



Control of body's center of mass motion relative to center of pressure during uphill walking in the elderly



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ABSTRACT

Uphill walking places more challenges on the locomotor system than level walking does when the two limbs work together to ensure the stability and continuous progression of the body over the base of support. With age-related degeneration older people may have more difficulty in maintaining balance during uphill walking, and may thus experience an increased risk of falling. The current study aimed to investigate using gait analysis techniques to determine the effects of age and slope angles on the control of the COM relative to the COP in terms of their inclination angles (IA) and the rate of change of IA (RCIA) during uphill walking. The elderly were found to show IAs similar to those of the young, but with reduced self-selected walking speed and RCIA ($P < 0.05$). After adjusting for walking speed differences, the elderly showed significantly greater excursions of IA in the sagittal plane ($P < 0.05$) and increased RCIA at heel-strike and during single limb support (SLS) and double limb support (DLS) in the sagittal plane ($P < 0.05$), and increased RCIA at heel-strike in the frontal plane ($P < 0.05$). The RCIA were significantly reduced with increasing slope angles ($P < 0.05$). The current results show that the elderly adopted a control strategy different from the young during uphill walking, and that the IA and RCIA during walking provide a sensitive measure to differentiate individuals with different balance control abilities. The current results and findings may serve as baseline data for future clinical and ergonomic applications.

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1. Introduction

More than 30% of the older population aged above 65 in the USA fall at least once during a year [1]. The risk of falling in the elderly is increased when facing increased challenges to balance control while walking on uneven terrains such as up stairs and slopes [2]. Walking up slopes was found to pose a greater risk of falling than walking up stairs with similar inclinations [3]. Uphill walking also places greater challenge on the locomotor system than level walking does [3–6], with different kinematics and between-limb load-sharing to ensure the stability and continuous progression of the body over the base of support (BOS) [5]. With age-related degeneration such as reduced muscle strength, limited range of joint motion and impaired perception [7], older people may have

more difficulty in maintaining balance during uphill walking, and are thus at greater risk of falling.

Previous studies on uphill walking have been mostly on young adults, analyzing spatio-temporal parameters, and joint kinematics and kinetics of the locomotor system [4,6,8]. Support moments have also been used to study the effects in young adults of slope angles on the demands on the lower limb joints in preventing the body from collapsing during uphill walking [5]. It was only more recently that studies on the lower limb mechanics and energy expenditure during uphill walking in the elderly were reported [9]. To the best knowledge of the authors, no study has examined the balance control of the body in terms of the motion of the body's center of mass (COM) relative to the center of pressure (COP) during uphill walking, neither in young adults, nor in older adults.

The motions of the body's COM and the COP of the supporting foot have been used to examine the stability of the body during level walking [10]. The relative motions between COM and COP, described using COM–COP separations in terms of COM–COP distances in the horizontal plane, have been used to study healthy older adults walking over obstacles [11]. Since the magnitudes of

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the COM motion, and thus the COM–COP distances, may be affected by a subject's stature [12] and/or the terrain conditions, COM–COP distances have frequently been normalized by leg length (LL) or body height (BH) [13]. However, such constant normalization factors do not include the effects of the COM height changes on the COM–COP distances, especially during slope walking. Another measure, minimum COM–BOS margin, has been suggested [14] but is not sufficient to describe the dynamic stability, which depends on both the position and the velocity of the COM with respect to the BOS [15]. Hof et al. [16] overcame the limitation by introducing the so-called XcoM (an extrapolated COM that considers both the COM position and its velocity) and the margin of stability as the minimum distance from the XcoM to the boundaries of the BOS. However, the continuously varying BOS during walking, during double limb support in particular, presents another difficulty in the definition of the COM–BOS margin or the margin of stability. Moreover, the above-mentioned methods were developed for level walking with the horizontal plane as reference. Therefore, their application to slope walking is either limited or has not yet been demonstrated.

Inclination angles (IAs) defining the orientation of the line connecting the COP and COM, and the rate of change of IA (RCIA) have been used to describe the body's dynamic control without the influence of stature differences or the need for defining dynamic BOS [17–19]. Moreover, they can be applied to slope walking without limitation. Generally, the COM does not have to be kept within the BOS during movement but appropriate COM velocity relative to the BOS is necessary for body stability [15]. For a COM position, there is a corresponding range of COM velocity for dynamic stability, a COM velocity outside of which would initiate falls. The IA and RCIA consider both the positions and velocities of the COM and COP, providing a better description of the body's balance control, thus enabling the study of the control strategies for dynamic stability between groups and between slope conditions during walking. Older adults were found to show an increased anterior RCIA with an IA similar to the young during obstacle-crossing, a task more challenging than walking [13]. Therefore, when walking up slopes, it is possible that, in order to maintain progression with similar IA, the elderly may also show increased anterior RCIA, which would likely be further increased with increasing slope angles.

