



Effect of walking speed on the intersegmental coordination of lower-limb segments in elderly adults

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ABSTRACT

Background: Ageing brings profound changes in walking gait. For example, older adults reduce the modification of pelvic and trunk kinematics with walking speed. However, the modification of the coordination between lower-limb segments with age has never been investigated across various controlled speeds.

Research question: Is the effect of speed on the intersegmental coordination different between elderly and young adults?

Methods: Nineteen senior and eight young adults walked on a treadmill at speeds ranging from 0.56 to 1.94 m s⁻¹. The motion of the lower-limb segments in the sagittal plane was recorded by cinematography. When the angles of the thigh, shank and foot during a stride are plotted one versus the other, they describe loops constrained on a plane. The coordination between lower-limb segments was thus evaluated by performing a principal component analysis between the thigh, shank and foot elevation angles. The effect of speed and age on the intersegmental coordination was examined using a two-level linear mixed model ANOVA.

Results: In both age groups the orientation of the plane changes with speed, due to a more in-phase shank and foot motion. However, the effect of speed on the covariation plane is lessened with age.

Significance: Our results demonstrate that there is an age-related specific adjustment of the intersegmental coordination to speed. In particular, older adults restrict their repertoire of angular segment motion. These differences in coordination are mainly related to different foot-shank coordination.

1. Introduction

It is well-documented that ageing brings profound changes in gait and the locomotion function decreases with age [1]. The age-related differences in gait mechanics between young and older adults occur in parallel with, among others, a decrease in muscular force [2], the emergence of proprioceptive deficits [3] or a change in the muscular activity during walking [4]. However, the natural process by which these systemic neurophysiological changes affect the resultant kinematic pattern of walking are not clear. Understand the difference in coordination between lower-limb segments could reveal how the neuromuscular system is organized to coordinate movement during walking [5]. Thus, determining modification of coordination with age may shed light on the process behind age-related changes in walking mechanics.

The coordination pattern of lower-limb segments during walking can be described by global variables having fewer degrees of freedom than the actual ones. For example, the thigh, shank and foot elevation angles (i.e. the angle between the segment and the vertical in the

sagittal plane) do not evolve independently during a stride. When plotted one versus the other in a 3-D space, the trajectory of the elevation angle co-vary along a ‘tear-drop’ shaped loop, which lies close to a plane [6]. This so-called planar covariation of intersegmental coordination provides a good prediction of the task being achieved [7]. Indeed, the characteristics of a coordinated movement can be quantified by spatial features of the loop: the modifications in the shape of the loop and the orientation of the covariation plane provide insight into the essential timing and amplitude relationship between elevation angles of lower limb segments [7]. For example, changes in intersegmental coordination occur when speed increases [8], mainly because of a modification of the phase shift between shank and foot angles [9].

Despite the neuromuscular modifications with age, planar covariation is maintained with age [10,11], even during uphill walking [12]. But, the modification of spatial characteristics of the covariation plane with age has never been investigated across various speeds. Although coordination between trunk and pelvis has been shown to be dependent of both age and walking speed [13], it is not clear how lower-limb segments are coordinated in elderly adults. Providing more quantitative

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details on intersegmental coordination at various controlled speeds can better pinpoint underlying mechanisms of age-related neuromuscular adaptations. Therefore, the purpose of this study was to investigate the effects of age on the intersegmental coordination in healthy young and elderly adults walking at matched speeds. Since older adults present a reduced ability to modify kinematics pattern between different walking speeds [13–15], we hypothesized that the effect of speed on the spatial characteristics of the covariation loop would be smaller in elderly than in young adults.

2. Methods

2.1. Subject and experimental procedure

Eight young (age: 24.5 ± 2.4 y, mass: 71.9 ± 16.7 kg, height: 1.76 ± 0.08 m, means \pm SD) and 18 elderly adults (age: 75.6 ± 6.7 y, mass: 65.9 ± 14.5 kg, height: 1.68 ± 0.09 m, mean \pm SD) participated to the study. No subject had a history of falling after the age of 60; they were able to walk at least one kilometer and didn't complain about musculoskeletal or neurological disorder. All subjects gave their informed consent. Experiments were performed according to the Declaration of Helsinki and were approved by the local ethics committee (Belgian Registration Number: B403201524765).

