Summer 2013

The Interplay Between Obesity, Biomechanics and Fitness Within the Reverse Causation Hypothesis

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Recommended Citation

Singh, Bhupinder. "The Interplay Between Obesity, Biomechanics and Fitness Within the Reverse Causation Hypothesis." PhD (Doctor of Philosophy) thesis, University of Iowa, 2013.
https://doi.org/10.17077/etd.0f00qet8

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THE INTERPLAY BETWEEN OBESITY, BIOMECHANICS AND FITNESS WITHIN THE REVERSE CAUSATION HYPOTHESIS

by

Bhupinder Singh

A thesis submitted in partial fulfillment of the
requirements for the Doctor of Philosophy degree in
Physical Rehabilitation Science
in the Graduate College of
The University of Iowa

August 2013

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CERTIFICATE OF APPROVAL

PH.D. THESIS

This is to certify that the Ph.D. thesis of

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ACKOWLEDGEMENTS

A thesis is by its nature not a project of a single person; there are many to thank and acknowledge at its conclusion. First and foremost, I owe many thanks to Dr. H. John Yack, for his ceaseless guidance and support. His faith in me as a doctoral student has motivated me throughout my time at the University of Iowa. I would like also to thank Dr. Kathleen Janz, for seeing the value of this work and joining its shaping. In addition, many thanks to Dr. Trudy Burns, Dr. Vanessa Curtis, and Dr. Laura Frey Law, for their guidance, insight and support as thesis committee members.

Thanks also to the faculty, staff, and students of the University of Iowa Physical Therapy Department. In particular, I wish to acknowledge Patricia Teran-Yengle and Shelby Francis, as well as the many students in the Gait Analysis Lab.

Additional thanks are due to the many people and communities at the University of Iowa that have so enriched my time here. Dr. Rajagopal and the Indian Winterim program, the University of Iowa Diversity Office, and the Graduate Student Senate have all, in their own way, expanded my horizons far beyond the lab.

Finally, I would like to thank my sister, Dr. Manpreet Kaur and my uncle Dr. Manohar Singh Batra for inspiring and guiding me, my many wonderful friends and comrades, Lindsay Fox for her love and support, and most importantly, my parents, Joginder Singh and Hardeep Kaur for raising their son.

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CHAPTER I
INTRODUCTION

Obesity levels remain high in the United States and continue to rise around the world, affecting all socioeconomic levels and ethnicities. Despite an increasing body of research, the relationships between obesity and social, environmental, genetic and biomechanical factors remain complex, and the underlying causes of the obesity epidemic are not sufficiently understood. Some of these causes may include biomechanical factors which may alter physical activity patterns. Biomechanics, in the form of joint stresses and range of motion, are thought to be altered in the presence of excess adiposity, though the nature of these alterations has not been fully explored. The changed biomechanical outcomes seen in obese individuals should be more closely observed to determine if excess adiposity inhibits or influences their capacity to perform physical activities.

The complexity of the relationship between biomechanics and obesity can be examined through the framework of the *reverse causation hypothesis*. A lack of physical activity is known to increase adiposity levels in adults and children (Must, 2005); the reverse causation hypothesis posits that the relationship between adiposity and physical activity is bidirectional. The reverse causation hypothesis describes a positive feedback loop in which obesity leads to physical inactivity, which leads to more obesity. Existing studies on the reverse causation hypothesis, both in adults (Peterson, 2004; Weiss, 2007; Godin, 2008) and in children (Janz, 2009; Kwon, 2011), have noted that emotional and social responses to physical activity impact an obese subject's motivation and perceived capacity to participate in physical activity. This thesis explored biomechanical factors that contribute to this feedback loop. In order to better understand the associations between obesity and biomechanics in children and adults, this thesis tested obesity-related biomechanical measures, in the form of joint stresses and restricted range of motion, with the goal of better understanding the biomechanical underpinnings of the positive feedback loop in the reverse causation hypothesis, and ultimately to improve obesity-related health promotion recommendations and interventions.

Obesity: An Epidemic

Weight issues are typically defined in adults as overweight (BMI >25.0 kg/m^2) and obese (BMI >30.0 kg/m^2). In children, overweight is defined as a BMI at or above the 85th percentile and obese at or above the 95th percentile for children of the same age and sex (Barlow, 2007). Globally, at least 400 million people were obese in 2005, with the number of obese expected to rise to 700 million by 2015 (Ogden, 2006). At present, 36% of adults and 17.1% of children and adolescents in the United States are obese (Flegal, 2012), with a significantly higher probability for overweight and obese adolescents to sustain weight problems into adulthood (Brio, 2010). The direct healthcare costs in the United States for obesity during 2010 were estimated to be $194 billion (U.S. dollars), in addition to the $59 billion that Americans spent on various weight loss programs and products (O'Brien, 2010).

The health repercussions of the obesity epidemic are staggering. The incidences of cardiovascular disease, type II diabetes, hypertension, stroke, cancers, and other morbidities have increased due to obesity. Musculoskeletal problems, like arthritis and lower back pain, are common amongst obese individuals; already, obesity is the single largest risk factor for knee osteoarthritis (Felson et al., 1988; Hunter, 2009). Type II diabetes mellitus, a disease strongly linked to obesity, already afflicts an estimated 285 million people world-wide and is on the rise (O' Brian, 2010; Neeland, 2012). Moreover, conditions such as hypertension and Type II diabetes mellitus, previously seen primarily in adults, have become more common amongst children and adolescents, mirroring the rise in childhood obesity. Not surprisingly, health-related quality of life and the subset of physical functioning have been found to be inversely related to weight status (Tsiros, 2011).

American College of Sports Medicine Recommendations for Weight Reduction

Walking, in combination with changes in diet, is commonly recommended as a convenient physical activity (PA) that can be used to expend a significant amount of metabolic energy (Browning, 2009). The American College of Sports Medicine (ACSM) recommends more than 30 minutes per day, five times a week (total 150 minutes/week), of moderate intensity PA (3.0-6.0 METs) for overweight and obese adults to improve health; additionally, more than 250 min/wk is recommended for long-term weight loss (Donnelly, 2009). The duration of recommended physical activity is even higher for children to maintain a healthy weight (Kwon, 2010). However, merely accumulating a weekly number of minutes of physical activity may not be sufficient for weight loss. Instead, it is likely that continuous exercise in at least 30 minute increments may be required to achieve significant weight loss. Donnelly, et al. compared physical activity performed continuously, for 30 min, 3 days per week (90 minutes/week) with physical activity performed in intermittent sessions totaling 30 minutes per day, 5 days a week (150 minutes/week) in women for 18 months (Donnelly, 2009). The results showed that the continuous activity group lost greater weight than the intermittent group, which underscores the value of continuous activity in efforts to lose weight.

However, some researchers have contended that 30 minute walking programs are too ambitious for obese individuals starting a walking program (Hill, 2005, Davis, 2006). Their claim is supported by recent data showing that only 3% of adult obese women meet the recommended physical activity guidelines for weight loss (420 min PA weekly) (Ekkekakis, 2010). Basic compliance with walking programs is often erratic, with individuals dropping out after a few follow up visits. Investigators have found several reasons for attrition, including diet issues (Yaas, 1993), weight loss expectations (Teixeira, 2004), lack of motivation, (Andersson, 1997) and lack of family support (Lantz, 2003). Musculoskeletal issues are also included as one of the precipitating factors that cause attrition and non-compliance in weight reduction programs that recommend physical activity (Grossi, 2006; Honas, 2003; Bish, 2002; Ekkekakis, 2010). One potential musculoskeletal issue reported by patients in walking programs is joint pain, with greater involvement of the knee joint (Saris, 1992; Carol, 2010). One of the potential causes of joint pain may be altered mechanics at the weight bearing joints, especially the knee. The external knee adduction moments are the primary modulator of load distribution in the medial compartments of the knee, widely used to predict progression and severity of knee osteoarthritis (Miyazaki, 2002). Higher knee adduction moments have been reported in obese as compared to non-obese individuals (Browning, 2007), increasing the risk factors for tissue deterioration, knee pain and knee osteoarthritis (Russel, 2010) in obese individuals participating in walking programs. For some obese individuals, the reality of musculoskeletal problems may outweigh the eventual benefits of physical activity and weight loss.

Biomechanical Impacts of Obesity

The relationship between biomechanics and obesity is complex. Obesity may increase joint stresses during physical activity, such as walking, and also restrict the range of motion during daily functional activities. Figure 1-1, below, has been constructed to illustrate the complexity of this relationship. This thesis directly tested decreased range of motion (Chapter 2), and the Changing Gait Mechanics associated with Increased Joint Stress (Chapter 3, Chapter 4). Additionally, this thesis tested the impact of cardiorespiratory fitness and fatigue on joint mechanics (Chapter 3, Chapter 4). The implications of some relationships as suggested in Figure 1-1 ("Discomfort and Reduced Capacity," "Decreased Movement and Function") are beyond the scope of this thesis, but are critical elements within the positive feedback loop established by the reverse causation hypothesis.

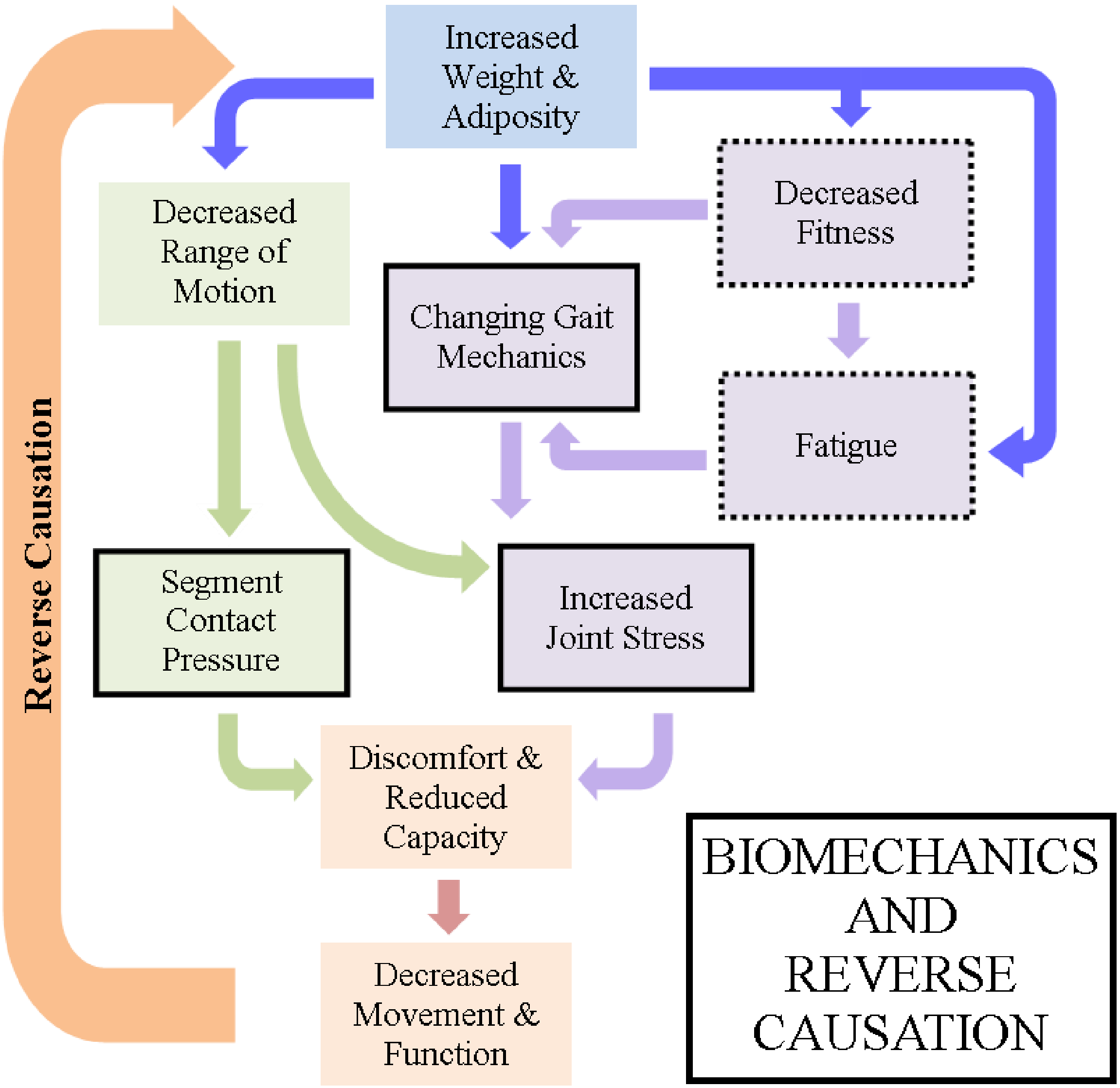

Figure 1-1: "Biomechanics and Reverse Causation" outlines the aspects of the reverse causation hypothesis feedback loop as explored by this thesis. The feedback loop is concluded by suggesting that increased contact pressure and joint stresses present in obese individuals subsequently inhibit future movement and function, resulting in the potential for ever higher incidence of obesity. The outside arrow, leading upwards from "Decreased Movement and Function" to "Increased Weight and Adiposity," summarizes the reverse causation hypothesis in action.

Gait and Biomechanical Stresses

In previous biomechanical studies investigating gait, obesity has been associated with slower gait speed (Lai, 2008; McGraw, 2000; Spyropoulos, 1991), wider step width, higher hip medial–lateral (ML) rotation and lower ankle anterior–posterior joint moments (Lai, 2008). Browning, et al. compared obese individuals to normal weight individuals by measuring ground reaction forces and found lower sagittal plane joint moments normalized to body weight in obese at different walking speeds (Browning, 2007). DeVita et al. reported less hip and knee flexion, and more ankle plantar flexion in obese individuals, producing a more erect walking pattern (DeVita, 2003). The moments and power at the knee were lower as compared to normal weight individuals, when normalized to body weight and height. The study also found that knee moments start to become coupled with body weight around a BMI of 30 kg/m^2, demonstrating a threshold where there is a neuromuscular adaptation to increased body weight.

Similar to obese adults, obese and overweight children (BMI > 85th percentile based on age and sex) typically walk slower than normal-weight children, and spend a longer time in stance phase (Nantel, 2006). In terms of biomechanical loads, obese children walk with a 40% higher peak normalized knee abduction moments during early stance (Davids 1996; Gushue, 2003). Nantel et al., found an increase in energy absorption by the hip flexors in obese as compared to normal weight children, indicating an increase in biomechanical stresses at the hip joint (Nantel, 2006). These studies investigated gait mechanics prior to any signs of fatigue; it can be argued that fatigue exaggerates gait abnormalities in overweight and obese children that might not be observed in the un-fatigued state, however no studies have been conducted to document changes in gait over time as obese children start to fatigue.

