mass (exCOM) introduced by Hof et al (2005), was used to quantify dynamic 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 = 9.81$m \cdot s^{-1}$, 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 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 dynamic stability.

Statistical analysis

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 preparatory phase. Statistical significance was assessed with an alpha level of 0.05. All data are presented as mean ± SD unless otherwise stated. All statistical analyses were carried out using IBM SPSS v19 (SPSS, Chicago, ILL).

Results

Variability of angular displacement

During the preparatory phase, older had a significantly higher AvgSD of AP 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 AvgSD of AP angular displacement of the pelvis and trunk between groups. During the execution phase, there were no differences in AvgSD of AP angular displacement of the pelvis, trunk or head between groups (Figure 1). During both the preparatory phase and execution phase, there were no differences in AvgSD of ML angular displacement of the pelvis, trunk or head between groups (Figure 1).

INSERT FIGURE 1 HERE

PCA of angular displacement is presented in Figure 2 and 3 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 2).

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.

INSERT FIGURE 2 HERE

INSERT FIGURE 3 HERE

Attenuation of upper body accelerations

During the preparatory and execution phase, young displayed significantly greater AP RMS acceleration for the pelvis, trunk and head compared to older ($p < 0.05$) (Figure 4A and B). During the preparatory 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 4C)). During the execution phase, there were no significant differences in acceleration attenuation between groups (Figure 4D).

During the preparatory and execution phases, there was no difference in ML RMS acceleration for the pelvis, trunk or head between age groups (Figure 5A and B). During the preparatory 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 5C). During the execution phase, both groups did not attenuate ML accelerations, however there were no significant differences between groups (Figure 5D).

INSERT FIGURE 4 HERE

INSERT FIGURE 5 HERE

Muscle activity

Older displayed a significantly delayed muscle activity onset of the SCM compared to young ($p < 0.05$) (Table 1). There were no differences in muscle activity onset time for the ES (T9) or ES (L3) between groups. (Table 1).

INSERT TABLE 1 HERE

Dynamic stability

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

INSERT FIGURE 6 HERE

Discussion

The purpose of the study was to examine any age-related change in the neuro-mechanical strategies underpinning head stabilisation and dynamic 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 delayed anticipatory activation of the SCM compared to young. Finally, older demonstrated reduced ML dynamic stability, while there was no difference between age groups for AP dynamic stability. Older participants showed greater variability of head angular displacement in AP direction compared to young participants during both the preparatory 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 et al., 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) and gait termination (Rum et al, 2017). 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, 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. This mechanism, however, may be impaired in older as they demonstrated a delayed onset of the SCM, which could explicate the decreased head stability and the inability to attenuate accelerations from the trunk to the head in the preparatory phase.

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, 1997). Interestingly, differences in upper body stabilisation between young and older were only observed in the AP direction during the present investigation. 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 dynamic 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 dynamic stability. This may have implication for fall risk as dynamic stability can be an indicator of fall risk (Lockhart and Liu, 2008; Toebes et al., 2012). Caderby et al (2014) observed that young were able to maintain ML dynamic stability during gait initiation, while ML dynamic stability in older during gait initiation warrants further research to generate an understanding of why ML dynamic 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 delayed activation of the neck flexor muscles. On the other hand, there was a discrepancy between head stabilisation and dynamic stability in the AP and ML direction, meriting further investigation.

Conflict of interest

The authors declare that they have no conflict of interest

References

Anders, C., Wagner, H., Puta, C., Grassme, R., Petrovitch, A., Scholle, H.-C., 2007. Trunk muscle activation patterns during walking at different speeds. Journal of Electromyography and Kinesiology 17, 245–252. doi:10.1016/j.jelekin.2006.01.002

Breniere, Y., Do, M.C., 1986. When and how does steady state gait movement induced from upright posture begin? Journal of biomechanics 19, 1035–40.

