Lower safety factor for old adults during walking at preferred velocity

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Abstract Older adults are more prone to falls during walking than young adults, although they walk more slowly and demonstrate higher stability state. This paradox of higher stability state but less safe locomotion let us hypothesize that older people may move closer to their dynamic stability limits. In order to investigate this hypothesis, the present study examined the safety factor of dynamic stability in old and young individuals when walking at their preferred velocity. Twelve older and 12 young male participants walked at their (a) walk-to-run transition velocity (WRV, i.e., maximum capacity) and (b) preferred walking velocity (PWV, i.e., actual applied load). Whole body kinematic data and ground reaction forces were captured. Dynamic stability was assessed using the “margin of stability (MoS)” as a criterion for the stability state of the body (extrapolated center of mass concept). The safety factor was calculated as the ratio between MoS at WRV and MoS at PWV. We found that, although older participants walked slower and provided a higher stability state compared to young ones, they showed a significantly reduced safety factor during preferred walking. This confirmed our hypothesis. Old adults do not walk slowly enough in relation to their

maximum walking velocity, resulting to a lower safety factor during normal locomotion. Apparently, the age-related muscle degeneration affects WRV more than PWV. The resulting lower safety factor for the older participants may partly explain the increased risk of falls in their daily life, in spite of slower locomotion.

Keywords Aging · Safety factor · Falling risk · Margin of stability

Introduction

Falls and their related injuries are among the most common and serious problems in senescence (AGS 2001; Baker and Harvey 1985; Blake et al. 1988). Their importance is highlighted when considering the constantly rising life expectancy. A key concern is not simply the high incidence of falls in older persons, but rather, the combination of high incidence, high susceptibility to injury, and the severity of their consequences (Lord et al. 2001). It becomes clear that it is important to identify mechanisms related to stability deficits, in order to prevent falls in senescence.

Most of the falls in the elderly occur after loss of stability in a forward direction during walking (Blake et al. 1988). It has been reported that the recovery performance after unexpected perturbation is notably decreased in senescence (Mademli et al. 2008a; Wojcik et al. 1999). This observed age-related deficit in ability to regain balance is mainly attributed to reduced muscle strength (Karamanidis et al. 2008; Schultz

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1995), modifications of muscle activation timing (Thelen et al. 1997), and lower contraction velocities (Hortobagyi et al. 1995; Larsson et al. 1979; Thelen et al. 1997).

During daily activities, older adults usually prefer to walk more slowly than young ones (Hamacher et al. 2011; Alexander 1996). This is either due to age related physical constraints such as reduced strength (DeVita and Hortobagyi 2000) and/or because they feel safer this way (Maki 1997). Slower walking speed leads to increased stability, irrespective of age (Kang and Dingwell 2008). Furthermore, it has been reported that when older and young people walk at self-selected velocity (Bohm et al. 2012), older adults walk slower and demonstrate higher stability state. However, although elderly individuals show a higher stability state during walking in everyday life, they exhibit a greater occurrence of falls (Bohm et al. 2012; Bierbaum et al. 2011; Lord et al. 2001), indicating a less safe locomotion compared to young ones. This paradox (i.e., higher stability state but less safe locomotion for the elderly population) led us to hypothesize, that old adults may move during their daily life closer to their limits of stability and thus be more prone to accidents and falls during walking.

In order to examine this hypothesis, the term of safety factor (Beer 2012), which is commonly used in engineering and biology to describe the structural demand of a system/material, is introduced in the context of human gait. Since there is evidence that the key mechanism of the transition velocity from walking to running in humans is related to the muscle force capacity of the plantar flexor muscles (Farris and Sawicki 2012; Neptune and Sasaki 2005), it is reasonable to suppose that the stability state in terms of margin of stability (MoS) (Hof et al. 2005; Hof 2008) at the walk-to-run transition velocity (WRV) would be the individual's maximum capacity (strength) while walking. Furthermore, the stability state (i.e., MoS) at the preferred walking velocity (PWV) can be defined as the actual applied load (designed load) of the human system while walking.

Accordingly, the objective of the present study is to examine the safety factor in old and young adults regarding their dynamic stability when walking at their preferred velocity. We hypothesized that, although old adults may walk slower so as to establish a higher stability state, they would demonstrate lower safety factor compared to young ones during walking at their preferred velocity.

