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► **To cite this version:**

Jérémie Begue, Nicolas Peyrot, Georges Dalleau, Teddy Caderby. Effect of increasing speed on whole-body angular momentum during stepping in the elderly. *Journal of Biomechanics*, 2021, 122, pp.110436. 10.1016/j.jbiomech.2021.110436 . hal-03338701

HAL Id: hal-03338701

<https://hal.science/hal-03338701>

Submitted on 9 May 2023

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TITLE PAGE

**Effect of increasing speed on whole-body angular momentum during
stepping in the elderly**

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Keywords (5): Step initiation; Balance; Aging; Rotational dynamics; Speed

Article type: Short communication

Word count in the abstract: 218/250

Word count in the manuscript (Introduction through Discussion): 1990/2000

Word count in the Acknowledgements section: 30

Number of figures: 2

1 **Abstract**

2 Recent evidence suggests that older adults may have difficulty controlling whole-body
3 angular momentum (H) during volitional stepping, which could impose a major challenge for
4 balance control and result in potential falls. However, it is not known if and how H is
5 influenced by speed when stepping. This study aimed to investigate the effect on H of
6 increasing speed during step initiation in older adults. Twenty-seven healthy individuals over
7 60 were enrolled in the current study and were instructed to perform a series of step initiations
8 with their dominant leg under two speed conditions: at preferred speed and as fast as possible.
9 Two force plates and a motion-capture system were used to record H and the components of
10 the net external moment (moment arms and ground reaction forces) during the double support
11 and step execution phases of stepping. Results revealed that increasing speed of stepping
12 affected H differently in both stepping phases and in the different planes. H ranges in all three
13 planes increased with speed during the double support phase. During the step execution
14 phase, while H ranges in frontal and transversal planes decreased, sagittal plane H range
15 significantly increased with speed. This increased H range in the sagittal plane, which may
16 result from the task demands, could impose a greater challenge for balance control in the
17 elderly.

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1 **1.Introduction**

2 Evidence suggests that whole-body angular momentum (H) is highly controlled by
3 humans to ensure safety and the efficient realization of daily motor tasks (Herr and Popovic,
4 2008; Maldonado et al., 2018; Robert et al., 2009). Poor H control has been associated with
5 poor balance (Vistamehr et al., 2016) and an increased risk of falling during locomotion
6 (Pijnappels et al., 2005). Thus, H constitutes a meaningful metric to better understand the
7 underlying mechanisms of falls in people suffering from balance disorders, such as the elderly
8 (Neptune and Vistamehr, 2018).

9 It has been previously reported that, contrary to young adults, older adults were unable
10 to fully reduce H after tripping, which decreased the balance recovery success and thus
11 predisposed them to a fall (Pijnappels et al., 2005). In line with these findings, a recent study
12 has shown that aging also altered the control of H during volitional stepping initiation (Begue
13 et al., 2019). Compared to younger adults, older adults exhibited smaller H ranges during the
14 double support phase. Conversely, they had higher H ranges during step execution, which
15 may impose a greater challenge for balance control and a potentially higher risk of falling
16 during this phase. However, this previous study only examined stepping movement at
17 preferred speed. Consequently, it remains unclear whether these age-related changes in H
18 were associated with the slower speed recorded in older adults rather than a poor H control.

19 Increasing gait speed has been found to decrease H ranges during walking (Bennett et
20 al., 2010; Silverman and Neptune, 2011) and running (Sepp et al., 2019), which could impose
21 a lower demand for balance control. However, to our knowledge, although previous studies
22 have investigated the influence of speed on step initiation and gait (e.g., Brenière et al., 1987;
23 Caderby et al., 2014; Ito et al., 2003; Lepers and Brenière, 1995), none have addressed the
24 question of if and how increasing speed influences H during stepping initiation in older
25 adults. Given the significant proportion of falls in the elderly when initiating stepping

1 (Robinovitch et al., 2013), such knowledge could be particularly useful in designing
2 interventions to help prevent falls. This study aimed, therefore, to examine the influence of
3 increasing speed on H during stepping initiation in healthy older adults.

4

5 **2.Methods**

6 Twenty-seven healthy individuals over the age of 60 (24 females and 3 males, 66 ± 3.3
7 years, $1.57 \pm 0.07\text{m}$, $60 \pm 9.8\text{kg}$) volunteered for this study. Participants were physically
8 active, free of any neurological or musculoskeletal disorders and did not report any falls
9 during the twelve months prior to the study. All gave written consent after being fully
10 informed of the testing procedure, which was approved by the local ethics committee.

