Influence of aging on the control of the whole-body angular momentum during volitional stepping: An UCM-based analysis

Teddy Caderby, Angélique Lesport, Nicolas A Turpin, Georges Dalleau, Bruno Watier, Thomas Robert, Nicolas Peyrot, Jérémie Begue

Experimental Gerontology 178 (2023) 112217

Contents lists available at ScienceDirect

Experimental Gerontology

journal homepage: www.elsevier.com/locate/expgero

Influence of aging on the control of the whole-body angular momentum during volitional stepping: An UCM-based analysis

Teddy Caderby [a,*], Angélique Lesport [a], Nicolas A. Turpin [a], Georges Dalleau [a], Bruno Watier [b], Thomas Robert [c], Nicolas Peyrot [a,d], Jérémie Begue [a]

[a] Laboratoire IRISSE, EA4075, UFR des Sciences de l'Homme et de l'Environnement, Université de la Réunion, Le Tampon, France
[b] LAAS-CNRS, CNRS, UPS, Université de Toulouse, Toulouse, France
[c] Laboratoire de Biomécanique et Mécanique des Chocs, LBMC UMR_T9406, Univ Lyon – Univ Gustave Eiffel, Lyon, France
[d] Mouvement - Interactions - Performance, MIP, Le Mans Université, EA 4334, 72000 Le Mans, France

ARTICLE INFO

Section Editor: Zsolt Radak

Keywords:
Balance
Elderly
Uncontrolled manifold theory
Gait
Rotational kinetics

ABSTRACT

Evidence suggests that whole-body angular momentum (WBAM) is a highly controlled mechanical variable for performing our daily motor activities safely and efficiently. Recent findings have revealed that, compared to young adults, older adults exhibit larger range of WBAM during various motor tasks, such as walking and stepping. However, it remains unclear whether these age-related changes are ascribed to a poorer control of WBAM with age or not. The purpose of the present study was to examine the effect of normal aging on WBAM control during stepping. Twelve young adults and 14 healthy older adults performed a series of volitional stepping at their preferred selected speed. An Uncontrolled Manifold (UCM) analysis was conducted to explore the presence of synergies among the angular momenta of the body segments (elemental variables) to control WBAM (performance variable); i.e., to stabilize or destabilize it. Results revealed the existence of a stronger synergy destabilizing the WBAM in the sagittal-plane older adults compared to young adults during stepping, while there was no difference between the two groups in the frontal and transversal planes. Although older participants also had a larger range of WBAM in the sagittal plane compared to young adults, we found no significant correlation between synergy index and the range of WBAM in the sagittal plane. We concluded that the age-related changes in WBAM during stepping are not ascribed to alterations in the ability to control this variable with aging.

1. Introduction

When performing volitional motor activities, such as standing up from a chair or walking, appropriate regulation of the overall body rotation is required for ensuring efficient realization of the task while maintaining balance. This overall body rotation may be characterized by a mechanical quantity that is the whole-body angular momentum (WBAM), usually expressed at the body's center of mass (CoM), which corresponds to the sum of angular momenta produced by the rotation of all body segments about the body's CoM. In particular, mounting evidence suggests that WBAM is a parameter tightly controlled by the central nervous system (Herr and Popovic, 2008). For example, during walking, it has been found that this mechanical quantity was kept small, i.e., near zero, through coordinated motions of all body segments (Herr and Popovic, 2008; Bennett et al., 2010). This ability to minimize WBAM during walking has been shown to be impaired in populations with mobility problems, such as amputees (Silverman and Neptune, 2011; Pickle et al., 2014; Pickle et al., 2016) and post-stroke patients (Nott et al., 2014; Vistamehr et al., 2016; Vistamehr et al., 2018). These patients exhibit a larger range of WBAM during walking than able-bodied individuals. Importantly, it has been found that a larger range of WBAM during walking was associated to lower clinical balance scores in these patients (Nott et al., 2014; Vistamehr et al., 2016), suggesting poorer balance ability and thus an increased risk of falling.

Several authors studied the effects of normal aging in the regulation of WBAM during various motor tasks. During an experimental tripping task, Pijnappels et al. (2005) showed that older adults were unable to sufficiently reduce WBAM during the push-off phase after tripping over

* Corresponding author at: Laboratoire IRISSE, EA4075, UFR des Sciences de l'Homme et de l'Environnement, Université de la Réunion, 117 rue du Général Ailleret, 97430 Le Tampon, Ile de la Réunion, France.
E-mail address: teddy.caderby@univ-reunion.fr (T. Caderby).

https://doi.org/10.1016/j.exger.2023.112217
Received 23 September 2022; Received in revised form 11 May 2023; Accepted 21 May 2023
Available online 25 May 2023

an obstacle. The larger WBAM after tripping in older adults reduced balance recovery success and led to falls in these individuals contrarily to younger individuals. During walking, Vistamehr and Neptune (2021) have shown that older adults exhibited a higher range of frontal-plane WBAM compared to younger adults, because of a higher destabilizing external moment during single-limb stance induced by a wider foot placement with respect to the CoM in the older adults. During volitional stepping, Begue et al. (2019, 2021) have shown that older people had a higher range of sagittal-plane WBAM than young adults. This difference was associated to larger angular momenta of trunk and legs in older adults compared to their younger counterparts (Begue et al., 2021). Taken together, these findings suggest that aging modifies the regulation of WBAM in daily motor tasks. This has been generally interpreted by the aforementioned authors as a poorer control of WBAM with advancing age. Nevertheless, this hypothesis has not been tested in these early studies. Thus, the question of whether the age-related changes in WBAM reflect a poorer control of this mechanical variable with advancing age remains open. Such knowledge could be particularly important in designing interventions to reduce falls in the elderly.

Maintaining a small magnitude of WBAM requires control, and implies specific organization between body segments. The Uncontrolled manifold (UCM) analysis is an approach commonly used to identify control variables, hereinafter referred to as the performance variables, and the degree to which these variables are controlled by a set of redundant elemental variables across repetitions of a motor task (Latash et al., 2007). This approach consists in partitioning the variance of the elemental variables around their mean trajectories, which are related to the performance variable, into two components: one that has no effects on the performance variable (variance within the UCM space, V_{UCM}) and one that affects the performance variable (variance orthogonal to the UCM space, V_{ORT}). From the relative amount of these two variance components, it is possible to identify the presence of synergy among the elemental variables aimed at maintaining or modifying the performance variable ($V_{UCM} > V_{ORT}$ or $V_{UCM} < V_{ORT}$, respectively), or to reveal the absence of control of this variable ($V_{UCM} = V_{ORT}$). The UCM approach has previously revealed the existence of synergies between body segments to control WBAM during various motor tasks, including walking (Robert et al., 2009), sit-to-stand (Reisman et al., 2002) or precision jump (Maldonado et al., 2018). Furthermore, this approach has also been employed to evaluate the effects of age on the control of a number of performance variables, such as the trajectory of the swing foot during walking (Krishnan et al., 2013), the trajectory of the fingertip during a pointing task (Verrel et al., 2012), ground reaction forces during sit-to-stand (Greve et al., 2013) and curb negotiation (Cui et al., 2020). Nevertheless, to the best of our knowledge, no study has investigated the effects of normal aging on the control of WBAM during stepping using the UCM approach.