Caderby, T., Yiou, E., Peyrot, N., Begon, M., Dalleau, G., 2014. Influence of gait speed on the control of mediolateral dynamic stability during gait initiation. Journal of biomechanics 47, 417–23. doi:10.1016/j.jbiomech.2013.11.011

Ceccato, J.C., de Sèze, M., Azevedo, C., Cazalets, J.R., 2009. Comparison of trunk activity during gait initiation and walking in humans. PLoS ONE 4. doi:10.1371/journal.pone.0008193

Davis, R.B., Õunpuu, S., Tyburski, D., Gage, J.R., 1991. A gait analysis data collection and reduction technique. Human Movement Science 10, 575–587. doi:10.1016/0167-9457(91)90046-Z

de Sèze, M., Falgairolle, M., Viel, S., Assaiante, C., Cazalets, J.-R., 2008. Sequential activation of axial muscles during different forms of rhythmic behavior in man. Experimental brain research 185, 237–47. doi:10.1007/s00221-007-1146-2

Doherty, T.J., 2003. Invited review: Aging and sarcopenia. Journal of applied physiology (Bethesda, Md. : 1985) 95, 1717–27. doi:10.1152/japplphysiol.00347.2003

Falla, D., Rainoldi, a., Merletti, R., Jull, G., 2004. Spatio-temporal evaluation of neck muscle activation during postural perturbations in healthy subjects. Journal of Electromyography and Kinesiology 14, 463–474. doi:10.1016/j.jelekin.2004.03.003

Greig, C.A., Young, A., Skelton, D.A., Pippet, E., Butler, F.M.M., Mahmud, S.M., 1994. Exercise Studies with Elderly Volunteers. Age and Ageing 23, 185–189.

Hof, A.L., Gazendam, M.G.J., Sinke, W.E., 2005. The condition for dynamic stability. Journal of biomechanics 38, 1–8. doi:10.1016/j.jbiomech.2004.03.025

Iosa, M., Fusco, A., Morone, G., Pratesi, L., Coiro, P., Venturiero, V., De Angelis, D., Bragoni, M., Paolucci, S., 2012. Assessment of upper-body dynamic stability during walking in patients with subacute stroke. Journal of rehabilitation research and development 49, 439–50.

Kavanagh, J., Barrett, R., Morrison, S., 2006. The role of the neck and trunk in facilitating head stability during walking. Experimental brain research 172, 454–63. doi:10.1007/s00221-006-0353-6

Kavanagh, J.J., Barrett, R.S., Morrison, S., 2005. Age-related differences in head and trunk coordination during walking. Human movement science 24, 574–587.

Kavanagh, J.J., Menz, H.B., 2008. Accelerometry: a technique for quantifying movement patterns during walking. Gait & posture 28, 1–15. doi:10.1016/j.gaitpost.2007.10.010

Khanmohammadi, R., Talebian, S., Hadian, M.R., Olyaei, G., Bagheri, H., 2015a. Characteristic muscle activity patterns during gait initiation in the healthy younger and older adults. Gait & Posture 43, 148–153. doi:10.1016/j.gaitpost.2015.09.014

Khanmohammadi, R., Talebian, S., Hadian, M.R., Olyaei, G., Bagheri, H., 2015b. Preparatory postural adjustments during gait initiation in healthy younger and older adults: Neurophysiological and biomechanical aspects. Brain research 1629, 240–9. doi:10.1016/j.brainres.2015.09.039

Kirkwood, R.N., Resende, R.A., Magalhães, C.M.B., Gomes, H.A., Mingoti, S.A., Sampaio, R.F., n.d. Application of principal component analysis on gait kinematics in elderly women with knee osteoarthritis. Revista brasileira de fisioterapia (Sao Carlos (Sao Paulo, Brazil)) 15, 52–8.

Laudani, L., Casabona, A., Perciavalle, V., Macaluso, A., 2006. 2006 Basmajian Student Award Paper: Control of head stability during gait initiation in young and older women. Journal of electromyography and kinesiology 16, 603–610. doi: 10.1016/j.jelekin.2006.08.001

Lockhart, T.E., Liu, J., 2008. Differentiating fall-prone and healthy adults using local dynamic stability. Ergonomics 51, 1860–1872. doi:10.1080/00140130802567079

Maki, B.E., 1997. Gait changes in older adults: predictors of falls or indicators of fear. Journal of the American Geriatrics Society 45, 313–20.

Mazzà, C., Iosa, M., Pecoraro, F., Cappozzo, A., 2008. Control of the upper body accelerations in young and elderly women during level walking. Journal Of Neuroengineering And Rehabilitation 5, 30.

