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TITLE PAGE

Age-related changes in the control of whole-body angular momentum during stepping

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Declarations of Interest

None of the authors have financial or other conflicts of interest with regard to this research.

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1 **Abstract:**

2
3 *Background:* Appropriate control of whole-body angular momentum (H) is crucial to
4 maintain dynamic balance and thus avoid falling during daily activities. Poor H control ability
5 during locomotion has been found in people with an increased risk of falling, such as post-
6 stroke patients and amputees. In contrast, little is known about the control of H during
7 locomotion in the elderly. The aim of this study was to investigate whether and how aging
8 influences three-dimensional H control during initiation of stepping.

9
10 *Methods:* Twenty-two healthy old and 22 healthy young individuals were instructed to
11 perform a series of initiation of stepping with their dominant leg and at their self-selected
12 preferred pace. Two force plates and a motion capture system were used to record H, the net
13 external moment about the body's center of mass and components of this net external moment
14 (moment arms and ground reaction forces) during the double support and step execution
15 phases of stepping.

16
17 *Results:* In the double support phase, older participants exhibited smaller peak-to-peak ranges
18 of H in the sagittal and transversal planes compared to their younger counterparts. These
19 results were explained by decreased net external moments in both planes in the older
20 participants. Conversely, during the step execution phase, older adults had higher peak-to-
21 peak ranges of H in the frontal and sagittal planes compared to the younger adults. These
22 higher ranges of H were associated with a longer duration of the step execution phase.
23 Furthermore, in the sagittal plane, a higher external moment also contributed to increasing
24 peak-to-peak ranges of H in older adults.

25
26 *Conclusion:* The current study revealed that older and younger adults exhibit different control
27 strategies of H during initiation of stepping. The age-related changes, which may emphasize a
28 higher difficulty to control H in the older adults, could impose a higher **challenge** for balance
29 control and a potentially higher risk of falling during the step execution phase in this
30 population.

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37 **Keywords:** Balance, Step initiation, Elderly, Angular momentum, Biomechanics
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1 **1. Introduction**

2
3 Maintaining balance is critical for performing efficiently in activities of daily living as well as
4 for avoiding falls. Balance maintenance implies appropriate regulation of the linear and
5 angular momenta of the body segments (Hof, 2007; Hofmann et al., 2009; Winter, 1995). The
6 sum of angular momenta produced by the rotation of body segments about the body's center
7 of mass (CoM), commonly referred to as whole-body angular momentum (H), is a mechanical
8 quantity that characterizes the rotational behavior of the whole body. This parameter has been
9 used for several decades by roboticists to explore approaches to maintaining balance and to
10 develop control strategies to enable biped robots to walk without falling down (Goswami and
11 Kallem, 2004; Kajita et al., 2003; Macchietto et al., 2009). More recently, researchers have
12 investigated H in humans during various motor tasks (Herr and Popovic, 2008; Imura and Iino,
13 2018; Maldonado et al., 2017; Martelli et al., 2013; Nolasco et al., 2019; Sepp et al., 2019).
14 Importantly, these works suggest that H is highly controlled by the central nervous system for
15 maintaining balance (Herr and Popovic, 2008; Maldonado et al., 2018; Robert et al., 2009).
16 Poor H control has been found in people with an increased risk of falling, such as amputees
17 (Pickle et al., 2014; Sepp et al., 2019; Silverman and Neptune, 2011), post-stroke patients
18 (Nott et al., 2014; Vistamehr et al., 2016, 2018), and also in able-bodied adults in situations
19 where balance is challenged by external perturbations (Martelli et al., 2013). Therefore, H
20 constitutes a powerful metric to gain insight into balance control strategies and underlying
21 mechanisms of falls in humans (Neptune and Vistamehr, 2018).

22
23 While H has been widely used to investigate balance control in populations with mobility
24 impairments, little attention has been given to understanding the effects of normal aging on
25 the control of H. To the best of our knowledge, only one study aimed to compare H between
26 young and older adults during the recovery phase of tripping (Pijnappels et al., 2005). In this
27 previous study, tripping was experimentally induced by a sudden appearance of an obstacle
28 while participants walked at their self-selected speeds on a walkway. Results revealed that,
29 contrary to young adults, older adults were unable to fully reduce H in the sagittal plane
30 during the push-off phase following tripping. This inability to adequately control H decreased
31 the balance recovery success after tripping and thus predisposed older adults to a fall. Despite
32 the efforts made by these authors, it remains unclear whether and how aging affects the
33 control of H during daily volitional activities. Yet, it has been observed that most falls in the
34 elderly occur during these activities, including walking and transition tasks such the initiation
35 of gait or stepping (Robinovitch et al., 2013). Identifying the presence and underlying
36 mechanisms of balance control impairments during these daily motor tasks can be useful for
37 designing interventions to reduce the incidence of falls among the elderly.

38
39 Initiation of stepping consists of the volitional transition from a **quiet standing** posture to a
40 dynamic phase during the step execution, which may challenge postural balance to a greater
41 extent than steady-state walking (Maslivec et al., 2018; Nagano et al., 2013). Stepping
42 initiation is generally composed of two phases: a double support phase preceding the foot-off
43 (FO) of the swing leg, during which “anticipatory postural adjustments” are developed,
44 followed by a step execution phase ending at the time of swing foot contact (FC) (Brunt et al.,
45 1999; Yiou and Do, 2011). During the double support phase, it is well established that the
46 center of pressure (CoP) initially shifts backward and laterally toward the swing foot, which
47 acts to propel the CoM forward and toward the stance limb. Afterwards, the CoP rapidly
48 shifts toward the future stance leg, which results in breaking the CoM motion toward that
49 limb (Elble et al., 2004; Jian et al., 1993; Yiou et al., 2017). This stereotypical pattern of CoP
50 and CoM displacement during the double support phase enables to create the mechanical

1 conditions for forward progression as well as promote mediolateral balance during the
2 subsequent step execution (Caderby et al., 2014; Hansen et al., 2016; Lepers and Brenière,
3 1995; Lyon and Day, 1997; Yiou et al., 2016). Nevertheless, it is worth noting that this may
4 lead to a rotational instability during stepping. The separation between CoM and CoP, which
5 is the point of application of the ground reaction force (GRF), may be responsible for a net
6 external moment about the CoM. This net external moment, which is equal to the moment
7 caused by the GRF plus the free vertical moment, is equivalent to the time rate of change of H
8 (Nott et al., 2014; Pijnappels et al., 2004; Vistamehr et al., 2014). Given that H is related to
9 rotational impulse, an increase in net external moment and/or its time of application will
10 result in a higher range of H, which in turn may impose a greater challenge to balance control
11 and a risk of falling (Neptune and Vistamehr, 2018; Pijnappels et al., 2005).