Limited Range of Motion

Obesity reduces joint range of motion as the adipose tissues around joints obstruct inter-segmental rotations (Gilleard, 2007; Singh, 2012). Range of motion reductions at the shoulder, lumbar spine and knee, suggest that obese individuals are likely to have limited functional reach capabilities in both the standing and seated positions when compared with normal weight individuals (Park, 2010). These limitations, due to excessive adipose tissue, could contribute to lower levels of physical activity due to discomfort, as well as a lower mobility (Larsson, 2001; Singh, 2012). The presence of excess soft tissue may impose different experiences upon obese individuals not only when they perform routine physical tasks (Larsson, 2001), but also during basic exercises like forward lunging, squatting or other common weight reduction exercises. Restriction in the range of motion may cause modification in the movement strategy, with implications for the associated biomechanical stresses. These biomechanically disadvantageous postures, combined with larger body masses would severely aggravate biomechanical stresses for obese individuals (Gilleard, 2007). Despite biomechanical differences, physical therapists and clinicians may not make different recommendations while prescribing exercises to obese individuals. Similar to walking, an increase in biomechanical stresses leading to pain and discomfort could be a reason for non-compliance with some weight reduction exercise programs (DeGrave, 2006).

Cardiorespiratory Fitness and Biomechanical Changes

Performance during gait and other daily activities can be affected by fitness levels. Of particular interest for the purposes of this thesis is cardiorespiratory fitness. Different submaximal and maximal tests (including the Nemeth, Ebelling, and Pacer protocols) assessing fitness levels by using measured or estimated $V0_2$ max have shown that obese individuals have lower fitness levels than their normal weight counterparts (McClain, 2006; Nemeth, 2008). Obesity is thus inversely related to cardiorespiratory fitness, and while obese individuals experience higher biomechanical loads as compared to their normal weight counterparts, the relationship between obesity, cardiorespiratory fitness and biomechanics has not been explored sufficiently in past studies.

Previous studies have related cardiorespiratory fatigue to VO_2 max and heart rate during an activity or exercise (Lollgen, 1980; Skurvydas, 2002), but there is no evidence of impact of cardiorespiratory fatigue on joint mechanics. Although it has been shown that muscular fatigue decreases muscle force generation and can reduce ability of muscle to attenuate ground reaction impact forces (Voloshin, 1998) leading to higher biomechanical stresses, it is unclear if cardiorespiratory fatigue will exhibit the same relationship with biomechanical stresses. While empirical clinical observation suggests fitness and fatigue in obese and overweight individuals are important issues that affect gait, these impressions have not been confirmed by scientific studies using biomechanical gait measures.

From the perspective of muscle function, there are physiological cues arising predominantly from active skeletal muscle to elicit changes in cardiorespiratory function. Changes in skeletal muscle structure have been observed in overweight and obese individuals compared to normal weight individuals, which may contribute to reduced cardiorespiratory fitness. These differences include a lower percentage of type I muscle fibers (Bercgren, 2004), impaired muscle oxidative capacity and microvascular function (Menshikova, 2005), inability to increase fat oxidation during exercise and increased intramuscular lipid storage (Menshikova, 2005; Blaak, 2004). In addition, other physiological effects of fatiguing exercise, especially the metabolic stress associated with muscle fatigue, might affect cardiorespiratory responses and performance during high-intensity exercise. Indeed, accumulation of various metabolites, such as lactic acid, is known to stimulate group IV and some group III muscle afferents that generate reflexes (the metaboreflex), which can significantly affect the cardiorespiratory responses to sustained exercise independently from increased central motor command (Smith, 2006; Marcora, 2007). While cardiorespiratory function can limit muscle function, it is the muscle that largely provides the stimuli for changes in cardiorespiratory function during exercise. Given the complex links and interactions between these factors, it is important to study how cardiorespiratory fatigue affects biomechanical outcomes.

Significance

Non-compliance by obese individuals to exercise recommendations suggests that current exercise regimes may not be meeting the needs of child and adult obese individuals. Obese individuals are more likely than normal weight individuals to suffer from increased biomechanical stresses while performing exercise, and even daily activities, resulting in pain, discomfort, and an increased potential for injury. Specifically, this thesis focused primarily on adult women, who are particularly at risk for developing musculoskeletal disorders like knee osteoarthritis. A better understanding of some of the specific biomechanical realities of obesity and how they contribute to the reverse causation hypothesis is critical to better address the physical needs of obese individuals and to interrupt the positive feedback loop of the reverse causation hypothesis.

Specifically, this thesis work made these primary contributions:

- Understanding the biomechanical stresses due to limitations in range of motion and strategies during common exercises inform rehabilitation approaches used with obese adults.
- Dependent changes in hip and knee joint loads during gait activities suggest biomechanical factors that could contribute to the reverse causation hypothesis.
- Understanding how cardiorespiratory fitness and cardiorespiratory fatigue, rather than adiposity, influence the biomechanical loads in children with excessive adiposity will better inform interventions and health promotion recommendations.

Purpose

The purpose of this thesis was to explore how segment biomechanics, in the form of joint moments and restricted range of motion, were influenced by obesity and fitness. The results from this work help to explain the biomechanical underpinnings of the reverse causation hypothesis in relation to obesity in both child and adult populations. The long-term goal of this research was to advance the reverse causation hypothesis as a framework in the development of better informed exercise interventions to promote physical fitness, and to achieve effective weight management for obese individuals. The purposes of this thesis were achieved by answering the following specific aims.

Specific Aim 1:

To analyze the biomechanics of obese and normal weight individuals, as measured by hip and knee moments while performing common physical therapy rehabilitation exercises.

Hypothesis 1:

It is hypothesized that restricted joint mobility in obese subjects will be associated with decreased hip and increased knee joint moments and that these differences will be more evident as the level of difficulty of squat and lunge increases.

Rationale 1:

Obese individuals have a limited range of motion as the adipose tissues around body joints are likely to interpose and obstruct inter-segmental rotations (Chaffin, 2006; Gilleard, 2007). Studies on sit-to-stand activity have also shown different strategies between obese and normal weight subjects: obese subjects tend to minimize trunk flexion while getting up from the chair (Sibella, 2003). The limitation could be attributed to the extra adipose tissue and less flexibility in obese individuals. It is expected that obese subjects will use similar strategies of limited trunk flexion during exercises like squat and forward lunge. This kinematic strategy brings to a minimization of hip joint torque, but it maximizes the moments at knee joint. The stress will further increase as the difficulty of activity increases, as obese individuals would try to keep their trunk erect as squat depth or lunge distance increases.

Specific Aim 2:

To assess the biomechanical gait changes in obese and normal weight adults following a 30-minute walking session.

Hypothesis 2:

It is hypothesized that the hip and knee adduction and extensor moments, which are the primary modulators of frontal and sagittal plane load distribution, will increase in obese individuals, following a 30-minute walking period, resulting in more stress across the hip and the knee joint.

Rationale 2:

The effect of continuous walking on biomechanics has not been documented in the past, but studies investigating running and drop-landing types of activities in adults have reported that muscular fatigue leads to an increase of ground impact forces (Christina, 2001). As the muscles starts to fatigue over a period of 30 minutes continuous walking, it results in less shock absorbency, an increase in loading rate, and a disproportionate increase in ground reaction force peaks over time (Parijat and Lockhart 2008). This increase in peak vertical ground reaction force during the weight acceptance phase of gait in a fatigued state, will lead to increased biomechanical stress on the joints. Furthermore, obese individuals have more percentage fat mass and lesser muscle mass than their normal weight counterparts, reducing ability of their muscles to attenuate ground reaction impact forces over time, leading to more stresses across the hip and knee joint.

Specific Aim 3:

To determine if gait biomechanics are associated with cardiorespiratory fitness and cardiorespiratory fatigue in overweight and obese children, aged 8-11 years.

Hypothesis 3a:

Gait biomechanics, as measured by lower limb moments, will be inversely related to cardiorespiratory fitness in overweight and obese children, in a non-fatigued state.

Hypothesis 3b:

Introduction of cardiorespiratory fatigue in overweight and obese children will be associated with an increase in lower limb moments as compared to the non-fatigued condition.

Hypothesis 3c:

The difference in lower limb moments between non-fatigued and fatigued states will not be related to the level of cardiorespiratory fitness.

Rationale 3:

Previous studies on the biomechanics of gait in overweight and obese children have shown that these children exhibit higher hip and knee joint stresses than their normal weight counterparts (Shultz, 2010; Strutzenberger, 2011). These studies did not account for the influence that cardiorespiratory fitness may have on biomechanical stresses. Cardiorespiratory fitness, quantified by VO_2 max, has been shown to have an inverse relationship with cardiovascular risk and adiposity in overweight and obese children (Nemeth, 2003). While cardiorespiratory fitness generally declines and biomechanical loads increase in the presence of obesity, some overweight and obese children clearly tolerate physical activity, pointing to a complex relationship between cardiorespiratory fitness and biomechanics. While no literature could be found on the effects of cardiorespiratory fatigue on gait biomechanics in obese children, several studies have noted the impact of muscular fatigue in obese individuals. There is a positive relationship between cardiorespiratory and muscular fatigue (Marcora, 2008). Furthermore, physiological effects of fatiguing exercise, especially the metabolic stress associated with muscle fatigue, might affect cardiorespiratory responses and performance during exercise (Smith, 2006). Therefore, understanding how muscular fatigue impacts gait biomechanics may assist in predicting outcomes due to cardiorespiratory fatigue.

These hypotheses were tested by collecting data during three different projects which make up the three core chapters of this thesis. The first study (Chapter 2), titled ‘Biomechanical Loads during Common Rehabilitation Exercises in Obese Individuals,’ analyzed the biomechanics of adult female subjects performing lunge and squat rehabilitation exercises. The second study (Chapter 3), titled “Changes in Gait over a 30 Minute Walking Session in Obese Females” assessed the biomechanical gait changes in obese subjects over a 30 minute walking session. The third study (Chapter 4), titled ‘Do Fitness and Fatigue affect Gait Biomechanics in Overweight and Obese Children?’ looked at obesity-related biomechanical issues in children. The fifth and final chapter gives the summary of the thesis and draws conclusions and public health implications.

CHAPTER II
BIOMECHANICAL LOADS DURING COMMON REHABILITATION EXERCISES IN OBESE INDIVIDUALS

Introduction

Squat and lunge exercises are classic exercises that have become an integral part of lower-extremity strengthening and postoperative rehabilitation programs and are universally used across the age and BMI spectrum of patients (Flanagan, 2004). The closed-chain, multi-joint, nature of these exercises is considered part of the basic rehabilitation strategy that has implications for improved performance in functional activities and gait. Gradation of these exercises not only challenges the torque requirements across the lower limb joints, but also challenges standing balance (Wilson, 2008).

Previous research on squat and lunge exercises has primarily focused on electromyographic (EMG) analysis to study muscle recruitment and strengthening issues (Jonhagen, 2009; Gorsuch, 2012) with few studies focusing on the biomechanics. Biomechanical analyses have demonstrated varying lower limb kinetic demands during rehabilitation of ACL reconstructive patients when performing the squat exercise (Salem, 2003). During the lunge exercise, the influence of forward trunk position on lower limb kinetics, specifically hip and knee joint moments, has been documented (Forrokhi, 2008). While these exercises are used across the age spectrum, most studies have been conducted in younger, normal weight, populations. Thus, the influence of age and obesity on performance has not been documented (Flanagan, 2003; Escamilla, 2001).

Although no studies of obese individuals performing these two activities have been conducted, previous studies underscore the potential for adiposity to influence activity performance. An increase in biomechanical stresses, as quantified by joint moments, has been reported during standing forward reaching tasks in obese subjects (Gilleard, 2007). This study suggested that increased moments were likely due to biomechanically disadvantageous postures used by obese individuals, rather than to their increased body mass. Underlying these postural deviations are reductions in joint range of motion (Park, 2010), which may cause modification in the movement strategy, with potential implications for increases in associated biomechanical stresses. Lower hip and higher knee extensor moments, attributed to limited trunk flexion were seen in obese, as compared to normal weight subjects when performing sit to stand activities (Sibella, 2003). It seems possible that obese subjects may use similar postural modifications and strategies when performing rehabilitation exercises, such as the squat and lunge, changing the biomechanical joint stresses, leading to unintended consequences.

Despite the potential for biomechanical differences in terms of increased joint stress and limited range of motion as compared to normal weight subjects, there are no published data showing that clinicians make different recommendations when prescribing exercises for obese individuals. Taking into consideration the biomechanical stresses and strategies during common exercises may help to inform rehabilitation approaches used for obese individuals. The purpose of this study was to analyze the biomechanics of obese and normal weight individuals, as measured by hip and knee moments, while performing common physical therapy rehabilitation exercises. It was hypothesized that restricted joint mobility in obese subjects would be associated with decreased hip and increased knee joint moments as compared to normal weight subjects and that these differences would be more evident as the level of difficulty of squat and lunge increased.

Methods

Ten obese females (BMI > 30 kg/m^2), age 37.4 ± 3.7 years, BMI 39.2 ± 3.7 kg/m^2 and ten normal weight (BMI<23 kg/m^2), age matched, female controls, age 38.1 ± 4.5 years, BMI 21.6 ± 2.3 kg/m^2, volunteered for the study that was approved by the University of Iowa Institutional Review Board. Height, weight, waist circumference; hip circumference and tibial length were recorded. Waist circumference was measured at the level of the right iliac crest and hip circumference was measured at the widest part of the hip with a tape measure (Gulick II, Country Technology Inc., Gays Mills, WI). Triads of infrared emitting diodes (IREDs) were placed on the pelvis and trunk, and bilaterally on the thighs, legs, and feet. Markers were affixed to the lateral aspect of the foot, to the shaft of the tibia, and to the lateral aspect of the thigh. Femoral epicondyle motion was tracked by two markers mounted on a custom femoral tracking device (Houck, 2000). Pelvic and trunk marker triads were attached to 5 cm extensions with base plates affixed over the sacrum and lower cervical vertebrae.

