Age-related changes in the center of mass velocity control during walking

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ABSTRACT

During walking, the body center of mass oscillates along the vertical plane. Its displacement is highest at mid-swing and lowest at terminal swing during the transition to double support. Its vertical velocity (CoMv) has been observed to increase as the center of mass falls between mid- and late swing but is reduced just before double support. This suggests that braking of the center of mass is achieved with active neural control. We tested whether this active control deteriorates with aging (Experiment 1) and during a concurrent cognitive task (Experiment 2). At short steps of <.4 m, CoMv control was low and similar among all age groups. All groups braked the CoMv at longer steps of >.4 m but older subjects did so to a lesser extent. During the cognitive task, young subjects increased CoMv control (i.e. increase in CoMv braking) while maintaining step length and walking speed. Older subjects on the other hand, did not increase CoMv control but rather maintain it by reducing both step length and walking speed. These results suggest that active braking of the CoM during the transition to double support predominates in steps >.4 m. It could be a manifestation of the balance control system, since the braking occurs at late stance where body weight is being shifted to the contralateral side. The active braking mechanism also appears to require some attentional resource. In aging, reducing step length and speed are strategic to maintaining effective center of mass control during the transition to double support. However, the lesser degree of control in older adults indicates a true age-related deficit.

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Walking is associated with the oscillation of the body’s center of mass (CoM). It rises during early swing and falls after mid-swing [29]. In healthy young adults, the velocity of the falling CoM is reduced before foot contact. The ability to reduce the vertical velocity of the CoM (CoMv) is not innate. It undergoes a developmental phase that matures at about 5–6 years old [4], and deteriorates in individuals with neurologic disorders [7,27].

As we age, there are clear changes in many aspects of the walking gait that may contribute to the greater occurrence of falls or slips. Extensive research has shown that the walking gait of older adults is characterized by the shortened stride length, slower walking speed and greater time spent in the double-support phase [2,14,15,17,29]. The mechanisms underlying these changes include a decrease in balance control ability during walking [24]. Since the control of the falling CoM occurs during the transition from the single- to the double-support phase, the magnitude of CoM braking has been suggested to represent the integrity of the balance control system [8].

Braking the falling CoM before foot contact can be ascribed to an active mode of CoM control. It provides evidence that the CoM is not falling under the force of gravity. Instead, it suggests that the central nervous system prepares for foot contact by decreasing the CoMv to achieve a soft landing. In contrast to the active mode of CoMv control, passive braking from the mechanical impact of the swing limb at foot contact is said to predominate if CoMv braking is not observed before foot contact. In both cases, supporting muscular activities also contribute to prevent the body from collapsing [29].

Thus, by analyzing the change in CoM velocity before foot contact, we can make some distinctions between the active and passive modes of CoM velocity control. In this study, we applied the CoM braking concept in two experiments to better understand the mechanisms of active CoM control. In Experiment 1, we tested the hypothesis that CoM braking declines with aging. Experiment 2 tested the hypothesis that active CoM braking deteriorates during a...
concurrent cognitive task. Although walking is often considered an over-learnt skill, it is not a fully subconscious or automatic motor activity [16]. For example, reaction time during single support was found to be slower compared to the double support phase [20]. Walking balance control also decreased during dual-tasking activities [26]. The results of the current study have been published in preliminary forms [10,11,13].

In Experiment 1, a convenience sample of 80 subjects reporting no significant musculoskeletal or neurological impairments participated in the study approved by the institutional review board. Subjects were divided into four age groups: Group 1 = 20–34 years (mean = 29 ± 4 years, range = 20–34, n = 25), Group 2 = 35–49 years (mean = 42 ± 3 years, range = 37–49, n = 10), Group 3 = 50–64 years (mean = 58 ± 5 years, range = 50–64, n = 18), Group 4 = >65 years (mean = 73 ± 5 years, range = 65–88, n = 27).

Subjects stood on a 2-m triangular force plate. A walkway extended from the force plate by an additional 4-m long wooden platform. Subjects were instructed to take 5–6 steps per trial. In each trial, they walked at either their self-selected, shorter or longer steps. No specific instructions were given regarding cadence or how to walk [3]. Each subject walked a total of 10–15 trials in random order of step lengths. In each trial, the first step was captured for analyses. There were three sets of strain gauges embedded in each corner of the force plate which recorded vertical and horizontal ground reaction forces along the three axes, and the center of foot pressure [5].

The various step lengths taken by all subjects were partitioned into six categories: .3–.4 m, .41–.5 m, .51–.6 m, .61–.7 m, .71–.8 m, and >.8 m. Instantaneous vertical and anteroposterior (A/P) CoM velocities were calculated using simple integration of the vertical and A/P acceleration components of the CoM extracted from the ground reaction force data. Step length was measured from the instantaneous coordinates of the center of foot pressure. Ground reaction analog signals were digitized at 250 Hz.

