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Barefoot vs. Shod: Effects of Trunk Loading and Body Mass Index on Walking Mechanics

Kevin D. Dames

UNIVERSITY OF NORTHERN COLORADO

Greeley, Colorado

The Graduate School

BAREFOOT VS. SHOD: EFFECTS OF TRUNK LOADING AND BODY MASS INDEX ON WALKING MECHANICS

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Kevin D. Dames

College of Natural and Health Sciences
School of Sport and Exercise Science
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has been approved as meeting the requirement for the Degree of Doctor of Philosophy in College of Natural and Health Sciences in School of Sports and Exercise Science, Program of Exercise Science

Accepted by the Doctoral Committee

____
Gary D. Heise Ph.D., Co-Research Advisor

____
Jeremy D. Smith Ph.D., Co-Research Advisor

____
David S. Hydock Ph.D., Committee Member

____
Kimberly A. Murza Ph.D., Faculty Representative

Date of Dissertation Defense ____________________________________

Accepted by the Graduate School

____

Linda L. Black, Ed.D.
Associate Provost and Dean
Graduate School and International Admissions

ABSTRACT

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In this dissertation, the impacts of increased mass and footwear on walking mechanics and energetics were investigated. In the first study, non-obese individuals were asked to walk on a treadmill with added load to the trunk (~15% of body mass) and with and without shoes. Metabolic costs of walking increased ~12% with added load, but walking barefoot did not significantly change metabolic costs. Trunk loading increased knee and hip range of motion but failed to alter spatiotemporal measures. In study 2, non-obese individuals were asked to complete the same tasks, but this time they walked overground instead of on a treadmill. The focus of this study was on lower extremity kinetics, which were not addressed in the first study. Loading increased stance and double support times, ground reaction forces, and joint moments and powers. Walking barefoot decreased spatiotemporal measures and ground reaction forces, but increased hip and knee moments and powers. Finally, in study three, rather than increasing body mass artificially by adding an external mass to the trunk, obese individuals with BMIs greater than 30 $kg \cdot m^{-2}$, but less than 40 $kg \cdot m^{-2}$, were recruited. Similar to Study 2, walking barefoot reduced stride length, stance time, and double support time. Barefoot walking also decreased vertical and anteroposterior ground reaction forces. However, joint moment and power responses to footwear conditions were dependent on body

morphology, as the Obese and Non-Obese groups responded differently to these footwear conditions. Therefore, footwear condition should be reported and considered when comparing conclusions of multiple studies. Statistical outcomes for kinetic dependent measures also differed with normalization. Four joint kinetic measures (including ankle dorsiflexor and hip extensor moments, and knee and hip powers), were larger in the Non-Obese group than the Obese group after normalization, but did not differ when considered in absolute units. On the other hand, ten joint kinetic measures, including ankle, knee, and hip joint moments and powers, were larger in the Obese group in absolute terms. All ten of these were not different from the Non-Obese group after normalization. Varying normalization schemes partially explains differing outcomes reported in the literature regarding obesity's impact on gait mechanics. Based on outcomes of the three studies presented here, ground reaction forces appear to scale with total weight, whether this is an external load (Study 2) or a consequence of obesity (Study 3). Walking barefoot decreased stride length, stance time, and double support time and ground reaction forces regardless of loading or obesity. However, joint kinetic responses to footwear appear to be dependent on body morphology, as the Obese and Non-Obese groups responded differently to these conditions.

ACKNOWLEDGEMENTS

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CHAPTER I

GENERAL INTRODUCTION

Obesity has become a major health issue in the United States. Data from recent national surveys report that 72.5 million American adults are obese (Hootman, Helmick, Hannan, & Pan, 2011). Other data show 33.9% and 35.1% of American adults are overweight and obese, respectively (Fryar, Carroll, & Ogden, 2014). The physiological and biomechanical complications associated with obesity generate substantial economic costs. Approximately $147 billion per year is spent in obesity-related health care costs (Hootman et al., 2011). Some musculoskeletal issues associated with obesity include knee and hip joint replacements, general pain of the low back and neck (Patterson, Frank, Kristal, & White, 2004) and arthritis (Hootman et al., 2011). Annual arthritis related health care costs are estimated at $128 billion (Hootman et al., 2011). The rate of obesity has been climbing over recent years (Fryar et al., 2014; Hootman et al., 2011), which will lead to even greater health care costs in the future. Excessive joint loads, such as those experienced during locomotion, are suspected to contribute to the greater prevalence of osteoarthritis in obese individuals (Hootman et al., 2011). While walking and running have positive effects on managing weight, these activities expose the overweight individual's body to the very impacts that they may be recommended to avoid.

Load carriage is a common task that, similar to obesity, increases the mass an individual must transport during locomotion. Unlike obesity, this extra mass is external to the body and often concentrated in a particular position (e.g., a backpack, single-strap satchel, or handheld load such as a grocery bag). The ability to carry heavy loads safely is an important task in many vocations. Research has focused on load carriage in diverse groups such as firefighters (Park, Hur, Rosengren, Horn, & Hsiao-Wecksler, 2010), military personnel (Knapik, Reynolds, & Harman, 2004; Majumdar, Pal, & Majumdar, 2010), hikers (Simpson, Munro, & Steele, 2012) and college students (Devroey, Jonkers, de Becker, Lenaerts, & Spaepen, 2007; Heuscher, Gilkey, Peel, & Kennedy, 2010). Previous work targeting students has focused on the relationship between frequency of backpack use, backpack weight, and back pain (Heuscher et al., 2010), as well as backpack design (Palmer, Bauer, Bowman, & Magleby, 2011). Even in normal, healthy individuals, walking with a heavy backpack presents physiological (Blacker, Fallowfield, Bilzon, & Willems, 2009; Quesada, Mengelkoch, Hale, & Simon, 2000) and biomechanical challenges (Quesada et al., 2000; H. Wang, Frame, Ozimek, Leib, & Dugan, 2013).

Footwear determines, in part, how an individual interacts with the environment during locomotion. The fit, thickness of cushion, style (e.g., sandal or laced shoe) and material influence comfort and ease of walking. Significant amounts of money are spent developing the perfect shoe for each sport, activity, and lifestyle. Footwear has been the focus of several health related research lines such as arch development in youths (Rao & Joseph, 1992), joint health in those afflicted with arthritis (Shakoor & Block, 2006), and plantar surface tissue health in older adults (Burnfield, Few, Mohamed, & Perry, 2004) and peripheral neuropathy in diabetics (Sarnow et al., 1994). Understanding the mechanical and physiological responses of healthy individuals to barefoot walking will help provide insights into the role of footwear in clinical populations.

Walking mechanics of those carrying loads and those who are obese are similar in many areas. Obesity and adding mass to non-obese individuals influences the location of the body's center of mass (Matrangola, Madigan, Nussbaum, Ross, & Davy, 2008; Smith et al., 2006). Artificially increasing body mass increases metabolic (Browning, Baker, Herron, & Kram, 2006) and mechanical (H. Wang et al., 2013) effort required during walking. Additionally, the most economical speed of walking (J/kg/m) is slower when body mass is increased (Browning & Kram, 2005).

Differences in metabolic cost due to increased mass can be attributed, in part, to differences in mechanics. For example, generating force to support body weight and accelerating the body's mass account for ~28% and ~45%, respectively, of the total metabolic cost of walking (Grabowski, Farley, & Kram, 2005). When body mass is increased, ground reaction forces (GRF) are also increased. Obesity increases vertical, anteroposterior, and medio-lateral GRFs compared to normal weight individuals (Browning & Kram, 2007). Joint kinetic measures, such as moments and powers, are also increased in obesity and load carriage even when these measures are normalized to total body mass. Some data show increased ankle joint plantarflexor moments, work, and power with obesity (DeVita & Hortobagyi, 2003). Others have reported larger sagittal plane hip and knee moments as consequences of obesity (Browning & Kram, 2007). In obese children, increases in knee abduction moments (Gushue, Houck, & Lerner, 2005) and hip abduction moments (McMillan, Auman, Collier, & Williams, 2009) have been reported. It has been suggested that increased frontal plane knee moment magnitudes are related to the severity of knee osteoarthritis (Sharma et al., 1998).

Similar lower extremity kinetic responses are available from the load carriage literature. With backpack loads, GRFs increase (Birrell & Haslam, 2010; Birrell, Hooper, & Haslam, 2007; H. Wang, Frame, Ozimek, Leib, & Dugan, 2012). Some suggest the vertical and anteroposterior GRF increases are proportional to the added mass (Birrell et al., 2007; Tilbury-Davis & Hooper, 1999; Y. Wang, Pascoe, & Weimar, 2001). Related to these forces, are increased pressures under the foot (Pau, Mandaresu, Leban, & Nussbaum, 2015). These extra lower limb joint loads may be a precursor to arthritis development (Sharma et al., 1998).

Similar spatiotemporal adjustments to load carriage have been observed between obese and non-obese individuals. These changes include shorter stride lengths (Blacker et al., 2009; LaFiandra, Wagenaar, Holt, & Obusek, 2003; Martin & Nelson, 1986), increased double support time (Browning & Kram, 2007; Kellis & Arampatzi, 2009; Martin & Nelson, 1986; Ranavolo et al., 2013), and increased stance time (Browning & Kram, 2007; Ranavolo et al., 2013). A longer double support time reduces the unique contributions of a single limb to body weight support (Ranavolo et al., 2013).

Many of these same spatiotemporal parameters are also impacted by footwear. Several positive adaptations that reduce GRFs and joint loads occur with barefoot walking. Reducing stride length (Keenan, Franz, Dicharry, Della Croce, & Kerrigan, 2011; Lythgo, Wilson, & Galea, 2009; Majumdar et al., 2006; Oeffinger et al., 1999; van Engelen et al., 2010; Wolf et al., 2008), single support time (Lythgo et al., 2009; Majumdar et al., 2006), and increasing double support time (Majumdar et al., 2006) are adaptations to walking without a protective shoe. These changes are likely related to the discomfort at initial contact, as plantar pressures are greater while barefoot, compared to shod (Burnfield et al., 2004). However, these increased pressures are related to the overall decrease in contact area, as the peak braking and initial vertical GRFs decrease while barefoot (Keenan et al., 2011). The second vertical GRF peak during toe-off may also decrease (Tilbury-Davis & Hooper, 1999). Perhaps more important than these changes for groups such as overweight individuals are the joint moment differences between barefoot and shod walking. A smaller knee varus moment and hip flexor moment during weight acceptance, and smaller ankle eversion moments at toe-off have been observed without shoes (Keenan et al., 2011). Other data show smaller knee flexor moments during weight acceptance and smaller plantarflexor moments at toe-off (Oeffinger et al., 1999). Given these noted reductions in hip, knee, and ankle joint moments, there may be benefits for overweight individuals to walk barefoot. Specifically, walking barefoot may decrease lower limb joint loads before any weight loss is achieved.

Understanding the relationships between load carriage and obesity under various footwear conditions can provide a better understanding of function. For example, walking barefoot may promote spatiotemporal patterns that reduce knee joint loads in overweight individuals. To address these issues, this dissertation will include three studies. The hypotheses for these projects are stated below.

Hypotheses

Study One Hypotheses – Load Carriage Economy

H1 Loading, regardless of footwear, will elicit shorter stride lengths, longer stance times, longer double support times and increased metabolic costs.

H2 Walking barefoot, regardless of load, will elicit longer stride lengths, shorter stance times, shorter double support times and reduced metabolic costs.

Therefore, it was expected that adding a backpack load to individuals walking barefoot would result in spatiotemporal patterns and metabolic costs similar to those of shod unloaded walking.

Study Two Hypotheses – Load Carriage Kinetics

H1 When footwear and loading changes were expected to occur in the same direction, their effects will be additive. For example, it is expected that the shortened stride length while barefoot will be even shorter while also loaded.

H2 When footwear and loading changes were expected to occur in opposite directions, their effects will cancel. That is, a dependent variable expected to decrease while barefoot and increase with load will result in a value similar to the shod, unloaded condition.

Study Three Hypotheses – Obesity

H1 Overweight individuals, regardless of footwear, will experience larger peak ground reaction forces, joint moments, and joint powers.

H2 While barefoot, regardless of body weight, barefoot walking will produce lower peak ground reaction forces, joint moments, and joint powers.

Overall Purpose of the Study

The purpose of this dissertation was to investigate the impact of footwear and increased mass on walking mechanics and energetics.

Methodology

Study 1 Methodology

Participants. Twelve individuals (7 female, 5 male) participated in this study (age = 24 ± 2 years, height = 1.73 ± 0.13 m, and mass = 71.1 ± 16.9 kg). All participants were healthy, recreationally active and free of any notable gait abnormalities. The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

Experimental protocol. Anthropometric data (including body mass and height) were collected based on VICON's full body plug-in-gait model with medial markers on the knee and ankle to better identify knee and ankle axes (Wong, Callewaert, Labey, Leardini, & Desloovere, 2009). Reflective markers were placed on various anatomical locations using double-sided tape based on the plug-in-gait model. Participants then walked on a level treadmill (Woodway, Waukesha, WI) at 1.5 $m \cdot s^{-1}$ for 6-min under four conditions: Barefoot Unloaded (BU), Shod Unloaded (SU), Barefoot Loaded (BL), and Shod Loaded (SL). This model of treadmill was selected because its rubberized slats allowed steady state barefoot walking to be accomplished without blister formation or undue discomfort. A moderately higher walking speed than previously used (Keenan et al., 2011) was selected in an effort to increase the demands on the system so that alterations in movement patterns would be more apparent. A backpack equal to 15% of the participant's body mass was worn during the two loaded conditions. A single textbook was placed in the pack against the participant's back to provide a solid, flat surface before adding lead weights until the desired mass of the backpack was achieved. Participants performed the shod conditions using their own athletic shoe (mean shoe mass

= 272 ± 68 g). The order of conditions was individually randomized and a brief rest was provided between successive walking bouts. The rest period was based on the time it took to change from one condition to the next and only lasted a couple of minutes. Randomization of all conditions across all participants was used in attempt to minimize any fatigue effects in this study. During all walking trials, metabolic (ParvoMedics, Sandy, UT) and motion (100 Hz) (VICON, Englewood, CO) data were collected. For metabolic data collection, expired gasses were passed into the gas analyzer via a hose and mouthpiece. A nose plug was worn to force all expired gasses to enter the mouthpiece. Motion data were collected during the last two minutes of each walking trial, which is where steady-state metabolic responses also occurred.

Data analysis. Mean rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) over the last 2-min of each 6-min trial (van Engelen et al., 2010; Warne & Warrington, 2014) were used to estimate average rate of energy consumption (Weir, 1949):

$$\dot{E} = (3.9)\dot{V}O_2 + (1.1)\dot{V}CO_2 \tag{1}$$

where $\dot{E}$ is energy cost in kcal/min, and $\dot{V}O_2$ and $\dot{V}CO_2$ in L/min. $\dot{E}$ was converted to units of J/s and normalized to body mass. Metabolic cost was not normalized to any additional mass added to the body. We felt not accounting for the additional passive mass reflected best the real world metabolic consequences of walking with additional mass.

For spatiotemporal and kinematic measures, marker data were processed using VICON Nexus. Marker coordinate data were filtered using a 4th Order, recursive digital Butterworth filter with a cut-off frequency of 6 Hz. Joint kinematics were determined using the built-in plug-in-gait model in VICON Nexus. Velocities were derived using finite difference approximations.

Foot contact events (i.e., heel strike and toe-off) for each leg were visually identified during post-processing by a single researcher. This researcher identified heel strike as the first frame in which the heel marker stopped moving downwards. Toe off was identified as the first frame in which the toe marker began moving upwards. The foot contact events were then used to determine spatiotemporal measures during the trial, which included stance time, double support time, and swing time. Stride time was determined as the sum of stance and swing times for a given leg. Stride length was determined based on the walking velocity relationship:

$$SL = V*ST \qquad (2)$$

where SL represents stride length in m, V represents the walking velocity ($1.5\ m \cdot s^{-1}$), and ST represents stride time in seconds.

Kinematic dependent measures included ankle, knee, and hip ranges of motion and peak sagittal plane angular velocities. Range of motion (ROM) for each lower limb joint was determined by:

$$ROM = \text{max flexion angle} - \text{max extension angle} \qquad (3)$$

Statistical analysis. Means for spatiotemporal and kinematic data were determined from three consecutive strides during the final two minutes of each trial. The average metabolic costs during the final two minutes of each of the four trials were compared. A series of 2x2 (loading, footwear) ANOVAs with repeated measures were performed using SPSS (version 20). Alpha was set at .05 for all tests. A Bonferroni post hoc test was performed where pairwise comparisons were appropriate.

Study 2 Methodology

Participants. Twelve young, healthy individuals (5 women, 7 men) with no known musculoskeletal or neurological issues that would compromise gait were recruited for this study (age = 23 ± 3 years, height = 1.73 ± 0.11 m, and mass = 70.90 ± 12.67 kg). The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

Data collection. Participants wore tight-fitting clothing throughout the experiment so that anatomical landmarks could be easily identified and to minimize marker movements. Anthropometric data (i.e., body mass, height and various segment widths and lengths) were measured based on VICON's Plug-in-Gait model. Retroreflective markers were attached various anatomical landmarks using double-sided tape. Participants then performed overground walking trials at 1.5 $m \cdot s^{-1}$, which is slightly faster than the preferred speed of young adults (Norris, Granata, Mitros, Byrne, & Marsh, 2007), in four walking conditions: Barefoot Unloaded (BU), Shod Unloaded (SU), Barefoot Loaded (BL), and Shod Loaded (SL). Two pairs of timing gates (BROWER Timing Systems, Draper, UT) were used to ensure walking speed was within ±5% of the target speed. A backpack loaded with lead weights equal to 15% of the participant's body mass was worn during the loaded conditions. Participants wore their own athletic shoe for the shod conditions (272 ± 68 g). The order of these four conditions was individually randomized. During each trial, 3D motion (100 Hz) (VICON, Englewood, CO) and ground reaction force (GRF) (2000 Hz) data were collected. GRFs were measured by a tandem-belt instrumented treadmill (AMTI, Watertown, MA) embedded in the center of the walkway. Trials included in the data analysis were within the expected velocity range and clean foot contacts were made with the force plates (i.e., a single, whole foot contact on each force plate).

Data analysis. Markers were labeled within VICON Nexus, but all subsequent processing of data was performed using a custom Visual 3D (C-Motion, Germantown, MD) script. Marker data were filtered using a recursive, Butterworth lowpass filter (F_c = 6 Hz). This cutoff frequency for filtering marker data was confirmed with a residual analysis performed in MATLAB (MathWorks, Natick, MA) as described by (Winter, 2009). GRF data were filtered using a recursive, Butterworth lowpass filter (F_c = 50 Hz). Motion and GRF data were combined through inverse dynamics to estimate joint reaction forces and moments for the ankle, knee, and hip in the sagittal plane. Joint powers were calculated as the product of the joint moment and angular velocity. Joint power peaks consistent with those selected by Winter (1987) were used in statistical analyses. Joint power peaks from the phases defined by Winter (1987) were used in statistical analyses. These include two ankle phases (A1, A2), four knee phases (K1-K4), and three hip phases (H1-H3). A1 is the initial weight acceptance and A2 the propulsive peak at toe-off. K1 is the energy absorption phase during weight acceptance. K2 is the only power generation phase and occurs during mid-stance. K3 is power absorption during terminal stance and early swing phase. The K4 phase is terminal swing power absorption. H1 phase occurs during early stance, H2 is an absorption phase during mid-stance, and H3 a power generation phase prior to toe-off. All joint moment and power data were normalized to body mass.

Statistical analysis. Dependent variables were determined from three successful strides and then averaged. A series of 2 x 2 (loading, footwear) ANOVAs with repeated measures were performed using IBM SPSS 23.0 (Armonk, NY). Alpha was set at .05 for all tests. A Bonferroni post hoc test was performed where pairwise comparisons were appropriate.

Study 3 Methodology

Participants. Twelve young, healthy-weight individuals (5 women, 7 men) with no known musculoskeletal or neurological issues were recruited as controls for this study. Ten obese individuals who had a body mass index (BMI) between 30 and 40 $kg \cdot m^{-2}$ were also recruited for this study. Besides being obese, this group was otherwise healthy. Participant characteristics can be found in Table 4.1. There was no difference in height or age of these two groups. To differentiate between those who were truly obese vs. those whose muscular build may lead to large BMIs, a waist circumference >100 cm for men, or >90 for women, provided a secondary means for placement in the Obese group. These circumference criteria place an individual in the High Risk for disease category, based on published guidelines from the American College of Sports Medicine (*ACSM's Guidelines for Exercise Testing and Prescription*, 2010). The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

Data collection. Anthropometric data (including body mass and height) were measured for use in VICON's full body plug-in-gait model. Retroreflective markers were attached to the appropriate anatomical landmarks for the same model using double-sided tape. Participants then performed overground walking trials at 1.5 $m \cdot s^{-1}$ ±5% while barefoot and shod. Two pairs of timing gates (BROWER Timing Systems, Draper, UT) spaced approximately 5 m apart were used to capture walking speed. Participants wore their own athletic shoes for the shod conditions (Non-Obese group = 272 ± 68 g; Obese group = 321 ± 90 g). The shoe mass was not significantly different between groups. The order of conditions was individually randomized. During each trial, 3D motion (100 Hz) (VICON, Englewood, CO) and ground reaction force (GRF) (2000 Hz) data were collected. GRFs were measured using a tandem-belt instrumented treadmill (AMTI, Watertown, MA) embedded in the center of the walkway with 2 individual force plates. Trials included in the data analysis were within the expected velocity range and clean foot contacts were made with the force plates (i.e., a single, whole foot contact on each force plate).

Data analysis. Markers were labeled with VICON Nexus, but all subsequent processing of data was performed using a custom Visual 3D (C-Motion, Germantown, MD) script. Marker data were filtered using a recursive, digital Butterworth lowpass filter (f_c = 6 Hz). This cutoff frequency was confirmed by a residual analysis as described by Winter (2009). GRF data were filtered using a recursive, digital Butterworth lowpass filter (f_c = 50 Hz). Motion and GRF data were combined through inverse dynamics to estimate joint reaction forces and moments for the ankle, knee, and hip in the sagittal plane. Joint moment peaks were selected based on the description by Winter (1987). These included two ankle peaks, five knee peaks, and three hip peaks. The ankle moments include a dorsiflexor peak during weight acceptance and late stance plantarflexor peak. Knee moment peaks selected included the weight acceptance flexor, early stance extensor, midstance flexor, late stance extensor, and late swing flexor peaks. Hip moment peaks included the early stance extensor, late stance flexor, and late swing extensor peaks. Joint powers were calculated as the product of the joint moment and angular velocity. Joint power peaks from the phases defined by Winter (1987) were used in statistical analyses. These included two ankle phases (A1, A2), four knee phases (K1-K4), and three hip phases (H1-H3). A1 is the initial weight acceptance and A2 the propulsive peak at toe-off. K1 is the energy absorption phase during weight acceptance. K2 is the only power generation phase and occurs during mid-stance. K3 is power absorption during terminal stance and early swing phase. The K4 phase is terminal swing power absorption. H1 phase occurs during early stance, H2 is an absorption phase during mid-stance, and H3 a power generation phase prior to toe-off.

Spatiotemporal and kinematic dependent measures were also identified. Spatiotemporal dependent variables included stride length, stance time, swing time, and double support time. Kinematic variables included angular ranges of motion (ROM) and joint angles at initial contact for the hip, knee, and ankle.