McCrum, C., Epro, G., Meijer, K., Zijlstra, W., Brüggemann, G.-P., Karamanidis, K., 2016. Locomotor stability and adaptation during perturbed walking across the adult female lifespan. Journal of Biomechanics 49, 1244–1247. doi:10.1016/j.jbiomech.2016.02.051

Menz, H.B., Lord, S.R., Fitzpatrick, R.C., 2003. Age-related differences in walking stability. Age and Ageing 32, 137–142.

Micera, S., Vannozzi, G., Sabatini, A.M., Dario, P., 2001. Improving detection of muscle activation intervals. IEEE Engineering in Medicine and Biology Magazine 20, 38–46. doi:10.1109/51.982274

Mickelborough, J., Van Der Linden, M.L., Richards, J., Ennos, a. R., 2000. Validity and reliability of a kinematic protocol for determining foot contact events. Gait and Posture 11, 32–37. doi:10.1016/S0966-6362(99)00050-8

Mickelborough, J., van der Linden, M.L., Tallis, R.C., Ennos, A.R., 2004. Muscle activity during gait initiation in normal elderly people. Gait & Posture 19, 50–57. doi:10.1016/S0966-6362(03)00016-X

Nagano, H., Begg, R., Sparrow, W.A., 2013. Ageing effects on medio-lateral balance during walking with increased and decreased step width, in: 2013 35th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC). IEEE, pp. 7467–7470. doi:10.1109/EMBC.2013.6611285

Pozzo, T., Berthoz, A., Lefort, L., 1990. Head stabilization during various locomotor tasks in humans. I. Normal subjects. Experimental brain research 82, 97–106.

Rum, L., Laudani, L., Macaluso, A., Vannozzi, G., 2017. Upper body accelerations during planned gait termination in young and older women. Journal of Biomechanics 65, 138–144.

Summa, A., Vannozzi, G., Bergamini, E., Iosa, M., Morelli, D., Cappozzo, A., 2016. Multilevel Upper Body Movement Control during Gait in Children with Cerebral Palsy. PloS one 11, e0151792. doi:10.1371/journal.pone.0151792

Toebes, M.J.P., Hoozemans, M.J.M., Furrer, R., Dekker, J., van Die?n, J.H., 2012. Local dynamic stability and variability of gait are associated with fall history in elderly subjects. Gait & Posture 36, 527–531. doi:10.1016/j.gaitpost.2012.05.016

Table 1. The time of the onset of muscle activity given as a percentage of total duration of the preparatory 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**
Upper spine (T9)			
Onset (%)	42.2 ± 20.5	63.3 ± 24.7	0.182
Lower spine (L3)			
Onset (%)	53.1 ± 25.6	60.7 ± 22.5	0.192

Legends

Figure 1. Young and older mean ± SD of variability of the pelvis (top row), trunk (middle row) and head (bottom row) segment angular displacement during preparatory 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.

