University of Memphis

University of Memphis Digital Commons

Electronic Theses and Dissertations

2-28-2014

Static and Dynamic Postural Stability of High Body Mass Index Subjects During Single-Leg Stance and Stair Descent

Tyler Ross Palumbo

Follow this and additional works at: https://digitalcommons.memphis.edu/etd

Recommended Citation

Palumbo, Tyler Ross, "Static and Dynamic Postural Stability of High Body Mass Index Subjects During Single-Leg Stance and Stair Descent" (2014). *Electronic Theses and Dissertations*. 864.
https://digitalcommons.memphis.edu/etd/864

This Thesis is brought to you for free and open access by University of Memphis Digital Commons. It has been accepted for inclusion in Electronic Theses and Dissertations by an authorized administrator of University of Memphis Digital Commons. For more information, please contact khggerty@memphis.edu.

STATIC AND DYNAMIC POSTURAL STABILITY OF HIGH BODY MASS INDEX SUBJECTS DURING SINGLE-LEG STANCE AND STAIR DESCENT

by

Tyler Ross Palumbo

A Thesis

Submitted in Partial Fulfillment of the

Requirements for the Degree of

Master of Science

Major: Biomedical Engineering

The University of Memphis

May 2014

Copyright © 2014 Tyler Ross Palumbo

All rights reserved

ACKNOWLEDGEMENTS

I would like to acknowledge and extend my sincere gratitude to a number of individuals for their guidance and support. First and foremost, I would like to thank my major advisor, Dr. John Williams, for his guidance, supervision, wisdom, and knowledge throughout my graduate research. Additionally, I would like to thank the members of my thesis committee, Dr. Audrey Zucker-Levin, Dr. William Mihalko, and Dr. Esra Roan for their help in my research.

I would also like to thank many other individuals for their contributions, without whom this research would not have been a success. Dr. Brooke Sanford provided vital knowledge, experience, and guidance regarding human motion analysis and the Visual3D program used for analysis. Kyle Hoffman was invaluable for discussing research ideas and questions. I am also especially grateful for my graduate school colleagues, Casey Hebert, Devin Conner, and Jason Lindsey for assisting with experiments, working through questions, and general laboratory training.

ABSTRACT

Palumbo, Tyler Ross. M.S. The University of Memphis. May 2014. Static and Dynamic Postural Stability of High Body Mass Index Subjects During Single-Leg Stance and Stair Descent. Major Professor: Dr. John L. Williams.

This study investigated the effects of body mass index (BMI) on stability and biomechanics during single leg stance (SLS) and stair descent (SD). A group of six high BMI subjects was compared with an age-matched control group of eleven young 'normal weight' (BMI < 25) adults. The high BMI individuals descended the stairs more slowly with longer support times. Their supporting limbs experienced larger hip, knee, and ankle sagittal-plane moments (normalized), smaller frontal plane hip moments, and larger frontal plane knee moments at toe-off of the swing limb, compared to controls. At swing limb touchdown, the supporting limb experienced hip flexion moments as opposed to extension moments, larger knee adduction moments, and lower normalized anterior ground reaction forces compared to controls. No differences were found for the investigated parameters during SLS. Stair descent differences in the high BMI participants suggest possible cumulative joint overloading, greater osteoarthritis risk, and decreased stability.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

ABBREVIATIONS

A/P – Anteroposterior

Abd – Abduction

Add – Adduction

Ant – Anterior

BMI – Body mass index

BoS – Base of Support

BW – Body weight

COM – Center of mass

COP – Center of pressure

Dorsi – Dorsiflexion

Ext – Extension

Flex – Flexion

GRF – Ground reaction force

Lat – Lateral

LTD – Left touchdown

LTO – Left toe-off

M/L – Mediolateral

Med – Medial

MoS – Margin of Stability

pCOM – Center of mass position

Plantar – Plantarflexion

Post – Posterior

RTD – Right touchdown

SLS – Single leg stance

TD – Touchdown

TO – Toe-off

TtC – Time to contact

TTS – Time to stabilization

vCOM – Center of mass velocity

vCOP – Center of pressure velocity

V-GRF SD – standard deviation of the vertical ground reaction force

XCOM – Extrapolated center of mass

CHAPTER I

INTRODUCTION

1.1 Background

The prevalence of obesity in the United States and worldwide is an increasing major health concern. In the United States, more than two thirds of the adult population is overweight or obese, while worldwide obesity rates have more than doubled in the past three decades (Del Porto et al., 2012, Dewan et al., 2013). The World Health Organization defines overweight as having a body mass index (BMI) ≥ 25 kg/m^2, while obesity is defined as having a BMI ≥ 30 kg/m^2 (WHO, 2008). BMI is defined as body mass divided by body height squared. The increase in obesity is especially troubling because obesity has been associated with various health problems including diabetes, stroke, heart disease, some types of cancers, and a generally lower quality of life (Del Porto et al., 2012; Wearing et al., 2006). Musculoskeletal impairments of function and mobility have also been linked to obesity, including osteoarthritis, osteoporosis, back pain, gout, and other disorders of the lower limbs and feet (Anandacoomarasamy et al., 2008; Wearing et al., 2006). These impairments can lead to difficulties in basic activities of daily living (Del Porto et al., 2012; Wearing et al., 2006). The risk of functional decline increases with increasing body mass (Del Porto et al., 2012; Himes et al., 2012). These physical limitations lead to decreased balance ability, altered gait patterns, and reduced muscle strength, which are three of the top risk factors for falls, especially in older adults (Del Porto et al., 2012; Zecevic et al., 2006). With the rise in prevalence of obesity and this group's known functional limitations, overweight and obese individuals may have more difficulty with daily activities such as standing balance and stair negotiation.

1.2 Postural Stability and Falls

Postural stability is defined as the ability to keep the body close to an equilibrium position when exposed to perturbations. Even though maintaining an upright posture is regarded as a simple task, the loss of balance resulting in a fall may occur many times throughout a person's life. Obesity has been shown to increase the risk of falling while performing a standing task or during ambulation, with middle-aged and older obese adults falling almost twice as often as aged matched non-obese adults (Corbeil et al., 2001; Fjeldstad et al., 2008; Himes et al., 2012; Wallace et al., 2002).

Studies on weight loss of both young and elderly obese subjects found that, before intervention, obese participants had impaired postural balance that was positively correlated with increasing body mass (Maffiuletti et al., 2005; Teasdale et al., 2007). After weight loss, the obese subjects had improved balance control and increased stability proportional to weight loss. In obese individuals that have a greater distribution of body fat in the their abdominal area, the body center of mass (COM) is shifted more anteriorly compared to lean individuals, potentially decreasing body balance which leads to a greater risk of falling, especially when combined with a relatively lower muscle mass. One study found that overweight individuals with this type of body fat distribution are at a greater risk of falling compared to non-obese when subjected to perturbations and other typical challenges of daily activities (Corbeil et al., 2001). A similar study found that obese children have a higher rate of injury to their incisors compared to non-obese children, indicating forward falls (Petti et al., 1997).

With increasing BMI being negatively associated with amount of physical activity as well as an increase in functional impairment, compromised balance and a risk of falls would seem likely to result. These limitations could lead to a fear of falling, which combined with the known sedentary nature of obese individuals (Fjeldstad et al., 2008), could lead to reduced physical activity, further functional impairment, and a greater falling risk.

1.3 Obesity in Static Balance

Research examining postural balance in adults is lacking, with a primary focus on anteroposterior measures of stability during quiet double leg stance. Teasdale et al. (2007) studied obese adults and McGraw et al. (2000) studied obese adolescents and found that reduced obesity correlated highly with improved quiet double stance balance control. Singh et al. (2009) reported that obese subjects had impaired postural control as measured by increased sway of the center of pressure (COP) during stance. Similarly, Hue et al. (2007) found an increase in body weight is highly correlated to a decrease in stability during balance using the velocity of the center of pressure (vCOP). Obese individuals had difficulty regaining bipedal balance in two perturbation studies (Berrigan et al., 2006; Matrangola et al., 2011). Nevertheless, many of these bipedal stance studies included or focused on elderly subjects that potentially had other risk factors for decreased functional performance, additionally, increased age is highly correlated with decreased postural control during standing (Chiari et al., 2002; Greve et al., 2013; Prieto et al., 1996; Wearing et al., 2006).

In two of the limited number of studies using young adults, Greve et al. (2013) and Chiari et al. (2002) both observed body weight as one of the most important factors that influenced postural test performance. Ledin et al. (1993) studied young and middle-aged normal weight adults with and without an added 20% body weight jacket and found that the added mass condition increased postural sway distance and velocity. A study on obese teenagers revealed COP displacements and path lengths were the same as non-obese on a hard surface, however, COP path length was increased in obese subjects compared to non-obese when standing on a foam surface (Bernard et al., 2003). The authors suggested that obese adolescents needed a more difficult task to expose the differences between the two groups. After testing quiet bipedal stance of one hundred obese subjects, Blaszczyk et al. (2009) only found differences in COP measures with eyes open or closed compared to controls for subjects with a BMI greater than 40. Their findings on obese individuals within the 25 to 40 BMI range contradicts many authors, and Handrigan et al. (2009) pointed out that their control group had COP sway and path length measures three or four times greater than those of controls found elsewhere in literature.

Single leg stance is a more difficult balancing posture than quiet bipedal stance due to the reduced base of support (BoS), which may help to elicit differences between groups of young and relatively healthy adults (Goldie et al., 1992; Riemann et al., 2003). Single leg standing is required for many daily activities including dressing, turning, kicking a ball, the single support phase of gait, and picking up an object. Limited ability during single leg balance is an indicator of fall risk (Hurvitz et al., 2000) and a predictor of injurious falls in the elderly (Vellas et al., 1997). Quantifying postural stability using various measures in single leg stance has given reproducible results in subjects with many different balance dysfunctions while also having good inter-rater and inter-subject reliability (Gerbino et al., 2007; Mancini et al., 2010).

Few studies, however, have compared the postural control abilities of obese subjects during single leg stance. In a study that explored the effects of body mass reduction by way of diet and light exercise in obese subjects, an increase in single leg stance time was found in individuals with a reduced body weight (Sartorio et al., 2001). Both Greve et al. (2007) and Ku et al. (2012) found that balance performance was negatively correlated with increasing BMI in young healthy obese and non-obese subjects performing double and single leg stance on a Biodex Balance System, which uses a circular tilting platform to record displacement of the COP. In a study that used force plate derived measures of mean vCOP and range of COP displacement, Mignardot et al. (2010) found that obese subjects strongly increased their vCOP and COP displacement range during both single leg stance by itself and during single leg stance combined with a time reaction test. The reaction time to an auditory signal was also increased in the obese subjects of that study, which combined with the vCOP and COP displacement range results, indicates that obese subjects had to devote more attentional resources to postural control, comparatively. Conversely, a study of single leg stance time in obese elderly adults found no differences between obese and non-obese individuals (Fjeldstad et al., 2008).

1.4 Obesity in Stair Descent

Most research on the daily activities of the obese has focused on gait, some on standing balance, little on sit-to-stand, and almost none on stair negotiation. In a study of joint kinematics of gait, obese subjects had more hip extension during the stance phase, less knee flexion at early stance and throughout the stance phase, and more plantar flexion at toe-off and throughout the stance phase compared to normal weight individuals (DeVita et al., 2003). Other researchers focusing on temporal measures of gait have found that obese individuals walk at a slower pace, have a longer stance phase duration, have a shorter swing phase duration, and greater time spent in double support compared to lean counterparts (Blaszczyk et al., 2011; Hulens et al., 2003; Spyropoulos et al., 1991). McGraw et al. (2000) found the same differences in gait of obese prepubertal boys, while also finding that obese boys had greater mediolateral COP displacement and variability. In addition, Fjeldstad et al. (2008) found obesity to be correlated with a higher incidence of stumbling and falls during walking.

Stair negotiation is a more challenging dynamic task than walking (Andriacchi et al., 1980; McFadyen et al., 1988; Kim et al., 2009; Lin et al., 2004), requiring larger muscle efforts, larger joint ranges of motion, and higher joint loads. However, very little research has been devoted to studying this activity with obese participants (Wearing et al., 2006). Falls on stairs is the third leading cause of accidental death in the United States, behind car accidents and poisonings (Jackson et al., 1995, Startzell et al. 2000), and 60% of those accidental stair falls occur during the transition phases on the first or last two steps (Jackson et al., 1995; Startzell et al., 2000). Falls are approximately three times more likely to occur during stair descent than ascent (Jackson et al., 1995; Svanstrom et al., 1974).

Many studies have investigated kinematics and kinetics of stair descent in healthy young and elderly adults. Literature is somewhat inconsistent in outcomes for this activity, although several studies have found similar results. Peak joint moments during stair descent of the stance leg typically occur near the time points of toe-off and touchdown of the swing leg (Andriacchi et al., 1980; Beaulieu et al., 2008; Christina et al., 2002; Kowalk et al., 1996; Novak et al., 2011; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002; Stacoff et al., 2007). Near the instance of swing leg toe-off, there is a hip adduction moment, knee flexion and adduction moment, and an ankle dorsiflexion and abduction moment (all moments are given as external) (Andriacchi et al., 1980; Beaulieu et al., 2008; Christina et al., 2002; Kowalk et al., 1996; Novak et al., 2011; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002; Stacoff et al., 2007). Also near the instance of swing leg toe-off, there are hip and knee flexion angles, ankle dorsiflexion angle, large vertical ground reaction force (GRF), and a posterior GRF of the stance leg (Andriacchi et al., 1980; Beaulieu et al., 2008; Christina et al., 2002; Kowalk et al., 1996; Novak et al., 2011; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002; Stacoff et al., 2007). Near the instance of swing leg touchdown, there is a hip adduction moment, knee flexion and adduction moment, and an ankle dorsiflexion and slight adduction moment leg (Andriacchi et al., 1980; Beaulieu et al., 2008; Christina et al., 2002; Kowalk et al., 1996; Novak et al., 2011; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002; Stacoff et al., 2007). Also near the instance of swing leg touchdown, there are larger hip and knee flexion angles, a larger dorsiflexion angle, slightly smaller vertical GRF, and an anterior GRF of the stance leg (Andriacchi et al., 1980; Beaulieu et al., 2008; Christina et al., 2002; Kowalk et al., 1996; Novak et al., 2011; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002; Stacoff et al., 2007). The hip flexion extension moment and mediolateral GRF, however, are highly variable throughout literature for stair descent.

Of the limited research done on obesity during stair descent, Spanjaard et al. (2008) found that normal weight subjects wearing a 20% added mass jacket had a longer first double support phase time and a larger first peak knee flexion moment near the touchdown phase, but no differences were found in step cadence, ankle flexion moment during the touchdown phase, or total stance phase duration. In a questionnaire study on difficulties during daily activities of obese women, trouble descending stairs was one of the most commonly checked tasks (Larsson et al., 2001). In an investigation of obese and normal weight children during stair descent, obese adolescents spent more time in double support, had a greater peak hip flexion moment, a smaller peak hip extension moment, and a greater peak knee extension moment (Strutzenberger et al., 2011). No differences were found in single support time, step width, joint angles in the sagittal or frontal planes (torso, pelvis, hip, knee, and ankle), GRF peaks normalized to body mass, or any other joint moments in the sagittal or frontal planes. It is important to note that the subjects in the Strutzenberger study were an average age of ten years.

1.5 Purpose

In daily activities, individuals must preserve postural stability in a multitude of dynamic environments that challenge the balance control system. Thus, there is a need to assess upright postural stability during various environmental situations that have the potential to cause instability. Studies analyzing biomechanical aspects of standing balance and stair climbing have been numerous in literature for normal weight individuals. Although single leg stance and stair descent have been shown to be more demanding tasks than bipedal stance or gait respectively, few studies have assessed the effects of balance performance and locomotion in the obese during these tasks (Hills et al., 2002; Wearing et al., 2006). The purpose of this study was to examine the stability and biomechanics of the torso, pelvis, and lower limbs during single leg stance and stair descent in normal and high BMI individuals. The primary research question was to discover whether there were differences in kinematic, kinetic, and temporospatial parameters between these groups that could affect their stability during these tasks. Based on the previous limited research on the obese performing these and other related activities, it was hypothesized that the overweight and obese group would exhibit diminished static and dynamic postural stability during single leg stance and stair descent, respectively.

CHAPTER II

MATERIALS AND METHODS

2.1 Subjects

Seventeen subjects participated in this study. A control group of eleven (seven male and four female) healthy young adults was compared with an aged matched group of six (five male and one female) overweight and obese subjects (referred to in the rest of this investigation as the high BMI group). The subjects were classified into two groups: a control group and a high BMI group. Subjects with a BMI between 18.5 and 24.9 were classified into the control group, while subjects with a BMI greater than 25 were classified into the high BMI group. Participants were excluded if they had a history of significant musculoskeletal injuries or surgeries, arthritis, or any other degenerative joint diseases, neuromuscular disorders, an inability to descend stairs without handrails, or were older than age 40 at the time of testing. Approval by the Institutional Review Board was granted for this study and participants signed an informed consent document prior to testing.

2.2 Equipment and Laboratory Layout

Data was collected by a licensed physical therapist and a graduate student prior to my involvement in this study. Prior to data collection, participants were asked to remove their shoes and socks and to change into tight fitting and minimal clothing to reduce the movement of and increase the visibility of the retroreflective markers. Retroreflective markers were placed on the skin over bony landmarks, palpated by an experienced physical therapist using double-sided tape. Markers were placed on the following landmarks (Appendix 4, Figure 15): left and right acromion processes, C7 of the spine, sacrum, left and right anterior and posterior superior iliac spines, inferior patellas, tibial tuberosities, medial and lateral femoral epicondyles, medial and lateral malleoli, calcanei, dorsum area of the foot placed so that a line with the calcaneus is parallel to the floor when the foot is flat, and 5th metatarsal heads. A virtual marker was created for the location of the first metatarsal heads using the anthropometric diagonal foot width method of Wunderlich et al. (2001). Rigid arrays of four markers each were placed on the thigh and shank segments using medical wrap.

A nine-camera motion analysis system (Qualisys AB, Gothenburg, Sweden) was used to collect 3D kinematic and kinetic data at 100 Hz. The cameras were calibrated before each trial using a wand that was moved through the lab space in view of the cameras. Stationary reference markers on an L-shaped frame were used to set the origin of the lab coordinate system. The experimental area consisted of a walkway with three imbedded force plates (OR6-7, AMTI, Watertown, MA, USA). Ground reaction force (GRF) data was collected from three force plates at 1000 Hz. A three-step wooden staircase (FP-Stairs, AMTI, Watertown, MA, USA) was used for the stair descent activity (Figure 1). The staircase has a step width of 61.0 cm, a rise of 17.8 cm, and a run of 28.0 cm (Croce et al., 2006). These dimensions are near the average of those found in literature and comply with the 2009 International Residential Code of residential staircase design (Cluff et al., 2011; IRC, 2009; Protopapadaki et al., 2007). The staircase was independently bolted to two of the force plates, with the first and third steps recorded by the same force plate and the second step recorded by the other. A mobile platform was placed above and behind the top step to mimic a fourth step/landing area with room to turn around to descend the staircase.

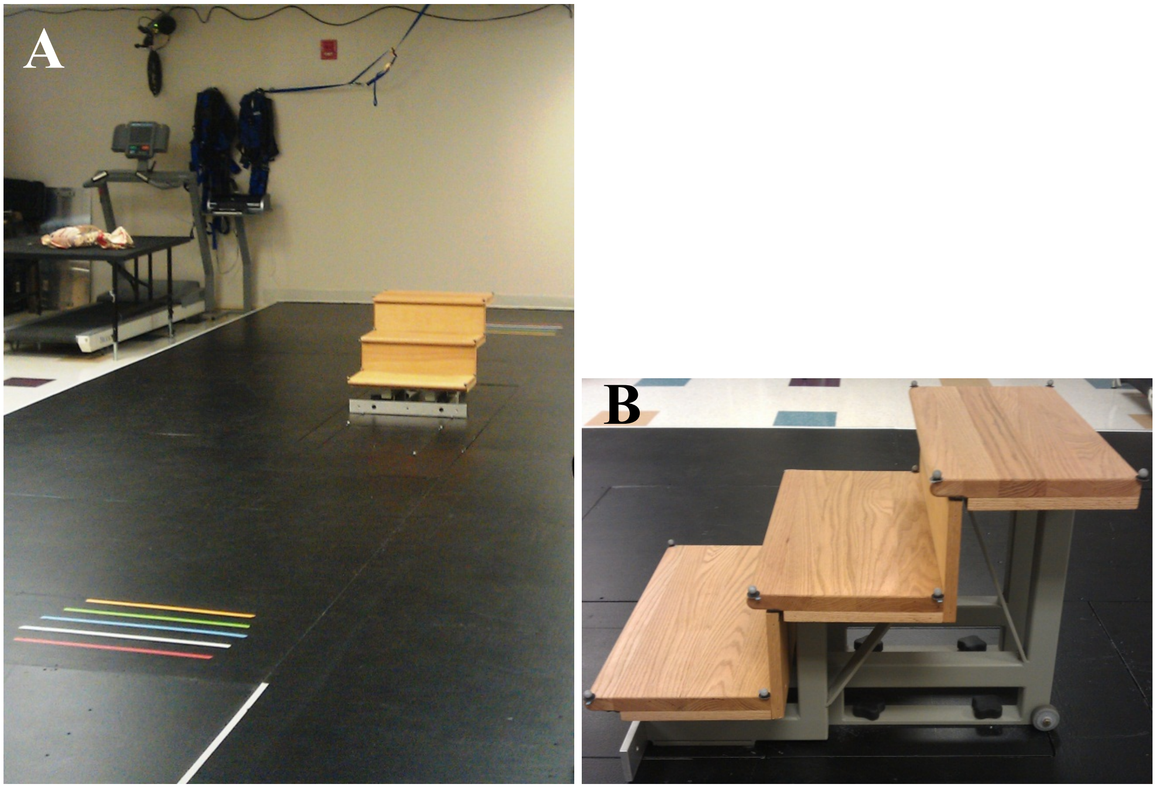


Figure 1. (A) Laboratory setup with walkway up to the staircase (minus the mobile platform). (B) Profile view of the staircase (FP-Stairs, AMTI); steps one and three are registered on force plate three under the third step and step two is registered on force plate two under step one.

2.3 Data Collection and Testing Protocol

Demographic and anthropometric measurements of the subjects were taken and BMI was calculated first. A static quiet standing trial of each subject was captured and used to create an anatomical model of the body segments that was applied to the movement files. Participants were allowed to practice single leg standing and stair negotiation prior to data collection. Subjects were then instructed to stand on two force plates, with one foot on each force plate, in a double support standing position. Concurrently at the initiation of data collection by the investigator, the subject was verbally asked to stand on one leg for a capture length of 30 seconds. A total of four single leg stance trials were captured, two for each leg. At the conclusion of the stance activity, subjects performed a stair negotiation task at a self-selected speed that consisted of walking on the walkway towards the staircase, ascending the stairs to the mobile platform above, descending the staircase, and then walking back to the starting position. Subjects then waited a brief period at the starting position until verbally asked by the investigator to repeat the task. Stair negotiation was done in a step over step manner and six trials of the task were collected. Rest periods were given if needed to avoid fatigue.

2.4 Dependent Variables

2.4.1 Single Leg Stance

The most commonly used model of postural control is the inverted pendulum model (Gage et al, 2004; Kuo et al., 2005; Kuo et al., 2007; Winter et al., 1995; Winter et al., 2003). The inverted pendulum model method, using the variables of margin of stability (MoS) and time to contact (TtC) (Hof et al., 2005), was used to help quantify single leg stance stability. This model's validity and balance discrimination capabilities have been proven during functional activities of previous studies (Abuzayan et al., 2013; Arampatzis et al., 2008; Bosse et al., 2012; Bruijn et al., 2013; Karamanidis et al., 2008; Mademli et al., 2008). MoS was defined as the shortest distance between the position of the extrapolated center of mass (XCOM) and the boundaries of the base of support (BoS) of the right foot. XCOM was calculated using the following equation:

$$\mathrm{XCOM} = \mathrm{pCOM} + \frac{\mathrm{vCOM}}{\sqrt{\mathrm{g}/l}} \tag{1}$$

where pCOM is the vertical projection of the position of the center of mass on the ground, vCOM is the velocity of the center of mass in the lab floor X-Y plane, g is gravitational acceleration, and *l* is the distance between the positions of the center of mass and the center of the right ankle joint (Figure 2). The five boundaries of the BoS of

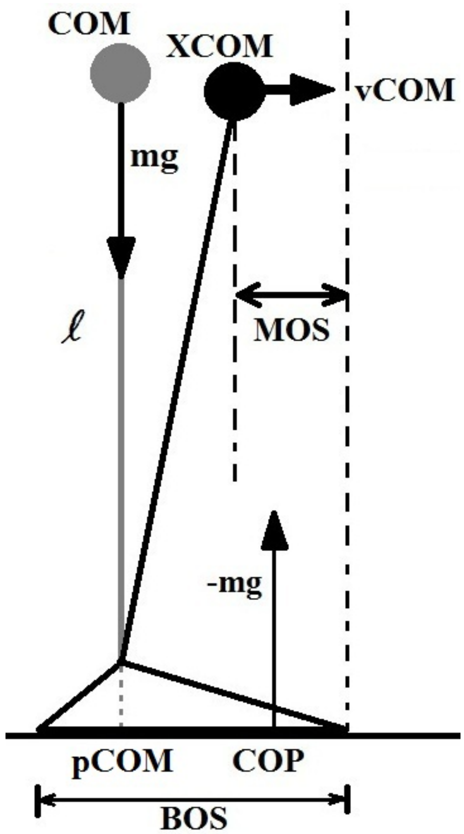


Figure 2. Inverted pendulum model where the body is modeled as a single mass (COM) rotating at the center of the ankle joint with a pendulum length of *l*. The base of support (BOS) is taken as the outline of the foot. The velocity of the center of mass (vCOM) is directed anteriorly, leading the extrapolated center of mass (XCOM) position to be anterior to that of the COM. The distance between the projection of the XCOM to the floor and the closest base of support is the variable called margin of stability (MOS).
COM = center of mass, XCOM = extrapolated center of mass, vCOM = velocity of the center of mass, m = mass, g = gravitational acceleration, *l* = pendulum length, MOS = margin of stability, pCOM = projection of the COM to the floor, COP = center of pressure, BOS = base of support.

the right foot (Figure 3) were defined as the lines between the marker positions of the 1st and 5th metatarsal heads, 5th metatarsal head and lateral malleoli, lateral malleoli and calcaneus, calcaneus and medial malleoli, and medial malleoli and 1st metatarsal head

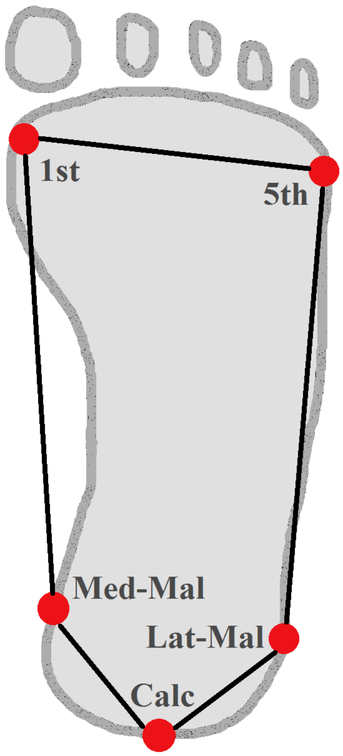


Figure 3. Base of support outline (solid black lines) connecting the markers of the right foot; 1st = first metatarsal head, 5th = fifth metatarsal head, Med-Mal = medial malleolus, Lat-Mal = lateral malleolus, Calc = calcaneus.

