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► **To cite this version:**

Bruno Watier, J er emie Begue, H el ene Pillet, Teddy Caderby. Instability during Stepping and Distance between the Center of Mass and the Minimal Moment Axis: Effect of Age and Speed. Applied Sciences, 2023, 13 (19), pp.10574. 10.3390/app131910574 . hal-04239243

HAL Id: hal-04239243

<https://laas.hal.science/hal-04239243>

Submitted on 12 Oct 2023

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Article

Instability during Stepping and Distance between the Center of Mass and the Minimal Moment Axis: Effect of Age and Speed

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Featured Application: This study is a first step for real-time monitoring the risk of falls during gait.

Abstract: The goal of this study was to analyze instability during stepping at different speeds in young and older adults. To this aim, the anteroposterior and the mediolateral distances between the body center of mass (COM) and the minimum moment axis (MMA) were computed. A total of 15 young adults (25 y.o. [19–29]) and 15 older adults (68.7 y.o. [63–77]) volunteered for this study. For the computation of the distances, a complete biomechanical protocol combining two force platforms and a 3D motion capture analysis system was setup. The subjects were equipped with 47 reflective markers and were modeled as a frictionless multibody system with 19 segments, 18 joints and 42 degrees of freedom. They were asked to perform a series of stepping tasks at fast and spontaneous speeds. The stepping was divided into five phases, with successive swing and double-stance phases. Greater instability was observed during the swing phases. The distances reveal a significant higher instability at fast speed for both groups ($p < 0.001$) for all the phases compared with spontaneous speeds. The anteroposterior distance was significantly greater for older adults, highlighting greater instability compared to young adults, while no differences were observed for the mediolateral distance all along the five phases, suggesting higher risks of backward and forward falls during stepping.

Keywords: biomechanics; gait; angular momentum; locomotion

Citation: Watier, B.; Begue, J.; Pillet, H.; Caderby, T. Instability during Stepping and Distance between the Center of Mass and the Minimal Moment Axis: Effect of Age and Speed. *Appl. Sci.* **2023**, *13*, 10574. <https://doi.org/10.3390/app131910574>

Academic Editor: Hanatsu Nagano

Received: 13 August 2023

Revised: 19 September 2023

Accepted: 20 September 2023

Published: 22 September 2023



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1. Introduction

Autonomous locomotion is largely related to quality of life. The loss of autonomous gait has been shown to be the first indicator of decline in motor skills but also in the cognitive abilities of individuals [1]. In this context, a better understanding of the factors that affect locomotion is essential to detect and prevent the loss of autonomy [2].

The ability to walk indoors and outdoors without falling is a key element in maintaining the autonomy of people whose locomotor system is impaired due to ageing, disability or pathology. Falling can occur for different reasons such as tripping (34%), slipping (25%) and incorrect weight shifting [3,4], which cause strong balance perturbations during the steady-state gait cycle. For non-impaired individuals, balance recovery following these perturbations is performed via changes in the moments generated at all joints of the support limb, resulting in a modification of the contact forces with the ground [5]. This control of the contact forces depends on the balance control system, which generates appropriate muscular actions based on sensory information from the visual, vestibular and somatosensory

systems. The vestibular system senses the acceleration and the angular velocity of the head, and the somatosensory system senses ground reaction forces and body positioning [6]. When impaired, the balance control system no longer performs its regulatory role, and recovery from the aforementioned perturbations is no longer guaranteed. Considering the mechanical nature of the signals implied in the balance control, it seems reasonable to focus the understanding of balance on biomechanical analysis of the body, which should provide essential information for personalized rehabilitation interventions [7].

During gait, the external mechanical actions applied to the body are gravity and ground contact forces. The latter are regularly broken (toe-off) and recovered (heel strike), and can be modulated so that the system efficiently performs a specific movement. Numerous studies have been carried out to define criteria for assessing the degree of gait instability. First, the trajectory of the center of pressure (COP) has been widely used as an indicator of healthy gait [8]. Since then, several indicators considering the whole-body dynamics of the musculoskeletal system have been proposed. Hof et al. [9], by using the “extrapolated center of mass”, defined the limit of the surface in which the center of mass (COM) should be located, and quantified the margin of stability during locomotion. This parameter is based on the model of the gait as an inverted pendulum, often proposed in the literature [10]. Angular momentum and the derivative of angular momentum at the COM have been particularly studied during walking and have been shown to increase significantly during unstable phases of gait [7,10,11]. In particular, the reduction in angular momentum variation during walking, consistent with the limitation of uncontrolled whole-body rotations, appears a realistic goal to achieve to improve overall balance abilities of impaired populations. It has previously been discussed in the works of Gomez et al. [12] and Vistamehr et al. [13] for people with Parkinson’s disease and post-stroke patients, respectively. In the same idea, Bailly et al. [14] studied the distance between the COM and the minimum moment axis (MMA), which is intimately related to the derivative of the angular momentum at the body COM: the greater the distance, the greater the angular acceleration of the body around the COM. This criterion could be easily understood in a clinical context. Its ability to discriminate healthy subjects walking in unstable conditions compared to level-ground walking has already been shown. In addition, MMA-COM index can be computed under various conditions, including when all contact points are not on the same horizontal surface (slope, stair, when using assistive devices, uneven terrain, etc.), contrary to classical COM or COP indices. Al. Abiad et al. [15] used this parameter to highlight the greater instability of locomotion in subjects with a transfemoral prosthesis.