12
13 Previous studies which examined the effect of aging on initiation of stepping reported that
14 older adults exhibited a slower progression velocity, shorter step and longer step duration than
15 their younger counterparts (Halliday et al., 1998; Hurt and Grabiner, 2015; Luchies et al.,
16 2002; Mercer et al., 1997; Patla et al., 1993; Singer et al., 2013). Furthermore, lower
17 magnitude of the anteroposterior GRF (Henriksson and Hirschfeld, 2005) and smaller
18 backward CoP shift during the double support phase have been found in the elderly compared
19 to young people (Halliday et al., 1998; Khanmohammadi et al., 2015; Polcyn et al., 1998).
20 These age-related motor adaptations could potentially result in a smaller net external moment
21 during the double support phase in the elderly and so have been interpreted as a compensatory
22 strategy for enhancing balance while stepping. However, during the subsequent phase, the
23 longer step execution duration (the longer time of application of net external moment)
24 observed in older adults could potentially induce a greater change of H and impose a greater
25 **challenge** for balance control. This could be supported by study results of Buckley et al.
26 (2010) which revealed that older adults had greater vertical axis body rotation and were
27 unable to exert the same control during the execution phase of stepping down when compared
28 with younger participants. While no age-related difference was found in the double support
29 phase, the findings suggested that older people might have difficulties controlling whole-body
30 rotational dynamics during stepping, particularly during the step execution (single support)
31 phase.

32
33 The current study aimed to examine the effects of aging on H control during initiation of
34 stepping. We hypothesized that aging would alter the control of H during step initiation.
35 Specifically, we assumed that the ranges of H in frontal, sagittal and transversal planes would
36 be different between young and older individuals.

37 38 **2. Methods**

39 40 **2.1 Participants**

41
42 A total of forty-four people participated in this study: 22 healthy young adults and 22 healthy
43 older adults. Characteristics of the participants are presented in Table 1. All individuals were
44 physically active and did not report having any falls during the previous twelve months before
45 the study began. They had no recent history of neurological, musculoskeletal or any disorders
46 that could affect their normal gait or balance. All participants gave their written consent after
47 being fully informed of the test procedure. The study protocol was in accordance with the
48 Declaration of Helsinki and approved by the local institutional review board (IRISSE,
49 EA4075).

2.2 Experimental procedure

Participants stood barefoot in a natural upright posture with their arms alongside their trunk on an initial force plate (60 x 40 cm, AMTI, USA). They were instructed to stand as still as possible with their body weight distributed evenly between their legs and to look forward. After receiving a verbal signal, participants stepped forward at their self-selected preferred pace onto a larger, second force plate (100 x 80 cm, SENSIX, France) located immediately in front of the first one. Both force plates, embedded in a walkway, measured GRFs and moments. Retroreflective spherical markers (14 mm diameter) were fixed on bony landmarks according to a simplified biomechanical model (Tisserand et al., 2016b): bilaterally on the scapular acromial tip, humeral lateral epicondyle, ulnar styloid process, anterior superior spine iliac, femoral lateral epicondyle, fibular apex of lateral malleolus, and a single marker at the center of posterior iliac spines. In addition, other markers were fixed bilaterally on the hallux, head of the fifth metatarsal and calcaneus in order to define the foot segment. A motion capture system equipped with 12 cameras (6 Bonita cameras and 6 Vero cameras; Vicon, UK) was used to simultaneously collect the kinematic data at 200 Hz and force plate data at 1000 Hz.

Participants were asked to initiate step with their dominant leg and to follow through with the non-dominant leg to stop on the second force plate in a comfortable upright posture. The dominant leg, determined as the limb used for kicking a ball (Tsuji et al., 2015), was maintained throughout the experiment. After each trial, the participants repositioned themselves in the standardized foot position previously marked on the first force plate, i.e., with the heels spaced 17 cm apart and the feet out-toeing by 14° (McIlroy and Maki, 1997). Data acquisition was triggered when participants were motionless and at least 1 s before the verbal signal from the experimenter. After two familiarization trials, each individual performed five trials from which data were collected.

2.3 Data analysis

All data were analyzed offline using a custom-made MatLab program (MatLab R2017a, The MathWorks, Inc., Natick, MA, USA). Both the kinematic and force plate data were low-pass filtered using a zero-lag fourth order Butterworth filter with a 10 Hz cut-off frequency. The anteroposterior and mediolateral CoP coordinates were calculated from force plate data in accordance with the manufacturer's instructions from the following equations:

$$\text{CoP}_{\text{AP}} = (M_{\text{ML}} + d_{\text{v}} \times F_{\text{AP}}) / F_{\text{v}} - d_{\text{AP}}$$

$$\text{CoP}_{\text{ML}} = (-M_{\text{AP}} + d_{\text{v}} \times F_{\text{ML}}) / F_{\text{v}} - d_{\text{ML}}$$

where M_{AP} and M_{ML} are respectively the moments about the anteroposterior and mediolateral axes measured at the center of the force plate; where F_{AP} , F_{ML} , and F_{v} are anteroposterior, mediolateral and vertical GRF, respectively; and where d_{AP} , d_{ML} and d_{v} are respectively anteroposterior, mediolateral and vertical distances between the geometric center of the force-plate surface and the true origin of the force-plate. These distances, which represent offsets due to problems in the manufacturing process, are provided by the manufacturer (AMTI). In the coordinate system used, the force along the mediolateral direction (X axis) is positive to the right, the force along the anteroposterior axis (Y) is positive in the direction of the step

1 movement and the force along the vertical axis (Z) is positive upwards; a positive moment
2 was defined as a moment in the anticlockwise direction.

3 The three-dimensional coordinates of the body's CoM position were computed as the
4 weighted sum of each body segment's CoM from a 9-segment biomechanical model validated
5 in young and older adults by Tisserand et al. (2016b). The velocity and acceleration of the
6 body's CoM were calculated as the first and second time derivatives of the body's CoM
7 position, respectively.

