The effect of aging on termination of voluntary movement while standing: A study on community-dwelling older adults

Satoshi Kasahara[a],[*], Hiroshi Saito[b]

[a] *Department of Rehabilitation Sciences, Faculty of Health Sciences, Hokkaido University, Sapporo, Japan*
[b] *Department of Physical Therapy, School of Rehabilitation, Tokyo Kasei University, Sayama, Japan*

ABSTRACT

For older adults, falls often occur when transitioning from motion to a complete stop, as the motor control required during this phase is very complex and challenging. The purpose of this study was to clarify the effect of aging on the motor control required to terminate motion. Twenty-five healthy older adults (aged > 65 years) and 25 healthy young adults (20–23 years) performed a rapid stopping task while standing on a force plate. The rapid stopping task was conducted by analyzing center of pressure (COP) on the force plate during a visually guided tracking experiment. To assess the ability to terminate motion, we measured the velocity waveform for the COP, along with the reaction, propulsion, braking, and total movement times. Both the reaction and movement times of the older-adult group were significantly longer than those of the younger-adult group (all, p < 0.05). There was no significant difference between the groups in regard to the initial backward propulsion time; however, in the subsequent sequence of backward braking, forward propulsion, and backward braking, all times were longer in the older-adult group than in the younger-adult group (p < 0.05). Our results show that the series of time delays shown by older adults when initiating and terminating motion is due to not only delayed reactions but also delayed stopping. Furthermore, our findings suggest that older adults have not only a diminished propulsion ability but also a diminished braking ability.

1. Introduction

The upright standing posture of humans is inherently unstable (Pai, 2003), and one in three community-dwelling older adults is likely to experience falls (Kannus, Sievänen, Palvanen, Järvinen, & Parkkari, 2005). Furthermore, the number of annual fall-related injuries in the United States is expected to be 5.7 million by 2030 (Hsieh, Roach, Wajda, & Sosnoff, 2019). Falls from a higher position (i.e., a standing posture) may cause a fracture of the neck or the femur (Tsuboi, Hasegawa, Suzuki, Wingstrand, & Thorngren, 2007). Another problem is that physical injuries can decrease the ability to engage in activities of daily living and can also reduce the quality of life in older adults (Hsieh et al., 2019).

The activities of daily living that we perform in a standing posture involve repeated, short, intermittent movements over relatively brief periods. These activities require frequent accelerations and decelerations, and considerably more complex motor control is required to initiate and terminate these sequences than to execute motions at a constant velocity (Cimolin et al., 2017; Peterson, Kautz, & Neptune, 2011). Further, during these transition phases, more complex motor skills are also required to maintain postural stability than during steady-state conditions (Cimolin et al., 2017). Given this high level of motor control required during transition

---

phases, it is logical to speculate that for older adults and individuals with motor disorders, the risk of falling increases during the frequent changes in posture that occur when moving while standing (Cimolin et al., 2017; Peterson et al., 2011).

Generally, instability in posture arises during transitions from a static to a dynamic state (and vice versa), which are often observed during the initiation (Breniere & Do, 1986) or termination (Cimolin et al., 2017) of a movement. The kinematic profiles of such initiation and termination (e.g., the center of pressure (COP) during a gait) movements virtually mirror each other (Jian, Winter, Ishac, & Gilchrist, 1993). However, different mechanisms underlie action initiation and termination (Morein-Zamir, Nagelkerke, Chua, Franks, & Kingstone, 2006), and these are presumed to be associated with the feed-forward system (Crenna & Frigo, 1991) and the feed-back system (Jaeger & Vanitchatchavan, 1992), respectively. However, compared to initiation movements, there is limited information regarding termination movements.

Most of the examinations of the termination phase of motion have investigated gait (Cao, Ashton-Miller, Schultz, & Alexander, 1998; Jung, Yi, & Song, 2016; Menant, Steele, Menz, Munro, & Lord, 2009; Tirosh & Sparrow, 2004). However, gait termination is influenced by last-minute gait parameters, because it is a process that culminates in the static standing posture from a steady gait (Bishop, Brunt, Pathare, & Patel, 2004; Chen, Ashton-Miller, Alexander, & Schultz, 1994; Hase & Stein, 1998; Sparrow & Tirosh, 2005). Although gait is undoubtedly an exceedingly complex movement, gait termination requires the fine control of both motor and postural control simultaneously (Sparrow & Tirosh, 2005). In the case of older adults, although physical functions decline with age, the true ability of the termination motion may be masked by decreased gait abilities. Furthermore, the ability to safely perform termination has a strong impact on patients with neurological diseases (Bishop, Brunt, & Marjama-Lyons, 2006; Kirker, Simpson, Jenner, & Wing, 2000; Morris & Iansek, 1997; Roeing, Wajda, Motl, & Sosnoff, 2015; Serrao et al., 2013), as it is expected to significantly affect fall risk (Meier, Desrosiers, Bourassa, & Blaszczyk, 2001). Therefore, it is necessary to address termination in a standing posture as the first step toward understanding it during gait.