The objective of this study was to examine the effect of normal aging on the control of WBAM during stepping based on the UCM approach. Based on our previous findings (Begue et al., 2019; Begue et al., 2021), we tested the hypothesis that normal aging would alter the control of WBAM during stepping, with the presence of a weaker inter-segmental synergy to control this variable in older adults compared to younger adults. Furthermore, we assume that the poorer control of WBAM would be correlated with the larger range of WBAM in older adults.

2. Methods

2.1. Population

Twelve healthy young adults (age: 25 ± 3 years; body mass: 60 ± 10 kg; height: 169 ± 10 cm) and 14 healthy older adults (age: 68 ± 4 years; body mass: 61 ± 12 kg; height: 160 ± 10 cm) volunteered for this experiment. All participants were physically active and free of any neurological, orthopedics or any disorders that could affect their normal balance and gait. Older adults did not report a fall in the 12 months before the start of the study. All subjects were informed for the study procedure and gave their written consent before their participation to the study, which was approved by the local ethics committee.

2.2. Procedure and experimental set-up

Initially, participants stood barefoot in a natural upright posture as still as possible with their arms alongside their body. After a verbal "go" signal from the experimenter, participants were asked to initiate step forward with their dominant leg (the leg used for kicking a ball (Begue et al., 2021)) and to take a second step with their non-dominant leg to stop in a comfortable upright posture with feet side-by-side. They were instructed to perform stepping movement at their preferred speed. After each trial, the participants repositioned themselves in the standardized feet position (McIlroy and Maki, 1997) marked on the ground. During the experiment, a total of 49 retro-reflective spherical markers (14 mm diameter) were fixed on bony landmarks of all participants in order to track the motion of 15 rigid body segments: pelvis, torso, head, right and left thighs, shanks, feet, toes, arms, forearms and hands (Begue et al., 2021). A motion capture system equipped with 14 cameras (Vicon, UK) was used to collected kinematic data at 200 Hz. Data acquisition was triggered when participants were motionless and at least 1 s before the verbal signal. After familiarization trials, each individual performed 20 trials.

2.3. Data analysis

Kinematic data were low-pass filtered using a zero-lag fourth-order Butterworth filter with a 6 Hz cut-off frequency. The optimal cut-off frequency was determined from a residual analysis (Winter, 2009).

2.3.1. Spatiotemporal parameters of stepping

Several temporal events were identified to calculate our dependent variables. The onset of the stepping movement was detected when the anteroposterior or mediolateral accelerations of the CoM went above 2 standard deviations (SD) from its baseline value (Caderby et al., 2017). The end of the stepping movement was defined as the time-point at which the mediolateral CoM velocity returned below the mean + 2 SD calculated during quiet standing after the end of the stepping (Singer et al., 2014). In addition to these parameters, stepping duration, forward progression velocity (peak anteroposterior CoM velocity during stepping) and spatial parameters, such as length and width of the first step (dominant leg) and the second step (non-dominant leg), were computed (for details, see Begue et al., 2021).

2.3.2. Whole-body and segment angular momenta computation

The whole-body angular momentum about the body's CoM position was calculated in the three dimensions using a 15-segment rigid body model of the whole body, as previously described (Maldonado et al., 2018; Begue et al., 2021). Briefly, the whole-body geometric model and lower limb, pelvis and upper limb anthropometry are based on the running model of Hamner et al. (2010). Mass properties of the hands were estimated from the regression equations of de Leva (1996). The anthropometric description of the torso and head segments were estimated from Dumas et al. (2007). Kinematics information was retrieved by inverse kinematics, using a multibody kinematic optimization approach in OpenSim software. The resulting kinematics data and the inertial parameters were exported to a custom-made Matlab program for calculation of the angular momentum. WBAM was calculated from the following equation:

$$\boldsymbol{WBAM} = \sum_{j=1}^{n} \left[\left(\boldsymbol{r_j} - \boldsymbol{r_{CoM}} \right) \times m_j \left(\boldsymbol{v_j} - \boldsymbol{v_{CoM}} \right) + I_j \boldsymbol{\omega_j} \right] \tag{1}$$

where n is the number of segments (n = 15); $\boldsymbol{r_j}$ and $\boldsymbol{r_{CoM}}$ are the vector positions of the CoM of the body segment j and the whole-body CoM (the weighted sum of each body segment's CoM), respectively, v_j and v_{CoM} are the vector velocities of the body segment j and the whole-body, respectively, m_j and I_j are the mass and inertia tensor of body segment j, respectively, and ω_j is the angular velocity vector of the body segment j. WBAM was expressed in a coordinate system (orthogonal coordinate system), in which the anteroposterior axis (X axis) was directed forward, the vertical axis (Y) was directed upward and the mediolateral axis (Z) was to the right of the subject. In order to decrease between-subject variability and provide non-dimensional measure WBAM was normalized by participants' height (l), mass and $\sqrt{g \cdot l}$ ($g = 9.81$ m·s^{-2}) (Begue et al., 2019; Vistamehr et al., 2016). The term $\sqrt{g \cdot l}$ (in m.s^{-1}) provides a normalization method similar to the concept of Froude number and independent of the progression velocity (Vistamehr et al., 2016; Vistamehr et al., 2018). Furthermore, we also calculated the angular momenta of five segment groups, all expressed at the body's CoM: right arm (upper arm, forearm, hand), left arm, right leg (thigh, shank, foot), left leg, and trunk (head, torso, pelvis). WBAM and segment angular momenta over the entire stepping movement were time normalized from 0 to 100 %, with 0 % being the onset of stepping movement and 100 % the end of stepping movement. Peak-to-peak ranges of WBAM in the three dimensions, determined as the difference between maximum and minimum values of WBAM, were also computed over the entire duration of the stepping movement.

2.3.3. Uncontrolled manifold analysis

The UCM approach was used to determine the control of WBAM during stepping. The performance variable in this study was WBAM and the elemental variables were the angular momenta of the five segments (right and left arms, right and left legs, and trunk). We purposely chose to retain these five segments for analysis by assuming that their angular momenta were independent of each other, which is a requirement of the UCM analysis. Nevertheless, we reported the results of the UCM analysis conducted with the angular momenta of the fifteen body segments in the supplementary material (Appendix A). The UCM analysis was conducted in the three planes (sagittal, frontal and transversal planes) separately. In each plane, the relation between the performance variable and the elemental variables was determined from the linear relation between WBAM and the segment angular momenta (SAM):

$$\boldsymbol{WBAM} = \boldsymbol{SAM}_{RightArm} + \boldsymbol{SAM}_{LeftArm} + \boldsymbol{SAM}_{RightLeg} + \boldsymbol{SAM}_{LeftLeg} + \boldsymbol{SAM}_{Trunk} \quad (2)$$