A link-based model was generated for tracking each segment. Anatomical landmarks were digitized, relative to segment local coordinate systems, with the subject standing in a neutral position, to create an anatomical model. Segment principal axes were defined by digitizing the following bony landmarks: Pelvis anterior and posterior superior iliac spines; Trunk: C-7 and L-1 vertebrae and glenohumeral joints; Thigh: functional hip joint center (Schwartz, 2005), lateral and medial condyles; Shank: lateral and medial condyles and malleoli; Foot: posterior heel, 5th metatarsal head, and second toe (Segal, 2009).

Kinematic data were collected using an Optotrak motion analysis system (Model 3020, Northern Digital Inc.,Waterloo, Ontario, Canada) operating at 60 Hz. Kinematic data were filtered at 6 Hz, using a zero phase lag, fourth-order, Butterworth low pass filter. Kinetic data were obtained using a Kistler force plate (Kistler Instruments, Inc., Amherst, NY). The force plate data were sampled at 300 Hz, and were filtered at 6 Hz. Visual 3D software (C-Motion Inc. Kingston, Ontario) was used to perform link-segment calculations.

Testing sessions included two trials each: Squatting down, feet shoulder width apart, with right foot on force plate and held for 3 s at three different knee angles: 60, 70, and 80 degrees (full knee extension being 0 degree). Real time feedback, showing a target line and a line representing the right knee angle in real-time, was used to achieve the desired knee angle (Teran-Yengle, 2011). Forward lunging, held for 3 s, with the feet shoulder width apart, positioned on the force plates at, three different distances between heel and toe: 1.0, 1.1, 1.2 times subject’s tibial length (Figure 1).

Figure 2-1: Shows the skeletal model of an obese female subject during squat exercise (left), placement of markers (center) and lunge exercise (right).

Data Analysis

Visual 3D software (C-Motion) was used for processing and the moments were normalized to body mass. Mean values, over 3 s while holding the positions, were calculated for lower limb joint range of motion, net joint moments and support moments (summation of the lower limb extensor moments). The mean of two trials for both activities was used for further analysis.

Statistical Analysis

Descriptive statistics in form of means and standard deviation were estimated. A repeated measures ANOVA model (3x2; joint moments by level of difficulty) with group (obese vs normal weight) as a between subject factor was fitted to investigate differences in hip, knee, ankle and support moments across three levels of difficulty for the squat and lunge activities. A group by level of difficulty interaction effect was included in the model. Pearson correlation coefficients were estimated to quantify the strength of the linear association between moments and range of motion. Regression analysis was performed to define relationships between BMI or other anthropometric measures and moments. SPSS 21.0 was used for analysis with p-value < 0.05 considered significant. All results are presented as means ± standard deviation.

Results

All the 20 subjects (10 obese and 10 normal weight) recruited for the study completed the study. No differences were seen in hip, knee and ankle range of motion between obese and normal weight subjects for squat and lunge activities (Table 2-1). Specifically, no difference was seen between the obese and normal weight groups for knee range of motion for squat 60°, squat 70° and squat 80° indicating that both groups performed the squat to the same depth. Also, no significant difference was seen in trunk flexion angle between the two groups for squat and lunge exercise (Table 2-1).

ROM	**Hip**		**Knee**		**Ankle**		**Trunk**	
	Obese	Normal	Obese	Normal	Obese	Normal	Obese	Normal
Squat 60°	64.7 (19.2)	62.3 (22.5)	59.6 (7.9)	57.5 (6.7)	119.6 (5.8)	119.3 (7.3)	33.7 (9.2)	34.4 (14.9)
Squat 70°	75.6 (23.2)	71.4 (25.0)	68.4 (8.5)	66.2 (7.5)	121.8 (6.3)	122.7 (7.3)	37.8 (11.0)	35.9 (15.6)
Squat 80°	85.0 (24.2)	82.4 (24.9)	78.3 (9.3)	75.3 (7.5)	124.1 (5.7)	125.5 (6.8)	41.2 (12.0)	40.6 (15.4)
Lunge 1.0	98.4 (12.2)	89.1 (20.9)	83.6 (12.7)	86.7 (9.3)	109.3 (6.9)	117.2 (10.6)	37.6 (7.8)	30.9 (12.0)
Lunge 1.1	102.4 (12.8)	91.4 (19.9)	88.0 (11.5)	85.9 (11.1)	109.2 (7.4)	114.1 (10.5)	37.3 (9.4)	31.2 (12.1)
Lunge 1.2	102.4 (14.3)	92.7 (17.7)	88.3 (13.4)	86.5 (10.0)	109.2 (8.9)	112.8 (9.3)	37.0 (9.2)	30.5 (11.6)

Table 2-1: Mean (standard deviation) hip, knee, ankle and trunk range of motion (ROM) for different levels of squat and lunge exercises in obese and normal weight subjects.

For the squat, hip and knee extensor moments in obese subjects were not different than normal weight subjects at any level of squat. Ankle extensor moments were higher in obese subjects for squat 80° ($p = 0.04$) (Table 2-2). The support moments were higher in obese subjects, as compared to the normal weight subjects, for squat 70° ($p = 0.03$) and squat 80° ($p = 0.01$), but not different for squat 60° ($p = 0.07$).

Moments (Nm/kg)	Hip Obese	Normal	Knee Obese	Normal	Ankle Obese	Normal	Support Obese	Normal
Squat 60°	0.22 (0.24)	0.12 (0.17)	0.67 (0.10)	0.59 (0.22)	0.28 (0.16)	0.19 (0.10)	1.18 (0.25)	0.92 (0.27)
Squat 70°	0.29 (0.28)	0.17 (0.18)	0.73 (0.12)	0.66 (0.23)	0.31 (0.19)	0.20 (0.13)	1.33 (0.32)	1.03 (0.30)
Squat 80°	0.37 (0.30)	0.24 (0.18)	0.82 (0.12)	0.75 (0.26)	0.34 (0.19)	0.20 (0.11)	1.53 (0.36)	1.18 (0.34)
Lunge 1.0	1.32 (0.27)	0.96 (0.39)	0.53 (0.15)	0.64 (0.30)	0.42 (0.20)	0.45 (0.26)	2.33 (0.36)	2.07 (0.65)
Lunge 1.1	1.41 (0.28)	1.07 (0.38)	0.53 (0.16)	0.56 (0.29)	0.43 (0.20)	0.42 (0.25)	2.44 (0.42)	2.05 (0.64)
Lunge 1.2	1.48 (0.32)	1.14 (0.39)	0.50 (0.22)	0.52 (0.24)	0.47 (0.21)	0.40 (0.22)	2.52 (0.47)	2.07 (0.59)

Table 2-2: Mean (standard deviation) hip, knee, ankle extensor and support moments for different levels of squat and lunge exercises in obese and normal weight subjects. The measures highlighted in grey color showed significant differences between two groups ($p<0.05$).

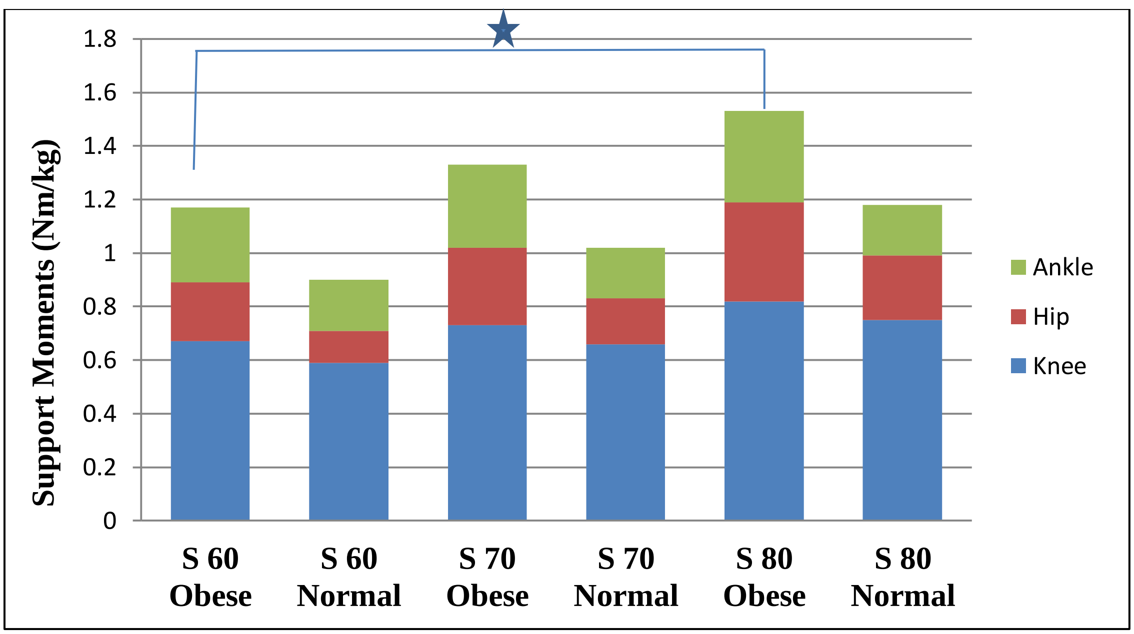

Figure 2-2: The support moments between squat 80° were greater (*) than squat 60° in obese subjects ($p = 0.01$). No differences were seen for normal weight subjects.

The higher support moments were primarily due to an increase in the knee extensor moments (correlation between knee and support moments was 0.71 and 0.64 for squat 60° and 80° respectively), which increased from 0.67 Nm/kg (squat 60°) to 0.82 Nm/kg (squat 80°) in obese subjects.

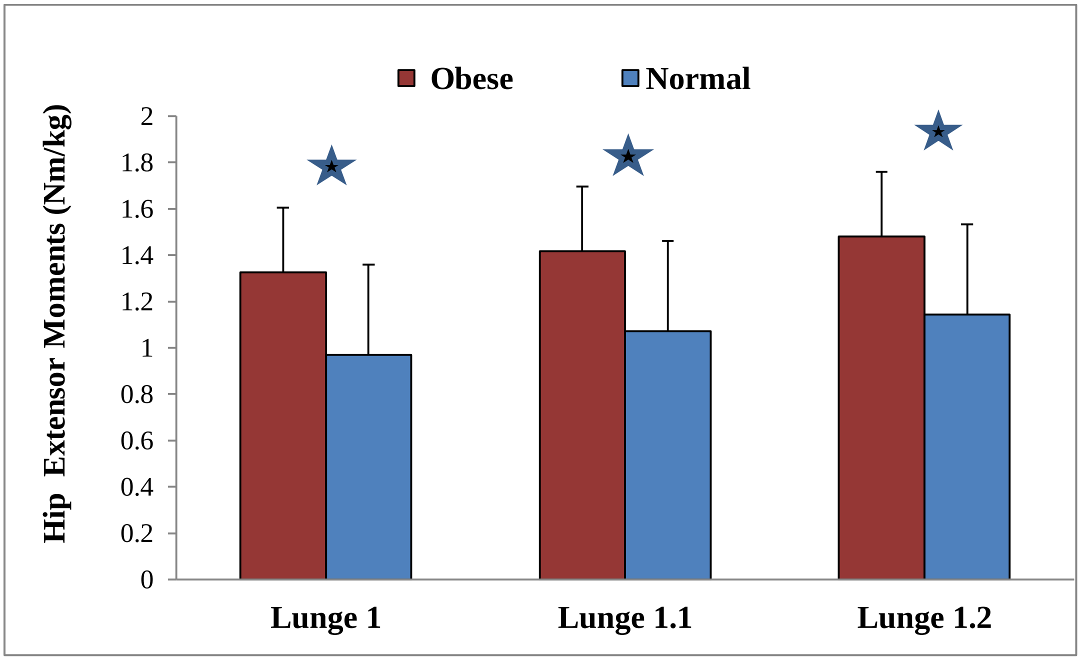

Figure 2-3: For the lunge, hip extensor moments were greater (*) in obese than normal weight subjects for level 1, 1.1 and 1.2 (p-values: 0.004, 0.003 and 0.007 respectively).

Knee and ankle extensor moments were not different between obese and normal weight groups (Table 2-2). Support moments showed an overall group effect between obese and normal weight subjects ($p = 0.01$). Correlations between extensor moments and range of motion were stronger in obese as compared to normal weight subjects for hip, knee and ankle during squat, as well as lunge trials (Table 2-3).

Squat	Hip			Knee		
Level	60°	70°	80°	60°	70°	80°
Obese	0.89	0.95	0.92	0.86	0.68	0.59
Normal	0.63	0.57	0.61	0.14	0.29	0.30
Lunge	**Hip**			**Knee**		
Level	Lunge 1.0	Lunge 1.1	Lunge 1.2	Lunge 1.0	Lunge 1.1	Lunge 1.2
Obese	0.67	0.77	0.75	0.64	0.77	0.64
Normal	0.38	0.36	0.28	0.04	0.19	0.01

Table 2-3: Pearson correlation coefficients between moments and range of motion at hip and knee joint for squat and lunge exercises. Correlation values above 0.48 are significant.

There was no linear association between BMI, waist circumference, and waist to hip ratio and joint moments for squat and lunge levels. However, when the data were split into obese and normal weight subjects based on BMI, obese subjects showed a stronger relationship (r=0.68) as compared to normal weight (r=0.51) for hip extensor moments for squat 60°.

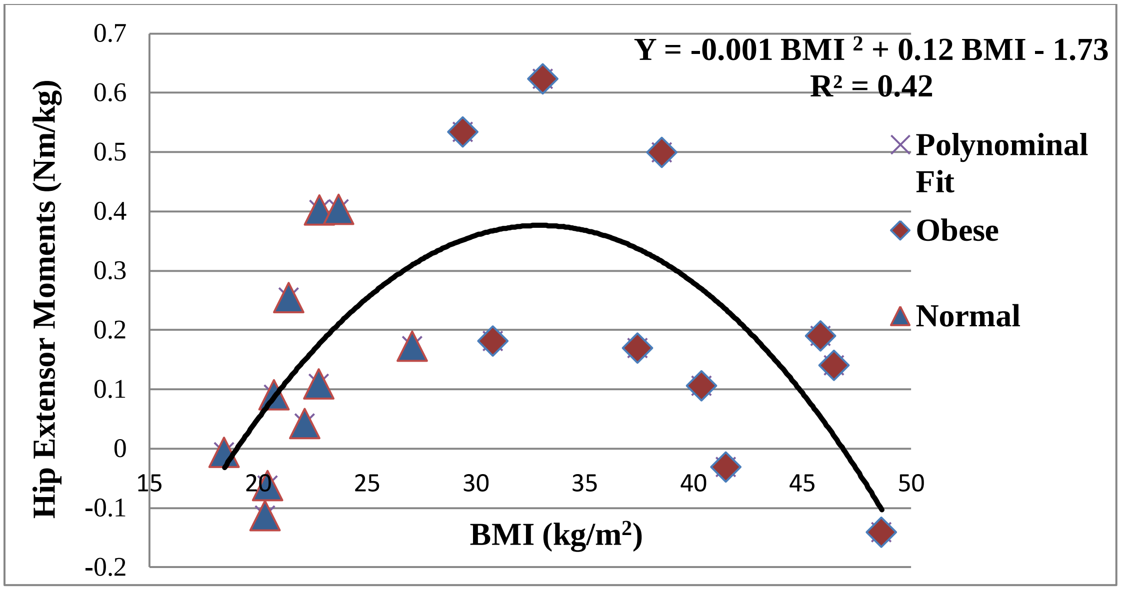

Figure 2-4: Relationship between peak hip extensor moments for obese and normal weight subjects for squat 60°, non-linear polynomial fit showed a moderate relationship between hip moments and BMI.