Subjects were asked to walk on an instrumented treadmill. The treadmill consisted of a modified commercial treadmill (h/p/Comos-Stellar, Germany, belt surface: 1.6×0.65 m, mass: ~ 240 kg) instrumented with four force transducers (Arsalis, Belgium), measuring the three components of the ground reaction forces (GRF) exerted by the tread-belt on the feet [16]. Subjects walked at six different selected speeds (0.56, 0.83, 1.11, 1.39, 1.67 and 1.94 m s^{-1}), presented in a randomized order. Trials were not recorded if the imposed speed exceeded 150% of the spontaneous speed: no elderly adults walked at 1.94 m s^{-1} . The number of subjects recorded in each condition is presented in Table 1. For each trial, data were recorded for 15 s after 3–5 min of waking and an average of 13.5 ± 2.6 strides (mean \pm SD) were recorded in each trial.

Reflective markers were glued on the subject's skin at chin-neck intersect, greater trochanter, lateral femoral condyle, lateral malleolus and fifth metatarsophalangeal joint. The position of the markers in the sagittal plane was recorded every 5 ms by means of high-speed video camera (BASLER piA 640-210), fixed on the ground, 3 m to the right side of the treadmill, perpendicular to its long axis. Horizontal and vertical coordinates of the reflectors in the sagittal plane were measured in each frame using a dedicated tracking software (LABVIEW 2010, National Instruments, Austin, TX, USA). The joint angles (hip, knee, and ankle) were computed from the elevation angle of adjacent segments.

2.2. Data analysis

The stride was defined as the period between two contacts of the right foot. The lateral (d_y) and fore-aft (d_x) position of the centre of

pressure was computed as in [8] by:

$$d_x = \frac{-M_y - h F_x}{F_z} \quad (1)$$

$$d_y = \frac{-M_x - h F_y}{F_z} \quad (2)$$

where F_x , F_y and F_z are the lateral, fore-aft and vertical ground reaction forces; M_x and M_y are the moment components in the force transducer coordinate system; and h is the vertical distance between the force transducers and the tread-surface. Foot-contact (FC) and toe-off (TO) were then estimated from the displacement of the centre of pressure [17].

From the markers location, the orientation of the thigh, shank, foot and trunk relative to the vertical axis (elevation angle) were computed as described in Borghese et al. [6]. The joint angles (hip, knee and ankle) were then computed from the elevation angle of adjacent segments. For each subject, the different strides of each trial were time-interpolated to fit a normalized 400-points time base (i.e. every 0.25% of the stride period) and then averaged.

To analyse the time-course of the elevation angle during the stride, a Fourier series component was performed [9]. The phase shift (P) and percentage of variance accounted for by the first harmonic were computed. The phase shift (P_{ij}) between two adjacent limb-segments i and j was computed as $P_{ij} = P_j - P_i$.

A principal component analysis was applied to determine the covariance matrix of the segment elevation angles. Eigenvalues and eigenvectors u_i were computed by factoring the covariance matrix from the set of original signals by using a singular value decomposition algorithm. The first two eigenvectors (u_1 and u_2) lied on the best-fitting plane of angular covariation and the data projected onto these axes corresponded to the first (PC_1) and second (PC_2) principal components. The planarity was evaluated for each condition by calculating the percentage of variance that was explained by u_1 (PV_1) and u_2 (PV_2). If the data were lying perfectly on a plane, $PV_1 + PV_2$ would be 100%. By definition, the third eigenvector u_3 is orthogonal to the plane defined by u_1 and u_2 . The parameters u_{3t} , u_{3s} , and u_{3f} corresponded to the direction cosines with the positive semi-axis of the thigh, shank, and foot angular coordinates, respectively. The effect of speed has been showed to be greater on u_{3t} [9]. Therefore, we specifically analysed and plotted the u_{3t} parameter, which provides a measure of the orientation of the plane. In order to evaluate the changes in the orientation of the plane with speed, the slope of the linear regression between u_{3t} and the speed of progression was measured.