From this relation, it is possible to express the changes in the performance variable (WBAM) as a function of changes in the elemental variables (SAM) at each normalized time:

$$\Delta WBAM = \boldsymbol{J} \cdot \boldsymbol{\Delta SAM} \quad (3)$$

where $\Delta WBAM$ (a scalar) and $\boldsymbol{\Delta SAM}$ (a 5×1 vector) are the changes in WBAM and the changes in the angular momenta of the five segments, respectively. $\Delta WBAM$ and $\boldsymbol{\Delta SAM}$ were computed across trials at a given time point and relative to the average WBAM and average segment angular momenta, respectively. $\boldsymbol{J}$ is a 1×5 vector containing the partial derivatives of the performance variable with respect to the elemental variables:

$$J = \left[\frac{\partial WBAM}{\partial SAM_{RightArm}} \; \frac{\partial WBAM}{\partial SAM_{LeftArm}} \; \frac{\partial WBAM}{\partial SAM_{RightLeg}} \; \frac{\partial WBAM}{\partial SAM_{LeftLeg}} \; \frac{\partial WBAM}{\partial SAM_{Trunk}} \right] \quad (4)$$

According to Eq. (4), it is noteworthy that $\boldsymbol{J}$ is a 1-by-5 vector of ones in the present study. A singular value decomposition was then performed on matrix $\boldsymbol{J}$ to identify the vectors associated with the null space (singular values = 0; UCM subspace, $\boldsymbol{\varepsilon_N}$) and the space orthogonal to it (singular value ≠ 0, ORT subspace, $\boldsymbol{\varepsilon_O}$). The variances (per degree of freedom) associated with the UCM and ORT subspaces, i.e., V_{UCM} and V_{ORT}, respectively, were computed at each normalized time as:

$$V_{UCM} = \frac{\sum_{i=1}^{N} \left\| \varepsilon_N{}^T \cdot \boldsymbol{\Delta SAM}_i \right\|^2}{(n-d) \cdot N} \quad (5)$$

and

$$V_{ORT} = \frac{\sum_{i=1}^{N} \left\| \varepsilon_O{}^T \cdot \boldsymbol{\Delta SAM}_i \right\|^2}{d \cdot N} \quad (6)$$

With $\boldsymbol{\varepsilon}_N^T$ and $\boldsymbol{\varepsilon}_O^T$ the transposed matrices of $\boldsymbol{\varepsilon_N}$ and $\boldsymbol{\varepsilon_O}$, respectively. $\boldsymbol{\varepsilon_N}$ is the null space matrix (a 4-by-5 matrix) and $\boldsymbol{\varepsilon_O}$ the vector of space orthogonal to it (a 1-by-5 matrix). N is the number of trials. n is the number of elemental degrees of freedom ($n = 5$) and d is the dimension of the orthogonal manifold ($d = 1$).

Afterwards, the synergy index (SI) was computed as:

$$SI = \frac{V_{UCM} - V_{ORT}}{V_{UCM} + V_{ORT}} \quad (7)$$

Note that this synergy index has the advantage of having symmetrical bounds ($-1 < \text{SI} < +1$), unlike other synergy indexes, such as the one based on the total amount of variance (Robert et al., 2009; Tillman and Ambike, 2018). Nevertheless, as this latter index is classically used in the literature, we present the results obtained for it in Appendix B.

Before statistical analysis, synergy index was transformed using a Fisher's Z-transformation (Robert et al., 2009; Verrel, 2010):

$$SI_Z = \frac{1}{2} \times log\left[\frac{1+SI}{1-SI}\right] \quad (8)$$

Note that $SI = 0$ translates to $SI_Z = 0$. Therefore, $SI_Z > 0$ indicates the presence of a synergy among the segment angular momenta stabilizing WBAM. Conversely, $SI_Z < 0$ indicates the presence of an anti-synergy destabilizing WBAM (Robert et al., 2009). $SI_Z = 0$ indicates the absence of synergy (or control).

2.4. Statistical analysis

Synergy index (SI_Z) and variances (V_{UCM} and V_{ORT}) over the entire stepping movement were time normalized from 0 to100%. Then, synergy index and variance curves were temporally aligned from the continuous curve registration method (Ramsay et al., 2009) using the open-source Functional Data Analysis Matlab Toolbox (www.functionaldata.org). This method aims to reduce timing variability between the participants and, therefore, allows for time-continuous analyses to provide outcomes based only on magnitude-related differences, rather than changes in timing (Honert and Pataky, 2021). Briefly, this method consists in transforming time series data into functions that are aligned using dynamic time wrapping functions, thus reducing phase variations between subjects while preserving the curves shape and amplitude (Ramsay et al., 2009). Afterwards, Statistical Parametric Mapping (SPM) was used to evaluate the aging effect on waveforms of synergy index and variances. SPM analysis was conducted by implementing the functions from the open-source spm1D package (www.spm1d.org) in a custom-made Matlab program (MathWorks, Inc., Natick, USA). Normality of the waveforms was beforehand assessed using the normality function of SPM for a two-sample t-test. Depending on data normality, parametric two-sample t-tests (SPM{t}, $\alpha = 0.05$) or non-parametric two-sample t-tests (SnPM{t}, $\alpha = 0.05$) were conducted independently on the waveforms of dependent variables. For all non-parametric t-tests, the number of iterations was set at 10,000 (Pataky et al., 2015). For each t-test, the critical t-statistic and areas of significance between the two groups were reported. Finally, the effect of aging on the temporo-spatial parameters of stepping, ranges of WBAM, mean synergy indexes, mean variances in the three planes were tested using t-tests for independent samples, after checking for data normality and homoscedasticity. Relationship between range of WBAM and mean synergy index in each plane were assessed using Pearson correlation coefficient. The level of statistical significance was set at $p < 0.05$.

3. Results

3.1. Spatiotemporal parameters

There was no significant difference in the temporal-spatial features of stepping movement between young and old participants ($p > 0.05$; Table 1).

3.2. Whole-body angular momentum

Statistical analysis revealed a significant larger WBAM range in the sagittal plane in older adults compared to young adults ($p = 0.01$; Fig. 1; Table 2). In contrast, no significant difference was found between the two age groups in both the fontal and transversal planes ($p > 0.05$).

3.3. Variances

Regarding the mean variances, statistical analysis revealed that older adults exhibited significant larger UCM and orthogonal variances in all planes compared to young adults (Table 2).

SPM analysis revealed significant differences in UCM variance between the two age groups in the three planes (Fig. 2). In the sagittal plane, old adults had higher UCM variance during the first double support phase (14–16 %; $p = 0.014$) and during the first single support phase (23–28 %; $p = 0.001$) compared to young adults. In the frontal plane, old adults had higher UCM variance during the second double support phase (34–39 %; $p = 0.001$) compared to young adults. In the transversal plane, old adults had higher UCM variance during the second double support phase (34–37 %; $p = 0.008$) compared to young adults.

Regarding the orthogonal variance, SPM analysis revealed significant differences between the two age groups in the three planes (Fig. 2). In the sagittal plane, SPM revealed that old adults exhibited higher orthogonal variance values during the first double support phase (13–16 %; $p = 0.004$), during the first single support phase (21–25 %; $p = 0.001$), during the second double support phase (33–37 %; $p = 0.006$) and during the restabilisation phase (85–91 %; $p = 0.005$) compared to young adults. In the frontal plane, old adults had higher orthogonal variance during the second double support phase (35–39 %; $p = 0.007$) compared to young adults. In the transversal plane, old adults had higher orthogonal variance during the first single support phase (22–25 %; $p = 0.008$).

