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TITLE PAGE

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

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Abstract

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

1.Introduction

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

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

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

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

2.Methods

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

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

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

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

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

$$\vec{\dot{H}} = \vec{r} \times \overrightarrow{GRF} + \vec{T}$$

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

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

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

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

3.Results

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

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

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

4.Discussion

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

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

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

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

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

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

Conflict of interest statement

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

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

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

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**	***P-value***	**Percent change (%)**
Double support phase				
Peak AP GRF (%BW)	0.085 ± 0.023	0.146 ± 0.031	***p<0.001****	+80.5 ± 48.7
Peak ML GRF (%BW)	-0.068 ± 0.012	-0.073 ± 0.013	***p<0.05****	+8.6 ± 18.1
Peak Vertical GRF (%BW)	1.070 ± 0.030	1.087 ± 0.051	***p<0.05****	+1.6 ± 3.8
Peak AP moment arm (%l)	0.042 ± 0.012	0.069 ± 0.013	***p<0.001****	+75.5 ± 59
Peak ML moment arm (%l)	0.040 ± 0.006	0.043 ± 0.007	***p<0.05****	+7.7 ± 15.9
Peak Vertical moment arm (%l)	0.562 ± 0.009	0.561 ± 0.010	*NS*	-0.3 ± 0.8
Peak T_v $\times 10^{-3}$ (%BW× l)	2.660 ± 0.686	3.750 ± 0.974	***p<0.001****	+46.1 ± 39
Step execution phase				
Peak AP GRF (%BW)	0.131 ± 0.037	0.195 ± 0.060	***p<0.001****	+51.7 ± 30.3
Peak ML GRF (%BW)	0.095 ± 0.012	0.116 ± 0.016	***p<0.001****	+23.4 ± 14
Peak Vertical GRF (%BW)	1.057 ± 0.031	1.133 ± 0.062	***p<0.001****	+7.2 ± 4.9
Peak AP moment arm (%l)	0.083 ± 0.021	0.120 ± 0.032	***p<0.001****	+43.5 ± 26.4
Peak ML moment arm (%l)	0.061 ± 0.008	0.072 ± 0.008	***p<0.001****	+19.1 ± 11.3
Peak Vertical moment arm (%l)	0.548 ± 0.009	0.544 ± 0.013	***p<0.05****	-0,6 ± 1.2
Peak T_v $\times 10^{-3}$ (%BW× l)	-4.146 ± 1.322	-4.313 ± 1.858	*NS*	+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

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

Figure Captions

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

Figure 2: The frontal, sagittal and transversal mean ranges of normalized whole-body angular momentum (H) in the two phases of step initiation for the preferred and fast speed conditions. H was normalized by body mass, body height and $\sqrt{g \cdot l}$ (g = 9.81 m·s^{-2} and l = body height). * Significant difference between speed conditions after using the Benjamini-Hochberg procedure.

A

Frontal Plane

Sagittal Plane

Transversal Plane

Vertical

ML

AP

CoM

ML-Moment arm

AP-Moment arm

Vertical-Moment arm

ML GRF

AP GRF

Vertical GRF

Tv

B

Preferred

Fast

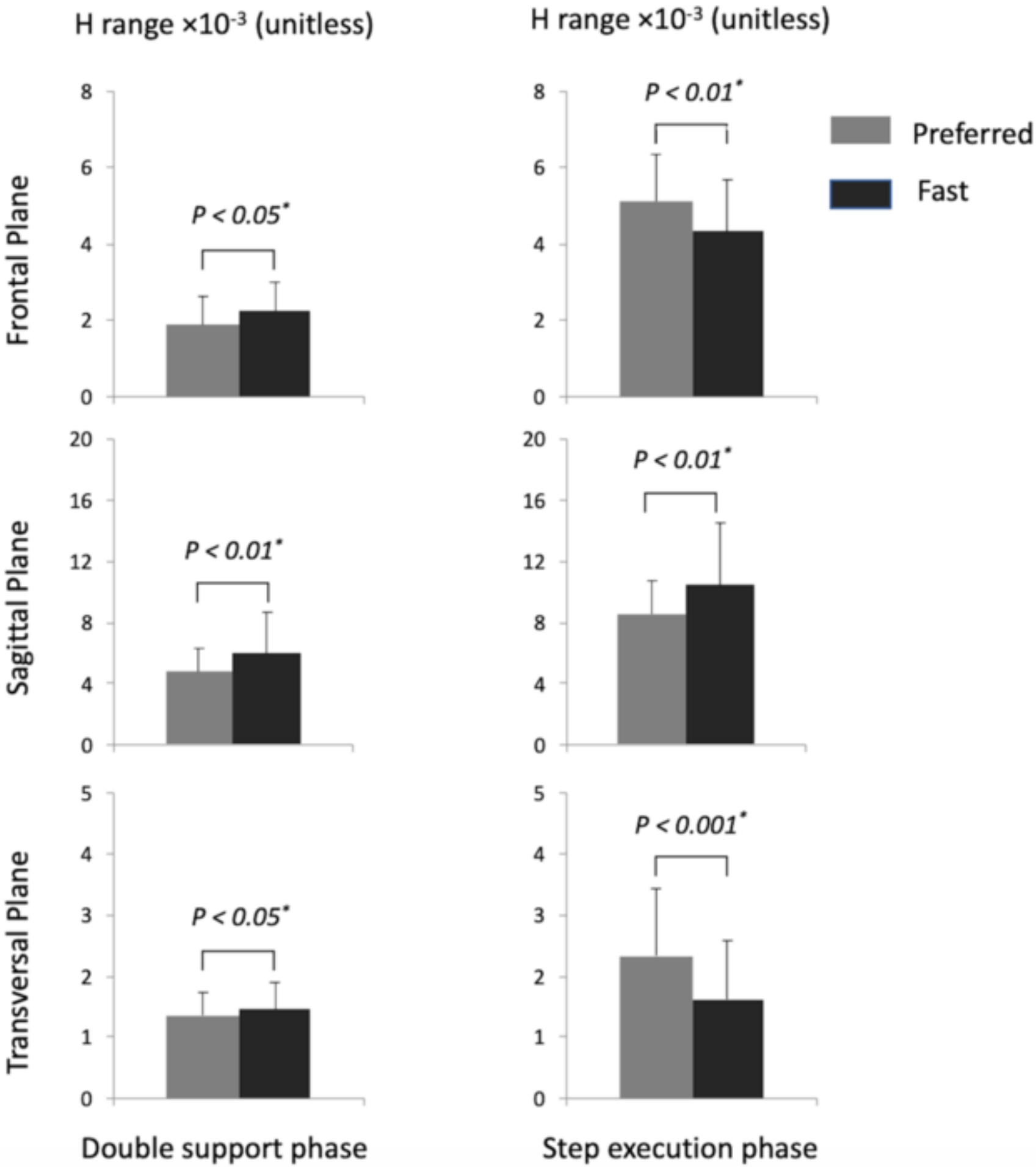
FO

FC

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

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

Percent of step initiation


H range ×10-3 (unitless)
H range ×10-3 (unitless)
Frontal Plane
Sagittal Plane
Transversal Plane
P < 0.05*
P < 0.01*
P < 0.01*
P < 0.01*
P < 0.05*
P < 0.001*
Preferred
Fast
Double support phase
Step execution phase