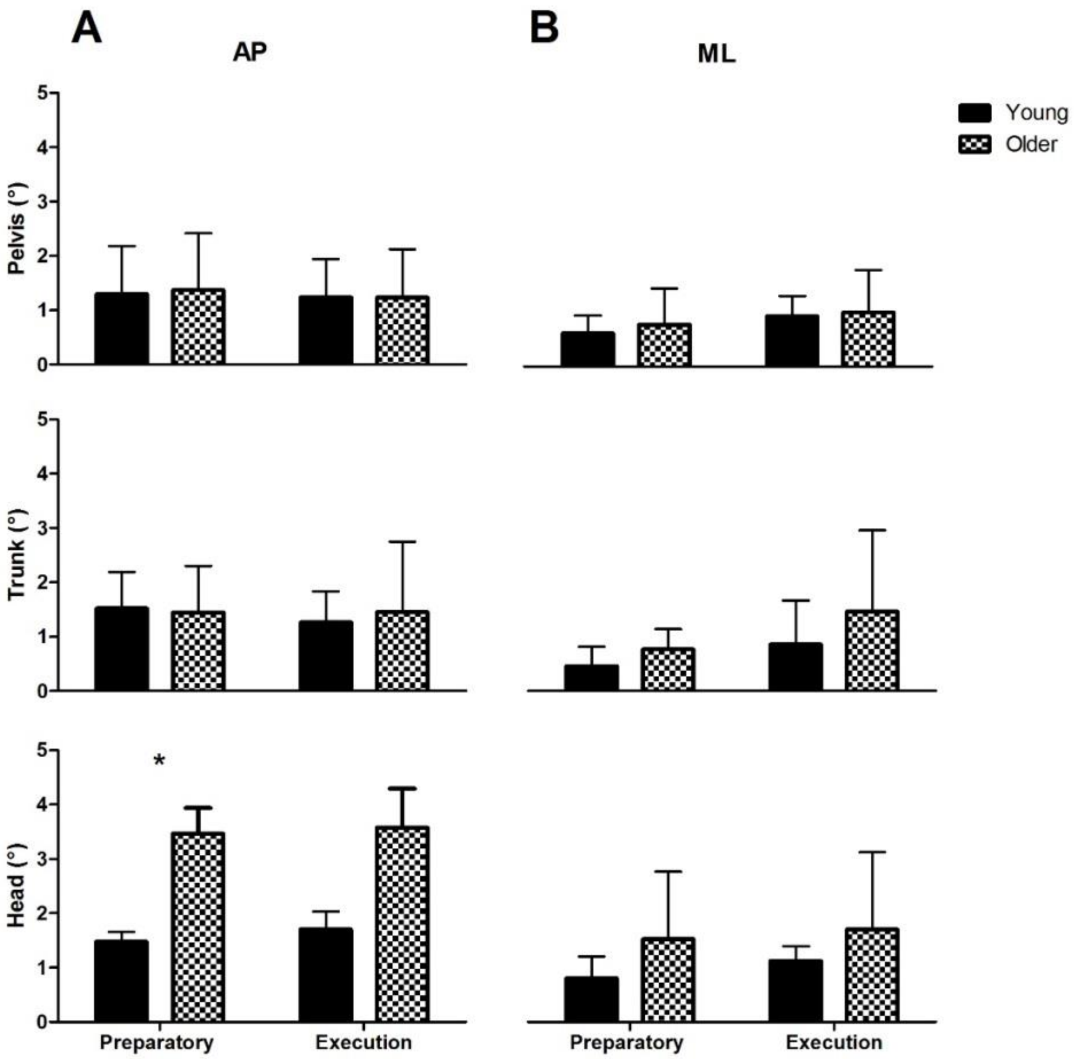
Figure 2. Principal component analysis on the data set of angular displacement of the pelvis, trunk, and head in the anteroposterior (AP) direction during the whole movement of gait initiation. Positive and negative values indicate flexion or extension, respectively (direction is indicated by the arrow to the left of the figures). The axes intersection (0) represents the COP onset, the first perforated line indicates the end of the preparatory 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 3. 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 preparatory 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 4. 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 (C_{PH}), pelvis-trunk (C_{PT}) and trunk-head (C_{TH}) (panel C & D)for young and older during the preparatory phase and execution phase in the anteroposterior (AP) direction. *indicates significance between groups

Figure 5. 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 (C_{PH}), pelvis-trunk (C_{PT}) and trunk-head (C_{TH}) (panel C & D) for young and older during the preparatory phase and execution phase in the mediolateral (ML) direction.

Figure 6 Margin of stability (MOS) at swing heel contact in the anteroposterior (AP) and mediolateral (ML) direction. * indicated significant difference between young and older.


A
AP
B
ML
Young
Older
5
4
3
2
1
0
Pelvis (°)
Trunk (°)
Head (°)
*
Preparatory
Execution
Preparatory
Execution