Similar relationships were seen for squat 70° (R^2=0.42) and squat 80° (R^2=0.39). A moderate relationship was seen for ankle (R^2=0.14) for squat 60° but no relationship was seen for knee (R^2=0.03).

Discussion

The purpose of this study was to analyze the biomechanics of obese and normal weight individuals, as measured by hip, knee and ankle moments, while performing squat and lunge exercises. For the squat exercise, the normalized support moments were higher in obese subjects when performing the two deeper squats (70° and 80°). The lunge exercise showed group differences in normalized hip moments and the support moments for all difficulty levels (1.0, 1.1 and 1.2 times tibial length). Joint range of motion was not different between obese and normal weight groups for either activity, however the obese group had higher correlations between range of motion and moments when compared to the normal weight group. The results suggest that obese individuals may have less flexibility in selecting movement strategies and experience higher biomechanical stress than normal weight subjects while performing basic rehabilitation exercises.

Previous studies of normal weight subjects reported similar magnitudes for knee extensor moments during the squat exercise (Salem, 2003). Support moments have also been used to characterize squat and stoop lifting techniques in normal weight individuals, however, peak magnitudes were higher than reported in the current study (Hwang, 2009). Subjects in this study, however, went into greater amounts of knee flexion (140^{o}) and when the moments were compared for the 60-80 degree range, the magnitudes were similar in both studies.

It was hypothesized that for squatting, obese subjects would have higher knee moments than normal weight subjects. While no significant differences were seen in the knee moments, there were increased support moments during squatting, with the main contribution appearing to come from the knee joint moments. Higher knee extensor moments were also reported in normal weight older adults (age 74.5 ± 4.39 yr) during squat as compared to a chair squat (Flanagan, 2003), even though the participants held onto a rail while performing the squat which could have implications of the kinetics. This increase in the net extensor moments, across the three lower limb joints, points to the possibility of higher generalized joint stress in obese subjects during squatting.

Analysis of the lunge exercise data showed an increase in the hip moments for the obese group and no differences in the knee and ankle moments. These results are contrary to the hypothesis that hip moments would decrease due to limited trunk flexion in obese individuals as has been reported in previous work on sit to stand in obese and normal weight individuals (Sibella, 2003). There was no difference in trunk flexion between obese and normal weight groups, so the increase in hip moments in the obese group could be due to mass distribution, bringing the center of mass forward. A recent study of the effect of adding external load on the biomechanics in young normal weight individuals during the lunge activity showed an increase in hip extensor moments with little change in the knee contributions (Reimann, 2012). Though there was no association between hip moments and waist circumference in the current study, a moderate correlation (r=0.46) was seen with waist to hip ratio in the obese group. It could be argued that the external weight simulates the excess adipose tissue in obese subjects, causing a similar increase in hip moments.

Although there were no significant differences in range of motion between obese and normal weight groups, stronger correlations between moments and range of motion were seen in the obese group. These higher correlations, in combination with higher support moments, might point to subtle restrictions in ROM (Park, 2010) or movement capability, giving the obese group less flexibility in how they accomplished the squat and lunge activities. Additionally, a non-linear relationship between hip moments and BMI was found (Figure 2-4). This suggests the possibility of a ceiling effect in subjects with higher BMIs. A similar ceiling effect was postulated for gait in obese individuals with BMI greater than 30 kg/m^2 (Devita, 2003) and was attributed to development of neuromuscular adaptations during gait in response to BMI or excess body mass. Similar adaptations or compensatory adjustments enables obese individuals to reduce the joint stress, relative to their BMI could explain the results in our study.

The present study had several limitations, e.g., it only examined the squat to a depth of 80 degrees of knee flexion, while some previous studies used a greater range of motion for squatting. The reduced range of motion for squatting was due to safety concerns for the obese individuals and their reluctance to perform deeper squats. Similar concerns limited the farthest lunge to 1.2 times the tibial length, which may not have challenged some subjects, particularly in the normal weight category. Additionally, the strategies subjects employed to reach the final position of the squat and the lunge exercise were not controlled. Though the analysis focused on the static result of motion, variation in the strategies to perform the motion might influence their static posture and thus the moments. Finally, the sample size may have limited our ability to find statistically significant differences in some of the outcome measures. As no previous studies on squat and lunge have been conducted in obese individuals, a sample size calculation was not feasible before the start of the study. However, post-hoc power analysis based on the means from the current study showed good power of 0.85 (85 percent) for lunge exercise, but low power (0.40) for squat trials.

Clinicians progress rehabilitation protocols by increasing the difficulty of the exercise: increasing the depth of the squat or increasing the distance between feet during the lunge. The current study identified significant increases in lower limb kinetics in obese individuals. These stressors may have consequences for the obese population where there is also an increased likelihood of joint pathology (Messier, 2005) and therefore using the same exercise progressions for both subject groups may not be justified. In addition, the sensitivity of joint moments to changes in ROM is greater for obese individuals, which would suggest that clinicians may need to be more sensitive to subtleties in performance. Finally, the non-linear associations that have been uncovered between anthropometric measures and kinetic measures make the assessment of how best to approach exercise in this population even more challenging. Thus, while this study advocates for the need to consider obesity as a factor in exercise prescription, it acknowledges the apparent complexity of issues that interact to bias the kinetic measures.

CHAPTER III
CHANGES IN GAIT OVER A 30-MINUTE WALKING SESSION IN OBESE FEMALES

Introduction

Thirty-four percent of adults in the United States are overweight (BMI 25.0 to ≤ 30.0 kg/m^2) and an additional 36% are obese (BMI >30.0 kg/m^2). Walking, in combination with changes in diet, is commonly recommended as a convenient physical activity (PA) that can be used to expend a significant amount of metabolic energy (Browning, 2009). Exercise programs typically recommend walking for at least 30-minutes per session to comply with the current recommendations from the American College of Sports Medicine (ACSM) (2009). However, some researchers have contended that 30 minute walking programs are too ambitious for obese individuals starting a walking program (Hill, 2005, Davis, 2006). Their claim is supported by recent data showing that only 3% of adult obese women meet the ACSM recommended physical activity guidelines for weight loss (Ekkekakis, 2010).

Musculoskeletal issues are one of the precipitating factors that cause attrition and non-compliance in weight reduction programs that recommend physical activity (Grossi, 2006; Honas, 2003; Bish, 2002; Ekkekakis, 2010). One musculoskeletal issue reported by patients in walking programs is joint pain, with greater involvement of the knee joint (Saris, 1992; Carol, 2010). One of the potential causes of joint pain may be altered mechanics at the weight bearing joints, especially the knee. The external knee adduction moments are the primary modulator of load distribution in the medial compartments of the knee, widely used to predict progression and severity of knee osteoarthritis (Miyazaki, 2002). Higher knee adduction moments have been reported in obese than non-obese individuals (Browning, 2007), increasing the risk factors for tissue deterioration, knee pain and knee osteoarthritis (Russel, 2010) in obese individuals participating in walking programs. DeVita, et al. found that absolute hip and knee peak extensor moments were greater in class III obese (BMI >40 kg/m^2) adults as compared to normal weight counterparts (DeVita, 2003). A further increase in the knee adduction and extensor moments, during a walking program, as the individuals start to fatigue, could help account for the discomfort/pain that is experienced by obese individuals. For some obese individuals, the reality of musculoskeletal problems may outweigh the eventual benefits of physical activity and weight loss. This information is important in determining how to optimize walking programs for obese individuals in order to minimize attrition and maximize compliance.

However, there is no concrete evidence that documents the time dependent effects of these walking programs on gait mechanics. While previous studies on the biomechanics of walking in obese individuals investigated differences in lower limb joint mechanics, these studies only examined individuals in a rested state (Lai, 2008; Spyropoulos, 1991). In addition, these studies categorized individuals based on BMI, but did not take into account their fitness levels. Uncertainty still remains regarding the exact nature of obese individuals' musculoskeletal problems during physical activity, and when, in the course of physical activity, these problems manifest. While physical activity is a necessary component for healthy weight management and weight loss, the relationship between current recommendations for activity and the subsequent biomechanical impact, is not sufficiently understood. Given the propensity for obese individuals to drop out of walking programs and the evidence that lower limb joint discomfort may be a contributing factor, it is important to investigate the short-term, time-dependent, changes in the gait pattern that may contribute to attrition and non-compliance.

One of the factors that remains unanswered is what happens over time during a walking program**.** The effect of continuous walking on biomechanics has not been documented in the past, but studies investigating running and drop-landing types of activities in adults have reported that repeated continuous trials lead to an increase in ground impact forces (Christina, 2001). Similarly, over a 30-minute continuous walking period, there may be less shock absorption, an increase in loading rate, and a disproportionate increase in ground reaction force peaks over time (Parijat and Lockhart, 2008). This increase in peak vertical ground reaction force during the weight acceptance phase of gait, may lead to increased biomechanical stress on the joints. Furthermore, obese individuals have more fat mass and muscle mass than their normal weight counterparts (Thibault, 2011), which may impact the coping strategies.

The purpose of this study is to assess the biomechanical gait changes in obese and normal weight adult subjects following a 30 minute walking session. It is hypothesized that the hip and knee adduction and extensor moments, which are the primary modulators of frontal and sagittal plane load distribution, will increase more in obese individuals, as compared to normal weight subjects following a 30-minute walking period, resulting in more stress across the hip and the knee joints.

Methods

Ten obese female subjects 38.3 ± 5.2 years, body mass index (BMI) 37.4 ± 5.4 kg/m^2 and ten normal weight control subjects 38.1 ± 4.5 years BMI 22.6 ± 2.3 kg/m^2

volunteered for the study that was approved by the University of Iowa Institutional Review Board. Height, weight, waist circumference, hip circumference, resting heart rate and blood oxygen saturation (SpO_2) were recorded. Waist circumference was measured at the level of the right iliac crest, with a Gulick II plus (Gulick II measuring tape; Country Technology Inc., Gays Mills, WI) tape measure. The subjects completed Jackson Non-Exercise test and PAR-Q questionnaire to estimate their fitness levels (see appendix). The Jackson non-exercise test has been validated in different sample populations. The standard error of estimate (SEE) for different regression models was 1.45 to 1.57 metabolic equivalents (MET's) and demonstrated a high level of cross-validity ($0.72 < R < 0.80$). The results indicate that fitness can be assessed from a non-exercise model which includes self-reported physical activity (Jurca, 2005).

Triads of infrared emitting diodes (IREDs) were placed on the pelvis and trunk, and bilaterally on the thighs, legs, and feet. Markers were affixed to the lateral aspect of the foot, to the shaft of the tibia, and to the lateral aspect of the thigh. Femoral epicondyle motion was tracked by two markers mounted on a custom femoral tracking device (Houck, 2000). Pelvic markers were affixed on the sacrum using a 5 cm extension. A similar extension was placed on the lower cervical vertebrae, to track the trunk segment.

A link-based model was generated for tracking each segment. Anatomical landmarks were digitized, relative to segment local coordinate systems, with the subject standing in a neutral position, to create an anatomical model. Segment principal axes were defined based on a single experienced clinician palpating and digitizing the following bony landmarks: Pelvis anterior and posterior superior iliac spines; Trunk or Head Arm Trunk (HAT): C-7 and L-1 vertebrae and glenohumeral joints; Thigh: hip joint center, lateral and medial condyles; Shank: lateral and medial condyles and malleoli; Foot: posterior heel, metatarsal head, and second toe (Segal, 2009). The hip joint center was estimated using the functional method based on the isolated motion of the femur relative to a stable pelvis during separate movement trials (Schwartz, 2005). The thigh segment was defined by the hip joint and medial and lateral condyles (Houck, 2000).

Gait data were collected using an Optotrak motion analysis system (Model 3020, Northern Digital Inc., Waterloo, Ontario, Canada) operating at 60 Hz. Kinematic data were filtered at 6Hz, using a zero phase lag fourth-order Butterworth low pass filter. Kinetic data were obtained using a Kistler force plate (Kistler Instruments, Inc., Amherst, NY). The force plate data were sampled at 300 Hz, and were filtered at 6 Hz, thus providing ground reaction forces. Visual 3D software (C-Motion Inc. Kingston, Ontario) was used to perform link-segment calculations.

Subject's walking speed and stride length was measured by using a GAITRite mat. GAITRite mat is a valid tool for measuring both averaged and individual step parameters of gait and has been shown to have excellent reliability (Menz, 2004). When compared with motion analysis system (Vicon), spatio-temporal variables like walking speed, cadence and step length from GAITRite showed an excellent level of agreement with intra-class correlation coefficients (ICC's) between 0.92 and 0.99 and repeatability coefficients (RC's) between 1.0% and 5.9% of mean values (Weber, 2005). The gait evaluation was conducted along an 8 m walkway. Start positions were set by putting tape marks on the floor, so that subjects naturally contact individual force plates with each foot while continuously walking back and forth at their predetermined self-selected speed. The treadmill session included total walking for a total 30 minutes out of which:

The Ebelling protocol (4 minutes at 0% incline and 4 minutes at 5% incline), was followed for the first 8 minutes (Ebelling, 1992). On the treadmill, target heart rate was set between 65-85% of estimated maximum (208 - 0.7*age), SpO_2 was maintained above 93 and perceived exertion below 17 on the Borg scale. HR, SPO_2 and Borg exertion scale were recorded every two minutes. A post-treadmill gait analysis was conducted immediately after the treadmill walking in the fatigued state at the same speed as the initial pre-treadmill test.

Data Analysis

Data were processed using Visual 3D software (C-Motion). Peak hip and knee adduction and extensor moments, normalized to body mass were calculated for five gait cycles during both pre and post treadmill trials. The moments were corrected for speed using the equations from previous literature (Rutherford, 2009; Goldberg, 2013).