Gait initiation and stepping, which consist of a voluntary transition from a quiet standing position to a dynamic phase, are known to be a particularly critical phase of locomotion. Observations revealed that most of the falls in older adults occurred during these transitional tasks [4]. In particular, research has shown that anticipatory postural adjustments performed during the double support phase are crucial to guarantee and prepare the correct execution of the first step [16]. Indeed, at the time of the takeoff of the foot, a mediolateral gap is created between the COP located under the foot of the stance leg and the projection of the COM. This natural gap induces a fall of the body toward the swing leg side, which stops at the time of swing foot contact [17]. The authors showed, during the anticipatory postural adjustments, an initial shift in the COP laterally towards the swing leg that accelerates the COM toward the contralateral leg in order to decrease the amplitude of the lateral fall [17,18]. Furthermore, they observed a forward displacement of the COM induced by the backward shift in the COP. The forward imbalance created by the anteroposterior gap between the COM and the COP, during the double support phase contributes to the generation of propulsive forces [19,20]. Segmental movements resulting from these anteroposterior and lateral movements generate variation in angular momentum, which helps to initiate and continue the gait.

In this context, some authors recently studied instability during stepping. Using the COP trajectory, Watanabe and Higuchi [21] demonstrated that action costs for maintaining postural stability are considered dominantly for planning the stepping. The margin of

stability [22] was also analyzed during stepping in different situations, and the higher risk of falls in frail individuals such as older adults were highlighted. Muijres et al. [23] also studied the stepping instability in older adults using foot placement error, step duration and mediolateral center of pressure path. The results showed that compared to young adults, older adults exhibited a higher foot placement error, suggesting a reduction in the ability to control balance with age. However, while foot placement error increased with step execution velocity, the authors did not find an age-related difference at higher speeds. On the contrary, age-related changes in angular momentum were found in other studies [24,25]. Compared to young adults, older adults exhibited higher values of angular momentum during stepping, emphasizing greater difficulties in the balance control during this task. It is noteworthy that these age-related changes were more pronounced in the sagittal plane. Furthermore, although both age groups increased angular momentum with increasing speed during stepping, the age-related changes previously observed at spontaneous speed were exacerbated at faster speeds [21]. Taken together, these results suggest that the greater variation in angular momentum with age and speed generates larger values of angular momentum derivatives and consequently contributes to increasing the distance between the COM and the MMA. However, to the best of our knowledge, no study has investigated instability during stepping using the distance between the COM and the MMA.

This study aimed, therefore, to investigate the distance between the MMA and the COM in young and older adults at different speeds during stepping. We hypothesized that, due to the greater variation in sagittal angular momentum in older adults, the derivative of the angular momentum should increase. As a consequence, the anteroposterior distance between the MMA and the COM should also increase. Similarly, we hypothesized that increasing speed should generate a greater distance between the MMA and the COM due to a greater variation in angular momentum. Finally, in this study, we tested the ability of the distance between the MMA and the COM to quantify this greater instability.

2. Materials and Methods

2.1. Participants

Thirty adults participated to this research and were assigned into two groups. Fifteen of them were young adults (25 y.o. [19–29]; 62.3 kg [44.8–76.1]; 170.2 cm [155–186]; 1 female; 3 males) and the fifteen others were older adults (68.7 y.o. [63–77]; 58.7 kg [37.1–77.1]; 157.6 cm [147–172]; 9 females; 6 males). None of them reported a history of locomotor pathologies. The study protocol was conducted in accordance with the Declaration of Helsinki and approved by the local institutional review board.

2.2. Experimental Procedure

Data collection methods have previously been described by Begue et al. [24] and are summarized here.

Initially, the volunteers stood barefoot and motionless on a force plate (AMTI, Watertown, MA, USA). After a verbal signal from the experimenter, they were asked to initiate a first step with their dominant leg up to a second force plate (SENSIX, Poitiers, France) located in front of the first force plate. All initiated the first step with their right leg, except for two young adults. Initial feet position was marked on the ground and standardized [26]. The participants were equipped with 49 reflective passive markers placed on anatomical bony landmarks based on the recommendations of the International Society of Biomechanics (ISB) [27,28]. Kinematics were collected with a Vicon system (200 Hz). At the end of the second step, participants had to stop in a natural upright posture with their arms alongside their body. Stepping was completed under two randomized speed conditions: at spontaneous speed and as fast as possible. For each participant, at least 9 trials were achieved at both speeds. Both force plates were synchronized with the Vicon system, and their data were collected at a frequency of 1000 Hz.

2.3. Skeletal Model

The human model included 42 degrees of freedom, 18 joints and 19 segments as presented by Maldonado et al. [29]. The skeleton was modeled as a friction-less articulated multi-body dynamic system. The pelvis, thighs, shanks and feet segment masses and inertia were set according to the model of [30]. Anthropometry of the torso and head segments, including the neck, were estimated from the regression equations of [31]. Hand anthropomorphic data were based on regression equations of De Leva [32].

2.4. Data Analysis

Kinematics and kinetics were processed with a cut-off frequency using a low-pass Butterworth digital filter of 4th order applied in a zero phase. A cutoff frequency of 6 Hz was used after a residual analysis for the kinematics and at 10 Hz for the force plate data. Euler XYZ body-fixed rotation angles are used to express the orientation using OpenSim [33] after an inverse kinematics process with body frames defined according to ISB recommendations [27,28]. The COM position of the whole body was also extracted after the inverse kinematics process. In the coordinate system used, X is the anteroposterior axis, Y is vertical, and Z is the mediolateral axis.

