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Lower limb joint forces during walking on the level and slopes at different inclinations

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ABSTRACT

Sloped walking is associated with an increase of lower extremity joint loading compared to level walking. Therefore, the aim of this study was to analyse lower limb joint compression forces as well as tibiofemoral joint shear forces during sloped walking at different inclinations. Eighteen healthy male participants (age: 27.0 ± 4.7 years, height: 1.80 ± 0.05 m, mass: 74.5 ± 8.2 kg) were asked to walk at a pre-set speed of 1.1 m/s on a ramp (6 m × 1.5 m) at the slopes of −18°, −12°, −6°, 0°, 6°, 12° and 18°. Kinematic data were captured with a twelve-camera motion capture system (Vicon). Kinetic data were recorded with two force plates (AMTI) imbedded into a ramp. A musculoskeletal model (AnyBody) was used to compute lower limb joint forces. Results showed that downhill walking led to significantly increased hip, tibiofemoral and patellofemoral joint compression forces ($p < 0.05$) and to significantly decreased ankle joint compression forces ($p < 0.05$). Uphill walking significantly increased all lower limb joint compression forces with increasing inclination ($p < 0.05$). Findings that downhill walking is a stressful task for the anterior cruciate ligament could not be supported in the current study, since anterior tibiofemoral joint shear forces did not increase with the gradient. Due to diverse tibiofemoral joint shear force patterns in the literature, results should be treated with caution in general. Finally, lower limb joint force analyses provided more insight in the structure loading conditions during sloped walking than joint moment analyses.

1. Introduction

Locomotion on slopes is a challenging task in daily life being even more pronounced during hiking. Positive effects of hiking have been shown [1], but this sport can also cause pain and injuries of the musculoskeletal system [2]. Most frequently, pain was reported in the knee joint during downhill walking [3]. Therefore, several research groups have analysed the effects of sloped walking on joint kinematics and kinetics [4,5].

Sloped walking is associated with an increase of lower extremity joint loading compared to level walking [3,4]. During downhill walking joint moments at the knee, and to a lesser extent the ankle and hip, were affected by the inclination. Knee extension moments showed large increases with increased inclination [4,6] and there is evidence that the anterior cruciate ligament is highly loaded during downhill walking [7]. During uphill walking, an increased hip extensor moment occurs with increasing inclination [3,4].

Joint moments, however, do not account for changes in muscle co-activation that may occur during uphill and downhill walking when compared to level walking and thus may not reflect the real joint loading situation [4,8]. The use of musculoskeletal models estimating lower limb muscle and joint loads during walking yields more detailed and valid information [8,9]. The computed forces may permit a better understanding of the relative musculoskeletal demands and potential risks of musculoskeletal injuries and pathologies [10]. Some data exists on *in-vivo* measurements of hip [11] and knee [12] joint forces from direct internal structure force measurements by using instrumented endoprostheses. This type of measurement, however, is limited to participants with impaired function. To analyze a non-impaired population, musculoskeletal models are commonly used to calculate joint loadings [10]. While early studies estimated knee joint forces using 2D-analyses [5,7], recent models can predict 3D muscle and joint forces by employing more complex quasi-static or dynamic optimization techniques [8,12,13].

Several studies analysed lower limb joint forces using musculoskeletal models during level walking [9,14,15], while for sloped walking, the number of studies has been limited [5,7,8]. The available studies on sloped walking only provide limited information. More

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specifically, Kuster et al. [7] and Schwameder et al. [5] used a model which included only the quadriceps, but no other muscles. These models were used to analyse downhill walking for one specific inclination [7] or to show that knee joint forces can be effectively controlled by varying step length and cadence during downhill and uphill walking on an 18° inclined ramp [5]. Haight et al. [8] found reduced early stance tibiofemoral joint loads during slow (0.75 m/s), 6° uphill walking compared to faster (1.50 m/s), level walking in obese and non-obese adults and suggested that uphill walking may be an appropriate exercise for obese individuals. Finally, these studies on sloped walking only focused on knee joint forces.

It appears that no study to date, has investigated the effect of uphill and downhill walking at different inclinations on hip, tibiofemoral, patellofemoral and ankle joint compression forces using a musculoskeletal model. Additionally, since downhill walking is stated to be a stressful task for the anterior cruciate ligament [7], tibiofemoral joint shear forces will also be analysed. Therefore, the purpose of this study was to analyse hip, tibiofemoral, patellofemoral and ankle joint compression forces as well as tibiofemoral shear forces comparatively during uphill and downhill walking on different inclinations and level walking.

2. Methods

2.1. Participants

Eighteen healthy male participants (age: 27.0 ± 4.7 years, height: 1.80 ± 0.05 m, mass: 74.5 ± 8.2 kg) volunteered to participate in this study. The study was approved by the ethics board and written informed consent was signed by all participants.