2.3. Statistics

The statistical analysis was designed to assess the effect of speed of progression, of the age group and of the interaction between these two factors. Since none of the older adults were able to walk at 1.94 m s^{-1} , the young adults walking at that speed were not considered in the statistical analysis. A two-level linear mixed model ANOVA was applied: speed and age group were set as fixed effects, and the subject was

Table 1

Stride length and range of motion (ROM) of the lower-limb joint angles as a function of walking speed in young (Y) and elderly (E) adults. N is the number of subjects in each age-speed class.

Speed (m s^{-1})	N	Stride length (m)		Hip ROM ($^\circ$)		Knee ROM ($^\circ$)		Ankle ROM ($^\circ$)		
		Y-E	Y	E	Y	E	Y	E	Y	E
0.56	8-18		0.94 ± 0.07	0.80 ± 0.15	31.3 ± 3.1	30.9 ± 5.2	54.4 ± 5.0	52.8 ± 6.6	20.7 ± 3.2	18.0 ± 2.8
0.83	8-18		1.13 ± 0.06	0.99 ± 0.14	33.6 ± 2.3	32.7 ± 5.4	61.4 ± 3.4	55.8 ± 5.4	24.5 ± 4.2	18.4 ± 3.2
1.11	8-18		1.31 ± 0.06	1.18 ± 0.14	36.7 ± 2.3	35.8 ± 5.4	64.0 ± 3.6	58.7 ± 5.0	26.7 ± 3.4	20.8 ± 4.4
1.39	8-17		1.51 ± 0.06	1.37 ± 0.10	38.5 ± 2.9	39.7 ± 5.7	64.8 ± 3.3	61.2 ± 5.4	28.2 ± 3.4	22.7 ± 4.3
1.67	8-14		1.69 ± 0.07	1.55 ± 0.12	40.3 ± 4.1	42.5 ± 5.8	64.7 ± 2.6	61.2 ± 4.5	31.4 ± 4.4	25.5 ± 4.5
1.94	8-0		1.79 ± 0.07	/	42.5 ± 4.7	/	63.2 ± 3.0	/	32.9 ± 5.1	/

