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Effect of sloped walking on lower limb muscle forces

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Lower limb joint loadings are increased during sloped walking compared to level walking and muscle forces are major contributors to lower limb joint forces. Therefore, the aim of this study was to analyze lower limb muscle forces during sloped walking at different inclinations. Eighteen healthy male participants (27.0 ± 4.7 y, 1.80 ± 0.05 m, 74.5 ± 8.2 kg) walked at a pre-set speed of 1.1 m/s on a ramp at the inclinations of 0°, $\pm 6^\circ$, $\pm 12^\circ$ and $\pm 18^\circ$. Kinematic data were captured with a motion capture system and kinetic data were recorded with two force plates imbedded into the ramp. A musculoskeletal model was used to compute lower limb muscle forces (normalized to body weight and gait cycle duration). During downhill walking gluteus maximus, quadriceps, soleus, peroneus and tibialis anterior muscle forces increased ($p \leq 0.002$) compared to level walking, while gluteus minimus, piriformis, adductor, iliopsoas, hamstrings and gastrocnemii muscle forces decreased ($p \leq 0.002$). Uphill walking decreased gluteus minimus, iliopsoas and tibialis anterior muscle forces ($p \leq 0.002$), while all other muscle forces increased ($p \leq 0.002$, except gluteus medius). Joint-muscle-force waveforms provided information on possible muscle contributions to joint compression forces. The most important muscles were: gluteus medius for hip forces, quadriceps and gastrocnemii for tibiofemoral forces, quadriceps for patellofemoral forces and triceps surae for ankle forces. The contribution of each muscle changed with the inclination during sloped walking compared to level walking. The current study provided important information on muscle forces during sloped walking that can be useful for rehabilitation and training procedures.

1. Introduction

Muscles have been shown to be the major contributors to the joint contact forces [1–3], while gravitational and centrifugal forces combined contribute less than 5% of the total contact force [1]. Estimation of individual muscle forces can provide insight on tissue loading, and may lead to a better understanding of the musculoskeletal demands and potential risks of musculoskeletal injuries [4]. Since direct measurement of muscle forces is generally not feasible, non-invasive methods based on musculoskeletal modelling should be considered [4]. Previous studies have confirmed that the use of musculoskeletal models estimating lower limb muscle and joint loads during walking can provide detailed information [5,6].

Sloped walking is associated with kinematic changes to raise the limb for toe clearance and heel strike, to lift the body during ascending, and for a controlled descent [7]. Furthermore, increases in lower extremity joint loadings have been reported compared to level walking [7,8]. Joint moments, however, do not account for changes in muscle activation that may occur during downhill and uphill walking when compared to level walking and thus may not reflect the real joint loadings [5,7]. Analysing sloped walking with the use of a musculoskeletal model has revealed increased hip, tibiofemoral and patellofemoral compression forces and decreased ankle joint compression forces during downhill walking, while uphill walking increased all lower limb joint forces with increasing inclinations [9]. Furthermore, mean tibiofemoral compression forces were lower or equal during downhill walking compared to the same inclination during uphill walking, while mean patellofemoral joint compression forces were found to be higher. Therefore, the analysis of knee joint forces has provided more insight into the structure loading conditions than joint moment analysis [9].

Several studies have analyzed lower limb muscle forces during level walking in able-bodied participants [1,10,11] as well as patients [6,11,12], but muscle forces during sloped walking have been minimally analyzed to date [5,13]. Haight et al. [5] found reduced gastrocnemius, quadriceps and gluteus maximus muscle forces during slow (0.75 m/s), 6° uphill walking compared to faster

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(1.50 m/s), level walking. Dorn et al. [13] performed predictive simulations on uphill walking at varied inclinations and observed muscle force patterns similar to level walking for smaller inclinations (5° and 10°). For the higher inclinations (15° and 20°), the peak forces developed by the hamstrings, vasti, and gluteus maximus increased, while iliopsoas, rectus femoris, and tibialis anterior did not change. Soleus and gastrocnemii muscle forces developed a double-peaked pattern for the higher inclines and the soleus peaks increased. The predictive simulations also revealed that joint kinematics and kinetics patterns at smaller inclinations were closer to level walking [13], which is not consistent with previous experiments [7]. Therefore, experimental analyses of muscle forces during uphill walking are necessary to evaluate the results of the predictive simulation. Furthermore, it appears that no study to date, has investigated the effect of downhill walking on lower limb muscle forces using a musculoskeletal model.

The purpose of this study was to analyze lower limb muscle forces spanning the hip, tibiofemoral, patellofemoral and ankle joint during sloped walking at various inclinations using a musculoskeletal model. It was hypothesized that sloped walking leads to altered lower limb muscle forces compared to level walking. Amongst others, downhill walking is expected to increase quadriceps and decrease gastrocnemii and hamstrings muscle forces while uphill walking would increase all of them in comparison to level walking.

2. Methods

2.1. Participants

Eighteen healthy male participants (age: 27.0 ± 4.7 y, height: 1.80 ± 0.05 m, mass: 74.5 ± 8.2 kg) were recruited. The study was approved by the institutional ethics board and written informed consent was signed by all participants.