8
9 To quantify the control of H, we analysed the time rate of change of H (\dot{H}) in the frontal,
10 sagittal and transversal planes (Nott et al., 2014; Vistamehr et al., 2018). The time rate of
11 change of H is equivalent to the net external moment about the body's CoM (\vec{H}) and was
12 calculated by the cross product of the moment arm vector and GRF vector plus the free
13 moment vector as follows:

$$15 \quad \vec{H} = \vec{r} \times \overline{GRF} + \vec{T}$$

16 where \vec{r} is the moment arm vector from the body's CoM to CoP, \overline{GRF} is the vector of GRF
17 and \vec{T} is the free moment vector (Figure 1). It is worth noting that the free moment
18 components about the anteroposterior (T_{AP}) and mediolateral (T_{ML}) axes were null (i.e., $T_{AP} =$
19 0 and $T_{ML} = 0$), unlike the free moment about the vertical axis (T_V). T_V is the frictional torque
20 about the vertical axis originating at the CoP and resulting shear forces between the foot and
21 the ground (Begue et al., 2018; Holden and Cavanagh, 1991).

22
23 Specifically, the net external moment about the body's CoM in the frontal plane, i.e., about
24 the anteroposterior axis (Y), was calculated as follows:

$$26 \quad \dot{H}_{\text{Frontal}} = (r_V \times F_{ML}) - (r_{ML} \times F_V)$$

27
28 where F_{ML} and F_V are the GRFs along the mediolateral (X) and vertical (Z) axes, respectively;
29 and where r_V and r_{ML} are the vertical and mediolateral moment arms (i.e., the distance
30 between CoM and CoP), respectively. By convention, the frontal external moment was
31 defined positive towards the swing leg (anticlockwise) and negative towards the stance leg
32 (clockwise). To preserve this sign convention, the frontal external moment calculation was
33 negated when the swing leg was the left side.

34
35
36 The net external moment about the body's CoM in the sagittal plane, i.e., about the
37 mediolateral axis (X), was calculated as follows:

$$39 \quad \dot{H}_{\text{Sagittal}} = (r_{AP} \times F_V) - (r_V \times F_{AP})$$

40
41 where F_V and F_{AP} are the GRFs along the vertical (Z) and anteroposterior (Y) axes,
42 respectively; and where r_{AP} and r_V are the anteroposterior and vertical moment arms (i.e., the
43 distance between CoM and CoP), respectively. By convention, the sagittal external moment
44 was considered positive backwards (anticlockwise) and negative forwards (clockwise).

45
46 The net external moment about the body's CoM in the transversal plane (about the vertical
47 axis (Z)) was calculated from the following formula:

1
2 $\dot{H}_{\text{Transversal}} = (r_{\text{ML}} \times F_{\text{AP}}) - (r_{\text{AP}} \times F_{\text{ML}}) + T_v$
3

4 where F_{AP} and F_{ML} are the GRFs along the anteroposterior (Y) and mediolateral (X) axes,
5 respectively; where r_{ML} and r_{AP} are the mediolateral and anteroposterior moment arms (i.e.,
6 the distance between CoM and CoP), respectively; and where T_v is the free vertical moment.
7 By convention, transversal external moment was defined positive towards the stance leg
8 (anticlockwise) and negative towards the swing leg (clockwise). To preserve this sign
9 convention, the transversal external moment calculation was negated when the swing leg was
10 the left side.

11
12 T_v was calculated from the following formula:

13
14 $T_v = M_v - (\text{CoP}_{\text{ML}} \times F_{\text{AP}}) + (\text{CoP}_{\text{AP}} \times F_{\text{ML}})$

15 where M_v is the moment acting about the vertical axis at the center of the force plate, CoP_{ML}
16 and CoP_{AP} are the CoP coordinates along the mediolateral and anteroposterior axes,
17 respectively; and F_{ML} and F_{AP} are the GRFs along the mediolateral and anteroposterior axes,
18 respectively.

19 Several temporal events of the stepping movement were determined. The onset of step
20 initiation (t_0) was detected from the anteroposterior and mediolateral accelerations of the CoM.
21 To be precise, as the onset of movement on both the anteroposterior and mediolateral axes did
22 not necessarily occur simultaneously (Caderby et al., 2014), t_0 was detected once one of these
23 acceleration signals deviated 2.5 standard deviations from its baseline value (Caderby et al.,
24 2017). Time of FO of the swing leg was determined when the vertical position of the fifth
25 metatarsal marker increased by 5 mm from its mean position during the initial upright posture.
26 FC corresponded to the instant when the vertical force signal of the second force plate
27 exceeded 10 N.

28 29 **2.4 Dependant variables** 30

31 **In the current study, we have voluntarily focused our analysis on the initiation phase of**
32 **stepping.** Step initiation movement was divided into a double support phase and a step
33 execution phase, as in Buckley et al. (2010). Duration of the double support phase
34 corresponded to the time delay between t_0 and the swing FO. Duration of the step execution
35 phase corresponded to the time between the swing FO and the swing FC. Forward progression
36 velocity was quantified by the peak of the anteroposterior CoM velocity reached at the end of
37 the first step (Brenière et al., 1987). Step length and step width were calculated respectively as
38 the anteroposterior and mediolateral distances between the heel markers of the dominant and
39 non-dominant legs at the instant of swing heel contact. In order to provide non-dimensional
40 measures and to correct as much as possible for unequal stature of the participants, these
41 spatiotemporal parameters were normalized by the person's height, according to Hof (1996).
42

43 To quantify the variation in H, we calculated the mean external moment about the body's
44 CoM (Vistamehr et al., 2018) and also the peak external moment components, i.e., peak GRFs,
45 moment arms and T_v (Silverman and Neptune, 2011) in both phases of stepping (double
46 support and step execution). The range of H, defined as the difference between the maximum
47 and minimum values of H, was also computed in both stepping phases, with H computed as

1 the area under the \dot{H} curve (Pijnappels et al., 2005). To provide non-dimensional measure and
2 decrease between-subject variability, the mean external moment was normalized by the
3 participant's weight and height. The range of H was meanwhile normalized by their mass,
4 height and $\sqrt{g \cdot l}$, where g corresponds to the gravitational constant and l is the person's
5 height (Vistamehr et al., 2014).