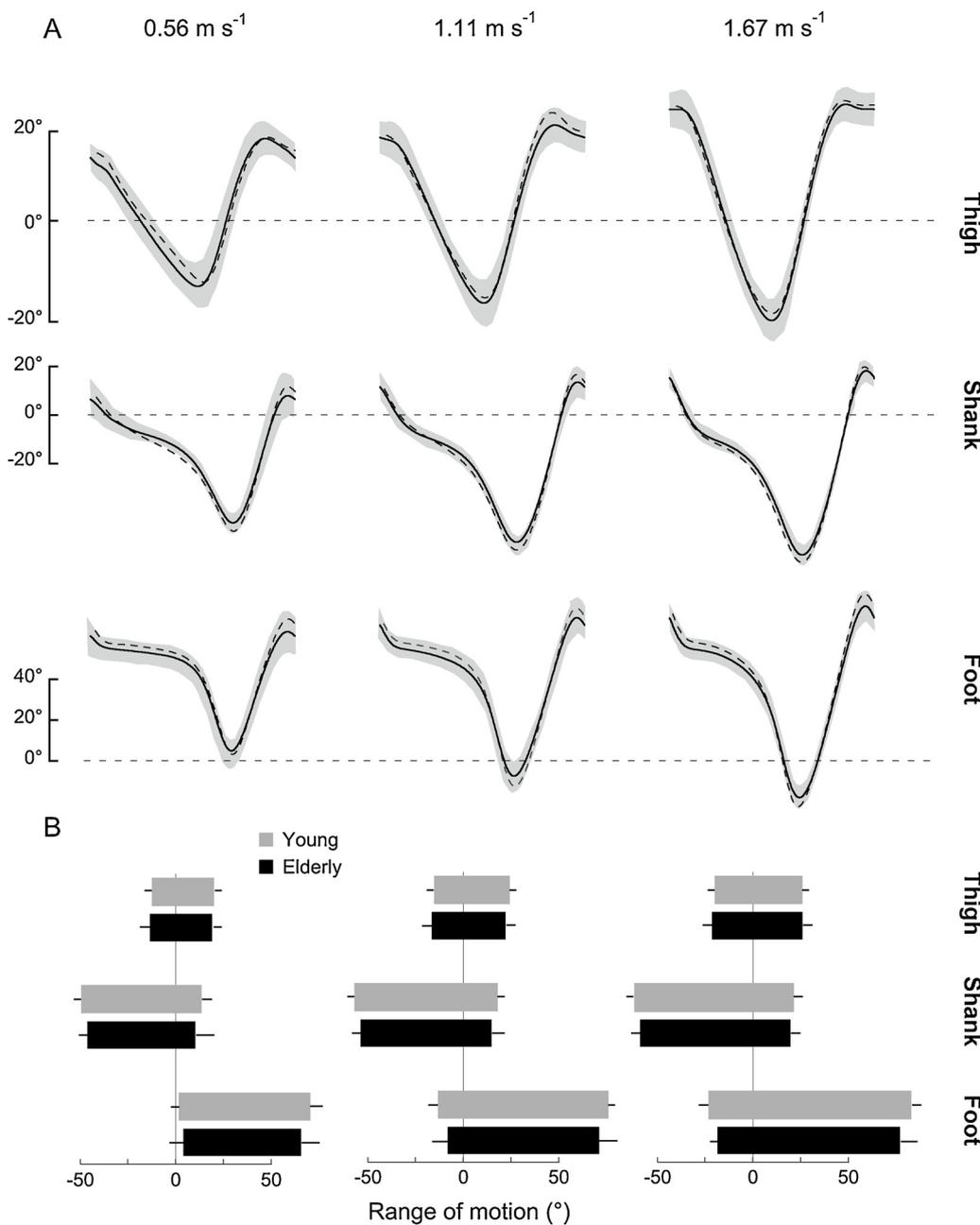


Fig. 1. Elevation angles of lower-limb segments during walking at slow, intermediate and fast walking speed in young and elderly adults (A): Elevation angles of the thigh, shank and foot over a stride at the speed of 0.56 (left), 1.11 (center) and 1.67 (right) m s⁻¹. All the curves of each subject walking at a given speed were first averaged (mean-curve). The curves presented here are the average of the mean-curves of all the young (interrupted lines) and elderly (continuous lines) adults. The grey zone represents ± 1 SD for the elderly adults. The horizontal interrupted lines correspond to the vertical position of the segment. (B): Average range of motion of the thigh, shank and foot over one stride at 0.56 (left), 1.11 (center) and 1.67 m s⁻¹ (right). All the strides of each subject walking at a given speed were first averaged. The rectangles represent the grand mean of all the young (grey) and the elderly (black) adults. Thin bars represent one standard deviation. In panels A and B, an angle of 0° correspond to the vertical position of the segment and a positive/negative angle corresponds to the forward/backward elevation of the segment.

set as a random effect. The normality of the residuals was checked by the Kolmogorov-Smirnov. Normality of the residuals was not assumed for 3 variables (range of motion of the thigh elevation angle, PV₁, phase shift between the thigh and the shank). In those cases, a log₁₀ transform was applied and the normality of the residuals was then assumed. Linear regression analysis, using Pearson’s correlation coefficient (*r*), was used to quantify the relationship between variables. Independent sample Student’s *t*-tests were used to compare the parameters of linear regression with speed between age groups. In all analyses, the significance level was fixed at *P* < 0.05.

3. Results

When speed increases, both the stride length and the range of motion (ROM) of the hip, knee and ankle joints increase (stride length: *F* = 124.8, *P* < 0.001; hip: *F* = 15.2, *P* < 0.001; knee: *F* = 10.2, *P* < 0.001; ankle: *F* = 18.3, *P* < 0.001; Table 1) in both groups. At all speeds, the stride length and the ROM of the knee and ankle joints are

smaller in elderly than in young adults (stride length: *F* = 36.4, *P* < 0.001; knee: *F* = 15.4, *P* < 0.001; ankle: *F* = 48.3, *P* < 0.001; Table 1).

In both groups, the ROM of the elevation angles increases with speed (thigh: *F* = 31.8, *P* < 0.001; shank: *F* = 31.2, *P* < 0.001; foot: *F* = 53.27, *P* < 0.001; Fig. 1A). The shape of the elevation angles time-course remains fairly similar across age groups (Fig. 1A). However, the ROM of the shank and the foot segments are slightly but significantly smaller in elderly than in young adults (shank: *F* = 6.6, *P* = 0.016; foot: *F* = 10.9, *P* < 0.003; Fig. 1B).