(Schloemer et al., 2013). Positive MoS values indicate that the XCOM position is within the base of support, yielding a posturally stable situation. Negative MoS values indicate that the XCOM position is outside of the BoS, yielding a posturally unstable situation and indicating a need to change the BoS. The TtC variable of this method was defined as the time it would take the XCOM to reach the closest BoS boundary and was calculated using the following equation:

$$\mathrm{TtC} = \frac{\mathrm{MoS}}{\mathrm{vCOM}} \tag{2}$$

This variable is the time that the BoS boundary would be reached without corrective action and corresponds to the time that corrections to the vCOM or COM position can be made without having to move the trunk or the arms, or change the BoS by taking a step.

Negative values of both margin of stability and time to contact theoretically indicate that postural stability will not be recovered without such actions (Bruijn et al., 2013).

Since the transition from double leg to single leg stance is the period of highest body movement during this activity (Dingenen et al., 2013; Levin et al., 2012), the speed of this transition will greatly affect the XCOM, MoS, and TtC variables. To help eliminate the chance that differences between groups in single leg stance could be due to the subjects' speed of this transition, a variable called the time to stabilization (TTS) (Colby et al., 1999; Wikstrom et al., 2005) was calculated to determine a relatively stable starting point for data analysis. TTS was determined using the signal of the vertical GRF starting at toe-off of the lift leg. Calculation of this variable was done by obtaining a cumulative average of the signal by successively adding one data point at a time, where the last calculation was the total average of all data points in the series. The signal was considered to be stable when the sequential average reached and stayed within one-quarter standard deviation of the series average. The TTS was defined as the time at which this point occurred. All variables were measured from the TTS to fifteen seconds after this point, for a total trial analysis time of fifteen seconds.

The anteroposterior and mediolateral positions of the center of pressure (COP) were obtained from the force plates and were used to calculate the total path distance of the COP throughout the trial. Velocity of the center of pressure (vCOP) was calculated by the total COP path distance divided by the trial length of fifteen seconds. This variable using this definition is one of the most commonly used parameters for analyzing standing (Kim et al., 2009; Lin et al., 2008; Piirtola et al., 2006; Raymakers et al., 2005; Prieto et al., 1996; Ruhe et al., 2010), has been found to be the most reliable and informative COP parameter (Lin et al., 2008; Piirtola et al., 2006; Raymakers et al., 2005), and has been shown to help predict fall risk in the elderly (Piirtola et al., 2006). The magnitude of the vCOM was calculated throughout the trial in the X-Y plane of the lab floor. The variability of the vertical GRF was also determined using the standard deviation. This variable has been found to be a sensitive and reliable measure to detect changes in steadiness during single leg standing (Goldie et al., 1989; Goldie et al., 1992).

The investigated parameters of single leg stance were transformed from the origin of the lab coordinate system to match the orientation of the right foot with the new origin at the center of the ankle joint projected onto the lab floor. The center of the ankle joint was defined as the point midway between the medial and lateral malleoli. Two trials per subject of single leg standing on the right limb were averaged and used for the analysis of this activity.

2.4.2 Stair Descent

All participants descended the stairs in a step over step manner. The analysis period of stair descent was the total stance phase of the right foot on the second step from the events of right touchdown to right toe-off (Figure 4, Figure 5). The XCOM, MoS, and TtC were calculated in the anterior-posterior direction using the anterior-posterior pCOM

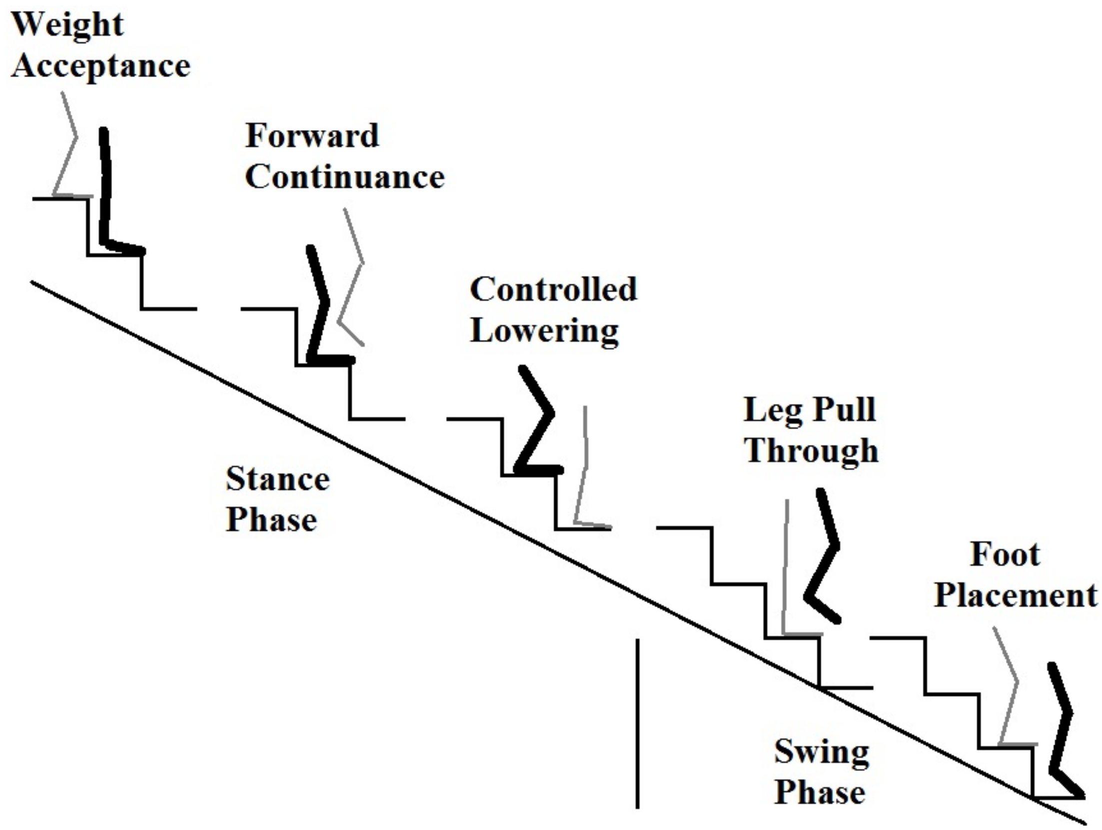


Figure 4. Schematic of the two phases and five sub-phases during step over step stair descent of a three-step staircase. The analysis period is the entire stance phase of the right leg. Bold leg is the lead (analyzed) leg.

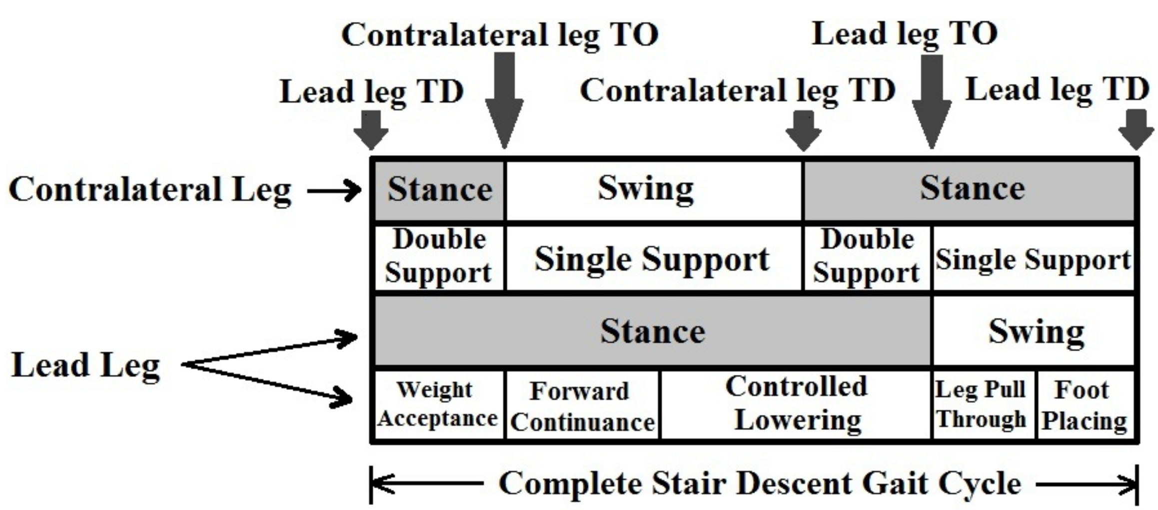


Figure 5. Phases of stair descent during one complete cycle of the lead leg (right leg). Analysis is of the right leg during the stance phase. TD = touchdown, TO = toe-off.

and vCOM. Due to the position of the COP on the stairs being limited by the length of edge of the second step (Bosse et al., 2012). The instantaneous values of anterior-posterior vCOM, MoS, and TtC were determined at the initiation of the first double support phase (i.e. at right foot touchdown, corresponding to 0% of the stance phase) and at the initiation of the single support phase (i.e. at left foot toe-off, near 25% of the stance phase). These instances correspond to time points during the stance phase when the anterior base of support is on the second step.

The anteroposterior and mediolateral velocities of the COP were obtained from the total path distance in each direction divided by the duration of the stance phase in seconds. Average anteroposterior vCOM was determined over the right foot stance phase. Step width was defined as the mediolateral distance between the center of the ankle of the right foot at left toe-off from the third step and the center of the ankle of the left foot at right toe-off from the second step (Stolze et al., 1998). The times of right leg stance phase, double support, and right leg single support were calculated. The external joint moments at the ankle, knee, and hip joints of the right leg were determined through the inverse dynamics model. The angles and ranges of motion of the ankle, knee, and hip joints, as well as the pelvis and torso, were also calculated. Joint moment and angle definitions are as follows: hip defined as the thigh relative to the pelvis, knee defined as the shank relative to the thigh, and ankle defined as the foot relative to the shank. Pelvis angle was defined relative to the lab coordinate system. Torso flexion-extension angle was defined as the angle between a line connecting the C7 and sacrum markers projected onto the sagittal plane relative to the vertical axis of the lab (van der Esch et al., 2011). Torso right-left tilt angle was defined as the angle between the X-Y plane of the lab floor and a line connecting the acromion markers. All moments and angles were calculated in the sagittal and frontal planes at the instances of left toe-off from the third step and left touchdown onto the first step (Figure 5). These time points were chosen because literature on healthy normal adults during stair descent reveals that joint moment peaks usually occur near these instances (Andriacchi et al., 1980; Beaulieu et al., 2008; Christina et al., 2002; Kowalk et al., 1996; Novak et al., 2011; Protopapadaki et al., 2007; Reeves et al., 2008; Riener et al., 2002; Stacoff et al., 2007).

2.5 Statistical Analysis

Marker data was interpolated over a maximum of 10 frames, while marker and GRF were both filtered using a fourth order 10 Hz Butterworth filter (Schmid et al., 2002). Visual3D software (C-Motion, Germantown, MD, USA) was used to analyze the kinematic and kinetic data. A 6-degree of freedom (DOF) linked rigid segment 3 dimensional (3D) model consisting of eight segments, including the feet, shanks, thighs, pelvis, and trunk, was constructed in Visual3D. Stair gait events were determined using a GRF threshold of 10 newtons.

The mean values from two trials of each subject were used for both single leg stance and stair descent, and an ensemble average was used to calculate the group's average kinematic, kinetic, and temporo-spatial data. Data were resampled to 101 values corresponding to 100% of the right foot stance phase. GRF variables were normalized by subject body weight in newtons and presented in units of body weight (BW), while moment variables were normalized by subject height in meters and mass in kilograms. GRF data were defined with posterior, lateral, and vertical directions as positive. Moment and angle data were defined with flexion, adduction, dorsiflexion, right tilt, anterior tilt, and obliquity up as positive. All variables with dimensions of length and velocity were normalized by subject height, due to the difference found in height between groups. Results are presented as means and standard deviations. Normality of the dependent variables was verified using the Kolmogorov–Smirnov test. Statistical analyses were carried out using GraphPad Prism V6.0 (GraphPad Software, San Diego, CA, USA). Observed differences between the control and high BMI groups were tested using a Student's *t*-test.

CHAPTER III

RESULTS

Participants' information and physical characteristics were compared for differences between groups (Table 1). No difference was found in age between the two groups ($p = 0.208$). Control and high BMI groups differed in body height, body mass, and body mass index (Table 1). Therefore, all force data was normalized to body weight and given as multiples of body weight; moments were normalized to body height and body mass, resulting in units of Newton meters per kilogram per meter; variables with dimensions of length and velocity were normalized by body height, yielding dimensionless units for length terms and units of seconds^{-1} for velocity terms.

Table 1. Participant information and physical characteristics: mean (standard deviation).

	Control	**High BMI**	**p-value**
Age (years)	24.6 (2.6)	26.9 (5.0)	0.208
Height (m)	1.72 (0.087)	1.85 (0.077)	0.010 *
Body Mass (kg)	65.2 (10.2)	106.4 (19.1)	< 0.001*
Body Mass Index (kg/m^2)	21.9 (1.9)	31.1 (4.2)	< 0.001*

* Indicates parameters at a value of $p < 0.05$.

3.1 Single Leg Stance

The trajectory of the center of mass (COM) and extrapolated center of mass (XCOM) inside the outlined base of support (BoS) has been given (Figure 6) for one representative subject of the control group during one 15 second single leg standing trial.

The beginning of the analysis period corresponds with the most medial positions of the COM and XCOM paths (Figure 6). This instance is when the body is still in transition from double leg to single leg stance and has not yet come to a relatively equilibrium single leg standing pattern. The XCOM follows the path of the COM quite closely, with deviations being due to the inclusion of the velocity of the center of mass (vCOM) in the calculation of the XCOM variable. For processing, the BoS boundary was approximated by a set of five straight lines connecting the five markers of the right foot.

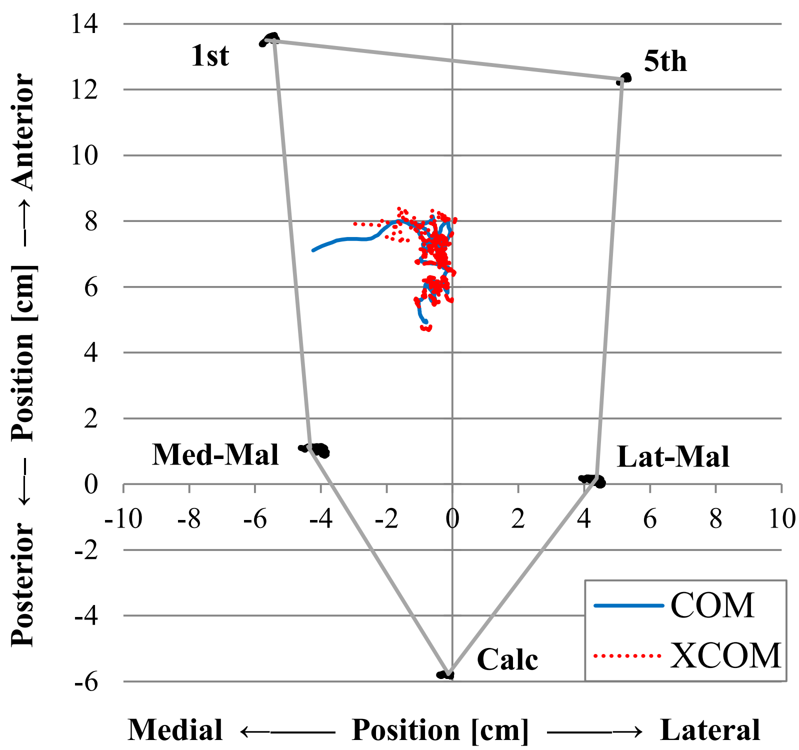


Figure 6. Mediolateral and anteroposterior path of the center of mass (COM, thick blue line) and extrapolated center of mass (XCOM, dashed red line), with the five markers of the right foot and the outline of the base of support (thin gray lines) for one representative subject during one 15 second single leg standing trial. The origin is the center of the ankle joint projected onto the lab floor. 1st = first metatarsal head, 5th = fifth metatarsal head, Med-Mal = medial malleolus, Lat-Mal = lateral malleolus, Calc = calcaneus.

An alternative representation of the COM and XCOM interaction with the base of support is included (Figure 7) for one representative subject of the control group during one 15 second single leg standing trial. The anterior and posterior boundaries of the BoS, defined as the line connecting the 1st and 5th metatarsal head markers and the line parallel to the mediolateral axis through the calcaneus marker respectively, are plotted together with the anteroposterior positions of the COM and XCOM over time (Figure 7 (A)). The lateral and medial boundaries of the BoS, defined as the line connecting the 5th metatarsal head and lateral malleolus markers and the line connecting the 1st metatarsal head and medial malleolus markers respectively, are plotted together with the

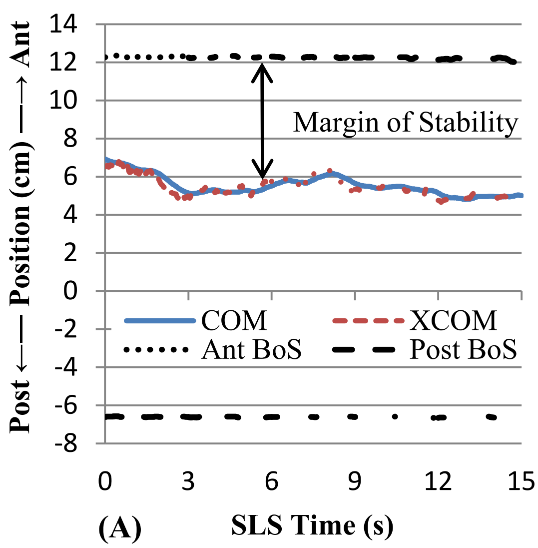


Figure 7. Data for one representative control group subject during one 15 sec. single leg standing (SLS) trial. (A) A/P position of the center of mass (COM), extrapolated center of mass (XCOM), anterior base of support (Ant BoS) defined as the line connecting the 1st and 5th metatarsal head markers, and posterior (Post) BoS defined as the line parallel to the M/L axis through the calcaneus marker, over the SLS time. The distance between the XCOM and the closest BoS is the margin of stability (MoS) at that instance. (B) M/L position of the COM, XCOM, lateral (Lat) BoS defined as the line connecting the 5th metatarsal head and lateral malleolus markers, and medial (Med) BoS defined as the line connecting the 1st metatarsal head and medial malleolus markers, over the SLS time. The distance between the XCOM and the closest BoS is the MoS at that instance.

mediolateral positions of the COM and XCOM over time (Figure 7 (B)). Another variable displayed (Figure 7) is a graphical representation of the definition of the margin of stability (MoS), which is defined as the distance between the XCOM position and the closest BoS line for every instance of the trial. For this trial of this subject, the position of the XCOM was always closest to the anterior and medial BoS boundaries.

For the calculation of the stability variables, the positions of markers, ground reaction force (GRF), center of pressure (COP), and COM were all transformed from the origin of the lab coordinate system to match the orientation of the right foot with the new origin at the center of the ankle joint projected onto the lab floor. No differences were found for any of the normalized single leg stance stability parameters (Table 2), indicating a state of similar postural stability between the normal weight and high BMI groups for the period of single leg stance between the time to stabilization and 15 seconds after.

Table 2. Single leg stance stability parameters.

	Control	**High BMI**	**p-value**
MoS (d.u.)	0.020 (0.0031)	0.020 (0.0032)	0.990
TtC (s)	6.98 (2.24)	6.34 (2.22)	0.580
vCOM (s^{-1})	0.0049 (0.00085)	0.0053 (0.00097)	0.368
COP Path (d.u.)	0.410 (0.134)	0.435 (0.103)	0.695
vCOP (s^{-1})	0.0273 (0.00896)	0.0290 (0.00690)	0.695
V-GRF SD (BW)	0.00598 (0.00210)	0.00521 (0.00115)	0.420
TTS (s)	2.22 (0.731)	2.81 (1.26)	0.234

Values are means (standard deviations). MoS = margin of stability, TtC = time to contact, vCOM = velocity of the center of mass in the mediolateral and anteroposterior directions, COP Path = total distance of the center of pressure path, vCOP = velocity of the center of pressure, V-GRF SD = vertical ground reaction force standard deviation, TTS = time to stabilization; d.u. = dimensionless units, s = seconds, BW = multiple of body weight. All parameters were transformed from the origin of the lab coordinate system to match the orientation of the right foot with the new origin at the center of the ankle joint projected onto the lab floor.
* Indicates parameters at a value of $p < 0.05$.

3.2 Stair Descent

The anteroposterior and mediolateral positions of the COM and XCOM in proximity to the outlined geometry of the right foot is given (Figure 8) for one representative subject during the stance phase of one stair descent trial. The instance of right touchdown (Figure 8 (A)) (initiation of double support, corresponding to the start of the analyzed stance phase) and the instance of left toe-off (Figure 8 (B)) (initiation of single support, corresponding to 23.7% and 25.6% of the analyzed stance phase for the control and high BMI groups respectively) were captured during the descent cycle. The

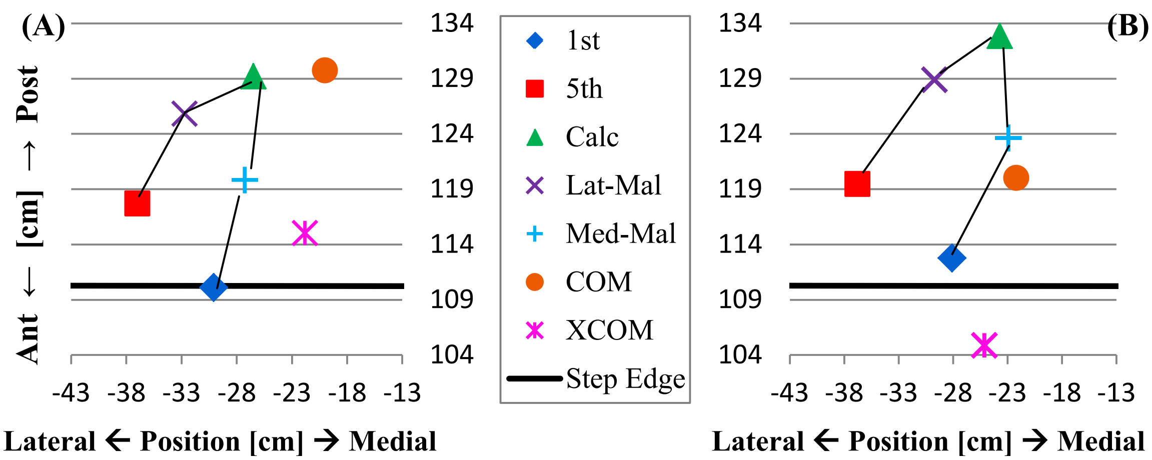


Figure 8. From one stair descent trial of one representative subject; anterior direction is decreasing. (A) Anteroposterior and mediolateral position of the center of mass (COM), extrapolated center of mass (XCOM), and right foot markers in relation to the edge of the step at the instance of right touchdown. (B) A/P and M/L position of the COM, XCOM, and right foot markers in relation to the edge of the step at the instance of left toe-off.

XCOM position is anterior to the position of the COM for both instances, as would be expected with the vCOM being directed anteriorly. For analysis, the BoS boundary was taken as the anteroposterior position of the edge of the step.

The ensemble averages of the anteroposterior path of the XCOM and COM for the control and high BMI groups during the stance phase of the right leg were calculated (Figure 9). Two interesting observations reveal (Figure 9) that the XCOM is always anterior to the COM position (due to the use of the vCOM in the XCOM calculation) and that the control group is always anterior to the high BMI group for both variables. The average position of the COM is posterior to the edge of the step, while the average position of the XCOM is anterior to the edge of the step at left toe-off for both groups. The average positions of the COM and XCOM are anterior to the edge of the step at left touchdown for both groups.

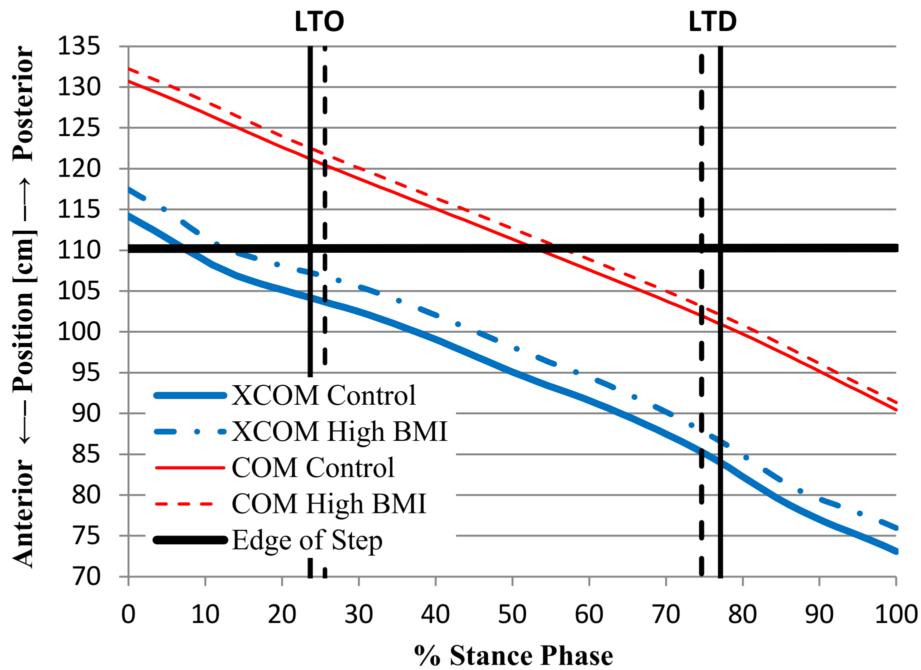


Figure 9. Ensemble average of the anteroposterior path of the extrapolated center of mass (XCOM) and center of mass (COM) during the stance phase of the right leg. Solid vertical lines are the points of left toe-off (LTO) and left touchdown (LTD) of the control group (23.7% and 77.1%). Dashed vertical lines are the points of LTO and LTD of the high BMI group (25.6% and 74.7%). 0% = right touchdown, 100% = right toe-off.