3.4. Synergy index

Regarding mean synergy indexes, statistical analysis revealed a significant smaller mean synergy index in the sagittal plane in older adults compared to young adults ($p = 0.017$; Table 1). In contrast, there was no significant difference between the two age groups for the mean synergy indexes in the fontal and transversal planes ($p > 0.05$).

SPM analysis revealed that old adults had smaller synergy indexes in the sagittal plane during the restabilisation phase at the end of movement (86–100 %; $p < 0.001$) compared to young adults (Fig. 3).

A significant negative correlation was found between the mean index synergy and the WBAM range in the frontal plane ($r = -0.77$; $p < 0.001$). In contrast, there was no significant correlation between the mean index synergy and the WBAM range in the sagittal and transversal planes ($p > 0.05$).

Table 1
Mean and standard deviation of the temporo-spatial features of stepping movement for young and old adults.

Parameters	Young	Old	p-Value
Stepping duration (s)	3.2 ± 0.2	3.3 ± 0.3	NS
Forward progression velocity (m·s^{-1})	0.68 ± 0.16	0.71 ± 0.13	NS
Length of 1st step (m)	0.57 ± 0.11	0.60 ± 0.09	NS
Width of 1st step (m)	0.16 ± 0.02	0.17 ± 0.03	NS
Length of 2nd step (m)	0.60 ± 0.13	0.65 ± 0.09	NS
Width of 2nd step (m)	0.17 ± 0.03	0.15 ± 0.04	NS

NS: Non significant ($p > 0.05$).

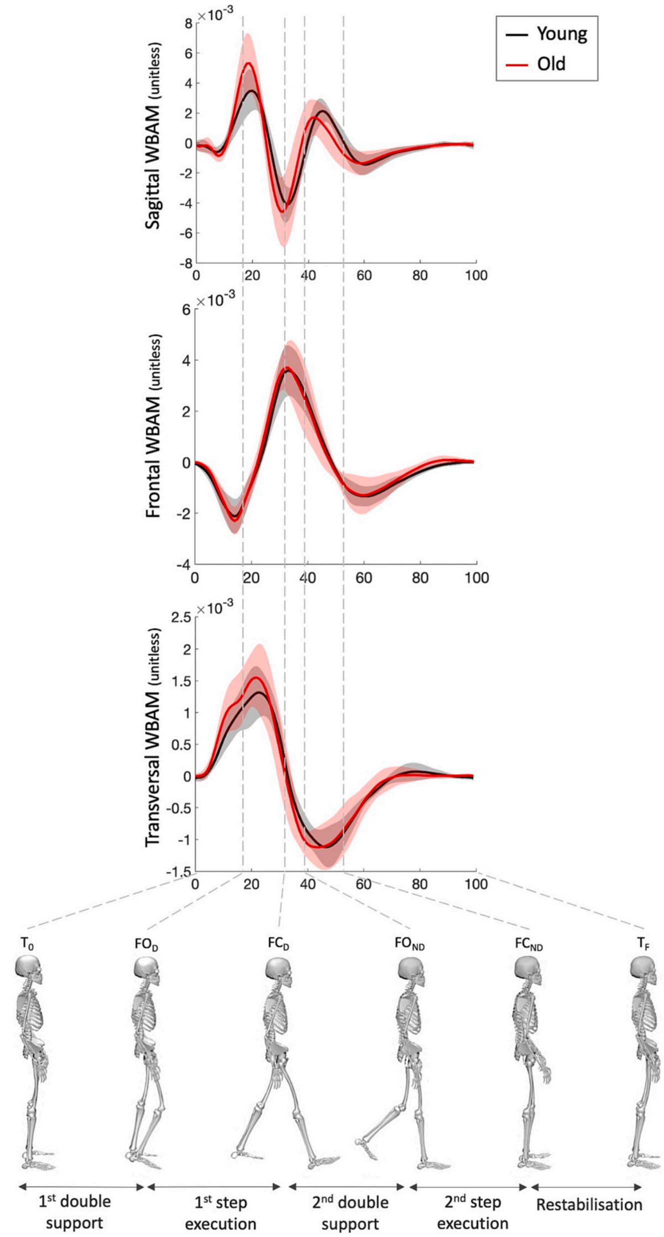

Fig. 1. Graphical representation of the mean whole-body angular momentum (WBAM) for young (black) and old adults (red) in the three planes during stepping. Dimensionless WBAM was normalized by the participant's height (m), mass (kg) and $\sqrt{g \cdot l}$ ($g = 9.81$ m·s^{-2} and l = participant's height (m)) T_0: onset of stepping movement. FO_D: foot-off of the dominant leg. FC_D: foot-contact of the dominant leg. FO_{ND}: foot-off of the non-dominant leg. FC_{ND}: foot-contact of the non-dominant leg. T_F: end of the stepping movement.

4. Discussion

To our knowledge, this is the first study to examine the control of WBAM during stepping in young and older adults using the UCM approach. Previously, the UCM framework has been used to investigate the control of WBAM during motor tasks, such as pointing (Greve et al.,

Table 2
Mean and standard deviation of the ranges of dimensionless WBAM, mean synergy indexes and mean variances in the three planes for young and old adults. Note that all parameters are unitless.

Parameters	Young	Old	p-Value
Whole body angular momentum (WBAM)			
Sagittal WBAM range ($\times 10^{-3}$)	11.5 ± 1.9	14.9 ± 3.9	**p = 0.01**
Frontal WBAM range ($\times 10^{-3}$)	7.3 ± 1.9	8.2 ± 1.5	NS
Transversal WBAM range ($\times 10^{-3}$)	3.1 ± 0.7	3.6 ± 0.7	NS
Mean UCM variance (VUCM)			
Sagittal V_{UCM} ($\times 10^{-5}$)	0.075 ± 0.025	0.114 ± 0.038	**p = 0.005**
Frontal V_{UCM} ($\times 10^{-7}$)	0.515 ± 0.140	0.727 ± 0.197	**p = 0.005**
Transversal V_{UCM} ($\times 10^{-7}$)	0.318 ± 0.144	0.461 ± 0.161	**p = 0.026**
Mean orthogonal variance (VORT)			
Sagittal V_{ORT} ($\times 10^{-5}$)	0.060 ± 0.018	0.102 ± 0.050	**p = 0.010**
Frontal V_{ORT} ($\times 10^{-7}$)	1.363 ± 0.789	2.149 ± 0.891	**p = 0.027**
Transversal V_{ORT} ($\times 10^{-7}$)	0.270 ± 0.120	0.417 ± 0.181	**p = 0.025**
Mean synergy index (SIZ)			
Sagittal SI_Z	−0.07 ± 0.09	−0.15 ± 0.07	**p = 0.017**
Frontal SI_Z	−0.46 ± 0.15	−0.56 ± 0.10	NS
Transversal SI_Z	−0.04 ± 0.11	−0.08 ± 0.13	NS

NS: Non significant ($p > 0.05$).