In each age-speed class, the planar covariation is maintained since PV₁ + PV₂ > 99%. However, PV₁ is smaller (and PV₂ is greater) in elderly than in young adults (*F* = 16.2, *P* < 0.001, Fig. 2A). Furthermore, PV₁ increases (and PV₂ decreases) with speed (*F* = 11.2, *P* < 0.001) both in young and elderly adults.

The orientation of the plane is modified when the walking velocity increases (Fig. 3), evidenced by the direction cosines *u*_{3*t*}, which decreases linearly with increasing speed (*F* = 7.3, *P* < 0.001). Note that

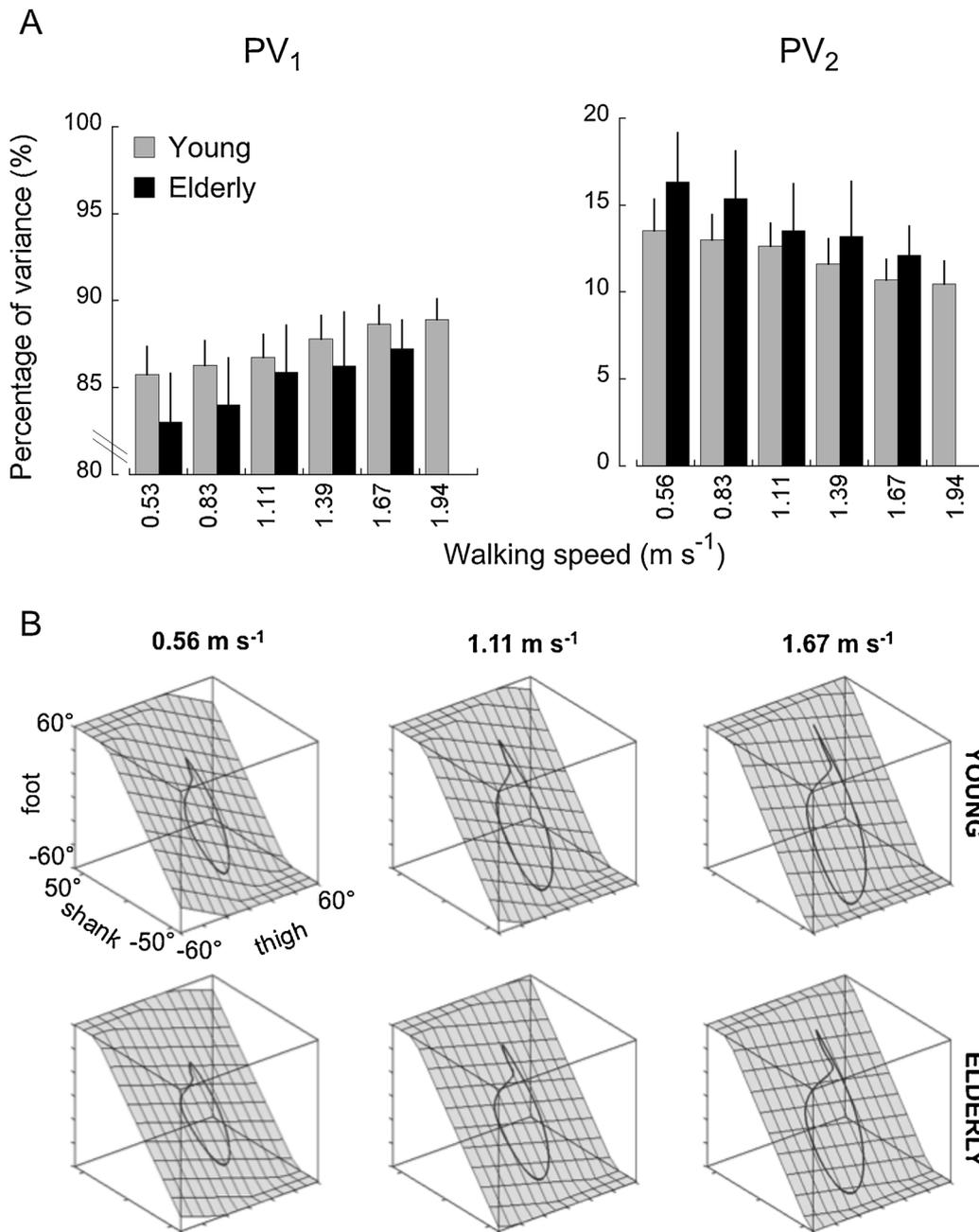


Fig. 2. Planar covariation of elevation angles (A): Percentage of variance accounted for by the first (PV₁, left) and the second (PV₂, right) eigenvector of the principal component analysis. Grey bars correspond to young adults, whereas black bars correspond to elderly adults. (B): Covariation of the ensemble-average limb-segment elevation angles during walking at 0.56 (left), 1.11 (center) and 1.67 (right) m s⁻¹ in young (up) and elderly adults (bottom). Note that when the elevation angles of thigh, shank and foot are plotted one versus the other in a x-y-z space, they co-vary along a loop constrained on a plane (x-y). Grids show the best-fitting plane. Other indications as in Fig. 1.

at all speeds, u_{3t} is smaller in elderly than in young adults ($F = 23.4$, $P < 0.001$) and the slope (s) and coefficient of correlation (r) of the linear regression are smaller in elderly than in young adults (s : $t = -2.19$, $P = 0.039$; r : $t = 2.49$, $P = 0.020$). Despite the fact that u_{3s} and u_{3f} also change linearly with increasing speed, the slope (s) of the linear regression are more than 3 times smaller as compared to the slope (s) of the linear regression of u_{3t} ($t = -3.66$; $P = 0.001$).

When performing a Fourier series analysis, the first harmonic of the elevation angles accounts for $85.9 \pm 4.0\%$ (means \pm SD) of the variance for all segments, at all speeds and in both age groups. Thus, the first harmonic of pairs of adjacent lower-limb segments (thigh-shank and shank-foot) captures the time relationship characteristics of the elevation angles (Fig. 4A). At each walking speed, the phase shifts between these first harmonics (P_{ts} and P_{sf}) are positive.