The spatio-temporal and dynamic stability parameters were analyzed for the right limb during stair descent (Table 3) and found to be different throughout the stance phase and at the instances of right touchdown and left toe-off. The high BMI group had 25% longer double support ($p = 0.021$) and 16% longer total stance phase times ($p = 0.022$) compared to the control group. The high BMI group also had a 19-23% slower anterior vCOM ($p = 0.0003 – 0.007$) on average throughout stance phase and at both right touchdown and left toe-off.

Table 3. Spatio-temporal and stability parameters during stair descent.

	Control	High BMI	p-value
Spatio-Temporal Parameters			
Single Support (s)	0.36 (0.039)	0.39 (0.034)	0.174
Double Support (s)	0.32 (0.070)	0.40 (0.050)	0.021 *
Total Stance Phase (s)	0.68 (0.090)	0.79 (0.077)	0.022 *
Step Width (d.u.)	0.0625 (0.0168)	0.0523 (0.0283)	0.362
A/P vCOP (s^{-1})	0.49 (0.13)	0.55 (0.089)	0.302
M/L vCOP (s^{-1})	0.19 (0.052)	0.23 (0.033)	0.207
Avg. Anterior vCOM (s^{-1})	0.36 (0.049)	0.29 (0.039)	0.007 *
At Instance of RTD			
MoS (d.u.)	0.022 (0.022)	0.039 (0.019)	0.138
TtC (s)	0.042 (0.042)	0.084 (0.044)	0.074
Anterior vCOM (s^{-1})	0.34 (0.42)	0.27 (0.034)	0.003 *
At Instance of LTO			
MoS (d.u.)	-0.037 (0.022)	-0.019 (0.021)	0.123
TtC (s)	-0.10 (0.055)	-0.060 (0.058)	0.180
Anterior vCOM (s^{-1})	0.35 (0.051)	0.27 (0.040)	0.006 *

Values are means (standard deviations). A/P vCOP = anteroposterior center of pressure velocity, M/L vCOP = mediolateral center of pressure velocity, Avg. = average, vCOM = center of mass velocity, MoS = margin of stability, TtC = time to contact; RTD = right touchdown, LTO = left toe-off.
* Indicates parameters at a value of $p < 0.05$.

Peak external joint moments, normalized to body mass and height, and peak angles were determined for the sagittal and frontal planes during the stance phase of stair descent (Figure 10 A-J, Figure 11 A-F, Table 4). The only angle that was different between the groups was peak hip flexion angle ($p = 0.059$, 95% confidence interval (CI) = -11.449 – 0.249), being 20% larger in the high BMI group and peaks for both groups occuring at the end of the stance phase. The high BMI group had a 6 times larger peak hip flexion moment ($p < 0.001$), 6 times larger peak knee extension moment ($p < 0.001$), more than twice as large peak knee adduction moment ($p = 0.019$), and almost twice as large peak ankle dorsiflexion moment ($p < 0.001$) compared to the control group.

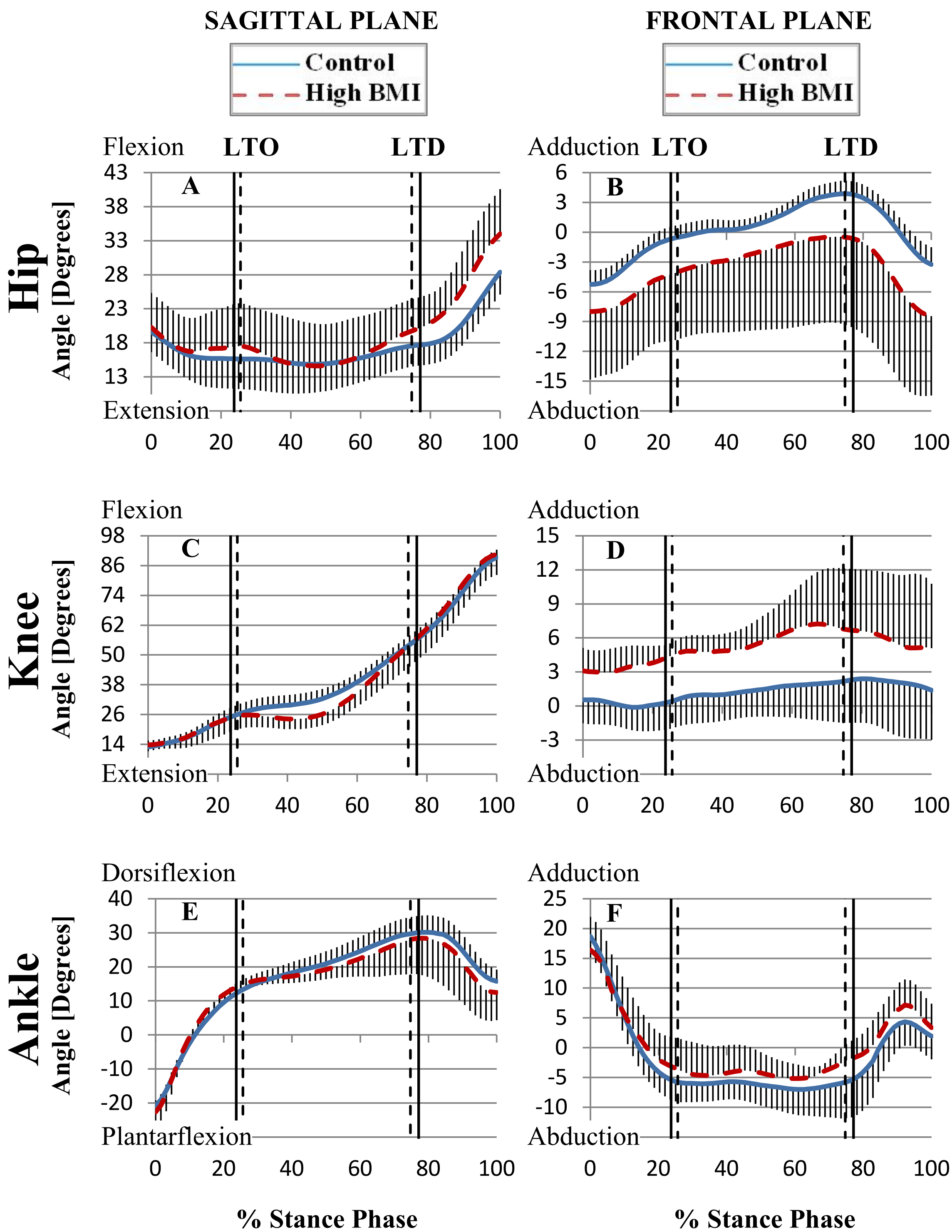


Figure 10 (A-F). Mean (with one-sided 95% confidence intervals) hip, knee, & ankle angles of right leg of control (blue solid line) & high BMI (dashed red line) groups in sagittal & frontal planes normalized to 100% stance phase. Solid vertical lines are the points of left toe-off (LTO) & left touchdown (LTD) of the control group (23.7% & 77.1%). Dashed vertical lines are the points of LTO & LTD of the high BMI group (25.6% & 74.7%). 0% = right touchdown, 100% = right toe-off.

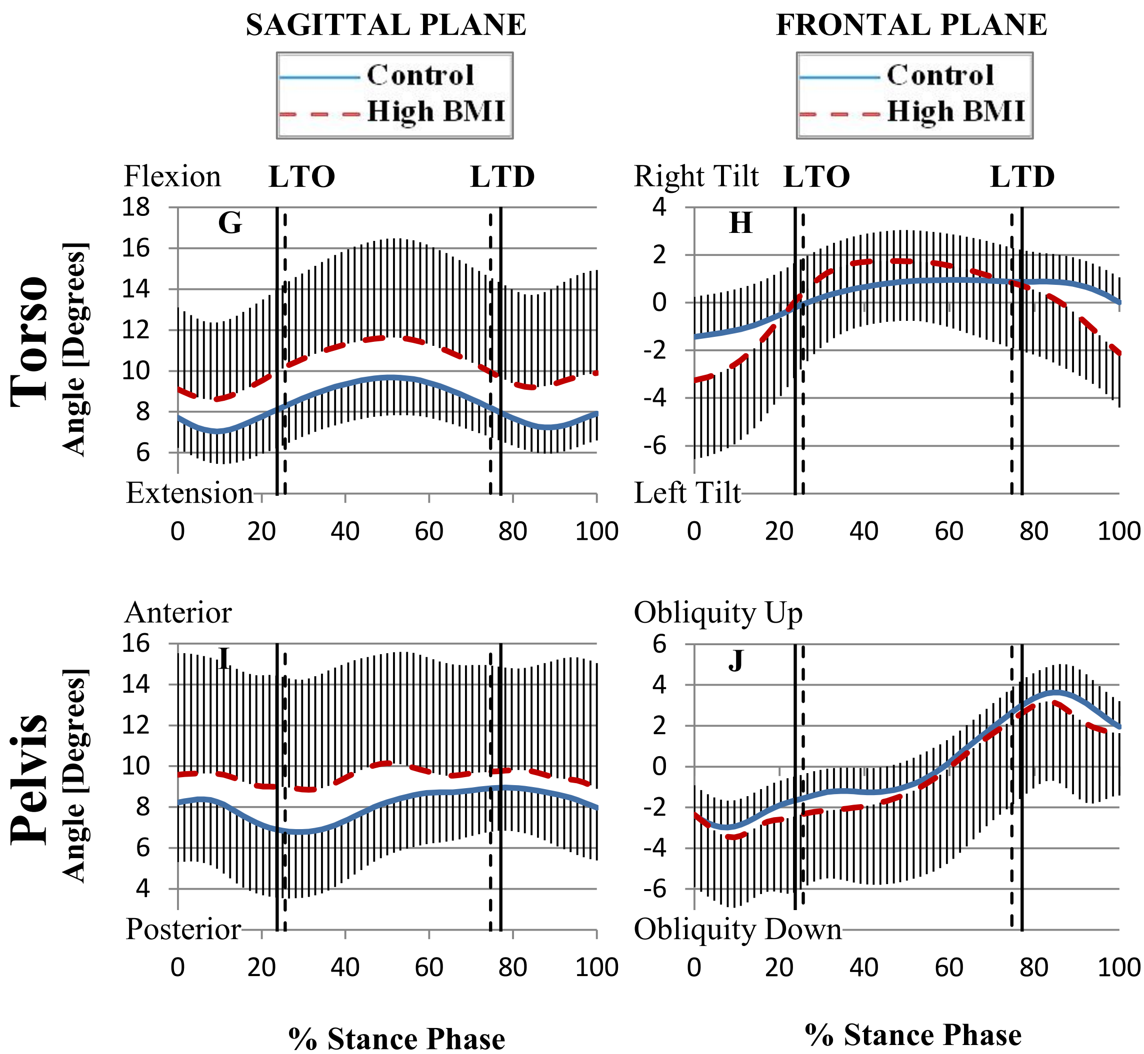


Figure 10 (continued G-J). Mean (with one-sided 95% confidence intervals) torso and pelvic angles of right leg of control (blue solid line) & high BMI (dashed red line) groups in sagittal & frontal planes normalized to 100% stance phase. Solid vertical lines are the points of left toe-off (LTO) & left touchdown (LTD) of the control group (23.7% & 77.1%). Dashed vertical lines are the points of LTO & LTD of the high BMI group (25.6% & 74.7%). 0% = right touchdown, 100% = right toe-off. Pelvic obliquity up = pelvis tilted to the left.

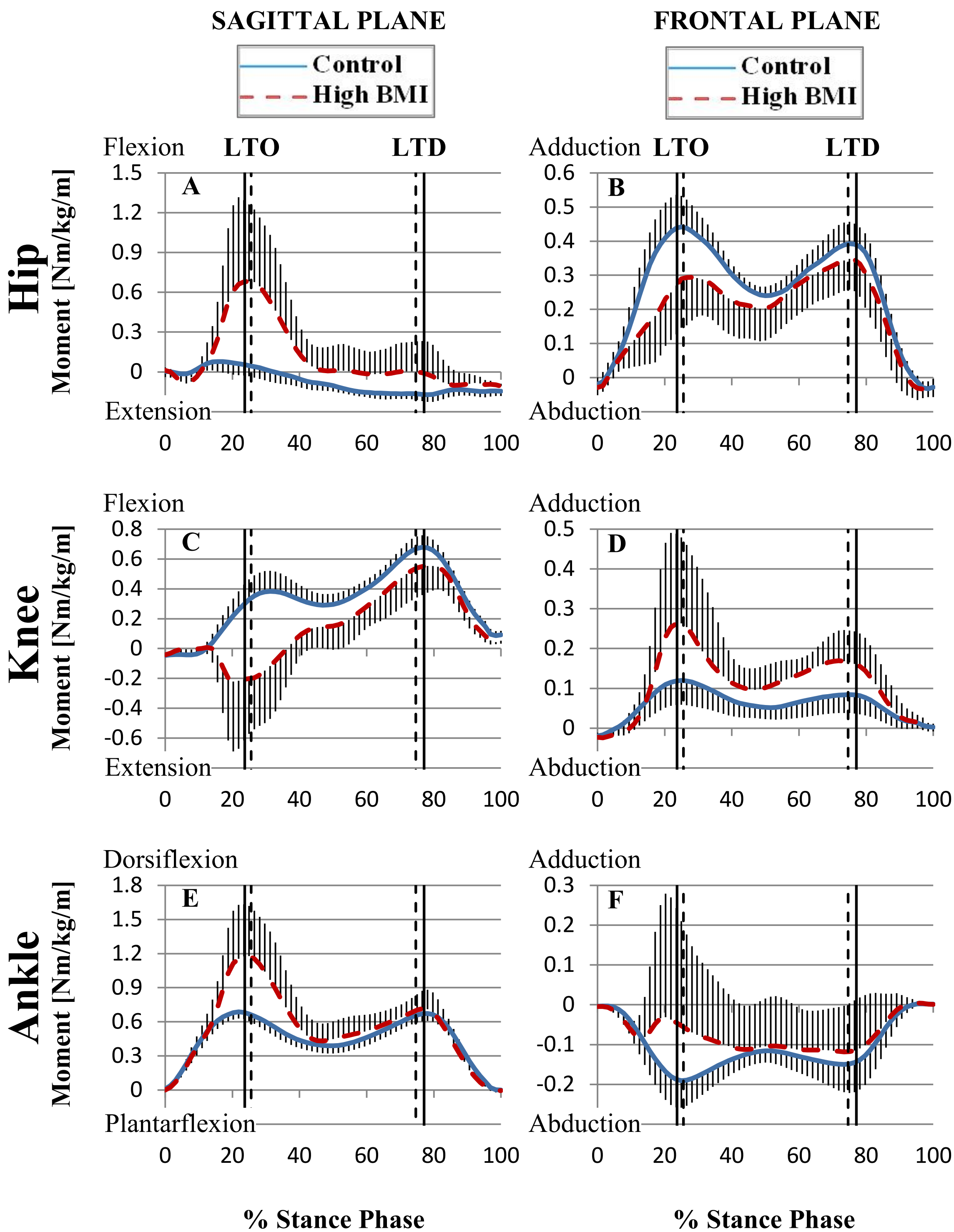


Figure 11. Mean (with one sided 95% confidence intervals) hip, knee, and ankle external joint moments (normalized to body mass and height) of the right leg of control (blue solid line) and high BMI (dashed red line) groups in both planes normalized to 100% stance phase. Solid vertical lines are the points of left toe-off (LTO) and left touchdown (LTD) of the control group (23.7% and 77.1%). Dashed vertical lines are LTO and LTD of the high BMI group (25.6% and 74.7%). 0% = right touchdown, 100% = right toe-off.

Table 4. Peak angles and peak external moments in the sagittal and frontal planes.

	Control	**High BMI**	**p-value**
Peak Angles (°)			
Hip Flexion	28.4 (4.9)	34.0 (6.3)	0.059
Hip Extension	13.4 (6.1)	14.1 (6.1)	0.829
Hip Adduction	4.1 (2.0)	0.6 (7.7)	0.168
Hip Abduction	-6.0 (1.9)	-10.3 (7.3)	0.079
Knee Flexion	89.3 (4.6)	90.6 (8.0)	0.674
Knee Extension	12.8 (3.7)	13.5 (2.2)	0.678
Knee Adduction	4.8 (4.7)	8.8 (3.9)	0.097
Knee Abduction	-2.7 (4.4)	0.9 (3.4)	0.104
Ankle Dorsiflexion	30.9 (7.4)	29.3 (10.1)	0.722
Ankle Plantarflexion	-20.9 (2.4)	-22.7 (5.3)	0.344
Ankle Adduction	18.7 (5.3)	16.4 (5.3)	0.412
Ankle Abduction	-9.5 (7.0)	-7.8 (1.8)	0.550
Torso Flexion	10.2 (2.4)	11.9 (4.5)	0.339
Torso Extension	6.3 (2.2)	8.0 (3.9)	0.248
Torso Right Tilt	2.3 (2.3)	2.7 (2.5)	0.790
Torso Left Tilt	-2.4 (2.1)	-3.9 (2.6)	0.217
Pelvic Anterior Tilt	10.3 (4.1)	11.2 (5.3)	0.698
Pelvic Posterior Tilt	5.8 (4.2)	7.9 (5.5)	0.403
Pelvic Obliquity Up	3.9 (1.9)	3.2 (3.7)	0.644
Pelvic Obliquity Down	-3.2 (1.9)	-3.8 (3.4)	0.636
Peak Moments (Nm/kg/m)			
Hip Flexion	0.133 (0.0941)	0.824 (0.547)	< 0.001*
Hip Extension	-0.223 (0.0533)	-0.192 (0.0858)	0.373
Hip Adduction	0.486 (0.106)	0.386 (0.102)	0.079
Hip Abduction	-0.0538 (0.0285)	-0.0551 (0.0207)	0.924
Knee Flexion	0.711 (0.108)	0.598 (0.150)	0.092
Knee Extension	-0.0714 (0.0348)	-0.445 (0.259)	< 0.001*
Knee Adduction	0.165 (0.0755)	0.333 (0.189)	0.019 *
Knee Abduction	-0.0323 (0.0119)	-0.0376 (0.0182)	0.478
Ankle Dorsiflexion	0.756 (0.0907)	1.35 (0.372)	< 0.001*
Ankle Plantarflexion	-0.00723 (0.00762)	-0.00799 (0.00225)	0.819
Ankle Adduction	0.0168 (0.0257)	0.118 (0.202)	0.115
Ankle Abduction	-0.218 (0.0984)	-0.188 (0.102)	0.563

Values are means (standard deviations). Moments were normalized to body mass and body height. Flexion, adduction, dorsiflexion, anterior tilt, right tilt, and obliquity up are positive. Pelvic obliquity up = pelvis tilted to the left.
* Indicates parameters at a value of $p < 0.05$.

Due to their larger peak hip flexion angle, the high BMI group had 33% more hip range of motion in the sagittal plane ($p = 0.032$) compared to the control group (Table 5). The high BMI group also had 26% less pelvic tilt range of motion ($p = 0.051$, CI = -0.008 – 2.408) in the sagittal plane.

Table 5. Range of motion (degrees) in the sagittal and frontal planes.

	Control	**High BMI**	**p-value**
Hip Flex/Ext	15.0 (3.8)	19.9 (4.6)	0.032 *
Hip Add/Abd	10.0 (2.8)	10.8 (3.3)	0.590
Knee Flex/Ext	76.5 (5.3)	77.1 (6.9)	0.848
Knee Add/Abd	7.5 (2.6)	7.9 (4.3)	0.811
Ankle Dorsi/Plantar	51.8 (6.2)	52.0 (8.3)	0.946
Ankle Add/Abd	28.2 (10.5)	24.2 (4.0)	0.383
Torso Flex/Ext	4.0 (1.4)	3.8 (1.7)	0.875
Torso Right/Left Tilt	4.7 (2.2)	6.5 (2.8)	0.158
Pelvic Ant/Post Tilt	4.5 (1.3)	3.3 (0.6)	0.051
Pelvic Up/Down Obliquity	7.0 (2.0)	7.0 (3.2)	0.986

Values are means (standard deviations).
* Indicates parameters at a value of $p < 0.05$.

In the sagittal plane at left toe-off (Table 6), the high BMI group had an eleven times larger knee adduction angle ($p = 0.015$) compared to controls. The high BMI group also had an eleven times larger hip flexion moment ($p < 0.001$), a knee extension instead of flexion moment ($p < 0.001$), and almost twice as large ankle dorsiflexion moment ($p < 0.001$) at left toe-off. In the case of the knee in the sagittal plane, the control group had a flexion moment while the high BMI group had an extension moment at left toe-off. In the frontal plane at left toe-off, the high BMI subjects had a 33% smaller hip adduction moment ($p = 0.050$, CI = -0.00016 – 0.28616). At left touchdown, the high BMI group had a hip flexion instead of extension moment ($p = 0.025$) and a 92% larger knee adduction moment ($p = 0.047$) compared to the controls. In the case of the hip in the sagittal plane at left touchdown, the control group had an extension moment while the high BMI group had a flexion moment. No differences were found in the angles of the joints or segments at left touchdown.

Table 6. External moments and angles in the sagittal and frontal planes at the instances of left toe-off (LTO) and left touchdown (LTD).

	Control	High BMI	p-value
Angles at LTO (°)			
Hip Flex/Ext	15.6 (6.6)	17.3 (5.7)	0.603
Hip Add/Abd	-0.8 (1.8)	-3.8 (6.1)	0.148
Knee Flex/Ext	24.7 (4.8)	25.0 (5.2)	0.879
Knee Add/Abd	0.4 (3.5)	4.4 (0.9)	0.015 *
Ankle Dorsi/Plantar	12.1 (4.4)	14.4 (1.0)	0.239
Ankle Add/Abd	-5.2 (4.2)	-3.7 (4.7)	0.521
Torso Flex/Ext	8.2 (2.8)	10.1 (3.9)	0.261
Torso Right/Left Tilt	-0.2 (3.0)	0.3 (3.2)	0.764
Pelvic Ant/Post Tilt	7.0 (5.2)	8.9 (5.0)	0.468
Pelvic Obliquity	-1.7 (1.9)	-2.2 (3.3)	0.690
Moments at LTO (Nm/kg/m)			
Hip Flex/Ext	0.0625 (0.133)	0.715 (0.508)	< 0.001*
Hip Add/Abd	0.438 (0.133)	0.295 (0.131)	0.050
Knee Flex/Ext	0.285 (0.166)	-0.247 (0.360)	< 0.001*
Knee Add/Abd	0.125 (0.0933)	0.265 (0.208)	0.072
Ankle Dorsi/Plantar	0.686 (0.104)	1.21 (0.404)	< 0.001*
Ankle Add/Abd	-0.184 (0.0907)	-0.0429 (0.269)	0.128
Angles at LTD (°)			
Hip Flex/Ext	17.7 (4.7)	19.7 (4.6)	0.391
Hip Add/Abd	3.8 (2.0)	-0.5 (8.4)	0.119
Knee Flex/Ext	56.1 (6.5)	53.6 (7.7)	0.483
Knee Add/Abd	2.3 (5.5)	6.7 (5.2)	0.129
Ankle Dorsi/Plantar	29.8 (7.4)	27.9 (9.2)	0.646
Ankle Add/Abd	-5.0 (9.5)	-2.2 (2.7)	0.491
Torso Flex/Ext	8.0 (2.0)	10.0 (4.3)	0.204
Torso Right/Left Tilt	0.8 (2.0)	0.9 (2.6)	0.909
Pelvic Ant/Post Tilt	8.9 (3.2)	9.7 (5.0)	0.689
Pelvic Obliquity	3.0 (1.8)	2.2 (3.8)	0.564
Moments at LTD (Nm/kg/m)			
Hip Flex/Ext	-0.173 (0.0722)	0.00851 (0.226)	0.025 *
Hip Add/Abd	0.381 (0.0997)	0.347 (0.0869)	0.485
Knee Flex/Ext	0.662 (0.103)	0.530 (0.190)	0.081
Knee Add/Abd	0.0875 (0.0734)	0.168 (0.0734)	0.047 *
Ankle Dorsi/Plantar	0.643 (0.111)	0.691 (0.149)	0.463
Ankle Add/Abd	-0.137 (0.117)	-0.113 (0.114)	0.686

Values are means (standard deviations). Moments were normalized to body mass and body height. Flexion, adduction, dorsiflexion, anterior tilt, right drop, and obliquity up are positive. LTO = left toe-off, LTD = left touchdown.
* Indicates parameters at a value of $p < 0.05$.

This study found multiple angles that were correlated to moments at left toe-off for high BMI subjects (Table 7). These variables were: hip abduction angle positively correlated with hip flexion and knee extension moments, torso flexion angle positively correlated with hip flexion, knee extension, and ankle dorsiflexion moments, pelvic obliquity down (pelvis tilted to the right side) angle positively correlated with hip flexion, knee extension, and ankle dorsiflexion moments. No other angles at left toe-off or left touchdown were highly correlated to the moments found to be different in the high BMI group at those instances.

Table 7. Regression analysis of angles with external moments found to be correlated for the high BMI group at left toe-off (LTO).

Angles at LTO	**Moments at LTO**		
	Hip Flexion	**Knee Extension**	**Ankle Dorsiflexion**
Hip Abduction	0.84 (0.007) *	0.63 (0.038) *	0.50 (0.078)
Torso Flexion	0.81 (0.009) *	0.86 (0.005) *	0.58 (0.048) *
Pelvic Obliq. Down	0.96 (< 0.001)*	0.87 (0.004) *	0.86 (0.005) *

Values are Adjusted R^2 (p-value). Pelvic Obliq. Down = pelvis tilted to the right.
* Indicates parameters at a value of $p < 0.05$.

The mediolateral, anteroposterior, and vertical GRFs were evaluated and normalized to body weight for dimensionless units given as multiples of body weight (Figure 12). A 17% smaller peak anterior GRF ($p = 0.033$) was found for the high BMI group compared to the control (Table 8). The anterior GRF at left touchdown was also 26% smaller ($p = 0.015$) for the high BMI group.