Statistical analysis. Dependent variables were determined from three successful strides and then averaged. A series of 2 x 2 (BMI, footwear) ANOVAs with repeated measures were performed using IBM SPSS 23.0 (Armonk, NY). Grouping based on BMI represented a between subjects factor (Obese, Non-Obese) and footwear a within subjects factor (barefoot, shod). The probability associated with a Type I error was set at .05 for all tests. Where a significant difference was detected, a percent increase was presented as the larger mean minus the smaller mean divided by the smaller mean, and a percent decrease was presented as the smaller mean minus the larger mean divided by the larger mean.

CHAPTER II

REVIEW OF LITERATURE

Introduction

Obesity has become a major health issue in the United States. Data from recent national surveys report that 72.5 million American adults are obese (Hootman et al., 2011). In terms of percentages, other report data show 33.9% and 35.1% of American adults are overweight and obese, respectively (Fryar et al., 2014). The significant physiological and biomechanical complications associated with obesity generate substantial economic costs. Approximately $147 billion per year is spent in obesity related health care costs (Hootman et al., 2011). Some musculoskeletal issues associated with obesity include knee and hip joint replacements, general pain of the low back and neck (Patterson et al., 2004) and arthritis (Hootman et al., 2011). Given these issues, it is unfortunate that the rate of obesity has been climbing over recent years (Fryar et al., 2014; Hootman et al., 2011). Excessive joint loads, such as those experienced during locomotion, are suspected to contribute to the greater prevalence of osteoarthritis in obese individuals (Hootman et al., 2011). While walking and running have positive effects on managing weight, these activities expose the overweight individual's body to the very impacts that they may be recommended to avoid. Despite this, walking is a common mode of locomotion that is part of daily life.

Overweight is defined as a Body Mass Index (BMI) >25 kg/m^2, and obesity as a BMI >30 kg/m^2 (*ACSM's Guidelines for Exercise Testing and Prescription*, 2010). This measure is routinely used to categorize an individual's morphology in the clinical setting (Fryar et al., 2014). Unfortunately, the BMI only accounts for height and mass of an individual, rather than estimating percent contributions of lean and fat masses to total mass. Despite its shortcomings, its simplicity, convenience, and minimal cost to calculate make the BMI more frequently used than other methods of describing body morphology in the literature. Indeed, the relationships between BMI and certain conditions (e.g., arthritis, knee replacements, diabetes) have strong statistical support (Anandacoomarasamy, Caterson, Sambrook, Fransen, & March, 2008; Patterson et al., 2004). Prevalence of arthritis in American normal weight adults over 40 years of age is between 25.7 and 33%, while in obese adults the prevalence is between 37.0 and 44.4% (Ong, Wu, Cheung, Barter, & Rye, 2013).

Load carriage is a common task that, similar to obesity, increases the mass that an individual must transport during locomotion. Unlike obesity, this extra mass is external to the body and often concentrated in a particular position (e.g., a backpack, single-strap satchel, or handheld load such as a grocery bag). The ability to carry heavy loads safely is an important task in many vocations. Research has focused on load carriage in diverse groups such as firefighters (Park et al., 2010), military personnel (Knapik et al., 2004; Majumdar et al., 2010), hikers (Simpson et al., 2012) and students (Devroey et al., 2007; Heuscher et al., 2010). Previous work targeting students has focused on the relationship between frequency of backpack use, backpack weight, and back pain (Heuscher et al., 2010), as well as backpack design (Palmer et al., 2011). Even in normal, healthy individuals, walking with a heavy backpack presents physiological (Blacker et al., 2009;

Quesada et al., 2000) and biomechanical challenges (Quesada et al., 2000; H. Wang et al., 2013).

Maximizing comfort and performance while minimizing musculoskeletal trauma is a concern of load carriage literature. Previous research has attempted to determine the optimal walking speed for a given load magnitude based on metabolic cost (Pal, Majumdar, Bhattacharyya, Kumar, & Majumdar, 2009) and assessed injury risk and incidence (Goh, Thambyah, & Bose, 1998; Heuscher et al., 2010; Myung & Smith, 1997). Recommendations for limiting load magnitudes have been developed to promote ability to carry heavy loads over time (Simpson, Munro, & Steele, 2011).

Footwear determines, in part, how an individual interacts with the environment during locomotion. The fit, thickness of cushion, style (e.g., sandal or laced shoe) and material influence comfort and ease of walking. Indeed, much money is spent developing the perfect shoe for each sport, activity, and lifestyle. Footwear has been the focus of several health related research lines such as arch development in youths (Rao & Joseph, 1992), joint health in those afflicted with arthritis (Shakoor & Block, 2006), and plantar surface tissue health in older adults (Burnfield et al., 2004) and diabetics (Sarnow et al., 1994). Understanding the response of normal, healthy individuals to barefoot walking is essential to determine what role footwear plays in these clinical populations.

There are many similarities between gait of overweight individuals and healthy individuals carrying loads. It is of interest to understand how footwear interacts with the walking performance of two groups, and how these groups compare given novel footwear conditions.

Gait

Walking gait has been studied for many years and in many different clinical populations to understand how performances of these individuals compare to those of normal, healthy individuals. This simple task of moving one foot in front of the other in cyclical fashion comprises many complex tasks for successful performance. When analyzing gait, several phases and events have been identified as having functional importance (J. Perry, 1992; Winter, 1987). The role of lower limb joints and individual muscle groups have been investigated to determine what a “normal” pattern looks like (J. Perry, 1992). However, “normal” patterns change depending on what conditions the individual experiences at the time of observation.

Footwear, load carriage, and obesity can each influence a person’s walking gait. Walking is a task of daily living that is commonly performed with backpack loads. Surveys have reported average (± standard deviation) backpack weights carried by university students to be 11.76 (4.17) lb. (Palmer et al., 2011), 12.98 (3.96) lb. (Smith et al., 2006), and 13.2 (7.26) lb. (Heuscher et al., 2010). Increasing body mass, via an external load or obesity, influences spatiotemporal, kinematic, kinetic, and metabolic cost of walking. It is important to understand the role of footwear during walking gait in obese individuals and those carrying trunk loads.

Spatiotemporal. Some easily observable characteristics of walking gait are related to length and time measures. Spatiotemporal parameters include such measures as walking speed, step length, and step time. These commonly reported measures provide basic descriptors of how the walking task was performed. For example, when carrying an external load, walking velocity decreases compared to the preferred unloaded speed

(Kellis & Arampatzi, 2009; Singh & Koh, 2009). Manipulating step length and frequency determine walking speed, and these measures are influenced by both body mass and footwear.

Spatiotemporal responses at fixed walking speeds between loaded and unloaded conditions are evident. For example, with incremental increases in load magnitude, there is a decrease in stride length (LaFiandra et al., 2003; Martin & Nelson, 1986; Myung & Smith, 1997). While this relationship appears to be more dramatic in females, that may be a consequence of not equally scaling the load applied to body mass for males and females (Martin & Nelson, 1986). Loads ranging from 1-52% body mass were imposed on the males, while the range for females was slightly greater, from 1-60%. This may partially explain the different responses.

The noted decrease in stride length is coupled with an increased stride frequency to maintain a fixed walking speed (Blacker et al., 2009; LaFiandra et al., 2003; Martin & Nelson, 1986). There are also significant decreases in swing time, increased time spent in double support (Kellis & Arampatzi, 2009; Martin & Nelson, 1986), and increased total stance time with added loads (Birrell & Haslam, 2010). In contrast, no spatiotemporal differences were reported by Devroey et al. (2007) with loads 5-15% of body weight. Goh et al. (1998) likewise found no significant differences in walking speed, stride length, or stride frequency with loads up to 30% body weight in male infantry personnel. In other samples, 15% body mass loads were sufficient to decrease walking speed, single support time, and increase double support time (Y. Wang et al., 2001). In male and female children carrying 17% of their body weight, significant decreases in stride length and single support time, and increased stance phase duration have been reported (Kellis

& Arampatzi, 2009). Load magnitude, experience with load carriage, age, and gender appear to determine, in part, gait spatiotemporal adaptations during load carriage.

At fixed walking speeds, obese individuals spend a greater time in stance, less time in swing, and more time in double support (Browning & Kram, 2007; Ranavolo et al., 2013). The longer stance and shorter swing times as a percent of stride were also reported earlier by DeVita and Hortobagyi (2003). These differences are also consistent across age groups, as similar shifts in these measures have been reported in obese children (McGraw, McClenaghan, Williams, Dickerson, & Ward, 2000). While stride length and stride frequency seem to be unaffected by body mass at fixed speeds (Browning & Kram, 2007; DeVita & Hortobagyi, 2003; Ranavolo et al., 2013), step width increase in obese adults (Ranavolo et al., 2013). This last measure is probably a factor of the increased circumference of the thigh in obese individuals. The spatiotemporal patterns presented above are thought to increase stability in obese individuals. For example, longer double support time reduces the relative contributions of a single limb to body weight support (Ranavolo et al., 2013).

Probably the most widely reported difference between shod and barefoot walking is that without shoes there is a decrease in stride length but an increase in stride frequency (Keenan et al., 2011; Lythgo et al., 2009; Majumdar et al., 2006; Oeffinger et al., 1999; van Engelen et al., 2010; Wolf et al., 2008). Stance time decreases (Lythgo et al., 2009; Majumdar et al., 2006; Zhang, Paquette, & Zhang, 2013) and toe-off may occur at an earlier percent of stride while barefoot (Wolf et al., 2008). Swing, as a percent of stride, also decreases while barefoot (Majumdar et al., 2006). A smaller percent of stride is spent in single support (Lythgo et al., 2009; Majumdar et al., 2006), and some report more time spent in double support while barefoot (Majumdar et al., 2006). In contrast, others have reported a decrease in double support time while barefoot (Lythgo et al., 2009). Variability in these spatiotemporal outcomes could be due to the type of footwear worn in these studies, walking speed, or age range of the participants. Majumdar et al. (2006) compared military boots to barefoot walking and included adult male military personnel, while Lythgo et al. (2009) compared barefoot to walking in the participants' own athletic footwear. The sample of Lythgo et al. (2009) also included a large range of ages (5 to ~20 years old) and included females and males.

Discomfort at initial contact, or during stance phase in general, may be a contributing factor in adopting these altered spatiotemporal parameters while barefoot. Interestingly, obese adults may have decreased plantar surface sensitivity, which may or may not play a role in gait adaptations given varying footwear conditions. Wu and Madigan (2014) reported 30% and 56% larger forces were necessary for the obese group to detect touch under the calcaneus and third metatarsal, respectively, compared to a normal weight group. If there is less sensitivity in the plantar surface and spatiotemporal differences are made through that mechanism, gait adaptations while barefoot may be masked. It is unknown whether this level of desensitization will influence gait patterns.

Kinematics. Kinematics covers a group of variables that can be identified visually. A few examples of kinematic variables are joint position, joint range of motion, and joint angular velocity. Even though the underlying mechanisms driving the kinematics are unknown without more advanced techniques, the visually observable patterns still provide some meaningful information regarding task performance.

A change in standing posture, or trunk flexion angle, is a common kinematic adaptation to load carriage. Backpack loads shift the body's center of mass posteriorly (Smith et al., 2006) and promote trunk flexion (A. B. Marsh, DiPonio, Yamakawa, Khurana, & Haig, 2006). A linear increase in trunk and hip flexion with applied loads from 0 to 30% of body mass have been observed (Devroey et al., 2007). The increased flexion of the trunk and hips requires larger activation of the rectus abdominis and external obliques, but lower activity in the erector spinae (Devroey et al., 2007). These kinematic and muscular activity changes are also evident during dynamic tasks, such as walking (Devroey et al., 2007). Other data suggest no changes in trunk angle during walking until load magnitudes exceeded 25% of body mass for males, and 28% of body mass for females (Martin & Nelson, 1986). Beyond these magnitudes, the increased flexion was also observed (Martin & Nelson, 1986). Similarly, non-linear increases in trunk flexion with added mass were reported by Majumdar et al. (2010). In that study the difference likely is a consequence of both the varying magnitudes of loads and the form of loads carried. Machine guns, rifles, and backpacks in various combinations were imposed, rather than backpacks alone. The posterior shift imposed by a trunk load may be countered by sufficient trunk flexion that the center of mass actually shifts anteriorly (Lloyd & Cooke, 2011).

Lower extremity kinematic changes are also observed with load carriage. Ankle and hip range of motion (ROM) increase with added loads, as seen in military personnel carrying military style equipment up to ~27% body mass (Majumdar et al., 2010). Majumdar et al. (2010) reported a more dorsiflexed ankle and greater knee flexion at initial contact with backpack loads. At toe-off, a more extended hip and knee with load were also observed. In another sample of military personnel, Tilbury-Davis and Hooper (1999) reported no sagittal plane differences with load carriage. It is somewhat surprising to see this lack of change given the large proportion of body mass (64%) the participants carried. However, the participants possibly had less influence of loading because they were "military personnel whose job involved load carriage" (Tilbury-Davis & Hooper, 1999). Level of training and method of load carriage may have produced these contrary results. Other populations who have received less training in load carriage are likely to be more effected by it. In college-aged males, for example, data show larger anterior tilt of the pelvis and more flexed hip and knee joints at initial contact (H. Wang et al., 2013). During stance, the knee remained more flexed (H. Wang et al., 2013). In females, load carriage is associated with smaller pelvic range of motion in the transverse and frontal planes (Smith et al., 2006). Though this population regularly carries backpack loads they are not of the same magnitude as those experienced in military personnel. The level of physical activity performed with these loads are also less, thus the extent of adaptation to load carriage is likely less.

Obese individuals are suggested to require more hip abduction and larger transverse plane pelvis range of motion to perform leg swing (Davids, Huskamp, & Bagley, 1996). This is a consequence of the larger thigh circumference in these individuals. Contact between the thighs prevents a more normal sagittal plane motion. During early stance, obese individuals have a more extended knee (DeVita & Hortobagyi, 2003; Gushue et al., 2005; McMillan, Pulver, Collier, & Williams, 2010) and hip (DeVita & Hortobagyi, 2003; McMillan et al., 2010). During mid-stance, Browning and Kram (2007) found no differences in joint angles of the hip, knee, or ankle between normal weight and obese individuals across a range of fixed speeds. At toe-off, DeVita and Hortobagyi (2003) observed greater dorsiflexion in obese persons. When considering average position throughout stance, DeVita and Hortobagyi (2003) found obese persons maintained greater hip and knee extension, and less dorsiflexion. Those authors did not provide an explanation of those observations, but commented that it was an attempt to maintain a “more erect walking pattern” (DeVita & Hortobagyi, 2003). These hip, knee, and ankle strategies probably assists with weight support. At preferred walking speeds, Lai, Leung, Li, and Zhang (2008) found no sagittal or transverse plane differences in joint positions between obese and normal weight individuals. Sagittal plane knee range of motion, however, may decrease at preferred speeds in obese individuals (Ranavolo et al., 2013). Given the shortened strides taken by overweight individuals it is not surprising that lower limb joint kinematics are also different.

There are many visually identifiable differences between barefoot and shod walking as well. Upon general inspection of angle vs. time curves, it is apparent that a phase shift exists between these two footwear conditions (Oeffinger et al., 1999). This is a consequence of toe-off occurring earlier in the gait cycle compared to when shod (Wolf et al., 2008). Besides this temporal misalignment, the shape of the kinematic profiles also differ. When walking barefoot, initial contact is made with a more neutral ankle angle (Oeffinger et al., 1999; Zhang et al., 2013). That is, the foot is closer to flat on the floor when contact occurs. This results in a smaller plantarflexion range of motion (ROM) during early stance (Zhang et al., 2013). Total ankle ROM in the sagittal plane is also decreased throughout a stride when not wearing shoes (Shakoor & Block, 2006; Wolf et al., 2008). Lastly, toe-out angle is decreased (Lythgo et al., 2009; Shakoor & Block, 2006). Reducing ROM and landing with a flatter foot may be a protective measure to avoid discomfort when a shoe does not help cushion or control pronation during weight acceptance.

Kinematic changes also occur above the ankle when footwear is manipulated. A decrease in knee and hip sagittal plane ROM have been reported for barefoot walking at equivalent speeds to the shod condition (Shakoor & Block, 2006). Initial contact is made with the knee in a more flexed position (Oeffinger et al., 1999; Zhang et al., 2013), and the knee remains more flexed throughout stance (Zhang et al., 2013).

Kinetics. Kinetics provide information of the forces involved in a task. These forces represent the underlying causes of the movement that can be observed as kinematics. Some kinetic variables of interest to be discussed include GRFs, pressure, and net joint moments.

The typical backpack design consists of a large compartment placed on the posterior torso and two straps that loop over the shoulders. Any objects placed inside the compartment pull posteriorly and inferiorly on the shoulders via straps. During quiet standing, pressures measured under the straps of a 25 kg backpack reach an average of 10 kPa on the chest, 12 kPa superior aspect of the shoulder, and 8 kPa for the posterior torso (Hadid, Epstein, Shabshin, & Gefen, 2012). Peak and average pressures of 55 and 6.5 kPa, respectively, can be reached with these loads (Hadid et al., 2012). Using a variety of pack configurations and weights, Mackie, Stevenson, Reid, and Legg (2005) reported overall backpack mass as the greatest contributor to pressure at the shoulder-strap interface. These authors (Mackie et al., 2005) reported 33.8 and 51.7 Newton mean and peak forces, respectively, applied to the shoulder by the strap with a 15% body weight load (7.9 kg) on a still manikin. These forces were associated with a total peak pressure (the sum total of each active sensor under the shoulder strap) of 446 kPa (Mackie et al., 2005). Walking was then simulated via 4.5 cm vertical oscillations of the manikin at a rate of 1.3 steps/sec using a 10% body weight pack (5.3 kg). The resulting peak force and pressure were 43.2 N and 390 kPa. It is not surprising then that shoulder and back pain are common complaints from individuals who routinely carry backpack loads (Smith et al., 2006).

At the distal end of the kinetic chain, load carriage increases the forces experienced at impact with the ground. There is general agreement that load carriage increases ground reaction force (GRF) magnitudes (Birrell & Haslam, 2010; Birrell et al., 2007; H. Wang et al., 2012). Some suggest the vertical and anteroposterior GRF increases are proportional to the added mass (Birrell et al., 2007; Tilbury-Davis & Hooper, 1999; Y. Wang et al., 2001). Others suggest only vertical GRFs scale with added mass (Kellis & Arampatzi, 2009), but this may be because those participants walked slower in the loaded conditions. In terms of force per area of the foot, standing with a backpack of ~16% body weight causes 16.9, 14.2, and 9.2% increases in mean pressures under the forefoot, midfoot, and rearfoot, respectively (Pau et al., 2015). Surprisingly, mean pressures only increased by 10.8, 10.7, and 4.6% in those three areas during walking, compared to the unloaded condition (Pau et al., 2015). In addition to the increased GRF profiles, Tilbury-Davis and Hooper (1999) reported ~41 and ~34% increase in braking and propulsive impulses, respectively, with a 20 kg load. These increased GRFs and pressures may promote the altered spatiotemporal and kinematic changes observed during load carriage.

Performing inverse dynamics allows estimates of the joint reaction forces and moments involved in performing movement. H. Wang et al. (2013) reported ~47% increase in hip extensor moments, ~83% increase in knee extensor moments, and a slight (~3%) but not significant increase in ankle dorsiflexor moment during weight acceptance while carrying a 32 kg load. Using the joint moments and angular velocities, joint powers may be calculated. H. Wang et al. (2013) reported ~64% increase in hip joint power production, ~98% increase in knee joint power absorption, and ~8% increase in ankle power absorption during weight acceptance. These mechanical measures can help explain the noted increase in metabolic cost associated with load carriage.

Given these large increases in GRFs, joint moments, and joint powers, it is not surprising that load carriage is associated with musculoskeletal trauma. One debilitating injury that can occur with load carriage is pack palsy, which causes muscle wasting and strength loss when the brachial plexus is compressed (Corkill, Lieberman, & Taylor, 1980). Ankle sprains, an injury specific to gait, are associated with load carriage (Yen, Gutierrez, Wang, & Murphy, 2015). Another serious health concern is the development of arthritis with long term exposure to increased mechanical loading of the body, such as that experienced with load carriage and obesity (Griffin & Guilak, 2005).

Obese individuals similarly experience greater vertical, anteroposterior, and medio-lateral GRFs than normal weight individuals (Browning & Kram, 2007). Additionally, the vertical and medio-lateral GRF increases scale to body weight (Browning & Kram, 2007). In comparison, when normalizing GRFs to body mass, obese individuals appear to experience lower peak vertical and anteroposterior forces (Lai et al., 2008). These differences may not be meaningful for two reasons: the obese participants walked 0.15 m/s slower than the normal weight group, and the magnitude of the differences were minimal (0.19 vs. 0.21 bodyweights for propulsive GRF peak, and 1.05 vs. 1.12 bodyweights for the 2nd vertical GRF peak for obese and normal weight, respectively).

Obese individuals present a challenge to the validity of kinetic data beyond the GRF profiles. The extra tissue and tissue movement during gait may perturb marker positions and introduce noise in the kinematic data used as inputs to inverse dynamics. Additionally, it may be difficult to palpate the underlying bony structures that are typically used for landmarks in motion capture experiments. Some researchers have attempted to minimize this issue by performing Dual X-Ray Absorptiometry on their participants (Browning & Kram, 2007). This allows the researcher to know more accurate and individualized inertial properties of the obese individual's body segments. Others have used data provided by the literature to estimate segment inertial properties (DeVita & Hortobagyi, 2003; Gushue et al., 2005).

In simple terms, larger masses require larger forces to create movement. This is true regarding joint moments in walking gait of the overweight population. Larger peak plantarflexor moments (DeVita & Hortobagyi, 2003; Lai et al., 2008), ankle work, and power have been reported in obese individuals (DeVita & Hortobagyi, 2003). A trend that did not reach significance for greater peak ankle moments was reported by Browning and Kram (2007). These samples seemed to adopt an ankle strategy, rather than a knee or hip strategy, to produce the extra work required to for walking with larger mass.

Anecdotally, obese individuals experience larger forces acting across lower extremity joints, which is one explanation of the increased prevalence of arthritis in that group (Felson, 1996). Significant increases in absolute hip and knee moments were indeed reported by Browning and Kram (2007). These authors chose to report absolute values because the joint contact surfaces do not increase in proportion to body mass. Normalizing to body mass is, in their opinion, inappropriate when dealing with obese individuals. The effect of normalizing to body mass can be seen in the data presented earlier by DeVita and Hortobagyi (2003). These authors reported significantly reduced knee extensor moments, normalized to mass, in obese compared to normal weight individuals.

No significant differences in knee extensor moments were observed between obese and healthy weight children (Gushue et al., 2005), and also in adults at preferred walking speeds while barefoot (Lai et al., 2008). However, knee abduction moments were greater in obese children (Gushue et al., 2005). Contrary to these findings, McMillan et al. (2009) reported significantly smaller peak knee abduction moments, normalized to body mass and height, during early and late stance. These authors also observed ~2 times the peak hip abductor moment of healthy weight children in obese children in early and late stance. In a follow-up to that study, McMillan et al. (2010) reported smaller ankle plantarflexor moments in late stance, knee moments in early and late stance, and hip moments in early stance in adolescents. The only joint moment larger in the overweight group was the hip moment in late stance.

The conflicting outcomes presented above are likely due to varying methodologies. A variety of normalization techniques, age ranges, sources of inertial property data (such as the use of adult parameters for children, as in Gushue et al. (2005)), footwear conditions, and speeds were chosen. Participants walked barefoot in two of these studies (Lai et al., 2008; Ranavolo et al., 2013), and shod in two others (Browning & Kram, 2007; DeVita & Hortobagyi, 2003). McMillan et al. (2010) did not report footwear condition. Ranavolo et al. (2013) did not fix walking speeds but instructed the normal weight adults to walk slowly, while the obese adults were told to walk naturally. No significant difference in speed between groups was reported but these conflicting instructions may have impacted walking strategies. Finally, Browning and Kram (2007) were interested in absolute joint moments and GRFs to understand the loads experienced by the lower extremity. DeVita and Hortobagyi (2003) reported moments normalized to mass, and others to mass and height (Lai et al., 2008; McMillan et al., 2010). This makes comparison between studies to understand the impacts of obesity on joint kinetics difficult. These factors confound the data and make it less obvious what is a product of obesity vs. a product of the imposed walking conditions.

