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TITLE PAGE

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

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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,  $66 \pm 3.3$   
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

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

	<b>Preferred</b>	<b>Fast</b>	<b><i>P-value</i></b>	<b>Percent change (%)</b>
Progression velocity (m·s <sup>-1</sup> )	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 (%)
<b>Double support phase</b>				
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 <sub>v</sub> × 10 <sup>-3</sup> (%BW × <i>l</i> )	2.660 ± 0.686	3.750 ± 0.974	<i>p</i> <0.001*	+46.1 ± 39
<b>Step execution phase</b>				
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 <sub>v</sub> × 10 <sup>-3</sup> (%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<sub>v</sub>) 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 ( $p > 0.05$ ).

4  
5

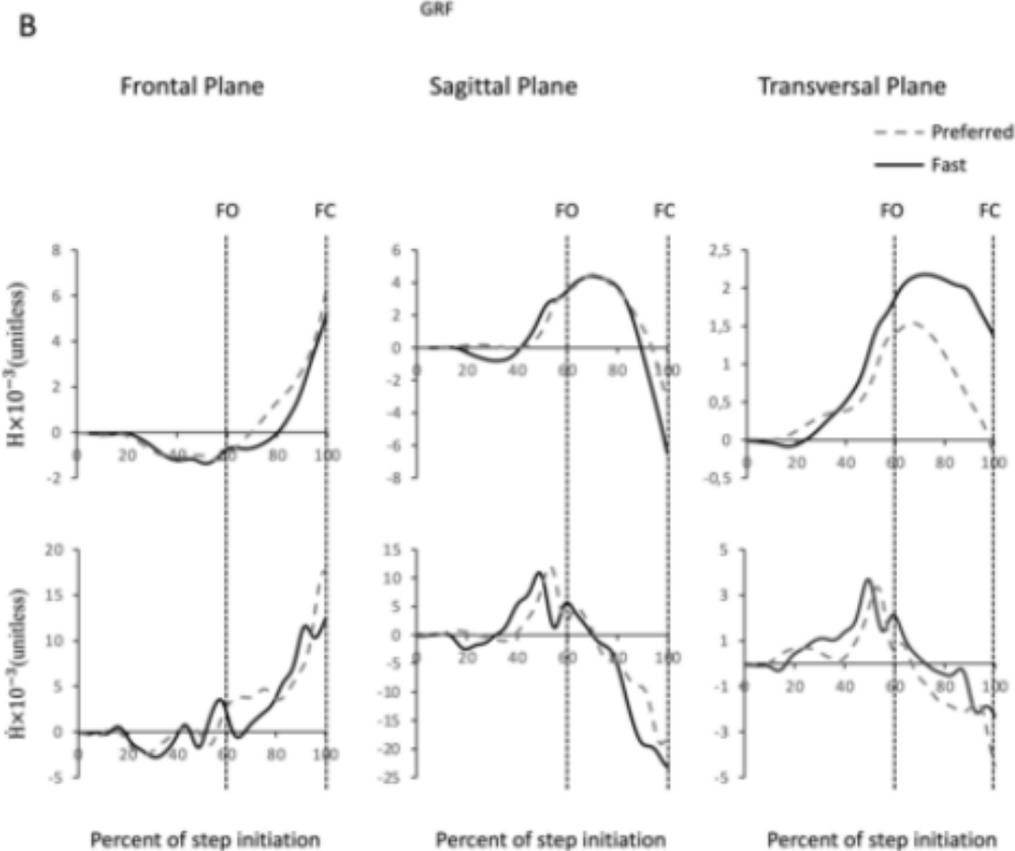
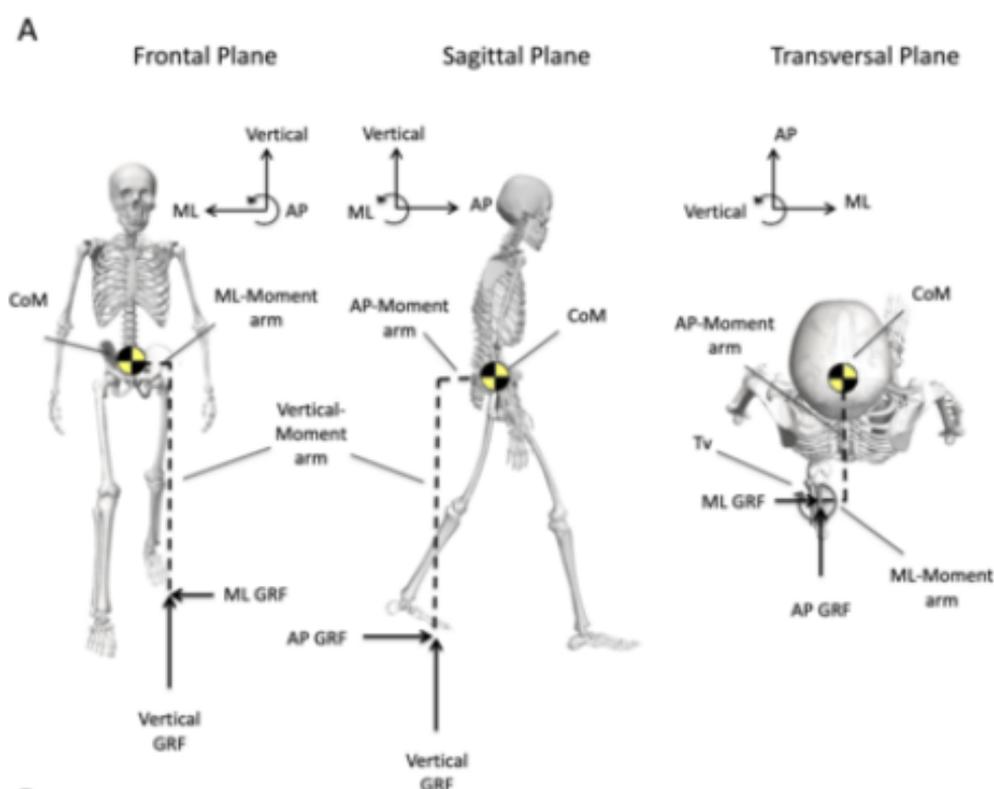
## 6 **Figure Captions**

