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Redistribution of intra- and inter-limb support moments during downhill walking on different slopes

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ABSTRACT

Downhill walking presents a greater risk of falling as a result of slipping or loss of balance in comparison with level walking. The current study aimed to investigate the effects of inclination angles on the intra-limb (inter-joint) and inter-limb sharing of the body support during downhill walking for a better understanding of the associated control strategy. Fifteen young male adults (age: 32.6 ± 5.2 years, height: 168.9 ± 5.5 cm, mass: 68.4 ± 8.7 kg) performed level and downhill walking while their kinematic and kinetic data were measured for calculating joint moments and total support moments of the lower limbs using inverse dynamics analysis. The peak total support moments of both the leading and trailing limbs increased with increasing inclination angles ($p < 0.05$) with different sharing patterns among individual joints. Being the major contributor to the peak total support moment during early single-limb support, the contribution of the knee remained unaltered ($p > 0.05$), but the contributions of the hip increased with reduced contributions from the ankle ($p < 0.05$). For the increased peak total support moment during late single-limb support, the intra-limb sharing changed from a major ankle contribution to a major knee contribution strategy. The hip contribution was also increased ($p < 0.05$) but the hip flexor moment remained unaltered ($p > 0.05$). During double-limb support, the main contributor to the whole body support changed from the trailing limb to the leading limb with increasing inclination angles ($p < 0.05$).

1. Introduction

Downhill walking presents a greater risk of falling as a result of slipping or loss of balance in comparison with level walking (Redfern, DiPasquale, 1997; Sheehan and Gottschall, 2012). During downhill walking, the joints of the stance limbs have to exert the necessary moments to prevent collapse while balancing and supporting the body under a downward gravitational shear force that increases with increasing inclinations. These demands have to be met by a well-controlled coordination of the joints of the lower limbs associated with different kinematic configurations of the body segments when facing different slopes. Implementation of these control strategies can be very challenging for individuals with muscle or joint pathologies. Knowledge of the intra-limb (inter-joint) and inter-limb sharing of the mechanical demands in providing the necessary body support during downhill walking is thus helpful for developing strategies for fall prevention.

Downhill walking has been studied using gait analysis in terms of spatio-temporal parameters, and angles and moments of individual joints of the locomotor system (Franz et al., 2012; Gao et al., 2008; Kang et al., 2002; Kawamura et al., 1991; Lay et al., 2006; Leroux et al., 2002; McIntosh et al., 2006; Mrozowski and Awrejcewicz, 2011; Prentice et al., 2004; Wall et al., 1981). While these data helped establish knowledge of the joint mechanics during downhill walking, they do not provide a direct description of the overall locomotor control strategy for body support because joint moments indicate loads at individual joints without revealing the inter-joint load-sharing for that purpose (Winter, 1980). A quantitative characterization of the coordination of the joint kinematics and kinetics within and between the lower limbs is needed to bridge the knowledge gap (Hong et al., 2014). The concept of the total support moment (Ms), defined as the numerical sum of the extensor moments at the hip and knee, and the plantarflexor moment at the ankle, presents a good approach (Winter, 1980).

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A wide range of combinations of the extensor moments of the hip and knee, and plantarflexor moments of the ankle, could be used by individuals to achieve similar kinematic patterns during gait (Winter, 1984). Proper combinations of joint moments with respect to a specific kinematic pattern are necessary to prevent collapse of the lower limbs while balancing and supporting the body weight (BW) (Winter, 1995). This may vary with different downhill inclinations. Comparisons of the contributions of individual joint to the total support moment (CMs) could also be useful for quantifying the possible strategies in normal walking and those in response to injury or intervention, such as in patients with stroke (Wren et al., 2011) and musculoskeletal pathology (Nadeau et al., 1997, 1998; Salsich et al., 2001).

A major difference between downhill and level walking is that the levels of the feet during double-limb support (DLS) of the former are not even, requiring different weight-sharing between the limbs. How the joint moments in the leading and trailing limbs are controlled during weight transfer to produce the necessary support for the whole body in response to different slopes remains unclear. Lay et al. (2006) reported the changes of total support moments in response to increased inclinations during downhill walking. However, the contributions of the individual joints to the total support moment during the stance phase, and the sharing of the whole body support moment between the two limbs during the DLS, have not been studied.

The current study aimed to investigate the effects of inclination angles on the intra- and inter-limb sharing of the demands for body support during downhill walking in terms of total support moment and contributions of individual joint moments to the total support moment.

2. Materials and methods

Fifteen young male adults (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.

Each subject wore 28 retro-reflective markers on specific anatomical landmarks (Chen and Lu, 2006) and walked at a self-selected pace on an 8-m walkway without slopes (level walking) and with 3-m slopes of 5°, 10° and 15° (slope conditions). Each slope was formed by four separate wooden units (A–D, Fig. 1). The first unit was a wedge (A) placed in front of the first forceplate while two wooden blocks (B and C) were placed on each of the two forceplates (AMTI, USA). The final unit (D) was a U-shaped wedge that was placed around the two forceplates from behind the second one so as to form a continuous slope with a horizontal surface at the end (Fig. 1). The top surfaces of the blocks were made to be consistent to the required slope surface. Each of the blocks was positioned so as not to make contact with any of the others. Thin anti-slip mats were attached to the bottom of the blocks to prevent them from sliding on the floor/force plates. Each subject took at least two steps on the slope before and after stepping on the forceplates. 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. Switching between test conditions was achieved by replacing the current blocks with those for the next slope or by removing them for level walking. The motions of the body segments were measured using a 7-camera motion analysis system (Vicon 512, OMG, UK) at a sampling rate of 120 Hz while the ground reaction forces (GRF) were measured by the forceplates at a sampling rate of 1080 Hz. Six successful trials were obtained for each slope condition.