The kinematic differences noted in the above section for barefoot and shod walking are driven by kinetic changes in these footwear conditions. One such parameter that likely influences spatiotemporal patterns is pressure under the foot. High plantar surface pressure is a variable investigated in diverse populations such as diabetics (Arndt, Ekenman, Westblad, & Lundberg, 2002), older adults (Burnfield et al., 2004), and those carrying a load (Arndt et al., 2002). Initiating stance with a flatter foot (Zhang et al., 2013) increases surface area in contact and may decrease discomfort. Wearing shoes further increases contact area, resulting in decreased pressure under the heel and central metatarsals (Burnfield et al., 2004). J. E. Perry, Ulbrecht, Derr, and Cavanagh (1995) reported increased pressures under the heel, all 5 metatarsal heads, and the hallux while walking in socks compared to leather walking shoes and running shoes. Walking faster also increases plantar pressures (Burnfield et al., 2004), so comparisons between footwear conditions must be made at comparable speeds.

Besides contact area, the force component of the pressure measure is different when walking barefoot. Keenan et al. (2011) reported a 6.9% decrease in the braking GRF, 4.6% increase in propulsive GRF, and 2.8% decrease in initial peak vertical GRF during treadmill walking at a preferred speed. Similarly, a 1.9% decrease in vertical GRF and 13.6% increase in propulsive GRF peak while barefoot were reported by Zhang et al. (2013). Despite the lower absolute force, the rate of force development was greater while barefoot because there was no cushion to absorb some of the force at impact (Zhang et al., 2013). However, Tilbury-Davis and Hooper (1999) reported no significant differences in anteroposterior GRF measures between barefoot walking and walking in military style boots. The difference in peak vertical forces at weight acceptance was much larger than that reported by Keenan et al. (2011) or (Zhang et al., 2013), at 19.8%. Additionally, Tilbury-Davis and Hooper (1999) reported a 3.1% decrease in the push-off peak while barefoot, while Keenan et al. (2011) actually saw a 0.4% increase in this measure. Once again, walking speed may have driven some of these differences. Participants walked at 1.3 m/s in Keenan et al. (2011) and (Zhang et al., 2013) while Tilbury-Davis and Hooper (1999) did not report the self-selected velocity of their participants for any condition. It is possible that their participants selected a different velocity for the barefoot and shod conditions, which would explain the large impact of footwear on the observed GRFs.

With the altered lower limb joint kinematics and GRF profiles, joint moments are also impacted by footwear condition. Keenan et al. (2011) reported that in healthy, young individuals, smaller ankle eversion moments at toe-off, a smaller knee varus moment and larger knee flexion moment just after initial contact were observed while barefoot. Other differences include a smaller hip flexor moment just after initial contact and smaller hip extensor moment at toe-off (Keenan et al., 2011). Oeffinger et al. (1999) found a decrease in the plantarflexor moment at toe-off, a decrease in the knee flexor moment just after initial contact, and an increased hip extensor moment just before initial contact while barefoot. In contrast, the only significantly different lower limb joint moment reported by Zhang et al. (2013) was a smaller dorsiflexor moment just after initial contact. Contrary to these studies, Shakoor and Block (2006) found no significant differences at the ankle, but reported barefoot walking decreased peak knee adduction and extension moments by 11.9% and 7.4%, respectively. In addition, hip adduction, internal rotation, and external rotation were decreased without shoes (Shakoor & Block, 2006). Differences among the outcomes of these studies are likely related to the characteristics of the sample groups. The participants include children (Oeffinger et al., 1999), young adult males (Zhang et al., 2013), young adult males and females (Keenan et al., 2011), and older adult men and women with knee osteoarthritis (Shakoor & Block, 2006). It is likely that age introduces a confounding factor when comparing these studies. It is possible that gender played a role in these differences as well, but some data suggests barefoot walking requires similar knee joint moments for males and females (Kerrigan, Riley, Nieto, & Della Croce, 2000).

Joint powers, on the other hand, have received minimal attention in the footwear literature. Oeffinger et al. (1999) appear to be the only group to have addressed these dependent variables between barefoot and shod conditions. These authors recruited normal, healthy children (7-10 years old), who wore their own athletic shoes during the shod conditions. Compared to the barefoot condition, the ankle in the shod condition absorbs more power in early stance, but produces less power near toe-off. At the knee, more power was generated during early stance, and less power absorbed in late stance while shod. No differences in hip power generation or absorption were reported. These differences were not attributed to differences in preferred walking speed (1.39 m/s vs. 1.43 m/s for barefoot and shod, respectively).

Metabolic cost. The energy input necessary to complete a given task is useful for comparing relative difficulty. There are many methods available to estimate the cost of an activity using mechanical and/or metabolic means. On the physiological side, the amount of oxygen consumed and carbon dioxide produced can be measured. The exchange of these gasses provides the means to determine caloric expenditure, or internal work. On the mechanical side, the amount of work performed by the segments themselves or on the center of mass define external work. This external work can also be segmented into positive, negative, net, or total work. The ratio of external work to internal (i.e., metabolic) work can be used to describe the efficiency of motion. Oftentimes, completing the most external work with the least internal work is desired.

The ability to carry large loads at quick speeds with low metabolic cost is desirable. In guinea fowl, backpack loads of 23% body weight increased metabolic cost by 17%, which was consistent across multiple speeds (R. L. Marsh, Ellerby, Henry, & Rubenson, 2006). In humans, a linear increase in metabolic cost with incremental increases in load has also been reported (Bastien, Schepens, Willems, & Heglund, 2005; Bastien, Willems, Schepens, & Heglund, 2005; Pal et al., 2009). Two mechanical tasks have been identified that account for a large portion of the increased metabolic cost of walking with increased mass. About 28% of the extra metabolic cost is accounted for by supporting a larger mass, while ~45% is accounted for by the additional work performed on the center of mass (Grabowski et al., 2005).

Minimalist shoes have become a popular form of footwear due to their lighter construction compared to standard athletic footwear. Data indicate that for every 100 g of mass added to the foot the cost of running increases by 1% (Franz, Wierzbinski, & Kram, 2012). The same linear relationship may or may not hold for walking, but data does show that reducing the mass of the foot segment decreases metabolic expenditure. For example, in middle aged women and men, cost of walking (J/kg/m) while barefoot was 13.7% less than walking in a MBT (Swiss Masai, Switzerland) shoes (van Engelen et al., 2010). In that study, a walking shoe was 3.37% more metabolically costly but this did not reach significance. Although metabolic work was lowest while barefoot, van Engelen et al. (2010) reported that positive and negative external mechanical work performed on the center of mass were greatest while barefoot. However, no differences in total external mechanical work was observed between the barefoot and shod conditions (van Engelen et al., 2010). Gjøvaag, Dahlen, Sandvik, and Mirtaheri (2011) also compared MBT shoes to standard athletic shoes, but did not include a barefoot condition. These authors reported no difference in level treadmill walking oxygen consumption at a freely chosen speed, nor at fixed speed, between the two shoe types despite a 216 g difference in shoe masses. This could be due to the young (mean of 22.9 years old), fit condition of the participants not being metabolically challenged by the task.

This review of literature reveals the need to continue researching gait mechanics of load carriage and obese individuals. The impact of footwear in these conditions will also be addressed to provide a clearer picture of adaptations made to an increased body mass. Understanding the responses to barefoot and shod walking will hopefully explain the conflicting outcomes of previous studies that have investigated walking mechanics in obese individuals. Walking without a highly cushioned shoe may be a means to alter lower extremity joint mechanics in obese individuals. Specifically, decreasing knee joint loads would be considered a positive adaptation.

CHAPTER III

STUDY 1: EFFECTS OF LOAD CARRIAGE AND FOOTWEAR ON SPATIOTEMPORAL PARAMETERS, KINEMATICS, AND METABOLIC COST OF WALKING[1]

Introduction

Backpack loads of 10% or greater of body mass are common among the college student population (Heuscher et al., 2010). Two-thirds of students in a recent survey reported daily backpack use (Heuscher et al., 2010) and walked an average of 9.04 miles weekly, most of which while carrying a backpack (Schwebel, Pitts, & Stavrinos, 2009). Loads as small as 12% of body mass have been shown to negatively influence pedestrian behaviors. For example, reduced walking speeds and reduced distances to an oncoming vehicle while crossing a street have been observed (Schwebel et al., 2009). Additionally, lower extremity injury and/or low back pain may be consequences of habitual load carriage (Heuscher et al., 2010; Martin & Nelson, 1986). With the significant distances and time spent walking with a backpack weekly, it is important to understand the unique responses of college-aged individuals to loaded walking. However, few studies have investigated load carriage in this population using loads similar to those that these individuals experience on a daily basis.

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Y. Wang et al. (2001) simulated the effect of typical backpack loads by adding 15% of body mass to backpacks in a group of college students. It was reported that while loaded single support time and step frequency decreased, whereas double support time increased. The increased double support time may be an attempt to increase stability when a load is applied to the trunk (Singh & Koh, 2009), while the decreased single support time presumably reduces the support contribution required by an individual leg (Y. Wang et al., 2001). Grabowski et al. (2005) suggested that the additional musculoskeletal effort required to maintain an upright body position and to generate forces necessary to propel the body during loaded walking are major contributors to the noted increases in metabolic cost of walking with an extra load. Martin and Nelson (1986) suggested that an increase in support time may also increase the risk of injury to the lower limbs.

Controversy exists in the literature about the effects of barefoot running on metabolic cost in part due to methodological differences across studies. For example, Hanson, Berg, Deka, Meendering, and Ryan (2011) have reported reductions in metabolic costs during barefoot running, while Divert et al. (2008) report no differences between barefoot and shod running. van Engelen et al. (2010) reported a 3.5% reduction in metabolic cost while walking barefoot compared to shod, but this difference was not significant. This suggests that even during walking there may be a potential effect on metabolic costs if the shoes are removed.

Barefoot locomotion conditions have also been shown to lead to alterations in locomotion mechanics (Keenan et al., 2011). Based on our observations, college students commonly wear unsupportive footwear that also have minimal cushioning between the foot and ground at contact. Observed spatiotemporal differences while walking barefoot, compared to shod, include: reduced speed, step length, double support time, and total support time (Lythgo et al., 2009). Increased step frequency and single support times have also been reported during barefoot walking (Lythgo et al., 2009). Zhang et al. (2013) reported that barefoot walking results in a higher loading rate than shod walking, which likely influences the adopted spatiotemporal gait characteristics presented above. Specifically, it is presumed that shortening the stride length reduces the discomfort experienced at foot strike without the cushioning of a standard shoe (Majumdar et al., 2006). It is currently unknown if these spatiotemporal differences would be further altered by a load when walking barefoot.

Given the noted gait adjustments made under novel conditions (i.e. loaded or barefoot) it was of interest to understand what effects these promote while simultaneously experienced. For example, adding a backpack load while walking shod increases stance time (Kellis & Arampatzi, 2009), but changing to barefoot walking from shod walking decreases stance time (Majumdar et al., 2006). It is unclear how stance time will respond when the shoe condition and load condition are simultaneously manipulated given the opposite effects of these conditions individually. The addition of a backpack load while barefoot may potentially increase the discomfort of initial contact and accentuate pain-reducing strategies and modify gait mechanics beyond those noted in barefoot walking without a load. However, the effect of carrying heavy loads without a supportive shoe on walking kinematics, spatiotemporal parameters, and economy is still unclear.

Thus, the purpose of this study was to investigate the simultaneous effects of loading and footwear changes on gait mechanics and walking economy. We hypothesized that loading, regardless of footwear, would elicit shorter stride lengths, longer stance times, longer double support times and increased metabolic costs. In contrast, we hypothesized that walking barefoot, regardless of load, would have the exact opposite effect on these measures. Therefore, it was our expectation that adding a backpack load to individuals walking barefoot would result in spatiotemporal patterns and metabolic costs similar to those of shod unloaded walking. Our first two hypotheses are based on previous findings from the literature where barefoot and load effects have been reported by themselves. Our last hypothesis, is simply a combination of the first two hypotheses with an expectation that barefoot and load effects observed individually will cancel each other out when experience simultaneously.

Methods

Participants

Twelve individuals (7 female, 5 male) participated in this study (age = 24 ± 2 years, height = 1.73 ± 0.13 m, and mass = 71.1 ± 16.9 kg). All participants were healthy, recreationally active and free of any notable gait abnormalities. The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

Experimental Protocol

Anthropometric data (including body mass and height) were collected based on VICON's full body plug-in-gait model with medial markers on the knee and ankle to better identify knee and ankle axes (Wong, Callewaert, Labey, Leardini, & Desloovere, 2009). Reflective markers were placed on various anatomical locations using double-sided tape based on the plug-in-gait model. Participants then walked on a level treadmill

(Woodway, Waukesha, WI) at 1.5 $m \cdot s^{-1}$ for 6-min under four conditions: Barefoot Unloaded (BU), Shod Unloaded (SU), Barefoot Loaded (BL), and Shod Loaded (SL). This model of treadmill was selected because its rubberized slats allowed steady state barefoot walking to be accomplished without blister formation or undue discomfort. A moderately higher walking speed than previously used (Keenan et al., 2011) was selected in an effort to increase the demands on the system so that alterations in movement patterns would be more apparent. A backpack equal to 15% of the participant's body mass was worn during the two loaded conditions. A single textbook was placed in the pack against the participant's back to provide a solid, flat surface before adding lead weights until the desired mass of the backpack was achieved. Participants performed the shod conditions using their own athletic shoe (mean shoe mass = 272 ± 68 g). The order of conditions was individually randomized and a brief rest was provided between successive walking bouts. The rest period was based on the time it took to change from one condition to the next and only lasted a couple of minutes. Randomization of all conditions across all participants was used in attempt to minimize any fatigue effects in this study. During all walking trials, metabolic (ParvoMedics, Sandy, UT) and motion (100 Hz) (VICON, Englewood, CO) data were collected. For metabolic data collection, expired gasses were passed into the gas analyzer via a hose and mouthpiece. A nose plug was worn to force all expired gasses to enter the mouthpiece. Motion data were collected during the last two minutes of each walking trial, which is where steady-state metabolic responses also occurred.

Data Analysis

Mean rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) over the last 2-min of each 6-min trial (van Engelen et al., 2010; Warne & Warrington, 2014) were used to estimate average rate of energy consumption (Weir, 1949):

$$\dot{E} = (3.9)\dot{V}O_2 + (1.1)\dot{V}CO_2 \tag{1}$$

where $\dot{E}$ is energy cost in kcal/min, and $\dot{V}O_2$ and $\dot{V}CO_2$ in L/min. $\dot{E}$ was converted to units of J/s and normalized to body mass. Metabolic cost was not normalized to any additional mass added to the body. We felt not accounting for the additional passive mass reflected best the real world metabolic consequences of walking with additional mass.

For spatiotemporal and kinematic measures, marker data were processed using VICON Nexus. Marker coordinate data were filtered using a 4th Order, recursive digital Butterworth filter with a cut-off frequency of 6 Hz. Joint kinematics were determined using the built-in plug-in-gait model in VICON Nexus. Velocities were derived using finite difference approximations.

Foot contact events (i.e., heel strike and toe-off) for each leg were visually identified during post-processing by a single researcher. This researcher identified heel strike as the first frame in which the heel marker stopped moving downwards. Toe off was identified as the first frame in which the toe marker began moving upwards. The foot contact events were then used to determine spatiotemporal measures during the trial, which included stance time, double support time, and swing time. Stride time was determined as the sum of stance and swing times for a given leg. Stride length was determined based on the walking velocity relationship:

$$SL = V*ST \qquad (2)$$

where SL represents stride length in m, V represents the walking velocity (1.5 $m \cdot s^{-1}$), and ST represents stride time in seconds.

Kinematic dependent measures included ankle, knee, and hip ranges of motion and peak sagittal plane angular velocities. Range of motion (ROM) for each lower limb joint was determined by:

$$ROM = \text{max flexion angle} - \text{max extension angle} \qquad (3)$$

Statistical Analysis

Means for spatiotemporal and kinematic data were determined from three consecutive strides during the final two minutes of each trial. The average metabolic costs during the final two minutes of each of the four trials were compared. A series of 2x2 (loading, footwear) ANOVAs with repeated measures were performed using SPSS (version 20). Alpha was set at .05 for all tests. A Bonferroni post hoc test was performed where pairwise comparisons were appropriate.

Results

Three participants (2 F, 1 M) were excluded from the joint kinematic analyses due to marker loss in at least one condition. The lower edge of the backpack obscured PSIS and sacral markers in these participants, not allowing us to compute joint kinematics of the lower extremities. Spatiotemporal measures were still able to be processed for these individuals, as well as metabolic data. Thus, these data were included in statistical analyses.

Loading increased metabolic costs of walking by about 12% ($p < .001$) (Figure 2.1), but had little to no effect on spatiotemporal measures (Table 3.1). Walking shod increased metabolic cost by ≈1% ($p = .124$), but this was not significant (Figure 3.1). A footwear*loading interaction ($p = .039$) for ankle ROM occurred as the effect of load was dependent on footwear. Loading decreased ankle ROM while barefoot, but increased ankle ROM while shod (Table 3.2). Figures 3.2 and 3.3 and Table 3.2 display hip and ankle joint velocities normalized to gait cycle (heel strike to heel strike). A significant increase in peak hip flexion velocity (p=.002) was observed between 60-80% of the gait cycle when walking with the backpack load. Peak plantar flexion velocity (p=.039) was greater between 50-60% of the gait cycle while walking barefoot. While walking with a backpack load, dorsiflexion velocity between 60-80% of the gait cycle was greater compared to no load (p=.024). Shod walking had longer stride lengths ($p < .001$), stance times ($p < .001$) and double support times ($p = .001$) (Table 3.1). ROM of the hip joint also decreased when walking barefoot ($p = .02$; Table 3.2). No significant kinematic changes were observed at the knee.

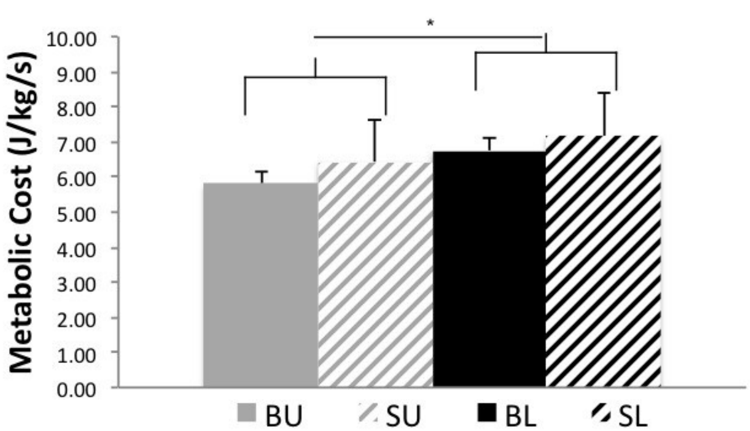

Figure 3.1. Means (+ SD) for metabolic cost according to condition. * indicates significant load effect (p < .05).

Table 3.1

Spatiotemporal parameters (means ± SD)

	BU	**SU**	**BL**	**SL**
Double Support (s)	0.069* (0.019)	0.095 (0.012)	0.072* (0.016)	0.098 (0.015)
Stance (s)	0.540* (0.038)	0.602 (0.034)	0.543 * (0.041)	0.584 (0.045)
Swing (s)	0.415 (0.024)	0.415 (0.030)	0.406 (0.038)	0.412 (0.025)
Stride Length (m)	1.43* (0.067)	1.53 (0.069)	1.42* (0.079)	1.49 (0.084)

* indicates significant footwear effect (*p* < .05).

Table 3.2

Sagittal plane kinematics (means ± SD)

	BU	**SU**	**BL**	**SL**
Range of Motion (deg)				
Ankle	37.50 (9.79)	29.29 (7.63)	35.15† (13.16)	33.02† (4.86)
Knee	58.36 (6.33)	59.88 (10.27)	56.20 (9.95)	59.34 (9.04)
Hip	47.44* (5.03)	51.80 (5.68)	51.66* (5.29)	53.69 (6.52)
Peak Angular Velocity (deg·s^{-1})				
Plantar Flexion	-487.44* (194.45)	-358.67 (140.7)	-436.44* (226.26)	-373.89 (105.16)
Dorsiflexion	213.33 (59.04)	172.89 (62.13)	225.33** (82.6)	204.56** (31.58)
Knee Flexion	346.33 (43.59)	343.11 (79.11)	337.00 (71.31)	362.00 (38.63)
Knee Extension	-395.89 (65.79)	-408.67 (73.15)	-389.67 (78.47)	-404.78 (50.88)
Hip Flexion	146.67 (13.53)	149.11 (16.79)	166.67** (18.63)	160.22** (15.47)
Hip Extension	-231.22 (28.5)	-257.33 (39.76)	-260.78 (36.43)	-265.67 (44.46)

* indicates footwear effect (*p* < .05).
** indicates load effect (*p* < .05).
† indicates a significant footwear x loading interaction (*p* < .05).

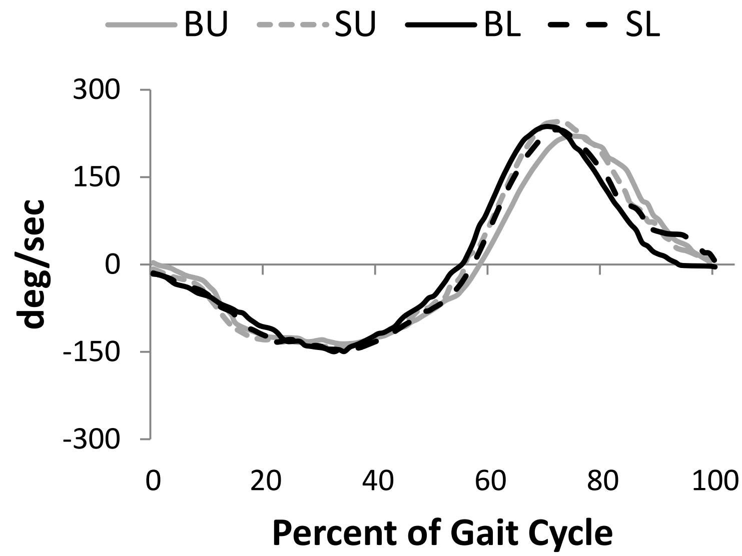

Figure 3.2. Mean hip joint angular velocities according to condition, normalized to percent of the gait cycle. The gait cycle was defined as initial heel strike of one leg to subsequent heel strike of the same leg. A significant increase in peak hip flexion velocity (p=.002) was observed between 60-80% of the gait cycle when walking with the backpack load.