Both P_{ts} and P_{sf} are reduced when speed increases (P_{ts} : $F = 12.1$, $P < 0.001$; P_{sf} : $F = 7.9$, $P < 0.001$). Note also that P_{ts} is not different between age groups ($F = 2.8$, $P = 0.097$), whereas P_{sf} is significantly reduced in elderly adults as compared to young ($F = 16.4$, $P < 0.001$).

The changes in u_{3t} with speed is correlated with the changes in P_{sf} (Fig. 4B) and the correlation coefficient of the linear regression between the two is $r = 0.94$ ($P < 0.001$).

4. Discussion

In this study, the effect of ageing on intersegmental coordination is investigated by comparing healthy young and elderly adults walking at matched speeds. We show that the effect of speed on the spatial characteristics of the covariation loop differs with age, shedding light on age-related modifications of the coordination strategies during walking.

To date, a handful of studies have dealt with the effects of ageing on intersegmental coordination (e.g. [11,12]). Nevertheless, most of them have been carried out at self-selected walking speed. In accordance with the literature, our analysis confirms that kinematic segmental covariation during walking observed in young adults [6] is maintained after the age of 65 [10–12] and that the speed of progression leads to a rotation of the principal plane [9]. The rotation occurs mainly along the

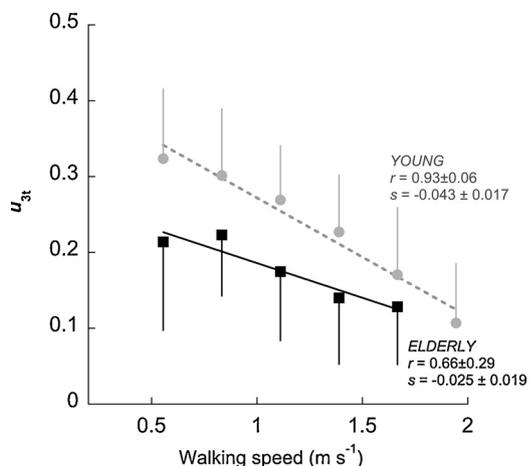


Fig. 3. Orientation of the principal plane in young and elderly adults as a function of walking speed. Direction cosines of the normal to the covariation plane with the positive semi-axis of the thigh angular coordinates (u_{3t}) as a function of walking speed in young (grey circles) and elderly (black squares) adults. Other indications as in Fig. 1B. The continuous grey (young) and black (elderly) lines correspond to the linear fit curves drawn through all the data (Kaleidagraph 4.5). The average coefficients of correlation (r) and slope (s) of the linear regression in young and elderly subject are given.

long axis of the gait loop (Fig. 2B), leading to a reduction of u_{3t} (Fig. 3). By separating the effects of both concomitant issues, such as age and speed, we show here that, as compared to young adults, u_{3t} is smaller and the effect of speed on u_{3t} is reduced in elderly adults (Fig. 3).