The main aim of this study was to quantitatively examine older adults' ability to cease motion. Specifically, to clarify the occurrences during the transition from movement to a complete stop, the rapid stopping that is performed in a standing posture was divided into phases based on the fundamental components of motion. We hypothesized that older adults would demonstrate poorer motion-termination abilities than younger adults. Additionally, from a motor-control perspective (Bishop et al., 2006; Crenna & Frigo, 1991; Jaeger & Vanitchatchavan, 1992; Morein-Zamir et al., 2006), we hypothesized that compared to younger adults, deceleration would be potentially more challenging for older adults than acceleration.

2. Methods

2.1. Participants

Twenty-five healthy young adults (YG: 25 men, aged 20–23 years) and 25 healthy older adults (OG: 25 men, aged > 65 years) participated in this study. The young participants were randomly selected from a pool of college students who volunteered to participate in this study; the older subjects were randomly selected from among community-dwelling older adults who registered with the recruitment agency used. All participants were physically active, lived independently in their communities, and had no disorders or injuries or any neurological, vestibular, orthopedic, or cognitive conditions that could interfere with their balance. Furthermore, since the motor task used in this study was initiated by a visual cue, participants were excluded if their visual acuity was below 1.0, which was determined using the Landolt ring chart (Kasahara & Saito, 2015). Also, all older adults reported having not experienced any falls for at least six months before enrollment in this study. The demographic characteristics of both groups are summarized in Table 1. The table shows similar traits throughout, aside from age. All participants provided written informed consent to participate, and the procedures were approved by the ethics committee of Hokkaido University School of Medicine (no. 11–03).

2.2. Experimental procedure

Participants stood, in bare feet, on a force plate (Kistler Type 9286A, Switzerland), with their arms crossed in front of their chests (Fig. 1A); their feet were approximately shoulder-width apart. The position of the participants' feet was kept constant through the entire session by tracing the location of each foot on a piece of white paper placed on top of the force plate (Kasahara & Saito, 2015). Before beginning the trials, participants were asked to fix their gaze on a computer screen in front of them (approximately 1 m away, at eye level).

The motor task was the forward movement of the COP during the standing posture on the force plate. Participants started the

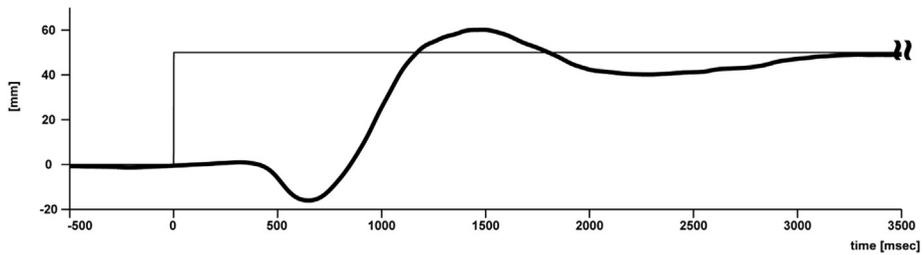**Table 1**
Participant characteristics.

	YG (n = 25)	EG (n = 25)
Age (y)	20.8 ± 0.7	70.4 ± 3.2 *
Height (cm)	172.2 ± 4.8	165.0 ± 5.1
Weight (kg)	62.4 ± 6.4	61.9 ± 8.7
BMI (kg/m^2)	21.0 ± 1.6	22.7 ± 2.7

Values are mean ± standard deviation. YG: younger group, OG: older group, BMI: body mass index.