Methods

Experimental design

Twelve older male (68.2 ± 4.2 years, 76.6 ± 7.7 kg; 172.5 ± 7.7 cm) and 12 matched regarding body mass and body height young male (25.2 ± 3.1 years; 77.6 ± 6.7 kg; 178.7 ± 8.7 cm) adults participated in the study, after giving informed consent to the experimental procedure according to the rules of the local scientific board. None of the subjects had any neurological or musculoskeletal impairment. They had to perform walking trials on a walkway in two different velocities: (a) WRV and (b) PWV.

WRV refers to the maximum walking velocity and was defined as the maximal gait velocity with double stance phase and checked through kinetic measurements, i.e., ground reaction forces. Each participant performed four to six walking trials on a walkway about 20-m long, starting at his self-chosen walking speed. Velocity was increased at each trial until the maximum walking velocity was achieved. Velocity was monitored by light barriers enclosing the capture volume and via kinematic data recorded by an infrared motion capture system (VICON Nexus, Vicon Motion Systems, Oxford, UK).

The subjects were instructed to walk at self-selected pace and at their maximum walking velocity on the walkway. They were asked to keep the required velocity all through the 20-m walkway. The gait protocol consisted of three valid walking trials at PWV followed by three valid trials at the WRV. Twelve infrared cameras operating at 250 Hz captured the kinematic data of the 21 reflective markers (14-mm diameter), fixed on whole body anatomical landmarks as described by Bierbaum et al. (2011). Briefly, the markers were positioned as follows: four markers on a headband; one marker at C7; and for both body sides, one marker at the acromion, tuberositas radii, processus styloideus ulna, trochanter major femoris, knee joint space lateral, malleolus lateralis, and tuber calcanei as well as the second caput ossis metatarsalis. The marker trajectories were smoothed using a Woltring filter routine with a noise level of 10 mm^2 . Segmental masses and the location of the segment center of mass were calculated based on the data reported by Zatsiorsky and Seluyanov (1983). Ground reaction forces of two successive steps were recorded (1,250 Hz) by means of two Kistler force plates ($60 \times 90 \text{ cm}^2$, Kistler, Winterthur, Switzerland).

Quantification of the dynamic stability

We used the “extrapolated center of mass” concept formulated by Hof et al. (2005) to quantify the stability state of the human body during walking. As a criterion for the stability state, the MoS was calculated as follows:

$$b_x = U_{x \max} - X_{cm}$$

where b_x is the MoS in the anteroposterior direction (MoS), U_{XCM} is the anterior boundary of the base of support (BoS), and X_{CM} is the position of the extrapolated center of mass (CM) in the anteroposterior direction, and it is calculated as follows:

$$X_{CM} = P_{XCM} + \frac{V_{XCM}}{\sqrt{g/l}}$$

where P_{XCM} is the horizontal (anteroposterior) component of the projection of the CM to the ground, V_{XCM} is the horizontal (anteroposterior) CM velocity, and the term $\sqrt{g/l}$ presents the eigenfrequency of a system of length l (inverted pendulum model, l is the distance from the ankle joint to the CM, Fig. 1).

A stable position of the human body is represented by positive values of the MoS, whereas an unstable position, i.e., when the human system has to perform additional actions to maintain balance, is indicated by negative values.

We investigated the average values of the dynamic stability parameters for each walking trial during a step cycle measured at the middle distance of the walkway (from foot strike to contralateral foot strike). Furthermore, we examined walking velocity, step length, and cadence during the walking trials. For the statistical analysis, we used the mean values for the three valid trials at WRV and PWV. We supposed that the MoS at WRV would be the individual’s maximum capacity (strength) and the MoS at PWV the actual applied load (designed load) of the examined participants while walking. Therefore, the safety factor during walking was calculated as follows:

$$\text{safety factor} = \frac{\text{MoS at WRV}}{\text{MoS at PWV}}.$$

Statistics

A two-way analysis of variance (age $\times$ walking velocity) was used to examine the age (young, old) and walking