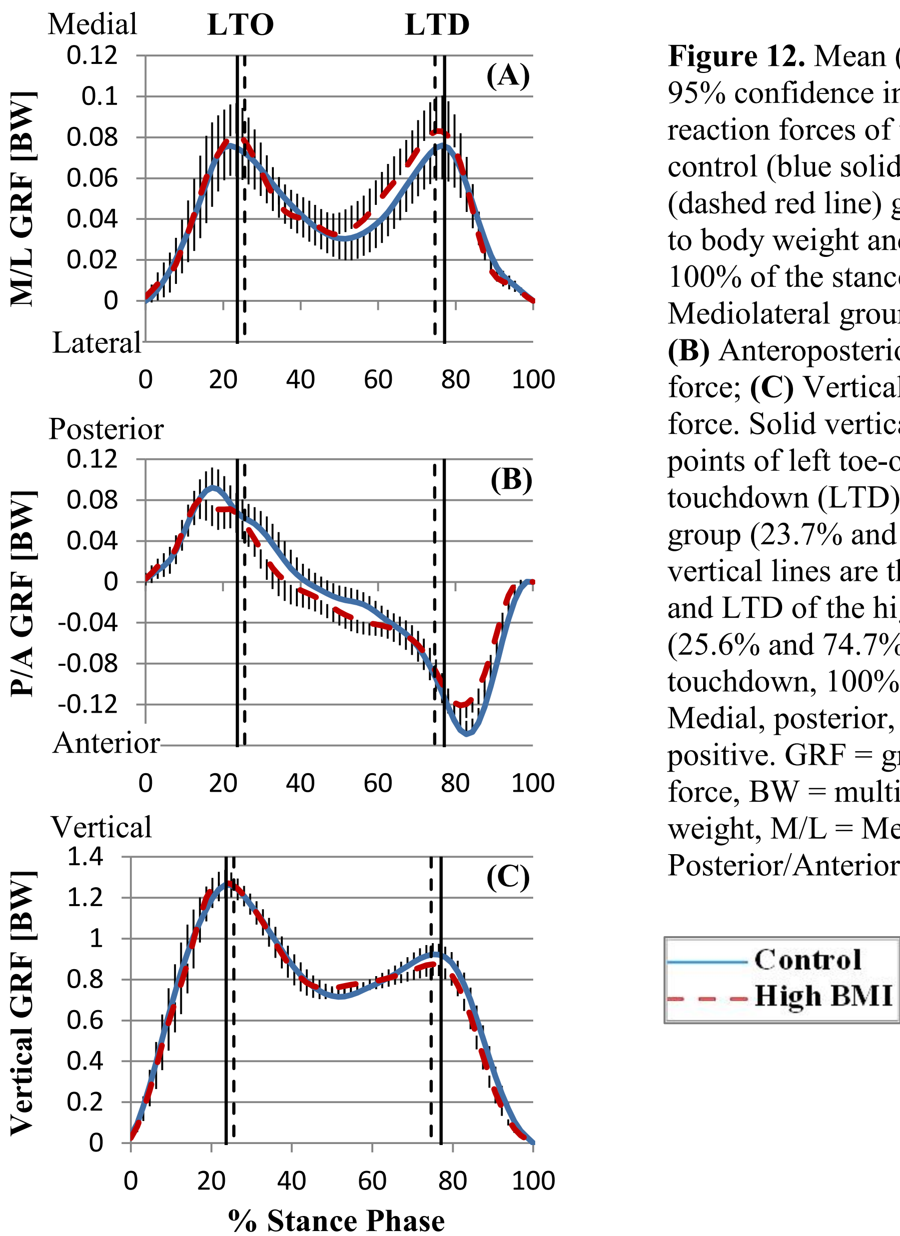


Figure 12. Mean (with one sided 95% confidence intervals) ground reaction forces of the right leg of the control (blue solid line) and high BMI (dashed red line) groups normalized to body weight and normalized to 100% of the stance phase. **(A)** Mediolateral ground reaction force; **(B)** Anteroposterior ground reaction force; **(C)** Vertical ground reaction force. Solid vertical lines are the points of left toe-off (LTO) and left touchdown (LTD) of the control group (23.7% and 77.1%). Dashed vertical lines are the points of LTO and LTD of the high BMI group (25.6% and 74.7%). 0% = right touchdown, 100% = right toe-off. Medial, posterior, and vertical are positive. GRF = ground reaction force, BW = multiples of body weight, M/L = Medial/Lateral, P/A = Posterior/Anterior.

Table 8. Ground reaction force (GRF) peaks and at the instances of left toe-off (LTO) and left touchdown (LTD).

	Control	High BMI	p-value
GRF Peaks (BW)			
Medial	0.0931 (0.0172)	0.0960 (0.00997)	0.705
Posterior	0.102 (0.0300)	0.0953 (0.0113)	0.584
Anterior	-0.156 (0.0176)	-0.130 (0.0290)	0.033 *
Vertical	1.30 (0.103)	1.33 (0.0973)	0.649
GRFs at LTO (BW)			
Medial/Lateral	0.0755 (0.0196)	0.0760 (0.0171)	0.955
Posterior/Anterior	0.0718 (0.0320)	0.0522 (0.0175)	0.189
Vertical	1.25 (0.0871)	1.28 (0.0870)	0.580
GRFs at LTD (BW)			
Medial/Lateral	0.0700 (0.0238)	0.0820 (0.0154)	0.288
Posterior/Anterior	-0.112 (0.0240)	-0.0826 (0.0136)	0.015 *
Vertical	0.898 (0.0985)	0.867 (0.104)	0.549

Values are means (standard deviations). GRFs were normalized to body weight to resulting in dimensionless units listed as multiples of body weight. Medial, posterior, and vertical are positive. GRFs = ground reaction forces, BW = multiples of body weight, LTO = left toe-off, LTD = left touchdown.
* Indicates parameters at a value of $p < 0.05$.

CHAPTER IV

DISCUSSION

The purpose of this study was to examine the effects of body mass on stability and biomechanical parameters of the lower limb for a group of normal weight and a group of high BMI individuals during single leg stance and stair descent. This is one of the few studies to provide a biomechanical analysis for the overweight population performing these tasks. No differences were found for the parameters of single leg standing, however analysis of stair descent revealed distinct differences in mechanics between the two groups.

4.1 Single Leg Stance

It was hypothesized that the overweight group would exhibit diminished static postural stability during single leg standing, but this study of a small number of subjects did not provide evidence to support this hypothesis. The variables of margin of stability (MoS) and time to contact (TtC) (Hof et al., 2005) using the inverted pendulum model method were used to help quantify single leg stance stability. The MoS and the center of mass velocity (vCOM), which is used in the MoS and TtC calculations, were normalized to subject height. No differences were found in any of these three parameters between groups. Total center of pressure (COP) path distance and velocity of the center of pressure (vCOP) were also normalized to subject height. Neither of these variables showed differences between groups. Differences were also not found for the parameters of standard deviation of the vertical ground reaction force (GRF) normalized to body weight or the time to stabilization (TTS). Contrary to the hypothesis, these results imply being overweight or obese did not impair postural stability during single leg standing.

4.1.1 Margin of Stability, Time to Contact, and Center of Mass Velocity

Many studies have used MoS and TtC to assess postural control during various activities, mainly comparing healthy young to elderly participants, but also in amputees, individuals with musculoskeletal conditions, and anterior cruciate ligament deficient subjects (Bierbaum et al., 2011; Curtze et al., 2011; Hof et al., 2010; Karamanidis et al., 2008; Lugade et al., 2011; McAndrew Young et al., 2012; Oberlander et al., 2012; Rosenblatt et al., 2010). Many of these investigators found differences in the elderly or impaired populations for these variables, implying diminished stability, in tasks such as gait over rough and smooth surfaces, quiet standing perturbations, and jumping down. Only one study looked at MoS and TtC for single leg standing, in which ten healthy young subjects were analyzed (Hof et al., 2005). Participants of their study had a non-normalized average MoS value of 1.55 cm and a TtC of 2.3 seconds for the 30 second single leg standing trial. In comparison, the control group of this current study had a non-normalized average MoS value of 3.36 cm and a TtC of 6.98 seconds for the 15 second single leg standing trial. Differences in reported values could be due to relatively small sample sizes and, more likely, the use of different methods for determining the base of support (BoS) of the foot. Hof et al. used the extreme boundaries of the COP under the foot to define the BoS, while the present investigation used the outline of five markers on bony landmarks of the foot. The vCOM in the combined anteroposterior and mediolateral directions was also calculated for the ten subjects in the Hof study, and was found to be a non-normalized average value of 0.51 cm/s. In comparison, the control group of this current study had a non-normalized average vCOM value of 0.85 cm/s, which was the combined magnitude of all three anatomical directions. This implies that the control group of the current study may have been less stable than the control group of the Hof study during single leg stance.

4.1.2 Center of Pressure Velocity and Center of Pressure Path Distance

vCOP during single leg standing has been investigated using healthy, young subjects compared to different populations in various studies (Donath et al., 2012; Clark et al., 2010; Croft et al., 2008; Hertel et al., 2002; Mignardot et al., 2010). These studies found average vCOP values between 3.1 and 4.22 cm/s for their control groups, while this study found a comparable non-normalized average vCOP value of 4.75 cm/s for the control group. Other studies, comparing stability of obese and non-obese subjects, have reported increased vCOP in the obese during quiet standing (Dutil et al., 2012; Hue et al., 2007; Teasdale et al., 2007) and single leg standing (Mignardot et al., 2010). However, this was not observed in the current study, for either absolute or normalized to height values. Differences in subject characteristics between these earlier studies and the present study may contribute to discrepancies in the findings. First, there was a difference in the mean subject age of other the studies, with previous studies including participants with an average age of 49, while the current study included participants with an average age of 25. Age has been shown to have an increasingly negative effect on standing balance of healthy adults (Hamacher et al., 2011; Hue et al., 2007; Karamanidis et al., 2008; Prieto et al., 1996; Tromp et al., 2001). Second, the overweight and obese group of the present study had BMI values averaging 31.1 kg/m^2 while the previous studies had an average BMI of 35.6 kg/m^2 for their overweight and obese groups, including values as high as 50.5 kg/m^2 (Teasdale et al., 2007) and 63.8 kg/m^2 (Hue et al., 2007). Hue et al. (2007)

and Greve et al. (2013) have shown that an increase in obesity correlates negatively with postural stability.

4.1.3 Standard Deviation of Vertical Ground Reaction Force and Time to Stabilization

The standard deviation of the vertical GRF was explored to quantify the vertical body oscillations. After normalization to body weight, there was no difference found between the control and high BMI groups for this variable. Two studies have reported vertical GRF standard deviation values of young and healthy subjects for single leg stance, finding non-normalized values of 3.1 N (Sell et al., 2011) and 3.6 N (Goldie et al., 1989). The current study found a comparable average vertical GRF standard deviation value of 3.82 N for the control group.

To help eliminate the influence of the speed of transition of double to single leg support on the dependent variables, the TTS was determined and used as the starting point for data analysis. No difference was found between the two groups of this study for TTS. Studies investigating TTS in young, healthy subjects performing step downs and jumps that transition into single leg stance found values between 1.71 and 2.95 seconds (Colby et al., 1999; Ross et al., 2003; Wikstrom et al., 2004). Two studies have reported TTS for the transition from double to single leg stance of young, healthy subjects, with values of 1.9 seconds (Dingenen et al., 2013) and 2.28 seconds (Levin et al., 2012). The current study found a comparable average TTS value of 2.22 seconds for the control group.

Overall, the single leg standing results of both groups were representative of healthy populations with adequate postural stability. It is possible that the young high BMI subjects in this study were able to compensate for any potential instability during the single leg stance activity. Although single leg standing has been shown to be an acceptable stability task for distinguishing between healthy and unhealthy groups, this task was unable to find stability differences between the control and high BMI groups of this study. Perhaps using a more difficult balancing task, such as single leg stance with eyes closed, on an unstable surface, or in combination with perturbations, is necessary to more clearly demonstrate reductions in postural stability of young overweight and obese individuals.

4.2 Stair Descent

It was hypothesized that the overweight group would exhibit diminished dynamic postural stability during stair descent. Even though falls on stairs is the leading category of all falls and nearly 75% of those falls happen during descent (Jackson et al., 1995; Svanstrom et al., 1974), indicating this activity to potentially cause instability, very little research has been done on obesity during stair descent. In support of the hypothesis, this study found differences between the control and high BMI groups for spatio-temporal, kinematic, and kinetic parameters. In the following discussion, all joint moments reported in this study and those listed for comparison from related literature are expressed as external joint moments, unless otherwise specified.

4.2.1 Spatio-Temporal Parameters

Obese subjects were found to descend at a 19% slower average vCOM than the non-obese subjects. This slower speed can be directly attributed to longer times spent in both double support and the total right limb stance phase. It has been shown that during gait the strength and power limitations of moving a larger mass, coupled with a presumed desire to limit muscle forces needed to balance moments at the joints, causes overweight individuals to take a shorter stride, have a wider step width, spend more time in double support and overall stance time, and have a slower velocity compared to normal weight individuals (Devita et al., 2003; McGraw et al., 2000; Spyropoulos et al., 1991). Along with potentially reducing musculoskeletal pain and osteoarthritis risk, these alterations can increase stability during gait due to an enlarged base of support and more time spent with both limbs better supporting the COM (Browning et al., 2007; Hicks-Little et al., 2012; McGraw et al., 2000; Spyropoulos et al., 1991). Although the present study found no difference between normal and high BMI in step width during stair descent for this younger population, the 16% longer right limb stance phase time, 25% longer double support time, and 19% slower vCOM observed in the high BMI group may be revealing a similar compensation phenomenon to those found during gait. Stair descent of obese children (Strutzenberger et al., 2011) and of normal adult subjects wearing an additional mass jacket (Spanjaard et al., 2008) demonstrated a longer double support time in the higher weight individuals, although no differences were found in step width. In a study of the effect of weight loss on gait (Hortobagyi et al., 2011), obese adults who lost a substantial amount of weight had a 7% increased swing time, 8% increased stride length, 83% increased knee flexion moment, and 12% increased gait speed.

4.2.2 Margin of Stability and Time to Contact

Normalized MoS and TtC were not different between groups at either the first initiation of double support (right touchdown) or at the initiation of single support (left toe-off), although there were large intersubject variabilities within the groups for these parameters. This possibly indicates a similar level of dynamic stability between groups during descent, most likely due to the overweight subjects reducing their vCOM and spending more time in double support to improve stability. In the only study employing MoS to investigate dynamic stability during stair descent, healthy older subjects were less stable with a 32% smaller MoS value at both right touchdown and left toe-off when compared to healthy younger subjects (Bosse et al., 2012). This was due in large part to the older subjects having a 14% faster vCOM, which led to a more anterior XCOM.

4.2.3 Kinematics and Kinetics

The cycle of stair descent is divided into two phases, stance and swing (Figure 13). The stance phase of the right leg, the period of analysis for this study, can be further subdivided into weight acceptance, forward continuance, and controlled lowering (McFadyen et al., 1988). During weight acceptance, the stance limb (right leg) is increasingly loaded until single limb support (right leg) is attained at swing limb (left leg) toe-off. The period of forward continuance involves forward progression of the COM without any vertical movement (McFadyen et al., 1988). Controlled lowering is the stage of weight shifting to the swing leg (left leg) with vertical lowering of the COM, and is the subphase with the most forward progression (McFadyen et al., 1988).

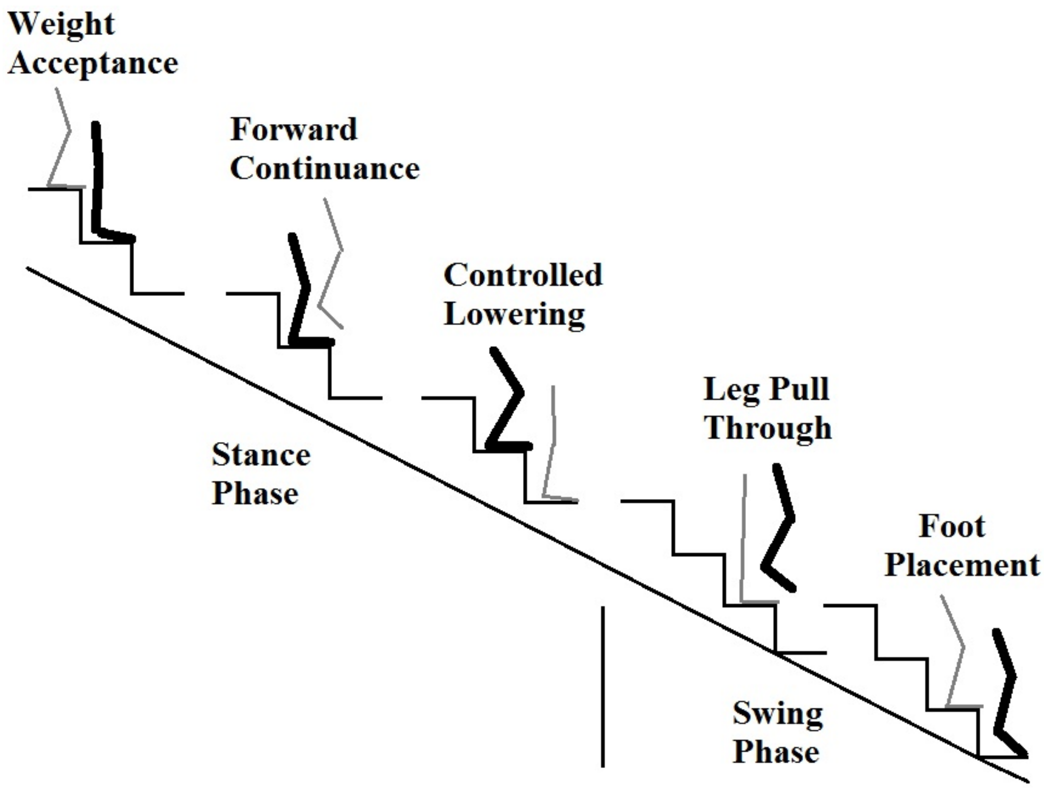


Figure 13. Schematic of the two phases and five sub-phases during step over step stair descent of a three-step staircase. The analysis period is the entire stance phase of the right leg. Bold leg is the right (analyzed) leg.

Weight Acceptance

In both groups, when the right limb touched down to initiate weight acceptance, the hip was slightly flexed, the knee was near full extension, the ankle was plantarflexed and adducted, and the torso and pelvis were tilted slightly forward, as they were throughout the right limb stance phase. Motion of the torso and pelvis in the frontal plane, though highly variable in both planes, can be described as slightly tilted to the right at touchdown, near neutral at mid-stance and at right toe-off. The plantarflexion and adduction of the ankle at right touchdown, and subsequent rotation to dorsiflexion and slight abduction, allows for a large amount of the gravitational energy that was transferred to kinetic energy during swing of the right leg to be absorbed (Cluff et al., 2011; Riener et al., 2002). These findings are comparable to reported angular motion of the joints and segments at lead limb touchdown and throughout the stance phase of healthy normal subjects during stair descent (Beaulieu et al., 2008; Cluff et al., 2011; Krebs et al., 1992; Lee et al., 2011; Lin et al., 2004; Mian et al., 2007; Novak et al., 2011; Powers et al., 1997; Protopapadaki et al., 2007).

Forward Continuance

At the transition from weight acceptance to forward continuance (left toe-off), the hip joint extended slightly with increasing flexion at the knee and a change to dorsiflexion at the ankle. These angular movements occurred simultaneously with a hip flexion moment, a knee flexion moment in normal weight and an extension moment in high BMI subjects, and an ankle flexion moment. At this time point during left toe-off, the high BMI group had an eleven times larger hip flexion moment, a knee extension instead of flexion moment, and a 76% larger dorsiflexion moment at the ankle. A study on gait of obese subjects also found a 43% lower sagittal plane knee joint moment normalized to body mass compared to normal weight participants (DeVita et al., 2003). However, studies using adult normals with 20% added mass jackets (Spanjaard et al., 2008) and on obese children (Strutzenberger et al., 2011) found an increase in knee flexion moment of 15% absolute and 21% normalized to body mass just after swing leg toe-off, respectively, but no difference in the ankle joint moment. Strutzenberger et al. (2011) also found a 26% larger peak hip extension moment normalized to body mass in the obese children, although it is important to note that the average age of their subjects was ten and the hip joint moment in the sagittal plane is highly variable throughout literature for healthy normal adults during stair descent and in studies comparing obese to normal weight gait (Browning et al., 2012; Lin et al., 2004). The knee extension moment of the high BMI group may have been used to slow the vertical descent and forward progression of the COM (Lin et al., 2004). Furthermore, it is possible that the knee moment difference is caused by an increase in knee stiffness by the high BMI group by contracting the knee flexors, which would reduce the external flexion moment but increase joint stability (Hortobagyi et al., 1999; Novak et al., 2013). Contrary to most stair descent literature, one study (Protopapadaki et al., 2007) found an extension moment normalized to body mass near swing leg toe-off in healthy normal subjects, which they attributed to possible differences in stair inclination, subject height, marker placement, trunk motion (not investigated by their study), and joint moment calculation methods between studies.

The 33% smaller hip adduction moment found in the high BMI individuals at left toe-off was possibly due to the larger knee adduction angle found for this group. An eleven times larger knee adduction angle near swing leg toe-off was also found in a study on obese gait (Lai et al., 2008). This first hip adduction moment peak that was smaller in the high BMI group corresponds with the acceptance of body weight by the right leg transferred from the left limb to stabilize the trunk over the support leg (right limb) (Lin et al., 2005; Novak et al., 2013). During gait, weakness of the hip abductor muscles to generate an internal hip abduction moment (external hip adduction moment) can cause frontal plane instability and increase the risk of falls (Krebs et al., 1998; Novak et al., 2011). In addition, normalized GRFs at this instance were similar for both groups, which is in agreement with normal weight compared to obese adults during walking (Browning et al., 2007; Lai et al., 2008) and normal weight compared to obese children during stair descent (Strutzenberger et al., 2011).

Controlled Lowering

As the right limb moved through mid-stance to left touchdown (Figure 10), the hip was still moderately flexed and the moment at the hip had gone into extension in the normal subjects while being near neutral in the high BMI subjects. The angle at the knee continued a steady climb of flexion and the knee moment was near maximum flexion at left touchdown. Prior to left touchdown, gravitational energy is dissipated and controlled lowering of the body is initiated mainly at the knee joint, with some assistance by muscles at the hip joint (Cluff et al., 2011). Additionally, the knee adduction moment in the high BMI group was 92% larger than that of the control group. This second adduction moment peak in the frontal plane corresponds with controlling the COM transfer back to the left leg at touchdown (Lin et al., 2005) and is important in providing propulsion and mediolateral stability (Kowalk et al., 1996). External adductor moments of the knee and hip have been found to be important in the control of the COM within the narrow BoS of the stance limb (right leg) during controlled lowering and in counteracting the destabilization produced by the upper body and mass of the swing leg (left limb) (Figure 14) (Novak et al., 2011). A larger knee adduction moment has been associated with an increase in medial compartment compressive loading, and has been reported in obese gait (Browning et al., 2007; Lai et al., 2008). An increase in compressive force at the knee can be a strategy to increase dynamic stability when experiencing a greater knee adduction moment (Messier et al., 2005).

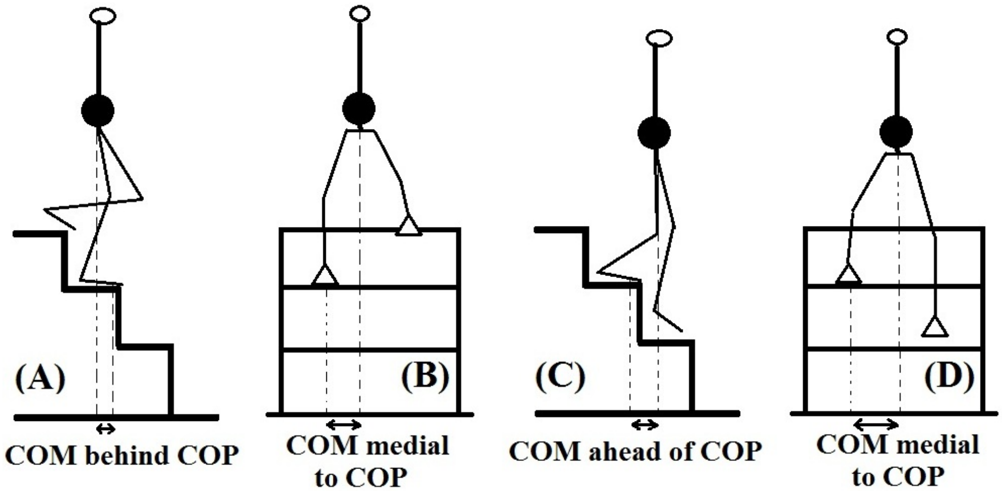


Figure 14. Anteroposterior and mediolateral separation of center of mass (● = COM) and center of pressure (COP) during normal stair descent. Vertical dashed lines from the foot are COP location, vertical dashed lines from sphere at the torso are COM location.

A larger knee extension moment, like that found in the high BMI group of the present study at left toe-off during stair descent, has also been found in osteoarthritic gait (Runhaar et al., 2011). This extension moment may improve stability by increasing compressive forces at the knee in the presence of a larger adduction moment (Messier et al., 2005), although this can increase the risk of developing osteoarthritis (Browning et al., 2012). Weight loss in obese individuals (Messier et al., 2005) was directly associated with a decrease in knee adduction moment normalized to body weight during gait. This weight loss study also found that there was a four pound reduction in knee joint load per step for every pound of bodyweight lost.

At touchdown of the left leg, the ankle joint of the right limb was close to being maximally flexed with a second flexion moment peak. Controlled lowering of the COM mainly involves movements at the knee and ankle for better control of the upper body, with less range of motion at the hip compared to the other joints (Lin et al., 2004). The anterior GRF at this point was 26% lower for the high BMI individuals, indicating relatively less of a propulsive force. This finding was in agreement with a previous study of obese individuals during gait (Lai et al., 2008) and could be due to the high BMI group wanting to generate less force so as to control the forward progression of the COM during weight acceptance of the left leg. In addition, intersegmental resultant force curves have shown that posteriorly directed forces are required at the hip during stair descent, which indicates that the anteroposterior components of the net inertial, gravitational, and ground reaction forces are directed anteriorly (Lin et al., 2005). This suggests that, for increased stability, the body is not accelerating excessively due to control of the posteriorly directed forces at the hip, most likely accomplished by keeping the COM behind the supporting limb (right leg) for as long as possible (Figure 14). The differences found in moments of the right limb at left toe-off and touchdown attributed to the differences found in the peak moments.