The braking of the CoMv was quantified by expressing the velocity at foot contact as a function of the maximum velocity as follows:

$$\text{braking index} = \frac{V_1 - V_2}{V_1}$$

where $V_1$ = maximum CoM vertical velocity occurring between mid- to late stance, and $V_2$ = vertical velocity of CoM at foot contact. The index that is obtained from this ratio is a dimensionless number. It indicates the amount of change (i.e., braking) in CoM vertical velocity (CoMv) at foot contact relative to its maximum value [27]. If there is very little active braking of the CoMv, $V_2$ will be similar to $V_1$, and the braking index will be a small value approaching zero. Conversely, if CoMv is actively decreased before foot contact, then $V_2$ will have a small value and the braking index will have a large value approaching 1.

The braking index, maximum CoM vertical velocity and CoM vertical velocity at foot contact were each analyzed with a 4 (Group) × 6 (Step) ANOVA with repeated measures on Step using the “Proc mixed” command code of the SAS statistical software (v. 9.1, SAS Institute Inc., North Carolina, USA). Simple effect comparisons were made with the “lsmeans” command code to account for the unbalanced design of the study. The significance level for all analyses was set at $p < .05$.

Fig. 1 shows a typical recording in a single trial of the kinematics of the CoM in a young adult from Group 1 and an older adult from Group 4 taking the first step. The traces in the left and middle columns show the young subject taking a short (.33 m) and long step (.62 m), respectively, while the traces in the right column are from the older subject walking at a similar long step length. As seen in the middle row displaying the vertical velocity traces ($V_z$), both subjects’ velocity increased steadily following toe-off (TO). At about mid-swing, vertical velocities reached a maximum value ($V_1$). Prior to foot contact ($V_2$), the young subject shows little braking of the CoMv at the short step. At the longer step however, there is a strong braking of the CoMv before foot contact, resulting in a reduced velocity at foot contact ($V_2$) while the older subject showed little braking of the CoMv at a similar step length. Compared to the young

![Fig. 1](image-url)
subject, the reduced braking before foot contact in the older subject is also evident from the steeper positive slope occurring between maximum velocity and velocity at foot contact.

The main effects of Group and Step were significant, $F(3, 76) = 10.39, p < .0001$ and $F(3, 76) = 47.74, p < .0001$, respectively. At short steps of <.4 m, the braking index in all subjects showed similar low values. Group differences appeared in the remaining five step categories of >.41 m (Fig. 2(a)). All groups increased the active mode of braking but the gain was more prominent in the two younger groups. Young subjects in Groups 1 and 2 were similar until the longest step category of >.8 m, where the amount of CoM velocity braking in Group 2 decreased to become similar with the two older groups. Both younger groups were different from the subjects in Groups 3 and 4 until step length was >.71 m. In Groups 3 and 4, the amount of CoM velocity braking was similar across all step lengths.

Analyses of the maximum CoMv (V1) and CoMv at foot contact (V2) among the groups provided important clues about the effect of step length on the braking index. Maximum CoMv in all four groups increased steadily and similarly with increasing step lengths until the longest step category of >.8 m where group differences appeared (Fig. 2(b)). This resulted in a significant interaction effect of Group and Step, $F(15, 76) = 2.51$, $p = .005$.

In the analyses of the CoMv at foot contact, the results were similar to that obtained with the analyses of the braking index (Fig. 2(c)). There was a significant interaction effect of Group and Step, $F(15, 76) = 2.51, p < .005$. CoMv at foot contact was similar among the four groups at the shortest step length category. As step lengths increased, the two younger groups maintained similar low velocities compared to the two older groups until step lengths reached >.71 m. In these long steps, the youngest group showed the greatest amount of active control of CoMv with keeping CoMv foot contact at the lowest values.

The results of the study showed that when taking short steps of <.4 m, the predominant strategy for braking the falling CoM was the more passive mode of control. As step lengths increased, the shift to a more active mode of control was observed. Aging affected the active CoMv braking at long steps of >.6 m. Although the experiment was not set up to study the neuromuscular mechanisms of balance control during late stance, the results of the analyses of the maximum CoMv and CoMv at foot contact gave important information as to the major source of CoMv control. We briefly discussed them here: when we compared how CoMv is controlled among the groups across the different steps, we assumed that for each category, maximum CoMv is similar among the groups, so that any group difference in the braking index can be attributed primarily to the change in CoMv at foot contact. The pattern of increasing CoMv in all groups as a function of increasing step length held true up to where subjects were asked to take very long steps which has been shown to require a significant amount of spinal rotation [19].