The stepping motion was divided into five phases [24]: anticipatory postural adjustments (APAs) and first double support phase (P1), swing phase of the dominant leg (P2), second double support phase (P3), swing phase of the second step (P4) and third double support phase until the end of the movement (P5), also named the restabilization phase. The APAs were considered to begin when one of the anteroposterior and mediolateral accelerations of the COM deviated 2.5 standard deviations from its baseline value until the toe-off of the dominant foot [25,34]. The end of the motion was defined as the time-point at which the mediolateral COM velocity remained within 2 standard deviations of the mean calculated during the terminal-stance quiet standing after the end of the stepping [25,35]. Each phase was normalized from 0 to 100%.

The resulting subject-specific kinematics data, i.e., positions, orientations and velocities of body segments, and the inertial parameters were exported to a custom-made Matlab program for the angular momentum computation. The whole-body angular momentum at the COM (\mathbf{H}^{COM}) was calculated in three dimensions, as described by Begue et al. [24]. Then, the derivative of the angular momentum was extracted using classical methodology via finite difference ($\mathbf{K}^{\text{COM}} = d\mathbf{H}^{\text{COM}}/dt$). Finally, the distance between the MMA and the COM was computed as explained below.

2.5. Theoretical Background

It is current in biomechanics to represent the resulting ground reaction forces as a vector located under the foot at the COP. The COP is defined as the barycenter of the vertical pressure under the foot. It does not take into account friction forces in the horizontal plane (XZ). Thus, at the COP, the ground reaction moments are null alongside X and Z. Only the vertical free moment is not null. However, the COP is not the point where the moment is minimal. At this point Q of the ground, the moment is not vertical but collinear to the reaction forces. Let us denote \mathbf{M}^{Q} as the moment at this point and \mathbf{R} as the reaction force.

$$\mathbf{M}^{\text{Q}} = \lambda \mathbf{R} \quad (1)$$

where λ is a given real value. According to the Varignon theorem, we can compute the moment at another point P:

$$\mathbf{M}^{\text{P}} = \mathbf{M}^{\text{Q}} + \mathbf{R} \times \mathbf{QP} = \lambda \mathbf{R} + \mathbf{R} \times \mathbf{QP} \quad (2)$$

where \times denotes the cross product. The square of Equation (2) leads to:

$$(\mathbf{M}^{\text{P}})^2 = (\lambda \mathbf{R} + \mathbf{R} \times \mathbf{QP})^2 = (\lambda \mathbf{R})^2 + 2\lambda \mathbf{R} \cdot (\mathbf{R} \times \mathbf{QP}) + (\mathbf{R} \times \mathbf{QP})^2 \quad (3)$$

where \cdot denotes the scalar product. It is clear that $2\lambda\mathbf{R} \cdot (\mathbf{R} \times \mathbf{QP}) = 0$ since $\mathbf{R} \times \mathbf{QP}$ is orthogonal to \mathbf{R} . Finally:

$$(\mathbf{M}^P)^2 = (\lambda\mathbf{R})^2 + (\mathbf{R} \times \mathbf{QP})^2 = (\mathbf{M}^Q)^2 + (\mathbf{R} \times \mathbf{QP})^2 \tag{4}$$

and

$$(\mathbf{M}^P)^2 \geq (\mathbf{M}^Q)^2 \tag{5}$$

Thus, the square-norm of the moment at point Q is minimal. One can note that this moment is constant alongside an axis colinear to \mathbf{R} , named minimal moment axis (MMA). Indeed, if M is a point along this axis:

$$\mathbf{QM} = a\mathbf{R} \quad \text{with } a \in [-\infty, +\infty] \tag{6}$$

$$\mathbf{M}^M = \mathbf{M}^Q + \mathbf{R} \times \mathbf{QM} = \mathbf{M}^Q + \mathbf{R} \times (a\mathbf{R}) = \mathbf{M}^Q \tag{7}$$

Now, if A is any point in space, it is possible to localize this minimal moment axis according to A. We define the point B as:

$$\mathbf{AB} = (\mathbf{R} \times \mathbf{M}^A) / \mathbf{R}^2 \tag{8}$$

The moment at point B is:

$$\mathbf{M}^B = \mathbf{M}^A + \mathbf{R} \times \mathbf{AB} = \mathbf{M}^A + \mathbf{R} \times (\mathbf{R} \times \mathbf{M}^A) / \mathbf{R}^2 \tag{9}$$

Using the properties of the double cross product:

$$\mathbf{R} \times (\mathbf{R} \times \mathbf{M}^A) / \mathbf{R}^2 = [(\mathbf{R} \cdot \mathbf{M}^A)\mathbf{R} - (\mathbf{R} \cdot \mathbf{R})\mathbf{M}^A] / \mathbf{R}^2 = (\mathbf{R} \cdot \mathbf{M}^A)\mathbf{R} / \mathbf{R}^2 - \mathbf{M}^A = \lambda\mathbf{R} - \mathbf{M}^A \tag{10}$$

Considering $\lambda = (\mathbf{R} \cdot \mathbf{M}^A) / \mathbf{R}^2$.

Finally, the moment at point B is:

$$\mathbf{M}^B = \mathbf{M}^A + \lambda\mathbf{R} - \mathbf{M}^A = \lambda\mathbf{R} \tag{11}$$

Accordingly, it is possible to determine one point of the MMA from any other point given the reaction force and the moment at this point. Moreover, this MMA exists regardless of the conditions of the gait. Indeed, this axis is not defined only when walking on flat ground like the COP and can take into account all external mechanical actions: swing phase, double stance, contact with a cane, staircase ramp, etc. Moreover, the vector $\mathbf{AB} = (\mathbf{R} \times \mathbf{M}^A) / \mathbf{R}^2$ represents the shortest distance between A and the MMA (the orthogonal projection of A on the MMA). Indeed, it is clear that \mathbf{AB} is perpendicular to \mathbf{R} and thus, to the direction of the MMA.