2.2. Experimental protocol

Participants were asked to walk at a pre-set speed of 1.1 m/s on a ramp (6 m × 1.5 m) at different inclination angles of 0°, ±6°, ±12° and ±18°. The ramp was tilted to each inclination (controlled by a digital goniometer) around the pivot point in the middle of the ramp using an electric-hydraulic device. Participants walked about five minutes at each inclination for familiarization. Usually four steps were made before the measurement. Speed was controlled via a timing device (Brower Timing Systems, Draper, Utah, USA). Trials were discarded if the participant's speed differed more than 2.5% from the target speed, if the foot was not completely planted on the force platform or if visually obvious stride alterations to contact the force platform were observed. The trial with the speed closest to the target speed was used for further analysis. The order of testing the various slopes was randomized among the participants.

2.3. Data collection

Reflective markers (diameter: 15 mm) were attached to the participants according to the Cleveland Clinic Marker set (Motion Analysis Corp, Santa Rosa, USA). Kinematic data were captured with a twelve-camera, marker based motion capture system (Vicon, Oxford, Oxford Metrics Ltd, UK; 250 Hz) and kinetic data were recorded with two force plates (AMTI, Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA; 1000 Hz) imbedded into the middle section of the ramp. The force plates were mounted on a separated supporting frame without any contact to the ramp.

2.4. Data analysis

Kinematic data were further processed with Vicon Nexus (Vicon, Oxford Metrics Ltd, UK). Processed kinematic and force plate data were imported into the AnyBody Modeling System (vers. 6.0, AnyBody Technology, Denmark) to calculate lower limb joint forces. The musculoskeletal model used in this study was a standard model (AMMR 1.6.2, MoCapModel) available in the software. The morphological data set for the lower extremities of Klein Horsman et al. [16] was used to model mass, moments of inertia, and muscle sites/geometry for all segments. Each leg contains 55 muscles and has seven degrees of freedom: three hip joint, one knee joint, one ankle joint, one subtalar joint and one patellar movement. The model was scaled to match each participant's anthropometry [17]. Inverse dynamics were performed and a third order polynomial muscle recruitment criterion was used [18]. The objective function allowed for synergies between muscles and a constraint preventing individual muscle forces from exceeding their physiological maximum was added [18]. Lower limb joint forces were calculated in the following reference frames: hip: femoral reference frame, tibiofemoral: tibial reference frame, patellofemoral: patella reference frame, and ankle: tibial reference frame. For the tibiofemoral joint shear force, anterior forces (positive values) represent a strain on the anterior cruciate ligament. Comparing estimated muscles activity (model) with measured EMG data, following was observed: correlation coefficients for ~69% of the conditions >0.7, rather high mean absolute errors (only 38% below 30%) and trend analysis revealing similar activities for all muscles and tasks (except tibialis anterior during uphill walking). Since EMG measurements have several limitations itself, the model was found to be suitable for sloped walking [19].

Kinematic and kinetic data were filtered using a Butterworth low pass filter with 10 and 15 Hz cut off frequencies, respectively, based on residual analysis [20]. Joint forces were normalized to body weight (×BW). Each trial was time normalized to stance phase duration and the ensemble mean was calculated over all participants. Mean and maximum hip, tibiofemoral, patellofemoral and ankle joint compression forces and mean and maximum tibiofemoral shear forces were calculated for each participant and mean values were then calculated over all participants. For additional information, ground reaction forces are presented in the Appendix.

2.5. Statistics

Statistical analysis was conducted using SPSS (version 22.0, IBM, Armonk, NY, USA). The significance level was set to $\alpha = 0.05$. Requirements for normality were achieved (Kolmogorov–Smirnov). Changes in mean and maximum joint forces with respect to different inclinations were analysed using one-way ANOVA with repeated measures. In cases of significance, a Bonferroni corrected ($\alpha = 0.00238$) post hoc test was performed for pairwise comparison (21 pairs, Appendix). Effect sizes for each comparison were quantified using Cohen's d_z to be small ($d = 0.20$–0.49), medium ($d = 0.50$–0.79) or large ($d > 0.80$) [21].

3. Results

The gait analysis revealed significant main effects of the inclination ($p < 0.05$) on all mean and maximum lower limb joint forces. Comparing sloped to level walking following results were found: downhill walking increased hip, tibiofemoral and patellofemoral compression forces and decreased ankle compression forces, while uphill walking increased all lower limb joint forces with increasing inclination.

Hip compression forces showed two peaks during stance phase (Fig. 1). Mean hip compression forces remained unaltered during downhill walking and ascending 6°, but were significantly increased during 12° (9.3%, $p < 0.001$, $d = 1.04$) and 18° (21.5%, $p < 0.001$, $d = 1.38$) uphill walking (Table 1). Maximum hip