A. The experimental environment
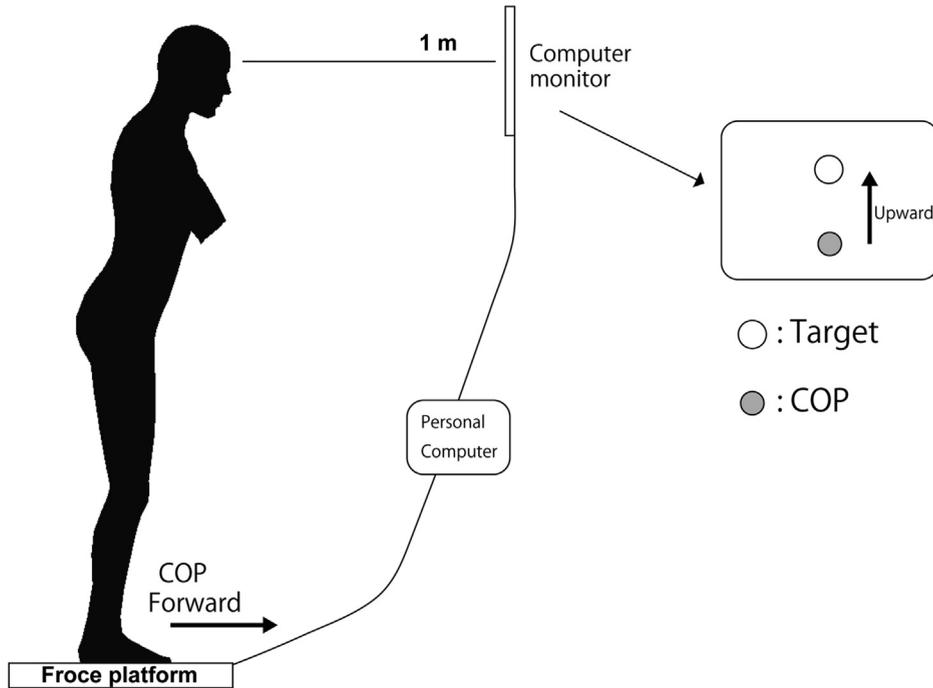
B. COP Position
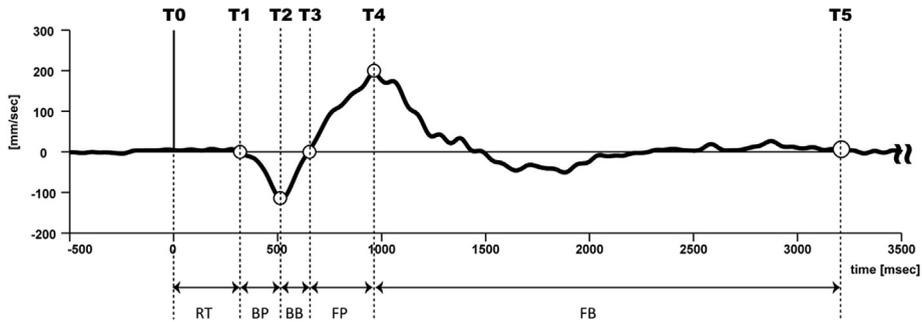

C. COP Velocity

Fig. 1. Experimental setup (A). The representative trajectory COP position (B) and velocity (C) in the horizontal plane for one young participant. The thin lines indicate target position and velocity, respectively, and the thick lines indicate COP position and velocity, respectively. The positive value indicates anterior; the negative value indicates posterior. T0 is the target onset, T1 is the COP onset, T2 is the COP maximum backward velocity, T3 is the turning point of the COP velocity, T4 is the COP maximum forward velocity, and T5 is the COP stop point. The RT is the reaction time, the BP is the backward propulsion time, the BB is the backward braking time, the FP is the forward propulsion time, and the FB is the forward braking time.

forward movement of the COP as fast as possible when they were given the visual cue (i.e., the shift of the target in the monitor) and then stopped rapidly to match the COP position with the target position on the monitor as accurately as possible (see Fig. 1A). The COP position on the force plate was calculated using a customized program (Labview 2009; National Instruments, Austin, TX, USA), and was then displayed in real time on the monitor as a virtual COP, positioned alongside a controlled target stimulus (Kasahara, Saito, Anjiki, & Osanai, 2015). After the examiner ensured that the COP was steady and within 1 cm of the start position, the task began with the target stimulus moving 5 cm upwards from the center of the monitor (see "Discussion"). Participants were asked to imitate the movements of the stimulus with the virtual COP as quickly as possible; the target stimulus moved at pseudorandom intervals between 10 and 30 s to eliminate the chance of participants predicting the moment each trial would start. Moving forward on the force plate caused the virtual COP to move upwards on the monitor; thus, the participants needed to move their COP 5 cm forward to match the target's movement. Although we instructed participants not to raise their toes or heels (if possible), we did not give any instructions regarding the movement of each leg joint; this was in order to avoid any bias related to postural strategies. Participants performed 10 trials of this task, and a 1–2-min rest was given between trials in order to minimize fatigue.

2.3. Data collection and analysis

Fig. 1B and C show the representative COP position and velocity of one young participant. All signals from the force plate were sampled at 1 kHz using a NI Compact DAQ (National Instruments, Austin, TX, USA) and digitally low-pass filtered using a zero-lag, second-order Butterworth filter at 10 Hz (LabView, National Instruments, Austin, TX, USA) (Kasahara et al., 2015). The grand reaction force and the position of the COP were calculated in real time using these signals. The COP positions in the anterior-posterior (AP) direction (COPx) and the medio-lateral (ML) direction (COPy) were calculated using following formulas (Saito, Yamanaka, Kasahara, & Fukushima, 2014).

$$COPx = -(My + Fx * d)/Fz$$

$$COPy = (Mx + Fy * d)/Fz,$$

where Fz is the vertical ground force, Fx and Fy are shear forces, Mx and My are moments around the AP and ML axes, respectively, and d is the distance from the origin of the force platform to the surface. As velocity is the most reliable parameter for determining postural and motor control (Jeka, Oie, Schöner, Dijkstra, & Henson, 1998), the averaged COP velocity of each participant was used in the following analysis.