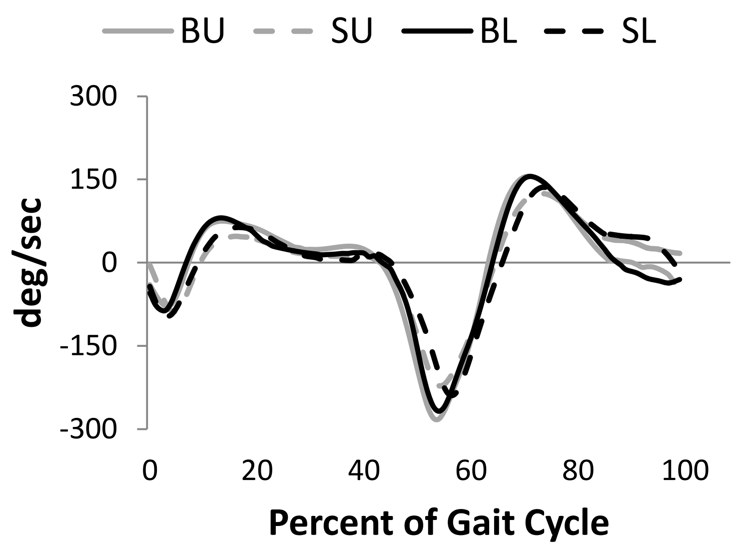

Figure 3.3. Mean ankle joint angular velocities for each condition, normalized to percent of the gait cycle. The gait cycle was defined as initial heel strike of one leg to subsequent heel strike of the same leg. Peak plantar flexion velocity (p=.039) was greater between 50-60% of the gait cycle while walking barefoot. While walking with a backpack load, dorsiflexion velocity between 60-80% of the gait cycle was greater compared to no load (p=.024).

Discussion

In order to investigate the combined effects of loading and footwear, we recruited college-aged individuals to perform steady state walking bouts on a treadmill in two footwear (i.e. shod, barefoot) and two loading (i.e. no load, 15% body mass backpack) conditions. Significant lower limb sagittal plane kinematic changes were observed at the ankle and hip joints in response to both loading and footwear. Spatiotemporal parameters were influenced by footwear, but not loading. Metabolic cost was significantly influenced by loading, but not footwear.

We observed a decrease in hip ROM while barefoot. Shakoor and Block (2006) reported this difference between shod and barefoot walking as well, but also reported a decreased ROM in the ankle and knee, which we did not observe. However, their population included individuals with osteoarthritis, while ours were all healthy, active, and young individuals. For ankle ROM in our study, an interaction of loading*footwear occurred. While barefoot, there was a decrease in ankle ROM with the addition of load (37.5 ± 9.8 degrees BU vs. 35.2 ± 13.2 BL), but an increase while shod (29.3 ± 7.6 degrees SU vs. 33.02 ± 4.9 SL). Unlike the clinical population (Shakoor & Block, 2006), our participants had greater ankle ROM while barefoot. This may be due to the fact that our population was significantly younger, had no musculoskeletal diseases, and/or were walking on a treadmill.

Spatiotemporal data for barefoot walking agree with previous work (Lythgo et al., 2009; Majumdar et al., 2006; van Engelen et al., 2010). The reduction in hip ROM and shorter strides while barefoot may indicate a change in walking strategy due to the discomfort associated with initial contact (Zhang et al., 2013). In the present study, this discomfort did not seem to be exacerbated by the load, as loading did not further alter spatiotemporal parameters while barefoot. Keenan, Franz, Dicharry, Della Croce, & Kerrigan (Keenan et al., 2011) reported that walking barefoot elicits lower initial peak vertical ground reaction forces (GRF) and braking GRFs, but higher propulsive GRFs, which they attributed to the decrease in stride length while barefoot. Although no kinematic changes were observed at the knee in this study, Keenan et al. (Keenan et al., 2011) observed differences in kinetic variables at the knee between barefoot and shod walking. Inclusion of these types of variables would provide a more complete picture of how humans cope with walking in different footwear conditions. It is currently unknown how kinetic variables would be influenced with the addition of a load during barefoot walking.

Metabolic cost in our study was approximately 1% lower while walking barefoot compared to shod, but this was not significant. Franz et al. (2012) reported a ≈1% increase in the metabolic cost of running for each 100 g added to the foot. Given the average mass of athletic shoes worn in the present study was 272 g, we had a slightly reduced effect due to shoe mass than we expected. Based on Franz et al., we should have observed approximately a 2-3% reduction in metabolic cost. Previous data (van Engelen et al., 2010) suggests barefoot walking cost is 3.5% lower than walking in a standard shoe. This was also not significant, but greater than the change observed in the present study. A potential limitation in our study was that we used a treadmill with comfortable, rubberized slats that likely reduced the discomfort associated with ground contact during barefoot walking. Thus, one might question whether our barefoot condition was more consistent with a minimalist shoe condition, given the rubberized nature of our treadmill belt. This would be consistent with recent findings that running barefoot is not the same as running in minimalist shoes (Bonacci et al., 2013).

The expected increase in metabolic cost while loaded was observed. This may be accounted for by increased activity of lower limb and trunk musculature. While standing, carrying a backpack elicits increased rectus abdominis activity (Al-Khabbaz, Shimada, & Hasegawa, 2008) and, while walking, a more flexed trunk angle to counter the large extensor moment induced by the backpack (Goh et al., 1998). Maintaining balance and supporting a larger mass while loaded requires additional muscular efforts that increase the steady state cost of walking (Grabowski et al., 2005). The short bouts of loaded walking in this study did not induce fatigue, but longer durations of loaded walking have been suggested to decrease stability and possibly increase risk of falls (Simpson et al., 2012) as lower limb muscles fatigue and the cost of walking continues to increase (Blacker, Williams, Fallowfield, & Willems, 2011). Besides tripping, load carriage may elevate the risk of lower limb trauma as a result of the increased demands placed on the lower limbs (Martin & Nelson, 1986) and/or be associated with the incidence of low back pain in college students (Heuscher et al., 2010). These risks seem to be especially a concern for females (Heuscher et al., 2010; Martin & Nelson, 1986).

One of the limitations of this study was the small sample size which limits the generalizability of our results. We also focused on a single walking speed and a single load condition for a short duration walk of 6-min, which further limits the generalizability of our results. Additionally, slight differences in foot marker positions between the barefoot and shod conditions were unavoidable. Barefoot walking and walking with minimalist shoes may not result in identical responses to load. Thus, further work should focus on responses specifically while wearing minimalist footwear. Finally, a treadmill with slightly compliant rubberized slats allowed barefoot walking to be performed without undue discomfort, but may have attenuated differences that may otherwise have been observed.

Conclusion

Our hypothesis that spatiotemporal parameters would be influenced by loading was not supported. However, our hypothesis that metabolic cost would increase while loaded was supported. Our hypothesis that barefoot walking would elicit spatiotemporal differences was supported, but the expected decrease in metabolic cost while barefoot did not occur. Therefore, a backpack load does not seem to influence spatiotemporal parameters in the college student population, regardless of footwear. Our results also suggest that spatiotemporal changes made during barefoot walking are accomplished by altering kinematics of the ankle and hip, but not the knee. The lack of change noted in spatiotemporal parameters while loaded and barefoot is consistent with our expectations that their individual effects would cancel each other out. Future work should focus on a more rigid surface than the treadmill used in this study and kinetic parameters to better understand the differences in these footwear conditions with a load.

CHAPTER IV

STUDY 2: BAREFOOT VERSUS SHOD: EFFECTS OF BACKPACK LOADS ON WALKING MECHANICS

Introduction

Walking is a common task that young adults perform with backpack loads. In some surveys, the average backpack carried by a university student is 11.76-13.2 lbs (Heuscher et al., 2010; Palmer et al., 2011; Smith et al., 2006). These loads influence spatiotemporal, kinematic, and kinetic parameters of level walking. For example, trunk loads promote shorter stride lengths at higher stride frequencies to maintain a fixed speed (Blacker et al., 2009; LaFiandra et al., 2003; Martin & Nelson, 1986). Kinematic differences with loads include greater hip and knee flexion during stance (H. Wang et al., 2013). Kinetic differences include increased ground reaction force (GRF) magnitudes that are proportional to the added mass (Birrell et al., 2007; Tilbury-Davis & Hooper, 1999) and greater hip and knee extensor moments, greater hip power generation, and greater power absorption at the knee and ankle (H. Wang et al., 2013).

Walking mechanics also differ between barefoot and shod conditions. Compared to walking shod, shorter stride lengths with an increased stride frequency (Dames & Smith, 2015; Keenan et al., 2011; Shakoor & Block, 2006; van Engelen et al., 2010) and decreased ranges of motion (ROM) at the hip, knee, and ankle joints have been reported (Shakoor & Block, 2006). These kinematic and spatiotemporal adjustments to barefoot walking are associated with decreased braking and vertical GRFs during early stance

(Keenan et al., 2011). Additionally, peak knee flexor moments increase, and hip flexor moments decrease during early stance while barefoot (Keenan et al., 2011). Those altered joint kinetics may be a response to the increased plantar pressures experienced while barefoot (Sarnow et al., 1994). Increasing body mass via an external load may exacerbate these changes, but currently the simultaneous impact of load carriage across footwear conditions is unclear.

Titchenal, Asay, Favre, Andriachi, and Chu (2015) compared three footwear conditions (athletic shoe, 3.8cm heels, and 8cm heels) with and without a 20% bodyweight load. They reported increased knee extensor moments in late stance, and larger abductor moments during early, middle, and late stance with the load. Rather than raising heel height, Dames and Smith (2015) investigated the kinematic and metabolic effects of treadmill walking barefoot vs. shod with trunk loads. Dames and Smith (2015) used a treadmill with a rubberized slat design that likely improved comfort during barefoot walking, but may have attenuated responses to the loading condition. Additionally, the treadmill used did not have force measuring capabilities, which limited the authors' ability to provide insights into lower extremity kinetics during the walking conditions.

The present study seeks to understand how simultaneously imposing external loads and varying footwear conditions impact overground walking mechanics in young, healthy adults. Understanding the underlying kinetic responses to footwear and load carriage would provide further insights into the observed spatiotemporal, kinematic, and metabolic responses previously reported (Dames and Smith, 2015). Based on the literature, it was hypothesized that ground reaction forces, joint moments, and joint powers would increase with load. It was also hypothesized that walking barefoot would reduce ground reaction forces, joint moments, and joint powers. Finally, due to the opposing effects of footwear and load carriage on these measures, it was hypothesized that when simultaneously barefoot and carrying a load, these measures would not be different than the shod, unloaded condition.

Methods

Participants

Twelve young, healthy individuals (5 women, 7 men) with no known musculoskeletal or neurological issues that would compromise gait were recruited for this study (age = 23 ± 3 years, height = 1.73 ± 0.11 m, and mass = 70.90 ± 12.67 kg). The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

Data Collection

Participants wore tight-fitting clothing throughout the experiment so that anatomical landmarks could be easily identified and to minimize marker movements. Anthropometric data (i.e., body mass, height and various segment widths and lengths) were measured based on VICON's Plug-in-Gait model. Retroreflective markers were attached various anatomical landmarks using double-sided tape. Participants then performed overground walking trials at 1.5 $m{\cdot}s^{-1}$, which is slightly faster than the preferred speed of young adults (Norris, Granata, Mitros, Byrne, & Marsh, 2007), in four walking conditions: Barefoot Unloaded (BU), Shod Unloaded (SU), Barefoot Loaded (BL), and Shod Loaded (SL). Two pairs of timing gates (BROWER Timing Systems, Draper, UT) were used to ensure walking speed was within ±5% of the target speed. A backpack loaded with lead weights equal to 15% of the participant's body mass was worn during the loaded conditions. Participants wore their own athletic shoe for the shod conditions (272 ± 68 g). The order of these four conditions was individually randomized. During each trial, 3D motion (100 Hz) (VICON, Englewood, CO) and ground reaction force (GRF) (2000 Hz) data were collected. GRFs were measured by a tandem-belt instrumented treadmill (AMTI, Watertown, MA) embedded in the center of the walkway. Trials included in the data analysis were within the expected velocity range and clean foot contacts were made with the force plates (i.e., a single, whole foot contact on each force plate).

Data Analysis

Markers were labeled within VICON Nexus, but all subsequent processing of data was performed using a custom Visual 3D (C-Motion, Germantown, MD) script. Marker data were filtered using a recursive, Butterworth lowpass filter (F_c = 6 Hz). This cutoff frequency for filtering marker data was confirmed with a residual analysis performed in MATLAB (MathWorks, Natick, MA) as described by (Winter, 2009). GRF data were filtered using a recursive, Butterworth lowpass filter (F_c = 50 Hz). Motion and GRF data were combined through inverse dynamics to estimate joint reaction forces and moments for the ankle, knee, and hip in the sagittal plane. Joint powers were calculated as the product of the joint moment and angular velocity. Joint power peaks consistent with those selected by Winter (1987) were used in statistical analyses. Joint power peaks from the phases defined by Winter (1987) were used in statistical analyses. These include two ankle phases (A1, A2), four knee phases (K1-K4), and three hip phases (H1-H3). A1 is the initial weight acceptance and A2 the propulsive peak at toe-off. K1 is the energy absorption phase during weight acceptance. K2 is the only power generation phase and occurs during mid-stance. K3 is power absorption during terminal stance and early swing phase. The K4 phase is terminal swing power absorption. H1 phase occurs during early stance, H2 is an absorption phase during mid-stance, and H3 a power generation phase prior to toe-off. All joint moment and power data were normalized to body mass.

Statistical Analysis

Dependent variables were determined from three successful strides and then averaged. A series of 2 x 2 (loading, footwear) ANOVAs with repeated measures were performed using IBM SPSS 23.0 (Armonk, NY). Alpha was set at .05 for all tests. A Bonferroni post hoc test was performed where pairwise comparisons were appropriate.

Results

Loading

Walking with load increased stance time (p = .008) and double support time (p < .001) (Table 4.1). Hip range of motion (ROM) increased (p < .001) with load (Table 2). There was a trend of increased knee ROM (p = .050), but no loading effect was observed for ankle ROM. Loading increased peak plantar flexor velocity during late stance (p = .031) and dorsiflexor velocity in early swing (p = .013) (Figure 4.2). At the knee, peak extension velocity in late stance (p = .047), and both hip flexion (p < .001) and extension (p = .012) velocities increased with load.

Braking GRF was 16.5% greater (p < .001), and propulsive GRF was 10.7% greater while walking with additional load (p < .001) (Figure 4.1). Loading increased the initial and second vertical GRF peaks by 12.6% (p < .001) and 13.8% (p < .001), respectively. The minimum vertical GRF (between the peaks) was 9.6% larger while loaded ($p < .001$).

In addition to altered GRFs, loading increased the peak ankle plantar flexor moment by 11.15% ($p < .001$) (Figure 4.2). At the knee, the first peak extensor moment was 17.78% greater ($p = .001$) while loaded. The peak hip extensor moment in early stance was 5.86% greater ($p = .001$) and the peak hip flexor moment just before toe-off 17.11% greater ($p < .001$) while loaded.

Peak hip power absorption just before toe-off (40-50% of stride) was 19.10% greater while loaded ($p < .001$) (Figure 4.2). Peak hip power generation at toe-off (~60% of stride) was 18.12% ($p = .001$) larger while loaded. Loading increased peak knee power absorption during early stance (5-15% of stride) by 19.94% ($p = .006$) and in late stance (50-60% of stride) by 18.65% ($p = .001$). Finally, peak power generation at the knee during stance (~20% of stride) increased by 24.60% ($p = .001$) with load.

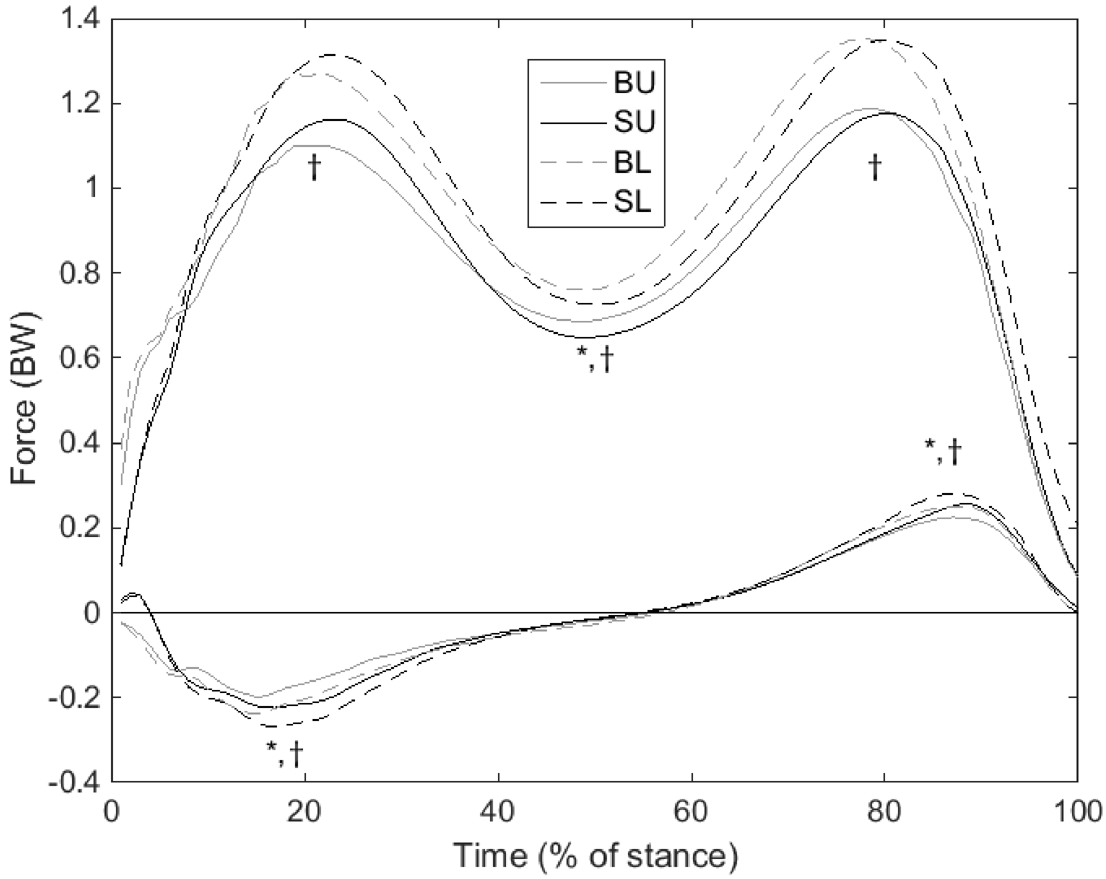

Figure 4.1. Vertical and anteroposterior ground reaction force profiles.* indicates significant footwear effect and † indicates significant load effect (p < .05).

Footwear

Walking barefoot decreased stride length ($p < .001$), stance time ($p < .001$), swing time ($p = .011$), and double support time ($p = .001$). Knee joint ($p = .01$) and hip joint ($p = .005$) ROM decreased while barefoot, but no footwear effect was observed for ankle ROM. Peak plantar flexion velocity in early stance ($p = .011$) and late stance ($p = .040$) were higher while shod. Hip and knee angular velocities were not influenced by footwear.

Walking barefoot decreased the peak braking GRF by 12.3% ($p = .004$) and peak propulsive GRF by 13.4% ($p = .001$). The minimum vertical GRF was 5.9% larger while barefoot ($p = .002$). Peak vertical GRFs (during weight acceptance, push-off) did not differ between footwear conditions.

Ankle moments and powers were not influenced by footwear conditions. The first hip extensor moment (just after initial contact) was 20.47% larger ($p = .001$), and the second peak knee extensor moment (~55% of stride) 23.36% ($p = .003$) larger while barefoot. Peak hip and knee power absorption in late stance (40-60% of stride) were 29.32% ($p < .001$) and 10.18% ($p = .010$) greater, respectfully, while barefoot. Peak hip power generation at toe-off (~60% of stride) was 34.00% ($p < .001$) larger while barefoot.

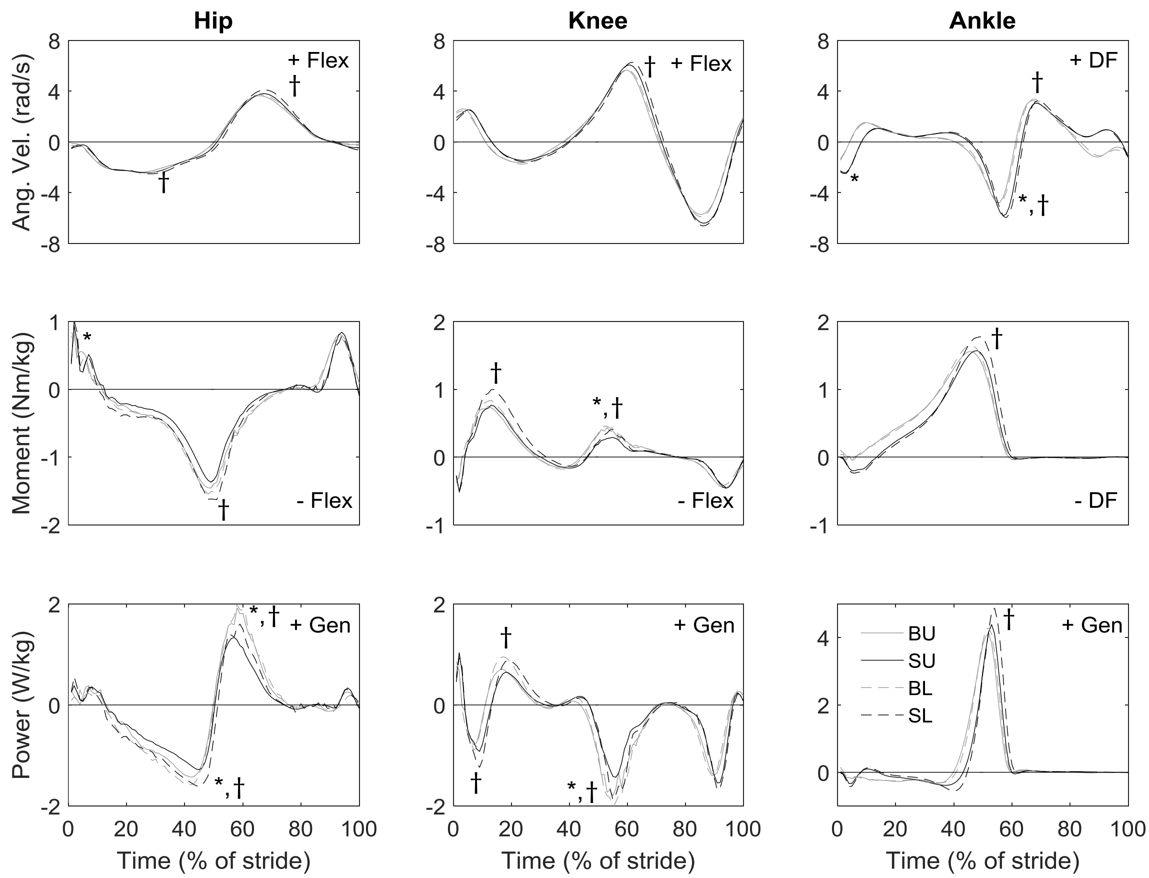

Figure 4.2. Hip (first column), knee (second column), and ankle (third column) angular velocity (first row), moment (second row), and power (third row). Data are presented as a group mean for all conditions. Positive = extension velocity, extensor moment, and power generation. * indicates significant footwear effect and † indicates significant load effect ($p < .05$).

Combined Loading & Footwear

While simultaneously increasing mass and removing the shoe, the expected increases with load and decreases while barefoot were offsetting for the braking and propulsive GRFs, double support time, and ROM at the hip. These counteracting responses resulted in no difference from the shod, unloaded condition for these measures (i.e., BL = SU). The only footwear by load interaction was observed for hip ROM ($p = .016$). Hip ROM increased when load was added while walking shod, but when load was added while walking barefoot, hip ROM did not change.

Table 4.1

Spatiotemporal parameters (means ± SD)

	BU	SU	BL	SL
Stride Length (m)*	1.48 (0.12)	1.60 (0.07)	1.45 (0.09)	1.59 (0.07)
Stance Time (s)*,†	0.56 (0.02)	0.59 (0.03)	0.57 (0.02)	0.60 (0.03)
Swing Time (s)*	0.40 (0.05)	0.41 (0.01)	0.38 (0.02)	0.40 (0.02)
Double Support Time (s)*,†	0.16 (0.02)	0.18 (0.03)	0.18 (0.02)	0.20 (0.03)

Note: * indicates significant footwear effect and † indicates significant load effect ($p <$ .05).