In the present study, we computed the derivative of the angular momentum at the COM as presented above. According to Newton's equation, the derivative of the angular momentum (\mathbf{K}^{COM}) at the COM is equal to the moment at the COM (\mathbf{M}^{COM}); i.e.,:

$$\mathbf{K}^{\text{COM}} = \mathbf{M}^{\text{COM}} \tag{12}$$

Thus, the shortest vector between the COM (point G) and its orthogonal projection on the MMA (point M) is denoted $\mathbf{d}_{\text{COM-MMA}}$ and was computed as follows:

$$\mathbf{d}_{\text{COM-MMA}} = \mathbf{GM} = (\mathbf{R} \times \mathbf{M}^{\text{COM}}) / \mathbf{R}^2 = (\mathbf{R} \times \mathbf{K}^{\text{COM}}) / \mathbf{R}^2 \tag{13}$$

where values of \mathbf{R} are the total reaction forces obtained with the force plates. Accordingly, all other things being equal, the greater the derivative of the angular momentum, the greater the distance. In this circumstance, the distance is associated with the whole-body variation of the angular momentum and could reflect the instability of the gait.

As a vector in space, $\mathbf{d}_{\text{COM-MMA}}$ was projected onto the anteroposterior axis d_{AP} , the vertical axis d_{VERT} and the mediolateral axis d_{ML} . According to our hypothesis, only d_{AP} and d_{ML} were studied here. Both components denoted hereafter anteroposterior and mediolateral distances were then divided by the subject's height to obtain them in a dimensionless form. For each phase and for the total duration of the motion, the range of each dimensionless distance was computed.

3. Results

3.1. Spatiotemporal Parameters

Spatiotemporal parameters are presented in Table 1. The ANOVA results highlighted a significant speed effect ($p < 0.001$), but no group effect for the forward speed. Forward speed increased with stepping speed. A significant Group \times Speed interaction was also found for the forward speed. Post hoc analysis revealed significant differences in this parameter for all pairwise comparisons, except for the comparison between age groups in the spontaneous speed condition ($p > 0.05$).

Table 1. Mean \pm Std spatiotemporal parameters for young and older adults at spontaneous and fast speed. ‘ns’ indicates a non-significant effect ($p > 0.05$).

Parameters	Young Adults		Older Adults		Speed	Effect	
	Spontaneous	Fast	Spontaneous	Fast		Group	Speed \times Group
Total duration (s)	3.21 \pm 0.15	2.75 \pm 0.17	3.18 \pm 0.32	2.89 \pm 0.28	$p < 0.001$	ns	ns
P1 duration (s)	0.62 \pm 0.09	0.49 \pm 0.08	0.51 \pm 0.08	0.43 \pm 0.08	$p < 0.001$	$p = 0.003$	ns
P2 duration (s)	0.43 \pm 0.06	0.34 \pm 0.04	0.40 \pm 0.06	0.33 \pm 0.04	$p < 0.001$	ns	ns
P3 duration (s)	0.29 \pm 0.07	0.13 \pm 0.03	0.25 \pm 0.05	0.16 \pm 0.02	$p < 0.001$	ns	$p = 0.004$
P4 duration (s)	0.46 \pm 0.05	0.36 \pm 0.03	0.48 \pm 0.05	0.38 \pm 0.06	$p < 0.001$	ns	ns
P5 duration (s)	1.42 \pm 0.11	1.42 \pm 0.13	1.54 \pm 0.17	1.58 \pm 0.18	ns	$p = 0.003$	ns
Forward speed (m/s)	0.71 \pm 0.16	1.18 \pm 0.19	0.73 \pm 0.12	0.99 \pm 0.21	$p < 0.001$	ns	$p < 0.001$

Regarding the temporal parameters, a significant group effect was found for the duration of the phases P1 and P5 ($p = 0.003$). Compared to young adults, older adults exhibited a smaller duration of P1 and, conversely, a longer duration of P5. The ANOVA revealed a significant speed effect for the total duration of the five phases and for the durations of the phases P1, P2, P3 and P4 ($p < 0.001$). The durations of these phases decreased with speed. A significant Group \times Speed interaction was found for the duration of the phase P3 ($p = 0.004$). The post hoc analysis indicated significant differences in P3 duration for all pairwise comparisons, except for the comparisons between young and older adults in both the spontaneous and fast speed conditions ($p > 0.05$).

3.2. Distances between the COM and the MMA

The time evolution of the anteroposterior and mediolateral distances over the five phases is presented in Figure 1.

Along the anteroposterior axis, for both age groups and both speed conditions, the minimal distance throughout the entire stepping is observed at the end of the first swing phase (P2), while the maximum distance is observed at the beginning of the second swing phase (P4). Statistical analysis revealed that the minimal and maximal distances are significantly impacted by group and speed conditions (Table 2). The negative minimal distance increased with age ($p = 0.009$) and speed ($p < 0.001$). Similarly, the maximal distance increased with age ($p = 0.004$) and speed ($p < 0.001$).

Table 2. Minimal and maximal dimensionless distances ($\times 100$) along the anteroposterior (AP) and mediolateral (ML) axis throughout the entire stepping. ‘ns’ indicates a non-significant effect ($p > 0.05$).