2013), sit-to-stand (Reisman et al., 2002), gait (Robert et al., 2009) and jump (Maldonado et al., 2018) tasks. Overall, these earlier studies brought the evidence that WBAM was a variable controlled by humans. During walking, Robert et al. (2009) revealed that the nature of the control of WBAM changed between the phases of gait cycle (double support and swing phases) and the planes (sagittal, frontal and transversal). In agreement with this latter study, we observed that the control of WBAM differed across the phases of movement and across the planes during stepping. Specifically, our results showed that sagittal-plane WBAM was stabilized (positive synergy index) at the end of the first double support phase, the onset of the first single support phase, and during the second double support and second single support phases, while it was negative during the other parts of the stepping movement (Fig. 3). In the transversal plane, WBAM was stabilized during the two single support phases, while it was negative in all double support phases (including the restabilisation phase). Finally, in the frontal plane, it was negative during the entire stepping movement. These negative values of the synergy index, which indicate that the orthogonal variance is greater than the UCM variance (Fig. 3), are often referred to as “anti-synergies” (Robert et al., 2009). They may be interpreted as a co-variation in the elemental variables contributing to destabilize the performance variable, i.e., to make it purposefully vary across task repetitions. This specific organization among the elemental variables suggests a control of the performance variable. Indeed, in the case of a random process, the variance is expected to be spread randomly between the two subspaces (UCM and orthogonal) resulting in a synergy index close to zero. In the literature, it was initially reported that anti-synergies appeared when a quick change in the performance variable was required during the motor task, e.g., a rapid change in finger force production or a quick displacement in the center of pressure under the feet while standing (Olafsdottir et al., 2005; Shim et al., 2005; Wang et al., 2006). However, during walking, Robert et al. (2009) noted that these anti-synergies were observed at specific phases of the gait cycle during which the performance variable (WBAM) did not vary rapidly. These authors instead interpreted these anti-synergies as a need to make adjustments of WBAM from stride to stride. Consistent with this interpretation, other authors have shown that anti-synergies may be aimed at correcting undesired and unintended deviations in the performance variable across task repetitions in order to achieve a known explicit goal, such as maintaining a similar horizontal body position during hopping (Yen et al., 2009) or maintaining a constant gait speed during treadmill walking (Toney and Chang, 2013). Taken together, these results indicate that these anti-synergies may be implemented by the controller either to achieve a quick change in the performance variable or to make adjustments (corrections) to that variable across tasks repetitions.

Our results showed that despite greater orthogonal and UCM variances (inter-trial variances), the synergy index did not differ between both age groups in both the frontal and transversal planes, indicating that WBAM was controlled in the old adults to the same extent as young adults. Results in these two planes are in line with those of Cui et al. (2020), who found that normal aging did not affect the control of the net external moment about the body's CoM (the equivalent of the time rate of WBAM) during double support phase in curb negotiation. In contrast with this previous study, we found that the synergy index in the sagittal plane is affected by age. Compared to their younger counterparts, older adults had a larger negative value of the synergy index during the restabilisation phase of stepping, which led to the larger negative value of the mean index synergy over the entire stepping in old adults. These larger negative synergy index values do not support our first hypothesis that aging is associated with less control of the WBAM in the sagittal plane during stepping. Indeed, the more the value of the synergy index deviates from 0 (positive or negative), the more the performance variable would be controlled. Specifically, we noted that the larger negative value of the synergy index in the restabilisation phase in old adults was ascribed to a higher orthogonal variance compared to young adults, while the UCM variance was not different between the two groups (Fig. 2). These results are in line with other studies that found an increased orthogonal variance and decreased synergy index in aging adults during balance recovery (Hsu et al., 2013), narrow base standing (Hsu et al., 2014) and treadmill walking (Decker et al., 2012). Overall, these previous studies interpreted these results as a reduced motor flexibility with aging. In the present study, our findings that the UCM variance did not differ between both groups suggest that there were no alterations in the use of motor redundancy in old adults, i.e., they exhibited a number of motor solutions relative to the UCM space similar to young adults. Rather, as the synergy index was negative during the restabilisation phase, the increased orthogonal variance suggests that old adults employed stronger WBAM-destabilizing synergies (i.e., anti-synergies) during this phase. During the restabilisation phase, WBAM must be restored to a value close to zero in order to stop the forward body movement and ensure a quasi-static standing posture. In particular, our results revealed that the age-related differences in the synergy index were specifically found at the end of the restabilisation phase (83–100 % of the stepping movement), i.e., when WBAM had already almost returned to zero (Fig. 1). This suggests that the anti-synergies implemented at these specific instants were not intended to achieve a quick change in the WBAM, but rather they were associated with the need to make adjustments of WBAM across trials. These trial-to-trial adjustments may be specifically aimed at counteracting unintentional deviations in WBAM in order to achieve a known explicit goal, which is to restore a quasi-static upright posture. Thus, the stronger anti-synergies during the restabilisation phase in older participants may reveal a greater need to correct deviations (or errors) in WBAM across trials in order to achieve a quasi-static standing posture. This may be viewed as a tighter control of WBAM (balance) in old adults compared to their younger counterparts, to ensure a successful and safe stepping termination. This is consistent with findings from previous studies that old adults have a greater difficulty to arrest the forward body motion during gait termination (Tirosh and Sparrow, 2004) and therefore adopt a more cautious strategy to ensure task success and mitigate risk of falling (Rum et al., 2017). Still, the present results should be considered