Fig. 1 Schematic diagram of the inverted pendulum model while walking. The human body is represented by a single mass m at the center of mass (CM) by a pendulum length (l). P_{XCM} is the projected CM on the ground, g the acceleration of gravity, and v the CM velocity. U_{MIN} and U_{MAX} indicate the boundaries of the base of support (BoS). CM trajectory is extrapolated in the direction of its anterior-posterior velocity denoted as X_{CM} . The parameter “margin of stability (b_x)” as the anterior-posterior distance of the X_{CM} to the anterior boundary of the BoS can be used as a criterion for the stability state of the human body (Hof et al. 2005)

velocity (preferred, maximum) related differences in the analyzed parameters. The “safety factor” and anthropometric data were compared with a t test for non-dependent samples. The level of significance for all statistical comparisons was set to $\alpha=0.05$.

Results

Old participants demonstrated greater MoS (i.e., greater stability state) ($p<0.001$, Table 1), but a significant lower ($\sim 11\%$) safety factor than the young ones, when they walked at their preferred velocity ($p=0.044$, Fig. 2). The main reason for the greater MoS in the old adults was the lower PWV compared to those in the young ones ($p<0.001$, table 1). The old individuals walked at significantly lower horizontal velocity of the CM and did shorter steps ($p<0.001$ and $p=0.011$, Table 1). However, there was a significant ($p<0.001$)

Table 1 Means $\pm$ SD of the spatial-temporal gait parameters and the components of dynamic stability during walking at walk-to-run and preferred velocity for the young and old participants

	Young ($n=12$)		Old ($n=12$)	
	WRV	PWV	WRV	PWV
V_{XCM} [m/s] ^{a,b,c}	3.58 $\pm$ 0.31	1.59 $\pm$ 0.11	3.07 $\pm$ 0.19	1.48 $\pm$ 0.11
Step Length [cm] ^{a,b}	100.6 $\pm$ 10.0	80.8 $\pm$ 4.7	94.2 $\pm$ 7.5	76.3 $\pm$ 5.4
Cadence [Hz] ^b	1.78 $\pm$ 0.19	0.99 $\pm$ 0.04	1.66 $\pm$ 0.18	0.97 $\pm$ 0.06
Base of support [cm] ^{a,b}	118.1 $\pm$ 5.1	107.6 $\pm$ 6.3	115.1 $\pm$ 7.7	103.4 $\pm$ 4.9
X_{CM} [cm] ^{a,b,c}	217.0 $\pm$ 12.5	135.3 $\pm$ 8.5	192.9 $\pm$ 10.8	127.9 $\pm$ 6.9
MoS [cm] ^{a,b,c}	-98.9 $\pm$ 10.7	-27.7 $\pm$ 3.1	-77.8 $\pm$ 7.4	-24.6 $\pm$ 3.2

WRV walk-to-run transition velocity, PWV preferred walking velocity, V_{XCM} Horizontal velocity of the center of mass, X_{CM} extrapolated center of mass, MoS margin of stability

^a Statistically significant age effect ($p<0.05$)

^b Statistically significant velocity effect ($p<0.05$)

^c Statistically significant interaction between age and velocity ($p<0.05$)

interaction between age and velocity on the MoS (Table 1), indicating lower differences in this parameter between PWV and WRV for the old participants compared to the young ones, accounting for the smaller safety factor found for the older adults. This interaction resulted from the fact that the differences in the values of horizontal velocity of the CM as well as of the position of the extrapolated CM between WRV and PWV were smaller for the older individuals. Similar to MoS, a significant interaction between age and velocity on horizontal velocity of the CM ($p=0.001$) as well as on the position of the extrapolated CM ($p=0.006$) has been found (Table 1).

As it was expected, both old and young adults demonstrated greater ($p<0.001$) step length as well as cadence when walking at WRV compared to PWV (Table 1). Accordingly, the values of the base of support and the position of the extrapolated CM ($p<0.001$, Table 1) were greater at WRV for both groups. There was an interaction between age and velocity on the extrapolated CM demonstrating lower differences between WRV and PWV for the old adults (Table 1).