11 Initially, participants stood barefoot on a first force plate (60x40cm, AMTI, USA) in a
12 natural upright posture, as still as possible, with their arms alongside their trunk. After a
13 verbal cue from the investigator, they were instructed to initiate stepping with their dominant
14 leg and to follow through with their non-dominant leg, stopping in a comfortable upright
15 position with arms alongside their trunk and feet parallel (Cimolin et al., 2017; Tirosh and
16 Sparrow, 2004) onto a larger, second force plate (100x80cm, Sensix, France) located
17 immediately in front of the first one. After each trial, the participants repositioned themselves
18 in their standardized foot positions previously marked on the first force plate (McIlroy and
19 Maki, 1997). Participants were asked to initiate step under two speed conditions: at preferred
20 speed and as fast as possible. The order of speed conditions was randomized. After two
21 familiarization trials, five trials per condition were completed.

22 Retro-reflective spherical markers were fixed on participants' bony landmarks
23 according to a simplified biomechanical model consisting of 9 segments (head + torso, right
24 and left arms, forearms + hands, thighs, legs + feet) (Tisserand et al., 2016). Three additional
25 markers fixed bilaterally on the hallux, calcaneus and the head of the fifth metatarsal defined

1 foot segment. A twelve-camera motion-capture system (Vicon, UK) was used to
2 simultaneously collect kinematic and force plate data (200Hz and 1000Hz, respectively)
3 during the stepping movement.

4 Data analysis methods were described in a previous study (Begue et al., 2019) and are
5 briefly described here. Kinematic and force plate data were low-pass filtered using a zero-lag
6 fourth order Butterworth filter with a 10Hz cut-off frequency. The center of pressure (CoP)
7 coordinates were calculated using force plate data according to the manufacturer's
8 instructions. Body center of mass (CoM) position in the three dimensions was computed as
9 the weighted sum of each body segment's CoM from the 9-segment model (Tisserand et al.,
10 2016). Time rate of change of H (\dot{H}), which is equivalent to the net external moment about the
11 body's CoM, was calculated as follows:

$$12 \quad \vec{\dot{H}} = \vec{r} \times \overline{\text{GRF}} + \vec{T}$$

13 where \vec{r} is the moment arm vector from the body's CoM to CoP, $\overline{\text{GRF}}$ is the ground reaction
14 force vector and \vec{T} is the free moment vector (Fig.1A). H was obtained using numerical
15 integration of \dot{H} (Pijnappels et al., 2005). H obtained from this method has been found to be
16 comparable with that computed from method using kinematic data (Collins et al., 2009; Herr
17 and Popovic, 2008).

18 Step initiation was divided into two phases: a double support phase (the time delay
19 between the onset of the movement to the toe-off of the swing leg) and a step execution phase
20 (the time between the swing toe-off and the swing heel-contact). Spatiotemporal parameters
21 such as progression velocity, step length, step width and duration of double support and step
22 execution phases were computed. Peak-to-peak ranges of H in the three dimensions
23 (difference between maximum and minimum values of H) and peak external components of
24 the net external moment (peak moment arms, GRFs, free vertical moment (T_v)) were
25 measured in both phases of stepping. **Net external moment (\dot{H}) was normalized by the**

1 participant's weight and height. H was normalized by the product of participant mass,
2 participant height (l) and $\sqrt{g \cdot l}$ (with $g=9.81\text{m}\cdot\text{s}^{-2}$) (Vistamehr et al., 2014).

3 After checking for data normality and homoscedasticity, paired t -tests were conducted
4 on each dependent variable in order to test the effect of speed. The Benjamini-Hochberg
5 procedure was used to correct for multiple comparisons (Benjamini and Hochberg, 1995).
6 Statistical significance was set at $p<0.05$.

7

8 **3.Results**

9 A significant speed effect was found on all spatiotemporal parameters (Table 1): step
10 width and step length increased, while durations of both the double support and execution
11 phases decreased with speed.

12 Typical traces of H and net external moment are represented for both speed conditions
13 in Figure 1.B. A speed effect was found on H ranges during both phases of stepping
14 movements (Fig.2). During the double support phase, H ranges in the three planes increased
15 significantly with speed; however, during step execution, H ranges decreased in the frontal
16 and transversal planes but increased in the sagittal plane as speed increased.