Initiation of Leg Pull Through

Near right toe-off, the hip and knee were maximally flexed with the hip in slight abduction and the knee in slight adduction. A 20% larger peak hip flexion angle was seen in the high BMI participants at this point that led to a 33% larger range of motion at the hip throughout the stance phase. This larger flexion angle may help to raise the right limb for clearance of the step as the limb progresses into swing, possibly minimizing tripping concerns (Novak et al., 2013). Furthermore, the high BMI group displayed a 26% smaller anteroposterior pelvic range of motion during stance. Research on gait has shown that slower walking speed can lead to reduced sagittal plane knee joint range of motion (Browning et al., 2007; Silvernail et al., 2013). The high BMI group showed a slower vCOM during stair descent, however this did not lead to a difference in range of motion at the knee.

Throughout most of the analyzed right limb stance phase, the normalized GRFs of both groups were similar. Slower walking speeds have been shown to reduce GRFs in normal weight and obese subjects (Browning et al., 2007; Messier et al., 2005). The slower descent of the high BMI subjects of this study seems to be a possible compensation mechanism to bring the GRFs within a normal range, as only the anterior GRF at left touchdown was found to be different between groups. When two force vectors have a similar magnitude and direction about a joint, the vector with the larger moment arm will generate a larger joint moment (Reeves et al., 2008). Differences in the moment arms could be influenced by differences in joint angles.

Several joint angles in the high BMI group were found to correlate well with joint moments that were different between groups at left toe-off. Torso flexion and pelvic obliquity down angles were positively correlated with hip flexion, knee extension, and ankle dorsiflexion moments. In addition, hip abduction angle also positively correlated with the hip flexion and knee extension moments of the high BMI group, although these correlations may have been due to the torso and pelvic angles. Torso flexion combined with an oblique down pelvis (tilted to the right) would most likely bring about hip flexion, knee extension, and ankle dorsiflexion, which would also cause the corresponding changes in the moments at those joints. Individuals with knee osteoarthritis and patients with total hip arthroplasty tend to lean their torso forward more than healthy normal subjects to reduce loading at the knee and increase their feeling of stability during stair negotiation (Asay et al., 2008; Lamontagne et al., 2011). This altered pattern of descent by differing trunk position to change the line of the GRF anteroposteriorly in relation to the hip joint could be a possible reason for the differences in hip flexion at left touchdown and toe-off, as well as the variability seen at the hip both in this study and throughout literature (Protopapadaki et al., 2007).

In obese individuals with a large proportion of mass in their abdomen, torso flexion may be used to improve visibility of the steps in the bottom of their visual fields during stair descent (Rosenbaum et al., 2009). Pelvic obliquity helps with both the mediolateral transfer of mass and the lifting of the swing leg for step clearance (Nadeau et al., 2003), and may be a compensatory mechanism to increase hip flexion range of motion in the high BMI group (Mian et al., 2007). By comparison, the pelvis was more anteriorly tilted in obese compared to normal weight children during stair descent (Strutzenberger et al., 2011). Moreover, this current study did not investigate the kinematics or kinetics of the transverse plane, which may have some influence on motion of the other planes. The few studies that have considered motion in the transverse plane during gait (Mian et al., 2007; Messier et al., 2005; Spyropoulos 1991) are inconclusive for alterations of normal weight and obese subjects.

Differences found in the mechanics of stair descent between groups may also be related to altered sensorimotor functioning and proprioception in the high BMI subjects. Foot proprioception strongly influences control of postural balance and some responses to perturbations (Kavounoudias et al., 2001; Meyer et al., 2004). Increased plantar pressure has been shown in obese subjects (Del Porto et al., 2012) during standing and walking, which can cause pain and tissue damage (Hills et al., 2002). This can lead to a reduction in proprioception under the foot, altering the sensing of when postural corrections are needed (Del Porto et al., 2012; Hills et al., 2002). Obesity has also been linked to an increase in sensory thresholds of all nerves, possibly affecting proprioception even in the absence of external pressure (Miscio et al., 2005). This decreased proprioception in the obese could hinder their balance control and increase their risk of falls.

Additionally, care should be taken when comparing different studies on stair negotiation as research has shown that subject height, stair dimensions, which step of the staircase is analyzed, and stepping cadence can have a large impact on the joint biomechanics (Andriacchi et al., 1980; Livingston et al., 1991; Riener et al., 2002; Spanjaard et al., 2008). The stair dimensions in this study were a rise of 17.8 cm and a run of 28.0 cm, giving an inclination angle of 32 degrees. These dimensions are near the average of those found in literature and comply with the 2009 International Residential Code of residential staircase design (Cluff et al., 2011; IRC, 2009; Protopapadaki et al., 2007).

In summary, and in concurrence with existing studies of overweight and obese individuals during gait and stair descent, the longer support times, slower velocity, and differences in moments at left toe-off and touchdown for the high BMI group indicate potential instability during the stair descent task.

CHAPTER V

LIMITATIONS, CONCLUSIONS, and FUTURE RESEARCH

5.1 Limitations

One limitation of the present study is the use of body mass index (BMI) classification for group separation. Although this measurement parameter has been used extensively in many clinical settings, literature suggests BMI to be an inaccurate measure of adipose tissue (Gallagher et al., 1996; Garn et al., 1986; Smalley et al., 1990). Differences in the distribution of fat content throughout the body and also different body types are also not considered in this measurement’s computation. BMI calculation may result in an overestimate of body fat content for individuals with more lean tissue and underestimate for those with less lean tissue.

The small samples sizes used are also of concern, as the control group had eleven subjects and the high BMI group had six subjects. This may result in these particular groups not accurately representing their respective populations. However, even with small samples sizes, many differences were found between groups for the analyzed activities. A difference in height between groups was also observed. Normalization by height was used as a method to account for the difference, however the parameters used to analyze the performed activities may still have been influenced.

In regards to equipment used, the experimental staircase consisted of only three steps. Research on stair negotiation of healthy normal subjects (Cluff et al., 2011) suggests that a minimum of five steps is needed to attain steady state stair descent. Hence, stair descent in this study may not be representative of stair negotiation in the real world, however most accidental stair falls occur during the transition phases of the first or last two steps (Jackson et al., 1995; Startzell et al., 2000).

An issue that has yet to be addressed by researchers in the field of biomechanics and has been recognized as a source of error in human motion analysis (Leardini et al., 2005) is soft tissue artifact from the movement of skin markers. Errors resulting from the movement of markers on skin sliding over bone could cause errors in the biomechanical calculations. Although a number of compensation methods have been developed, a viable solution of minimizing this error has yet to be found (Wearing et al., 2006).

Another limitation is the use of inverse dynamics in the calculation of kinetic variables. This method assumes that the segments of the body can be modeled as rigid structures, which may not be the case during human motion as some structures of the body are more rigid than others (Pandy et al., 2001). Joint moment calculations for more flexible structures, like those of the feet, may be less accurate due to this assumption.

5.2 Conclusions

The purpose of this study was to examine the stability and biomechanics of the torso, pelvis, and lower limbs during single leg stance and stair descent in normal and high body mass index individuals. The primary research question was to determine whether there were differences in various biomechanical parameters between these groups that could affect their stability during the performed tasks. Based on the previous limited research on obese participants performing these and other related activities, it was hypothesized that the high BMI would exhibit diminished static and dynamic postural stability. In regards to single leg standing, no differences were found ($p = 0.234 – 0.990$) between groups for the variables investigated. This implies that, contrary to the hypothesis with respect to the stance task, overweight and obese subjects did not have impaired postural stability compared to normal weight subjects. The relatively young population used in this study may have been able to effectively compensate for their potential instability, with a more challenging postural task perhaps needed to reveal differences. In regards to stair descent, and in support of the hypothesis, many differences were found between groups that could indicate a reduction in stability for the high BMI group. A 19% slower rate of descent ($p = 0.007$) and 25% more time spend in double support ($p = 0.021$) and 16% more time spent in the stance phase ($p = 0.174$) was found for the high BMI subjects, presumably to better control their center of mass. At left toe-off, the high BMI group had an eleven times larger hip flexion moment ($p < 0.001$), 33% smaller hip adduction moment ($p = 0.050$), knee extension instead of a flexion moment ($p < 0.001$), 76% larger ankle dorsiflexion moment ($p < 0.001$), and an eleven times larger knee adduction angle ($p = 0.015$). At left touchdown, the high BMI group had a slight hip flexion instead of extension moment ($p = 0.025$), a 92% larger knee adduction moment ($p = 0.047$), and a 26% lower anterior ground reaction force normalized to body weight ($p = 0.015$). Differences at these two instances also contributed to peak differences seen in these variables. A 20% larger peak hip flexion angle ($p = 0.059$) caused 33% more sagittal plane hip range of motion ($p = 0.032$) seen for the high BMI group, and this group also had 26% less sagittal plane pelvic tilt range of motion ($p = 0.051$). All of these biomechanical differences taken together suggest, in agreement with existing literature, possible cumulative overloading of the joints, greater risk of osteoarthritis, and decreased stability during stair descent. These differences also reveal potential compensation mechanisms of the larger mass in the high BMI group due to their increased gravitational energy that must be dealt with.

5.3 Future Research

Future research should include a replication of this study for stair ascent and descent with a range of obese individuals to build a consensus of how individuals with a high BMI negotiate stairs. Movement analysis combined with EMG data would reveal the concurrent contributions of the muscles in the lower limbs. Future studies should also have larger sample sizes to ensure greater power to detect differences between groups. As previously mentioned, additional research on static postural stability of obese subjects should be investigated, possibly with more challenging tasks for young populations. Overall, further research on obese performing various tasks needs to be undertaken, as there is very limited existing literature on this population for activities other than gait and quiet standing.

REFERENCES

Abuzayan, K., Alabed, H., Ezarrugh, J., Agila, M., 2013. Developing the methods for the study of static and dynamic balance. International Journal of Medical, Pharmaceutical Science and Engineering 7, 432-443.

Anandacoomarasamy, A., Caterson, I., Sambrook, P., Fransen, M., March, L., 2008. The impact of obesity on the musculoskeletal system. International Journal of Obesity 32, 211-222.

Andriacchi, T., Andersson, G., Fermier, R., Stern D., Galante, J., 1980. A study of lower-limb mechanics during stair-climbing. The Journal of Bone and Joint Surgery 62-A, 749-757.

Arampatzis, A., Karamanidis, K., Mademli, L., 2008. Deficits in the way to achieve balance related to mechanisms of dynamic stability control in the elderly. Journal of Biomechanics 41, 1754-1761.

Asay, J., Mundermann, A., Andriacchi, T., 2009. Adaptive patterns of movement during stair climbing in patients with knee osteoarthritis. Journal of Orthopaedic Research 27, 325-329.

Beaulieu, F., Pelland, L., Robertson, D., 2008. Kinetic analysis of forwards and backwards stair descent. Gait and Posture 27, 564-571.

Bernard, P., Geraci, M., Hue, O., Amato, M., Seynnes, O., Lantieri, D., 2003. Influence of obesity on postural capacities of teenagers. Annales de readaptation et de Medecine Physique 46, 184-190.

Berrigan, F., Simoneau, M., Tremblay, A., Hue, O., Teasdale, N., 2006. Influence of obesity on accurate and rapid arm movement performed from a standing posture. International Journal of Obesity 30, 1750-1757.

Bierbaum, S., Peper, A., Karamanidis, K., Arampatzis, A., 2011. Adaptive feedback potential in dynamic stability during disturbed walking in the elderly. Journal of Biomechanics 44, 1921-1926.

Blaszczyk, J., Cieslinska-Swider, J., Plewa, M., Zahorska-Markiewicz, B., Markiewicz, A., 2009. Effects of excessive body weight on postural control. Journal of Biomechanics 42, 1295-1300.

Blaszczyk, J., Plewa, M., Cieslinska-Swider, J., Bacik, B., Zahorska-Markiewicz, B., Markiewicz, A., 2011. Impact of excess body weight on walking at the preferred speed. Acta Neurobiologiae Experimentalis 71, 528-540.

Bosse, I., Oberlander, K., Savelberg, H., Meijer, K., Bruggemann, G., Karamanidis, K., 2012. Dynamic stability control in younger and older adults during stair descent. Human Movement Science 31, 1560-1570.

Browning, R., Kram, R., 2007. Effects of obesity on the biomechanics of walking at different speeds. Medicine and Science in Sports and Exercise 39, 1632-1641.

Browning, R., 2012. Locomotion mechanics in obese adults and children. Current Obesity Reports 1, 152-159.

Bruijn, S., Meijer, O., Beek, P., Dieen, J., 2013. Assessing the stability of human locomotion: a review of current measures. Journal of the Royal Society Interface 10, 20120009.

Chiari, L., Rocchi, L., Cappello, A., 2002. Stabilometric parameters are affected by anthropometry and foot placement. Clinical Biomechanics 17, 666-677.

Christina, K., Cavanagh, R., 2002. Ground reaction forces and frictional demands during stair descent: effects of age and illumination. Gait and Posture 15, 153-158.

Clark, R., Bryant, A., Pua, Y., McCrory, P., Bennell, K., Hunt, M., 2010. Validity and reliability of the Nintendo wii balance board for assessment of standing balance. Gait and Posture 31, 307-310.

Cluff, T., Robertson, D., 2011. Kinetic analysis of stair descent: part 1. forwards step-over-step descent. Gait and Posture 33, 423-428.

Colby, S., Hintermeister, R., Torry, M., Steadman, J., 1999. Lower limb stability with acl impairment. Journal of Orthopaedic and Sports Physical Therapy 29, 444-454.

Corbeil, P., Simoneau, M., Rancourt, D., Tremblay, A., Teasdale, N., 2001. Increased risk for falling associated with obesity: mathematical modeling of postural control. IEEE Transactions on Neural Systems and Rehabilitation Engineering 9, 126-136.

Croce, U., Bonato, Paolo., 2007. A novel design for an instrumented stairway. Journal of Biomechanics 40, 702-704.

Croft, J., Tscharner, V., Zernicke, R., 2008. Movement variability and muscle activity relative to center of pressure during unipedal stance on solid and compliant surfaces. Motor Control 12, 283-295.

Curtze, C., Hof, A., Postema, K., Otten, B., 2011. Over rough and smooth: amputee gait on an irregular surface. Gait and Posture 33, 292-296.

Del Porto, H., Pechak, C., Smith, D., Reed-Jones, R., 2012. Biomechanical effects of obesity on balance. International Journal of Exercise Science 5, 301-320.

DeVita, P., Hortobagyi, T., 2003. Obesity is not associated with increased knee joint torque and power during level walking. Journal of Biomechanics 36, 1355-1362.

Dewan, A., 2013. Related to: obesity and total joint arthroplasty: a literature based review. The Journal of Arthroplasty 28, 714-721.

Dingenen, B., Staes, F., Janssens, L., 2013. A new method to analyze postural stability during a transition task from double-leg stance to single-leg stance. Journal of Biomechanics 46, 2213-2219.

Donath, L., Roth, R., Zahner, L., Faude, O., 2012. Testing single and double limb standing balance performance: comparison of cop path length evaluation between two devices. Gait and Posture 36, 439-443.

Dutil, M., Handrigan, G., Corbeil, P., Cantin, V., Simoneau, M., Teasdale, N., Hue, O., 2012. The impact of obesity on balance control in community-dwelling older women. Age 35, 883-890.

Fjeldstad, C., Fjeldstad, A., Acree, L., Nickel, K., Gardner, A., 2008. The influence of obesity on falls and quality of life. Dynamic Medicine 7, 1-6.

Gage, W., Winter, D., Frank, J., Adkin, A., 2004. Kinematic and kinetic validity of the inverted pendulum model in quiet standing. Gait and Posture 19, 124-132.

Gallagher, D., Visser, M., Sepulveda, D., Pierson, R., Harris, T., Heymsfield, S., 1996. How useful is body mass index for comparison of body fatness across age, sex, and ethnic groups. American Journal of Epidemiology 143, 228-239.

Garn, S., Leonard, W., Hawthorne, V., 1986. Three limitations of the body mass index. The American Journal of Clinical Nutrition 44, 996-997.

Gerbino, P., Griffin, E., Zurakowski, D., 2007. Comparison of standing balance between female collegiate dancers and soccer players. Gait and Posture 26, 501-507.

Goldie, P., Bach, T., Evans, O., 1989. Force platform measures for evaluating postural control: reliability and validity. Archives of Physical Medicine and Rehabilitation 70, 510-517.

Goldie, P., Evans, O., Bach, T., 1992. Steadiness in one-legged stance: development of a reliable force-platform testing procedure. Archives of Physical Medicine and Rehabilitation 73, 348-354.

Greve, J., Alonso, A., Bordini, A., Camanho, G., 2007. Correlation between body mass index and postural balance. Clinics 62, 717-720.

Greve, J., Cug, M., Dulgeroglu, D., Brech, G., Alonso, A., 2013. Relationship between anthropometric factors, gender, and balance under unstable conditions in young adults. BioMed Research International 27, 1-5.

Guo, M., Axe, M., Manal, K., 2007. The influence of foot progression angle on the knee adduction moment during walking and stair climbing in pain free individuals with knee osteoarthritis. Gait and Posture 26, 436-441.

Hamacher, D., Singh, N., Van Dieen, J., Heller, M., Taylor, W., 2011. Kinematic measures for assessing gait stability in elderly individuals: a systematic review. Journal of the Royal Society Interface 8, 1682-1698.

Handrigan, G., Corbeil, P., Simoneau, M., Teasdale, N., 2009. Balance control is altered in obese individuals. Journal of Biomechanics 43, 383-384.

Hertel, J., Gay, M., Denegar, C., 2002. Differences in postural control during single-leg stance among healthy individuals with different foot types. Journal of Athletic Training 37. 129-132.

Hicks-Little, C., Peindl R., Hubbard, T., Scannell, B., Springer, B., Odum, S., Fehring, T., Cordova, M., 2011. Lower extremity joint kinematics during stair climbing in knee osteoarthritis. Medicine and Science in Sports and Exercise 43, 516-524.

Hicks-Little, C., Peindl R., Fehring, T., Odum, S., Hubbard, T., Cordova, M., 2012. Temporal-spatial gait adaptations during stair ascent and descent in patients with knee osteoarthritis. The Journal of Arthroplasty 27, 1183-1189.

Hills, A., Henning, E., Byrne, N., Steele, J., 2002. The biomechanics of adiposity – structural and functional limitations of obesity and implications for movement. Obesity Reviews 3, 35-43.

Himes, C., Reynolds, S., 2012. Effect of obesity on falls, injury, and disability. Journal of American Geriatric Society 60, 124-129.

Hof, A., Gazendam, M., Sinke, W., 2005. The condition for dynamic stability. Journal of Biomechanics 38, 1-8.

Hof, A., Vermerris, S., Gjaltema, W., 2010. Balance responses to lateral perturbations in human treadmill walking. The Journal of Experimental Biology 213, 2655-2664.

Hortobagyi, T., DeVita, P., 1999. Altered movement strategy increases lower extremity stiffness during stepping down in the aged. Journal of Gerontology: Biological Sciences 54A, B63-B70.

Hortobagyi, T., Herring, C., Pories, W., Rider, P., DeVita, P., 2011. Massive weight loss-induced mechanical plasticity in obese gait. Journal of Applied Physiology 111, 1391-1399.

Hue, O., Simoneau, M., Marcotte, J., Berrigan, F., Dore, J., Marceau, P., Marceau, S., Tremblay, A., Teasdale, N., 2007. Body weight is a strong predictor of postural stability. Gait and Posture 26, 32-38.

Hulens, M., Vansant, G., Claessens, A., Lysens, R., Muls, E., 2003. Predictors of 6-minute walk test results in lean, obese and morbidly obese women. Scandinavian Journal of Medicine and Science in Sports 13, 98-105.

Hurvitz, E., Richardson, J., Werner, R., Ruhl, A., Dixon, M., 2000. Unipedal stance testing as an indicator of fall risk among older outpatients. Archives of Physical Medicine and Rehabilitation 81, 587-591.

IRC: International Residential Code for One- and Two-Family Dwellings, 2009 Edition. International Code Council, Inc., Falls Church, VA, 2009.

Jackson, P., Cohen, H., 1995. An in-depth investigation of 40 stairway accidents and the stair safety literature. Journal of Safety Research 26, 151-159.

Karamanidis, K., Arampatzis, A., Mademli, L., 2008. Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. Journal of Electromyography and Kinesiology 18, 980-989.

Kavounoudias, A., Roll, R., Roll, J., 2001. Foot sole and ankle muscle inputs contribute jointly to human erect posture regulation. Journal of Physiology 532, 869-878.

Kim, H., 2009. A comparison of the center of pressure during stair descent in young and healthy elderly adults. Journal of Physical Therapy Science 21, 129-134.

Kowalk, D., Duncan, J., Vaughan, C., 1996. Abduction-adduction moments at the knee during stair ascent and descent. Journal of Biomechanics 29, 383-388.

Krebs, D., Wong, D., Jevsevar, D., Riley, P., Hodge, W., 1992. Trunk kinematics during locomotor activities. Physical Therapy 72, 505-514.

Krebs, D., Jette, A., Assmann, S., 1998. Moderate exercise improves gait stability in disable elders. Archives of Physical Medicine and Rehabilitation 79, 1489-1495.

Ku, P., Osman, N., Yusof, A., Abas, W., 2012. Biomechanical evaluation of the relationship between postural control and body mass index. Journal of Biomechanics 45, 1638-1642.

Kuo, A., Donelan, M., Ruina, A., 2005. Energetic consequences of walking like an inverted pendulum: step-to-step transitions. Exercise and Sport Sciences Reviews 33, 88-97.

Kuo, A., 2007. The six determinants of gait and the inverted pendulum analogy: a dynamic walking perspective. Human Movement Science 26, 617-656.

Lai, P., Leung, A., Li, A., Zhang, M., 2008. Three-dimensional gait analysis of obese adults. Clinical Biomechanics 23, S2-S6.

Lamontagne, M., Varin, D., Beaule, P., 2011. Does the anterior approach for total hip arthroplasty better restore stair climbing gait mechanics. Journal of Orthopaedic Research 29, 1412-1417.

Larsson, U., Mattson, E., 2001. Perceived disability and observed functional limitations in obese women. International Journal of Obesity 25, 1705-1712.

Leardini, A., Chiari, L., Croce, U., Cappozzo, A., 2005. Human movement analysis using stereophotogrammetry part 3. soft tissue artifact assessment and compensation. Gait and Posture 21, 212-225.

Ledin, T., Odkvist, L., 1993. Effects of increased inertial load in dynamic and randomized perturbed posturography. Acta Otolaryngology 113, 249-252.

Lee, J., Park , E., 2011. 3d spinal motion analysis during staircase walking using an ambulatory inertial and magnetic sensing system. Medical and Biological Engineering and Computing 49, 755-764.

Levin, O., Nevel, A., Malone, C., Deun, S., Duysens, J., Staes, F., 2012. Sway activity and muscle recruitment order during transition from double to single-leg stance in subjects with chronic ankle instability. Gait and Posture 36, 546-551.

Lin, H., Lu, T., Hsu, H., 2004. Three-dimensional analysis of kinematic and kinetic coordination of the lower limb joints during stair ascent and descent. Biomedical Engineering Applications, Basis, and Communications 16, 101-108.

Lin, H., Lu, T., Hsu, H., 2005.Comparisions of joint kinetics in the lower extremity between stair ascent and descent. Journal of Mechanics 21, 41-50.

Lin, D., Seol, H., Nussbaum, M., Madigan, M., 2008. Reliability of cop-based postural sway measures and age-related differences. Gait and Posture 28, 337-342.

Livingston, L., Stevenson, J., Olney, S., 1991. Stairclimbing kinematics on stairs of differing dimension. Archives of Physical Medicine and Rehabilitation 72, 398-402.

Lugade, V., Lin, V., Chou, L., 2011. Center of mass and base of support interaction during gait. Gait and Posture 33, 406-411.

Mademli, L., Arampatzis, A., Karamanidis, K., 2008. Dynamic stability control in forward falls: postural corrections after muscle fatigue in young and older adults. European Journal of Applied Physiology 103, 295-306.

Maffiuletti, N., Agosti, F., Proietti, M., Riva, D., Resnik, M., Lafortuna, C., Sartorio, A., 2005. Postural instability of extremely obese individuals improves after a body weight reduction program entailing specific balance training. Journal of Endocrinological Investigation 28, 2-7.

Mancini, M., Horak, F., 2010. The relevance of clinical balance assessment tools to differentiate balance deficits. European Journal of Physical and Rehabilitation Medicine 46, 239-248.

Matrangola, S., Madigan, M., 2011. The effects of obesity on balance recovery using an ankle strategy. Human Movement Science 30, 584-595.

McAndrew Young, P., Wilken, J., Dingwell, J., 2012. Dynamic margins of stability during human walking in destabilizing environments. Journal of Biomechanics 45, 1053-1059.

McFadyen, B., Winter, D., 1988. An integrated biomechanical analysis of normal stair ascent and descent. Journal of Biomechanics 21, 733-744.

McGraw, B., McClenaghan, B., Williams, H., Dickerson, J., Ward, D., 2000. Gait and postural stability in obese and nonobese prepubertal boys. Archives of Physical Medicine and Rehabilitation 81, 484-489.

Messier, S., Gutekunst, D., Davis, C., DeVita, P., 2005. Weight loss reduces knee-joint loads in overweight and obese adults with knee osteoarthritis. Arthritis and Rheumatism 52, 2026-2032.

Meyer, P., Oddsson, L., DeLuca, C., 2004. Reduced plantar sensitivity alters postural responses to lateral perturbations of balance. Experimental Brain Research 157, 526-536.

Mian, O., Thom, J., Narici, M., Baltzopoulos, V., 2007. Kinematics of stair descent in young and older adults and the impact of exercise training. Gait and Posture 25, 9-17.

Mignardot, J., Olivier, I., Promayon, E., Nougier, V., 2010. Obesity impact on the attentional cost for controlling posture. PLoS ONE 5, e14387.

Miscio, G., Guastamacchia, G., Brunani, A., Priano, L., Baudo, S., Mauro, A., 2005. Obesity and peripheral neuropathy risk: a dangerous liaison. Journal of the Peripheral Nervous System 10, 354-358.

