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Uphill walking: Biomechanical demand on the lower extremities of obese adolescents

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ABSTRACT

The number of obesity prevalence in adolescents is still increasing. Obesity treatment programs typically include physical activity with walking being recommended as appropriate activity, but limited information exists on the demand uphill walking places on the joint loading and power of obese adolescents. Therefore, the purpose of this study was to investigate the effect of different inclinations on step characteristics, sagittal and frontal joint angles, joint moments and joint power of obese adolescents in comparison to their normal-weight peers. Eleven obese (14.5 ± 1.41 years, BMI: 31.1 ± 3.5 kg/m^2) and eleven normal-weight adolescents (14.3 ± 1.86 years, BMI: 19.0 ± 1.7 kg/m^2) walked with 1.11 m/s on a ramp with two imbedded force plates (AMTI, 1000 Hz) at three inclinations (level, 6°, 12°). Kinematic data were collected via an infrared-camera motion system (Vicon, 250 Hz). The two-way (inclination, group) ANOVA indicated a significant effect of inclination on almost all variables analysed, with the hip joint being the most affected by inclination, followed by the knee and ankle joint. The obese participants additionally spent less time in swing phase, walked with an increased knee flexion and valgus angle and an increased peak hip flexion and adduction moment. Hip joint power of obese adolescents was especially in the steepest inclination significantly increased compared to their normal-weight peers. Obese adolescents demonstrate increased joint loading compared to their normal-weight peers and in combination with a musculoskeletal malalignment they might be prone to an increased overuse injury risk.

1. Introduction

In the European Union at least 3 million children are obese, increasing by more than 85,000 children each year [1]. In the USA one fifth of all 12- to 19-year-olds is obese [2]. Additionally, the number orthopedic disorders increases in obese pediatric patients, with increased body weight contributing to increased wear within the skeletal structure [e.g. 3,4–8]. As such, obese adolescent have reported pain in the lower back [5], hip, knee and ankle joint [3] and shown foot deformities [e.g. 7], valgus/varus deformities [5,8,9], genu recurvatum and tight quadriceps [5,9] as well as cartilage lesions [3,4]. These alterations might contribute to the higher occurrence of osteoarthritis in the adulthood of obese individuals compared to normal-weight controls [6][e.g. 6]. Together with dietary and psychological treatment typical physical activity is included in a regular obesity treatment program [e.g. 10]. Walking is often recommended as appropriate activity, due to its easy accessibility and the adequate metabolic cost at which the patients are exposed to low risk of musculoskeletal injury [11,12]. However, physical activity is often restricted in this population [12] due to existing conditions such as knee pain [3]. Understanding of movement strategies in obese children with respect to kinematics and joint loading would allow for an informed exercise program, but is currently limited [13]. Research of level walking for obese children/adolescents and adults has highlighted kinematic and kinetic alterations [6,14,15][e.g. 6,14,15], but the understanding of uphill walking in obese children is very limited. Obese children have demonstrated an increased genu valgum [6] and increased muscular contraction force to maintain normal gait function [15] in level gait. Also, altered sagittal and frontal plane movement patterns occur when climbing stairs [6,13][e.g. 6,13]. Hence, it is

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likely that alterations of kinematics and joint loading will occur for the more challenging task of uphill walking than in level walking. Since the implementation of a more active life style could also include outdoor walking and hiking activities, or training on a treadmill, where metabolic cost can effectively be modulated by the inclination [16,17], investigating uphill walking is of fundamental relevance.

Uphill walking however requires movement modifications compared to level gait: amongst others, swing limb trajectories must be modified to ensure safe toe clearance and foot placement as the inclination changes. As such, the limb needs to be raised higher for toe clearance and heel strike and the body needs to be lifted to overcome the ground elevation whilst ascending [18]. In a healthy population, uphill walking has been associated with an increase of lower extremity joint loading [18–20] and muscle forces [21] compared to level gait.

In obese adults, some evidence exists that uphill walking could be a useful tool to moderate joint loading. Haight et al. [17] demonstrated that slow uphill (6° incline) walking with the same metabolic cost as fast level walking reduces the early stance tibiofemoral loading of obese adolescents by 23%. In an earlier study, Ehlen et al. [22] reported reduced peak knee extensor net muscle moments and peak knee frontal plane moments in slow uphill walking versus faster level walking in an obese adolescent group. The dependency of walking speed and inclination was systematically investigated by Browning et al. [16], demonstrating that metabolic rate of obese adults can be modulated by either increasing walking speed (0.5–1.75 m/s), or/and walking inclination (−3° to 9°). It is not known, however, to which extent excess body mass influences kinematics and joint loading in obese adolescents compared to their non-obese peers when walking uphill at the same speed. It is hypothesized that, similar to level walking, obese children would choose smaller step lengths and demonstrate different sagittal and frontal plane joint moments and joint power, while walking uphill compared to their non-obese peers. Furthermore, it is hypothesized that the differences are more pronounced with increasing inclination.