Statistical Analysis

The side with greater moments for post treadmill trials was selected as the side of interest and used for further analysis. Descriptive statistics in the form of means and standard deviation were estimated. A two-way repeated measures ANOVA model (2x2; joint moments, pre vs. post treadmill) with group (obese vs. normal weight) as a between subject factor was fitted. Due to the small sample size, additional analysis was conducted using change in moments from pre to post treadmill walking as the main outcome measure. Pearson correlation coefficients were used to identify factors associated with changes in moments. Linear regression models were fitted using BMI and VO_2 max as predictor variables. The alpha level was set at 0.05, and SPSS (Version 19) was used for statistical analysis with p-value < 0.05 considered significant. All results are presented as mean ± standard deviation

Results

All twenty subjects (ten obese and ten normal weight) recruited for the study completed the study. Obese subjects walked at an average self-selected speed 1.36 m/s (3.06 miles/hour) and normal weight subjects walked at 1.47 m/s (3.3 miles/hour) on the treadmill. Over ground speed was not different between pre 1.30 m/s and post 1.31 m/s for obese (p= 0.9) and normal weight pre 1.43 m/s, post 1.44 m/s (p= 0.8) subjects. The normal weight subjects had higher speed as compared to obese subjects at pre (p= 0.03) and post treadmill trials (p= 0.04). Estimated $V0_2$ max was higher for normal weight subjects (35.8 ± 3.3 ml/min/kg) than obese subjects (32.1 ± 3.3 ml/min/kg) (p-value = 0.02). Mean fitness level score on the Jackson non-exercise test was 1.6 for obese and 2.3 for normal weight subjects. Subject characteristics are described in Appendix (Table A3-1).

Moments result:

No significant interaction effects were seen between obese and normal weight groups. Knee extensor moments showed a significant main effect for time from pre to post treadmill walking (p= 0.04) (Fig 3-1).

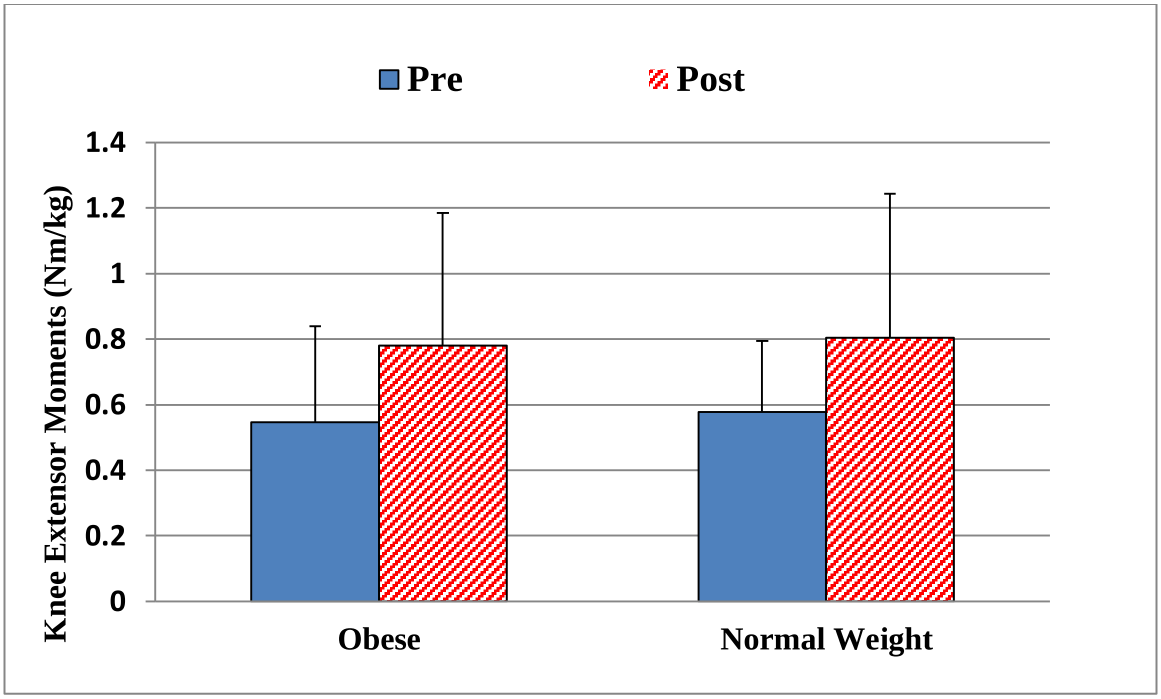

Figure 3-1: The mean and standard deviation peak knee extensor moments during weight acceptance for pre (blue) and post (red) 30 minute treadmill walking.

The second peak knee extensor moments, associated with controlling the knee prior to the swing phase also increased in both obese subjects pre-treadmill 0.50 ± 0.30 Nm/kg to post-treadmill 1.09 ± 0.50 Nm/kg and normal weight subjects 0.40 ± 0.30 Nm/kg to 0.90 ± 0.40 Nm/kg .

Hip extensor moments on the other hand decreased and also showed a time effect (p= 0.02), however no interaction effect was seen between obese and normal weight groups (Fig 3-2).

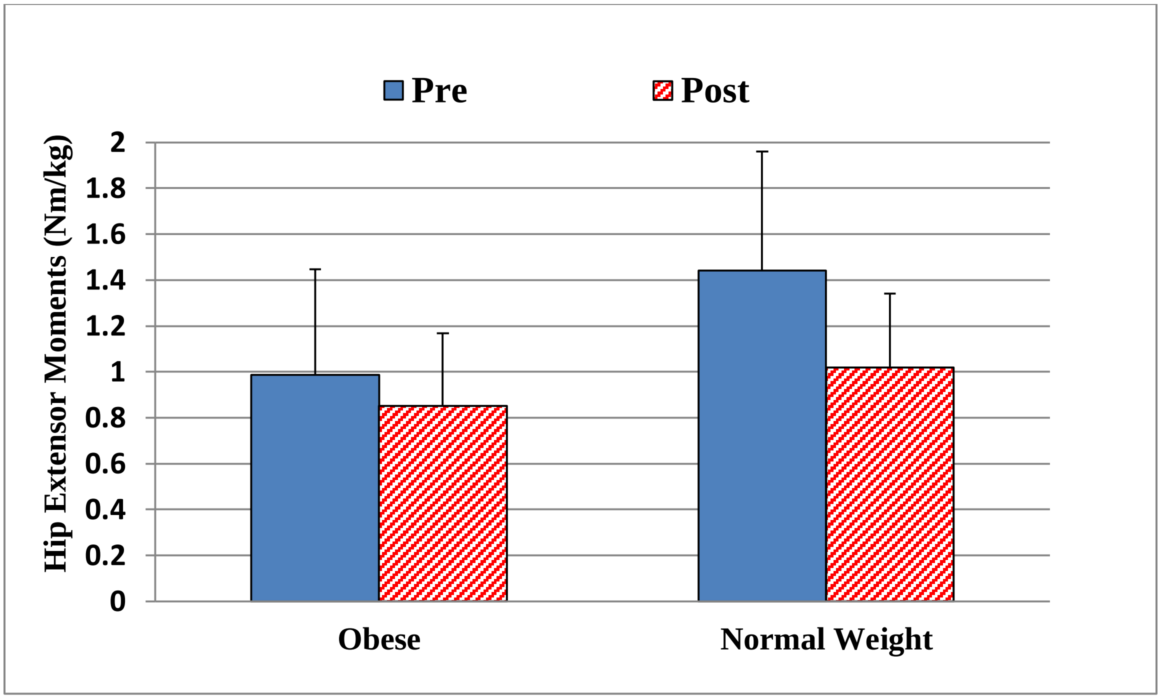

Figure 3-2: The mean and standard deviation peak hip extensor moments during weight acceptance for pre (blue) and post (red) 30 minute treadmill walking.

The knee adduction moments, did not change in obese subjects pre-treadmill 0.37 ± 0.13 Nm/kg to post-treadmill 0.40 ± 0.16 Nm/kg or normal subjects 0.38± 0.13 Nm/kg to 0.43 ± 0.2 Nm/kg. Similarly, hip adduction moments, did not change in obese subjects pre-treadmill 0.77 ± 0.19 Nm/kg to post-treadmill 0.97 ± 0.24 Nm/kg and normal subjects 0.82 ± 0.19 Nm/kg to 0.89 ± 0.24 Nm/kg.

Linear regression models were used to assess the effect of BMI and anthropometric measures like hip and waist circumference on change in peak moments from pre to post treadmill.

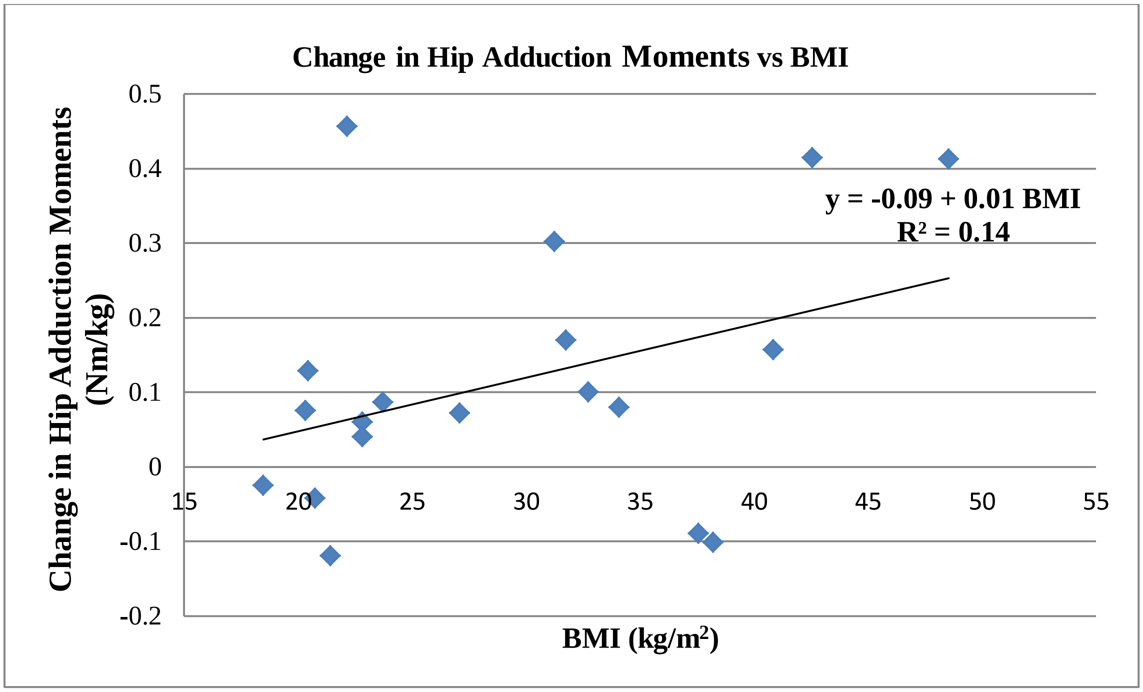

Figure 3-3: The weak relationship between BMI and change in hip adduction moments for 20 (10 obese and 10 normal) subjects.

Knee adduction moments showed no relationship between change in moments and BMI (r-square= 0.04).Similar trends were seen for extensor moments. BMI showed a weak relationship with changes in hip extensor moments (R^2= 0.13) and no relationship was seen between changes in knee extensor moments and BMI (R^2= 0.01).

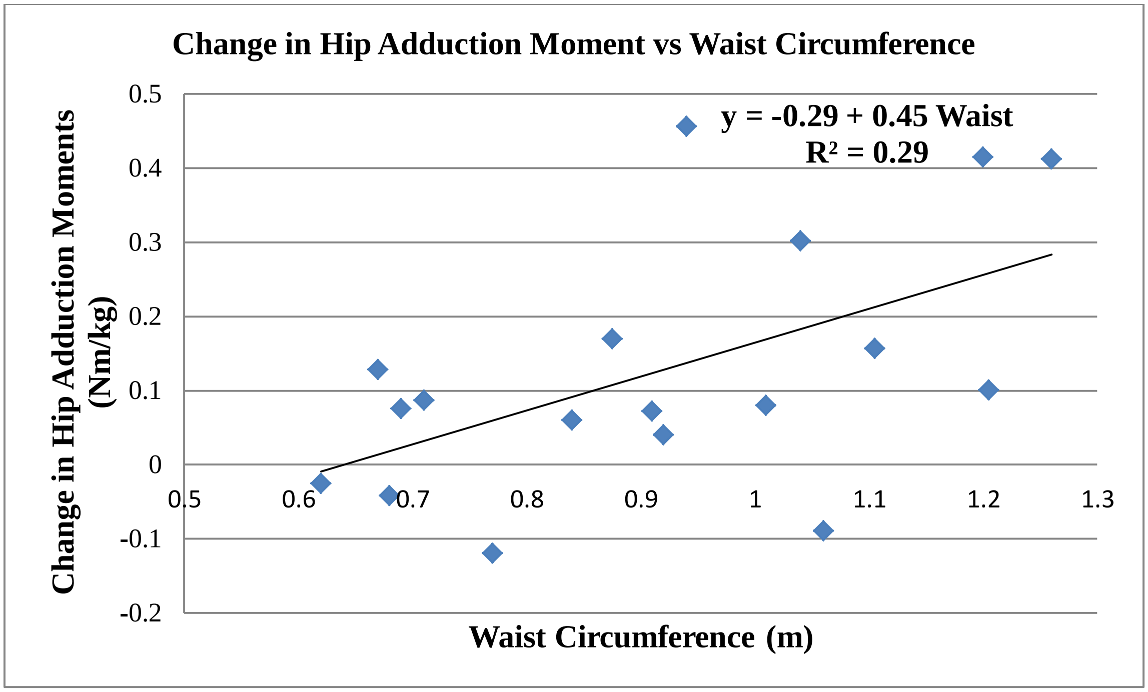

Figure 3-4: The relationship between waist circumference and change in hip adduction moments for 20 (10 obese and 10 normal) subjects.

Similar trends were seen for hip circumference and change in moments and waist circumference (R^2= 0.26).In addition to waist circumference, waist to hip ratio is also used to characterize adiposity. Table 3-1 shows the correlation between all anthropometric measures and change in hip and knee moments.