6 7 8 **2.5 Statistical analysis**

9
10 Dependant variables were averaged over five trials per adult. After checking for data
11 normality and homoscedasticity, Student t -tests were conducted on each of these variables to
12 test for differences between young and older groups. A Mann-Whitney U test was applied for
13 non-normally distributed variables. The level of statistical significance was set at $p < 0.05$. The
14 statistical analysis was performed using Statistica (Version 8.0, StatSoft Inc., Tulsa, USA).

15 16 **3. Results**

17 18 **3.1 Spatiotemporal parameters**

19
20 The non-normalized and normalized values of the spatiotemporal parameters are presented in
21 Table 2. Without normalization, both the peak forward progression velocity and step length
22 were significantly smaller in older adults than in the young individuals. Furthermore, the
23 duration of the step execution phase was longer in older adults than in the young controls. No
24 age effect was found for the other non-normalized spatiotemporal parameters ($p > 0.05$). After
25 normalization, only the duration of step execution was significantly different between both
26 groups. This duration was longer in older adults than in the young individuals. No age effect
27 was found for the other normalized parameters ($p > 0.05$).

28 29 **3.2 Whole-body angular momentum and net external moment**

30
31 *Description of the biomechanical traces.* Net external moment and H in each plane are
32 presented in Figure 2. Net external moment and H described a biphasic pattern during
33 stepping in young and old participants alike. In the frontal plane (about the anteroposterior
34 axis), net external moment was mostly negative during the double support phase, whereas it
35 was positive during the step execution phase. Thus, in the first part of the step initiation
36 (double support phase), the negative external moment in the frontal plane resulted in a body
37 rotation toward the stance leg (counter clockwise angular momentum); and in the second part
38 of the step initiation (step execution phase), the positive external moment acts to rotate the
39 body toward the swing leg (clockwise angular momentum). In the sagittal plane (about the
40 mediolateral axis), the positive external moment produced in the double support phase leads
41 to a clockwise angular momentum, i.e., a body rotation backwards. During the step execution
42 phase, the previous positive net external moment progressively decreases to become negative,
43 thus leading to a decrease in H that reaches a value near zero at the time of swing FC. In the
44 transversal plane, the net external moment is positive in the first part of the step initiation and
45 is negative in the second part. Thus, the body rotates toward the support leg (counter
46 clockwise angular momentum) during the double support phase and toward the swing leg
47 (clockwise angular momentum) during the step execution phase.

48
49 *Range of H .* During the double support phase, older adults exhibited significantly smaller
50 ranges of H in both the transversal and sagittal planes than their young counterparts (Figure 3).

1 However, in the frontal plane, no significant difference was found between the two age
2 groups ($p>0.05$). Conversely, during the step execution phase, the ranges of H in the frontal
3 and sagittal planes were significantly higher in older adults than in the young controls. No
4 statistical difference was revealed in the transversal plane between both groups ($p>0.05$).

5
6 *Net external moment.* During the double support phase, mean values of the net external
7 moment in both the sagittal and transversal planes were significantly smaller in the older adult
8 group (Figure 4). In contrast, no difference was found in the frontal plane between both age
9 groups ($p>0.05$). During the step execution phase, the mean value of the net external moment
10 in the sagittal plane was more significant in older compared to young individuals, while no
11 striking difference was observed in the frontal or transversal planes ($p>0.05$).

12
13 *GRFs and moment arms.* The values of peak GRFs and moment arms, i.e., the components
14 that contribute to net external moment, are presented in Table 3. Statistical analysis revealed
15 that older adults generated smaller anteroposterior moment arm, anteroposterior GRF and T_v
16 during the double support phase compared to young participants. During the step execution
17 phase, older adults generated significantly smaller anteroposterior and vertical GRFs. No age
18 effect was found on the other components of the external moment in both the double support
19 and step execution phases ($p>0.05$).

20 21 22 **4. Discussion**

23
24 The aim of this study was to investigate the age-related changes in the control of H during
25 initiation of stepping. Our main results revealed that older compared to younger individuals
26 had smaller peak-to-peak ranges of H in the sagittal and transversal planes during the double
27 support phase but had higher ranges of H in the frontal and sagittal planes during the step
28 execution phase, which supports our hypothesis.

29
30 During the double support phase, we observed that both the anteroposterior GRF peak and
31 anteroposterior CoM-CoP distance, i.e., anteroposterior moment arm, were smaller in older
32 compared to younger participants, which is consistent with findings from previous studies
33 (Halliday et al., 1998; Henriksson and Hirschfeld, 2005; Khanmohammadi et al., 2015;
34 Polcyn et al., 1998). It is important to consider that a reduced anteroposterior GRF would
35 decrease the positive (backward) net external moment in the sagittal plane. Conversely, a
36 decrease in the anteroposterior moment arm (posteriorly directed) during this phase would
37 result in a higher positive net external moment in the sagittal plane, due to a reduction in the
38 negative (forward) sagittal plane moment generated by vertical GRF. Our results found that
39 the effect of anteroposterior GRF reduction (by 18 %) was larger than that of the decrease in
40 anteroposterior moment arm (16 %), resulting in a significantly smaller positive external
41 moment in the sagittal plane in older adults compared to their young counterparts during the
42 double support phase. Similarly, in the transversal plane, we observed a smaller positive
43 (directed toward the stance leg) external moment in older participants during the double
44 support phase.