Nadeau, S., McFadyen, B., Malouin, F., 2003. Frontal and sagittal plane analyses of the stair climbing task in healthy adults aged over 40 years: what are the challenges compared to level walking. Clinical Biomechanics 18, 950-959.

Novak, A., Reid, S., Costigan, P., Brouwer, B., 2010. Stair negotiation alters stability in older adults. Lower Extremity Review Oct, 1-7.

Novak, A., Brouwer, B., 2011. Sagittal and frontal lower limb joint moments during stair ascent and descent in young and older adults. Gait and Posture 33, 54-60.

Novak, A., Brouwer, B., 2013. Kinematic and kinetic evaluation of the stance phase of stair ambulation in persons with stroke and healthy adults: a pilot study. Journal of Applied Biomechanics 29, 443-452.

Oberlander, K., Bruggemann, G., Hoher, J., Karamanidis, K., 2012. Reduced knee joint moment in acl deficient patients at a cost of dynamic stability during landing. Journal of Biomechanics 45, 1387-1392.

Pandy, M., 2001. Computer modeling and simulation of human movement. Annual Review in Biomedical Engineering 3, 245-273.

Petti, S., Cairella, G., Tarsitani, G., 1997. Childhood obesity: a risk factor for traumatic injuries to anterior teeth. Endodontics and Dental Traumatology 13, 285-288.

Piirtola, M., Era, P., 2006. Force platform measurements as predictors of falls among older people – a review. Gerontology 52, 1-16.

Powers, C., Boyd, L., Torburn, L., Perry, J., 1997. Stair ambulation in persons with transtibial amputation: an analysis of the seattle lightfoot. Journal of Rehabilitation Research and Development 34, 9-18.

Prieto, T., Myklebust, J., Hoffmann, R., Lovett, E., Myklebust, B., 1996. Measures of postural steadiness: differences between healthy young and elderly adults. IEEE Transactions on Biomedical Engineering 43, 956-966.

Protopapadaki, A., Drechsler, W., Cramp, M., Coutts, F., Scott, O., 2007. Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals. Clinical Biomechanics 22, 203-210.

Raymakers, J., Samson, M., Verhaar, H., 2005. The assessment of body sway and the choice of the stability parameters(s). Gait and Posture 21, 48-58.

Reeves, N., Spanjaard, M., Mohagheghi, A., Baltzopoulos, V., Maganaris, C., 2008. The demands of stair descent relative to maximum capacities in elderly and young adults. Journal of Electromyography and Kinesiology 18, 218-227.

Reid, S., Lynn, S., Musselman, R., Costigan, P., 2007. Knee biomechanics of alternate stair ambulation patterns. Medicine and Science in Sports and Exercise 39, 2005-2011.

Riemann, B., Myers, J., Lephar, S., 2003. Comparison of the ankle, knee, hip, and trunk corrective action shown during single-leg stance on firm, foam, and multiaxial surfaces. Archives of Physical Medicine and Rehabilitation 84, 90-95.

Riener, R., Rabuffetti, M., Frigo, C., 2002. Stair ascent and descent at different inclinations. Gait and Posture 15, 32-44.

Rosenbaum, D., Walking down memory lane: where walkers look as they descend stairs provides hints about how they control their walking behavior. The American Journal of Psychology 122, 425-430.

Rosenblatt, N., Grabiner, M., 2010. Measures of frontal plane stability during treadmill and overground walking. Gait and Posture 31, 380-384.

Ross, S., Guskiewicz, K., 2003. Time to stabilization: a method for analyzing dynamic postural stability. Athletic Therapy Today 8, 37-39.

Ruhe, A., Fejer, R., Walker, B., 2010. The test-retest reliability of centre of pressure measures in bipedal static task conditions – a systematic review of the literature. Gait and Posture 32, 436-445.

Runhaar, J., Koes, B., Clockaerts, S., Bierma-Zeinstra, S., 2011. A systematic review on changed biomechanics of lower extremities in obese individuals: a possible role in development of osteoarthritis. Obesity Reviews 12, 1071-1082.

Salsich, G., Brechter, J., Powers, C., 2001. Lower extremity kinetics during stair ambulation in patients with and without patellofemoral pain. Clinical Biomechanics 16, 906-912.

Samuel, D., Rowe, P., Hood, V., Nicol, A., 2011. The biomechanical functional demand placed on knee and hip muscles of older adults during stair ascent and descent. Gait and Posture 34, 239-244.

Sartorio, A., Lafortuna, C., Conte, G., Faglia, G., Narici, M., 2001. Changes in motor control and muscle performance after a short-term body mass reduction program in obese subjects. Journal of Endocrinological Investigation 24, 393-398.

Schloemer, S., Cotter, J., Jamison, S., Chaudhari, A., 2013. Time-to-contact demonstrates modulation of postural control during a dynamic lower extremity task. Gait and Posture 38, 658-662.

Schmid, M., Conforto, S., Camomilla, V., Cappozzo, A., D'Alessio, T., 2002. The sensitivity of posturographic parameters to acquisition settings. Medical Engineering and Physics 24, 623-631.

Sell, T., 2012. An examination, correlation, and comparison of static and dynamic measures of postural stability in healthy, physically active adult. Physical Therapy in Sport 13, 80-86.

Silvernail, J., Milner, C., Thompson, D., Zhang, S., Zhao, X., 2013. The influence of body mass index and velocity on knee biomechanics during walking. Gait and Posture 37, 575-579.

Singh, D., Park , W., Levy, M., Jung, E., 2009. The effects of obesity and standing time on postural sway during prolonged quiet standing. Ergonomics 52, 977-986.

Smalley, K., Knerr, A., Kendrick, Z., Colliver, J., Owen, O., 1990. Reassessment of body mass indices. The American Journal of Clinical Nutrition 52, 405-408.

Spanjaard, M., Reeves, N., van Dieen, J., Baltzopoulos, V., Maganaris, C., 2008. Lower-limb biomechanics during stair descent: influence of step-height and body mass. The Journal of Experimental Biology 211, 1368-1375.

Spyropoulos, P., Pisciotta, J., Pavlou, K., Cairns, M., Simon, S., 1991. Biomechanical gait analysis in obese men. Archives of Physical Medicine and Rehabilitation 72, 1065-1070.

Stacoff, A., Quervain, I., Luder, G., List, R., Stussi, E., 2007. Ground reaction forces on stairs part ii: knee implant patients versus normal. Gait and Posture 26, 48-58.

Startzell, J., Owens, A., Mulfinger, L., Cavanagh, P., 2000. Stair negotiation in older people: a review. Journal of the American Geriatric Society 48, 567-580.

Stolze, H., Kuhtz-Buschbeck, J., Mondwurf., C., Johnk, K., Friege, L., 1998. Retest reliability of spatiotemporal gait parameters in children and adults. Gait and Posture 7, 125-130.

Strutzenberger, G., Richter, A., Schneider, M., Mundermann, A., Schwameder, H., 2011. Effects of obesity on the biomechanics of stair-walking in children. Gait and Posture 34, 119-125.

Svanstrom, L., 1974. Falls on stairs: an epidemiological accident study. Scandinavian Journal of Public Health 2, 113-120.

Teasdale, N., Hue, O., Marcotte, J., Berrigan, F., Simoneau, M., Dore, J., Marceau, P., Marceau, S., Tremblay, A., 2007. Reducing weight increases postural stability in obese and morbid obese men. International Journal of Obesity 31, 153-160.

Teasdale, N., Simoneau, M., Corbeil, P., Handrigan, G., Tremblay, A., Hue, O., 2013. Obesity alters balance and movement control. Current Obesity Reports 2, 235-240.

Tromp, A., Pluijm, S., Smit, J., Deeg, D., Bouter, L., Lips, P., 2001. Fall-risk screening test: a prospective study on predictors for falls in community-dwelling elderly. Journal of Clinical Epidemiology 54, 837-844.

Van der Esch, M., Steultjens, M., Harlaar, J., van den Noort, J., Knol, D., Dekker, J., 2011. Lateral trunk motion and knee pain in osteoarthritis of the knee: a cross-sectional study. Musculoskeletal Disorders 12, 141-147.

Vellas, B., Wayne, S., Romero, L., Baumgartner, R., Rubenstein, L., Garry, P., 1997. One-leg balance is an important predictor of injurious falls in older persons. Journal of the American Geriatrics Society 45, 735-738.

Wallace, C., Reiber, G., LeMaster, J., Smith, D., Sullivan, K., Hayes, S., Vath, C., 2002. Incidence of falls, risk factors for falls, and fall-related fractures in individuals with diabetes and a prior foot ulcer. Diabetes Care 25, 1983-1986.

Wearing, S., Henning, E., Byrne, N., Steele, J., Hills, A., 2006. The biomechanics of restricted movement in adult obesity. Obesity Reviews 7, 13-24.

World Health Organization: Obesity and Overweight Fact Sheet (2008). Available at <http://www.who.int/mediacentre/factsheets/fs311/en/>. Accessed December 2013.

Wikstrom, E., Powers, M., Tillman, M., 2004. Dynamic stabilization time after isokinetic and function fatigue. Journal of Athletic Training 39, 247-253.

Winter, D., 1995. Human balance and posture control during standing and walking. Gait and Posture 3, 193-214.

Winter, D., Patla, A., Ishac, M., Gage, W., 2003. Motor mechanisms of balance during quiet standing. Journal of Electromyography and Kinesiology 13, 49-56.

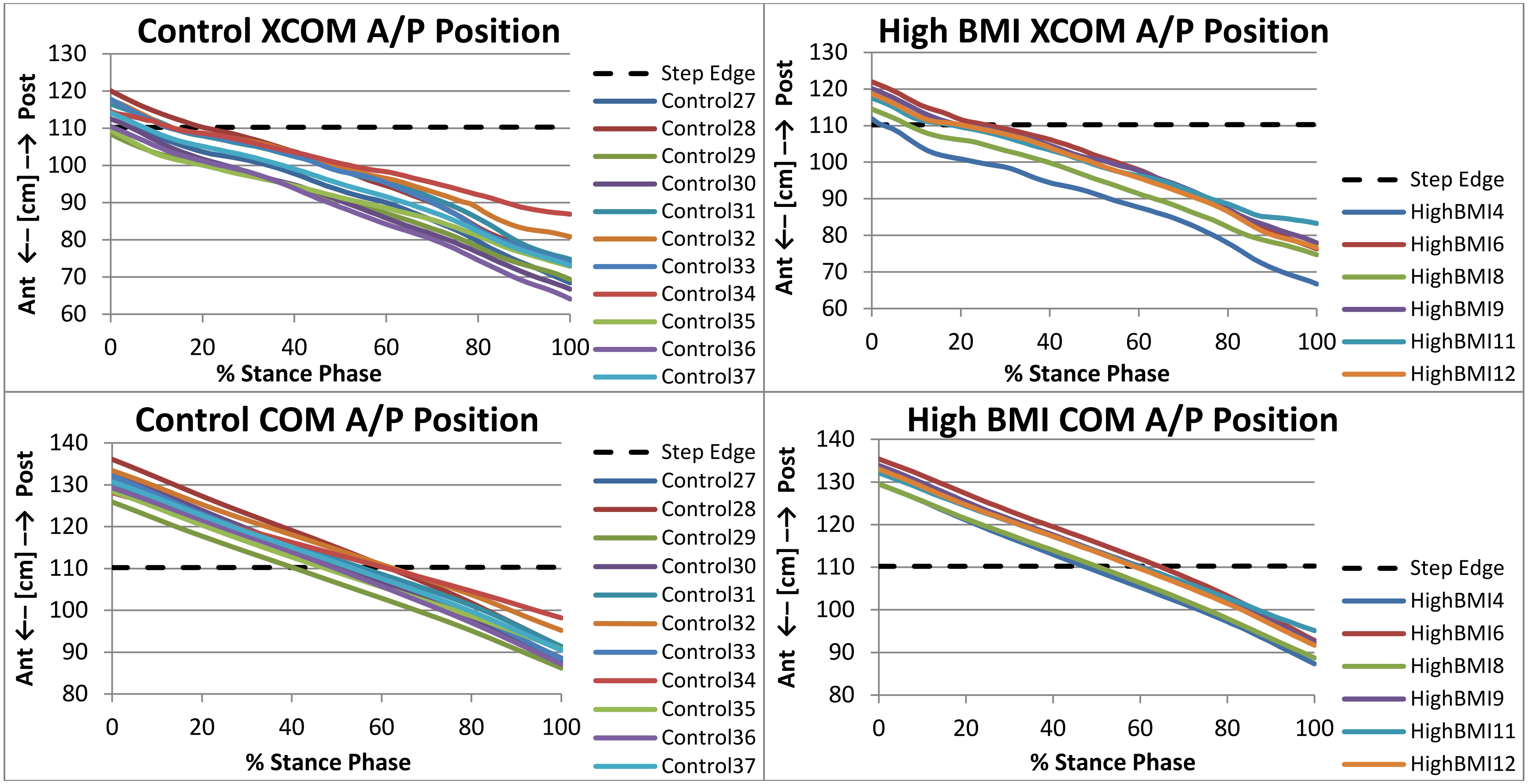
Wunderlich, R.,Cavanagh, P., 2001. Gender differences in adult foot shape: implications for shoe design. Medicine and Science in Sports 33, 605-611.

Yu, B., Stuart, M., Kienbacher, T., Growney, E., An, K., 1997. Valgus-varus motion of the knee in normal level walking and stair climbing. Clinical Biomechanics 12, 286-293.

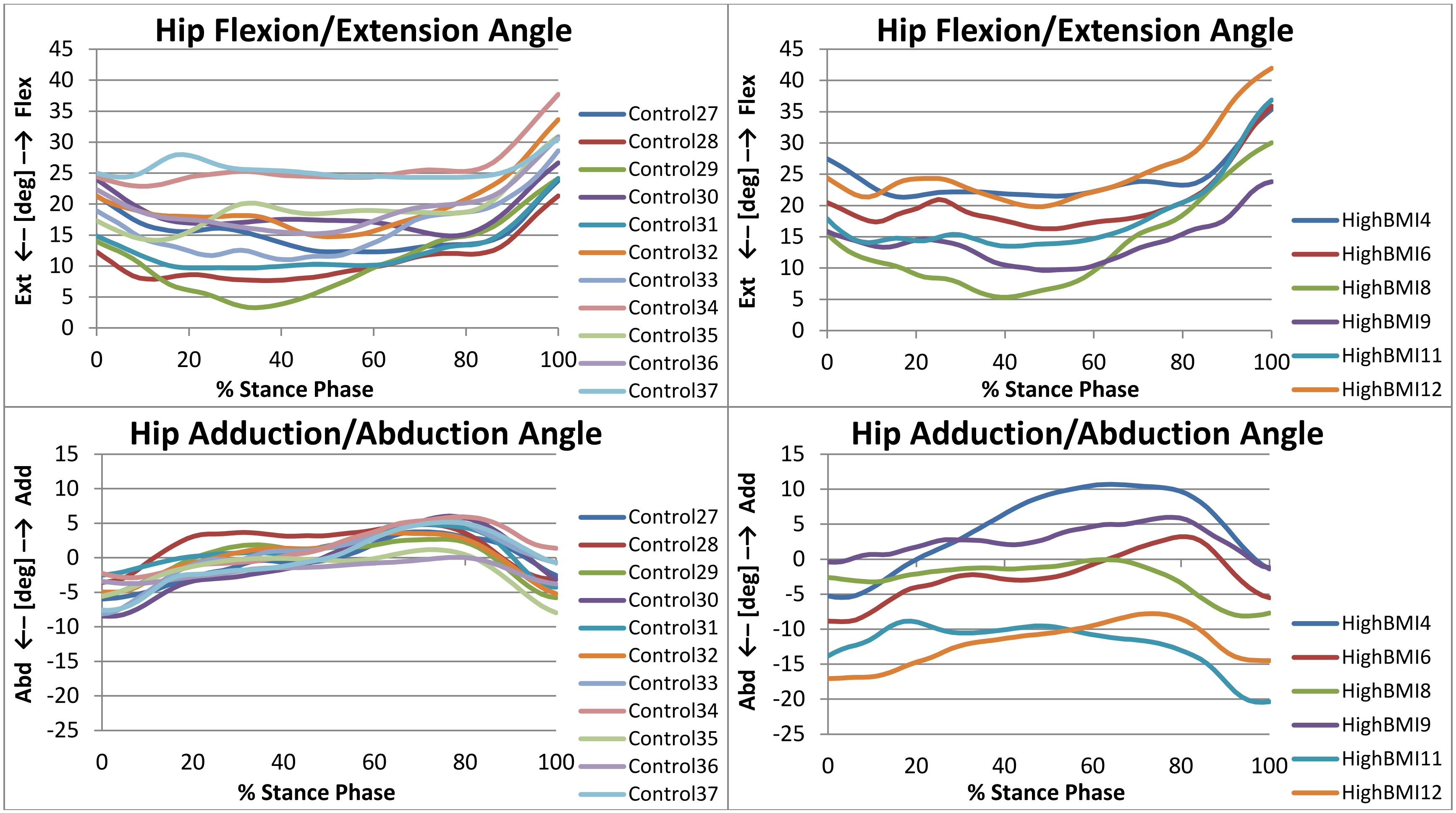
Zecevic, A., Salmoni, A., Speechley, M., Vandervoort, A., 2006. Defining a fall and reasons for falling: comparisons among the views of seniors, health care providers, and the research literature. The Gerontologist 46, 367-376.

APPENDIX A

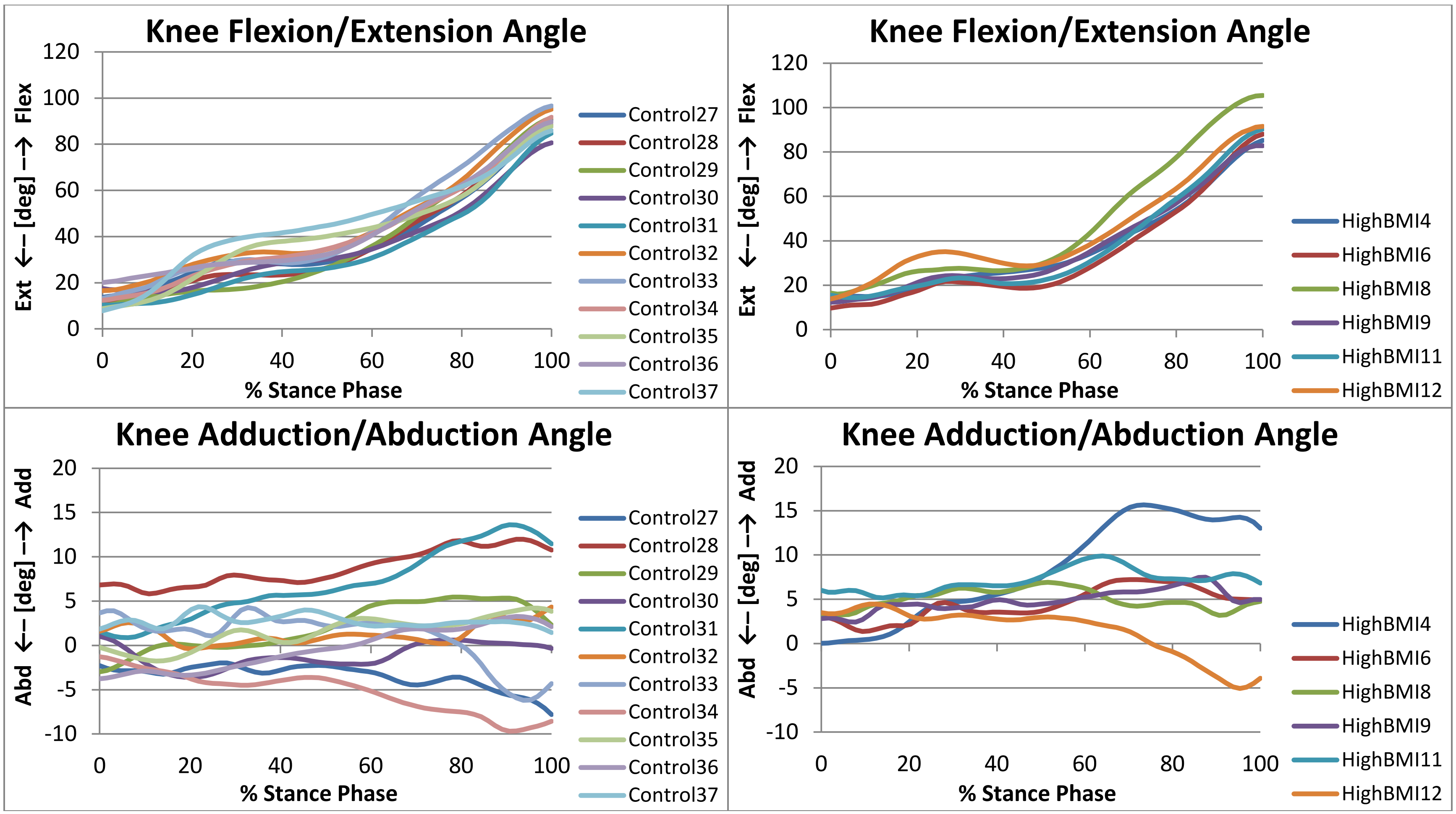
Extrapolated center of mass (XCOM) and center of mass (COM) anteroposterior (A/P) position:



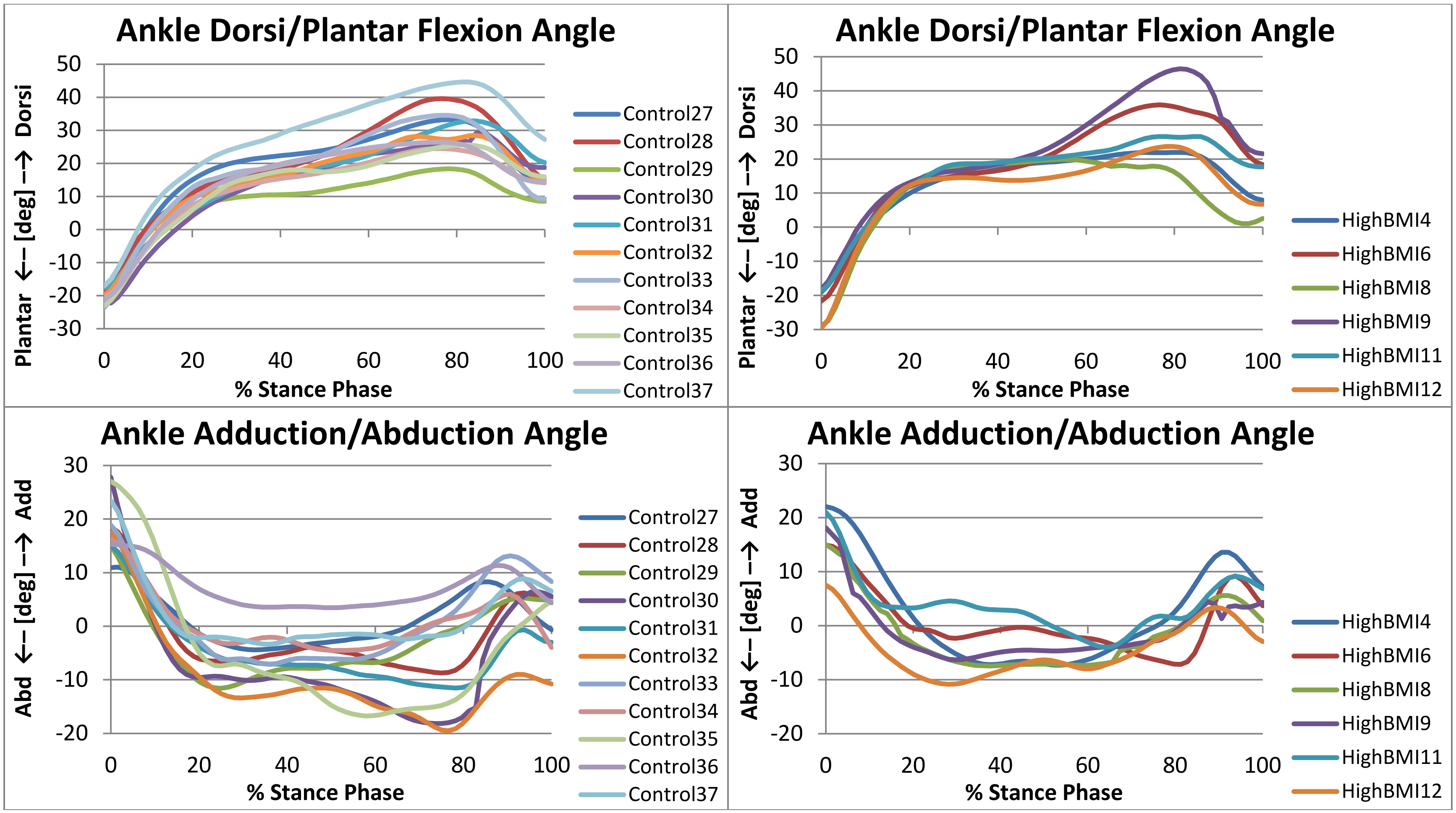
Hip angles in the sagittal and frontal planes:



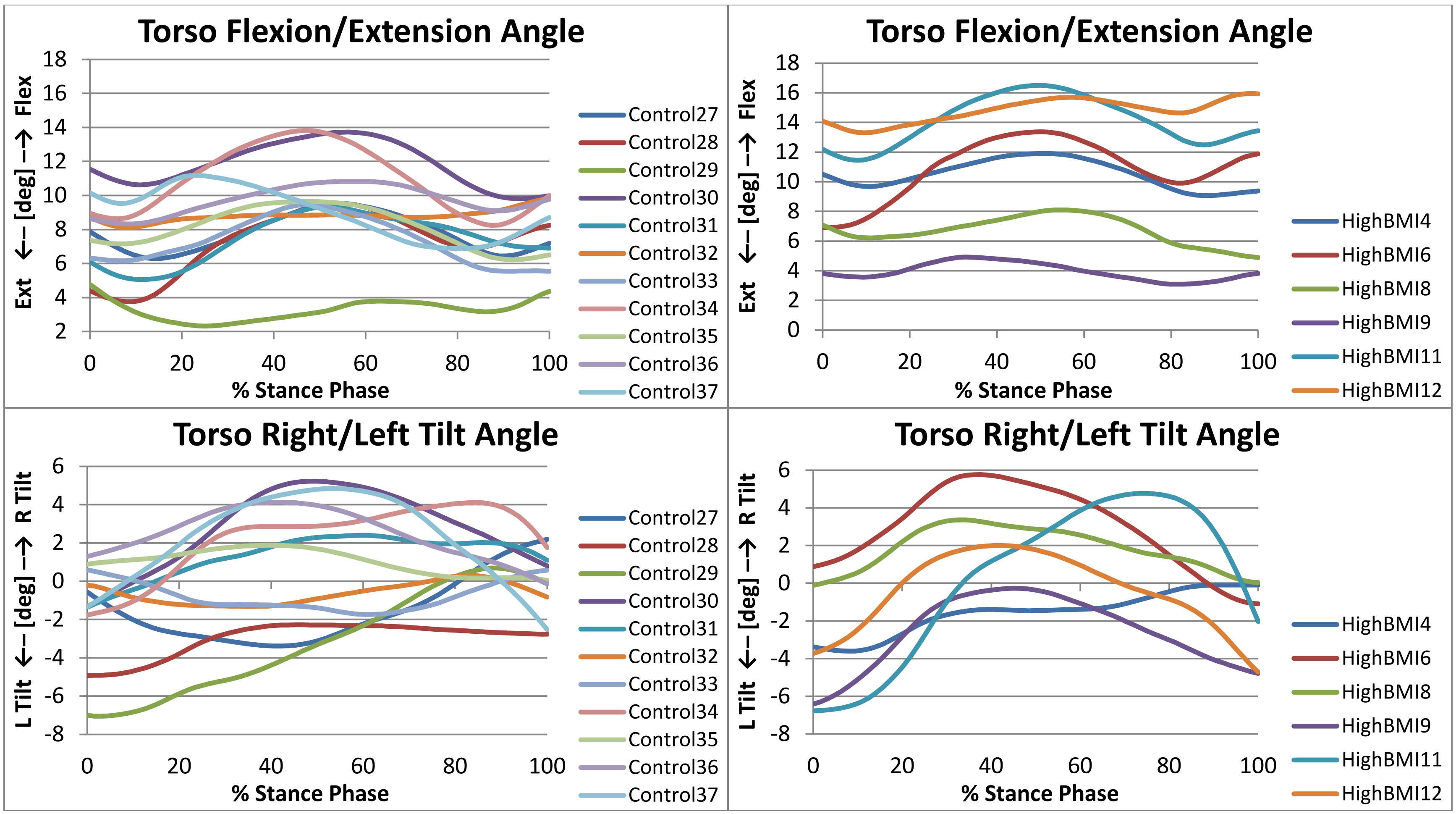
Knee angles in the sagittal and frontal planes:



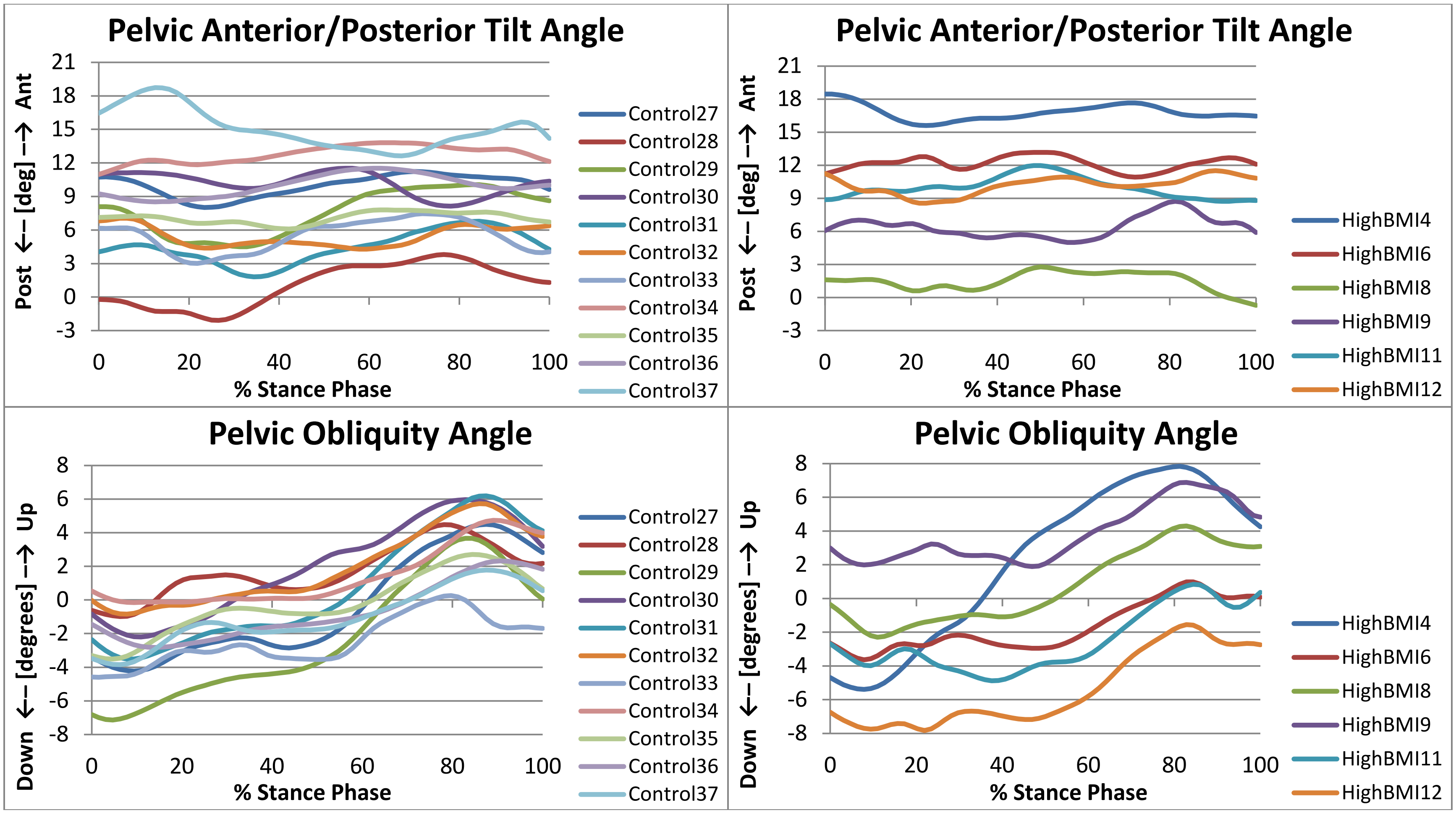
Ankle angles in the sagittal and frontal planes:



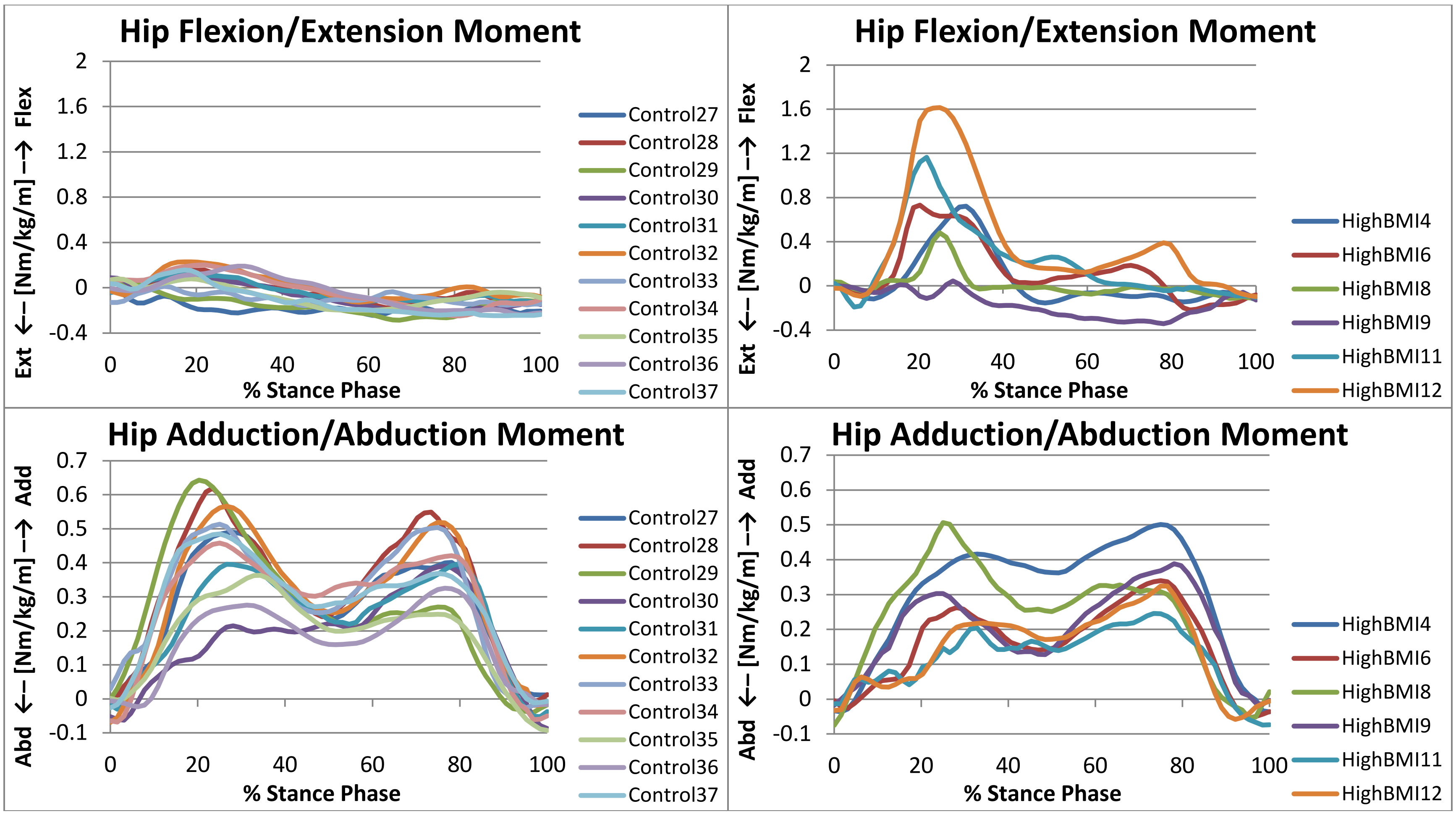
Torso angles in the sagittal and frontal planes:



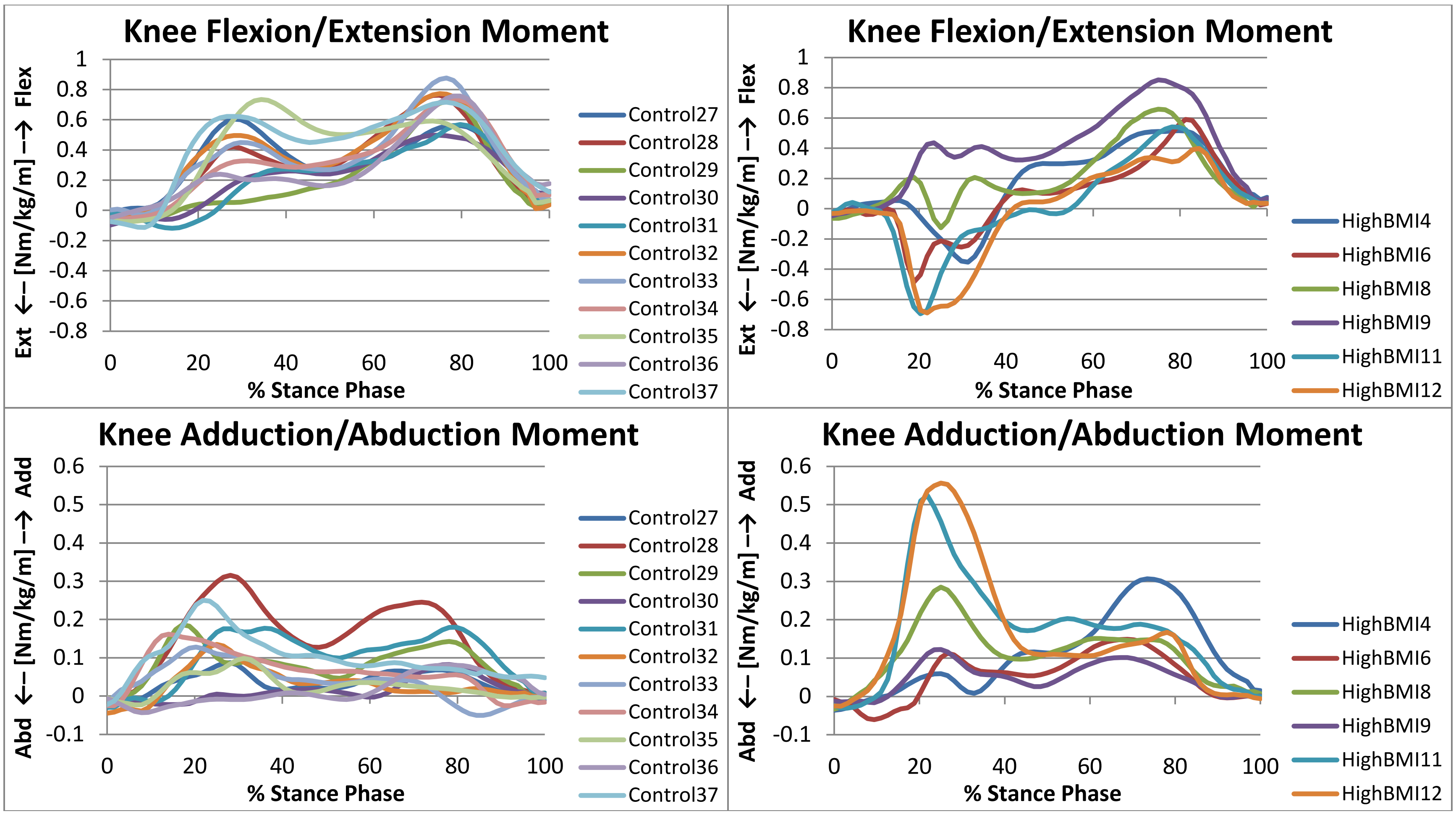
Pelvic angles in the sagittal and frontal planes:



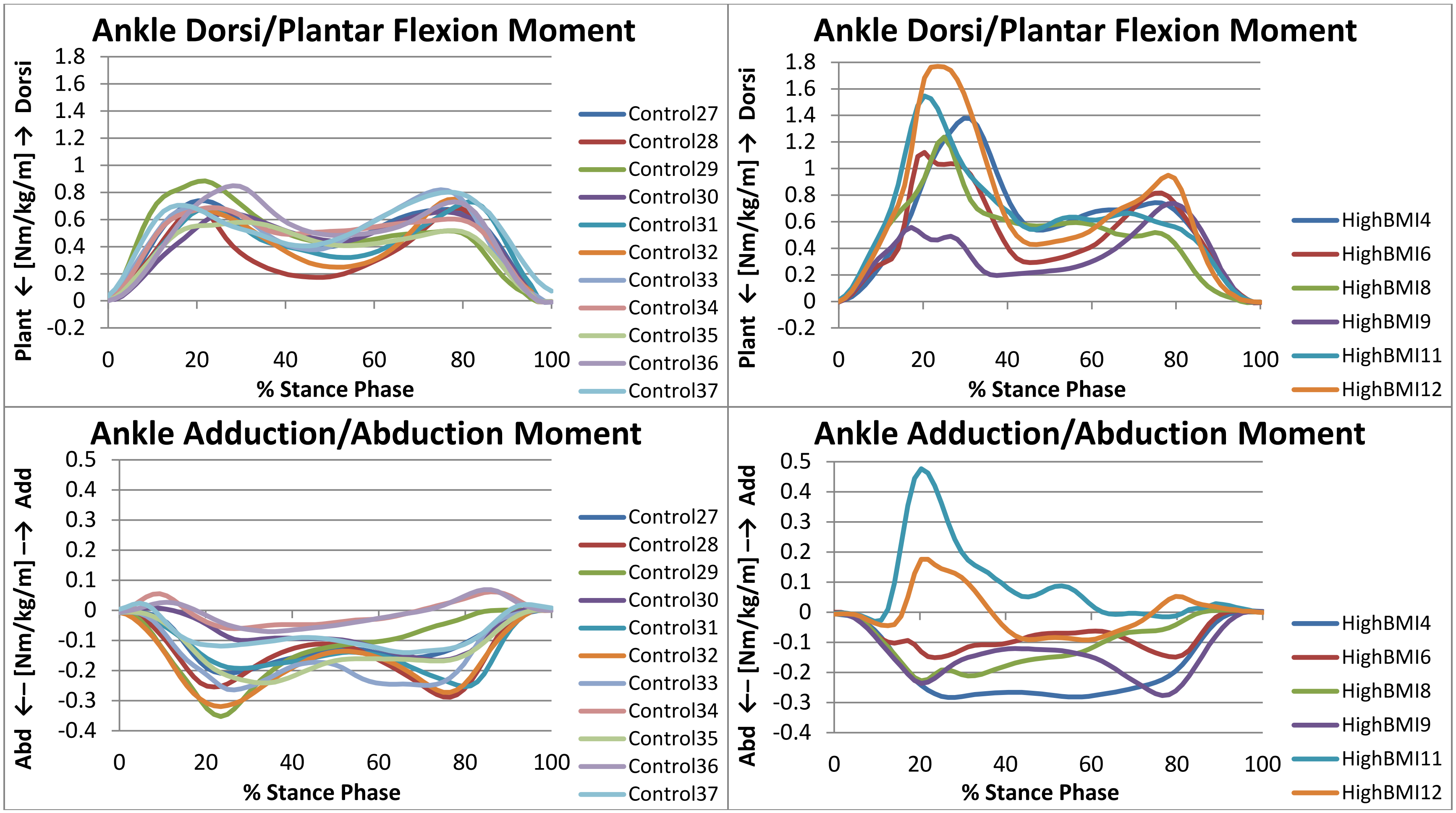
Hip moments normalized to body mass and body height in the sagittal and frontal planes:



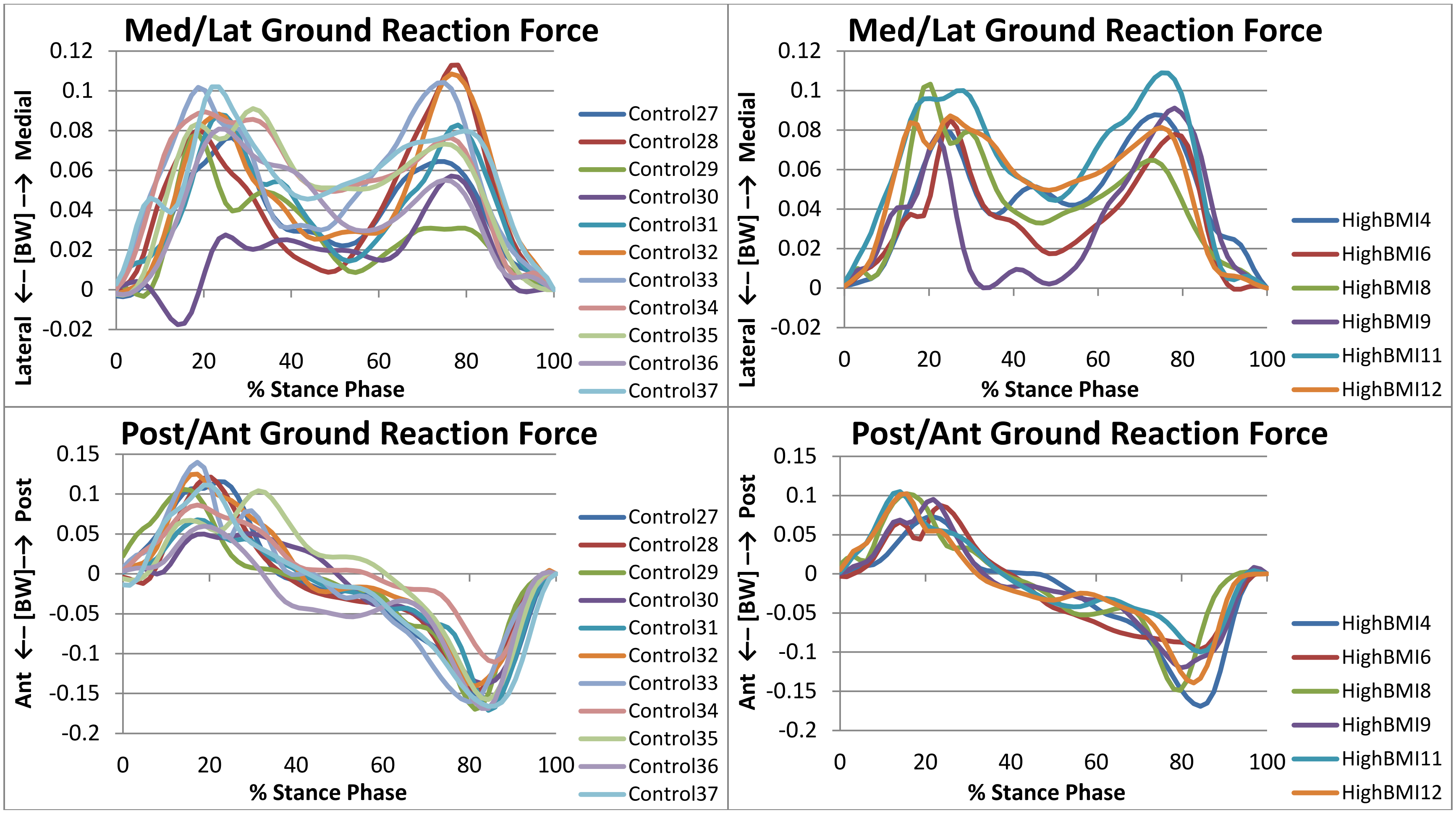
Knee moments normalized to body mass and body height in the sagittal and frontal planes:



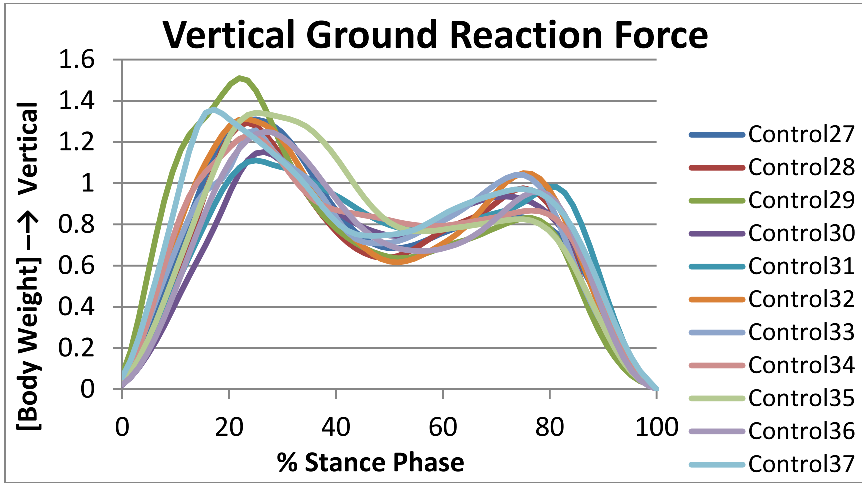
Ankle moments normalized to body mass and body height in the sagittal and frontal planes:

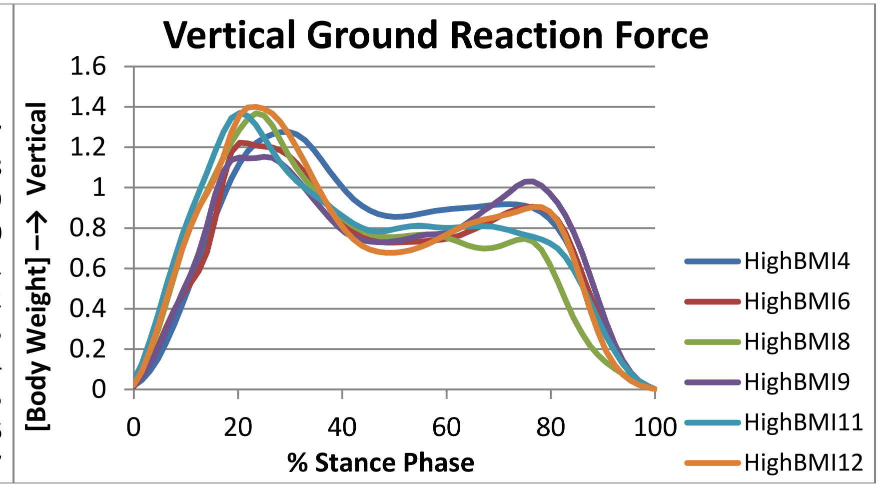


Ground reaction forces normalized to body weight:



Ground reaction forces normalized to body weight:





APPENDIX B

Study IRB approval letter.

THE UNIVERSITY OF MEMPHIS

Institutional Review Board

To: Brooke Sanford, John Williams, and Audrey Zucker-Levin
Biomedical Engineering

From: Chair, Institutional Review Board
For the Protection of Human Subjects
irb@memphis.edu

Subject: High BMI Impacts on Joint Function in Daily Activities (060911-781)

Approval Date: July 1, 2011

This is to notify you of the board approval of the above referenced protocol. This project was reviewed in accordance with all applicable statuses and regulations as well as ethical principles.

Approval of this project is given with the following obligations:

1. At the end of one year from the approval date, an approved renewal must be in effect to continue the project. If approval is not obtained, the human consent form is no longer valid and accrual of new subjects must stop.
2. When the project is finished or terminated, the attached form must be completed and sent to the board.
3. No change may be made in the approved protocol without board approval, except where necessary to eliminate apparent immediate hazards or threats to subjects. Such changes must be reported promptly to the board to obtain approval.
4. The stamped, approved human subjects consent form must be used. Photocopies of the form may be made.

This approval expires one year from the date above, and must be renewed prior to that date if the study is ongoing.

[redacted]

Digitally signed by [redacted]
DN: cn=[redacted], o=The University of Memphis, ou=IRB, email=[redacted]@memphis.edu, c=US
Date: 2011.07.03 13:58:08 -05'00'

Chair, Institutional Review Board
The University of Memphis

Cc: Dr. John Williams

APPENDIX C

Control and high BMI subjects' consent forms:

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER

Main Consent Form

Normative Data Base for Biomechanical Data in Healthy Subjects

Principal Investigator:	Audrey Zucker-Levin PhD, PT, MBA, GCS 930 Madison Avenue Memphis, TN 38163
Co-Principal Investigators:	Phyllis Richey, PhD John Williams, PhD
Sub-Investigators:	Kyle Huffman, BS Casey Hebert, BS Tyler Palumbo, BS Brook Sanford, PhD

1. INTRODUCTION:

In this consent form, the word "you" means you and/or your child.