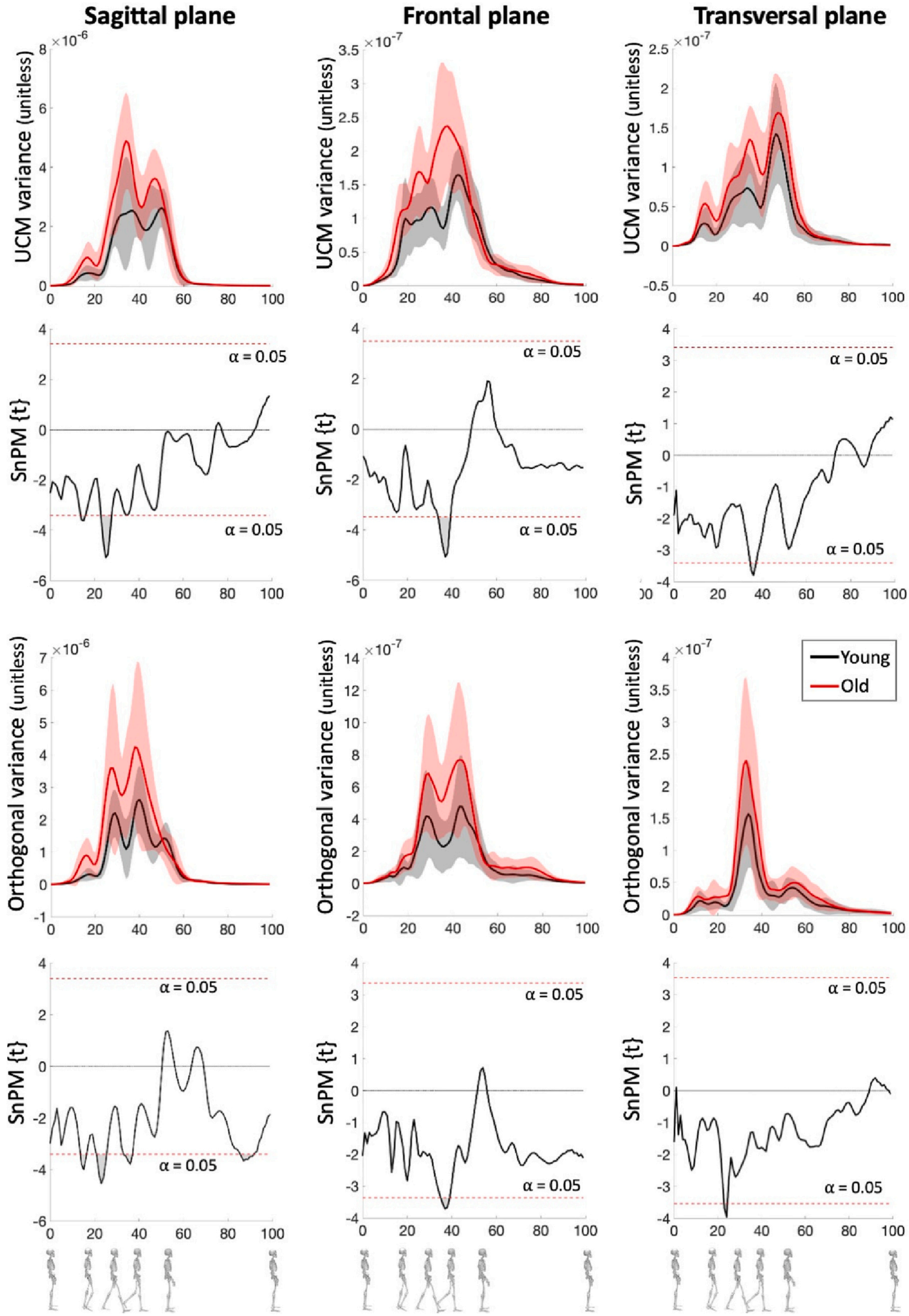

Fig. 2. Significant differences in UCM and orthogonal variances in the three planes between young adults (black) and old adults (red) using spm1D. SPM{t} was used for normally distributed waveforms and SnPM{t} for not normally distributed waveforms. The shaded area within the waveforms represents the area where variance is significantly different (i.e., in which the SPM{t} and SnPM{t} exceeds the critical threshold).

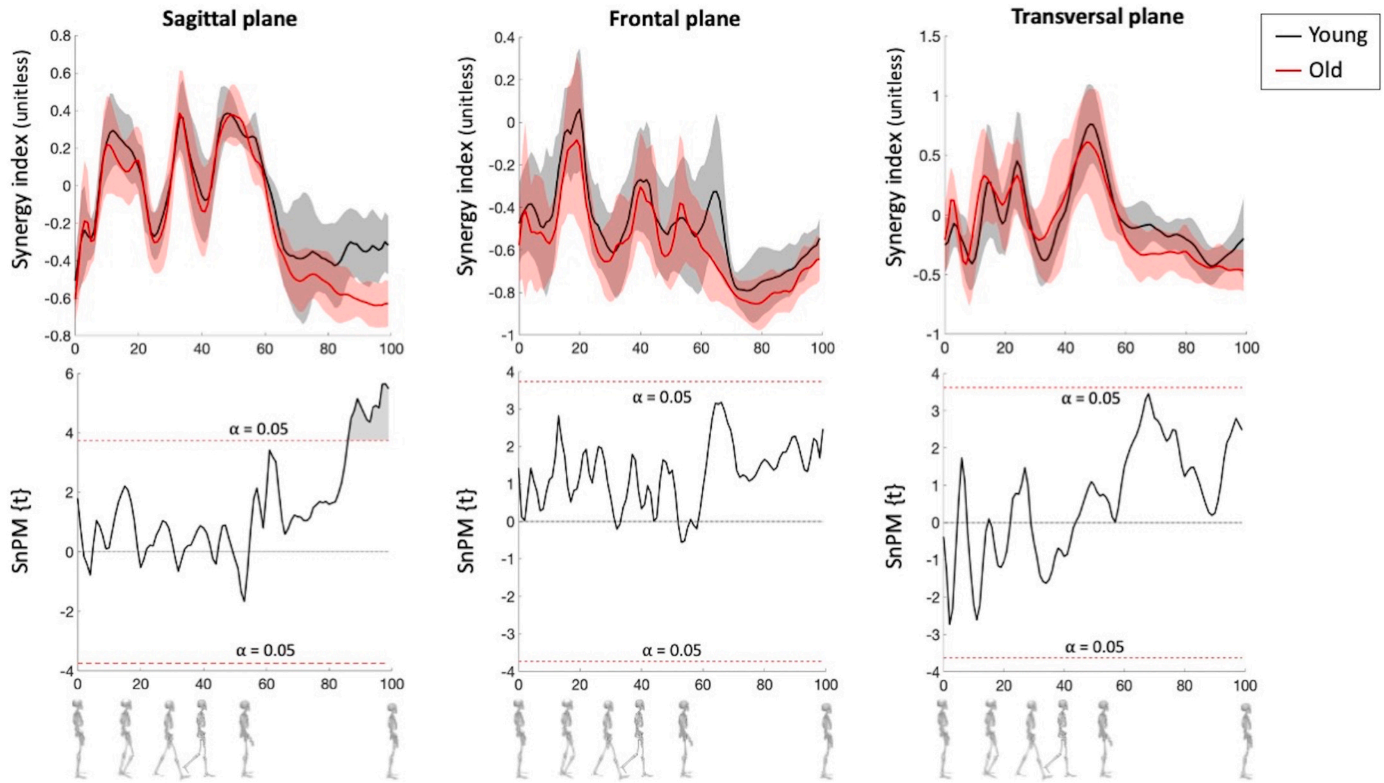

Fig. 3. Significant differences in synergy index in the three planes between young adults (black) and old adults (red) using spm1D. SPM{t} was used for normally distributed waveforms and SnPM {t} for not normally distributed waveforms. The shaded area within the waveforms represents the area where synergy index is significantly different (i.e., in which the SPM{t} and SnPM{t} exceeds the critical threshold).

with caution as both angular momentum and the associated variabilities were very low during the restabilisation phase, leading to a less accurate estimate of the synergies. Further investigation is required to confirm these results.

In agreement with our previous study (Begue et al., 2021), our results showed that old adults had a higher range of sagittal-plane WBAM compared to young adults, while there was no difference between both groups in the other two planes. This increased range of sagittal-plane WBAM in old adults, which has been shown to be the result of changes in trunk and leg rotational dynamics and less segment-to-segment angular momentum cancellation (Begue et al., 2021), poses a greater challenge for balance maintenance during stepping. In the current study, we tested the hypothesis that this larger WBAM range in older adults would be associated to a less control of this variable with age. The results obtained for the synergy index, discussed above, do not support this assumption. Consistently, we noted that there was no significant correlation between the WBAM range and the mean synergy index in the sagittal plane. Taken together, these results suggest that the age-related changes in WBAM during stepping are not related to alterations in the ability to control this variable with aging. Consequently, the age-related differences in WBAM during stepping could instead result from an adaptive strategy aimed to compensate for leg muscle weakness, as previously hypothesized (Begue et al., 2021). Nevertheless, we found a negative correlation ($r = -0.77$) between mean synergy index and WBAM range in the frontal plane during stepping. The synergy index in this plane being negative during the entire stepping movement, these finding suggest that WBAM in the frontal plane was adjusted during the entire stepping movement, and consequently changes in the nature or degree of its control (increase or decrease in synergy index) lead to alterations in WBAM range and potentially affect balance maintenance in this plane. This is consistent with the idea that maintaining balance during walking requires an active control in the frontal plane, whereas this could be done either passively or by very low-level control in the sagittal plane (Donelan et al., 2004).