45
46 It should also be noted that the reduction in anteroposterior moment arm in older participants
47 would result in a greater positive (directed toward the stance leg) net external moment in the
48 transversal plane, due to a decrease in the negative (directed toward the swing leg) transversal
49 plane moment generated by mediolateral GRF. However, this was counteracted mainly by the
50 larger effect of the decrease in the positive T_v (by 28 %) and also by the effect of the decrease

1 in anteroposterior GRF, which reduced the positive external moment in the transversal plane
2 in older compared to young participants. Nevertheless, there was no significant difference in
3 the net external moment in the frontal plane during the double support phase (directed toward
4 the stance leg), as no age-related change was observed in the underlying components, i.e.,
5 mediolateral and vertical GRF and moment arms. Thus, as the duration of the double support
6 phase (the time of application) did not differ between the two age groups, the decrease in net
7 external moments in both the sagittal and transversal planes explained the lower ranges of H
8 in these two planes in the older adults. This decrease in H ranges may impose a lower demand
9 for balance control during the double support phase in older compared to young adults. These
10 results could be interpreted as a compensatory strategy implemented by older adults to
11 enhance balance, which is consistent with previous observations made by some authors
12 (Luchies et al., 2002; Patla et al., 1993). Interestingly, it was pointed out that such a strategy
13 could be indicative of a poorer control of balance in the elderly (Tisserand et al., 2016a). By
14 extrapolation, it could be hypothesized that these age-related changes in H during the double
15 support phase could reveal a reduced H control ability in older adults. **It is not ruled out that
16 these age-related changes in H during the double support phase may contribute to the
17 differences in H between young and older adults during step execution, since the initial
18 mechanical conditions created during the double support phase predetermine the body motion
19 during the subsequent execution phase (Lepers and Brenière, 1995; Lyon and Day, 1997).**
20

21 During the step execution phase, we noted that the ranges of H in the frontal and sagittal
22 planes were higher in older adults than in younger participants. These results may in part be
23 due to the longer duration of step execution in the older adult group. By lengthening the step
24 execution duration (by 10 %), older adults increased the time of application of the external
25 moment during this phase, which may contribute to the higher change in H compared to
26 young adults. Further, higher H ranges may to a greater extent be the result of larger net
27 external moments in the sagittal (45 % higher in older adults) and frontal (27 %) planes
28 compared to young adults, although this was statistically significant only in the sagittal plane.
29 It should be noted that the higher negative (forward) net external moment in the sagittal plane
30 may theoretically be a result of an increase in moment induced by the vertical GRF
31 (depending on both the vertical GRF and the anteroposterior moment arm) and/or a decrease
32 in external moment generated by the anteroposterior GRF (depending on the anteroposterior
33 GRF and the vertical moment arm). In the current study, we found that the anteroposterior
34 and vertical moment arms during step execution did not differ between the young and older
35 groups. In contrast, we noted that both vertical and anteroposterior GRF peaks were reduced
36 in older participants compared to young controls during step execution. From a biomechanical
37 point of view, reduced vertical GRF would decrease negative net external moment in the
38 sagittal plane, while decreased anteroposterior GRF peak would increase it. Thus, it seems
39 that the greater negative net external moment in the sagittal plane in the older adults would be
40 mainly a result of decreased anteroposterior GRF, which could potentially be attributed to
41 muscular weakness, reduced flexibility and/or a conservative strategy (Franz, 2016; Iosa et al.,
42 2014). Our results suggest that anteroposterior GRF reduction in the elderly contributes to
43 increasing the sagittal plane H range during stepping. Anteroposterior GRF reduction was
44 also found during the prosthetic leg stance phase of walking in below-knee amputees, leading
45 to greater ranges of H in the sagittal plane at a range of gait speeds compared to non-amputees
46 (Silverman and Neptune, 2011).
47
48

49 From a mechanical perspective, net external moment and H are related to body's CoM
50 dynamics (Herr and Popovic, 2008; Hof et al., 2007). It has been shown that H may be

1 modulated, for example by rotating the arms or the trunk, to control the state of the body's
2 CoM relative to the base of support and thus, to ensure balance during motor tasks (Hof et al.,
3 2007; Hofmann et al., 2009). Conversely, poor H control, as highlighted by large H variations,
4 may place a greater challenge on the control of body CoM and consequently may disrupt
5 balance (Kent et al., 2019; Martelli et al., 2013; Neptune and Vistamehr, 2018). In the current
6 study, the larger sagittal and frontal H ranges during step execution could impose a higher
7 challenge for balance control, which could potentially increase the risk of falling in the
8 elderly. This is supported by findings from Pijnappels et al. (2005) who observed that the
9 range of H in the sagittal plane was larger in older compared to young adults during the push-
10 off phase after tripping, i.e., the single leg stance phase following the collision of the swing
11 foot with an obstacle. These authors revealed that an insufficient reduction in H after tripping
12 was associated with a higher rate of falls in older compared to young adults. Other studies
13 have made similar conclusions. In particular, prior studies in post-stroke patients have
14 observed that individuals with a higher range of H (Vistamehr et al., 2016) and a higher time
15 rate of change of H (Nott et al., 2014) during walking had lower clinical balance scores,
16 indicating poorer balance control. In addition, it was found that post-stroke patients
17 categorized as fallers had a larger change of H in the frontal plane during single leg support of
18 walking than their non-faller peers, suggesting that this phase was associated with a high risk
19 of falling (Nott et al., 2014; Vistamehr et al., 2018). Similarly, the results of the current study,
20 in terms of increase in H ranges, could reveal a decreased ability of older adults to control H
21 during the step execution phase (single leg stance) of stepping initiation. Supporting this
22 hypothesis, Buckley et al. (2010) observed that - compared to young adults - older adults were
23 unable to exert the same level of control of vertical-axis angular momentum during the single
24 leg support of stepping down initiation, which was characterized by a greater vertical-axis
25 body rotation and a smaller T_v peak during stepping down. Furthermore, these authors
26 reported that there was no age-related difference in whole-body kinetics during the double
27 support phase. In our study, in contrast with Buckley et al. (2010), we found no difference
28 between age groups in either the transversal plane H or T_v during step execution. This
29 discrepancy could be due to the task demands. In particular, compared to level step initiation,
30 initiation of stepping down may exacerbate whole-body rotation about the vertical axis, which
31 could then pose a greater challenge for H control in the transversal plane.