The trajectory of the COP in the forward shift task was backward, followed by forward (Klous, Mikulic, & Latash, 2012; Cau et al., 2014) (see Fig. 1B). The onset of the COP was taken as the first point where the averaged COP velocity dropped two standard deviations (SDs) from the mean, which was calculated as 1 s prior to the target onset (i.e., the baseline period) and continued for 200 ms (see Fig. 1C). Reaction time (RT) was defined as the time between the target onset to the onset of the COP. The end of the COP was determined to be the moment the averaged COP velocity fell within ± 2 SDs of the mean of the baseline and remained there for 1 s (Perry, Santos, & Patla, 2001). The total movement time was defined as the time from the onset to the end of the COP.

Initially, the COP trajectory was divided into the backward shift phase and the forward shift phase (see Fig. 1C). Then, to examine motor control ability in detail, each shift phase was divided into two sub-phases: the propulsive phase and the braking phase. The first propulsive phase was defined as the period from the trial onset to the peak of the backward COP velocity, and the first braking phase was defined as the period from the backward peak to the inflection point of the COP velocity. Meanwhile, the second propulsive phase was defined as the period from the inflection point to the peak of the forward COP velocity, and the second braking phase was defined as the period from the forward peak to the termination of the COP velocity. Additionally, because this study focused on pure braking ability, braking time (BT) was calculated (by subtracting RT from forward BT). These points and phases were measured using a customized program (Kasahara et al., 2015) and, finally, were re-confirmed through visual inspection (Klous et al., 2012).

2.4. Statistical analysis

All statistical analyses were performed using SPSS Statistics 18.0 (IBM Inc., Armonk, NY). The demographic parameters of the subjects were presented as mean ± SD and data for the COP were presented as mean ± standard error (SE). Independent-samples t-tests were used for comparisons between the YG and OG regarding demographic and COP parameters. If data were not normally distributed, a Mann-Whitney U test was applied. All significance levels were set at $p < 0.05$.

3. Results

Both groups showed a similar COP trajectory, with the COP moving backward and then forward (see Fig. 2, top). Although the pattern of velocity was similar in both groups, the end point of termination was later in the older participants compared to that in the young participants (Fig. 2, bottom). There was no significant difference between the groups in terms of the final forward position of the COP at the stopping point (YG: 5.7 ± 0.2 cm; OG: 5.8 ± 0.2 cm). There was also no significant difference between the groups in regard to the maximum backward velocity of the COP (YG: 15.7 ± 1.5 cm/s; OG: 18.7 ± 1.6 cm/s, p = 0.17). On the other hand, there was a significant difference between the groups regarding the maximum forward velocity of the COP, with the OG showing a higher maximum forward velocity than the YG (YG: 20.1 ± 1.8 cm/s; OG: 25.8 ± 1.9 cm/s, p < 0.05).

Table 2 shows the groups' times for each phase relating to the stopping task. As expected, the OG showed a longer RT than did the
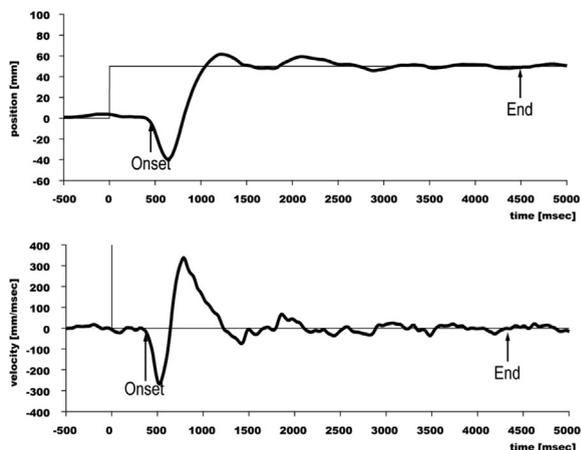
A. Younger
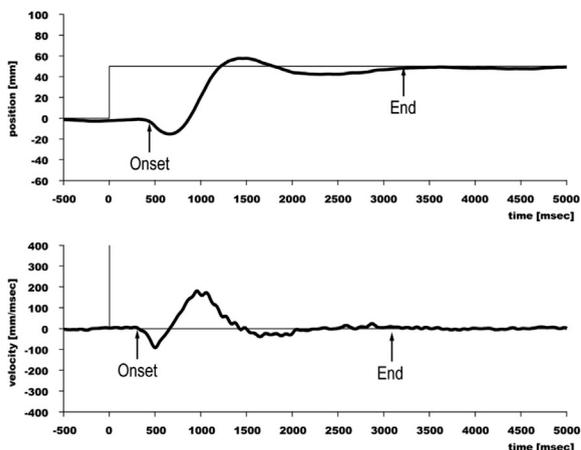
B. Older

Fig. 2. Representative example of time series of position (top) and velocity (bottom) of center of pressure (COP) in one younger participant (A, left panels) and one older participant (B, right panels). The COP onset and end point were shown in each group.