It should be acknowledged that the present study contains some limitations. A main limitation was that the old participants were healthy and also physically active, i.e., engaging in >150 min of moderate-intensity physical activity per week. This probably explains the absence of major differences in the control of WBAM between young and old adults. Interestingly, previous studies showed that old adult fallers exhibited a significantly higher UCM variance associated with swing foot placement (Yamagata et al., 2019) and a larger orthogonal variance associated with the vertical CoM position (Yamagata et al., 2021) during walking compared to non-fallers. Thus, it would be interesting to study the control of WBAM in more frail old adults with a higher risk of falling. Furthermore, another perspective would be to study what are the factors involved in the age-related alterations in the stabilization of WBAM during the restabilisation phase of stepping, e.g., peripheral (muscle weakness, loss of flexibility, etc.) and central (anxiety) factors.

To conclude, the present study stresses that normal aging did not affect the control of WBAM in both the frontal and transversal planes during stepping. In contrast, old adults exhibited a larger negative synergy index in the sagittal plane during the restabilisation phase compared to young adults, suggesting an increased control of this variable during this phase with aging. Furthermore, we observed that the decrease of the synergy index was not related to the higher range of sagittal-plane WBAM in old adults. Taken together, these results suggest that the age-related changes in sagittal-plane WBAM during stepping are not ascribed to poorer control of this mechanical variable in aging adults. Nevertheless, the age-related differences in the synergy index observed during the restabilisation phase should be considered with caution as both angular momentum and the associated variabilities are very low during this phase, leading to a less accurate estimate of the synergies. Further investigation is therefore required to confirm these results.

Funding

This work was supported by a grant from the University of La Reunion.

CRediT authorship contribution statement

Teddy Caderby: Conceptualization, Methodology, Writing – original draft, Funding acquisition. **Angélique Lesport:** Conceptualization, Investigation, Methodology, Formal analysis, Writing – review & editing. **Nicolas A. Turpin:** Methodology, Formal analysis, Writing – review & editing. **Georges Dalleau:** Methodology, Writing – review & editing, Funding acquisition. **Bruno Watier:** Methodology, Writing – review & editing. **Thomas Robert:** Methodology, Writing – review & editing. **Nicolas Peyrot:** Conceptualization, Writing – review & editing, Funding acquisition. **Jérémie Begue:** Conceptualization, Investigation, Methodology, Formal analysis, Writing – review & editing.

Declaration of competing interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Acknowledgments

The authors thank Lorie Eclogrin for his technical support.

Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.exger.2023.112217.

References

Begue, J., Peyrot, N., Dalleau, G., Caderby, T., 2019. Age-related changes in the control of whole-body angular momentum during stepping. Exp. Gerontol. 127, 110714 https://doi.org/10.1016/j.exger.2019.110714.

Begue, J., Peyrot, N., Lesport, A., Turpin, N.A., Watier, B., Dalleau, G., et al., 2021. Segmental contribution to whole-body angular momentum during stepping in healthy young and old adults. Sci. Rep. 11 (1), 19969. https://doi.org/10.1038/s41598-021-99519-y.

Bennett, B.C., Russell, S.D., Sheth, P., Abel, M.F., 2010. Angular momentum of walking at different speeds. Hum. Mov. Sci. 29 (1), 114–124. https://doi.org/10.1016/j.humov.2009.07.011.

Caderby, T., Yiou, E., Peyrot, N., de Vivies, X., Bonazzi, B., Dalleau, G., 2017. Effects of changing body weight distribution on mediolateral stability control during gait initiation. Front. Hum. Neurosci. 11, 127. https://doi.org/10.3389/fnhum.2017.00127.

Cui, C., Kulkarni, A., Rietdyk, S., Barbieri, F.A., Ambike, S., 2020. Synergies in the ground reaction forces and moments during double support in curb negotiation in young and older adults. J. Biomech. 106 https://doi.org/10.1016/j.jbiomech.2020.109837.

de Leva, P., 1996. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. J. Biomech. 29 (9), 1223–1230. https://doi.org/10.1016/0021-9290(95)00178-6.

Decker, L.M., Cignetti, F., Potter, J.F., Studenski, S.A., Stergiou, N., 2012. Use of motor abundance in young and older adults during dual-task treadmill walking. PLoS One 7 (7), e41306. https://doi.org/10.1371/journal.pone.0041306.

Donelan, J.M., Shipman, D.W., Kram, R., Kuo, A.D., 2004. Mechanical and metabolic requirements for active lateral stabilization in human walking. J. Biomech. 37 (6), 827–835. https://doi.org/10.1016/j.jbiomech.2003.06.002.

Dumas, R., Chèze, L., Verriest, J.P., 2007. Adjustments to McConville et al. and Young et al. body segment inertial parameters. J. Biomech. 40 (3), 543–553. https://doi.org/10.1016/j.jbiomech.2006.02.013.

Greve, C., Zijlstra, W., Hortobagyi, T., Bongers, R.M., 2013. Not all is lost: old adults retain flexibility in motor behaviour during sit-to-stand. PLoS One 8 (10), e77760. https://doi.org/10.1371/journal.pone.0077760.

Hamner, S.R., Seth, A., Delp, S.L., 2010. Muscle contributions to propulsion and support during running. J. Biomech. 43 (14), 2709–2716. https://doi.org/10.1016/j.jbiomech.2010.06.025.

Herr, H., Popovic, M., 2008. Angular momentum in human walking. J. Exp. Biol. 211 (Pt 4), 467–481. https://doi.org/10.1242/jeb.008573.

Honert, E.C., Pataky, T.C., 2021. Timing of gait events affects whole trajectory analyses: a statistical parametric mapping sensitivity analysis of lower limb biomechanics. J. Biomech. 119, 110329 https://doi.org/10.1016/j.jbiomech.2021.110329.

Hsu, W.L., Chou, L.S., Woollacott, M., 2013. Age-related changes in joint coordination during balance recovery. Age (Dordr.) 35 (4), 1299–1309. https://doi.org/10.1007/s11357-012-9422-x.

Hsu, W.L., Lin, K.H., Yang, R.S., Cheng, C.H., 2014. Use of motor abundance in old adults in the regulation of a narrow-based stance. Eur. J. Appl. Physiol. 114 (2), 261–271. https://doi.org/10.1007/s00421-013-2768-7.

Krishnan, V., Rosenblatt, N.J., Latash, M.L., Grabiner, M.D., 2013. The effects of age on stabilization of the mediolateral trajectory of the swing foot. Gait Posture 38 (4), 923–928. https://doi.org/10.1016/j.gaitpost.2013.04.023.

Latash, M.L., Scholz, J.P., Schoner, G., 2007. Toward a new theory of motor synergies. Mot. Control. 11 (3), 276–308. https://doi.org/10.1123/mcj.11.3.276.