17 Regarding the components that contribute to the time rate of change of H we found
18 that peak GRFs, moment arms and T_v increased with speed during the double support phase,
19 while vertical moment arm was unchanged (Table 2; $p>0.05$). During step execution,
20 anteroposterior and mediolateral moment arms increased, while vertical moment arm
21 decreased significantly. Similarly, anteroposterior, mediolateral and vertical GRFs increased
22 with speed, while T_v did not vary ($p>0.05$).

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1 **4.Discussion**

2 This study aimed to examine the effect of speed increase on H during volitional
3 stepping initiation in healthy older adults. The reduction of the double support phase duration
4 as a function of speed is in agreement with previous findings of Caderby et al. (2014) and
5 indicates a decrease in the time of application of the net external moment. At the same time,
6 we also noted that H ranges in the three planes increased with speed. These speed-related
7 changes in H ranges are the result of increased net external moments in all three planes, which
8 are dependent on GRFs, moment arms and T_v . Specifically, in the frontal plane, an increase in
9 both the mediolateral moment arm (+7.7%) and vertical GRF (+1.6%) tends to result in a
10 higher negative external moment (directed toward the stance leg), despite the larger negative
11 mediolateral GRF (+8.6%). In the sagittal plane, an increase in anteroposterior GRF (+80.5%)
12 leads to a higher positive sagittal external moment (backward), despite the increase in both
13 the vertical GRF and anteroposterior moment arm (+75.5%). Finally, in the transversal plane,
14 the increase in the anteroposterior GRF, mediolateral moment arm and T_v (+46.1%) result in a
15 higher positive external moment (directed toward the stance leg), despite an increase in both
16 the mediolateral GRF and the anteroposterior moment arm.

17 Current results are consistent with previous findings showing that CoM-CoP distances
18 (moment arms) along the anteroposterior and mediolateral directions increase with
19 progression velocity during the postural phase of stepping (Brenière et al., 1987; Caderby et
20 al., 2014; Ito et al., 2003; Lepers and Brenière, 1995). It is generally accepted that the
21 separation between CoM and CoP during this phase contributes to creating the propulsive
22 forces necessary to reach the desired velocity at the end of the first step (Lepers and Brenière,
23 1995) and to promote mediolateral stability during step execution (Caderby et al., 2014;
24 McIlroy and Maki, 1999). Thus, one may speculate that the increase in H ranges as a function

1 of speed during the double support phase reflects a strategy to develop a faster progression
2 velocity during stepping.

3 During step execution, we observed that frontal and transversal H ranges decreased
4 with speed, which is consistent with results of previous studies on walking and running
5 (Bennett et al., 2010; Sepp et al., 2019; Silverman and Neptune, 2011). Nonetheless, it should
6 be noted that the normalization of H by speed could have contributed to the decrease in H
7 range as a function of speed in these previous studies, which was not the case in the current
8 study. In our study, the decrease in frontal and transversal H ranges are the result of both a
9 reduction in step execution duration (-20.4%) and changes in the net external moments. In
10 particular, the decrease in frontal H range is associated with the increase in mediolateral GRF
11 (+23.4%). This results in a smaller positive external moment (directed toward the swing leg),
12 despite the small decrease in vertical moment arm (-0.6%) and the increase in both the
13 vertical GRF (+7.2%) and mediolateral moment arm (+19.1%). In the transversal plane, the
14 decrease in H range as a function of speed is associated with increases in mediolateral GRF
15 and anteroposterior moment arm (+43.5%), resulting in a smaller negative external moment
16 (directed toward the swing leg), despite increases in mediolateral moment arm and
17 anteroposterior GRF (+51.7%).

18 Unlike the frontal and transversal planes, the range of H in the sagittal plane increased
19 as a function of speed despite the reduction in step execution duration. This is the result of the
20 increase in vertical GRF and anteroposterior moment arm eliciting a higher negative external
21 moment (forward) with the increase in speed. Thus, the increase in anteroposterior GRF was
22 not sufficient to compensate for the higher forward external moment when participants
23 increased their speed. At this stage, it is not known whether this increase in sagittal H range is
24 due to insufficient propulsive force production ability in the elderly, as emphasized in the
25 literature (Franz and Kram, 2013), or to the task demands. In any case, as H must be restored

1 in the subsequent phases to halt the body's rotation in the forward direction, the increase in H
2 during step execution may place a greater demand on the balance control system. Importantly,
3 Pijnappels et al. (2005) have shown that older adults have difficulties restoring H in the
4 sagittal plane during the single leg phase following tripping, which decreased the balance
5 recovery success and thus predisposed them to a fall. In the present study, there were no
6 recorded falls or difficulty stopping after initiating step. Nevertheless, it should be noted that
7 our participants were active and healthy. It would be worthwhile to examine the speed-related
8 changes in H with more vulnerable elderly people, to determine whether such changes may
9 impose a greater risk of falling during stepping.