Discussion

The present study investigated the safety factor during preferred walking in old and young adults regarding their dynamic stability. We found that although old participants walked at lower velocity and provided a higher stability state, they showed a significantly lower safety factor compared to young ones during preferred walking. Therefore, our hypothesis has been confirmed. The findings demonstrate that old adults moved closer to their maximum stability capacity. Although it cannot be discarded that participants may have walked differently in the motion lab than at home or on the street, the protocol of our study minimized this possibility, by involving a floor walkway, along with a series of trials ensuring familiarization of subjects with the experimental environment. Therefore, it is reasonable to expect that old people would move closer to their maximum stability capacity also in everyday life, being therefore

Fig. 2 Mean values and standard deviation of the safety factor during walking at the preferred walking velocity (PWV) in young ($n=12$) and old ($n=12$) adults. *Statistically significant differences between young and old adults ($p<0.05$)

more prone to accidents and falls during walking. Indeed, several epidemiological studies report high frequency of falls in the everyday life of the elderly, despite a slower walking velocity (Baker and Harvey 1985; Lord et al. 2001; Tinetti et al. 1995). Therefore, we can argue that the lower safety factor may increase the risk of falls within the elderly population and partly explain the high frequency of falls in old adults.

In accordance with previous studies, we found that the PWV of older adults is significantly slower compared to young ones (Hamacher et al. 2011; Alexander 1996). The aging process is associated with a general degeneration of the musculoskeletal system and a decline in its capacities such as loss in muscle strength (Frontera et al. 2000; D'Antona et al. 2003; Morse et al. 2005) and stiffness of collagenous tissues like tendons (Karamanidis and Arampatzis 2005; Kubo et al. 2003; Mademli et al. 2008b). During walking, the plantar flexor muscles are the primary contributors for forward propulsion in late stance (Anderson and Pandy 2003; Neptune and McGowan 2011). Hence, the age-related reduction in the capacities of the plantar flexor muscles (Karamanidis and Arampatzis 2006; Mademli and Arampatzis 2008), which leads to lower joint moment and mechanical power during walking, contributes to the slower WRV found for the older adults. Furthermore, it has been reported that shortening velocity of the gastrocnemius medialis muscle may be a key factor that limits the speed humans choose to walk at and explains the transition from walking to running (Farris and Sawicki 2012). The maximum muscle shortening velocity depends on muscle fiber length (Hill and White 1968; Narici 1999) and fiber type (Scott et al. 2001), both negatively affected by aging (Lexell et al. 1988; Mademli and Arampatzis 2008). Consequently, the known age-related deterioration in muscle strength and muscle morphology can be associated with the reduced WRV in senescence.

It has been reported that walking slower may be a proactive strategy to improve walking stability (Shkuratova et al. 2004). However, the lower safety factor indicates a less stable locomotion in the old adults compared to young ones, despite the slower PWV. The lower safety factor of the older people may be explained by the fact that WRV and PWV are controlled and regulated by criteria, which can be differently affected by aging. While the key factor for the transition from walking to running seems to be the fascicle shortening velocity of the gastrocnemius muscle (Farris and

Sawicki 2012), the metabolic energy cost per distance is the main factor which regulates the PWV by humans (Srinivasan 2009; Martin et al. 1992; Browning et al. 2006). It is generally accepted that individuals regardless of age, body weight, or physical activity status tend to select a preferred gait speed at or near the speed at which metabolic energy expenditure is minimized, i.e., most energetically efficient speed (Ralston 1958; Martin et al. 1992; Browning et al. 2006). Such speed values range according to the literature from 1.2 to 1.5 m/s (Martin et al. 1992; Kang and Dingwell 2008; Ralston 1958). The velocities measured as PWV in the present study coincide with this range (Table 1). Moreover, we found that the differences in the horizontal velocity of CM between the two groups were smaller at the PWV than at the WRV, i.e., there was a significant interaction between age and velocity (Table 1). This implies that the age related deterioration in the capacities of the musculoskeletal system may affect the mechanisms responsible for determining the preferred and maximum walking velocity differently, resulting to a reduced safety factor for the older adults.

Concluding, we found that although old participants walked with a lower preferred velocity and thus with a greater stability state compared to the young ones, they showed a lower safety factor, since they walked closer to their stability limits. The reduced safety factor for the old individuals may increase the risk of falls in their everyday life and thus partly explain the greater frequency of falls in this population, despite their slower locomotion. Therefore, we suggest that the development of interventions aiming to improve the stability limits of the older people during walking, resulting this way to an increase of the safety factor, would contribute toward reducing the risk of falling.

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