Maldonado, G., Bailly, F., Soueres, P., Watier, B., 2018. On the coordination of highly dynamic human movements: an extension of the uncontrolled manifold approach applied to precision jump in parkour. Sci. Rep. 8 (1), 12219. https://doi.org/10.1038/s41598-018-30681-6.

McIlroy, W.E., Maki, B.E., 1997. Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. Clin. Biomech. 12 (1), 66–70. Bristol, Avon. doi: S026800339600040X [pii].

Nott, C.R., Neptune, R.R., Kautz, S.A., 2014. Relationships between frontal-plane angular momentum and clinical balance measures during post-stroke hemiparetic walking. Gait Posture 39 (1), 129–134. https://doi.org/10.1016/j.gaitpost.2013.06.008.

Olafsdottir, H., Yoshida, N., Zatsiorsky, V.M., Latash, M.L., 2005. Anticipatory covariation of finger forces during self-paced and reaction time force production. Neurosci. Lett. 381 (1–2), 92–96. https://doi.org/10.1016/j.neulet.2005.02.003.

Pataky, T.C., Vanrenterghem, J., Robinson, M.A., 2015. Zero- vs. one-dimensional, parametric vs. non-parametric, and confidence interval vs. hypothesis testing procedures in one-dimensional biomechanical trajectory analysis. J. Biomech. 48 (7), 1277–1285. https://doi.org/10.1016/j.jbiomech.2015.02.051.

Pickle, N.T., Wilken, J.M., Aldridge, J.M., Neptune, R.R., Silverman, A.K., 2014. Whole-body angular momentum during stair walking using passive and powered lower-limb prostheses. J. Biomech. 47 (13), 3380–3389. https://doi.org/10.1016/j.jbiomech.2014.08.001.

Pickle, N.T., Wilken, J.M., Aldridge Whitehead, J.M., Silverman, A.K., 2016. Whole-body angular momentum during sloped walking using passive and powered lower-limb prostheses. J. Biomech. 49 (14), 3397–3406. https://doi.org/10.1016/j.jbiomech.2016.09.010.

Pijnappels, M., Bobbert, M.F., van Dieen, J.H., 2005. Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait Posture 21 (4), 388–394. https://doi.org/10.1016/j.gaitpost.2004.04.009.

Ramsay, J.O., Hooker, G., Graves, S., 2009. Registration: aligning features for samples of curves. In: Functional Data Analysis With R And MATLAB, pp. 117–130.

Reisman, D.S., Scholz, J.P., Schoner, G., 2002. Coordination underlying the control of whole body momentum during sit-to-stand. Gait Posture 15 (1), 45–55. https://doi.org/10.1016/s0966-6362(01)00158-8.

Robert, T., Bennett, B.C., Russell, S.D., Zirker, C.A., Abel, M.F., 2009. Angular momentum synergies during walking. Exp. Brain Res. 197 (2), 185–197. https://doi.org/10.1007/s00221-009-1904-4.

Rum, L., Laudani, L., Macaluso, A., Vannozzi, G., 2017. Upper body accelerations during planned gait termination in young and older women. J. Biomech. 65, 138–144. https://doi.org/10.1016/j.jbiomech.2017.10.019.

Shim, J.K., Olafsdottir, H., Zatsiorsky, V.M., Latash, M.L., 2005. The emergence and disappearance of multi-digit synergies during force-production tasks. Exp. Brain Res. 164 (2), 260–270. https://doi.org/10.1007/s00221-005-2248-3.

Silverman, A.K., Neptune, R.R., 2011. Differences in whole-body angular momentum between below-knee amputees and non-amputees across walking speeds. J. Biomech. 44 (3), 379–385. https://doi.org/10.1016/j.jbiomech.2010.10.027.

Singer, J.C., McIlroy, W.E., Prentice, S.D., 2014. Kinetic measures of restabilisation during volitional stepping reveal age-related alterations in the control of mediolateral dynamic stability. J. Biomech. 47 (14), 3539–3545. https://doi.org/10.1016/j.jbiomech.2014.08.022.

Tillman, M., Ambike, S., 2018. Cue-induced changes in the stability of finger force-production tasks revealed by the uncontrolled manifold analysis. J. Neurophysiol. 119 (1), 21–32. https://doi.org/10.1152/jn.00519.2017.

Tirosh, O., Sparrow, W.A., 2004. Gait termination in young and older adults: effects of stopping stimulus probability and stimulus delay. Gait Posture 19 (3), 243–251. https://doi.org/10.1016/S0966-6362(03)00063-8.

Toney, M.E., Chang, Y.H., 2013. Humans robustly adhere to dynamic walking principles by harnessing motor abundance to control forces. Exp. Brain Res. 231 (4), 433–443. https://doi.org/10.1007/s00221-013-3708-9.

Verrel, J., 2010. Distributional properties and variance-stabilizing transformations for measures of uncontrolled manifold effects. J. Neurosci. Methods 191 (2), 166–170. https://doi.org/10.1016/j.jneumeth.2010.06.016.

Verrel, J., Lovden, M., Lindenberger, U., 2012. Normal aging reduces motor synergies in manual pointing. Neurobiol. Aging 33 (1), 200 e201–210. https://doi.org/10.1016/j.neurobiolaging.2010.07.006.

Vistamehr, A., Neptune, R.R., 2021. Differences in balance control between healthy younger and older adults during steady-state walking. J. Biomech. 128, 110717 https://doi.org/10.1016/j.jbiomech.2021.110717.

Vistamehr, A., Kautz, S.A., Bowden, M.G., Neptune, R.R., 2016. Correlations between measures of dynamic balance in individuals with post-stroke hemiparesis. J. Biomech. 49 (3), 396–400. https://doi.org/10.1016/j.jbiomech.2015.12.047.

Vistamehr, A., Balasubramanian, C.K., Clark, D.J., Neptune, R.R., Fox, E.J., 2018. Dynamic balance during walking adaptability tasks in individuals post-stroke. J. Biomech. 74, 106–115. https://doi.org/10.1016/j.jbiomech.2018.04.029.

Wang, Y., Asaka, T., Zatsiorsky, V.M., Latash, M.L., 2006. Muscle synergies during voluntary body sway: combining across-trials and within-a-trial analyses. Exp. Brain Res. 174 (4), 679–693. https://doi.org/10.1007/s00221-006-0513-8.

Winter, D.A., 2009. Biomechanics And Motor Control of Human Movement. John Wiley & Sons.

Yamagata, M., Tateuchi, H., Shimizu, I., Saeki, J., Ichihashi, N., 2019. The relation between limb segment coordination during walking and fall history in community-dwelling older adults. J. Biomech. 93, 94–100. https://doi.org/10.1016/j.jbiomech.2019.06.010.

Yamagata, M., Tateuchi, H., Shimizu, I., Saeki, J., Ichihashi, N., 2021. The relation between kinematic synergy to stabilize the center of mass during walking and future fall risks: a 1-year longitudinal study. BMC Geriatr. 21 (1), 240. https://doi.org/10.1186/s12877-021-02192-z.

Yen, J.T., Auyang, A.G., Chang, Y.H., 2009. Joint-level kinetic redundancy is exploited to control limb-level forces during human hopping. Exp. Brain Res. 196 (3), 439–451. https://doi.org/10.1007/s00221-009-1868-4.