Table 2
Comparison between the younger group (YG) and older group (OG) for each time and phase.

	YG	EG
Reaction time	340.4 ± 5.3	366.7 ± 9.1**
Backward propulsion phase	154.5 ± 9.3	134.3 ± 7.5
Backward braking phase	119.1 ± 7.2	205.0 ± 39.7*
Forward propulsion phase	298.6 ± 32.8	147.6 ± 46.3**
Forward braking phase	2148.2 ± 144.1	4877.5 ± 203.5**
Total movement time	2720.5 ± 144.6	5364.4 ± 205.6**

Values are means ± standard deviations. Units are msec. *: $p < 0.05$, **: $p < 0.01$.

YG ($p < 0.05$, Fig. 2). Further, for the OG, the first backward propulsive phase was shorter than that for the YG; however, the difference between the groups in this regard was not significant ($p = 0.17$). There were, however, significant differences between the groups in regard to other phases. For the OG, the first backward braking phase was significantly longer than that for the YG ($p < 0.05$). For each forward phase, the OG showed a shorter propulsive phase and a longer braking phase than did the YG ($p < 0.01$ for both). Also, the OG showed a longer BT than the YG ($p < 0.01$, Fig. 3). Finally, there was a significant difference between the groups in regard to the total movement time: the OG took longer to stop completely than did the YG ($p < 0.01$, Fig. 2).
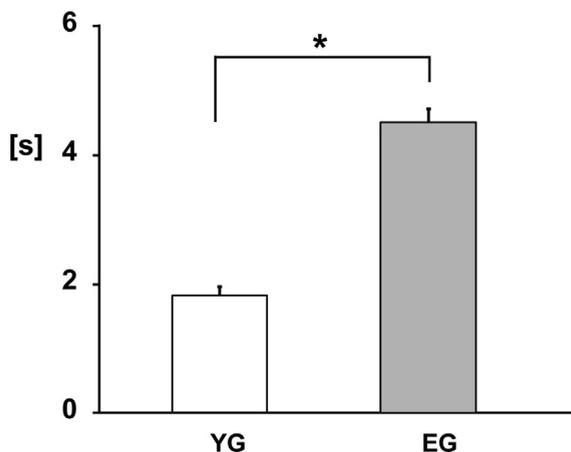

Fig. 3. Comparison of the braking time between the younger (YG) and older (OG) groups. The error bars represent the standard errors. *$p < 0.01$ between groups.

4. Discussion

Older adults often experience falls during both the initiation and termination of movements (Cao et al., 1998; Jung et al., 2016; Menant et al., 2009; Sparrow & Tirosh, 2005; Tirosh & Sparrow, 2004; Tucker, Kavanagh, Morrison, & Barrett, 2010a). In the current study, we focused on the termination of movements, aiming to quantitatively examine the effect of aging on the ability to cease motion. Our results showed that termination of motion in a standing posture was significantly delayed in the OG compared to the YG, which suggests that the ability to rapidly cease motion decreases with age. Furthermore, our findings suggest that this extended time required by older individuals is essentially caused by not only a delay in the reaction to a stopping cue but also a delay in the on-line processing of motor control through the central nervous system (CNS).

Regarding the transition from a dynamic to a quasi-static condition, most previous studies in this regard have used a walking task (Bishop et al., 2004; Cao et al., 1998; Chen et al., 1994; Hase & Stein, 1998; Jung et al., 2016; Menant et al., 2009; Sparrow & Tirosh, 2005; Tirosh & Sparrow, 2004). However, several previous studies demonstrated that gait parameters (e.g., velocity, step length, cadence) influenced the performance of the gait termination (Cao et al., 1998; Tirosh & Sparrow, 2004). The stopping task during a standing task may allow for exploration of motor control in older adults. In this study, we asked participants to perform a rapid stopping task while in a standing posture to reduce the influence of gait parameters. Consequently, there was no significant difference in the first key velocity of COP (i.e., the maximum backward velocity of the COP) between the younger group and the older group. Additionally, although differences in COP displacement might affect various time parameters (Tucker, Kavanagh, Morrison, & Barrett, 2010b), there was no significant difference between the groups regarding forward COP distance; this was because the displacement of our target had been set to a constant distance (see the Methods section for more information). However, the second COP velocities (i.e., the maximum forward velocity of the COP) revealed a significant difference between groups. The second forward velocity in older adults seems to increase in order to recover delayed responses to the stop cue and to shorten extended times in the backward braking phase.