You are being given the opportunity to participate in this research study because you are a normally developing individual and information on how you walk will be valuable for comparison to individuals who are not normally developing or have an injury. Research studies include only people who choose to take part. Please read this consent form carefully and take your time making your decision. As your study doctor discusses this consent form with you, please ask him/her to explain any words or information that you do not clearly understand. We encourage you to talk with your family and friends before you decide to take part in this research study. The nature of the study, risks, inconveniences, discomforts, and other important information about the study are listed below.

Please tell the study doctor if you are taking part in another research study.

The purpose of this study is to develop a data base of information that specifically describes how normal people walk. This data base will be used in the future as a comparison with people who have problems walking. Approximately 250 subjects will be participating in this study.

The study will take place at the Tennessee Health Management – Rehab America Biomechanics Laboratory (Motion Analysis Lab) located on the concourse level of 930 Madison, room EC013F. The Motion Analysis Lab is part of the University of Tennessee Health Science Center, Department of Physical Therapy.

Your participation in this study will last approximately 2 hours.

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER IRB NUMBER: 10-00746-XP
IRB APPROVAL DATE: 07/23/2013
IRB EXPIRATION DATE: 06/27/2014

2. PROCEDURES TO BE FOLLOWED:

The following procedures will occur during your evaluation:

- You will be asked to put on shorts and a t-shirt for the testing. If you have not brought shorts and a t-shirt with you, you will be provided with a pair of elastic waist athletic shorts and a t-shirt in your appropriate size. The shorts will be worn low on the hips to expose your hip bones (iliac crests).
- You will receive a standard physical therapy evaluation that includes the following:
    - Your height and weight will be measured with a standard physicians scale. The length of both legs will be measured with a standard tape measure from a point on your hip bone (Anterior Superior Iliac Spine or ASIS) to the inside ankle bone (medial malleolus).
    - A screening of the mobility of all of your joints will be performed to be sure you have no limitations in motion. If you appear to have restricted movement, a goniometer may be used to measure how mobile your joints are. A goniometer is a tool that is commonly used by doctors and physical therapists. It is placed on the skin surrounding the joint measured. An example of restricted movement is tight hamstrings. People with tight hamstrings are not able to stand and touch their toes while keeping their knee straight. If you have tight hamstrings, the amount of motion in your hip and knees will be measured with a goniometer.
    - The strength of your muscles will be tested using standard muscle testing procedures. The overall strength of your arm, leg, and trunk muscles will be tested by asking you to resist a force applied by the examiner at specific parts of your body. For example, to test the strength of your thigh muscles (quadriceps) the examiner will have you sit on a mat table with your feet dangling and ask you to resist pressure applied at the ankle.
    - Your reflexes will be tested with a reflex hammer. The examiner will use the reflex hammer to tap the tendons at your knees, ankles and elbows. Your posture will then be examined in standing. You will then be observed walking barefoot for approximately one minute.
- After the standard physical therapy evaluation is completed, the following will take place:
    - Approximately 20 reflective markers will be placed on specific parts of your body. These markers will be placed using double sticky tape and/or specially designed Velcro straps. If your skin is oily, alcohol will be used to assure sticking of the tape. These markers will provide information on the location of your limbs when you are walking.
    - In addition to reflective markers, small electrodes will be placed on 8 muscles of your legs (front and back of upper leg, front and back of lower leg). Like the reflective markers, these electrodes will be placed using double sticky tape and/or specially designed Velcro straps. Alcohol will be used on the skin that will be under the electrodes to improve the ability of the electrodes to pick up the electrical signals produced by your muscles. If you have excessive hair at the site

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER IRB NUMBER: 10-00746-XP
IRB APPROVAL DATE: 07/23/2013
IRB EXPIRATION DATE: 06/27/2014

that the electrode needs to be placed, the hair will be shaved with a disposable razor. Shaving cream will be used to condition the skin prior to shaving. These electrodes will provide information on when your muscles turn on and turn off during walking.

- After the markers and electrodes are placed, you will be asked to walk over a specially designed 25 foot-long platform positioned in the center of the room. You will be asked to walk approximately 10 times at your normal comfortable walking speed, 10 times at a slow walking speed, and 10 times at a fast walking speed. After walking, the markers and electrodes will be removed.

3. RISKS ASSOCIATED WITH PARTICIPATION:

As a result of your participation in this study, you are at risk for the following side effects:

Rare (1-5%)

- Mild skin irritation from the tape used to adhere the reflective markers and electrodes
- Mild skin irritation from the use of the disposable razor or shaving cream
- Minor cut if a razor is used

Very Rare (<5%)

- Muscle soreness due to the muscle testing procedure
- Tripping and falling when walking

You should discuss these with the examiner and your regular health care provider if you choose.

Any significant new findings developed during the course of this research project, which may impact upon the safety and efficacy of the procedure or treatment under study and consequently influence your willingness to continue participation, will be provided to you.

4. BENEFITS ASSOCIATED WITH PARTICIPATION:

There are no direct benefits to you for participating in this study. There will be benefit to society from your participation. People with disabilities will be compared to the data obtained from your test. Treatment decisions and research findings may be made based upon the comparison.

5. ALTERNATIVES TO PARTICIPATION:

None of the procedures will be performed if you choose not to participate in the study.

6. CONFIDENTIALITY:

All your paper research records will be stored in locked file cabinets in the motion analysis lab and will be accessible only to research personnel.

There will be no identifiable information associated with electronic research records. You will be provided an identifying code number that will be used for all electronic research records. Additionally, the computer containing the coded data will be computer password protected and accessible only to research personnel.

Under federal privacy regulations, you have the right to determine who has access to your personal health information (called "protected health information" or PHI). PHI collected in this study may include your medical history, the results of physical exams, lab tests, x-ray exams, and other diagnostic and treatment procedures, as well as basic demographic information. By signing this consent form, you are authorizing the researchers at the University of Tennessee to have access to your PHI collected in this study. The Institutional Review Board (IRB) at the University of Tennessee Health Science Center may review your PHI as part of its responsibility to protect the rights and welfare of research subjects. Your PHI will not be used or disclosed to any other person or entity, except as required by law, or for authorized oversight of this research study by other regulatory agencies, or for other research for which the use and disclosure of your PHI has been approved by the IRB. Your PHI will be used only for the research purposes described in the Introduction of this consent form. Your PHI will be used until the study is completed.

You may cancel this authorization in writing at any time by contacting the principal investigator listed on the first page of the consent form. If you cancel the authorization, continued use of your PHI is permitted if it was obtained before the cancellation and its use is necessary in completing the research. However, PHI collected after your cancellation may not be used in the study. If you refuse to provide this authorization, you will not be able to participate in the research study. If you cancel the authorization, you will be withdrawn from the study. Finally, the federal regulations allow you to obtain access to your PHI collected or used in this study.

You will not be identified in any presentations or publications based on the results of this research study.

7. COMPENSATION AND TREATMENT FOR INJURY:

You are not waiving any legal rights or releasing the University of Tennessee or its agents from liability for negligence. In the event of physical injury resulting from research procedures, the University of Tennessee does not have funds budgeted for compensation, either for lost wages or for medical treatment. Therefore, the University of Tennessee does not provide for treatment or reimbursement for such injuries.

If you suffer a research related injury, your study examiner will provide acute medical treatment, and will provide you with a subsequent referral to appropriate health care facilities. You and/or your insurance carrier will be billed for the costs associated with the medical treatment of a research related injury.

There will be no compensation available for any ancillary expenses incurred as a result of research related physical injuries, such as additional hospital bills, lost wages, travel expenses, etc.

There will be no compensation available for any non-physical injuries that may be incurred as a result of research participation, such as exposure to criminal or civil liability, or damage to your reputation, financial standing, or employability.

8. QUESTIONS:

If you have any questions about this research study you may contact Audrey Zucker-Levin, PhD, PT, at 901.448.5888.

In the event of a research related injury, contact Audrey Zucker-Levin, PhD, PT, at 901.830.6954. This is a cell phone number accessible 24/7.

You may contact Terrence F. Ackerman, PhD, UTHSC IRB Chairman, at 901.448.4824 or visit the IRB website at http://www.uthsc.edu/research/research_compliance/IRB/participant_complaint.php if you have any questions about your rights as a participant in this study or your rights as a research subject.

9. PAYMENT FOR PARTICIPATION:

You will not be paid for participation in this research study. Your parking costs will be paid by a voucher given to you after this consent is reviewed.

10. COSTS OF PARTICIPATION:

There are no costs for participation in this study.

11. PREMATURE TERMINATION:

Your participation in this research study may be terminated by the investigator without regard to your consent for the following reasons:

- Failure of equipment
- Finding of excessive weakness in your legs, significantly diminished range of motion in your joints, and/or significant deviation of walking when observed; in which case, you will be recommended to see your primary care physician for further evaluation.

IRB NUMBER: 10-00746-XP
IRB APPROVAL DATE: 07/23/2013
IRB EXPIRATION DATE: 06/27/2014

12. VOLUNTARY PARTICIPATION:

Your participation in this research study is voluntary and your refusal to participate or your decision to withdraw will involve no penalty or loss of benefits to which you are otherwise entitled.

If you decide to stop being part of the study, you should tell your study investigator. In addition, any information that you have already provided will be kept in a confidential manner.

If you are a student of the University of Tennessee, participating or not participating in this study will in no way influence your grade in any course. If you are an employee of the University, participating or not participating will not affect your employment status.

THE UNIVERSITY of TENNESSEE HEALTH SCIENCE CENTER

13. CONSENT OF SUBJECT:

You have read or have had read to you a description of the research study as outlined above. The investigator or his/her representative has explained the study to you and has answered all the questions you have at this time. You knowingly and freely choose to participate in the study. A copy of this consent form will be given to you for your records.

Signature of Research Subject	**Date**	**Time**

Printed Name of Research Subject

Signature of Person Obtaining Consent	**Date**	**Time**

Printed Name of Person Obtaining Consent

In my judgment, the subject or the legally authorized representative has voluntarily and knowingly given informed consent and possesses the legal capacity to give informed consent to participate in this research study.

Signature of Investigator	**Date**	**Time**
Signature of Legally Authorized Representative	**Date**	**Time**

Relationship of Legally Authorized Representative

Assent of Minor (Ages 14-17)	**Date**	**Time**

IRB NUMBER: 10-00746-XP
IRB APPROVAL DATE: 07/23/2013
IRB EXPIRATION DATE: 06/27/2014

Normative Data Base for Biomechanical Data in Healthy Subjects

Assent Discussion for Subjects 8-13 years of Age

A. Assent Obtained

☐ The assent discussion was initiated on _____________ (date) at _________ (time).

The information was presented in age-appropriate terms.

The minor: ____ (Subject's Name)

☐ Agreed to take part in the study on _______________ (date) at _____________ (time).
☐ Declined to take part in the study. The minor declined for the following reason(s):

RESEARCHER/DESIGNEE STATEMENT: I hereby certify that I have discussed the research project with the research participant and/or his/her parent(s) or legal guardian(s). I have explained all the information contained in the informed consent document, including any risks that may be reasonably expected to occur. I further certify that the research participant was encouraged to ask questions and that all questions were answered.

____
Researcher/Designee Printed Name

____________________________ ______________ ______________
Researcher/Designee Signature Date Time (AM/PM)

____
Minor Subject Printed Name

____________________________ ______________ ______________
Minor Subject Signature (8-13 years) Date Time (AM/PM)

B. Assent Not Obtained

☐ An assent discussion was not initiated with ____________________________ (Subject's Name) for the following reason(s):
- ☐ Minor is physically incapacitated
- ☐ Minor is cognitively or emotionally unable to participate in an assent discussion
- ☐ Minor refused to take part in the discussion
- ☐ Other____

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER IRB NUMBER: 10-00746-XP
IRB APPROVAL DATE: 07/23/2013
IRB EXPIRATION DATE: 06/27/2014

BMI Consent Form

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER

Principal Investigator: Audrey Zucker-Levin PhD, PT, MBA, GCS
930 Madison Avenue
Memphis, TN 38163

Co-Investigators: Brooke Sanford MS
930 Madison Avenue
Memphis, TN 38163

John L Williams PhD
330 Engineering Technology
University of Memphis
Memphis, TN 38103

1. INTRODUCTION:

In this consent form, the word "you" means you.

You are being given the opportunity to participate in this research study because you are a normally developing individual who has a body mass index (BMI) of 25 or greater. Information on how you walk will be valuable for comparison to individuals who are normally developing who have a BMI less than 25. Research studies include only people who choose to take part. Please read this consent form carefully and take your time making your decision. As your study doctor discusses this consent form with you, please ask him/her to explain any words or information that you do not clearly understand. We encourage you to talk with your family and friends before you decide to take part in this research study. The nature of the study, risks, inconveniences, discomforts, and other important information about the study are listed below.

Please tell the study doctor if you are taking part in another research study.

The purpose of this study is to compare how obese people walk and function compared to how normal people walk and function. Approximately 100 subjects will be participating in this study.

The study will take place at the Tennessee Health Management – Rehab America Biomechanics Laboratory (Motion Analysis Lab) located on the concourse level of 930 Madison, room EC013F. The Motion Analysis Lab is a part of the University of Tennessee, Health Science Center, Department of Physical Therapy.

Your participation in this study will last approximately 2 hours.

IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

2. PROCEDURES TO BE FOLLOWED:

The following procedures will occur during your evaluation:
First, you will receive a standard physical therapy evaluation that includes the following:

- Your height and weight will be measured with a standard physicians scale. The length of both legs will be measured with a standard tape measure by measuring a point on your hip bone (Anterior Superior Iliac Spine or ASIS) to the inside ankle bone (medial malleolus). A screening of the mobility of all of your joints will be performed to be sure you have no limitations in motion. If you appear to have restricted movement, a goniometer may be used to measure how mobile your joints are. A goniometer is a tool that is commonly used by doctors and physical therapists. It is placed on the skin surrounding the joint measured. An example of restricted movement is tight hamstrings. People with tight hamstrings are not able to stand and touch their toes while keeping their knee straight. If you have tight hamstrings, the amount of motion in your hip and knees will be measured with a goniometer. The strength of your muscles will be tested using standard muscle testing procedures. The overall strength of your arm, leg, and trunk muscles will be tested by asking you to resist a force applied by the examiner at specific parts of your body. For example, to test the strength of your thigh muscles (quadriceps) the examiner will have you sit on a mat table with your feet dangling and ask you to resist pressure applied at the ankle.
- Your reflexes will be tested with a reflex hammer. The examiner will use the reflex hammer to tap the tendons at your knees, ankles and elbows.Your posture will then be examined in standing You will then be observed walking barefoot for approximately 1 minute. After the standard physical therapy evaluation is completed, approximately 20 reflective markers will be placed on specific parts of your body. These markers will be placed using double sticky tape and/or specially designed Velcro straps. If your skin is oily, alcohol will be used to assure sticking of the tape. These markers will provide information on the location of your limbs when you are walking. In addition to reflective markers, small electrodes will be placed on 10 muscles of your legs (front and back of upper leg, front and back of lower leg). Like the reflective markers, these electrodes will be placed using double sticky tape and/or specially designed Velcro straps. Alcohol will be used on the skin that will be under the electrodes to improve the ability of the electrodes to pick up the electrical signals produced by your muscles. If you have excessive hair at the site that the electrode needs to be placed, the hair will be shaved with a disposable razor. Shaving cream will be used to condition the skin prior to shaving. These electrodes will provide information on when your muscles turn on and turn off during walking.After the markers and electrodes are placed, you will be asked to walk over a specially designed 25 foot-long platform positioned in the center of the room. You will be asked to walk approximately 10 times at your normal comfortable walking speed, 10 times at a slow walking speed, and 10 times at a fast walking speed. You will also be

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

asked to perform activities such as sitting in a chair, standing from a chair, squatting, and stair ascent and descent. After walking, the markers and electrodes will be removed.

- All procedures are performed for research purposes only.

3. RISKS ASSOCIATED WITH PARTICIPATION:

As a result of your participation in this study, you are at risk for the following side effects:

Rare (1-5%)

- mild skin irritation from the tape used to adhere the reflective markers and electrodes.
- mild skin irritation from the use of the disposable razor or shaving cream.
- minor cut if a razor is used.

Very Rare (<5%)

- Muscle soreness due to the muscle testing procedure.
- tripping and falling when walking and performing daily activities.

You should discuss these with the examiner and your regular health care provider if you choose.

Any significant new findings developed during the course of this research project, which may impact upon the safety and efficacy of the procedure or treatment under study and consequently influence your willingness to continue participation, will be provided to you.

4. BENEFITS ASSOCIATED WITH PARTICIPATION:

There are no direct benefits to you for participation in this study.

5. ALTERNATIVES TO PARTICIPATION:

Your participation is voluntary; you do not need to participate if you choose not to. You will not undergo any of the procedures if you do not participate in this study.

6. CONFIDENTIALITY:

All your paper research records will be stored in locked file cabinets in the motion analysis lab and will be accessible only to research personnel.

There will be no identifiable information associated with electronic research records. Each participant will be provided an identifying code number that will be used for all electronic research records. Additionally, the computer containing the coded data will be computer password protected and accessible only to research personnel.

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER
IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

Under federal privacy regulations, you have the right to determine who has access to your personal health information (called "protected health information" or PHI). PHI collected in this study may include your medical history, the results of physical exams, lab tests, x-ray exams, and other diagnostic and treatment procedures, as well as basic demographic information. By signing this consent form, you are authorizing the researchers at the University of Tennessee to have access to your PHI collected in this study The Institutional Review Board (IRB) at the University of Tennessee Health Science Center may review your PHI as part of its responsibility to protect the rights and welfare of research subjects. Your PHI will not be used or disclosed to any other person or entity, except as required by law, or for authorized oversight of this research study by other regulatory agencies, or for other research for which the use and disclosure of your PHI has been approved by the IRB. Your PHI will be used only for the research purposes described in the Introduction of this consent form. Your PHI will be used indefinitely.

You may cancel this authorization in writing at any time by contacting the principal investigator listed on the first page of the consent form. If you cancel the authorization, continued use of your PHI is permitted if it was obtained before the cancellation and its use is necessary in completing the research. However, PHI collected after your cancellation may not be used in the study. If you refuse to provide this authorization, you will not be able to participate in the research study. If you cancel the authorization, then you will be withdrawn from the study. Finally, the federal regulations allow you to obtain access to your PHI collected or used in this study. However, in order to complete the research, your access to this PHI may be temporarily suspended while the research is in progress. When the study is completed, your right of access to this information will be reinstated.

You will not be identified in any presentations or publications based on the results of this research study.

7. COMPENSATION AND TREATMENT FOR INJURY:

You are not waiving any legal rights or releasing the University of Tennessee or its agents from liability for negligence. In the event of physical injury resulting from research procedures, the University of Tennessee does not have funds budgeted for compensation either for lost wages or for medical treatment. Therefore, the University of Tennessee does not provide for treatment or reimbursement for such injuries.
If you suffer a research related injury, your study examiner will provide acute medical treatment and will provide you with a subsequent referral to appropriate health care facilities. You and/or your insurance carrier will be billed for the costs associated with the medical treatment of a research related injury.
No compensation will be available to you for any ancillary expenses incurred as the result of research related physical injuries, such as additional hospital bills, lost wages, travel expenses, etc.

THE UNIVERSITY of TENNESSEE UT HEALTH SCIENCE CENTER IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

No compensation will be available to you for any non-physical injuries that may be incurred as a result of research participation, such as exposure to criminal or civil liability, or damage to their reputation, financial standing, or employability.

8. QUESTIONS:

If you have any questions about this research study you may contact Audrey Zucker-Levin PhD, PT, at 901.448.5888.

In the event of a research related injury, contact Audrey Zucker-Levin PhD, PT, at 901.830.6954 This is a cell phone number accessible 24/7.

You may contact Dr. Terrence F. Ackerman, Ph.D., UTHSC IRB Chairman at 901-448-4824 or visit the IRB website at http://www.uthsc.edu/research/research_compliance/IRB/participant_complaint.php if you have any questions about your rights as a participant in this study or your rights as a research subject.

9. PAYMENT FOR PARTICIPATION:

You will be paid $25 for participation in this research study. Your parking costs will be paid by a voucher given to you after this consent is reviewed.

10. COSTS OF PARTICIPATION:

There are no costs for participation in this study.

11. PREMATURE TERMINATION:

Your participation in this research study may be terminated by the investigator without regard to your consent for the following reasons:

- Failure of equipment
- Finding of excessive weakness in your legs; significantly diminished range of motion in your joints; and/or significant deviation of walking when observed; in which case, you will be recommended to see your primary care physician for further evaluation.

12. VOLUNTARY PARTICIPATION:

Your participation in this research study is voluntary and your refusal to participate or your decision to withdraw will involve no penalty or loss of benefits to which you are otherwise entitled.

IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

If you decide to stop being part of the study, you should tell your study investigator. In addition, any information that you have already provided will be kept in a confidential manner.

If you are a student, you understand participating or not participating in this study will in no way influence your grade in any course. If you are an employee of the university, you should realize that participating or not participating will not affect your employment status.

Subject Initials ____

IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

THE UNIVERSITY of TENNESSEE HEALTH SCIENCE CENTER

13. CONSENT OF SUBJECT:

You have read or have had read to you a description of the research study as outlined above. The investigator or his/her representative has explained the study to you and has answered all the questions you have at this time. You knowingly and freely choose to participate in the study. A copy of this consent form will be given to you for your records.

Signature of Research Subject	**Date**	**Time**

Printed Name of Research Subject

Signature of Person Obtaining Consent	**Date**	**Time**

Printed Name of Person Obtaining Consent

In my judgment, the subject has voluntarily and knowingly given informed consent and possesses the legal capacity to give informed consent to participate in this research study.

Signature of Investigator	**Date**	**Time**

IRB NUMBER: 11-01282-XP
IRB APPROVAL DATE: 5/24/2011
IRB EXPIRATION DATE: 04/02/2012

APPENDIX D

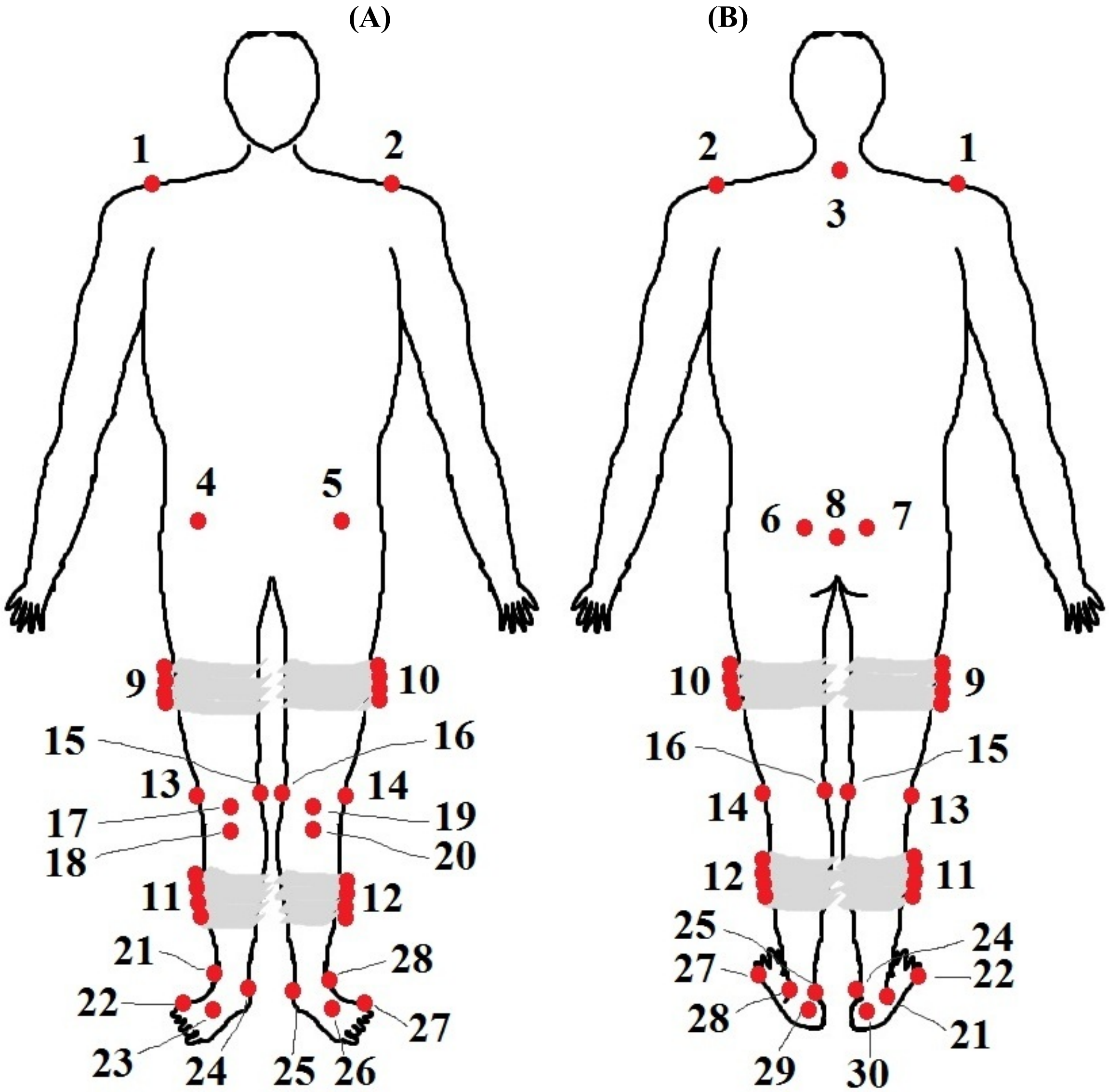


Figure 15. Retroreflective marker locations (A) Front (B) Back. 1) right acromion, 2) left acromion, 3) C7 of spine, 4) right anterior superior iliac spine, 5) left anterior superior iliac spine, 6) left posterior superior iliac spine, 7) right posterior superior iliac spine, 8) sacrum, 9) right thigh rigid array, 10) left thigh rigid array, 11) right shank rigid array, 12) left shank rigid array, 13) right lateral femoral epicondyle, 14) left lateral femoral epicondyle, 15) right medial femoral epicondyle, 16) left medial femoral epicondyle, 17) right inferior patella, 18) right tibial tuberosity 19) left inferior patella, 20) left tibial tuberosity, 21) right lateral malleolus, 22) right 5th metatarsal head, 23) right dorsum of foot, 24) right medial malleolus, 25) left medial malleolus, 26) left dorsum of foot, 27) left 5th metatarsal head, 28) left lateral malleolus, 29) left calcaneus, 30) right calcaneus