Table 4.2

Joint range of motion (means ± SD)

	BU	SU	BL	SL
Ankle	29.68 (6.92)	29.28 (10.80)	30.33 (6.24)	33.22 (6.57)
Knee*,†	56.67 (4.51)	61.03 (5.64)	57.00 (4.00)	62.19 (5.14)
Hip*,†,**	46.10 (4.91)	47.36 (4.59)	47.29 (5.58)	49.61 (5.00)

Note: * indicates significant footwear effect, † indicates significant load effect, and ** indicates interaction between loading and footwear ($p < .05$).

Discussion

The purpose of this study was to investigate the influences of footwear and increased mass on overground walking mechanics in young, healthy individuals. In general, walking without shoes reduced peak AP GRFs, lower extremity joint ranges of motion, and spatiotemporal parameters, while loading tended to have the opposite effect on these measures. Control of the lower limb in response to footwear conditions was dominated by the hip and knee joints, while loading impacted hip, knee, and ankle kinetic parameters.

Loading Effects

The 15% body mass load imposed in the present study resulted in GRF increases of 9.6-16.5%, which is consistent with the expectation that GRFs increase nearly in proportion to the added load (Birrell et al., 2007; Tilbury-Davis & Hooper, 1999). With 15% body weight loads, Quesada et al. (2000) reported a 94.5% increase in knee extensor moments in early stance, a 15.3% increase in hip flexor moments in late stance, and a 27.2% increase in ankle plantar flexor moments just before toe-off, compared to the unloaded condition. In the present study, the early stance extensor moment increased in the loaded condition by 5.86% at the hip and 17.78% at the knee, with an 11.15% increase in the plantar flexor moment. These relative changes differ in magnitude to those of Quesada et al. (2000), but both our data and theirs suggest the largest increase in joint moments with load occurs at the knee, next largest at the ankle, and lowest at the hip. The disagreement in the relative increases of these lower extremity joint moments could be related to walking speed, as participants in Quesada et al. (2000) walked at 1.67 $m{\cdot}s^{-1}$, compared to the 1.5 $m{\cdot}s^{-1}$ speed imposed in our study.

Footwear Effects

Peak vertical GRFs were not different between footwear conditions, while braking and propulsive forces were smaller while barefoot. Thus, our participants attempted to minimize the shear component of the GRF (i.e., braking and propulsive forces) in the barefoot condition, rather than the vertical, compressive forces. At slower walking speeds, this may not be the case, as Zhang et al. (2013) reported increased propulsive GRFs and a trend of increased braking forces while barefoot, compared to shod, at a 1.3 $m{\cdot}s^{-1}$ walking speed. It is possible that differences in walking speed (1.3 vs.

1.5 m$\cdot$s^{-1}) produced these opposing responses, as the subject characteristics between studies are similar. Keenan et al. (2011) observed a decrease in braking GRFs and increased propulsive GRFs while barefoot, also walking ~1.3 m$\cdot$s^{-1}. However, their participants walked on a treadmill rather than overground.

Extensor moments at the hip joint in early stance and the knee during late stance were greater while barefoot. This partially agrees with Keenan et al. (2011), who reported larger knee moments, but smaller hip moments while barefoot. However, those authors also did not find significantly different sagittal plane ankle moments. In older adults (mean age 59 years), a decreased knee extensor moment while barefoot, but no differences in hip moments, were reported (Shakoor & Block, 2006). At the other end of the age spectrum, children 7-10 years old decreased knee flexor moments during early stance and increased plantar flexor moments during late stance without shoes (Oeffinger et al., 1999). These conflicting results could be a result of differing participant characteristics and/or methodologies, as noted with the GRF data above (i.e., walking speed differences, treadmill vs. overground).

While barefoot, there was greater hip and knee power absorption in the latter half of stance (40-60% of stride) and hip power generation just before toe-off (~60% of stride). Thus, it seems the hip joint is important for control of the lower limb during stance when the shoes are removed. Oeffinger et al. (1999) also observed greater knee power absorption during stance while barefoot, but observed both ankle power generation and absorption differences in footwear conditions, which we did not detect. However, we observed significant kinetic differences at the hip, while they did not. These differences are likely due to walking at a preferred speed (Oeffinger et al., 1999) vs. the fixed, slightly challenging speed imposed here.

The unique responses to barefoot walking observed in children (Oeffinger et al., 1999), young adults (as in the present study), and older adults (Shakoor & Block, 2006) could suggest that lower extremity kinetic adaptations to footwear are dependent, in part, on age.

Combined Loading and Footwear Effects

The kinematic results of this study are consistent with those previously reported in steady-state treadmill walking (Dames & Smith, 2015) , except for the ROM data. In our previous study, hip ROM decreased while barefoot, and a footwear*load interaction was found for ankle ROM, as ankle ROM decreased with load while barefoot but increased with load while shod. Here, however, no ankle ROM effect was found for footwear or load, but knee ROM decreased while barefoot and increased with load, and a footwear*load interaction was found for hip ROM. It appears that these footwear and loading conditions imposed on overground walking promote altered ROM as compared to treadmill walking, even at comparable speeds. Differences exist in kinematic measures obtained from treadmill vs. overground (Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007), but these are reasonably small, and comparing the effects of footwear and loading between treadmill and overground conditions was not the purpose of this study. Furthermore, spatiotemporal differences with load were not observed in our previous study, but were observed here, so it is not surprising to see other kinematic differences between these studies. As opposed to the compliant treadmill used in our previous study, the discomfort associated with walking overground on a hard surface likely promoted the spatiotemporal and kinematic differences observed here. Future work will compare the impact of these footwear conditions between groups who differ in body mass (i.e., obese vs. non-obese).

One limitation of this study that is worth mentioning is the slightly different position of the heel and second metatarsal head markers between barefoot and shod conditions. Participants provided their own athletic shoes, thus we could not place markers directly on the skin by cutting into the shoe material. These differences may have impacted the kinematic data, and subsequent joint moment and power estimates. However, ankle joint motion from shoe-mounted and skin-mounted markers have been shown to have a high level of agreement, with coefficients of multiple correlation in all three planes ≥0.974 (Sinclair, Taylor, Hebron, & Chockalingam, 2014).

Conclusion

Our hypothesis that spatiotemporal measures and joint ROM would decrease while barefoot was supported, but our expectation for joint moments to decrease while barefoot was not supported. As hypothesized, loading increased longer stance and double support times, greater hip and knee ROM, and GRFs were observed. Finally, braking and propulsive forces increased with load, but decreased while barefoot, resulting in similar magnitudes of these measures between the shod, unloaded condition and the barefoot, loaded condition. This supports our third hypothesis, that the expected increase with loading and decrease while barefoot would offset one another. In general, lower extremity kinetic responses to load carriage were observed at the ankle, knee, and hip joints, whereas footwear responses were found only at the knee and hip.

CHAPTER V

STUDY 3: BAREFOOT VERSUS SHOD: EFFECTS OF OBESITY ON WALKING MECHANICS

Introduction

Obesity has become a major health issue in the United States. Data from a recent CDC report estimates that 72.5 million American adults are obese, with approximately $147 billion per year spent in obesity related health care costs (Hootman et al., 2011). Musculoskeletal issues associated with obesity include knee and hip joint replacements, general pain of the low back and neck (Patterson et al., 2004) and arthritis (Hootman et al., 2011). Increased joint loads are suggested to contribute to the greater prevalence of osteoarthritis in obese individuals (Felson, 1996; Hootman et al., 2011). Weight loss is suggested as a means to decrease these joint loads (Messier, Gutekunst, Davis, & DeVita, 2005).

Increased body mass promotes walking patterns that differ from average weight adults. In comparison to non-obese individuals walking at a similar speed, obese individuals use longer stance and double support times (Browning & Kram, 2007; Ranavolo et al., 2013), increased step widths (Ranavolo et al., 2013), and decreased swing times (DeVita & Hortobagyi, 2003). In terms of joint kinematics, obese individuals generally have a more extended leg at initial foot contact and throughout stance (DeVita & Hortobagyi, 2003; McMillan et al., 2010). Increasing body mass also leads to different kinetic profiles that accompany changes in the spatiotemporal and kinematic patterns.

Absolute peak vertical, anteroposterior, and mediolateral GRFs increase with obesity, with the increase in vertical and anteroposterior GRFs nearly proportional to the increase in total mass (Browning & Kram, 2007). The impact of obesity on joint kinetics is less clear as some have reported increased joint moment magnitudes at the knee (Browning & Kram, 2007), whereas others have reported decreased joint moment magnitudes at the knee (DeVita & Hortobagyi, 2003) or no difference compared to non-obese individuals (Lai et al. (2008). Thus, further investigation of joint kinetics in an obese population is needed.

Previous contradictory joint kinetic outcomes might be due to differences in methodology, such as the use of different walking speeds and footwear, and/or normalization methods. Participants walked barefoot in two of these studies (Lai et al., 2008; Ranavolo et al., 2013), and shod in two others (Browning & Kram, 2007; DeVita & Hortobagyi, 2003), while McMillan et al. (2010) did not report a footwear condition. Ranavolo et al. (2013) did not control walking speed but instructed the normal weight adults to walk slowly, while the obese adults were told to walk naturally. No significant difference in speed between groups was reported, but these conflicting instructions may have impacted walking strategies. Finally, Browning and Kram (2007) were interested in absolute joint moments and GRFs to understand the loads experienced by the lower extremity. DeVita and Hortobagyi (2003) reported moments normalized to mass, and others normalized to mass and height (Lai et al., 2008; McMillan et al., 2010).

In summary, contradictory outcomes from previous gait studies of overweight individuals were likely due to methodological differences (e.g., normalization approaches), footwear conditions, and gait speed disparities. The present study was designed to examine two footwear conditions (barefoot versus shod), control gait speed, and explore the influence of normalization approaches. The primary purpose of the present study was to compare the influence of footwear between normal weight and obese individuals with a fixed gait speed. It was hypothesized that larger peak ground reaction forces, joint moments, and joint powers, regardless of footwear, would be observed in overweight/obese individuals compared with healthy weight individuals. Additionally, it was hypothesized that, regardless of body weight, barefoot walking would lead to reduced ground reaction forces, joint moments, and joint powers compared with shod walking. A secondary purpose of this study was to explore the influence of normalization on study outcomes.

Methods

Participants

Twelve young, healthy-weight individuals (5 women, 7 men) with no known musculoskeletal or neurological issues were recruited as controls for this study. Ten obese individuals who had a body mass index (BMI) between 30 and 40 $kg{\cdot}m^{-2}$ were also recruited for this study. Besides being obese, this group was otherwise healthy. Participant characteristics can be found in Table 5.1. There was no difference in height or age of these two groups. To differentiate between those who were truly obese vs. those whose muscular build may lead to large BMIs, a waist circumference >100 cm for men, or >90 for women, provided a secondary means for placement in the Obese group. These circumference criteria place an individual in the High Risk for disease category, based on published guidelines from the American College of Sports Medicine (*ACSM's Guidelines for Exercise Testing and Prescription*, 2010). The university's Institutional Review Board approved this study and all participants provided informed written consent prior to participation.

Table 5.1

Participant characteristics. Mean ± SD

Group	**Age (years)**	**Height (m)**	**Mass (kg)***	**BMI (kg·m^{-2})***
Non-Obese	23 (3)	1.73 (0.11)	70.90 (12.67)	23.55 (2.09)
Obese	26 (3)	1.79 (0.10)	108.46 (13.25)	33.75 (2.91)

Note: * indicates a significant difference between groups. $p < .001$

Data Collection

Anthropometric data (including body mass and height) were measured for use in VICON's full body plug-in-gait model. Retroreflective markers were attached to the appropriate anatomical landmarks for the same model using double-sided tape. Participants then performed overground walking trials at 1.5 m·s^{-1} ±5% while barefoot and shod. Two pairs of timing gates (BROWER Timing Systems, Draper, UT) spaced approximately 5 m apart were used to capture walking speed. Participants wore their own athletic shoes for the shod conditions (Non-Obese group = 272 ± 68 g; Obese group = 321 ± 90 g). The shoe mass was not significantly different between groups. The order of conditions was individually randomized. During each trial, 3D motion (100 Hz) (VICON, Englewood, CO) and ground reaction force (GRF) (2000 Hz) data were collected. GRFs were measured using a tandem-belt instrumented treadmill (AMTI, Watertown, MA) embedded in the center of the walkway with 2 individual force plates. Trials included in the data analysis were within the expected velocity range and clean foot contacts were made with the force plates (i.e., a single, whole foot contact on each force plate).

Data Analysis

Markers were labeled with VICON Nexus, but all subsequent processing of data was performed using a custom Visual 3D (C-Motion, Germantown, MD) script. Marker data were filtered using a recursive, digital Butterworth lowpass filter (f_c = 6 Hz). This cutoff frequency was confirmed by a residual analysis as described by Winter (2009). GRF data were filtered using a recursive, digital Butterworth lowpass filter (f_c = 50 Hz). Motion and GRF data were combined through inverse dynamics to estimate joint reaction forces and moments for the ankle, knee, and hip in the sagittal plane. Joint moment peaks were selected based on the description by Winter (1987). These included two ankle peaks, five knee peaks, and three hip peaks. The ankle moments include a dorsiflexor peak during weight acceptance and late stance plantarflexor peak. Knee moment peaks selected included the weight acceptance flexor, early stance extensor, midstance flexor, late stance extensor, and late swing flexor peaks. Hip moment peaks included the early stance extensor, late stance flexor, and late swing extensor peaks. Joint powers were calculated as the product of the joint moment and angular velocity. Joint power peaks from the phases defined by Winter (1987) were used in statistical analyses. These included two ankle phases (A1, A2), four knee phases (K1-K4), and three hip phases (H1-H3). A1 is the initial weight acceptance and A2 the propulsive peak at toe-off. K1 is the energy absorption phase during weight acceptance. K2 is the only power generation phase and occurs during mid-stance. K3 is power absorption during terminal stance and early swing phase. The K4 phase is terminal swing power absorption. H1 phase occurs during early stance, H2 is an absorption phase during mid-stance, and H3 a power generation phase prior to toe-off.

Spatiotemporal and kinematic dependent measures were also identified. Spatiotemporal dependent variables included stride length, stance time, swing time, and double support time. Kinematic variables included angular ranges of motion (ROM) and joint angles at initial contact for the hip, knee, and ankle.

Statistical Analysis

Dependent variables were determined from three successful strides and then averaged. A series of 2 x 2 (BMI, footwear) ANOVAs with repeated measures were performed using IBM SPSS 23.0 (Armonk, NY). Grouping based on BMI represented a between subjects factor (Obese, Non-Obese) and footwear a within subjects factor (barefoot, shod). The probability associated with a Type I error was set at .05 for all tests. Where a significant difference was detected, a percent increase was presented as the larger mean minus the smaller mean divided by the smaller mean, and a percent decrease was presented as the smaller mean minus the larger mean divided by the larger mean.

Results

Results are divided into four sections: Footwear, Group, Footwear*Group interactions, and the impact of Normalization. Walking barefoot decreased spatiotemporal measures, knee and hip ranges of motion, and GRFs. Obese individuals had increased stride length and stance time, GRFs, and absolute joint moments and powers. However, Obese and Non-Obese groups did not respond the same to footwear conditions.

Footwear

Data in this section are collapsed across Obese and Non-Obese groups. Walking barefoot was accomplished with shorter stride lengths ($F = 32.616$, $p < .001$), stance times (F = 82.022, $p < .001$) and double support times (F = 54.851, $p < .001$) (Table 5.2). At initial contact, a significant difference in ankle angle was observed (F = 25.959, $p < .001$) (Figure 5.1). The ankle was plantarflexed at contact while barefoot, but dorsiflexed while shod. The knee (F = 12.304, $p = .002$) and hip (F = 5.486, $p = .030$) joints were more flexed at contact while barefoot. Ankle ROM between footwear conditions did not differ (Table 5.3), but knee (F = 18.627, $p < .001$) and hip (F = 14.924, $p = .001$) ROMs were smaller while barefoot.

Table 5.2

Spatiotemporal measures. Mean ± SD

	Non-Obese		**Obese**	
	Barefoot	**Shod**	**Barefoot**	**Shod**
Stride Length (m)*	1.48 (0.12)	1.60 (0.07)	1.55 (0.10)	1.63 (0.09)
Stance Time (s)*,†	0.56 (0.02)	0.59 (0.03)	0.60 (0.04)	0.65 (0.04)
Swing Time (s)	0.40 (0.05)	0.41 (0.01)	0.40 (0.02)	0.40 (0.03)
Double Support Time (s)*,†,**	0.16 (0.02)	0.18 (0.03)	0.21 (0.03)	0.25 (0.04)

Note: * indicates significant footwear effect, † indicates significant group effect, and ** indicates a group*footwear interaction. $p < .05$

Table 5.3

Joint range of motion in degrees. Mean ± SD

	Non-Obese		**Obese**	
	Barefoot	**Shod**	**Barefoot**	**Shod**
Ankle	29.68 (6.92)	29.28 (10.80)	27.49 (6.13)	34.02 (4.34)
Knee*	56.67 (4.51)	61.03 (5.64)	52.90 (5.46)	59.56 (4.41)
Hip*	46.10 (4.91)	47.36 (4.59)	47.30 (5.54)	48.66 (4.96)

Note: * indicates significant footwear effect. $p < .05$

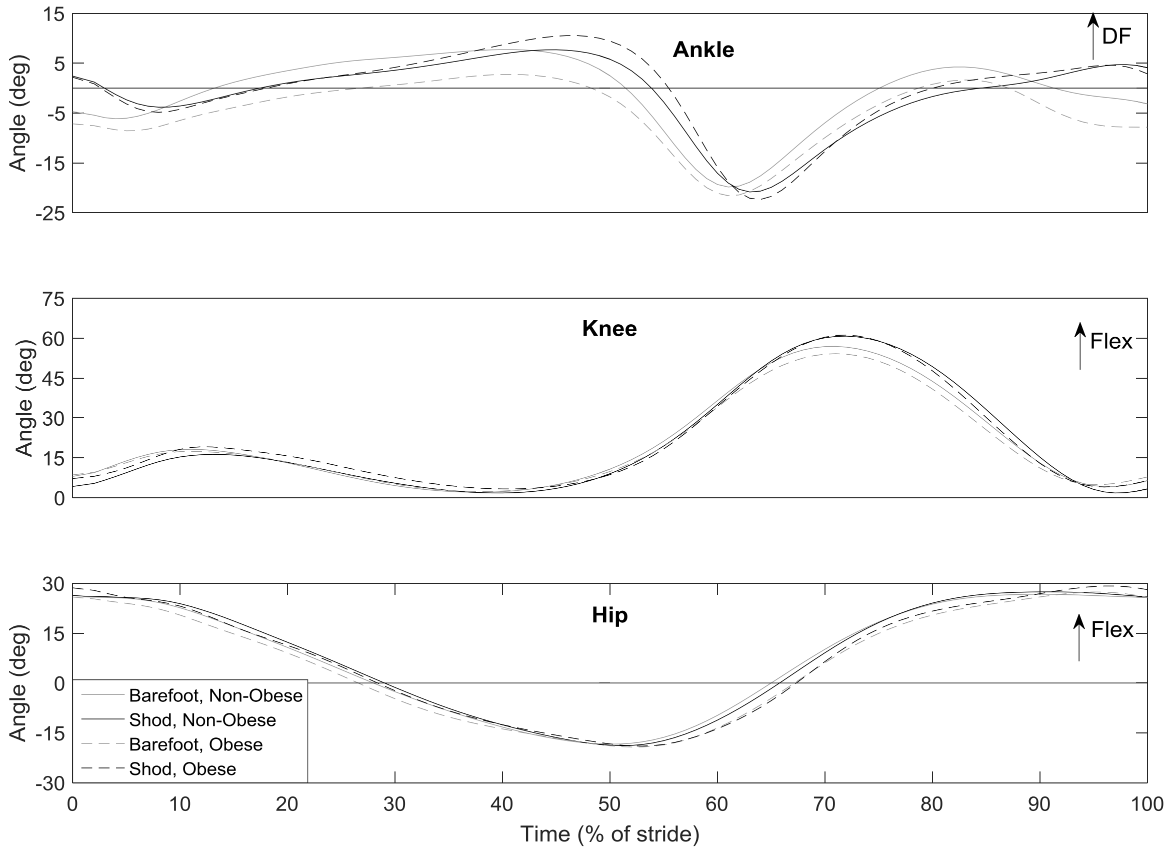

Figure 5.1. Group averages by condition for ankle (top), knee (middle), and hip (bottom) angular positions in degrees.

The magnitude and direction of responses to footwear conditions were similar across the normalized and absolute data for all peak GRF dependent measures, with *p*-values differing only slightly. Thus, the outcomes presented in text (F and *p*-values) are for the normalized data, but the relative changes between conditions apply to the absolute and normalized GRF measures. Walking barefoot reduced peak braking force by 10.91% ($F = 14.260$, $p = .001$) and propulsive force by 10.02% ($F = 35.090$, $p < .001$) (Figure 5.2). The initial vertical GRF peak was 2.76% smaller ($F = 11.119$, $p = .003$), but no differences were observed between footwear conditions for the second peak vertical GRF.

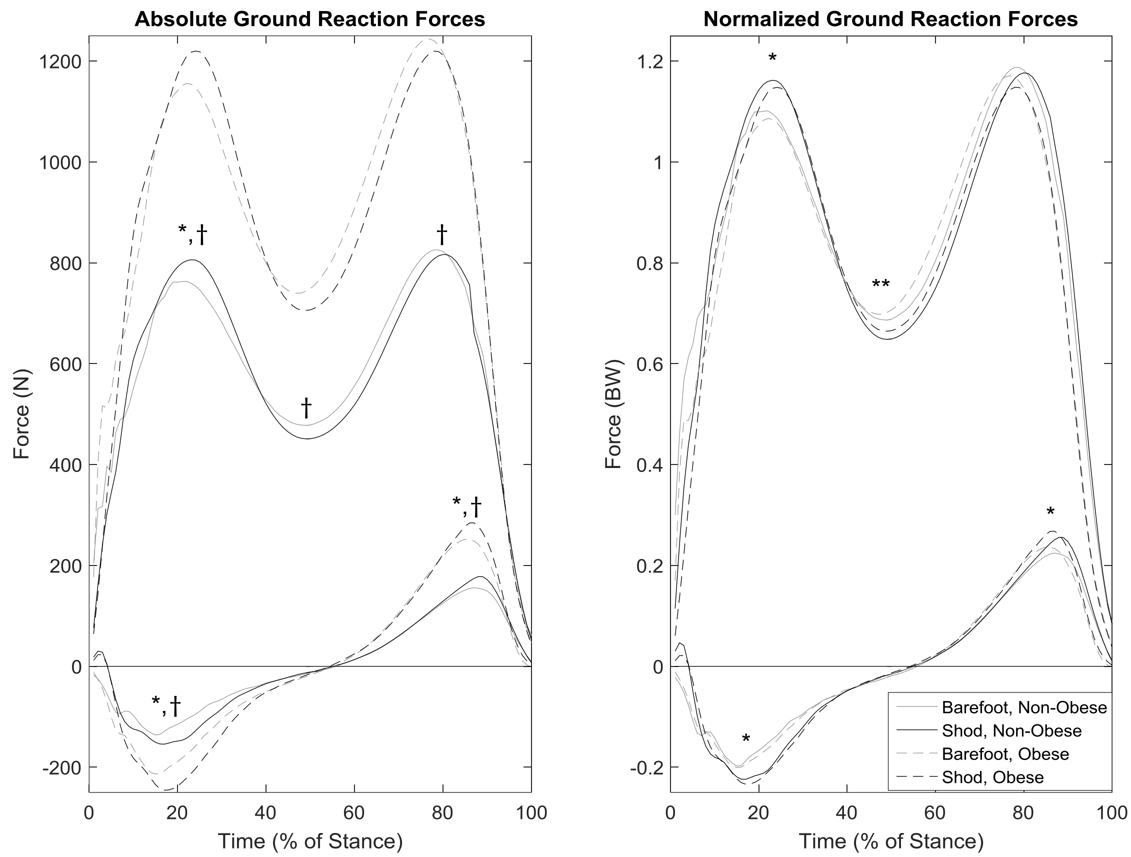

Figure 5.2. Group averages by condition for absolute (left) and normalized (right) ground reaction forces. * indicates significant footwear effect and † indicates significant group effect. $p < .05$

Walking barefoot increased hip extensor moments in early stance by ~26% in normalized (F = 7.939, $p = .011$) and absolute (F = 6.865, $p = .016$) units (Figure 5.3). Walking barefoot increased hip power generation around toe-off by ~30% in both normalized (F = 15.587, $p = .001$) and absolute (F = 10.321, $p = .005$) terms. All other main effects for joint moments and powers were coincident with Footwear*Group interactions, and are discussed in a following section that addresses interactions.