Parameters	Young Adults		Older Adults		Speed	Effect	
	Spontaneous	Fast	Spontaneous	Fast		Group	Speed \times Group
Minimal AP distance	-2.27 \pm 0.35	-3.87 \pm 0.78	-2.81 \pm 0.47	-4.71 \pm 1.29	$p < 0.001$	$p = 0.009$	ns
Maximal AP distance	2.00 \pm 0.37	3.48 \pm 0.97	2.75 \pm 0.58	4.39 \pm 1.33	$p < 0.001$	$p = 0.004$	ns
Minimal ML distance	-1.15 \pm 0.35	-2.21 \pm 0.64	-1.35 \pm 0.17	-1.92 \pm 0.36	$p < 0.001$	ns	$p = 0.017$
Maximal ML distance	1.10 \pm 0.36	1.85 \pm 0.61	1.32 \pm 0.25	1.89 \pm 0.60	$p < 0.001$	ns	ns

Along the mediolateral axis, the minimal distance occurred at the end of the first swing phase (P2) and the maximal occurred at the transition between the double-stance phase (P3) and the second swing phase (P4), as observed along the anteroposterior axis. The ANOVA revealed no group effect for both minimal and maximal distances (Table 2). Nevertheless, a significant speed effect was found (Table 2). Both minimal ($p < 0.001$) and maximal ($p < 0.001$) absolute distances increased significantly with increasing speed. Also, a significant Group \times Speed interaction was found for the minimal mediolateral

distance. The post hoc analysis revealed significant differences in the minimal mediolateral for all pairwise comparisons, with the exception of the comparison between young and older adults in the fast speed condition ($p > 0.05$).

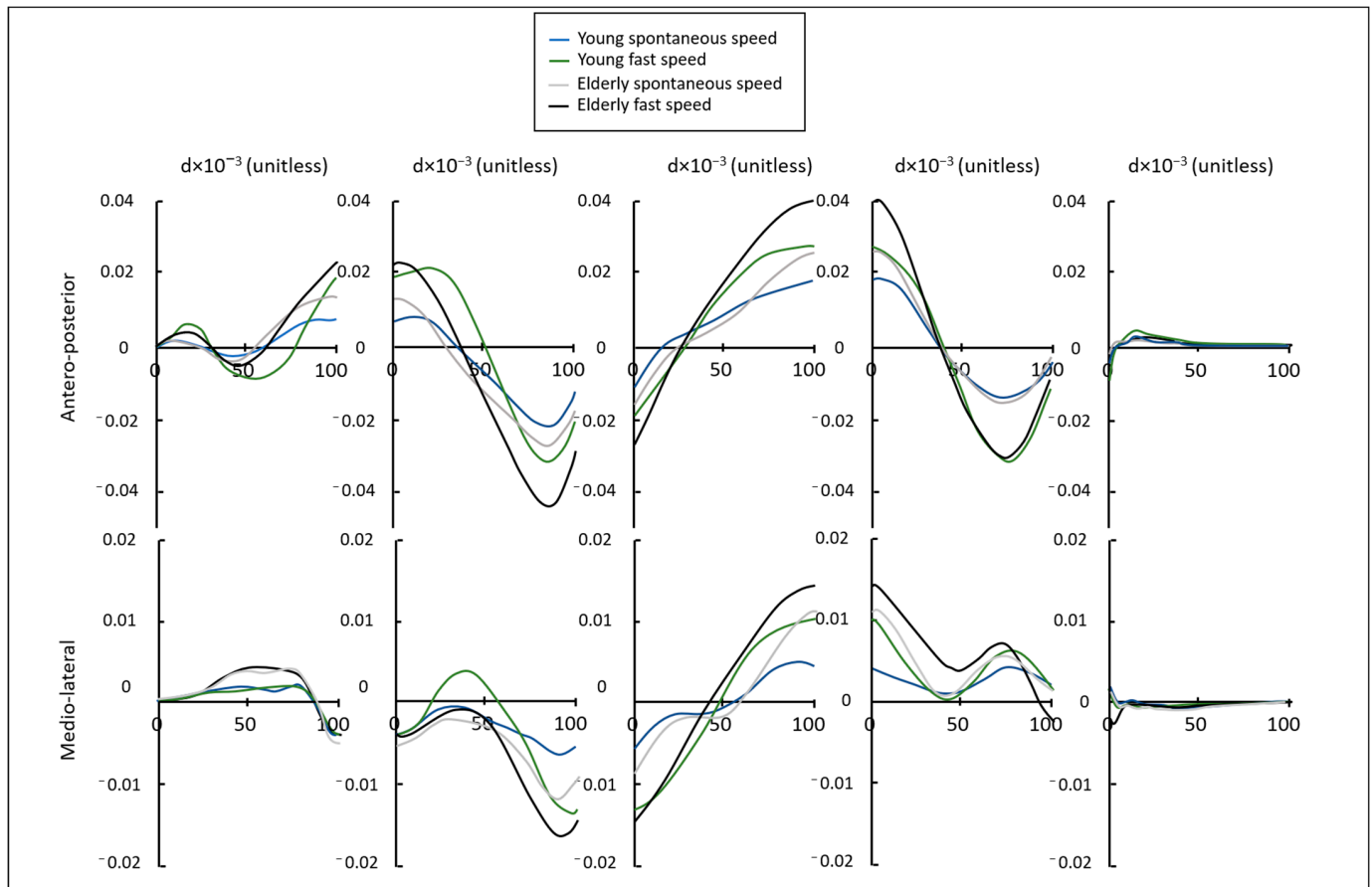


Figure 1. Means of the dimensionless anteroposterior and mediolateral distances between the COM and the MMA for older and young participants in both speed conditions. The five phases (P1 to P5) are represented from left to right. Each phase is normalized from 0 to 100%. Minimal distances are observed at the end of the first swing phase (P2). Maximal distances are observed at the transition between the phases P3 and P4. The amplitudes on the ordinate axes are different in the anteroposterior $[-0.04; 0.04]$ and mediolateral $[-0.02; 0.01]$ directions.