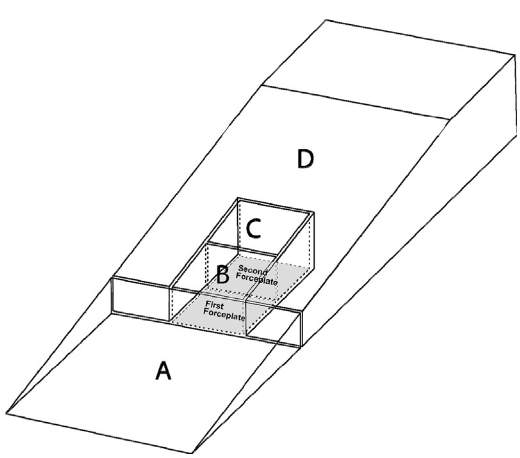

Fig. 1. A schematic diagram showing the sloped walkway formed by four separate wooden units (A–D). The first unit was a wedge (A) placed in front of the first forceplate while two wooden blocks (B and C) were placed on each of the two forceplates. The final unit (D) was a U-shaped wedge that was placed around the forceplates from behind the second one so as to form a continuous slope with a horizontal surface at the end. The top surfaces of the blocks were made to be consistent to the required slope surface. Each of the blocks was positioned so as not to make contact with any of the others.

For dynamic analysis the pelvis-leg apparatus was modeled as a system of seven rigid segments, namely the pelvis, thighs, shanks and feet, connected by model joints (Chen and Lu, 2006). Each body segment was embedded with an orthogonal coordinate system with the positive x-axis directed anteriorly, the positive y-axis superiorly and the positive z-axis to the right. A Cardanic rotation sequence (z–x–y) was used to describe the rotational movements of each joint (Grood and Suntay, 1983). With the measured kinematic and GRF data, the internal moments at the lower limb joints were calculated using inverse dynamics analysis. Effects of soft tissue artifacts were reduced by using a global optimization method (Lu and O'Connor, 1999). Body segmental inertial properties were obtained using an optimization-based method (Chen et al., 2011).

In order to account for anatomical variations between the subjects, all the calculated joint moments were normalized to body weight (BW, unit: N) and leg length (LL, unit: m) and all the GRFs were normalized to BW. The lever-arm length of the GRF at the each joint was also calculated as the joint moment divided by the magnitude of the GRF in the sagittal plane (unit: cm).

The total support moment (Ms) of a limb was calculated as the sum of the net joint moments at the hip, knee and ankle during stance phase, which showed a characteristic double peak pattern similar to that of the ground reaction curve during level walking (Winter, 1980). The contributions of the hip, knee and ankle to the first and second peaks of Ms (Ms_1 and Ms_2) were calculated by dividing the joint moment value by the corresponding peak values of Ms. The whole body support moment (WMs) was also calculated as the summation of Ms of both limbs. For studying the coordination between the limbs in providing WMs for overall support of the body's COM during downhill walking, the time integral of Ms for each limb during DLS was calculated as a percentage of the time integral of WMs. The angle and moment components for each joint at which Ms_1 and Ms_2 occurred (i.e., P_1 and P_2) were extracted for each trial for subsequent statistical analysis. The spatial–temporal variables, namely walking speed, cadence, step length, and durations of single-limb support (SLS) and DLS, were also calculated.

A one-way repeated measures analysis of variance (ANOVA) with inclination angle as the main factor was performed to analyze the effects of inclination on walking speed, cadence and step length during slope walking. A mixed-model one-way analysis of co-variance (ANCOVA) with walking speed as a covariate was used to analyze the effects of inclination on each of the other dependent variables. Whenever an inclination effect was found for a dependent variable, a polynomial test (Kutner et al., 2005) was carried out to determine whether the changes of the dependent variable showed a linear trend with respect to increasing inclination angles (5°, 10°, and 15° slopes), with or without including the condition of level walking (i.e., 0° slope). All significance levels (level walking and 5°, 10°, 15° slopes) were set at $\alpha = 0.05$. All statistical analyses were performed using SAS version 9.2 (SAS Institute Inc., NC, USA).

3. Results

3.1. Spatial–temporal variables

The walking speed had no covariate effects on any of the other dependent variables. The spatial–temporal variables were not significantly affected by slope angles except for step length that varied with a linearly decreasing trend with increasing slope angles (Table 1).

3.2. Joint angles and moments, and lever-arm lengths of GRF

The joint angle and moment patterns in the lower limb during downhill walking were similar to those during level walking but with different magnitudes. The hip extension at P_2 was found to decrease linearly with increasing inclination angles but the corresponding moments were not significantly affected (Table 3). The knee flexor and extensor moments increased linearly at both P_1