	Change in Moments			
Anthropometric Measures	Hip Adductor	Hip Extensor	Knee Adductor	Knee Extensor
Waist Circumference (m)	0.40	0.29	-0.10	0.09
Hip Circumference	0.39	0.32	-0.14	-0.06
Waist to Hip ratio	0.17	0.03	0.03	0.32

Table 3-1: Pearson correlation (r) for anthropometric measures and hip and knee moments.

Linear regression models were also used to assess the effect of fitness, assessed as VO_2 max from the Ebelling protocol, on the hip and knee moments. Change in peak moments from pre to post treadmill was used as the outcome measure.

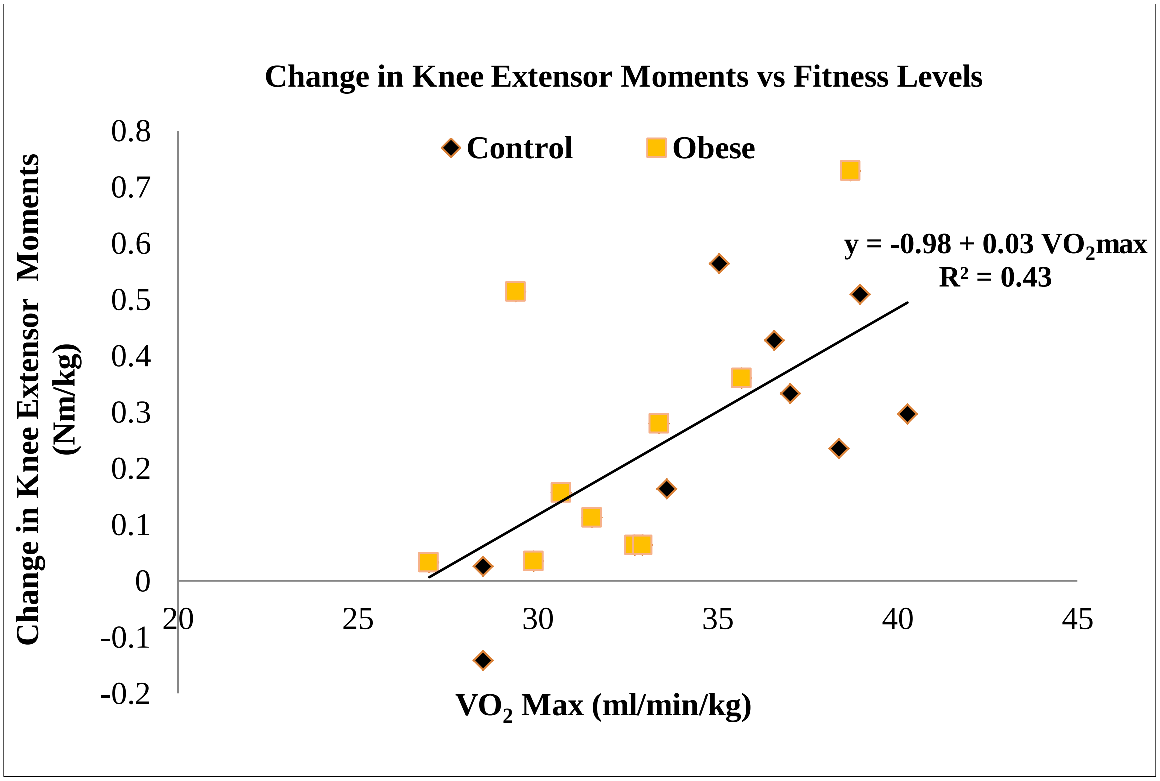

Figure 3-5: Change in knee extensor moments showed a good relationship with $V0_2$ max.

Knee adduction moments also had a moderate relationship with VO_2 max (r^2=0.33). On the other hand, change in the hip extensor moments showed an inverse relationship with VO_2 max (r^2=0.46) indicating that fitness might be an important variable to be considered in this population.

Discussion

The purpose of this study was to assess biomechanical gait changes in obese and normal weight adult women following a 30 minute walking session. No significant group effects were identified between obese and normal weight groups. There was an increase in the extensor moments at the knee after the treadmill walking session. The hip and knee adduction moments did not show any increase. Changes in hip adduction moments showed a weak relationship with BMI and a moderate relationship with waist circumference. Knee extensor and adductor moments, on the other hand, showed good to moderate relationships with $V0_2$ max, but not BMI or waist circumference. This indicated that fitness, and not BMI, may be the important factor in judging the implications of exercise on joint mechanics. Clinicians can use these results to help judge how changing biomechanics may affect compliance with walking programs.

It was hypothesized that hip and knee extensor moments would increase after 30-minute treadmill walking in obese subjects. There was an increase in knee extensor moments, but the increase was seen in both obese and normal weight subjects. Recent work has looked at the influence of load carriage and fatigue on mechanical loading during walking in healthy normal weight males (Huan, 2010). The study reported that the knee extensor moments increase as a partial mechanism to absorb the increased ground impact forces during loaded walking. The increase in the knee extensor moments in the current study may also be attributed to an increase in the ground reaction force that may occur over time, during the acceptance phase of the gait cycle, although the ground force data were not analyzed as part of this study. An increase in the extensor moments at the knee for both obese and control subjects, points to the possibility of increased muscle work and knee joint stress which might affect the ability of individuals to maintain a walking program.

Another possible explanation for the increase in knee extensor moments is that the hip extensor moments decreased after the 30-minute walking. Winter et al. demonstrated a trade-off mechanism between hip and knee extensor moments, where knee extensor moments increases are associated with hip extensor moments decreases and vice versa (Winter, 1989). This trade-off mechanism, which recognizes variable lower limb strategies for supporting the trunk during single limb stance, may explain the increase in knee extensor moments as the hip extensor moments decreased (Figure 3-6). In addition, hip flexor moments also increased from pre to post treadmill walking in our study and showed a strong positive relationship with the corresponding second knee extensor moment (r-square =0.81) following similar trends as hip extensor moments.

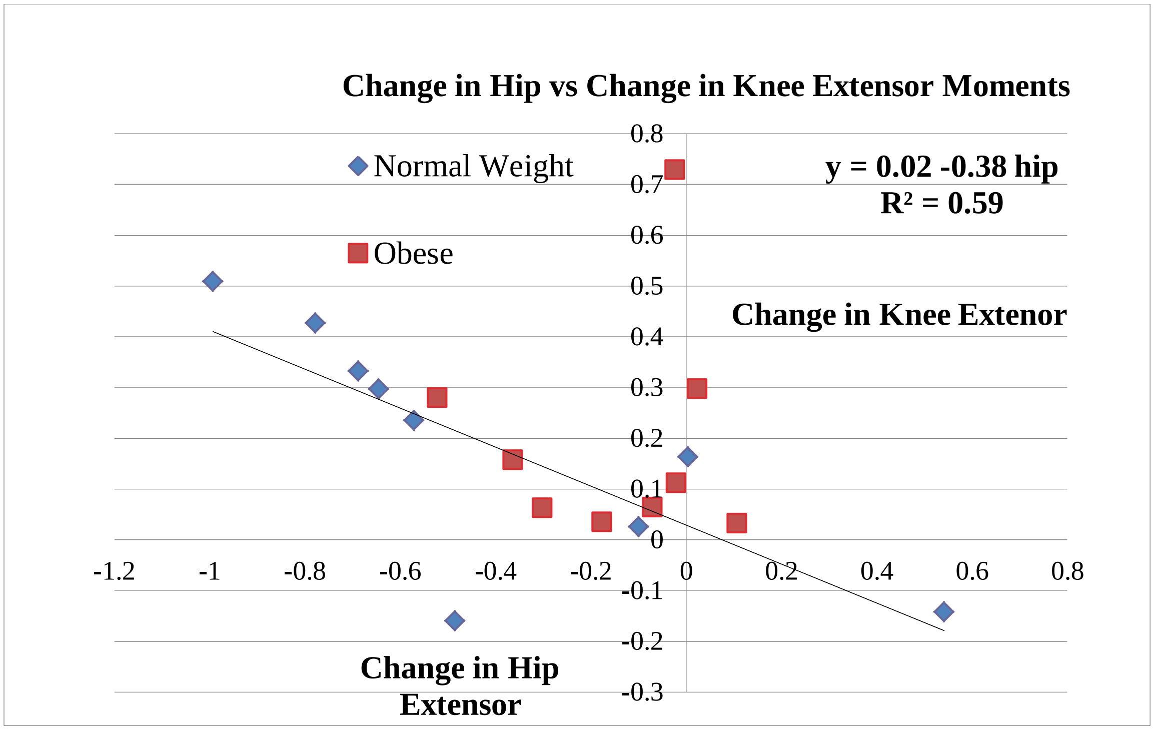

Figure 3-6: Knee Extensor Moments during weight acceptance increased, but hip extensor moments decreased for most subjects demonstrating a trade-off mechanism between hip and knee moments shown in the previous studies.

A few studies have quantified frontal plane moments in healthy obese individuals, particularly at the knee, due to the relationship between knee adduction and medial compartment loading (Andriacchi, 2004; Shelburn, 2008). Higher non-normalized peak knee adduction (Nm) moments have been reported in obese adults as compared to non-obese adults. Our study reported similar normalized hip and knee adduction moments for obese and normal weight subjects at baseline, before the 30-minute treadmill walking. These results are consistent with recent studies on three dimensional gait in obese individuals (Lai, 2008; Segal, 2009). In the Lai, et al. study no differences were reported for hip and knee adduction moments in obese and normal weight subjects (Lai, 2008). Segal, et al., 2009 also reported that knee adduction moments were not different between normal weight and obese individuals with varying weight distributions.

It was hypothesized that hip and knee adduction moments would also increase over time in obese individuals, but the moments did not change significantly. Another recent study looked at biomechanical changes at the knee after lower limb fatigue in healthy young women (Longpré, 2013). The magnitude of knee adduction moments normalized to body mass was identical to our study (0.35 Nm/kg) and showed that fatigue did not alter the first or second peak knee adduction moments, or peak knee flexion angles. Although it has been shown that obesity may increase the risk of developing medial compartment OA, results from this study indicate that there are no differences in obese and normal weight normalized moments and that these moments did not change over time.

This study hypothesized an increase in moments in obese subjects, but not in normal weight subjects. The results showed that normal weight subjects experienced changes in moments similar to those of obese subjects. It was further hypothesized that the measure of obesity in this study, BMI, may not provide the most precise association. In the current study, the knee adduction moments did not show any relationship with waist to hip ratio (Table 3-1), which is in agreement with previous work on knee adduction moments and weight distribution in obese individuals (Segal, 2009). Segal et al. showed that increased weight, due to obesity, is responsible for increased medial compartment loading, rather than thigh or abdominal fat distribution. However, the relationship between change in knee extensor moments and waist to hip ratio was stronger than that with BMI. Similarly, waist circumference had stronger relationship, than BMI, with changes in some lower limb moments (e.g. hip adduction moments). These findings suggest that BMI alone may not be the best way to characterize obesity, and waist circumference and waist to hip ratio should be taken into consideration while evaluating obese populations. Other studies have shown that measures of absolute and relative waist size, such as waist circumference and waist to hip-ratios, have been suggested as more relevant clinical measures of adiposity (Lemieux, 1996). It has also been shown that a given BMI may not correspond to same degree of body fat across different populations (Freedman, 2009). This study thus expands the discussion of the relationship between moments and anthropometric measures and suggests the need for further research.

Recognizing that fitness is an important variable that could affect biomechanical changes over time, an effort was made to recruit both obese and normal weight subjects with similar fitness levels by having the subjects complete the Jackson non-exercise test to judge their own fitness levels.. Normal weight subjects reported higher scores on the Jackson test as compared to obese subjects, which corresponded with higher VO_2 max estimated in normal weight subjects by the Ebelling protocol. There was a positive association between VO_2 max and knee sagittal and frontal moments, indicating that fitness, not BMI, may be the more important factor in judging the implications of exercise on knee joint mechanics. One possible explanation for these findings could be that the lesser fit subjects already have higher moments and do not have the capacity, unlike fit subjects, to change their gait pattern to counteract fatigue. In addition, knee extensor moments showed an inverse relationship with hip extensor moments which points to the possibility of different coping strategies based on fitness levels during continuous walking. These strategies might affect the ability of lesser fit individuals to maintain a walking program.

When normalized for body weight, obese and normal weight subjects experienced similar increases in knee moments after 30 minutes of walking. However, the magnitude of non-normalized moments was significantly higher for obese subjects (82.5 ± 32.2 Nm) than their normal weight (49.2 ± 26.6 Nm) counterparts (p= 0.001). The joint stress due to higher moments can be experienced as discomfort and pain. Although pain was not reported in this study, joint pain has been reported as a barrier to the obese individuals participating in walking programs or physical activity. Furthermore, pain has been shown to increase with obesity levels (Heo, 2010). Given the propensity for obese individuals to drop out of walking programs and the evidence that lower limb joint discomfort may be a contributing factor, it is important to investigate the short-term, time-dependent, changes in the gait pattern that may contribute to attrition and non-compliance.

The study had some limitations. Although an attempt was made to recruit subjects with similar fitness levels to ensure that fitness was not a confounding issue, normal weight subjects ended up with higher fitness levels than obese subjects. The fitness levels were still in the general range of moderate fitness levels in adult women. However, some promising relationships were seen in the current study by considering these differences in fitness levels, and future research should be designed with different high and low fitness groups to explore the effect of fitness as a variable. Another potential limitation was that kinetic data were collected using over-ground force plates before and after continuous 30 minute walking on the treadmill. Although the post treadmill walking trials were immediately collected and tape marks were used to make sure the subjects naturally hit the force plates to capture the fatigue state, studies have shown a change in gait pattern from treadmill to over-ground walking (Lee, 2007). While an instrumented treadmill was used, it gives only vertical ground reaction and sagittal plane moments. Frontal plane mechanics was one of the important outcome measures of the study necessitating complete three dimensional gait analyses. Future research should be performed on a fully-equipped three dimensional gait treadmill for continuous 30 minute walking and should also look at changes in gait over smaller increments of time.

Finally, normalized moments were used as the main outcome measure in this study; however it must be kept in mind that moments are an estimate of joint forces, rather than a direct measure of the joint forces. Though an estimate, the moments have been associated with joint pathology; for example, high knee adduction moments are highly correlated with risk for progressive knee OA. To conclude, kinetics at the hip and knee joints should be taken into consideration, in the context of fitness, to improve compliance to walking programs. The clinicians can use the results to help judge how changing biomechanics may affect compliance with walking programs.

CHAPTER IV
DO FITNESS AND FATIGUE AFFECT GAIT BIOMECHANICS IN OVERWEIGHT AND OBESE CHILDREN?