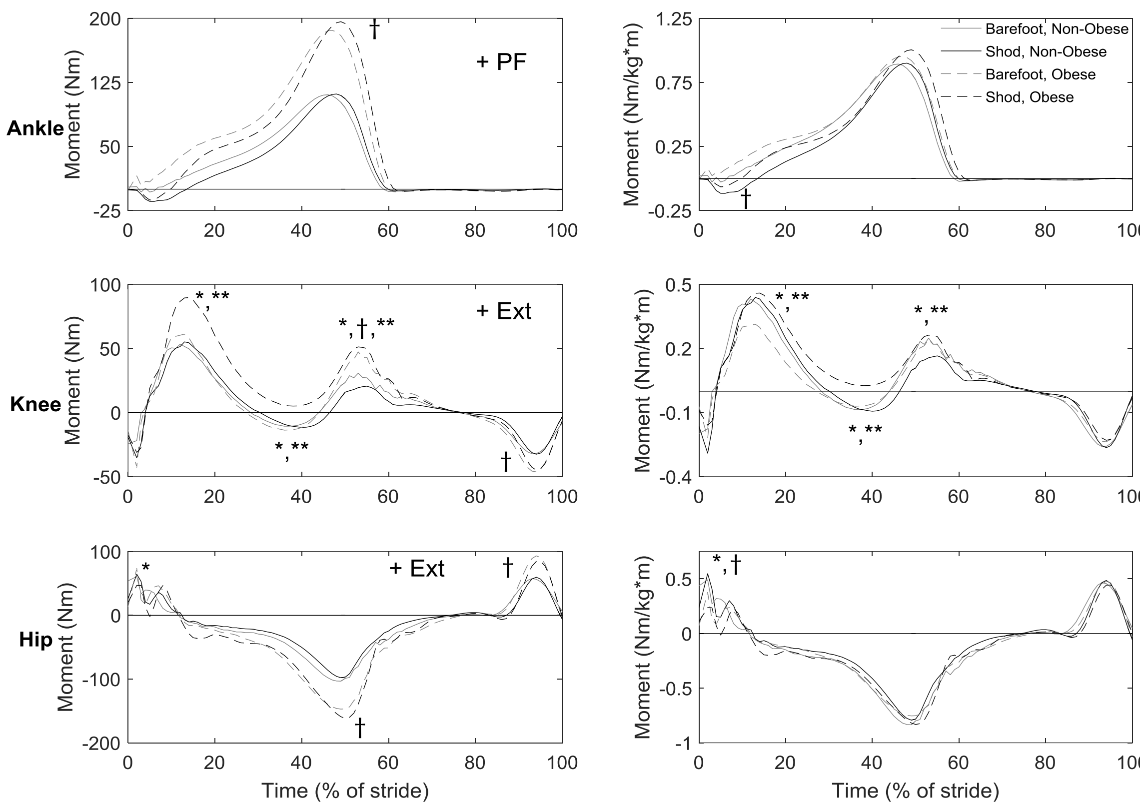

Figure 5.3. Group averages by condition for absolute (left column) and normalized (mass*height) (right column) joint moments. * indicates significant footwear effect, † indicates significant group effect, and ** indicates a group*footwear interaction. $p < .05$

Group

The Obese group spent more time in double support (F = 24.106, $p < .001$) and stance (F = 13.038, $p = .002$) than the Non-Obese group (Table 5.2). An interaction occurred for double support time, as the Obese group had a greater increase in this measure while shod. Range of motion and angle at initial contact of all lower extremity joints were similar between groups (Table 5.3).

The Obese group was 53% heavier than the Non-Obese group, and the differences in peak vertical GRFs were proportional to this extra mass when comparing absolute data. These increases were 53.11% and 48.29% for the initial and second vertical peaks, 63.54% for the minimum (between the peaks), and 57.64% and 59.11% for the propulsive and braking force peaks, respectively ($p < .001$ for all) (Figure 5.2). These differences disappeared when GRFs were normalized to body weight.

As a group, the Non-Obese tended to produce more hip power, but generated and absorbed less knee power during stance than the Obese group (Figure 5.4). That is, hip power absorption (H2) ($F = 5.413$, $p = .031$, absolute data) and generation (H3) ($F = 4.591$, $p = .045$, normalized data) were higher in the Non-Obese group, while the Obese group demonstrated greater knee power absorption (K3) ($F = 7.927$, $p = .011$, absolute data) and generation (K2) ($F = 5.141$, $p = .035$, absolute data). Other joint moment and power comparisons between groups are dependent on normalization and are presented in the section dedicated to comparison of normalized and absolute data outcomes.

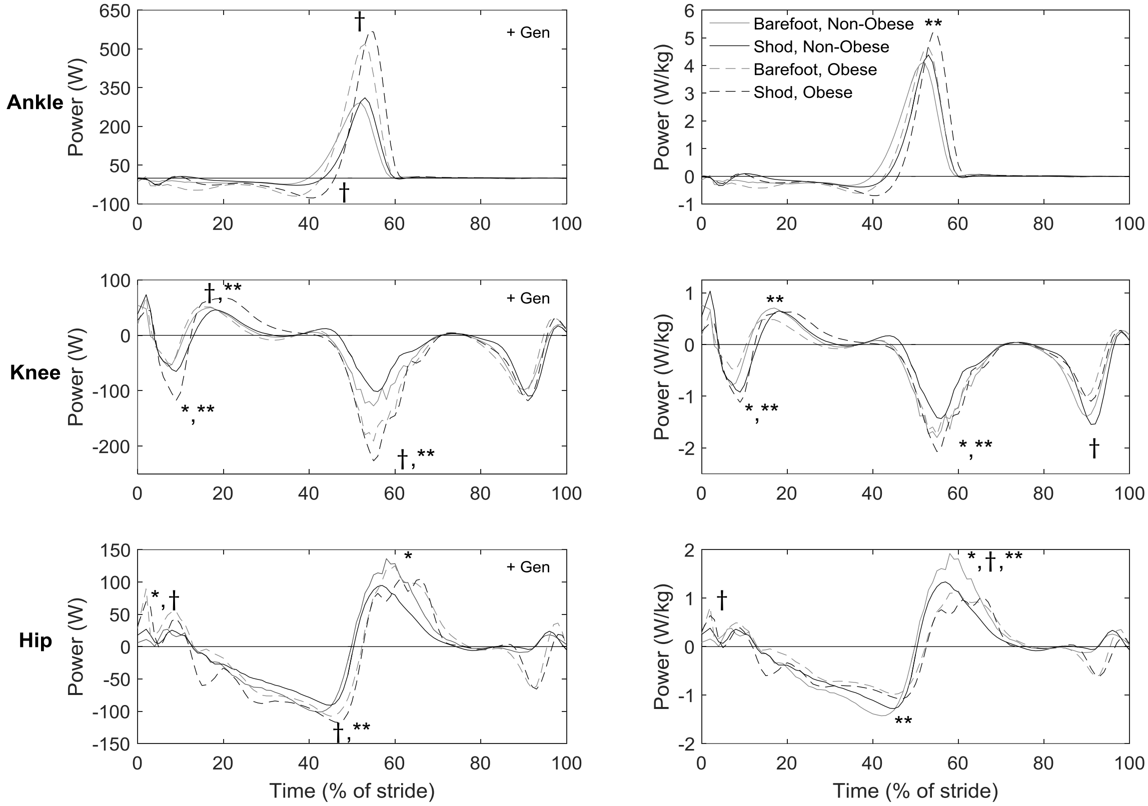

Figure 5.4. Group averages by condition for absolute (left column) and normalized (mass) (right column) joint powers. * indicates significant footwear effect, † indicates significant group effect, and ** indicates a group*footwear interaction. $p < .05$

Footwear*Group Interaction

The joint kinetic responses in the Obese and Non-Obese groups were different across footwear conditions. For the knee joint, interactions were seen in the peak moments and powers, whereas the ankle and hip joints only showed interactions in joint powers (Figure 5.5). Seven of the nine interactions were identified in both the absolute and normalized data. Two of nine (A2 and H3 peaks) interactions were only evident in normalized data. Thus, the emphasis in this section will be on the normalized data in order to address each interaction in the same units.

Three Group*Footwear interactions were observed in knee joint moment responses. First, an interaction was observed for the knee joint extensor moment in early stance (F = 4.441, p = .048). The Obese group reduced the magnitude of the early stance knee extensor moment magnitude by 40.02% while barefoot, but footwear had no impact in the Non-Obese group. Second, the Obese group never demonstrated flexor function during midstance while shod, but did exhibit flexor function during midstance while barefoot. In the Non-Obese group footwear had no impact on this measure, so this Footwear*Group interaction was significant (F = 6.507, p = .020) due to a shift in the control strategy in the Obese group when shoes were removed. Third, the Non-Obese group increased the peak late stance knee extensor moment by 51.60% (F = 9.978, p = .005) while walking barefoot, while this measure did not differ between footwear conditions for the Obese group.

For joint powers, the Non-Obese group generated more power at the ankle, knee, and hip in the barefoot condition than while shod. The Non-Obese group also absorbed more power at the knee and hip joints while barefoot than shod. In contrast, the Obese group generated less power at the ankle and knee while barefoot, but more at the hip while barefoot. The Obese group also absorbed less power at the knee and hip without shoes. This resulted in interactions for the A2 (F = 4.964, p = .038), K1 (F = 7.204, p = .014), K2 (F = 7.114, p = .015), K3 (F = 11.582, p = .003), H2 (F = 10.380, p = .004), and H3 (F = 6.738, p = .018) phase peaks.

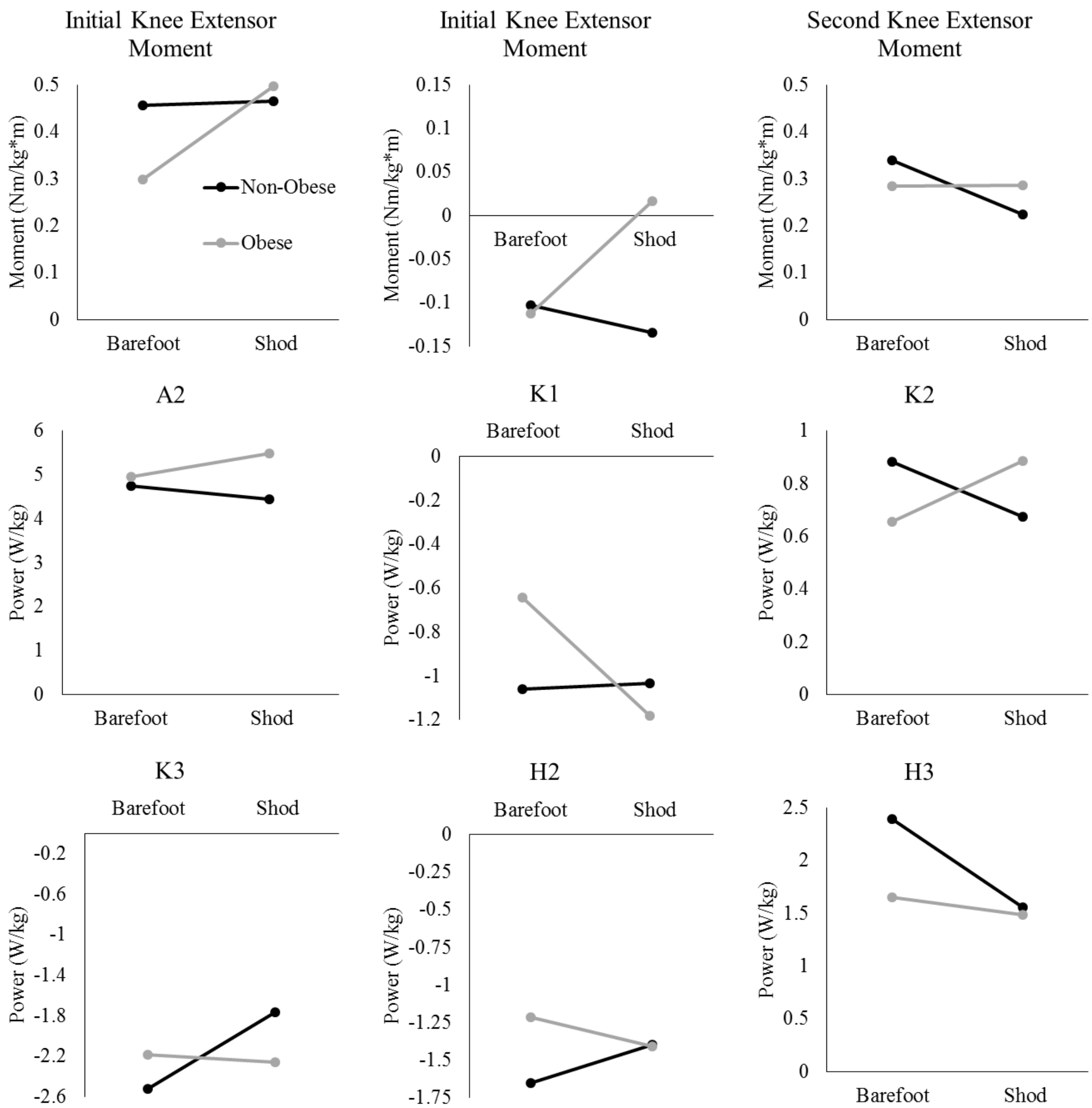

Figure 5.5. Joint kinetic measures where Footwear*Group interactions were observed.

Normalization

Statistical outcomes varied between normalized and absolute data for GRFs (Figure 5.2), joint moments (Figure 5.3), and joint powers (Figure 5.4). The Obese group was 53% heavier than the Non-Obese group, with an average product of height and mass 57.75% larger than the Non-Obese group. For several kinetic variables, a Group main effect was observed in the absolute data, but no differences were observed after normalization. This was the case for the peak vertical and anteroposterior GRFs (normalization metric = body weight), peak joint moments at the hip, knee and ankle (normalization metric = mass*height), and joint powers at the hip, knee, and ankle (normalization metric = body mass) (Table 5.4). With the exception of the ankle and hip power absorption and ankle power generation, the Obese group increased these kinetic measures approximately equal to their larger size relative to the Non-Obese group. When data were normalized, four dependent measures were larger in the Non-Obese group than the Obese group (Table 5.5). However, these measures were not different between groups in absolute terms.

Table 5.4

DVs that had a Group main effect only in the absolute data

DV	Footwear	Obese	Non-Obese	% Diff.	*p* - value
GRFs (N)					
$VGRF_1$	Shod	1230.09 (153.29)	798.91 (142.07)	53.11	< .001
	Barefoot	1186.04 (173.18)	779.15 (126.83)		
$VGRF_2$	Shod	1221.64 (153.32)	837.84 (198.24)	48.29	< .001
	Barefoot	1249.96 (154.46)	828.92 (151.26)		
Braking	Shod	-252.04 (16.11)	-158.04 (28.75)	59.11	< .001
	Barefoot	-227.61 (39.18)	-143.41 (27.41)		
Propulsive	Shod	287.47 (37.66)	183.72 (40.49)	57.64	< .001
	Barefoot	260.10 (40.67)	163.64 (34.28)		
Joint Moments (Nm)					
Plantarflexor	Shod	197.51 (25.20)	139.13 (83.15)	51.72	.001
	Barefoot	187.44 (61.91)	114.59 (31.87)		
Knee Extensor	Shod	55.95 (22.90)	27.68 (11.57)	61.40	.008
	Barefoot	55.57 (21.07)	41.41 (14.78)		
Knee Flexor	Shod	-46.52 (6.79)	-33.24 (9.24)	42.72	< .001
	Barefoot	-48.04 (6.54)	-33.02 (7.95)		
Hip Flexor	Shod	-160.87 (57.42)	-101.77 (35.11)	48.03	.009
	Barefoot	-151.11 (47.10)	-108.99 (28.67)		
Hip Extensor	Shod	90.39 (20.50)	62.21 (31.57)	52.41	.004
	Barefoot	99.60 (22.50)	62.44 (22.49)		
Joint Powers (W)					
Ankle Abs. (A1)	Shod	-87.26 (52.17)	-43.48 (26.86)	89.91	.005
	Barefoot	-93.43 (46.33)	-51.67 (35.98)		
Ankle Gen. (A2)	Shod	601.18 (151.27)	313.01 (102.62)	77.27	< .001
	Barefoot	550.47 (215.30)	336.64 (104.60)		
Knee Gen. (K2)	Shod	93.18 (45.43)	47.21 (23.30)	50.01	.035
	Barefoot	70.16 (37.86)	61.68 (30.78)		
Knee Abs. (K3)	Shod	-247.15 (88.99)	-126.21 (43.22)	58.56	.011
	Barefoot	-238.71 (98.41)	-180.23 (67.24)		
Hip Abs. (H2)	Shod	-150.36 (32.32)	-99.15 (34.44)	30.40	.031
	Barefoot	-131.30 (45.51)	-116.86 (30.97)		

Note: % Diff. = relative increase in measure in the Obese group compared to the Non-Obese group, collapsed across Footwear. $VGRF_1$ = initial vertical GRF peak, $VGRF_2$ = second vertical GRF peak, Knee Extensor = late stance extensor peak (~50% of stride), Knee Flexor = late swing peak (~90% of stride), Hip Extensor = late swing peak (~90% of stride), Abs. = absorption, Gen. = generation. *p* - value is for Group differences, collapsed across Footwear. $p < .05$

Table 5.5

DVs that had a Group main effect only in the normalized data

DV	Footwear	Obese	Non-Obese	% Diff.	*p* - value
Joint Moments (Nm/kg*m)					
Dorsiflexor	Shod	-0.08 (0.03)	-0.13 (0.09)	-49.81	.011
	Barefoot	-0.04 (0.06)	-0.11 (0.08)		
Hip Extensor	Shod	0.39 (0.08)	0.63 (0.12)	-35.68	< .001
	Barefoot	0.51 (0.19)	0.78 (0.24)		
Joint Powers (W/kg)					
Knee Abs. (K4)	Shod	-1.18 (0.29)	-1.67 (0.31)	-20.75	< .001
	Barefoot	-1.09 (0.31)	-1.55 (0.19)		
Hip Gen. (H3)	Shod	1.48 (0.62)	1.56 (0.26)	-29.51	.045
	Barefoot	1.65 (0.65)	2.39 (0.44)		

Note: % Diff. = relative decrease in measure in the Obese group compared to the Non-Obese group, collapsed across Footwear. Hip Extensor = early stance peak. *p* - value is for Group differences, collapsed across Footwear. $p < .05$

Discussion

The purpose of this study was to investigate the impact of footwear and obesity on walking mechanics in young adults. In general, kinematic responses to footwear were similar regardless of mass. Across the Obese and Non-Obese groups, walking barefoot promoted a plantarflexed ankle angle and greater knee and hip flexion at initial contact. These kinematic differences were associated with reduced GRFs while walking barefoot. However, joint kinetic responses to barefoot and shod conditions were not consistent across groups. Specifically, Footwear*Group interactions were observed for peak knee extensor and flexor moments, and ankle, knee, and hip joint powers.

Footwear

Walking barefoot produced spatiotemporal responses consistent with the literature (Keenan et al., 2011; Shakoor & Block, 2006; Zhang et al., 2013). We observed shorter strides, stance times, and double support times regardless of Group when walking barefoot compared to shod. It is commonly suggested that discomfort at initial contact without the cushion of a shoe promotes these altered spatiotemporal measures. Supporting this idea, the hip and knee joints were more flexed and the ankle was in a plantarflexed position at contact during the barefoot condition, which likely represents an attempt to avoid heel contact. This ankle position resulted in a lack of dorsiflexor moment in the Obese group for the barefoot condition (Figure 4.3). However, the difference in magnitudes of the early stance ankle moments was not significant. A similar kinematic response at the ankle and knee during weight acceptance has been reported elsewhere (Zhang et al., 2013). Additionally, the initial vertical GRF, and braking and propulsive AP GRFs were reduced while walking barefoot. Collectively, this suggests both Obese and Non-Obese groups adopted a lower extremity posture that reduced compressive and shear forces during stance. Other authors have reported increased anteroposterior forces while walking barefoot at ~1.3 $m \cdot s^{-1}$ (Keenan et al., 2011; Zhang et al., 2013). Zhang et al. (2013) suggested the heel-toe height difference in a standard shoe explain the decreased the propulsive forces produced to maintain speed while shod. Differences in these kinetic outcomes could be due to some participants walking overground, as in the present study and also Zhang et al. (2013), versus on a treadmill (Keenan et al., 2011).

Walking barefoot did not influence ankle moments, but did reduce knee extensor moments during early stance and increase knee extensor moments in late stance. Early stance hip extensor moments also increased while barefoot. Thus, the segment directly impacted by altering footwear (i.e., the foot) was the only segment whose joint moments were not impacted by footwear. Keenan et al. (2011) observed similar ankle moment responses between barefoot and shod conditions. However, the proximal joint data from their sample differs from ours. Keenan et al. (2011) reported a reduction in hip extensor and flexor moments and an increase in knee flexor moments. Our participants walked overground at 1.5 $m \cdot s^{-1}$, while participants in Keenan et al. (2011) walked on a treadmill set to their preferred velocity (average = 1.28 $m \cdot s^{-1}$). The differing outcomes may be related to known kinematic and kinetic differences between overground and treadmill walking (Lee & Hidler, 2008; Riley et al., 2007).

Perhaps the most important outcome is that the Obese group adopted a similar neuromuscular control pattern to the Non-Obese group while barefoot, but not while shod. Absolute peak knee extensor moments in early stance were similar between the Obese group while barefoot and Non-Obese group, while in the normalized data this measure was actually smaller in the Obese group. During midstance, the Non-Obese group demonstrated flexor moments while barefoot and while shod, but the Obese group only experienced flexor function in the barefoot condition. This difference was not reflected in the power curves because the angular velocity of the knee at this point is close to zero. Figure 5.6 shows the knee joint angular velocity curves for one representative Obese participant in the barefoot and shod conditions.