The range of the distances along the anteroposterior and mediolateral axes for all phases and over the entire stepping are presented in Table 3.

Along the anteroposterior axis, a significant group effect was found for the ranges of the distance COM-MMA in all phases, with the exception of P1 ($p > 0.05$). Compared to young adults, older adults exhibited larger range of the anteroposterior distance for P2, P3, P4, P5 and throughout the entire stepping. In addition, the results highlighted a significant effect of the speed on the ranges of the distance along the anteroposterior axis. Ranges of the distances increased with speed for all the phases. Statistical analysis revealed no effect of group \times speed interaction for the ranges of the distance between COM and MMA along the anteroposterior axis ($p > 0.05$).

Along the mediolateral axis, no group effect was revealed for the range of the distance between COM and MMA. However, the ANOVA revealed a significant speed effect for the ranges of the distance along the mediolateral axis, with an increase in this distance with increasing speed for all phases. Furthermore, a significant effect of the Group \times Speed interaction was observed for the ranges of mediolateral distance during P2 and P3. For these two phases, post hoc analysis indicated significant differences in the mediolateral distance for all pairwise comparisons, except for the comparison between young and older adults at fast speed ($p > 0.05$).

Table 3. Ranges of the dimensionless distances along the anteroposterior (AP) and mediolateral (ML) axis ($\times 100$) for each phase and throughout the entire stepping. ‘ns’ indicates a non-significant effect ($p > 0.05$).

Phase	Parameters	Young Adults		Older Adults		Speed	Effect	
		Spontaneous	Fast	Spontaneous	Fast		Group	Speed \times Group
P1	AP	1.48 \pm 0.48	3.10 \pm 1.02	2.18 \pm 0.78	3.62 \pm 1.34	$p < 0.001$	ns	ns
	ML	0.97 \pm 0.28	1.21 \pm 0.56	1.23 \pm 0.42	1.40 \pm 0.54	$p = 0.026$	ns	ns
P2	AP	3.14 \pm 0.59	6.02 \pm 1.79	4.19 \pm 0.95	7.02 \pm 2.26	$p < 0.001$	$p = 0.036$	ns
	ML	1.13 \pm 0.31	2.69 \pm 0.68	1.42 \pm 0.32	2.36 \pm 0.71	$p < 0.001$	ns	$p = 0.013$
P3	AP	3.08 \pm 0.73	5.00 \pm 1.70	4.30 \pm 0.81	6.95 \pm 2.58	$p < 0.001$	$p = 0.004$	ns
	ML	1.64 \pm 0.47	3.47 \pm 1.44	2.11 \pm 0.40	3.06 \pm 1.01	$p < 0.001$	ns	$p = 0.04$
P4	AP	3.54 \pm 0.56	6.23 \pm 1.21	4.46 \pm 0.93	7.49 \pm 1.83	$p < 0.001$	$p = 0.007$	ns
	ML	1.07 \pm 0.43	2.13 \pm 0.87	1.53 \pm 0.32	2.33 \pm 0.72	$p < 0.001$	ns	ns
P5	AP	1.09 \pm 0.21	1.85 \pm 0.51	0.90 \pm 0.21	1.52 \pm 0.49	$p < 0.001$	$p = 0.031$	ns
	ML	0.69 \pm 0.25	0.86 \pm 0.25	0.59 \pm 0.14	0.82 \pm 0.24	$p < 0.001$	ns	ns
Total	AP	4.27 \pm 0.64	7.35 \pm 1.64	5.56 \pm 0.95	9.10 \pm 2.53	$p < 0.001$	$p = 0.005$	ns
	ML	2.21 \pm 0.63	4.06 \pm 1.27	2.67 \pm 0.34	3.82 \pm 0.87	$p < 0.001$	ns	ns

4. Discussion

The aim of the present study was to investigate the distance between the MMA and the COM in young and older adults at different speeds during stepping. Consistent with our hypotheses, the results revealed that (1) older adults exhibited higher anteroposterior distances than their younger counterparts and (2) both anteroposterior and mediolateral distances increased with increasing speed during stepping. Our results also provide an answer to our second objective regarding the ability of the distances to discriminate the most unstable phases of the stepping. These results are further discussed in the following paragraphs.

Regarding the spatiotemporal results, it is interesting to note that the forward speed of the stepping was not different between both groups at spontaneous speed, but it was slower in older adults than their young counterparts in the fast speed condition. This result is in line with the literature, which has already shown that there were no significant differences in spontaneous walking speeds up to the age of 80 [36]. Thus, it appears justified to compare the dynamics of both groups throughout the entire motion. Although there is no group effect for the total duration, the results highlight a significant group effect for the duration of the phases P1 (APAs) and P5 (restabilization phase). Compared to young adults, older adults decreased the duration of the first phase ($p = 0.003$) and, on the contrary, increased the duration of the restabilization phase ($p = 0.003$). This result confirms previous findings of the literature [37], which showed that the time required to stop the locomotion was longer in older adults. Conversely, one can note that speed effect is observed for all the spatiotemporal parameters, meaning that the volunteers correctly followed the instructions accelerating the stepping during the fast speed condition.