Our results showed that older adults require more time than younger adults to perform sudden cessation of motion, and this accords with the results of previous related studies (Cao et al., 1998; Menant et al., 2009; Tirosh & Sparrow, 2004). Rapid stopping is a complex and time-critical task that involves swift visual processing, strategy planning, and motor execution (Cao et al., 1998). Normal physiological aging results in a slowing of neural processing and of the transmission of information through the CNS (Meier et al., 2001; Spirduso, 1975; Grouios, 1991); furthermore, skeletal muscle volume, peripheral nerve conduction velocities (Dorfman & Bosley, 1979), and peripheral sensation (Menant et al., 2009) also decrease with advancing age. Therefore, the observed delayed reaction time in the older participants was expected, and could have been caused by both the central and the peripheral functional decline with aging.

This delayed reaction time affects subsequent motions (Meier et al., 2001). In this study, participants were instructed to rapidly move their COP in accordance with a corresponding cue on a computer monitor, and then to stop at the target position; then, following a change in the target position, they needed to move their COP backward once (i.e., the reversal phenomenon, see Figs. 1B and 2 (Cau et al., 2014; Klous et al., 2012)). In this backward shift phase, despite the fact that there was no significant difference between the groups in backward COP velocity, the OG required more time to cease the backward motion than did the YG. This may suggest the difference in braking abilities between the YG and the OG. In the following shift phase (i.e., the forward shift phase), the older participants increased the forward COP velocity more than the younger participants. It is possible that the older adults increased their forward velocity in order to compensate for the preceding delay, caused by their longer RT and first backward BT. However, in the OG, such increased velocities appeared to produce increased stopping time and did not contribute to a recovery of the delay. Conversely, the higher maximum forward velocity compared to that of the YG may impose greater motor-control demands on the OG. As a result of decreasing absolute muscle strength due to aging, older adults are routinely forced to exert high levels of control over muscle power (Hortobágyi, Mizelle, Beam, & DeVita, 2003). Although more muscle power is needed to increase velocity, it has been determined that, for older adults, such increased muscle power results in increased errors in motor output (Christou & Carlton, 2001). Furthermore, the greater kinematic momentum produced by the increased velocity (which requires increased acceleration) consequently increases the amount of internal force requiring control. Therefore, our findings suggest that older adults have less ability to control their movement velocity, and during rapid movements they require more processing time to correct errors in motor outputs and to control internal forces.

A major finding of this study was that older adults have a longer braking phase than do younger adults. In particular, even when forward BT was calculated (by measuring the RT to the stopping of the target stimulus; Cao et al., 1998), the termination was later in the OG. Since propulsion force continues in the braking phase, both a reduction in propulsion force and an increase in braking force are required for full, stable termination (Tirosh & Sparrow, 2004). The group members' modulation of these two forces is expected to mirror the age-related decline in the force and speed of muscular contraction (Tirosh & Sparrow, 2004). In fact, there were significant differences in the braking times with respect to termination between groups (see Table 2). The biomechanical requirements for propulsion involve producing a forward movement and maintaining body balance, and the biomechanical requirements for braking include eliminating forward movement, producing a backward movement, and maintaining body balance. Compared to propulsion, motor control related to braking is more complex as a result of the interaction between the neural (Jaeger & Vanitchatchavan, 1992) and biomechanical factors (Cao et al., 1998). Consequently, rapid stopping of movement is more difficult for older adults.

Our previous study used the visually guided cyclic tracking task with the set of the sine wave (i.e., smooth pursuit movement; Kasahara & Saito, 2015). In our previous study, participants seemed to rely on the visual feedback at a first few cycles under the condition of the slow sine velocity. After that, participants had opportunities to perform with the feed-forward during the latter cycle. On the other hand, the current study used the step-like motion induced by the visual cue (i.e., a visually guided, step-tracking task;

Cooke, Brown, & Cunningham, 1989) and this task was performed discretely. Given the discrete movement, the initiation relies on the feed-forward system and the termination relies on the feed-back system (Crenna & Frigo, 1991; Morein-Zamir et al., 2006). The visual feedback system during the termination was required to match the COP position to the moved target position because visual information is one of the major inputs in termination control (Jian et al., 1993). Therefore, the involved visuo-spatial transformation may incur a higher loading of information processing during the deceleration phase (Temprado et al., 2013). However, since older adults require more time to process the available on-line information through visual feedback (Chaput & Proteau, 1996), their accurate termination would take more time compared with young adults.

Difficulty terminating movement appears not only in older adults but also in patients with Parkinson's disease (Bishop et al., 2006; Morris & Iansek, 1997), cerebellar ataxias (Serrao et al., 2013), multiple sclerosis (Roeing et al., 2015), and diabetic peripheral neuropathy (Meier et al., 2001). Studies conducted in clinical settings have shown that patients with motor disorders have lower braking ability than healthy older adults, people of the same age, and younger adults (Bishop et al., 2006; Meier et al., 2001; Morris & Iansek, 1997; Roeing et al., 2015; Serrao et al., 2013). Tucker et al. (2010b) measured RT in regard to stopping, and could not find any significant difference between older adults who had never fallen and those who had only fallen once previously, but a more detailed analysis of termination and braking ability, such as that conducted in this study, may contribute to the early identification of older adults who have a potentially high risk of falling. Our findings would also be helpful for understanding the reason for the high risk of falls among older adults and patients with motor disorders during transitions in movements.