10

11 **Conflict of interest statement**

12 The authors declare that they have no financial or other conflicts of interest in regards to this
13 research.

14

15 **Acknowledgements**

16 Research grant from the Réunion Regional Council and from the European Regional
17 Development Fund (FEDER) supported this work. The authors wish to thank Mrs Leila
18 Laipala for her technical assistance.

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Table 1. Mean of spatiotemporal parameters for preferred and fast speeds.

	Preferred	Fast	<i>P-value</i>	Percent change (%)
Progression velocity (m·s ⁻¹)	0.69 ± 0.13	0.91 ± 0.21	<i>p</i> <0.001*	+32.9 ± 15.7
Step width (cm)	17.7 ± 2.26	18.4 ± 2.37	<i>p</i> <0.01*	+4.0 ± 7.5
Step length (cm)	52.9 ± 8.88	59.3 ± 12.5	<i>p</i> <0.001*	+11.8 ± 11.2
Double support phase duration (s)	0.58 ± 0.10	0.50 0.11	<i>p</i> <0.001*	-13.6 ± 11.1
Step execution phase duration (s)	0.41 ± 0.08	0.32 ± 0.06	<i>p</i> <0.001*	-20.4 ± 14.1

Percent change (%) corresponds to the difference between fast and preferred speed conditions, expressed in percentage of the value measured at preferred speed condition. * Significant difference between speed conditions after using the Benjamini-Hochberg procedure.

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Table 2. The normalized components of time rate of change of H (net external moment) for preferred and fast speeds in the two phases of step initiation.

	Preferred	Fast	<i>P</i> -value	Percent change (%)
Double support phase				
Peak AP GRF (%BW)	0.085 ± 0.023	0.146 ± 0.031	<i>p</i> <0.001*	+80.5 ± 48.7
Peak ML GRF (%BW)	-0.068 ± 0.012	-0.073 ± 0.013	<i>p</i> <0.05*	+8.6 ± 18.1
Peak Vertical GRF (%BW)	1.070 ± 0.030	1.087 ± 0.051	<i>p</i> <0.05*	+1.6 ± 3.8
Peak AP moment arm (% <i>l</i>)	0.042 ± 0.012	0.069 ± 0.013	<i>p</i> <0.001*	+75.5 ± 59
Peak ML moment arm (% <i>l</i>)	0.040 ± 0.006	0.043 ± 0.007	<i>p</i> <0.05*	+7.7 ± 15.9
Peak Vertical moment arm (% <i>l</i>)	0.562 ± 0.009	0.561 ± 0.010	<i>NS</i>	-0.3 ± 0.8
Peak $T_v \times 10^{-3}$ (%BW × <i>l</i>)	2.660 ± 0.686	3.750 ± 0.974	<i>p</i> <0.001*	+46.1 ± 39
Step execution phase				
Peak AP GRF (%BW)	0.131 ± 0.037	0.195 ± 0.060	<i>p</i> <0.001*	+51.7 ± 30.3
Peak ML GRF (%BW)	0.095 ± 0.012	0.116 ± 0.016	<i>p</i> <0.001*	+23.4 ± 14
Peak Vertical GRF (%BW)	1.057 ± 0.031	1.133 ± 0.062	<i>p</i> <0.001*	+7.2 ± 4.9
Peak AP moment arm (% <i>l</i>)	0.083 ± 0.021	0.120 ± 0.032	<i>p</i> <0.001*	+43.5 ± 26.4
Peak ML moment arm (% <i>l</i>)	0.061 ± 0.008	0.072 ± 0.008	<i>p</i> <0.001*	+19.1 ± 11.3
Peak Vertical moment arm (% <i>l</i>)	0.548 ± 0.009	0.544 ± 0.013	<i>p</i> <0.05*	-0.6 ± 1.2
Peak $T_v \times 10^{-3}$ (%BW × <i>l</i>)	-4.146 ± 1.322	-4.313 ± 1.858	<i>NS</i>	+3.7 ± 31.2

Peak ground reaction forces (GRFs) were normalized by body weight (BW), moment arms were normalized by body height (*l*), and free vertical moment (T_v) was normalized by the product of *l* and BW. ML: mediolateral; AP: anteroposterior. Percent change (%) corresponds to the difference between fast and preferred speed conditions, expressed in percentage of the

1 value measured at preferred speed condition. * Significant difference between speed
2 conditions after using the Benjamini-Hochberg procedure. *NS*: non-significant difference
3 ($p>0.05$).