Older adults appear to be able to compensate for a reduction in the range of pelvic rotation and flexibility of the trunk by increasing rotation of the spine about the vertical axis [25]. However, age-related changes in the complex coordination among the lower extremities, hip and trunk may make it difficult to adapt to long steps. Taking very long steps is an area that has received little attention in research and it is unclear as to how to account for the group differences found in this study. Nevertheless, the finding of increasing maximum CoMv with increasing steps is not surprising because step length has a positive correlation with progression velocity. The longer the steps, the faster a person tends to walk [23] and the faster the CoM falls between mid- to late stance as we have shown in this study. The older adults’ lower gain in braking index values may be explained by their inability to maintain a relatively constant CoMv at foot contact with increasing step lengths.

In Experiment 2, a convenience sample of 20 new health subjects reporting no significant musculoskeletal or neurological impairments participated in the study approved by the institutional review board. Subjects were divided into two age groups, each comprising 10 subjects: a Young group: 20–34 years old (mean = 29 ± 4 years, range 20–34), and an Older group >64 years old (mean = 73 ± 5 years, range 65–88).

Subjects were instructed to walk at their preferred (self-selected) pace and step length. Each subject walked a total of 3–7 trials. In each trial, the first three to four steps were captured for analyzes. The same protocol was repeated with a cognitive task: subjects were asked to quickly and accurately subtract backward by seven from a randomly given number. No instructions were given regarding how or whether to prioritize the walking and cognitive tasks. Subjects called out their response each time and the command to start walking was given after the second response. The experiment was recorded with the 6-camera 3D PEAK Motus motion capture system (120 Hz).

Coordinate data from the raw markers were low-passed filtered using a dual pass fourth-order Butterworth filter and a 6 Hz cutoff frequency. The kinematics of the CoM were derived from an 11-segment CoM model [28]. The braking index, step length and walking speed were analyzed separately with a $2 \times (Age) \times 2 \times 2$ (Dis-
In older adults, the results of the present study showed that CoM braking capacity is altered and related to the atrophy of the mesencephalon [7]. The high frequency electrical stimulation applied in either the subthalamic nucleus or the substantia nigra pars reticulata induced an increased in this mechanism, independent of levodopa supplementation [9]. This suggests that basal ganglia and descending projecting pathways to the brainstem structures are involved in the control of this active CoMv braking. The improvement from the stimulation could result from better coordination among the muscles in the lower extremities [9] including for example, between the hamstring muscle of the swing limb and the gastrocnemius muscle of the stance limb [22]. Somatosensory inputs also appear to contribute to CoM braking as its magnitude decreased when subjects walked on a foam or took a single step [8]. Conversely, removal of vision has no effect on CoMv braking [8], suggesting that vision is not necessary to brake the falling CoM observed during the single-support phase of the walking gait cycle.

In older adults, the results of the present study showed that CoMv braking is as if they increased their balance control during the late stance period of the gait cycle. It remains to be studied whether healthy young adults can actually increased their attentional capacity to produce the results obtained in this study. It would be difficult to fit such a post hoc explanation to theories of attentional resource that subscribe to the general idea of a limited attentional capacity which predicts a deterioration in performance when attention is so-called divided. The results of the study may be more compatible with alternative theories that incorporate the deployment of adaptive sensorimotor strategies as a mechanism to explain the wide ranging capacity for humans to multi-task [6,18,21]. We will explore such ideas in follow-up experiments that extend from the current study.

General discussion

The mechanisms underlying the control of CoM vertical velocity remains to be studied. In Parkinsonian patients, the CoM braking capacity is altered and related to the atrophy of the mesencephalon [7].

Due to space constraints, our subjects’ walking was restricted to 3–4 steps per trial. As a result, we were unable to analyze the speed and accuracy of the subjects’ subtraction response. The results nevertheless strongly suggest that the cognitive task did indeed achieve its intended effect in distracting the subjects from the walking activity [1].

The results obtained in the younger group were interesting and warrant some consideration here. Attentional demand using the simple reaction time protocol has been shown to be higher in the single-limb support phase of the walking gait cycle compared to the double-support phase [20]. Yet in the presence of cognitive distraction, the young adults in the present study not only maintained their step length and speed but produced higher braking index values. It is as if they increased their balance control during the late stance period of the gait cycle. It remains to be studied whether healthy young adults can actually increased their attentional capacity to produce the results obtained in this study. It would be difficult to fit such a post hoc explanation to theories of attentional resource that subscribe to the general idea of a limited attentional capacity which predicts a deterioration in performance when attention is so-called divided. The results of the study may be more compatible with alternative theories that incorporate the deployment of adaptive sensorimotor strategies as a mechanism to explain the wide ranging capacity for humans to multi-task [6,18,21]. We will explore such ideas in follow-up experiments that extend from the current study.

General discussion

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Fig. 3. Mean and standard error of (a) braking index, (b) walking speed, and (c) step length as a function of age and distraction. * p < .05.
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