2.2. Data collection

Participants walked at a pre-set speed of 1.1 m/s on a ramp ($6 \text{ m} \times 1.5 \text{ m}$) at different inclination angles of 0°, ±6°, ±12° and ±18° [9]. All participants wore the same type of standard indoor sport shoes (Puma SE, Germany). Speed was controlled via a timing device (Brower Timing Systems, Draper, Utah, USA). Participants walked about 5 min at each inclination for adjustment. Usually four steps were made before and after the measurement. Reflective markers 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 motion capture system (Vicon, Oxford, Oxford Metrics Ltd, UK; 250 Hz) and kinetic data were recorded with two force plates imbedded into the ramp (AMTI, Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA; 1000 Hz).

2.3. Data analysis

Kinematic data were further processed using Vicon Nexus (Vicon, Oxford Metrics Ltd, UK). The processed kinematic and kinetic data were imported into the AnyBody Modelling System (vers. 6.0, AnyBody Technology, Denmark) to calculate lower limb muscle forces. Kinematic and kinetic data were filtered using a Butterworth low pass filter with 10 and 15 Hz cut off frequencies, respectively. The musculoskeletal model used in this study was a standard model (AMMR 1.6.2, MoCapModel-LowerBody including the trunk). Each leg has seven degrees of freedom: three hip joint, one knee joint, one ankle joint, one subtalar joint and one patellar movement. Mass, moments of inertia, and muscle sites/geometry of all segments were modelled according to the morphological dataset for the lower extremities provided by Klein Horsman et al. [14]. Each leg contained 55 muscles and mechanical effects were modelled by 159 simple muscle slips [14]. The model was scaled to match each participant's anthropometry [15]. Inverse dynamics were performed and a third order polynomial muscle recruitment criterion, including a constraint preventing individual muscle forces from exceeding their physiological maximum, was used [16]. Comparing estimated muscles activity (model) with measured electromyography (EMG) data of five muscles (biceps femoris, rectus femoris, vastus lateralis, gastrocnemius lateralis and tibialis anterior), following was observed: correlation coefficients above 0.6 for all conditions except tibialis anterior at +18° and rectus femoris at −6°, 0° and +6° (~69% of the conditions >0.7), rather high mean absolute errors (only 38% below 30%, highest for tibialis anterior) and trend analysis revealing similar activities for all muscles and tasks (except tibialis anterior during uphill walking). Since EMG measurements have several limitations itself (e.g. muscle crosstalk and no linear relationship between muscle forces and muscle activity, amongst others), the model was found to be suitable for sloped walking [17].

Following muscles were analyzed: gluteus maximus, medius and minimus, piriformis, adductors (sum of adductor longus, brevis and magnus, pectineus and gracilis), iliopsoas, rectus femoris, vastus medialis, lateralis and intermedius, biceps femoris, semitendinosus, semimembranosus, gastrocnemius medialis and lateralis, soleus, peroneus and tibialis anterior. Muscle forces were normalized to body weight (× BW). Mean muscle forces were calculated for each participant and average values were then calculated over all participants.

Furthermore, hip, tibiofemoral, patellofemoral and ankle compressive forces were calculated using the same model [9] to analyze possible muscle contributions to the respective joint loading. Therefore, muscle and joint forces of each trial were time-normalized to gait cycle duration and the ensemble means were calculated over all participants.

2.4. 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 in the data using the Shapiro–Wilk test were only partly achieved. Therefore, changes in mean muscle forces with respect to various inclinations were analyzed using the Friedman test. In cases of significance, Wilcoxon tests with Bonferroni correction ($\alpha = 0.00238$) were used for pairwise post hoc comparisons (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 ($d > 0.80$) [18] and are shown for significant values in results section.

3. Results

Participants had to walk at a pre-set speed of 1.1 m/s and the mean walking speed at all inclinations was 1.1 ± 0.01 m/s. The gait analysis revealed significant main effects of inclination ($p < 0.05$) on all mean lower limb muscle forces. In general the pairwise comparisons revealed that all lower limb muscle forces, except gluteus medius, were significantly altered ($p \leq 0.02$) during sloped compared to level walking (Table 1). During downhill walking, muscle forces of gluteus maximus ($d > 0.69$), quadriceps ($d > 1.17$), soleus ($d > 1.40$), peroneus ($d > 0.86$) and tibialis anterior ($d = 0.87$) were significantly increased ($p \leq 0.02$) compared to level walking, while gluteus minimus ($d > 0.96$), piriformis ($d > 0.73$), adductors ($d > 0.81$), iliopsoas ($d > 0.61$), hamstrings ($d > 1.12$) and gastrocnemii ($d > 1.20$) were significantly decreased ($p \leq 0.02$). Uphill walking significantly decreased ($p \leq 0.02$) gluteus minimus ($d > 0.83$), iliopsoas ($d > 0.66$) and tibialis anterior ($d > 0.84$) muscle