Introduction

The implications of childhood obesity are profound: from the health of the individual child to the overall cost on the healthcare system of the United States. The consequences of childhood obesity can be long lasting and considerable. Annual health care costs for an obese child are estimated to be $2,500 to $4,200 more than the annual health care costs for a normal weight child. The acute cost of treating children with obesity-related conditions has been estimated to be $127 million (NIH, 2000). These costs result from a variety of obesity-related conditions. Attention to childhood obesity is of particular importance, not only for the immediate impact obesity may have upon the health and developmental factors of the child, but also for the continued impact obesity may have upon quality of life and health into adulthood.

Obese adolescents are more likely to have musculoskeletal disorders than age-matched normal weight controls (Jannini, 2011). These disorders have been linked to aberrant lower limb mechanics, associated with functional and structural limitations, imposed by obesity (Wearing, 2006). Altered mechanical modulation of bone growth, due to children being overweight, can interfere with growth plate development resulting in boney deformities in the spine and long bones of the lower limbs, and has been linked to Blount's disease (Gushue, 2005; Villemure, 2009). In addition, evidence of cardio respiratory disease has been documented in overweight six year olds; and strong negative associations have been shown in overweight children between cardiorespiratory fitness and cardiovascular disease (Anderssen, 2007). Based on these observations it is not surprising that health-related quality of life and the subset of physical functioning is inversely related to weight status (Tsiros, 2011).

Defining the nature of obesity amongst children poses challenges. Body weight classifications in children have traditionally been based on generally accepted cut-points of BMI: overweight children in the 85th – 94th percentile and obese children in the 95th – 99th percentile, based on 2000 CDC growth charts. Other methods have also been proposed to define obesity: weight >20% above predicted weight for height (Cooper, 1990), triceps skinfold thickness >85th percentile of children of the same age and sex (Elliot, 1999), and percentage body fat >25% for boys and >30% for girls (Williams, 1994). While BMI is a general measure of obesity, it does not distinguish differences in adiposity levels in children of the same BMI (Freedman, 2004). As the level of adiposity may influence the degree to which heavy children can control excess body mass during physical activities, measuring adiposity could provide important information to address potential gait-related issues in obese children.

Measures of strength can provide additional clarifying data in the prediction of biomechanical loads in obese children in non-fatigued and fatigued states. Little is known about how muscular strength influences fatigue in obese children. However, lower limb strength may attenuate ground reaction forces, influencing mechanical loads in both fatigued and unfatigued conditions. Lower limb strength, as measured by jumping tasks, has been shown to be associated with improved performance on functional tasks in obese children (Riddford-Harland, 2006). However, lower limb strength does not necessarily correlate with upper limb strength (Ducher, 2009). Upper limb strength, as measured by handgrip strength, has been shown to have a strong positive correlation with predicted

VO_2 max, and weak negative correlation with fat mass (Wallymahmed, 2007). Given muscle strength and adiposity are not strongly related, (Cawthon, 2011) independent measures of strength of the upper and lower limbs may further improve models to predict gait biomechanical loads.

Increases in childhood obesity, while certainly emerging from a variety of causes, are clearly related to a lack of physical activity. While the American Heart Association recommends that children should participate in 60 minutes of moderate – intense exercise each day, recent studies have reported that obese children achieve only a fraction of this level of physical activity (Troiano, 2008). As in adults, children may find it difficult to perform sufficient physical activity for a variety of reasons, some of which may be biomechanical in nature. If so, the discomfort associated with biomechanical stresses observed in obese individuals may also be contributing to the positive feedback loop that is at the core of the Reverse Causation hypothesis. The Reverse Causation hypothesis suggests that physical inactivity is not only a result of obesity, but may also hinder further physical activity, contributing to increased weight gain, and increased susceptibility to disease and pathology (Must, 2005; Kwon, 2011). This cycle can be observed as some obese children tolerate physical activity better than some of their counterparts: fitter children being more likely to persist in physical activity than their less fit counterparts with similar BMIs.

Lower levels of physical activity have been related to lower levels of cardiorespiratory fitness (Vancampfort, 2010) and it has been established that obesity is inversely related to VO_2 max. Clinically, it may be difficult to get a 'complete' evaluation of gait in some overweight and obese children because the motor patterns that contribute to higher mechanical loads and resulting pain or injury, may not be present when they are briefly examined during a typical clinic visit. It can be argued that fatigue exaggerates gait abnormalities in overweight and obese children that might not be observed in the unfatigued state. While empirical clinical observation suggests fitness and fatigue in obese and overweight children are important issues that affect gait, these impressions have not been confirmed by scientific studies using biomechanical measures to assess gait.

The purpose of this project is to determine how cardiorespiratory fitness and fatigue influence gait biomechanics in overweight and obese children (aged 8-11 years). The unique aspect of this project is that it examines cardiorespiratory fitness (an attribute) and cardiorespiratory fatigue (a temporary state), in overweight and obese children and their association with physical performance (gait biomechanics). It was hypothesized that:

1. Gait biomechanics, as measured by lower limb moments, will be inversely related to cardiorespiratory fitness in overweight and obese children, in a non-fatigued state.
2. Introduction of cardiorespiratory fatigue in overweight and obese children will be associated with an increase in lower limb moments as compared to the non-fatigued condition. .
3. The difference in lower limb moments between non-fatigued and fatigued states will not be related to the level of cardiorespiratory fitness.

The secondary purpose of this project is to explore the effect of including measures of adiposity and muscular strength in predicting the relationship between cardiorespiratory fitness and gait biomechanics.

4. It was hypothesized that including measures of adiposity and muscular strength will improve models to predict the relationship between cardiorespiratory fitness and lower limb moments in overweight and obese children, aged 8-11 years in the fatigue state and non-fatigue state.

Methods

Thirty subjects (15 females, 15 males) age 8-11 years (9.7+/- 0.9), with BMI measures above the 85^{th} percentile, volunteered for the study. Based on pilot data 30 subjects were needed to detect significant correlations and differences with 80% power. Children with any musculoskeletal injuries, neurological syndromes and cardiopulmonary disease were excluded from the study. The study was approved by the University of Iowa Institutional Review Board.

Data Collection

The data collection was divided into two visits**.**

During the first visit: Height, waist circumference; hip circumference and leg length were recorded. Waist circumference was measured at the level of the right iliac crest and hip circumference was measured at the widest part of the hip with a tape measure (Gulick II, Country Technology Inc., Gays Mills, WI).

Adiposity**,** percent body fat, was estimated by air displacement plethysmography (Bod Pod). A standard two-point calibration was performed using an empty chamber and a known volume. Children were asked to wear tightly fitting bathing suits and a swim cap

(Dempster, 1995). Subjects practiced the breathing protocol (huffing) before sitting in the Bod Pod device. Subjects were instructed to remain still inside the Bod Pod and breathe into a mouth tube using a standard protocol. Percent body fat and lean mass were estimated based on measured thoracic volume using Lohmann equations (Lohmann, 1992). The Bod Pod has been shown to have good within-day and between-days reliability. Test-retest reliability for assessing %fat using the Bod Pod was shown to be 0.994 (Tseh, 2010). In addition, percentage fat estimation from Bod Pod has been validated in overweight and obese individuals by comparing the results to underwater weighing (Ginde, 2005).

Cardiorespiratory fitness was accessed in two ways (Nemeth and Pacer protocols) in order to establish the validity of the testing protocol. During the first visit, VO_2max was estimated using the Nemeth submaximal treadmill walking protocol (Nemeth, 2009) which has been shown to be valid and reliable in 130 overweight and obese children (Nemeth, 2009). The mean square error was 241.06 with the predicted VO_2max within 10% of the observed value in 67% of subjects. For this model, the observed VO_2max was predicted with R^2 =0.75 and an adjusted R^2 =0.73, with a cross-validity coefficient of 0.85. Subjects walked on the treadmill at a brisk, but comfortable, velocity for four minutes on a level (0% grade) surface and four minutes at 5% grade. Heart rate and perceived exertion, Wong-Baker Face Pain Scale (see appendix), were monitored every two minutes. Heart rate, documented prior to walking (at rest) and at the end of the eight minutes, in combination with walking speed, was used to calculate an estimate $V0_2$ max to predict fitness level (see data analysis and appendix).

<u>During the second visit:</u> Right lower limb isometric strength was assessed, with the hip and knee at 90^{o} flexion, using a custom made device, similar to a leg press. Subjects were asked to push as hard as possible with their right foot on a load cell plate. Following a warm-up, three trials were recorded. The right lower limb was the dominant limb for 22 of the 29 subjects as determined by asking subjects which leg they used to kick a ball. Hand Grip was measured using a 100 kg/220 lb hand grip dynamometer (Flaghouse, Inc NJ). After adjusting for hand size, subject was asked to squeeze it as hard as possible while seated with their elbow flexed at 90^{o}. Three trials were recorded each for right and left hand.

Subject's walking speed and stride length were measured using a GAITRite mat. The GAITRite mat is a valid tool for measuring both averaged and individual step parameters of gait and has been shown to have excellent reliability (Menz, 2004). Following three practice trials, walking along an 8 m segment of a hallway, subjects were instructed to maintain a constant walking pace and walk three times back and forth on the gait mat. These data were used to determine walking speed and step length measures. In preparation for the gait data collection, subjects practiced hitting marks on the floor during walking and jogging, corresponding to the previously collected walking step length data, in order to make it more likely that they would naturally land on the force platforms during the gait trials.

Triads of infrared emitting diodes (IREDs) were placed on the pelvis and trunk, and bilaterally on the thighs, legs, and feet. Markers were affixed to the lateral aspect of the foot, to the shaft of the tibia, and to the lateral aspect of the thigh. Pelvic and trunk marker triads were attached to 5 cm extensions with base plates affixed over the sacrum and lower cervical vertebrae.

A link-based model was generated for tracking each segment. Anatomical landmarks were digitized, relative to segment local coordinate systems, with the subject standing in a neutral position, to create an anatomical model. Segment principal axes were defined by digitizing the following bony landmarks: Pelvis anterior and posterior superior iliac spines; Trunk: C-7 and L-1 vertebrae and glenohumeral joints; lateral and medial condyles; Shank: lateral and medial condyles and malleoli; Foot: posterior heel, 5^{th} metatarsal head, and second toe (Segal, 2009). The hip joint center was estimated using the functional method based on the isolated motion of the femur relative to a stable pelvis during separate movement trials (Schwartz, 2005). The thigh segment was defined by the hip joint and medial and lateral condyles (Houck, 2000).

Kinematic data were collected using an Optotrak motion analysis system (Model 3020, Northern Digital Inc., Waterloo, Ontario, Canada) operating at 60 Hz. Kinematic data were filtered at 6Hz, using a zero phase lag, fourth-order, Butterworth low pass filter. Kinetic data were obtained using a Kistler force plate (Kistler Instruments, Inc., Amherst, NY). The force plate data were sampled at 300 Hz, and were filtered at 6 Hz. Visual 3D software (C-Motion Inc. Kingston, Ontario) was used to perform link-segment calculations. Subjects performed walking and jogging along an 8m walkway. Three trials of jogging and five trials of walking biomechanics were assessed twice: prior to and immediately following the fatigue activity.

Fatigue

The PACER protocol was used to fatigue subjects and estimate VO_2max (Mahar, 2011). The protocol required subjects to move between two markers, placed 15 m apart, within a progressively decreasing time interval. The allotted time to complete the 15 m distance decreased every minute. Running speed was 8.5 km/hr (5.3 miles/hr) for the first level and increased by 0.5 km/hr at each level. The activity was terminated if the subject failed to reach the 15 m marker in the allotted time twice or could no longer maintain the required speed. A research team member ran with the subjects to keep them motivated, in an attempt to get a maximum effort. The number of laps, heart rate, and perceived exertion (Wong-Baker Face Pain Scale, see appendix) at end of PACER were recorded. Cardiorespiratory fitness was estimated as VO_2max using the PACER estimation equations (see data analysis).

Data Analysis:

Visual 3D software (C-Motion) was used for processing lower limb kinematic and kinetic data. Net joint moments were normalized to body mass. Peak hip and knee moments were calculated for each of the three jogging and five walking gait cycles during both data collections: pre/post fatigue. As the first and last trial for jogging and walking were not different, average values for three jogging and five walking trials were used for further analysis. As the magnitude of moments is affected by the speed (Shultz, 2010), the moments were corrected for speed using equations from the previous literature (Rutherford, 2009). The primary focus of the analysis was Adduction (in the frontal plane) and Extensor (in the sagittal plane) moments.

Cardiorespiratory fitness was estimated by the Nemeth Protocol as follows: VO_2max = -1772.81 + 318.64 ×Sex (F= 0, M= 1) + 18.34 × Weight (kg) + 24.45 × Height (cm) - 8.74 × 4minHR - 0.15 Weight (kg) × HR difference + 4.41 × Speed (mph) × HR difference. Cardiorespiratory fitness was also estimated using the PACER estimation equations: Quadratic Model VO_2max = 41.76 + (0.49 × PACER laps) – (0.01 × PACER squared) – (0.61 × BMI) + (0.34 × gender × age). There was a strong association between two methods of $V0_2$ max estimation and PACER protocol was used for further analysis.

Strength Measurements:

The right hand and left grip force were not different p-value= 0.58 and highly correlated r= 0.89, so only right hand grip was used for further analysis. Also, right leg strength was equally correlated to right hand grip force r= 0.52 and left grip force r= 0.53.

Statistical Analysis:

All results are presented as mean ± standard deviation. The right side was used as the side of interest for analysis. Pearson correlation coefficients between fitness and biomechanics measures were estimated for initial descriptive analysis to address Hypothesis 1. Graphical displays were used to show association between peak moments, corrected for speed, and fitness levels, as measured by estimated VO_2 max in a non-fatigue state.

The outcome measures to address hypothesis 2 were lower limb moments at non-fatigued and fatigued states. Moments, corrected for speed, were used for both walking and jogging trials. Repeated measures analysis of variance models were fitted to compare the pre and post moments for walking and jogging trials.

Moments, corrected for speed, were used to determine the association between cardiorespiratory fitness and change in moments to address hypothesis 3. Data were graphically displayed; correlation coefficients were estimated using general linear models. To address the secondary aim, for hypothesis 4, final models included moments, corrected to pre PACER speed, as dependent variables and cardiorespiratory fitness, adiposity (% body fat) and right lower limb strength, as the three independent (predictor) variables. A stepwise regression approach was used to define the best model.