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6 **Figure Captions**

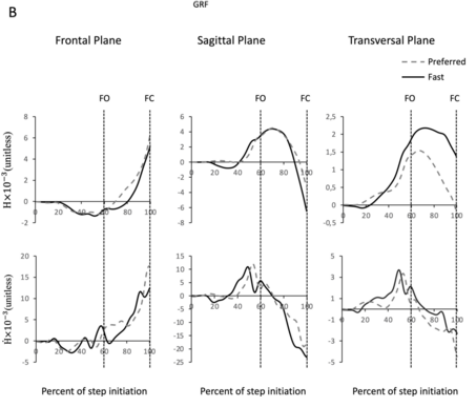
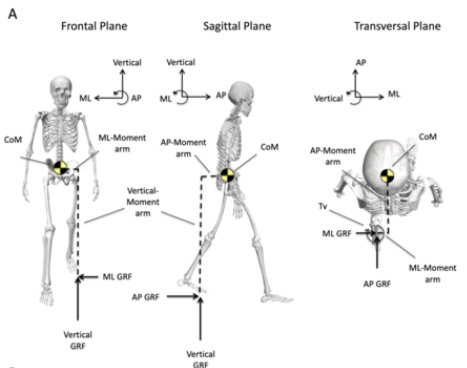
7 **Figure 1:** (A) Representation of net external moment components about the body's center of
8 mass (CoM) in the frontal, sagittal and transversal planes during stepping. The frontal angular
9 momentum, sagittal angular momentum and transversal angular momentum were defined
10 about anteroposterior, mediolateral and vertical axes, respectively. The GRFs, the free vertical
11 moment (T_v) and the moment arms appear in black arrows, grey arrows and dashed lines,
12 respectively. The current figure only represents the left leg contribution to the net external
13 moment. However, the right leg also contributes to net external moment, particularly when
14 the limb is in contact with the ground and must be considered in the calculation of the net
15 external moment. (B) Typical traces of normalized whole-body angular momentum (H) and
16 normalized net external moment (\dot{H}) obtained for one representative participant during the
17 step initiation movement for preferred and fast speed conditions (one trial). FO: instant of the
18 swing foot-off; FC: instant of the swing foot contact.

19

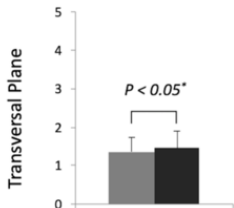
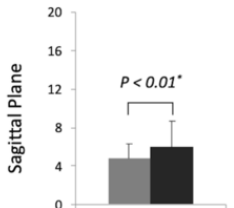
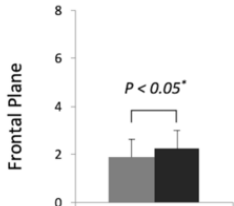
20 **Figure 2:** The frontal, sagittal and transversal mean ranges of normalized whole-body angular
21 momentum (H) in the two phases of step initiation for the preferred and fast speed conditions.

22 H was normalized by body mass, body height and $\sqrt{g \cdot l}$ ($g = 9.81 \text{ m}\cdot\text{s}^{-2}$ and $l = \text{body height}$).

23 * Significant difference between speed conditions after using the Benjamini-Hochberg
24 procedure.

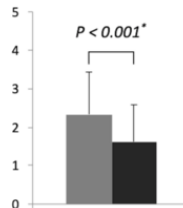
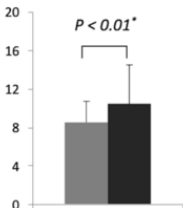
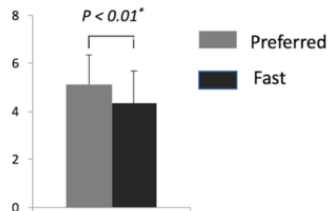


H range $\times 10^{-3}$ (unitless)



Double support phase

H range $\times 10^{-3}$ (unitless)



Step execution phase