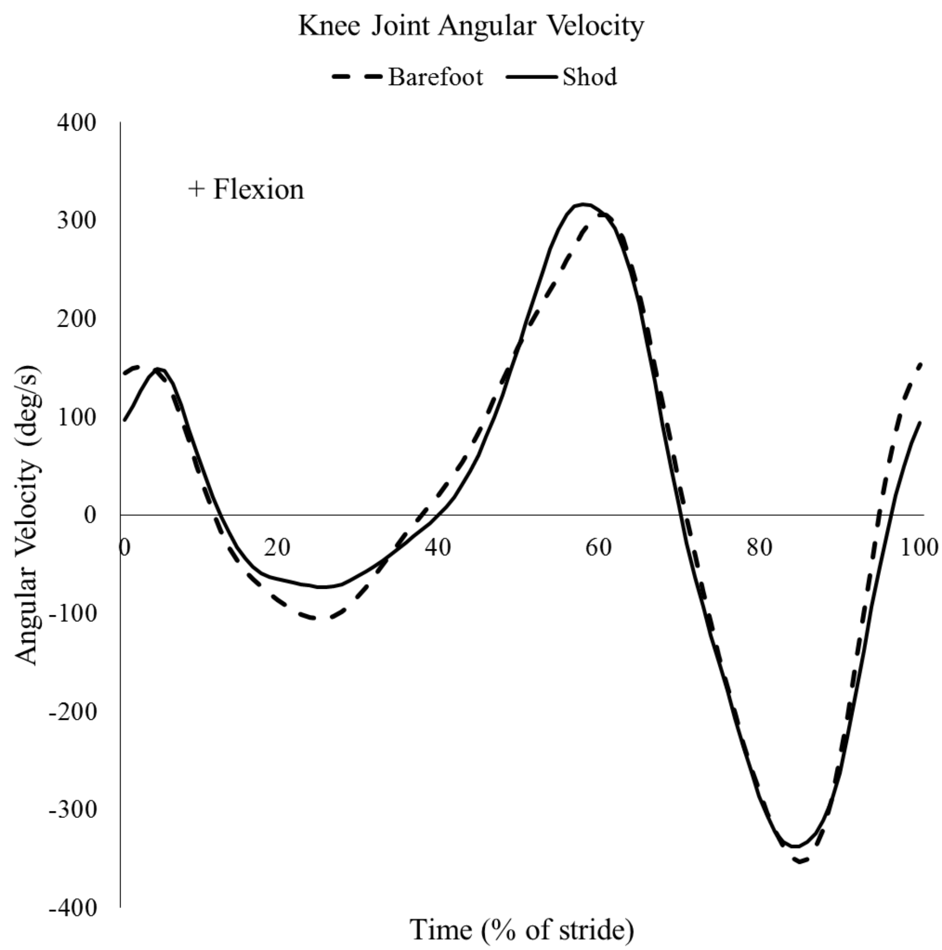

Figure 5.6. Knee joint angular velocity curves of one participant.

Removing cushion and support from under the foot in the Obese group promoted walking mechanics more similar to the Non-Obese group by decreasing joint moment magnitudes and altering knee kinetic patterns. These adaptations could be beneficial in reducing knee joint damage, as high joint loads are thought to contribute to joint degradation and development of osteoarthritis (Anandacoomarasamy et al., 2008). However, it is common practice for obese persons to wear thickly cushioned, heavily supportive shoes for physical activity. Not wearing shoes during activity might offset some of the additional load on the joint, which over the long term could help reduce the incidence of joint pain which could discourage physical activity. The obese adults in this sample did not experience pain with physical activity, so further study would be needed to understand the role of footwear in a cohort of obese adults who do experience pain during walking.

The Obese group absorbed less power at the knee in early stance in the barefoot condition than the shod condition, and was also lower than that of either footwear condition for the Non-Obese group regardless of normalization. Walking barefoot also decreased knee joint power generation and absorption later in stance, as well as hip power absorption during late stance in the Obese group. Walking barefoot increased hip power generation during late stance and early swing regardless of body mass. In a younger sample (7-10 years) walking at their preferred speed, walking barefoot decreased ankle power absorption but increased ankle power generation (Oeffinger et al., 1999). Oeffinger et al. (1999) also found walking without shoes decreased both knee power generation during early stance and knee power absorption during late stance, which agrees with our data. However, Oeffinger et al. (1999) did not observe any hip power changes between footwear conditions, which we did find. Thus, walking barefoot seems to promote consistent decreases in knee power peaks, but ankle and hip joints adaptations may be dependent on age and/or walking speed.

Group

The Obese group spent a longer time in stance and in double support than the Non-Obese group, which agrees with previous findings (Browning & Kram, 2007; Malatesta et al., 2009; Ranavolo et al., 2013). There were no kinematic differences between groups at initial contact or for range of motion across ankle, knee, and hip joints. In a sample of more obese adults (BMI range = 32.4-58.7 $kg \cdot m^{-2}$), initial contact was made with a more extended limb, which remained straighter until toe-off, compared to the non-obese cohort (DeVita & Hortobagyi, 2003). It is possible kinematic differences are not observed until BMI reaches a threshold above that of our sample (BMI range = 30.19 – 40 $kg \cdot m^{-2}$). Because the kinematic responses were similar between groups in the present study, differences in the GRF vector magnitude likely produced the observed changes in joint kinetic measures between groups.

In the absolute joint power data, the Obese group generated more power at the ankle and knee, and absorbed more power at the ankle, knee, and hip joints than the Non-Obese group. In children, no differences in ankle or knee powers (W/kg) were detected between normal and obese groups, while the latter produced less power and absorbed more power at the hip (Nantel, Brochu, & Prince, 2006). Both groups walked barefoot at similar speeds (0.98 and 1.01 $m \cdot s^{-1}$ for normal weight and obese groups, respectively). When comparing power relative to mass in the present study, our outcomes are consistent with those of Nantel et al. (2006). The Non-Obese group generated more power (W/kg) at the hip than the Obese group, with no differences in knee or ankle stance phase powers between groups. With load carriage in a healthy sample of adults similar in age to our participants, there was an increase in hip power generation, and knee and ankle joint power absorption, with the greatest increase in the knee measure (H. Wang et al., 2013). This partially agrees with our results, as the ankle dominated the power generation differences between Obese and Non-Obese groups, while the hip and knee joints dominated the differences in power absorption. In Study 2 of this dissertation, loading increased power generation at the hip, knee, and ankle, and power absorption at the hip and knee. Similar to the results of H. Wang et al. (2013), the magnitude of the knee power responses were greater than those of the hip or ankle joints. Therefore, lower extremity joint power responses to increased mass via load carriage is similar, but not quite the same as increasing mass by obesity.

Footwear*Group Interaction

One of the intents of this study was to identify the impact of footwear on walking mechanics between Obese and Non-Obese adults to address the lack of consistent outcomes in the literature. In general, our data show that lower extremity kinetic responses do differ between footwear conditions depending on body weight, which supports our hypothesis that varying footwear conditions have partially driven conclusions developed in the literature. For example, mass*height normalized knee extensor moments in early stance were similar between the Obese group while shod and both footwear conditions of the Non-Obese group, but this measure in the Obese group was lower while barefoot. Lai et al. (2008) reported no differences in hip or knee moments (as Nm/kg*m) between normal and overweight youths when walking barefoot at their preferred speeds. These authors did find smaller plantarflexor moments however in the overweight group (Lai et al., 2008). In contrast, while walking shod at a fixed speed, larger ankle moments (as Nm/Nm) have been observed in obese children (Gushue et al., 2005). In adolescents matched for walking speed, mass and height normalized knee moments in early stance were larger in the obese group, but footwear condition was not reported (McMillan et al., 2010). Thus, walking speed and footwear both influence outcomes regarding obesity’s impact on lower extremity kinetic measures.

In the present study, walking barefoot reduced joint moments at the knee and joint powers at the ankle, knee, and hip in obese adults. The knee moment results were especially promising, as the absolute peak knee extensor moment in early stance in the Obese group was equal to that of the Non-Obese group. Additionally, the Obese group produced extensor function throughout stance while shod, but while barefoot they adopted a flexor moment during midstance similar to that of the Non-Obese group. Thus, removing the shoe of an obese adult seems to promote walking mechanics that mimic a healthy population in both magnitude of kinetic parameters and control strategy. This adaptation to reduce joint loads occurs without any decrease in weight, which previous investigations have shown is a mechanism for reducing knee joint loads (Aaboe, Bliddal, Messier, Alkjaer, & Henriksen, 2011; Messier et al., 2005). Interestingly, this phenomenon of reduced joint moments and powers while barefoot did not occur in the Non-Obese group. Thus, there is likely a weight threshold above which removing cushion from under the foot promotes a reduction in knee joint loads.

Normalization

Significant increases in GRFs and ankle (A1, A2), knee (K2, K3) and hip (H2) joint powers in the Obese group were approximately equal to the larger mass of that group. Similarly, the ankle plantarflexor moment, and knee and hip flexor and extensor moments increased approximately in proportion to the larger size of the Obese group. Therefore, these variables were only different between groups in the absolute data. In a sample of adults with similar characteristics to those in the present study, significantly smaller body weight normalized peak vertical and propulsive forces were observed in the obese individuals, but the average preferred speed of that group was significantly slower than the non-obese group (1.12 vs. 1.27 $m \cdot s^{-1}$) (Lai et al., 2008). If the obese group had walked at a similar speed as the normal group, perhaps they would have had similar body weight normalized peak forces, as was the case in Browning and Kram (2007). Similar to the present study, the 63.5% heavier obese group in Browning and Kram (2007) experienced ~61% larger absolute vertical and anteroposterior GRFs when walking at 1.5 $m \cdot s^{-1}$.

Absolute values of four kinetic variables were equivalent between groups, but normalizing these data made them appear smaller in the Obese than the Non-Obese group. These variables included the early stance dorsiflexor moment, the early stance hip extensor moment, knee power absorption during swing, and hip generation prior to toe-off. However, the erratic response of the center of pressure immediately following foot strike is compounded as the estimates of joint moments progress proximally from ankle to hip. This makes the early stance hip moment peak less reliable than the other moment peaks mentioned above. A statistical difference between groups was observed for these variables only after normalization. For example, the average absolute ankle dorsiflexor moments were -16.58 Nm for the Non-Obese Shod condition and -15.22 Nm for the Obese Shod condition. The average height and mass product for the two groups were 123.74 kg*m and 195.19 kg*m, for the Non-Obese and Obese groups, respectively. After normalization, these dorsiflexor moments became -0.13 Nm/kg*m (Non-Obese) and -0.08 Nm/kg*m (Obese). Even though no difference in the dorsiflexor moment was present in the absolute data, a difference in this measure was imposed once data were normalized as a consequence of the larger size of the Obese group.

Joint moment comparisons between non-obese and obese groups have produced varying conclusions. DeVita and Hortobagyi (2003) reported that obese adults walking at ~1.5 $m \cdot s^{-1}$ produce larger absolute ankle moments but similar knee and hip joint moments as non-obese adults. Normalizing by mass resulted in significantly smaller knee moment magnitudes for the obese group (DeVita & Hortobagyi, 2003). This outcome mirrors the impact of normalization in the present study, as presented in the previous paragraph. In contrast to DeVita and Hortobagyi (2003), Browning and Kram (2007) reported larger hip extensor moments and a trend that did not reach significance of larger knee moments in the obese group, but only when these measures were compared in absolute terms. Once normalized by mass, knee moments appeared smaller across the range of speeds in the obese group (0.5-1.75 $m \cdot s^{-1}$) (Browning & Kram, 2007). Disagreement between these studies for the absolute measures may be a factor of body size, as the participants in Browning and Kram (2007) had an average BMI of 35.6 $kg \cdot m^{-2}$, while those in DeVita and Hortobagyi (2003) had an average BMI of 42.3 $kg \cdot m^{-2}$, with a maximum of 58.7 $kg \cdot m^{-2}$.

In children, larger body weight and height normalized plantar flexor moments in the obese have been reported when walking at equivalent speeds (Gushue et al., 2005). No difference in knee extensor moments were observed though. However, this lack of difference in knee moments may have been related to variability within the sample rather than an artifact of normalization. The standard deviation in the overweight group was equivalent to the mean (25.8 ± 25.6 Nm), whereas the non-obese group was much less variable (16.5 ± 8.4 Nm) (Gushue et al., 2005). when normalized, these values were 2.3 ± 1.1 Nm/Nm and 2.0 ± 1.6 Nm/Nm for the normal and overweight groups, respectively (Gushue et al., 2005). This demonstrates that larger absolute, but smaller normalized joint moments, of obese individuals is also a phenomenon present in children. In a group of adolescents (12-17 years) walking at their preferred speed, the height and mass normalized ankle plantarflexor moment was reduced in the obese group (McMillan et al., 2010). Early stance and late stance knee flexor moment, and early stance hip extensor moment were also significantly smaller in the obese group (McMillan et al., 2010). Thus, normalized joint moment magnitudes seem to be reduced at preferred speeds in the obese group, but are similar to non-obese individuals when speeds are also similar. The preferred speed of obese individuals is slower than that of their normal weight peers and this likely influences these outcomes (Lai et al., 2008).

When comparing obese and non-obese children, Shultz, Hills, Sitler, and Hillstrom (2010) reported no differences in sagittal plane joint powers at the knee or ankle joints when body weight was a covariate. At the hip, the early stance power generation was higher in the obese group even after accounting for weight (Shultz et al., 2010). In contrast to studies mentioned previously, where walking speed was manipulated and cadence was freely chosen, Shultz et al. (2010) imposed a fixed cadence 130% of the preferred value but allowed speed to vary. Speed did not differ between groups (average = 1.18 $m \cdot s^{-1}$ vs. 1.23 $m \cdot s^{-1}$ for obese and normal weight groups, respectively), and the higher stride rate, compared to the preferred, increased hip and knee sagittal plane powers in both normal weight and obese participants (Shultz et al., 2010). In the present study, mass normalized joint powers were also not significantly different between groups for the ankle or knee during stance. Mass normalized hip power generation was 56.33% larger in the Obese group during early stance, but 20.75% lower in late stance compared to the Non-Obese group. These adaptations at the hip joint in our sample may be a response to the slightly challenging fixed speed (1.5 $m \cdot s^{-1}$) that was imposed, in contrast to the slightly challenging cadence imposed by Shultz et al. (2010).

Two joint moments (ankle dorsiflexor and early stance hip extensor) and two joint powers (K4 and H3) were not increased in the Obese group for the absolute data, suggesting obesity did not impact those measures. However, these became statistically smaller in the Obese group after normalization due to the larger denominator (i.e., mass, weight, or mass*height) used for individuals in that group. Thus, where no differences in strategy were present in absolute data for those measures, a difference was observed after normalization.

Browning and Kram (2007) highlighted absolute joint loads. Their reasoning was based on data suggesting that absolute joint loads more accurately represent the impact of obesity on joint kinetics. Joint damage is likely related to the overall magnitude of forces experienced at a joint, especially for the knee (Felson, 1996). Other evidence for an emphasis on absolute forces and moments is that knee joint compressive forces increase 2-3 lb per 1 lb of weight gained (Felson et al., 2000). A study on weight loss found a decrease in knee joint compressive forces of 4 lb for every 1 lb lost (Messier et al., 2005). Based on these data, normalizing forces and moments by body weight does not seem to provide a true picture of loads occurring within a joint.

In addition, joint contacting surface area does not scale proportionally with body mass. Ding, Cicuttini, Scott, Cooley, and Jones (2005) reported that obese individuals had ~9% and ~6% more bone area on the medial tibia and lateral tibia surfaces than normal weight adults, despite being ~50% heavier. Therefore, stress ($\sigma = \text{Force} \cdot \text{Cross} \cdot \text{sectional Area}^{-1}$) on articular cartilage will be higher in obese individuals because the increase in joint compressive forces are greater than the increase in joint contact area. This is an important consideration for mechanical means of joint damage.

The exact mechanisms causing osteoarthritis are under investigation, but Piscoya, Fermor, Kraus, Stabler, and Guilak (2005) demonstrated an increase in molecular signals associated with osteoarthritis development after compressive stress. Given these disproportional responses to increases or decreases in body weight, it seems that absolute, rather than normalized, joint kinetic measures may be more appropriate when comparing obese and non-obese individuals. At a minimum, joint kinetic data should be presented both as absolute and normalized when contrasting these groups.

One limitation of this study is the lack of data on the physical activity patterns of these two groups. Even though the one group is obese, they were otherwise healthy individuals and may routinely participate in exercise. An obese, sedentary individual may respond differently than a physically active obese individual. Future investigations should characterize the physical activity levels of participants to further understand the role of obesity in walking.

Conclusion

To our knowledge, this is the first study to compare the impact of footwear on walking gait in obese and non-obese adults. Our first hypothesis that obese individual would experience larger peak ground reaction forces, joint moments, and joint powers was supported in the absolute data. Our hypothesis that footwear would reduce these measures was partially supported. Ground reaction forces were decreased in the barefoot condition for both groups. However, joint kinetic responses were dependent on body morphology, as the Obese and Non-Obese adults responded differently to footwear conditions. Finally, comparisons between these groups are dependent on normalization scheme. Statistical outcomes, and conclusions drawn from them, are not identical between absolute and normalized data when comparing groups who vary greatly in mass.

CHAPTER VI

GENERAL SUMMARY AND CONCLUSIONS

Summary

The purpose of this dissertation was to investigate the impact of footwear and increased mass on walking mechanics and energetics. In general, walking barefoot produces different spatiotemporal, kinematic, and kinetic responses than walking while shod. Loading alters these same measures and also increases metabolic cost. Obese individuals display similar kinematics, but dissimilar spatiotemporal and kinetic responses as non-obese persons.

In Study 1, two footwear conditions (barefoot, athletic shoe) and two loading conditions (no load, 15% body mass backpack load) during treadmill walking were compared. Adding load to the trunk required ~12% extra metabolic energy expenditure. Across loading conditions, removing the shoes resulted in a nonsignificant, ~1% decrease in this measure. The treadmill used in this study was not capable of capturing ground reaction forces, so Study 2 was devised to address mechanical differences in walking gait in response to these footwear and load conditions.

In Study 2, participants walked overground in the same footwear and loading conditions as Study 1 while motion and ground reaction forces were collected. Walking barefoot decreased ground reaction forces, but loading increased ground reaction forces in proportion to the added mass, which agrees with previous data from load carriage

(Tilbury-Davis & Hooper, 1999) and obesity (Browning & Kram, 2007) studies. Walking barefoot did not influence ankle moments, but increased hip and knee extensor moments. Loading increased plantar flexor and knee and hip extensor moments. These extensor moments prevent collapse of the lower extremity, so it makes sense that increasing mass would promote larger joint loads that act to keep the body upright. Based on these outcomes, Study 3 was designed to compare the impact of an internal load (i.e., obesity) on walking mechanics, while addressing methodological differences among other studies.

In Study 3, a comparison between obese and non-obese adults was made to investigate the role of footwear in these groups. The outcomes of previous studies who used different footwear conditions were inconsistent (DeVita & Hortobagyi, 2003; Lai et al., 2008), and in some cases footwear conditions were not reported (McMillan et al., 2010; Nebel et al., 2009). In Study 3 it was found that spatiotemporal and kinematic responses to walking barefoot are similar between these groups, but kinetic responses are dependent on body composition. Footwear*Group interactions were identified for nine kinetic measures, including three knee joint moment peaks and three knee power peaks. For example, the Obese group, when walking barefoot, had similar knee joint flexor moment function during midstance as the Non-Obese group. However, when shod, the Obese group did not demonstrate any flexor moment function during this period. Based on the multiple interactions identified, it is important for future work to consider and report the footwear used in walking trials. Knee kinetics have been the primary focus of obesity related work, but despite its apparently higher sensitivity to footwear than the ankle or hip joints, the influence of footwear has largely been ignored. Because of this, differences in footwear condition among studies has confounded the varying outcomes observed. Comparisons among previous studies should take into account the footwear conditions along with other participant characteristics and methodologies that may have played a role in the observed outcomes. From an applied perspective, some consider barefoot running to be a healthy alternative to running while shod. In a similar vein, some barefoot physical activity could be considered beneficial for obese adults. This suggestion obviously merits future investigation.

A secondary focus of Study 3 was the impact of normalization on comparisons of kinetic variables between obese and non-obese. Conflicting results of previous studies are confounded by varying normalization methodologies, as joint moments have been reported as absolute (Browning & Kram, 2007), normalized by mass (DeVita & Hortobagyi, 2003), and normalized by the product of mass and height (Lai et al., 2008). In Study 3, absolute ankle plantar flexor and knee and hip extensor moments were greater in the Obese group. Hip and knee flexor moments were also larger in that group. In the normalized data, none of these measures were different between groups, and ankle dorsiflexor and early stance hip extensor moments became larger in the Non-Obese group. When considering that the absolute forces across a joint are associated with cartilage damage (Felson, 1996) and that joint contacting area does not scale with the increased mass of an obese person (Ding et al., 2005), it seems that normalizing joint moments may be inappropriate for obese individuals.

In summary, a non-obese person whose mass is increased via an external load responds to walking barefoot differently than an obese person. Chronic exposure to their larger mass, or the distribution of mass, in the obese population may account for these differences. Footwear differences partially explain the varied conclusions in the literature regarding the impact of obesity, as conclusions have been based on responses in differing footwear conditions. Finally, normalization method influences statistical outcomes when comparing groups who differ greatly in body mass. If data is to be normalized, absolute data should also be presented.

Conclusions

Study 1 demonstrated that metabolic cost of walking increases nearly in proportion to added mass, but that footwear does not significantly influence this measure. This supported the hypothesis that loading would increase metabolic cost, but the hypothesis that walking barefoot would reduce this measure was not supported.

The hypothesis that increasing mass would require larger GRFs was supported. The results from Study 2 and 3 show that GRFs scale with total mass, whether this mass is externally added or a consequence of obesity. The hypothesis that walking barefoot would decrease GRFs was also supported. Again, Study 2 and 3 illustrated that walking without shoes decreased vertical and anteroposterior forces regardless of loading or obesity.

The hypothesis that increasing mass would increase joint moments was supported. In Study 2, ankle plantar flexor and knee extensor moments were increased with added load. In Study 3, ankle plantar flexor, and knee and hip extensor moments were larger in the Obese group. Collectively, these joint actions prevent collapse of the lower extremity, so it makes sense that a greater body weight would require these measures to be larger. The Obese group had decreased knee extensor moments in early stance and knee flexor moment function during midstance while barefoot, in contrast with maintained knee extensor function during this period while shod. Thus, removing the shoe promoted a knee moment profile more similar to that of the Non-Obese group, which may be a positive adaptation.

Future Directions

Based on the data presented here, several directions remain open for future work. The rate of obesity in the United States is steadily climbing (Fryar et al., 2014), and more work is necessary to understand its effects on locomotion. Based on the results of the present work, future gait related obesity questions should consider footwear condition as an important factor. The chosen footwear type must be reported to clarify subsequent findings for the reader.

Second, this dissertation focused on increasing mass, but the impact of decreased mass through weight-loss is an important avenue for future investigations. As noted here, obesity is not the same as acutely adding mass to the body via an external load. Thus, literature investigating unweighting of the body (e.g., Fischer and Wolf (2015)) also may not fully capture adaptations made to an actual decrease in total body mass. Quantifying adaptations made in response to weight loss are important for understanding the positive impacts of physical activity. For example, weight loss reduces symptoms of arthritis (Aaboe et al., 2011), increases balance (Teasdale et al., 2007), and reduces joint loading (Hortobagyi, Herring, Pories, Rider, & DeVita, 2011) and metabolic cost of walking (Delextrat, Matthew, & Brisswalter, 2015). Given the estimated $147 billion yearly spending on obesity related health issues (Finkelstein, Trogdon, Cohen, & Dietz, 2009), it is essential to more fully appreciate the mechanical and metabolic changes associated with weight loss.