There are some limitations to this study. First, the sample size was small, and all the participants in this study were male. Cooke et al. (1989) described that although it would be preferable to have equal numbers of male and female subjects in such studies, it was their experience that older males are more willing than females to participate in kinetic studies. Generally, the fall risk for older females is higher than that for older males (Jung et al., 2016) and, thus, examinations of older females in this regard are warranted. Another limitation was that this study only analyzed kinetic data using a force plate, and could not directly examine the different mechanisms used by older adults to perform movement initiation and termination. Thus, further studies should address this issue in-depth using other measurements (e.g., joint motion, muscle activities of the lower limbs, the center of mass) during various forms of the termination task.

5. Conclusion

In summary, older adults have more difficulty terminating their motion rapidly and fully than younger adults. The present study revealed that the delay in total motion shown by older adults is due not only to extended RT but also to extended BT, and these findings add new insight to understanding older adults' high risk of falling when moving in a standing posture. Based on our findings, it can be suggested that interventions to improve the braking abilities of older adults would contribute to reducing the risk of falls in this population.

Acknowledgments

Conflict of interest

There are no conflicts of interest to declare.

References

Bishop, M., Brunt, D., & Marjama-Lyons, J. (2006). Do people with Parkinson's disease change strategy during unplanned gait termination? *Neuroscience Letters, 397*, 240–244.
Bishop, M., Brunt, D., Pathare, N., & Patel, B. (2004). The effect of velocity on the strategies used during gait termination. *Gait & Posture, 20*, 134–139.
Breniere, Y., & Do, M. C. (1986). When and how does steady state gait movement induced from upright posture begin? *Journal of Biomechanics, 19*, 1035–1040.
Cao, C., Ashton-Miller, J. A., Schultz, A. B., & Alexander, N. B. (1998). Effects of age, available response time and gender on ability to stop suddenly when walking. *Gait & Posture, 8*, 103–109.
Chaput, S., & Proteau, L. (1996). Aging and motor control. *Journal of Gerontology: Psychological Sciences, 51B*, 346–355.
Cau, N., Cimolin, V., Galli, M., Precilios, H., Tacchini, E., Santovito, C., & Capodaglio, P. (2014). Center of pressure displacements during gait initiation in individuals with obesity. *Journal of NeuroEngineering and Rehabilitation, 11*, 82.
Chen, H. C., Ashton-Miller, J. A., Alexander, N. B., & Schultz, A. B. (1994). Age effects on strategies used to avoid obstacles. *Gait & Posture, 20*, 134–139.
Christou, E. A., & Carlton, L. G. (2001). Old adults exhibit greater motor output variability than young adults only during rapid discrete isometric contractions. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 56*, B524–532.
Cimolin, V., Cau, N., Galli, M., Santovito, C., Grugni, G., & Capodaglio, P. (2017). Gait initiation and termination strategies in patients with Prader-Willi syndrome. *Journal of NeuroEngineering and Rehabilitation, 14*, 44.
Cooke, J. D., Brown, S. H., & Cunningham, D. A. (1989). Kinematics of arm movements in elderly humans. *Neurobiology of Aging, 10*, 159–165.
Crenna, P., & Frigo, C. (1991). A motor programme for the initiation of forward-oriented movements in humans. *The Journal of Physiology, 437*, 635–653.
Dorfman, L. J., & Bosley, T. M. (1979). Age-related changes in peripheral and central nerve conduction in man. *Neurology, 29*, 38–44.
Grouios, G. (1991). Ageing effects on reaction time. *Journal of Physical Education, 28*, 18–22.
Hase, K., & Stein, R. B. (1998). Analysis of rapid stopping during human walking. *Journal of Neurophysiology, 80*, 255–261.
Hsieh, K. L., Roach, K. L., Wajda, D. A., & Sosnoff, J. J. (2019). Smartphone technology can measure postural stability and discriminate fall risk in older adults. *Gait & Posture, 67*, 160–165.
Hortobágyi, T., Mizelle, C., Beam, S., & DeVita, P. (2003). Old adults perform activities of daily living near their maximal capabilities. *The Journals of Gerontology Series*

A, Biological Sciences and Medical Sciences, 58, 453–460.

Jaeger, R. J., & Vanitchatchavan, P. (1992). Ground reaction forces during termination of human gait. *Journal of Biomechanics, 25*, 1233–1236.

Jeka, J., Oie, K., Schöner, G., Dijkstra, T., & Henson, E. (1998). Position and velocity coupling of postural sway to somatosensory drive. *Journal of Neurophysiology, 79*, 1661–1674.

Jian, Y., Winter, D. A., Ishac, M. G., & Gilchrist, L. (1993). Trajectory of the body COG and COP during initiation and termination of gait. *Gait & Posture, 1*, 9–22.