32
33 While some previous studies found an inverse relationship between H and progression
34 velocity during walking (the faster the gait speed, the lower the range of H) (Bennett et al.,
35 2010; Silverman and Neptune, 2011), other studies have shown that the range of H was
36 unchanged as a function of gait speed (Bruijn et al., 2011). This discrepancy could be mainly
37 due to the method used for normalizing the H values. In their studies, Bennett et al. (2010)
38 and Silverman and Neptune (2011) normalized the H values by taking into account gait speed,
39 while the other studies did not. This method could cause a decrease in H when speed value
40 increases. In the current study, we did not normalize the H by the progression velocity in
41 order to emphasize the effect of the slower pace in the elderly on balance control during
42 stepping. Indeed, we found that older adults initiated stepping with a lower progression
43 velocity and shorter step compared to younger participants. These significant differences
44 disappeared, however, when progression velocity and step length were normalized by
45 participant height (see Table 2). This indicates that differences in progression velocity and
46 step length between the groups could in part be associated to their significant difference in
47 height (older adults were significantly smaller than the younger adults; see Table 1), although
48 it is not excluded that this may also be the result of a conservative strategy implemented by
49 older adults (as attested by the decreased anteroposterior GRF and anteroposterior moment
50 arm). Nevertheless, upon examining whether the fact of normalizing H by progression

1 velocity influenced our results, we observed that the age-related changes in ranges of H in the
2 double support phase disappeared when H was normalized by progression velocity, while
3 those in the step execution phase persisted (see Table S1). This suggests that the age-related
4 alterations in ranges of H during the double support phase could be associated to a difference
5 in progression velocity between both groups i.e., slower in older adults. The fact that this was
6 not the case for age-related changes found during the step execution phase reinforces the
7 hypothesis that these age-related differences may be due to a higher difficulty to control H, as
8 suggested by other authors (Buckley et al., 2010; Pijnappels et al., 2005).

9
10 Previously, Pijnappels et al. (2005) argued that the inability to sufficiently reduce H during
11 the recovery phase of tripping in older adults could be particularly related to muscle force
12 deficit. Muscle force generation has been shown to be a primary mechanism for controlling H
13 during walking (Neptune and McGowan, 2011, 2016). Specifically, these previous studies
14 reported that H in the sagittal plane was mainly controlled by ankle plantar flexors throughout
15 the gait cycle, while frontal plane H was controlled by gluteus medius, soleus and
16 gastrocnemius muscles. Furthermore, a recent study (Alcazar et al., 2018) observed that
17 deficits in both muscle force and muscle contraction velocity in the elderly impaired the
18 force-velocity profiling and both were negatively associated with physical function, quality of
19 life and frailty. These results suggest that muscle contraction velocity could also influence H
20 control during gait. In addition to muscle impairments, the decreased ability to control H in
21 older adults could be explained by other physiological alterations associated with aging, such
22 as cognitive (Montero-Odasso et al., 2012) and sensory impairments (Borel and Alescio-
23 Lautier, 2014), which lead to poorer balance and a higher risk of falling (Horak et al., 2002;
24 Sheridan and Hausdorff, 2007). Finally, the age-related changes in H control in our study
25 could also be due to the adoption of a different stepping strategy. Our results revealing a
26 decrease in both anteroposterior GRF and anteroposterior moment-arm suggest that older
27 adults may adopt a conservative strategy (Halliday et al., 1998; Henriksson and Hirschfeld,
28 2005; Khanmohammadi et al., 2015). As above mentioned, this anteroposterior GRF
29 reduction may particularly contribute to the age-related alterations in the sagittal plane H
30 ranges during both the double support and step execution phases. These findings may
31 potentially help to better understand why - despite the adoption of such conservative
32 strategies in many tasks of daily living (Galna et al., 2009; Menz, 2003) - older adults have a
33 higher rate of falling than their younger counterparts (Talbot et al., 2005). Further
34 investigation is required to confirm this suggestion.

35
36 There are a few limitations in the current study. One limitation is that we used a simplified
37 biomechanical model (including 9 segments) to compute the three-dimensional location of the
38 body's CoM. Nevertheless, this model has yielded satisfactory accuracy during stepping
39 movements in both young and older adults (Tisserand et al., 2016b). Another limitation is that
40 we examined age-related changes in H control in only one speed condition (preferred pace).
41 Investigating a wider range of speeds may provide additional insight into controlling H in the
42 elderly, as older people may choose to initiate step more slowly or more quickly than the
43 speed tested in this study. Furthermore, our older participants were still relatively young (65.3
44 ± 3.2 years old) and were physically active. It would be valuable to examine how H is
45 controlled in much older and sedentary people, who present a higher risk of falling (Talbot et
46 al., 2005). In particular, these people may potentially develop fear of falling, which could
47 have a detrimental effect on their balance performance and increase their risk of falling
48 (Haertner et al., 2018; Young and Mark Williams, 2015). Finally, because we have not
49 assessed the relationship between H ranges and risk of falling, further study is necessary to
50 ascertain this relationship in the elderly. From a clinical point of view, this could be

1 particularly relevant in order to establish normative values based on H values for
2 identifying/assessing older individuals at high risk of falling and also to design H-focused
3 interventions aiming to reduce these falls.
4

5 In conclusion, the results of the current study revealed that, compared to young adults, older
6 people had smaller ranges of H in the sagittal and transversal planes during the double support
7 phase of stepping, and they had higher ranges of H in the frontal and sagittal planes during the
8 step execution phase. These age-related variations underscore that older adults may have
9 difficulties to control H during stepping, which may impose a higher **challenge** for balance
10 control during the step execution phase compared to young adults. Given that falls in elderly
11 people occur frequently during initiation of stepping (Robinovitch et al., 2013), these findings
12 could provide a basis for future studies aiming to reduce the incidence of falls in this
13 population.
14

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17
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23 **Declarations of Interest**

24 None of the authors have financial or other conflicts of interest with regard to this research.
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1 **Figures captions**

2
3 Figure 1(2-column): Graphical representation of ground reaction forces (GRFs) and moment
4 arms which contribute to the external moment about the body's center of mass (CoM). The
5 frontal angular momentum, sagittal angular momentum and transversal angular momentum
6 were defined about the anteroposterior (AP), mediolateral (ML) and vertical axes,
7 respectively. The black arrows represent the GRFs, the grey arrow signifies free vertical
8 moment (T_v) and the dashed lines represent the moment arms. Only the left leg contribution
9 to the external moment (not the right leg) is displayed, which must be considered in the
10 calculation of the net external moment about the CoM when the limb is in contact with the
11 ground.

12
13 Figure 2(1,5-column): Graphical representation of the mean normalized external moment (\dot{H})
14 and the mean normalized whole-body angular momentum (H) for old (full line) and young
15 (dashed line) individuals in the three planes. \dot{H} was normalized by body weight and height
16 and H by body mass, body height and $\sqrt{g \cdot l}$ ($g = 9.81 \text{ m}\cdot\text{s}^{-2}$ and $l = \text{body height (m)}$). FO:
17 instant of the swing foot-off; FC: instant of the swing foot contact.