495 Figure 1

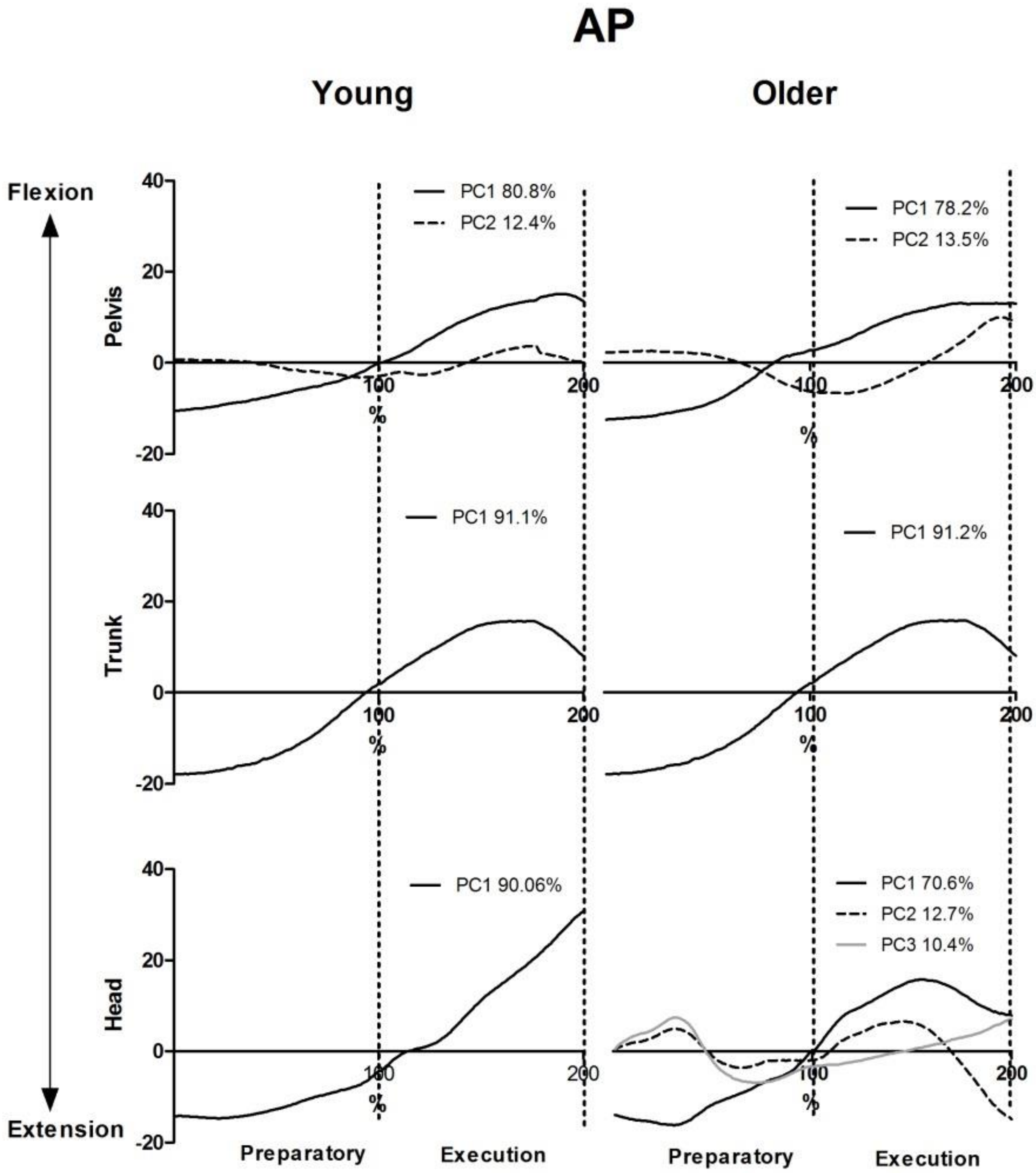
