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Effects of obesity on lower extremity muscle function during walking at two speeds

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ABSTRACT

Walking is a recommended form of physical activity for obese adults, yet the effects of obesity and walking speed on the biomechanics of walking are not well understood. The purpose of this study was to examine joint kinematics, muscle force requirements and individual muscle contributions to the walking ground reaction forces (GRFs) at two speeds (1.25 m s^{-1} and 1.50 m s^{-1}) in obese and nonobese adults. Vasti (VAS), gluteus medius (GMED), gastrocnemius (GAST), and soleus (SOL) forces and their contributions to the GRFs were estimated using three-dimensional musculoskeletal models scaled to the anthropometrics of nine obese (35.0 (3.78 kg m^{-2})); body mass index mean (SD)) and 10 nonobese (22.1 (1.02 kg m^{-2})) subjects. The obese individuals walked with a straighter knee in early stance at the faster speed and greater pelvic obliquity during single limb support at both speeds. Absolute force requirements were generally greater in obese vs. nonobese adults, the main exception being VAS, which was similar between groups. At both speeds, lean mass (LM) normalized force output for GMED was greater in the obese group. Obese individuals appear to adopt a gait pattern that reduces VAS force output, especially at speeds greater than their preferred walking velocity. Greater relative GMED force requirements in obese individuals may contribute to altered kinematics and increased risk of musculoskeletal injury/pathology. Our results suggest that obese individuals may have relative weakness of the VAS and hip abductor muscles, specifically GMED, which may act to increase their risk of musculoskeletal injury/pathology during walking, and therefore may benefit from targeted muscle strengthening.

1. Introduction

Obesity is a worldwide public health concern and obese adults and children are advised to engage in daily physical activity. Walking is a recommended form of physical activity for obese adults because it is convenient and suitable to elicit a moderate-vigorous metabolic response [1]. However, obese individuals have lower relative muscle strength compared to nonobese individuals [2]. Weakness and susceptibility to fatigue of certain key muscles (e.g. vasti (VAS) and gluteus medius (GMED)) can result in an abnormal gait pattern due to their critical role in locomotor tasks [3], predisposing individuals to musculoskeletal injury or pathology (e.g. large joint osteoarthritis (OA) and low back pain) [4,5]. In addition, muscle force requirements increase with walking speed [6], so at the faster walking speeds used during exercise, certain muscles, including those responsible for forward progression (e.g. the gastrocnemius (GAST) and soleus (SOL)), may be unable to effectively perform their respective functions, resulting in gait deviations that may increase the risk of musculoskeletal injury/pathology.

Surprisingly, the degree to which obesity affects gait kinematics and kinetics is not clear. Some studies report that kinematics are similar in obese and nonobese groups [7,8], while others report that obese individuals walk with a more extended leg and similar knee extensor moments during stance and greater step width compared to their nonobese counterparts [9], particularly at faster walking speeds. Unfortunately, there is limited information regarding how investigators did or did not account for the peripheral adiposity that obscures the motion of the underlying skeleton. Thus, differences in methodology may explain these equivocal kinematic results. In addition, studies that have reported lower extremity gait biomechanics in obese individuals [8,9] have not provided a quantitative assessment of individual muscle function, which may help explain the observed gait patterns.

Musculoskeletal simulations can provide us with an improved understanding of the force requirements and roles that individual

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muscles play during locomotor tasks [10]. Recent studies have estimated the contributions of individual muscles to the ground reaction force (GRF) during walking in nonobese adults [11,12]. These studies have shown that during early stance, VAS and GMED muscles are significant contributors to the vertical GRF (GRF_V), and function to decelerate and support the body, while during mid-late stance, the gastrocnemius (GAST) and soleus (SOL) are the primary contributors to the GRF_V and the anterior-posterior GRF (GRF_{AP}). In the frontal plane, GMED acts to maintain mediolateral (ML) stability and balance, and has been shown to be the primary contributor to the ML GRF (GRF_{ML}) [13]. Unlike in the sagittal plane, where a more aligned skeleton would reduce knee extensor muscle requirements, support and stability of the body in the frontal plane is largely accomplished by the hip abductor muscles (e.g. GMED). The effect of GMED weakness may be altered frontal plane kinematics of the pelvis (e.g. increased pelvic obliquity, an increase in pelvic drop of the contralateral hip) resulting in pathological hip joint articulation [14]. For this study, we focused our investigation on the muscles that have large contributions to all three components of the GRF (VAS, GMED, GAST, and SOL) [12].

The purpose of this study was to quantify joint kinematics, estimate individual muscle forces (VAS, GMED, GAST, SOL), and the individual muscle contributions to the walking GRFs at two speeds (1.25 and 1.50 m s^{-1}) in obese and nonobese adults. It has been reported that obese adults walking with a more erect posture and similar knee extensor joint torques compared to nonobese adults [9], suggesting reduced knee extensor muscle forces. We hypothesized that (1) peak knee flexion during stance would be less, while pelvis obliquity would be greater in the obese vs. nonobese group, and the differences between the obese and nonobese groups would be greater at the faster walking speed; (2) absolute and lean mass normalized forces for all muscles, except VAS, which we predict to be similar, would be greater in the obese vs. nonobese adults at both speeds; and (3) VAS contribution to the GRF_V would be similar between the obese and nonobese individuals at a velocity of 1.25 m s^{-1} but would be reduced at a velocity of 1.50 m s^{-1} in the obese group.