The time evolution of the anteroposterior distance observed during stepping describes a profile consistent with that presented by Al-Abiad et al. [15] for healthy people. The minimal distance (i.e., the largest negative distance) is observed at the end of the single support phase of the first step (i.e., the swing phase of the dominant leg—phase P2). This result is consistent with the fact that during the swing phase, the base of support is reduced, thus increasing the difficulty to control the instability. In addition, just before the contralateral heel strike (i.e., during the swing phase of the non-dominant leg—phase P4), the significant shifts in the COP and the COM, which progress between the two supporting feet during the step transition, further increase instability. In the context of stepping, a lower negative anteroposterior distance is observed during the second swing phase (P4). This result is probably related with the beginning of the deceleration phase and the decrease in the dynamics of the participants. Indeed, participants were instructed to stop just after phase P4 at phase P5. The negative values observed during P2 and P4 indicate that the MMA is behind the center of mass. Knowing that the external vertical force acts upwards, this means that, according to Equation (13), the derivative of the angular momentum is negative and generates global forward angular acceleration of the body. This angular acceleration is stopped at the time of the heel strike, which marks the onset of the double support phase. Concerning the maximum values of the anteroposterior distance, they are observed at the beginning of the swing phases (P2 and P4). At this time, the MMA shifts beyond

of the COM and the positive value corresponds to a backward angular acceleration, meaning that an insufficient global positive angular speed of the body at the beginning of the swing phase could lead to falling backwards. This result is in accordance with the previous work of Manckoudian et al. [38].

In a similar manner, the minimal distance along the mediolateral axis is observed at the end of the first swing phase (P2). Accordingly, this means that the MMA is located more laterally towards the non-dominant leg than the COM, while the dominant leg is swinging. It confirms global positive angular acceleration around the anteroposterior axis and the fall of the body toward the swing leg. This lateral fall stops at the beginning of the double-stance phase (P3). Conversely, the maximal mediolateral distance is observed during the swing phase of the non-dominant leg (P4). During this phase, the mediolateral distance is positive, meaning an ipsilateral fall of the body, i.e., toward the non-dominant leg. In the middle of the double-stance phase P3, the sign of the distance changes from negative to positive value. This result highlights the transition of the COM from the dominant to the non-dominant side. All these results are in accordance with previous data [39,40]. Another interesting result is the evolution of the mediolateral distance during the APAs. During a large part of the phase P1, the distance along the mediolateral axis is positive and becomes negative at the time of heel lift of the dominant leg. This positive value highlights a displacement of the COM towards the non-dominant side before the swing of the dominant leg, which is in line with previous observations during the APAs [16–18].

Finally, whatever the phase, the variations in the distance along the anteroposterior axis are approximatively twice the variations along the mediolateral axis. This reflects the larger variations in the anteroposterior forces which orient alternatively the MMA forwards and backwards in the direction of the propulsive and braking forces.

Interestingly, our results showed a significant effect of age for anteroposterior distances, but not for mediolateral distances. Indeed, compared to their younger counterparts, older adults increased both minimal and maximal anteroposterior distances, which results in the larger range of the anteroposterior distance between COM and MMA during stepping. The larger range of anteroposterior distance in older adults is potentially associated with the larger range of angular momentum in the sagittal plane already observed in a previous study [24]. The detailed analysis of each phase also revealed an effect of age for the range of the distance along anteroposterior axis. These significant differences are observed for all stepping phases, except for phase 1, during which APAs are performed. This interesting result reveals larger reorientations of the MMA with respect to the position of the COM that could explain a poorest ability of older adults to control propulsion and braking forces. This result is in accordance with Al-Abiad et al. [15], who previously demonstrated larger anteroposterior distances between the COM and the MMA for populations with transfemoral amputations, which are known as being at risk of fall. The authors also demonstrated the preponderant role of the control of the anteroposterior force for the instrumented and healthy lower limb. Given the absence of significant differences with age along the mediolateral axis, present results suggest that the risk of falling forwards or backwards is higher than the risk of falling laterally during stepping in older adults. This result could explain the contradictory results obtained by the authors regarding the gait instability in the elderly compared to young adults [41,42]. However, it is worth noting that the lack of difference along the mediolateral axis could be explained by the fact that older adults included in the study were relatively young (68.7 years on average) and physically active.

Our results revealed a significant speed effect on the maximal and the minimal values of the anteroposterior and mediolateral distances. The results highlighted an average increase of 66% of the absolute values of the minimal and maximal distances between the spontaneous and the fast speed conditions. This result is consistent with those of previous studies, which highlighted an increase in the variation of angular momentum with speed during stepping [25]. Indeed, a larger angular momentum during the same time interval will lead to a higher derivative of the angular momentum and, according to Equation (13), to a higher distance. These results are also corroborated by the previous works of Bailly et al. [14], who also observed higher distances during locomotion at fast speed compared to self-selected one.

The detailed analysis of each phase revealed that during the phase P1, the range of mediolateral and anteroposterior distances increased with speed. The increase with speed is particularly more important for young adults (+109%) than older adults (+ 66%) along the anteroposterior axis. Conversely, along the mediolateral axis, the speed effect is more moderate, with an average increase of 19% for both age groups combined. As previously described [19,43], this period presents dynamics adjustments with variations of both the anteroposterior and mediolateral distances. This result confirms the COM displacement and the reorientation of the external forces before the toe-off. It allows

the subject to generate the force propulsion and the placement of the COM above the supporting foot at the beginning of the swing phase.