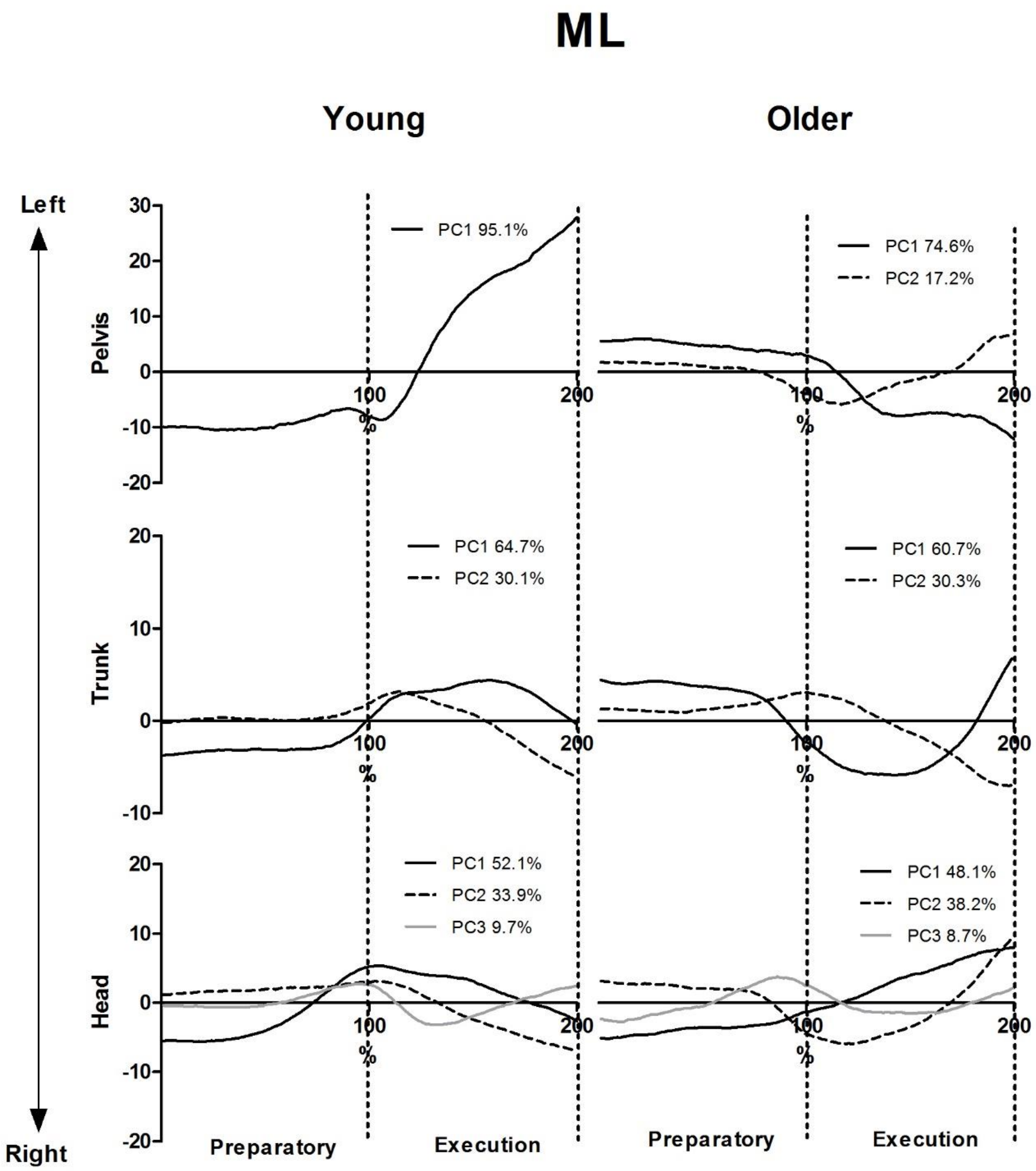