Regarding the impact of normalization on joint moment data, alternative methods of normalization may be investigated in the future. One potential method for obtaining relative kinetic parameters could be to adopt the common procedure performed on metabolic data of normalizing a dependent measure by lean body mass, rather than total body mass (Delextrat et al., 2015). This would normalize a magnitude only by the non-fat tissues (e.g., bone, muscle, blood, etc.). In so doing, the larger amount of adipose tissue in an obese person would not be part of the denominator used in providing the relative score. As a consequence, the influence of the extra fat tissue would still be reflected in the value of a joint moment or power, while still scaling these values to the height and/or mass of the person to compare individuals of varying sizes.

Much of the previous obesity work has focused on older adults and those with other musculoskeletal issues or diseases in addition to obesity. Since gait mechanics differ between those who are obese and those who are obese and also have arthritis (Harding, Hubley-Kozey, Dunbar, Stanish, & Wilson, 2012), confounding comorbidity factors make direct comparisons between some studies difficult. This also clouds the implications for those who are obese but do not have other health problems. Therefore, more research is needed in obese, but otherwise healthy, children and young adults to specifically target the impact of obesity itself on locomotion and other daily living tasks.

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APPENDIX A

INSTITUTIONAL REVIEW BOARD DOCUMENTATION

UNIVERSITY of
NORTHERN COLORADO

CONSENT FORM FOR HUMAN PARTICIPANTS IN RESEARCH
UNIVERSITY OF NORTHERN COLORADO

Project Title: The effect of load carriage and footwear on gait kinematics and walking economy in college students

Investigators: Kevin Dames, Sutton Richmond, and Sherilyn Sommerville
School of Sport & Exercise Science

Research Advisor: Dr. Jeremy D. Smith (970) 351-1761, School of Sport & Exercise Science

Purpose: to investigate the differences in walking kinematics and economy in different loading and footwear conditions. With the increasing popularity of shoes that provide minimal foot protection or support (e.g. sandals, minimalist running shoes etc.) it is our goal to understand if humans walk differently when carrying a load without ample foot support.

One visit will be made to the Human Performance Laboratory (Gunter 1740). After signing the informed consent, your body mass and height will be recorded. Following this, reflective markers will be placed on specific locations of your body, a heart monitor will be fitted to you, and a mask to collect your expired breaths for analysis in the metabolic cart will be adjusted to you. Once you are set up, you will complete the 4 walking conditions, each separated from subsequent trials by a 3-minute rest that allows you to remove the mask and the backpack, if it were worn during the trial. The total time of involvement will be about 2 hours.

The walking trials will last a total of 24 minutes on the treadmill, split into four different conditions (6 minutes each, 3.5 mi./hr.): unloaded-shod, unloaded-barefoot, loaded-barefoot, and loaded-shod. The loaded conditions will require you to carry a backpack equal to 15% of your body mass and the shod conditions will be performed in your own shoes. The order of the four conditions will be randomized. Data collected from the trials will include metabolic cost, stride characteristics, and heart rate.

Participation in this study will not exceed the intensity of activity already experienced while walking across campus with a typical backpack. You will be familiarized with the treadmill and allowed to try walking on it prior to actual testing. A technician will be present at the side of the treadmill as a spotter. The barefoot conditions may cause some discomfort, but the treadmill is made of a rubber material and should cause only minimal discomfort. Fatigue, muscle soreness, strains or sprains associated with physical activity may occur but should resolve themselves within a couple days. If an injury requiring medical attention should occur, the researchers will contact the necessary personnel.

There are no direct benefits to you as a participant in this project. However, the information gained from this study will provide further understanding of how humans cope with load carriage while walking, and the influence footwear has in that task. Understanding the influence of load carriage and footwear on kinematics in the college-aged population is important because excessive stress on the body, specifically the low back, may contribute to musculoskeletal pain. If walking barefoot significantly alters gait, trunk orientation, or metabolic cost, inferences can be made to walking in sandals, minimalist shoes and other unsupportive footwear that mimic the barefoot condition. Information from this study may be used as healthy controls to compare with populations who have unique gait and posture issues, such as amputees.

Results will be used for academic purposes only (e.g. papers or presentations), but any identifying information will be removed in an effort to maintain your privacy. All information will be locked in the Human Performance Laboratory.

Participation is voluntary. You may decide not to participate in this study and if you begin participation you may still decide to stop and withdraw at any time. Your decision will be respected and will not result in loss of benefits to which you are otherwise entitled. Having read the above and having had an opportunity to ask any questions, please sign below if you would like to participate in this research. A copy of this form will be given to you to retain for future reference. If you have any concerns about your selection or treatment as a research participant, please contact the Office of Sponsored Programs, 25 Kepner Hall, University of Northern Colorado Greeley, CO 80639, 970-351-2161.

____

Subject's Signature Date

____

Researcher's Signature Date

A. Purpose

College students regularly carry heavy loads in their backpacks while in transit to class. However, this population has received much less attention in the load carriage literature than other groups (e.g. military and children). In addition, college students often wear minimal, unsupportive footwear (e.g. flip-flops, sandals, minimalist shoes etc.) that do little to protect the feet. The effect of carrying heavy loads without a supportive shoe on walking kinematics and economy in this population is still unclear.

The influences of carrying a heavy load on the back have been previously investigated. Doing so, tends to promote increased time of double support (Singh & Koh, 2008; Wang, Pascoe, & Weimar, 2001), increased time of stance phase, decreased single support time (Wang et al., 2001), decreased preferred walking velocity (Singh & Koh, 2008;Wang et al., 2001), increased forward lean of the trunk (Lloyd & Cooke, 2011; Singh & Koh, 2008; Simpson, Munro, & Steele, 2011; Smith et al., 2006) and increased metabolic cost (Lloyd & Cooke, 2000). These kinematic changes seem to be made in an attempt to increase dynamic stability when the body's center of mass is elevated due to a heavy load on the posterior surface of the body (Singh & Koh, 2008).

Alterations in kinematics have also been noted in barefoot walking. Observed changes in gait while barefoot, compared to shod, include: slower preferred walking speed, faster step cadence, shorter step length, shorter double support time, shorter total support time, and higher single support time (Lythgo, Wilson, & Galea, 2009). It is presumed that shortening the stride length reduces the discomfort experienced at foot strike without the cushioning of a standard shoe. As mentioned above, it is assumed that this is also the case when using the minimalist style shoes regularly worn by many college individuals. In addition, while commuting on foot to class, the added load of a backpack increases the reaction forces experienced at foot strike. This may potentially modify gait mechanics even further in the loaded condition compared to the unloaded condition.

Given the noted gait adjustments made under novel conditions (i.e. loaded, barefoot) it is of interest to understand what kinematic and metabolic affects these promote while simultaneously experienced. The combined influence of footwear and load carriage on gait in adults has received little attention. Thus, an attempt will be made in this study to reveal any interactions that may exist between loaded walking and shoe selection on gait mechanics and walking economy in college-aged individuals.

This research project will investigate:

1. How do footwear and backpack loads influence gait mechanics in college-aged individuals?
2. How do footwear and backpack loads influence walking economy in college-aged individuals?

This research project qualifies as exempt because the intensity and type of activities will not exceed the daily activities of the population investigated. All participants will exceed 18 years of age and will have the capacity to voluntarily engage in the study. No data collected will be of a sensitive nature so accidental disclosure of information will not be harmful to the participants. No vulnerable populations will be included.

B. Methods

1. **Participants**

Fifteen healthy, non-smoking adults (male and female), 18-30 years of age, who are accustomed to wearing a traditional style backpack will be recruited for this study from graduate and undergraduate classes at UNC and from the surrounding community. Participants must be free of lower extremity or low back injury for at least 6 months before participation in this study. Prior to participation, volunteers will give written, informed consent. Participants will not receive any course credit or tangible incentive as an enticement to participate in this study.

2. **Data Collection Procedures**

The participants will be asked to perform 4, 6-minute walking trials on the treadmill in the Human Performance Laboratory (Gunter Hall 1740). Participant height (m), mass (kg), and age (years) will be recorded and tight fitting clothing will be provided. A traditional backpack will be fitted to the person so that the bottom edge is at the level of L1 so that the positioning is standardized between subjects. Reflective markers will be placed on specific locations of the body in order to create a digital model of the person for analysis. No actual video of the person will be collected; the cameras will capture the position of the markers.

All four walking trials will be completed at a slightly challenging walking speed of 1.5 m/s: two loading conditions (unloaded and 15% body mass) and two footwear conditions (standard athletic shoe and barefoot). The order in which these conditions are completed will be randomized and performed wearing the participant's own shoes. During each trial, an average VO_2 between minutes 4-6, gait kinematics, and trunk orientation will be recorded. The first complete gait cycle at minutes 4 and 5 will be collected for analysis. Three minutes of rest will be given between trials to allow the person to remove the backpack, sit, and disconnect themselves from the metabolic cart prior to subsequent walking efforts. The total time of involvement in the study will be about 1.5-2 hours.

3. **Data Analysis Process**

All dependent variables will be analyzed using a Repeated Measures ANOVA. A Bonferroni post hoc test will be performed where pair-wise comparisons are warranted. Means and standard deviations for all variables will be presented. An alpha level of .05 will be set for all statistical procedures.

4. **Data Handling Process**

The data will be collected privately, without any outside observers present with exception to the primary investigator, research advisor, and research assistants. Each participant will be assigned an identification number that will be used for all tests and data collection. There will be no identifiable information connecting the participants to the data other than the informed consent. Consent forms will be kept in a locked file cabinet in the Biomechanics Lab (Gunter Hall 1750) and will only be accessible by the researchers. Any electronic data will be password protected and will only be identifiable via the subject's identification number. Any information that may be considered personally identifiable will be stored for a period of five years in the locked filing cabinet. After the five years, it will be removed and destroyed. Any unidentifiable information will be kept indefinitely. Every possible precaution will be taken to protect the participant's identity.

C. Risks, Discomforts and Benefits

The participants will be familiarized with the treadmill and allowed to try walking on it in a brief warm-up prior to testing. The researchers will provide a visual and verbal demonstration of treadmill use. The tasks required of the participants will not exceed their normal daily walking commutes across campus. The participants will be recreationally active and free from any musculoskeletal or head injury for at least 3 months. However, certain physical discomforts are possible, as with any type of physical activity. Muscle soreness and fatigue associated with any type of physical activity is possible but these should dissipate within 48 hours of testing without need for medical attention. Strains and sprains are a possibility but should also resolve themselves without seeking treatment. There may be some discomfort when walking on the treadmill barefoot, but the treadmill belt is slightly compliant. A technician will be positioned at the side of the treadmill as a spotter throughout testing in case of a trip. However, the population selected is not particularly prone to falls. Potential injuries resulting from a fall range from abrasions, contusions, or bone fractures. A second technician will be in charge of emergency treadmill shut-off as well.

Risks associated with participation in this study also include potential psychological discomfort (e.g. feeling self-conscious while wearing the tight fitting clothing during testing). To minimize this, only the researchers and research advisor will be present during data collection. If at any time a participant decides to withdraw from the study he/she may do so without any fear of negative consequences. If in the unlikely event that someone is injured, the researchers will contact the appropriate medical authorities.

There are no direct benefits to the participants but they will help contribute to a growing body of knowledge regarding the interaction of load carriage and footwear and their influences on gait mechanics. Understanding the influence of load carriage and footwear on kinematics in the college-aged population is important because excessive stress on the body, specifically the low back, may contribute to musculoskeletal pain. If walking barefoot significantly alters gait, trunk orientation, or metabolic cost, inferences can be made to walking in sandals, minimalist shoes and other unsupportive footwear that mimic the barefoot condition. This information may also be used as a control group for comparison to populations with unique gait and posture issues, such as amputees, in the future. The data obtained in this study will be for academic purposes only (e.g. research papers and presentations in the classroom, and potentially at professional conferences).

D. Costs and Compensations

There are no costs to the participants involved in this study apart from their time commitment (approximately 1.5-2 hours). Participants will not receive any kind of compensation for their involvement in the study. Should potential participants decide not to become involved in the study there will be no cost or penalty to them.

E. Grant Information

At this time, there is no grant funding for this project.

Attached Relevant Materials

The Informed Consent document is attached to this document.

UNIVERSITY of
NORTHERN COLORADO

CONSENT FORM FOR HUMAN PARTICIPANTS IN RESEARCH
UNIVERSITY OF NORTHERN COLORADO

Project Title: Effects of load carriage and footwear on kinetic and spatiotemporal parameters of walking

Investigators: Kevin D. Dames

Research Advisor: Dr. Jeremy D. Smith (970) 351-1761, School of Sport & Exercise Science

Purpose: to investigate the differences in walking kinetic and spatiotemporal parameters under different loading and footwear conditions. With the increasing popularity of shoes that provide minimal foot protection or support (e.g. sandals, minimalist running shoes etc.) it is our goal to understand if humans experience higher peak forces and/or loading rates under barefoot conditions, compared to shod, with and without trunk loads in order to provide insight into the relative risk of injury associated with these conditions.

One visit will be made to the Biomechanics Laboratory (Gunter 1750). After signing the informed consent, you will complete walking trials at your preferred walking speed and at a fixed speed of 1.5 m/s. You will complete a series of 10-meter walking trials with 4 different combinations of 2 load conditions (unloaded, loaded) and 2 footwear conditions (barefoot, shod) at each speed (i.e. preferred, fixed). You may rest between each trial.

During the loaded conditions you will carry a backpack equal to 15% of your body mass. The shod conditions will be performed in your own shoes. The order in which these conditions are performed will be randomized. Data will be collected from the force plates in the center of the walkway. Reflective markers will be placed on anatomical landmarks using double sided tape for motion capture. Total involvement will be about 1.5 hours.

Participation in this study will not exceed the intensity of activity already experienced performing tasks such as carrying groceries. You will be given verbal and visual demonstration of the tasks as well as provided time to become comfortable walking within the testing environment. The barefoot conditions may cause some discomfort, but rest will be provided between conditions to minimize this risk. Fatigue, muscle soreness, strains or sprains normally associated with physical activity may occur but should resolve themselves within a couple days. If an injury requiring medical attention should occur, the researchers will contact the necessary personnel.

There are no direct benefits to you as a participant in this project. However, the information gained from this study will provide further understanding of how humans cope with load carriage and the role footwear plays in performance of that task. Understanding the influence of load carriage and footwear on kinetic and spatiotemporal parameters is important because high loading rates, as observed in both loaded and barefoot walking, respectively, compared to normal walking, may be associated with lower extremity injury when repeatedly experienced. If walking barefoot significantly alters any gait parameters involved in this study, inferences may be made to walking in sandals, minimalist shoes and other unsupportive footwear that attempt to mimic the barefoot condition. Results will be used for academic purposes only (e.g. papers or presentations), but any identifying information will be removed in an effort to maintain your privacy. All information will be locked in the Biomechanics Laboratory (Gunter 1750).

Participation is voluntary. You may decide not to participate in this study and if you begin participation you may still decide to stop and withdraw at any time. Your decision will be respected and will not result in loss of benefits to which you are otherwise entitled. Having read the above and having had an opportunity to ask any questions, please sign below if you would like to participate in this research. A copy of this form will be given to you to retain for future reference. If you have any concerns about your selection or treatment as a research participant, please contact the Office of Sponsored Programs, 25 Kepner Hall, University of Northern Colorado Greeley, CO 80639, 970-351-2161.

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Subject's Signature Date

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Researcher's Signature Date

A. Purpose

College students regularly carry heavy backpacks. However, this population has received much less attention in the load carriage literature than other groups (e.g. military personnel and children). In addition, college students often wear minimal, unsupportive footwear (e.g. flip-flops, sandals, minimalist shoes etc.) that attempt to mimic the barefoot condition by providing little cushion. The kinetic and spatiotemporal changes in walking without a supportive shoe while also carrying a load are still unclear.

The influences of trunk loading on spatiotemporal and kinetic parameters have been previously investigated. Noted spatiotemporal differences in walking with trunk loads include increased time of double support (Singh & Koh, 2008; Wang, Pascoe, & Weimar, 2001), increased time of the stance phase, decreased single support time (Wang et al., 2001), and decreased preferred walking velocity (Singh & Koh, 2008; Wang et al., 2001). Compared to walking without a load, Wang et al. (2012) reported load carriage to elicit higher peak values for vertical ground reaction force (GRF), braking GRF, vertical ground reaction loading rate, and braking ground reaction loading rate. In their review of the literature, Zapdoor & Nikooyan (2011) report that an elevated risk of lower extremity injury is associated with high loading rates.

Alterations in spatiotemporal parameters have also been noted in barefoot walking. Observed changes in gait while barefoot, compared to shod, include slower preferred walking speed, faster step cadence, shorter step length, shorter double support time, and higher single support time (Lythgo, Wilson, & Galea, 2009). A key kinetic finding is that while walking barefoot the foot experiences a higher loading rate than shod walking (Zhang et al. 2013). The addition of a trunk load while barefoot may potentially modify gait mechanics even further.

Given the noted gait adjustments made under novel conditions (i.e. loaded, barefoot) it is of interest to understand what kinetic and spatiotemporal effects these elicit while simultaneously experienced. The combined influence of footwear and load carriage on gait in adults has received little attention. As previously mentioned, loading rate is associated with lower extremity injury; both load carriage and barefoot walking seem to elicit higher loading rates than unloaded and shod walking, respectively. Thus, this research project will investigate the effects of load carriage and footwear on walking gait in order to make inferences about the relative risk of lower extremity injury associated with each.

This research project will attempt to provide insight into the following questions:

3. How do kinetic and spatiotemporal parameters of barefoot walking compare to shod walking?
4. How do kinetic and spatiotemporal parameters of walking with a trunk load compare to walking without a trunk load?
5. What are the combined effects of loading and footwear on kinetic and spatiotemporal parameters of walking gait?

This research project qualifies as expedited because the intensity and nature of the tasks involved will not exceed the daily activities of the population investigated. All participants will exceed 18 years of age and will have the capacity to engage in the study voluntarily. No data of a sensitive nature will be collected so accidental disclosure of information would not pose a risk to participants. No vulnerable populations will be included.

B. Methods

5. **Participants**

Fifteen recreationally active, non-smoking adults (male and female), 18-30 years of age, who are accustomed to wearing a backpack and are comfortable walking barefoot will be recruited for this study from the UNC campus and surrounding area. Eligible participants will be free of lower extremity, low back, or head injury for at least 6 months prior to involvement. Before any activity is performed volunteers will provide written, informed consent. Volunteers will not receive any academic, monetary, or tangible benefit in exchange for their participation.

6. **Data Collection Procedures**

Participants will be asked to perform a series of walking trials under four conditions: Barefoot Unloaded (BU), Shod Unloaded (SU), Barefoot Loaded (BL), and Shod Loaded (SL). Participants will walk in each of the four conditions at a target pace of 1.5 (+/- 5%) m/s as well as at their preferred speed. All walking trials will be performed in the UNC Biomechanics Lab (Gunter 1750) across a 10-meter walkway. A tandem-belt instrumented treadmill imbedded in the center of the walkway will be used to collect kinetic and spatiotemporal data. Kinetic data will be normalized to body mass. Retroreflective markers taped to specific anatomical landmarks will be used along with infrared cameras for motion capture. All kinetic, spatiotemporal, and kinematic data will be averaged across three trials per condition for statistical analyses.

The order in which conditions are completed will be randomized. For the shod conditions participants will wear their own athletic footwear. A backpack loaded with 15% of the participant's body mass will be provided by the researchers for the loaded conditions. Total time of involvement will be about 1-1.5 hours.

7. **Data Analysis Process**

All dependent variables will be analyzed using a series of 2x2 ANOVAs with repeated measures. A Bonferroni post hoc test will be performed where pair-wise comparisons are warranted. Means and standard deviations for all variables will be presented. An alpha level of .05 will be set a priori for all statistical procedures.

8. **Data Handling Process**

The data will be collected privately, without any outside observers present with exception to the primary investigator, research advisor, and research assistants. Each participant will be assigned an identification number that will be used for all tests and data collection. There will be no identifiable information connecting the participants to the data other than the informed consent. Consent forms will be kept in a locked file cabinet in the UNC Biomechanics Lab (Gunter Hall 1750) and will only be accessible by the researchers. Any electronic data will be password protected and will only be identifiable via the subject's identification number. Any information that may be considered personally identifiable will be stored for a period of five years in the locked filing cabinet. After the five years, it will be removed and destroyed. Any unidentifiable information will be kept indefinitely. Every possible precaution will be taken to protect the participants' identities.

C. Risks, Discomforts, and Benefits

Participants will be familiarized with the activities involved in this study via visual and verbal demonstration by the researchers. The tasks required of the participants will not exceed their normal, daily activity. Participants will be recreationally active and free from any musculoskeletal or head injury for at least 6 months. However, certain physical discomforts are possible, as with any type of physical activity. Potential activity related physical discomforts include muscle soreness and fatigue, but these should dissipate within 48 hours of testing without need for medical attention. Strains and sprains are a possibility but should also resolve themselves without seeking treatment. There may be some discomfort when walking barefoot but rest between trials will be allowed to minimize this possibility. Potential injuries resulting from a fall range from abrasions, contusions, or bone fractures. If at any time a participant decides to withdraw from the study he/she may do so without any fear of negative consequences. If in the unlikely event that someone is injured, the researchers will contact the appropriate medical authorities.

There are no direct benefits to the participants but they will help contribute to a growing body of knowledge regarding the unique effects of trunk loading and footwear on gait mechanics. Understanding the influences of load carriage and footwear on walking gait is important because excessive stress on the body, specifically the loading rate and GRFs experienced by the foot during walking, may contribute to musculoskeletal pain. If walking barefoot significantly alters any of the parameters included in this study then inferences may be made to minimalist shoes that attempt to mimic the barefoot condition. Data from this population may be used as control data for comparison to other populations. The data obtained in this study will be for academic purposes only (e.g. research papers and presentations at professional conferences).

D. Costs and Compensations

There are no costs to the participants involved in this study apart from their time commitment (approximately 1-1.5 hours). Participants will not receive any kind of compensation for their involvement in the study. Should potential participants decide not to become involved in the study there will be no cost or penalty to them.

E. Grant Information

At this time, there is no grant funding for this project.

Attached Relevant Materials

The Informed Consent document is attached to this document.

Institutional Review Board

DATE: May 14, 2014

TO: Kevin Dames, B.S., M.A
FROM: University of Northern Colorado (UNCO) IRB

PROJECT TITLE: [604072-1] Effects of load carriage and footwear on kinetic and spatiotemporal parameters of walking

SUBMISSION TYPE: New Project

ACTION: APPROVED
APPROVAL DATE: May 14, 2014
EXPIRATION DATE: May 14, 2015
REVIEW TYPE: Expedited Review

Thank you for your submission of New Project materials for this project. The University of Northern Colorado (UNCO) IRB has APPROVED your submission. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on applicable federal regulations. Please remember that informed consent is a process beginning with a description of the project and insurance of participant understanding. Informed consent must continue throughout the project via a dialogue between the researcher and research participant. Federal regulations require that each participant receives a copy of the consent document.

Please note that any revision to previously approved materials must be approved by this committee prior to initiation. Please use the appropriate revision forms for this procedure.

All UNANTICIPATED PROBLEMS involving risks to subjects or others and SERIOUS and UNEXPECTED adverse events must be reported promptly to this office.

All NON-COMPLIANCE issues or COMPLAINTS regarding this project must be reported promptly to this office.

Based on the risks, this project requires continuing review by this committee on an annual basis. Please use the appropriate forms for this procedure. Your documentation for continuing review must be received with sufficient time for review and continued approval before the expiration date of May 14, 2015.

Please note that all research records must be retained for a minimum of three years after the completion of the project.

If you have any questions, please contact Sherry May at 970-351-1910 or Sherry.May@unco.edu. Please include your project title and reference number in all correspondence with this committee.
Really well written Kevin!
Best Wishes, Maria

This letter has been electronically signed in accordance with all applicable regulations, and a copy is retained within University of Northern Colorado (UNCO) IRB's records.