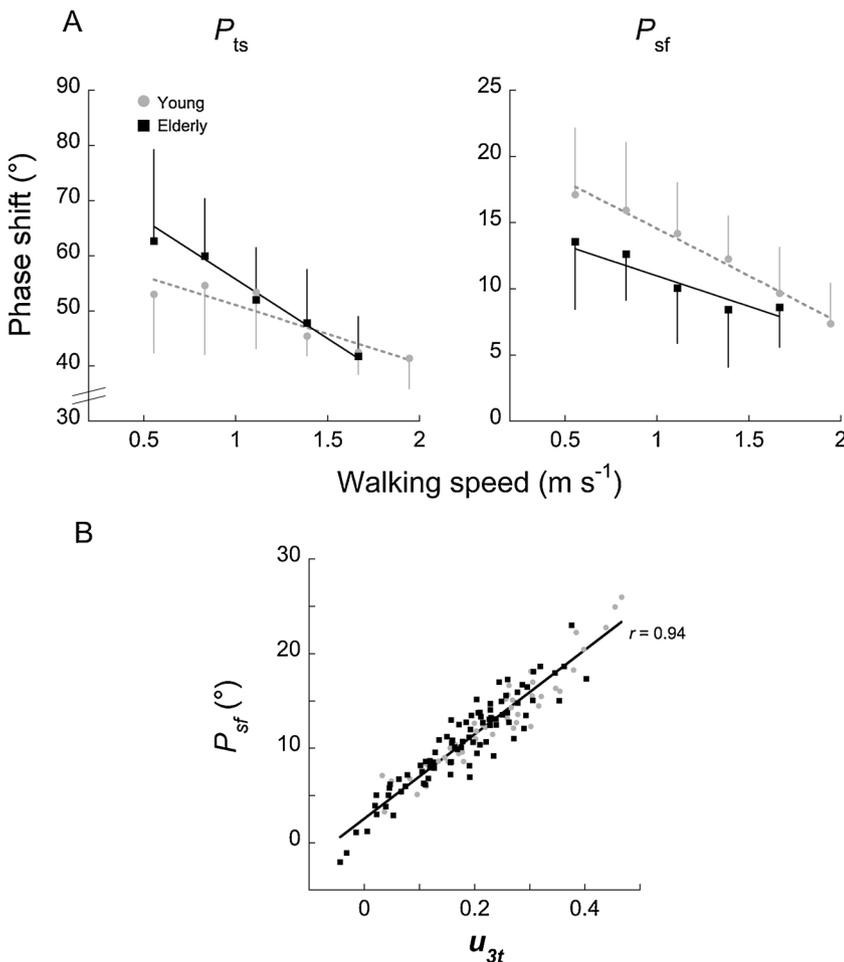


Fig. 4. Orientation of the principal plane in young and elderly adults as a function of walking speed (A): Phase shift between the 1st harmonic of thigh relative to shank elevation angles (subscript ts , left) and between the 1st harmonic of shank relative to foot elevation angles (subscript sf , right) as a function of walking speed in young (grey circles) and elderly (black squares) adults. Other indications as in Fig. 1B. The continuous grey (young) and black (elderly) lines correspond to the linear fit curves drawn through all the data (Kaleidagraph 4.5). (B): Correlation between P_{sf} and the direction cosines of the normal to the covariation plane with the positive semi-axis of the thigh angular coordinates (u_{3t}). Correlation coefficient is based on the least-squares linear regression on all trials from all subjects (Kaleidagraph 4.5).