AP
Young
Older
Flexion
Extension
Pelvis
Trunk
Head
PC1 80.8%
PC2 12.4%
PC1 78.2%
PC2 13.5%
PC1 91.1%
PC1 91.2%
PC1 90.06%
PC1 70.6%
PC2 12.7%
PC3 10.4%
40
20
0
-20
100
200
%
Preparatory
Execution

Figure 2



502

503 Figure 3

504

505

506

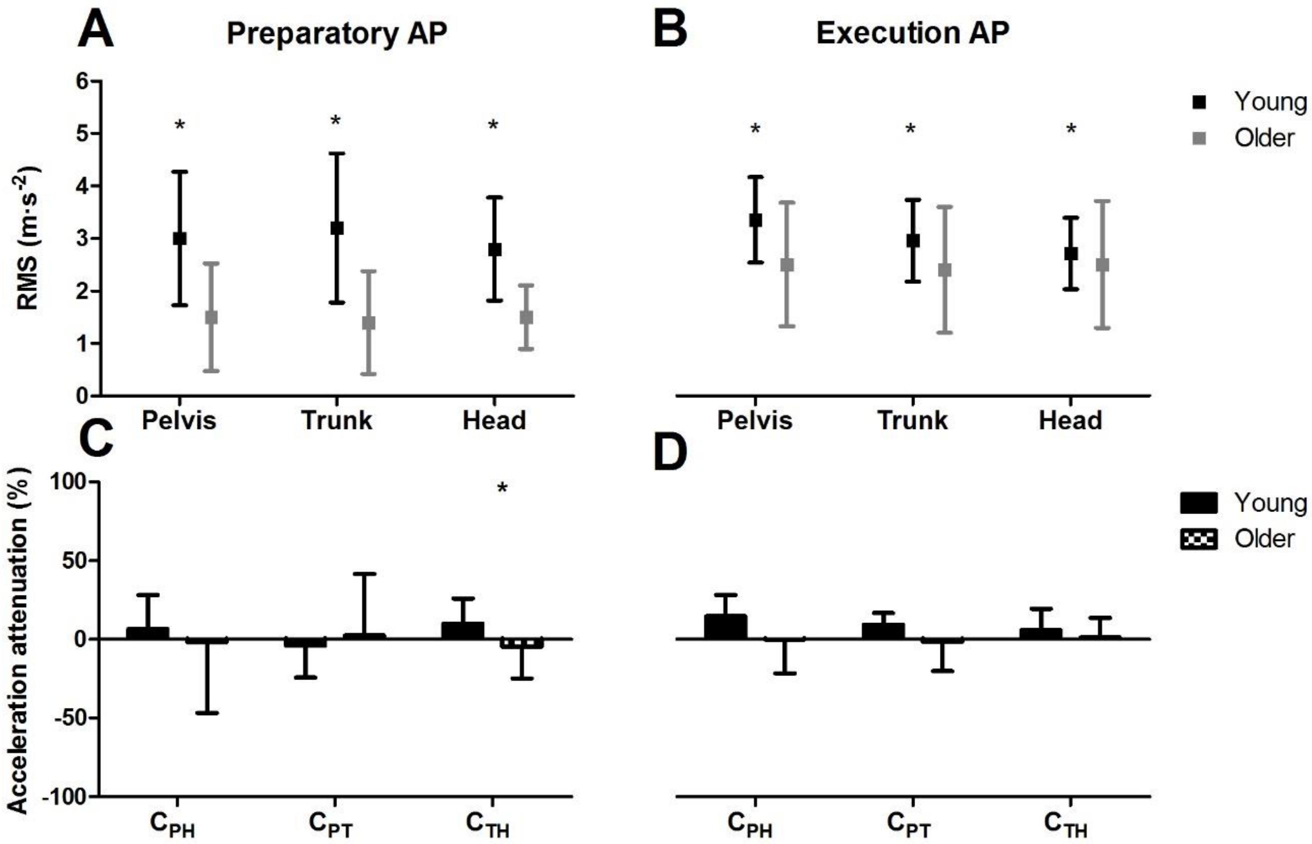

A
Preparatory AP
B
Execution AP
RMS (m·s-2)
Pelvis
Trunk
Head
Young
Older
C
D
Acceleration attenuation (%)
CPH
CPT
CTH
Young
Older