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.

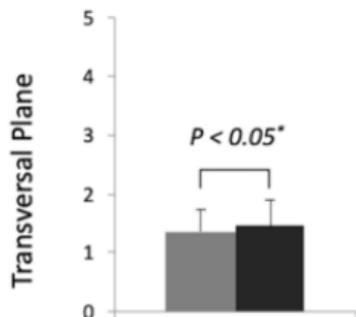
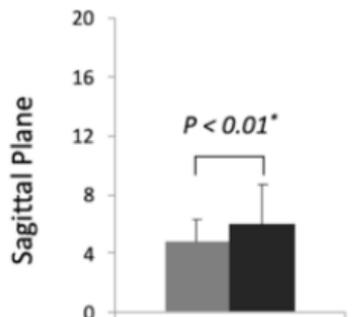
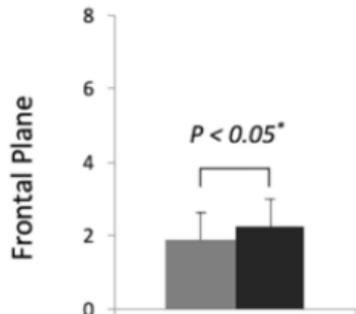
19

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.

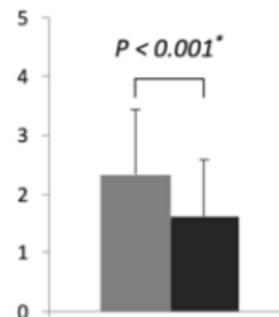
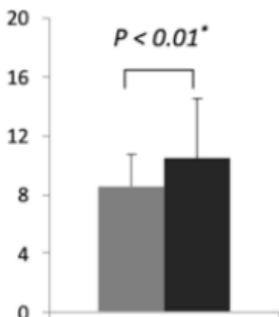
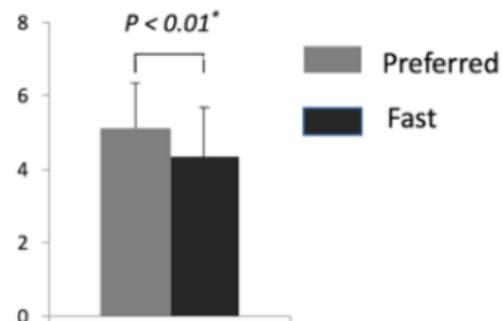


H range  $\times 10^{-3}$  (unitless)



Double support phase

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



Step execution phase