2. Methods

2.1. Subjects

A convenience sample of nine obese (8 female) adults and 10 nonobese adults (5 female) participated in our study. Inclusion criteria included a BMI of <25 kg m^{-2} (nonobese) and 30–40 kg m^{-2} (obese), age 18–45, and sedentary to moderately active (< 2–3 bouts of exercise/week or participation in any sporting activities < 3 h/week), while exclusion criteria included orthopedic, metabolic, or neurologic impairments, other than obesity, that would hinder movement and prevent safe participation in the study. Subject characteristics and anthropometrics are presented in Table 1. All subjects gave written informed consent approved by Colorado State University's Human Research Institutional Review Board.

Table 1
Physical characteristics of obese and nonobese participants.

Participant characteristics	Obese	Nonobese
Body mass (kg)	96.8 (11.5)	63.7 (4.47)
Lean mass (kg)	51.4 (8.55)	46.9 (6.83)
Height (m)	1.66 (0.069)	1.69 (0.051)
BMI (kg m^{-2})	35.0 (3.78)	22.1 (1.02)
Age (years)	35 (7.6)	26 (6.0)

Values are mean (SD).

2.2. Experimental protocol

We quantified body mass composition for each subject via dual x-ray absorptiometry (DEXA, Hologic Discover, Bedford, MA). As part of a larger study, participants walked at nine randomized speed grade combinations (speeds: 0.50–1.75 m s^{-1}, grades: 0°–9°). Trials lasted 6 min with 5 min of rest between trials. During an acclimatization period, before the first trial, subjects walked on the treadmill at a self-selected speed. The acclimation period ended when participants had walked for at least 5 min and were observed to have a normal gait pattern (all participants walked 10 min or less during the acclimation period). Here, we are reporting only the results from two level (0° grade) trials: 1.25 and 1.50 m s^{-1}. The 1.25 m s^{-1} walking speed was selected as it is very near the self-selected speed for obese adults reported by DeVita and Hortobagyi (1.29 m s^{-1}) [9], while a walking speed of 1.5 m s^{-1} was selected because it is considered an appropriate exercise walking speed for obese adults to meet physical activity guidelines and achieve proper physiological benefits [15].

2.3. Experimental data

Whole body kinematics and kinetics were collected using a 10-camera motion capture system (Nexus, Vicon, Centennial, CO) recording at 100 Hz and a dual-belt, force measuring treadmill (Fully Instrumented Treadmill; Bertec Corp., Columbus, OH) recording at 1000 Hz. We used an obesity-specific marker set methodology, which was utilized to attenuate the effects of subcutaneous adiposity obscuring the motion of the underlying skeleton, particularly the anterior pelvis. Physical reflective markers were placed over the following anatomical landmarks: 7th cervical vertebrae, acromion processes, right scapular inferior angle, sterno-clavicular notch, xiphoid process, 10th thoracic vertebrae, and bilaterally over posterior-superior iliac spines, medial and lateral epicondyles of the femurs, medial and lateral malleoli, calcanei, first metatarsal heads, second metatarsal heads, and proximal and distal heads of the 5th metatarsals. Marker clusters (four non-collinear markers affixed to a rigid plate) were adhered to the sacrum, and bilaterally to the thighs and shanks to aid in three-dimensional tracking. We also digitally marked the anterior superior iliac spines (ASIS) and iliac crests using a digitizing pointer (C-Motion, Germantown, MD). Borhani et al. showed improved repeatability and good reliability in tracking the movement of the pelvis with a cluster placed on the sacrum (similar design and placement as in our study) as compared to the "traditional' method of tracking via anterior and posterior ASIS markers in nonobese, overweight, and obese individuals [16]. Electromyographic (EMG) data (Noraxon, Scottsdale, AZ) from bipolar surface electrodes recording at 1000 Hz was collected for the soleus, lateral gastrocnemius, vastus lateralis, vastus medialis, biceps femoris long head, and semimembranosus muscles using International Society for Electrophysiology and Kinesiology standard procedures [17]. The EMG signal was band-pass filtered (16–380 Hz), fully rectified and finally low-pass filtered at 7 Hz. All biomechanics data was collected during the final 30 s of each trial. Marker trajectory and GRF data were digitally low-pass filtered at 5 and 12 Hz, respectively, using fourth-order zero-lag Butterworth filters.

2.4. Musculoskeletal modeling

We scaled a generic OpenSim musculoskeletal model for each subject to account for individual anthropometrics. The mass and inertial properties of each body segment were scaled as a function of segment length, determine by anatomical landmarks, and total body mass. The model was comprised of 12 body segments with