Figure 4

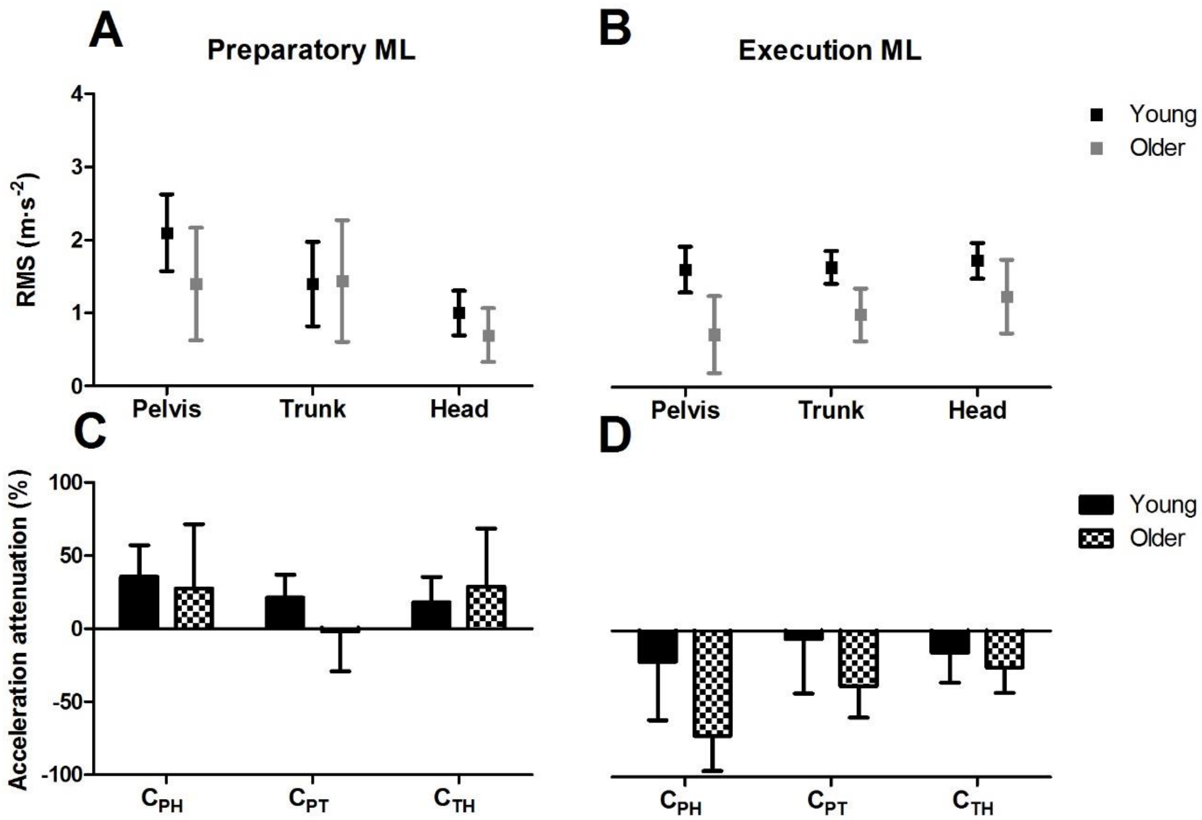

A
Preparatory ML
B
Execution ML
RMS ($m \cdot s^{-2}$)
4
3
2
1
0
Pelvis
Trunk
Head
Young
Older
C
D
Acceleration attenuation (%)
100
50
0
-50
-100
C_{PH}
C_{PT}
C_{TH}

Figure 5

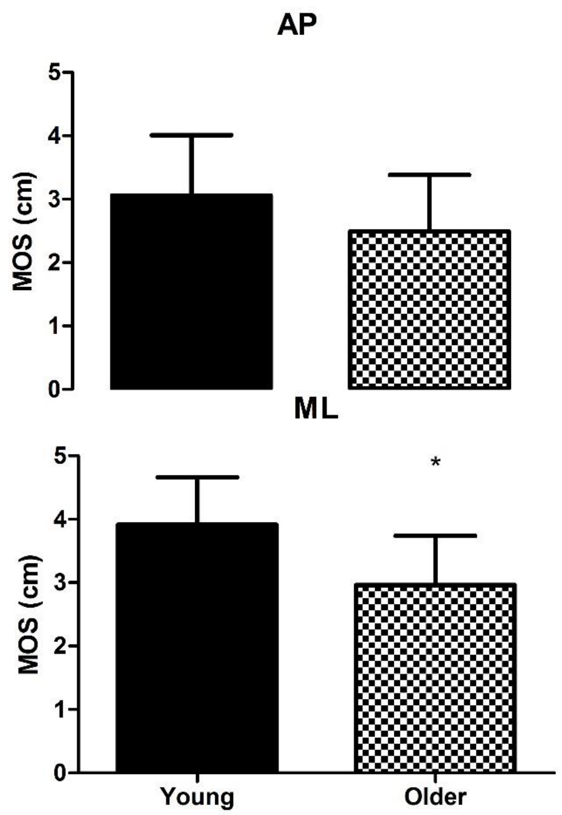


Figure 6