Jung, S., Yi, J., & Song, C. (2016). Biomechanical alterations of gait termination in middle-aged and elderly women. *Journal of Physical Therapy Science, 28*, 861–867.

Kannus, P., Sievänen, H., Palvanen, M., Järvinen, T., & Parkkari, J. (2005). Prevention of falls and consequent injuries in elderly people. *The Lancet, 366*, 1885–1893.

Kasahara, S., & Saito, H. (2015). Effect of loading parameters on motor performance during a dynamic weight-shift task. *Gait & Posture, 41*, 100–105.

Kasahara, S., Saito, H., Anjiki, T., & Osanai, H. (2015). The effect of aging on vertical postural control during the forward and backward shift of the center of pressure. *Gait & Posture, 42*, 448–454.

Kirker, S. G. B., Simpson, D. S., Jenner, J. R., & Wing, A. M. (2000). Stepping before standing: Hip muscle function in stepping and standing balance after stroke. *Journal of Neurology, Neurosurgery, and Psychiatry, 68*, 458–464.

Klous, M., Mikulic, P., & Latash, M. L. (2012). Early postural adjustments in preparation to whole-body voluntary sway. *Journal of Electromyography and Kinesiology, 22*, 110–116.

Meier, M. R., Desrosiers, J., Bourassa, P., & Blaszczyk, J. (2001). Effect of type II diabetic peripheral neuropathy on gait termination in the elderly. *Diabetologia, 44*, 585–592.

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., & Lord, S. R. (2009). Rapid gait termination: Effects of age, walking surfaces and footwear characteristics. *Gait & Posture, 30*, 65–70.

Morein-Zamir, S., Nagelkerke, P., Chua, R., Franks, I., & Kingstone, A. (2006). Compatibility effects in stopping and response initiation in a continuous tracking task. *The Quarterly Journal of Experimental Psychology, 59*, 2148–2161.

Morris, M. E., & Iansek, R. (1997). Gait disorders in Parkinson's disease: A framework for physical therapy practice. *NeuroReport, 21*, 125–131.

Pai, Y. C. (2003). Movement termination and stability in standing. *Exercise and Sport Sciences Reviews, 31*, 19–25.

Perry, S., Santos, L. C., & Patla, A. E. (2001). Contribution of vision and cutaneous sensation to the control of centre of mass (COM) during gait termination. *Brain Research, 913*, 27–34.

Peterson, C. L., Kautz, S. A., & Neptune, R. R. (2011). Braking and propulsive impulses increase with speed during accelerated and decelerated walking. *Gait & Posture, 33*, 562–567.

Roeing, K. L., Wajda, D. A., Motl, R. W., & Sosnoff, J. J. (2015). Gait termination in individuals with multiple sclerosis. *Gait & Posture, 42*, 335–339.

Saito, H., Yamanaka, M., Kasahara, S., & Fukushima, J. (2014). Relationship between improvements in motor performance and changes in anticipatory postural adjustments during whole-body reaching training. *Human Movement Science, 37*, 69–86.

Sparrow, W. A., & Tirosh, O. (2005). Gait termination: A review of experimental methods and the effects of ageing and gait pathologies. *Gait & Posture, 22*, 362–371.

Serrao, M., Conte, C., Casali, C., Ranavolo, A., Mari, S., Di Fabio, R., ... Sandrini, G. (2013). Sudden stopping in patients with cerebellar ataxia. *Cerebellum, 12*, 607–616.

Spirduso, W. W. (1975). Reaction and movement time as a function of age and physical activity level. *Journals of Gerontology, 30*, 435–440.

Temprado, J. J., Sleimen-Malkoun, R., Lemaire, P., Rey-Robert, B., Retornaz, F., & Berton, E. (2013). Aging of sensorimotor processes: A systematic study in Fitts' task. *Experimental Brain Research, 228*, 105–116.

Tirosh, O., & Sparrow, W. A. (2004). Gait termination in young and older adults: Effects of stopping stimulus probability and stimulus delay. *Gait & Posture, 19*, 243–251.

Tsuboi, M., Hasegawa, Y., Suzuki, S., Wingstrand, H., & Thorngren, K. G. (2007). Mortality and mobility after hip fracture in Japan: A ten-year follow-up. *The Journal of Bone and Joint Surgery, 89*, 461–466.

Tucker, M. G., Kavanagh, J. J., Morrison, S., & Barrett, R. S. (2010a). What are the relations between voluntary postural sway measures and falls-history status in community-dwelling older adults? *Archives of Physical Medicine and Rehabilitation, 91*, 750–758.

Tucker, M. G., Kavanagh, J. J., Morrison, S., & Barrett, R. S. (2010b). Differences in rapid initiation and termination of voluntary postural sway associated with ageing and falls-risk. *Journal of Motor Behavior, 42*, 277–287.