2. Methods

2.1. Participants

Eleven obese adolescents (5 female, 6 male, age: 14.5 ± 1.41 y, height: 1.67 ± 0.07 m, mass: 87.24 ± 16.3 kg, BMI: 31.1 ± 3.5 kg/m^2) and a control group of eleven age and height matched normal-weight adolescents (5 female, 6 male, age: 14.3 ± 1.86 y, height: 1.66 ± 0.07 m, mass: 52.41 ± 6.3 kg, BMI: 19.0 ± 1.7 kg/m^2) were recruited. Obesity was defined as the age matched BMI being above the 97th percentile and the control group was defined by an age-matched BMI being between the 10th and 85th percentile. The study was approved by the ethics board of the University of Salzburg and written informed consent was signed by all participants and their legal guardian.

2.2. Data collection

After marker placement and a 10 min warm-up on a treadmill with self-selected speed, all participants walked with the same type of indoor sport shoes on an instrumented ramp (6.0 m × 1.4 m) (Fig. 1) at different uphill inclinations of 0° (level), 6° and 12° in a randomized order at a pre-set speed of 1.11 m/s (±2.5%) in all conditions, which was controlled via a timing device (Brower Timing Systems, Draper, Utah, USA) [19,21]. Reflective markers were attached to all participants by the same researcher according to the Cleveland Clinic Marker set (Motion Analysis Corp, Santa Rosa, USA), which is a cluster based maker placement with an

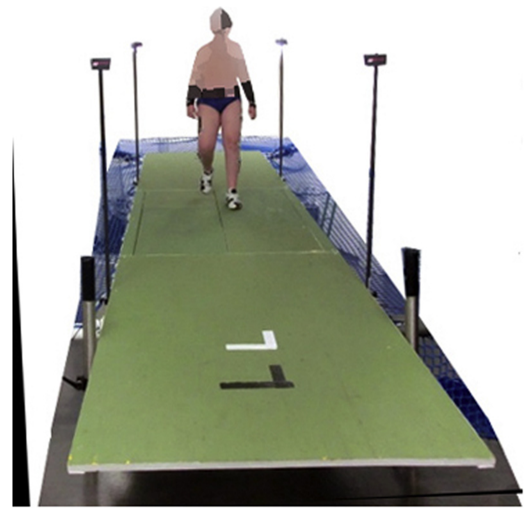

Fig. 1. Measurment set-up with the custom built ramp.

global optimization approach [23]. In case of soft tissue overlapping the left and right anterior superior iliac spina (ASIS) landmark in the obese group the respective markers were placed more laterally and the ASIS distance was measured [13]. Kinematic data were collected with an eight-camera, marker based motion capture system (Vicon, Oxford, Oxford Metrics Ltd, UK; 250 Hz) and kinetic data were recorded with two force plates embedded into the ramp (AMTI, Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA; 1000 Hz), allowing analysis of the gait cycle following the first force plate contact of the right limb. At each inclination approximately five minutes of familiarization with the new inclination was given. Participants were instructed to walk in a straight line with a freely chosen walking speed. In case the walking speed was not within the range (1.08–1.14 m/s) with six trials, feedback was given to the participant to either walk slower or faster. Three valid trials, where participants walked within the given speed-range and hit the force plates without any visual interruption of the gait cycle were collected. Rest periods were given between inclinations to avoid fatigue of the participants. If the speed requirements and valid foot contacts were not met within twelve trials, a valid trial with the walking speed closest to the acceptable speed range was chosen for further analysis.

2.3. Data analysis

Out of three trials, the trial with the walking velocity closest to the pre-set speed was taken for further analysis [21]. Kinematic data were labeled using Vicon Nexus (Vicon, Oxford Metrics Ltd, UK) and further processed using Visual3D (V3D; Cmotion, Rockville, MD, USA). Kinematic and kinetic data were filtered using a Butterworth low pass filter with 10 and 15 Hz cut off frequencies, respectively. The hip joint center was determined via functional calibration [24]. Spatio-temporal parameters and right sagittal and frontal ankle, knee and hip joint angles, internal moments and sagittal plane powers were calculated via the six degree of freedom model in Visual 3D [25]. Kinetic data were normalized to body mass, all data were plotted against stance phase and peak values were identified. Due to the high possibility of measurement errors in the transversal plane for obese participants this plane was omitted for analysis.

2.4. Statistics

Statistical analysis was conducted using SPSS (version 22.0, IBM, Armonk, NY, USA). The significance level was set to $\alpha = 0.05$.