18
19 Figure 3(1,5-column): The frontal, sagittal and transversal mean ranges of normalized whole-
20 body angular momentum (H) for old (light grey) and young individuals (dark grey) in the two
21 phases of step initiation (double support phase and step execution phase). H was normalized
22 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 age difference with $p < 0.05$, $p < 0.01$ and $p < 0.001$, respectively.

24
25 Figure 4(1,5-column): The frontal, sagittal and transversal mean external moment (\dot{H}) for old
26 (light grey) and young individuals (dark grey) in the two phases of step initiation (double
27 support phase and step execution phase). \dot{H} was normalized by body mass and body height. *,
28 **, ***: Significant age difference with $p < 0.05$, $p < 0.01$ and $p < 0.001$, respectively.

1 **Tables**

2

3 Table 1. Characteristics of the participants.

	Young <i>n</i> = 22 (17 F / 5 M)	Old <i>n</i> = 22 (18 F / 4 M)	<i>P-Value</i>
Age (years)	22 ± 3.2 (18-28)	65.3 ± 3.2 (60-71)	<i>p</i> < 0.001
Weight (kg)	61.2 ± 10 (42-86)	62.5 ± 11 (45-83)	NS
Height (m)	1.66 ± 0.07 (1.55-1.82)	1.59 ± 0.08 (1.46-1.81)	<i>p</i> < 0.01

4 Values are presented as mean ± SD (Range).

5 NS: non-significant difference (*p* > 0.05).

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1 Table 2. Spatiotemporal parameters (non-normalized and normalized)
 2 for the young and old individuals.
 3

	Young n= 22	Old n = 22	<i>P-Value</i>
<i>Non-normalized parameters</i>			
Progression velocity (m.s ⁻¹)	0.79 ± 0.14	0.70 ± 0.13	<i>p<0.05</i>
Step width (m)	0.17 ± 0.02	0.18 ± 0.02	<i>NS</i>
Step length (m)	0.61 ± 0.08	0.54 ± 0.09	<i>p<0.05</i>
Double support phase duration (s)	0.65 ± 0.07	0.68 ± 0.08	<i>NS</i>
Step execution phase duration (s)	0.39 ± 0.06	0.43 ± 0.08	<i>p<0.05</i>
<i>Normalized parameters</i>			
Progression velocity	0.20 ± 0.03	0.18 ± 0.03	<i>NS</i>
Step width	0.10 ± 0.01	0.11 ± 0.01	<i>NS</i>
Step length	0.36 ± 0.04	0.34 ± 0.05	<i>NS</i>
Double support phase duration	1.58 ± 0.17	1.69 ± 0.20	<i>NS</i>
Step execution phase duration	0.94 ± 0.14	1.07 ± 0.19	<i>p<0.05</i>

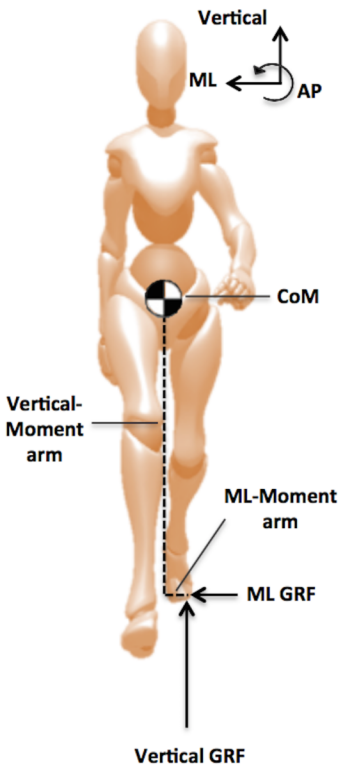
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 5 *NS*: non-significant difference ($p>0.05$).
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1 Table 3. The normalized components of external moment for young and older participants in
 2 the two phases of step initiation.

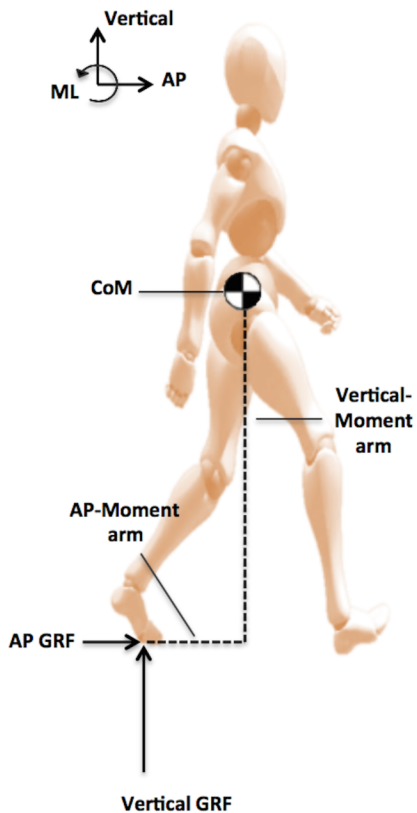
	Young n=22	Old n = 22	<i>P-value</i>
Double support phase			
Peak AP GRF (%w)	0.100 ± 0.021	0.082 ± 0.019	<i>p<0.01</i>
Peak ML GRF (%w)	-0.064 ± 0.011	-0.067 ± 0.014	<i>NS</i>
Peak vertical GRF (%w)	1.085 ± 0.033	1.068 ± 0.031	<i>NS</i>
Peak AP moment arm (%h)	0.049 ± 0.012	0.041 ± 0.011	<i>p<0.05</i>
Peak ML moment arm (%h)	0.037 ± 0.007	0.039 ± 0.007	<i>NS</i>
Peak Vertical moment arm (%h)	0.562 ± 0.009	0.560 ± 0.008	<i>NS</i>
Peak $T_v \times 10^{-3}$ (%w*h)	3.644 ± 0.857	2.610 ± 0.613	<i>p<0.001</i>
Step execution phase			
Peak AP GRF (%w)	0.164 ± 0.050	0.133 ± 0.040	<i>p<0.05</i>
Peak ML GRF (%w)	0.095 ± 0.013	0.092 ± 0.013	<i>NS</i>
Peak vertical GRF (%w)	1.092 ± 0.046	1.057 ± 0.034	<i>p<0.01</i>
Peak AP moment arm (%h)	0.094 ± 0.021	0.084 ± 0.022	<i>NS</i>
Peak ML moment arm (%h)	0.055 ± 0.009	0.057 ± 0.009	<i>NS</i>
Peak Vertical moment arm (%h)	0.547 ± 0.008	0.545 ± 0.009	<i>NS</i>
Peak $T_v \times 10^{-3}$ (%w*h)	-4.486 ± 1.140	-3.989 ± 1.451	<i>NS</i>