The change in the orientation of the plane, evaluated by u_{3t} , is mainly related to a change of the phase shift between shank and foot elevation angles [6]. At all ages and at all speeds, the phase shift between the motions of the shank and the foot (P_{sf}) is positive, meaning that the motion of the shank precedes that of the foot. When speed increases, P_{sf} decreases, which in turn affects the orientation of the plane (Fig. 4B).

Note that P_{sf} is smaller and change less with speed in elderly than in young adults, resulting in the modifications of intersegmental coordination observed with ageing (Fig. 4B). In addition, the more in-phase oscillation of the shank and the foot in elderly adults may explain the reduction of ankle ROM with ageing (Table 1), most likely because seniors “lock their ankle” during mid-stance [14]. This reduced angular excursion at the ankle in elderly has already been ascribed to a stiffening of the ankle joint due, among others, to co-contractions of antagonist muscles [18] or to a reduced ankle joint flexibility [19].

In addition, the smaller ROM has been associated with a loss of ankle power generation to propel the centre of mass of the body in late stance [20,21]. However, the decrease of the ankle power generation cannot only be explained by a decrease in the muscular capacities, since Franz [22] has observed that elderly adults are able to develop higher ankle power when walking uphill. Instead, the smaller ankle power and ROM and the modified intersegmental coordination might reflect a different walking strategy adopted to reduce ankle efforts at the expense of proximal extensors [21].

Another feature of intersegmental coordination in older adults consists of a greater PV_2/PV_1 ratio than young adults (Fig. 2). During walking, a greater PV_2 has already been observed when crossing obstacles, suggesting that a higher PV_2/PV_1 ratio may reflect central nervous system control mechanisms to facilitate an increased toe

clearance [23]. Similarly, the greater PV_2/PV_1 ratio observed here could indicate a different motor strategy in elderly adults to control toe clearance in level walking. This is further supported by a prior study showing that as compared to young adults, elderly display different strategies to avoid toe catching [24].

The modification of shank and foot coordination in elderly may also be undertaken to increase in the postural stability during walking [19]. For example, Kerrigan, Lee [25] showed that elderly with an history of falls generate 22% less ankle power during push-off than those without history of falls. Furthermore, the adaptive changes of intersegmental coordination in patients with balance disorder, like patients with cerebellar ataxia [26], are similar to those observed here in elderly adults (Figs. 2A & 3). Indeed, as in elderly adults, ataxic patients have a reduced stride length and their intersegmental coordination is modified: they present a reduced u_{3c} and a greater PV_2/PV_1 ratio [26].

Coordination of limb segments plays a central role in the production of effective movement [27,28]. The different intersegmental coordination strategy used by elderly as compared to young adults may be due to multifactorial reasons. Among other factors, older adults might adopt a different control strategy in an attempt to redistribute the joint efforts between the hip, knee and ankle [21] and/or to preserve stability since ageing would involve greater fall risk [13]. However, our results do not allow to establish a direct link between lower-limb segments coordination and walking dynamics.

In this study, intersegmental coordination was analysed only in the sagittal plane, as movements in this plane represent the major and most systematic component of walking gait [6]. However, movements in the other dimensions of space may change with age and should be considered in further studies. Another limitation of the study to mention is the use of a treadmill, which may have imposed constraints on limb coordination.

Despite these limitations, the present paper demonstrates that there is an age-related adjustment of the intersegmental coordination to walking speed. Our results suggest that the modification of the orientation of the covariation plane with speed is lessened in elderly as compared to young adults, suggesting that senior restricts the repertoire of angular segment motion. These differences in coordination could mainly be related to different foot-shank coordination. Future research focusing on motor control strategies would be of great interest to understand the difference in coordination with age.

Conflict of interest statement

All authors disclose any financial and personal relationships with other people or organizations that could inappropriately influence (bias) their work.

CRediT authorship contribution statement

A.H. Dewolf: Conceptualization, Formal analysis, Methodology, Software, Validation, Visualization, Writing - original draft, Writing - review & editing. **G.M. Meurisse:** Conceptualization, Data curation, Formal analysis, Investigation, Validation, Visualization, Writing - review & editing. **B. Schepens:** Conceptualization, Data curation, Formal analysis, Investigation, Project administration, Supervision, Validation, Visualization, Writing - review & editing. **P.A. Willems:** Project administration, Supervision, Validation, Visualization, Writing - review & editing.

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