During the phases P2, P3 and P4, a speed effect was observed for the range of the mediolateral and the anteroposterior distances. However, this speed effect is higher along the mediolateral axis (+85%) compared to the anteroposterior axis (+70%). These increases in range occur while phase durations decrease, with a mean value of 29%. It therefore corresponds to larger dynamics variations in shorter times. Along the anteroposterior axis, the largest variations of the distances are observed during the second swing phase P4, while the largest variations along the mediolateral axis occur during the double-stance phase P3. This result corroborates previous studies which revealed the large lateral displacement of the COM during this phase [39]. Finally, the analysis of the restabilization phase (phase P5) showed that the subjects anticipated the end of the stepping. Indeed, in this phase, the variations of the anteroposterior and mediolateral distances reach the lowest values until the end of the motion. However, in this phase, there is no decrease in duration with the speed condition. For the entire stepping, the differences between the maximal and the minimal values of the distances confirm our hypothesis about the speed effect, while a group effect is observed only along the anteroposterior direction. Moreover, these larger variations of the distance are globally realized with a shorter time (−13%) at fast speed leading to greater dynamics variables of the subject and potentially higher instability.

While our results demonstrate higher instability during the swing phases, it is interesting to note that these swing phases represent the largest part of the stepping time. Indeed, at spontaneous speed, the swing phase corresponds to approximately 60% of the step composed of one swing phase and one double-stance phase. This result is reinforced at a fast speed, where the most unstable phase represents 70% of the step time without any group effect. Thus, it is an interesting result to note that the most unstable phases represent larger time percentage during the fast speed condition.

Moreover, as mentioned above, our results show that age affects the duration of APAs and the restabilization phase, often referred to as “corrective postural adjustments (CPA)”. Specifically, older adults exhibit a shorter APAs duration and, conversely, a longer restabilization duration compared to their young counterparts. It is an interesting result meaning that older adults are able to prepare their step more quickly, while they take longer to restabilize compared to young adults. These results are in line with the findings of the literature, which reveal that age modifies postural control strategies, with greater reliance on reactive versus anticipatory control mechanisms with aging [44]. This shift from anticipatory to reactive control with aging is thought to predispose the elderly to greater instability, making it more difficult for them to maintain balance during voluntary motor tasks or after an external perturbation [44]. This is corroborated by the present results, which show that the distances between COM and MMA increase with age. Nevertheless, it is important to note that some studies have reported an increase in APAs duration in people at high risk of falling, such as post-stroke people [45], compared to able-bodied adults. Taken together, these results suggest that the duration of APAs is a parameter that should be interpreted with caution, in particular when it is used to assess the instability of patients and therefore the risk of falls. In a clinical context, it therefore appears that the restabilization duration, or APC duration, as well as the distances between the COM and the MMA could be better criteria for quantifying instability. However, it remains evident that the restabilization time could not be used to monitor immediate fall risks, while the real-time analysis of the distances between the COM and the MMA could help to prevent falls if some thresholds can be established in the future.

Indeed, the distances which are calculated in this study are closely related to the variation of angular momentum, which has been previously identified as a good candidate to monitor the instability of gait motion [7,10]. However, it is important to insist on the necessity for the human being to generate these variations of angular momentum to initiate and produce the motion of the COM. Considering the nature of the human joints that essentially work in rotation, the variation in angular momentum is inherent to the generation of human motion. Thus, as human locomotion is naturally unstable, angular momentum and therefore the distances cannot remain zero during stepping. However, it is also obvious from a dynamic point of view that the variations in the angular momentum must be performed within certain limits to prevent an uncontrolled movement which could lead to falls. The challenge is to determine the proper limits for angular momentum variation and the distances that allow us to respect both the mobility and control of this inherent instability.

Finally, it is also interesting to note that the distances between the COM and the MMA were able to discriminate unstable conditions whatever the anthropometry of the subjects. Indeed, as a dimensionless parameter, no correlation was observed in the results presented here between the weight and the height of the subjects and the dimensionless distances.

5. Conclusions

The main results of this study revealed an effect of age and speed on the distance between COM and MMA during a stepping task. Older adults exhibited larger distances between COM and MMA along the anteroposterior direction, compared to young adults. In particular, older adults displayed larger minimal and maximal anteroposterior distances, resulting in a larger range of the distance over the entire stepping movement. These age-related differences underscore that older adults may experience higher difficulties in controlling instability during stepping movement. With increasing speed, both older and younger adults increased their body dynamic, resulting in larger anteroposterior and mediolateral distances. Finally, these results confirm that the distance between COM and MMA appears to be able to discriminate individuals at risk of falling and that these appear to be the most unstable phases of the locomotion, as the distance increases during the most critical phase with reduced support surface. Therefore, further works should determine the thresholds of the distance along both the anteroposterior and mediolateral axis, which could inevitably lead to falls.

Author Contributions: Conceptualization, B.W., J.B., T.C. and H.P.; methodology, B.W., J.B., T.C. and H.P.; validation, B.W. and H.P.; formal analysis, B.W. and H.P.; investigation, B.W.; resources, J.B. and T.C.; data curation, J.B. and B.W.; writing—original draft preparation, B.W.; writing—review and editing, B.W., J.B., T.C. and H.P.; visualization, B.W.; supervision, B.W.; project administration, B.W. and T.C.; funding acquisition, T.C. All authors have read and agreed to the published version of the manuscript.

Funding: This work was partially supported by the French national research agency (ANR), through the HoBiS [ANR-18-CE27-0010], eWALKING [ANR-22-CE19-0009-01] and BAC2WALK [ANR-22-CE19-0003] projects.

Institutional Review Board Statement: The study was conducted in accordance with the Declaration of Helsinki and approved by the Institutional Review Board of IRISSE, EA 4075.

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study. Written informed consent has been obtained from the volunteers to publish this paper.

Data Availability Statement: Data are available on reasonable request from the corresponding author.

Conflicts of Interest: The authors declare no conflict of interest.

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