3 Peak ground reaction forces (GRFs) were normalized by body weight (w), moment arms were
 4 normalized by body height (h), and free vertical moment (T_v) was normalized by the product
 5 of h and w. ML: mediolateral; AP: anteroposterior. *NS*: non-significant difference ($p>0.05$).
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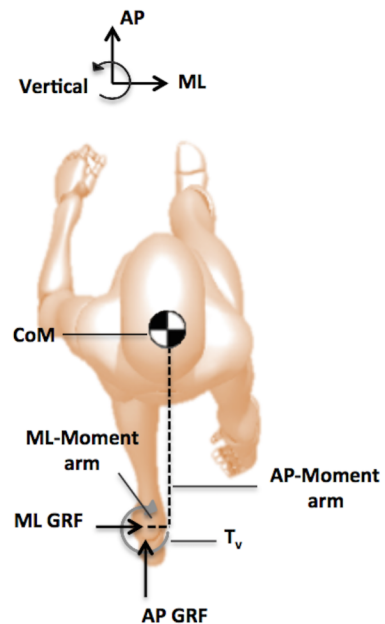
Frontal Plane



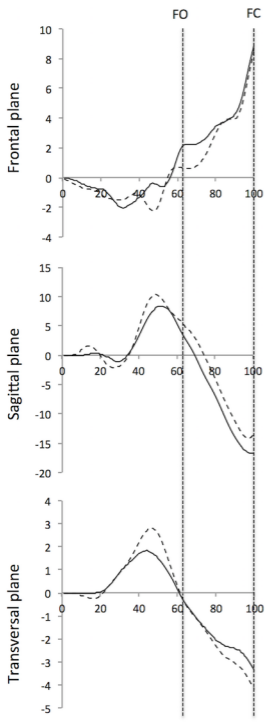
Sagittal Plane



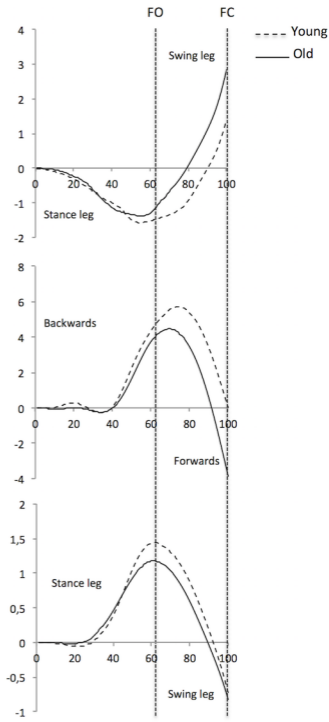
Transversal Plane



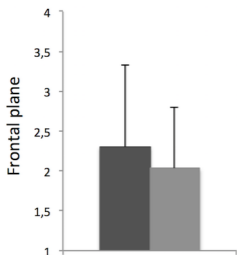
$\dot{H} * 10^{-3}$ (Unitless)



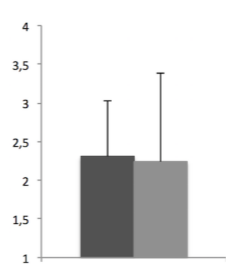
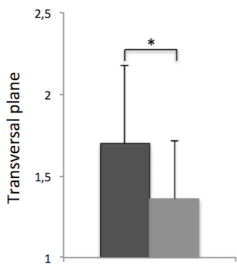
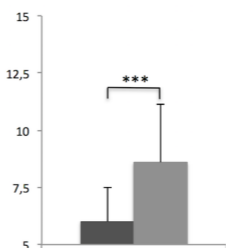
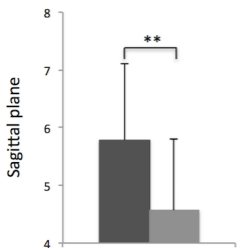
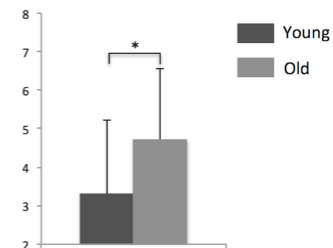
$H * 10^{-3}$ (Unitless)



H range *10⁻³ (Unitless)



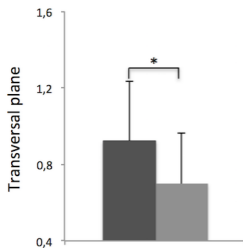
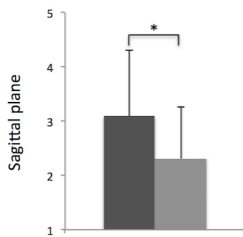
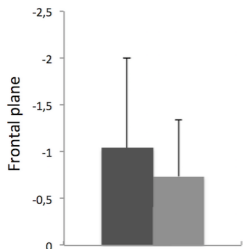
H range *10⁻³ (Unitless)



Double support phase

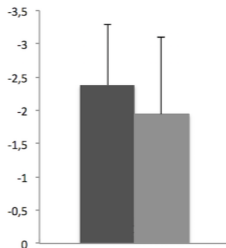
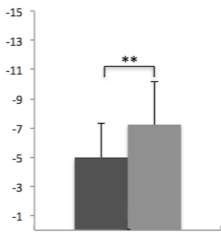
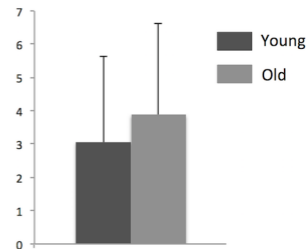
Step-execution phase

\dot{H} mean $\cdot 10^{-3}$ (Unitless)



Double support phase

\dot{H} mean $\cdot 10^{-3}$ (Unitless)



Step-execution phase