The current study aimed to investigate the effects of age and slope angles on the control of the COM relative to the COP in terms of IA and RCIA during uphill walking. It was hypothesized that compared to young adults, older people would show increased RCIA while maintaining similar IA when walking uphill, and that the RCIA would be increased with increasing slope angles.

2. Materials and methods

Fifteen older male adults (age: 69.9 ± 4.3 years, height: 165.3 ± 6.5 cm, mass: 63.3 ± 4.4 kg) and fifteen young male controls (age: 32.6 ± 5.2 years, height: 168.9 ± 5.5 cm, mass: 68.4 ± 8.7 kg) participated in the current study with informed written consent, as approved by the Institutional Research Board. All subjects were free of neuromusculoskeletal dysfunction and had normal or corrected-to-normal vision. A subject would be excluded if he had any distinct gait asymmetries, or was taking medication that might influence gait.

Each subject walked at a self-selected pace on an 8-meter walkway without slopes (level walking) and with 3-m slopes of 5° , 10° and 15° (slope conditions). Each slope was formed by two wedged wooden blocks placed on two adjacent force plates (AMTI, USA) and two further blocks each placed next to one of the force plates along the walkway, i.e., one leading up to the first force plate and the other behind the second force plate (Fig. S1) [5,20]. Each subject took at least two steps on the slope before and after

stepping onto the force plates. The sequence of the test conditions (level walking and slopes) for each subject was randomized. Before the tests, each subject was allowed to walk on the level or sloped walkway several times. The motions of the body segments, namely head and neck, trunk, pelvis, arms, forearms, thighs, shanks and feet, were measured using 39 infrared retro-reflective skin markers [21]. Three-dimensional trajectories of the markers were measured using a seven-camera motion analysis system (Vicon 512, Oxford Metrics Group, UK) at a sampling rate of 120 Hz, and the ground reaction forces (GRF) were measured using the force plates at a sampling rate of 1080 Hz. A fourth-order Butterworth low-pass filter with a cut-off frequency of 15 Hz was used to filter both the kinematic and force plate data [22]. Before and after the foot was on the force plate, heel-strikes and toe-offs of the limb were determined according to a 5N-threshold for the GRF or by a foot marker-based algorithm [23]. The limb used to step onto the first force plate was taken as the reference limb. Six successful trials, three for each limb as reference, were obtained for each test condition and for each subject.

A 13-body-segment model of the whole body [24] was used to calculate the position of the body's COM as the weighted sum of the segmental COMs of all body segments. Body segmental inertial properties were obtained using an optimization-based method [25]. Effects of soft tissue artifacts were reduced using a global optimization method [26]. The COP position was calculated using forces and moments measured by the force plates [27]. The medial/lateral positions of the COM and COP were described relative to the line of progression that bisected the medial/lateral range of motion of the COM during a gait cycle, a positive value being to the side of the contralateral limb [13,19]. The anterior/posterior positions of the COM and COP were described parallel to the direction of progression, a zero value being the position of heel-strike and a positive value being anterior to that position. The IA in the sagittal plane (α) and frontal plane (β) were calculated as follows [18,19]

$$\vec{t} = \left(\frac{\vec{P}_{\text{COM-COP}} \times \vec{Z}}{|\vec{P}_{\text{COM-COP}}|} \right) \quad (1)$$

$$\alpha = \sin^{-1}(t_x) \quad (2)$$

$$\beta = \sin^{-1}(t_y) \quad (3)$$

where $\vec{P}_{\text{COM-COP}}$ was the vector pointing from the COP to the COM, and $\vec{Z}$ was the unit vector of the vertical axis of the global coordination system. With the current force plate setup, α and β were calculated from the beginning of the SLS to the subsequent heel-strike. The rates of changes of α and β were also calculated by smoothing and differentiating their trajectories using the generalized cross-validatory spline method [28].

The values of the IA and RCIA were averaged over the double-limb support (DLS) and SLS, and the maximum and minimum of RCIA were also obtained. The values of the IA and RCIA at heel-strike of the contralateral leg and toe-off of the ipsilateral leg were also extracted because the transitions between SLS and DLS are critical instances at which maintaining body stability is expected to be more difficult [18]. Spatio-temporal parameters including gait speed, step length, step width and cadence for all conditions were also calculated. The values of each of the variables from the six successful trials were averaged for each slope condition and for each subject.

A two-way repeated measures analysis of variance (ANOVA) was performed to analyze the effects of age and slope on the spatio-temporal variables, namely walking speed, cadence, step length and step width. A previous study has shown that walking speed directly affects the magnitudes of the IAs during walking [17]. Therefore, if the results of ANOVA showed a significant effect

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