Results

A total of 28 out of 30 children completed the study (15 boys, 13 girls). One subject did not meet the BMI criteria during the first visit and one other subject did not complete the first stage of the PACER and withdrew from the study. Subjects anthropometric characteristics were as follows: mean height 1.48 ± 0.91 m, mass 60.4 ± 17.2 kg. Mean BMI percentile was 96.1 ± 4.1 percentile (BMI percentile was not different for boys 96.3 ± 3.1 and girls 95.8 ± 4.7). Mean percentage body fat was 32.3 ± 7.6 % body mass (range: 16.2 - 46.7). Other subject characteristics are described in Appendix (Table A4-1).

The mean VO_2 max estimated using the Nemeth protocol, was: 35.3 ± 6.5 mL/min/kg (range: 24.13 to 49.1). During the PACER fatigue protocol, subjects completed an average of 17.5 ± 8.5 laps (range: 4-45). The average cardiorespiratory fitness, assessed by the PACER protocol was: 34.1 ± 6.0 mL/min/kg (range 22.6 to 46.5). All subjects reached a target heart rate of 170 beats per minute, mean 182 ± 12 beats per min (range 172-206) at the end of the PACER protocol. All subjects reported a score of 1 or more on perceived exertion Wong-Baker Face Pain Scale. Walking speed pre- PACER was 1.31 ± 0.14 m/sec and post PACER was 1.38 ± 0.11 m/sec. The mean time between the end of the PACER and start of the first jogging trial was 38 ± 12 sec.

Average right lower limb strength, normalized to body mass, was 7.54 ± 2.29 N/kg. Average right hand grip strength, normalized to body mass, was 2.68 ± 0.79 N/kg.

Effect of Fitness:

The first hypothesis explored the association between lower limb moments and fitness levels in the non-fatigue state. The peak hip and knee adduction moments showed moderate association with fitness levels prior to fatigue (Figure 4-1 and 4-2). Peak hip and knee extensor moments did not show any relationship with fitness levels. The R-square values for all lower limb moments are shown in Table 4-1.

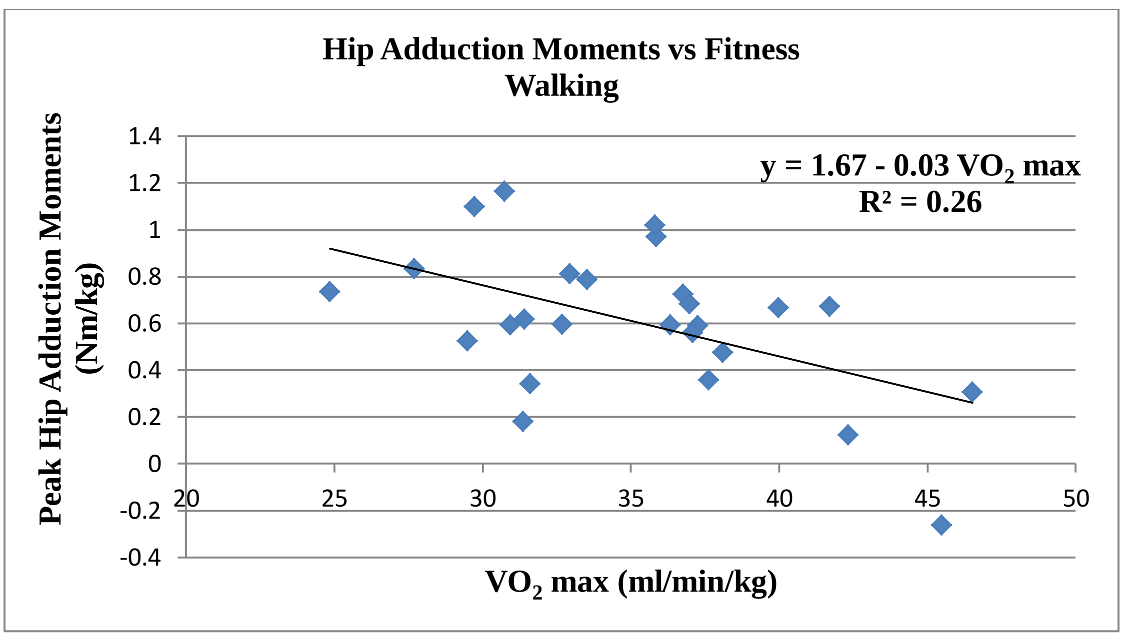

Figure 4-1: Shows inverse relationship between peak hip adduction moments and fitness levels, as measured by estimated VO_2 max in a non-fatigue state during walking for 28 subjects.

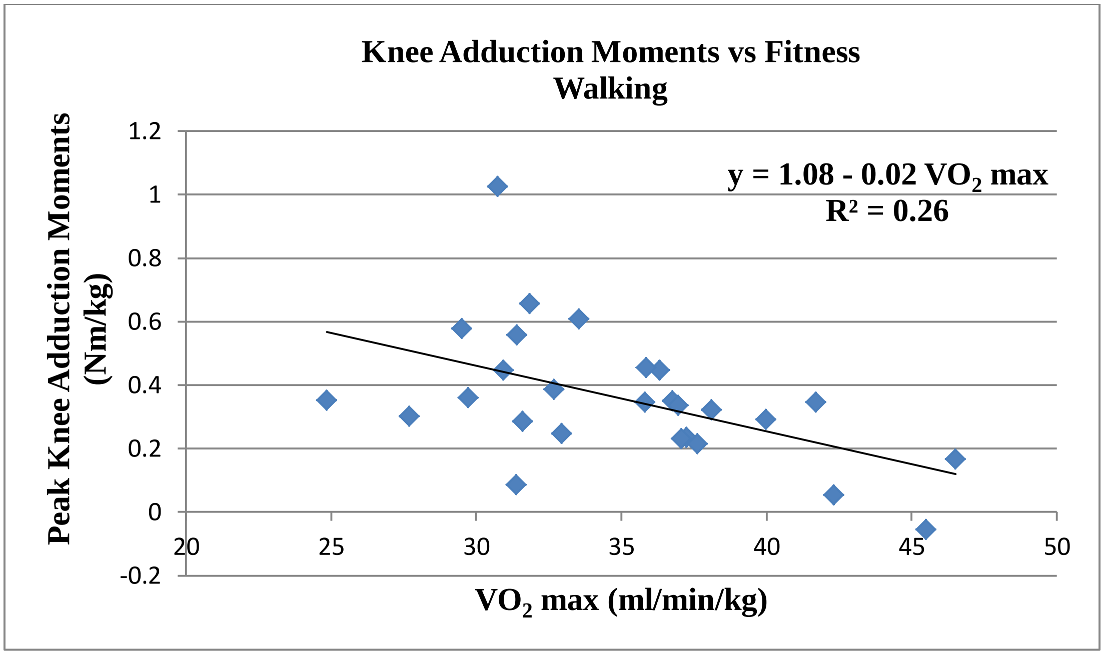

Figure 4-2: Shows inverse relationship between peak knee adduction moments and fitness levels, as measured by estimated VO_2 max in a non-fatigue state during walking for 28 subjects.

Walking	Moment	Ankle Extensor	Knee Adductor	Knee Extensor	Hip Adductor	Hip Extensor	Hip Flexor
R^2	Pre Pacer Walking	0.05	0.26	0.05	0.26	0.06	0.13

Table 4-1: Coefficients of determination (R-square values) for association between moments and $V0_2$ max for pre-PACER walking.

After the walking trials, subjects completed three jogging trials in a non-fatigued state. Knee adduction moments had the highest r-square value (Figure 4-3), with only weak associations seen between peak moments and fitness levels. Fitness levels did not show any association with peak hip and knee extensor moments and hip adduction moments. The r-square values for all the lower limb moments for jogging are shown in Table 4-2.

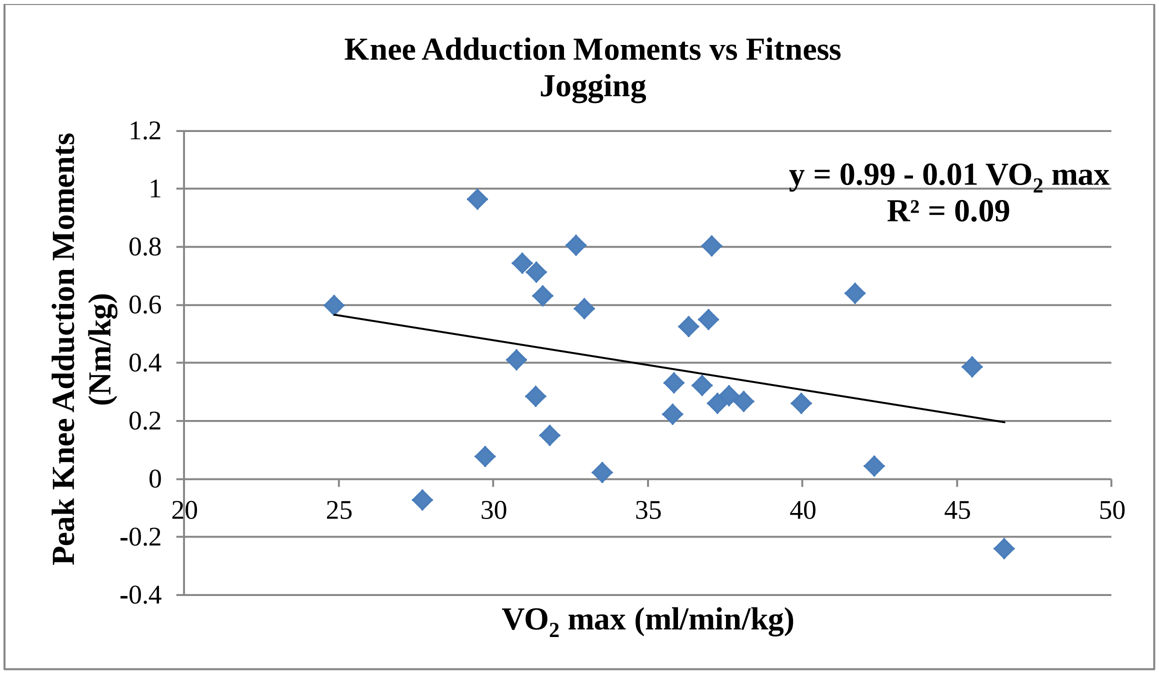

Figure 4-3: Shows inverse relationship between peak knee adduction moments and fitness levels, as measured by estimated VO_2 max in a non-fatigue state during jogging for 28 subjects.

Jogging	Moments	Ankle Extensor	Knee Adductor	Knee Extensor	Hip Adductor	Hip Extensor	Hip Flexor
R^2	Pre Pacer Jogging	0.07	0.09	0.003	0.04	0.01	0.005

Table 4-2: Coefficients of determination **(**R-square values)for association between moments and $V0_2$ max for pre-PACER jogging.

<u>Effect of Cardiorespiratory Fatigue:</u>

The second hypothesis looked at the effect of cardiorespiratory fatigue on lower limb moments. Fatigue, induced by the PACER protocol, resulted in an increase in the knee adduction moments (p= 0.01), knee extensor moments (p= 0.02) and hip extensor moments (p=0.01) as measured during post-fatigue walking trials (Figure 4-4). Hip adduction moments did not increase from pre to post-fatigue trials (p=0.66). The mean and standard deviations for all the lower limb moments pre to post-fatigue are shown in Table 4-3.

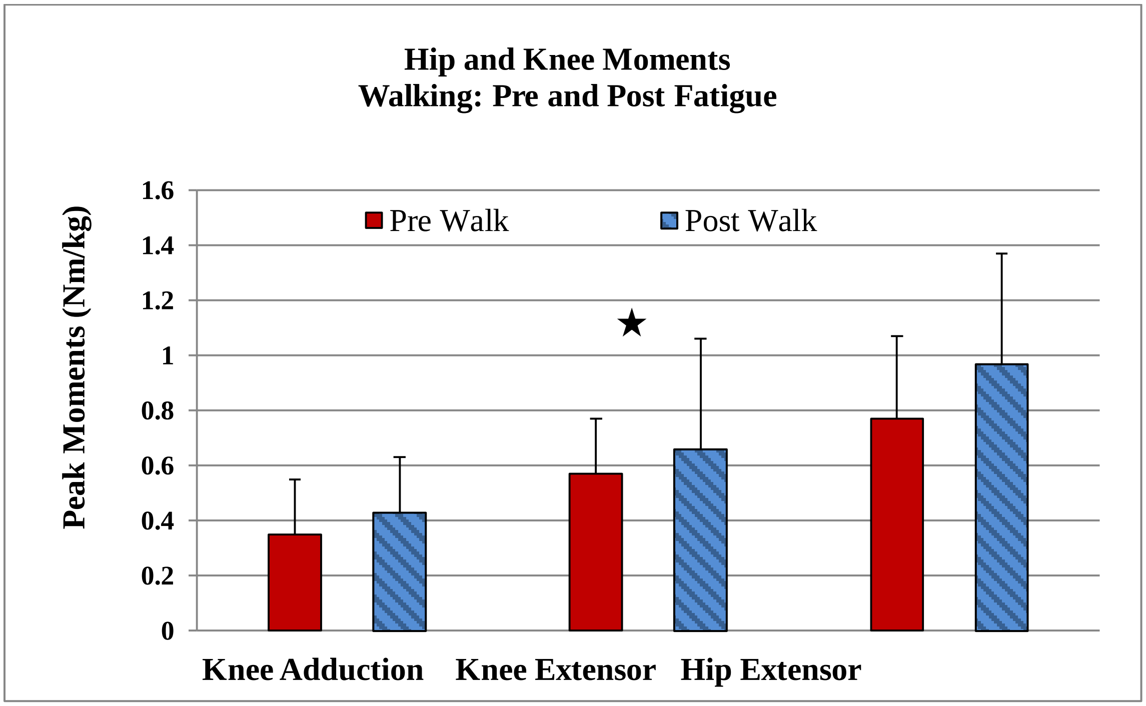

Figure 4-4: Mean and standard deviation comparing knee adduction, and hip and knee extensor moments for pre and post-fatigue walking trials. Following fatigue there was a significant (*) increase in the knee adduction moments, knee extensor moments and hip extensor moments.

Immediately after the completion of PACER protocol to achieve cardiorespiratory fatigue, subjects performed three jogging trials. There was an increase only in the hip extensor moments between pre- and post-fatigue jogging trials (p=0.003) (Figure 4-5). There was no increase in other moments (Table 4-3). The mean and standard deviation for all the lower limb moments comparing pre to post-fatigue for jogging trials are shown in Table 4-3.

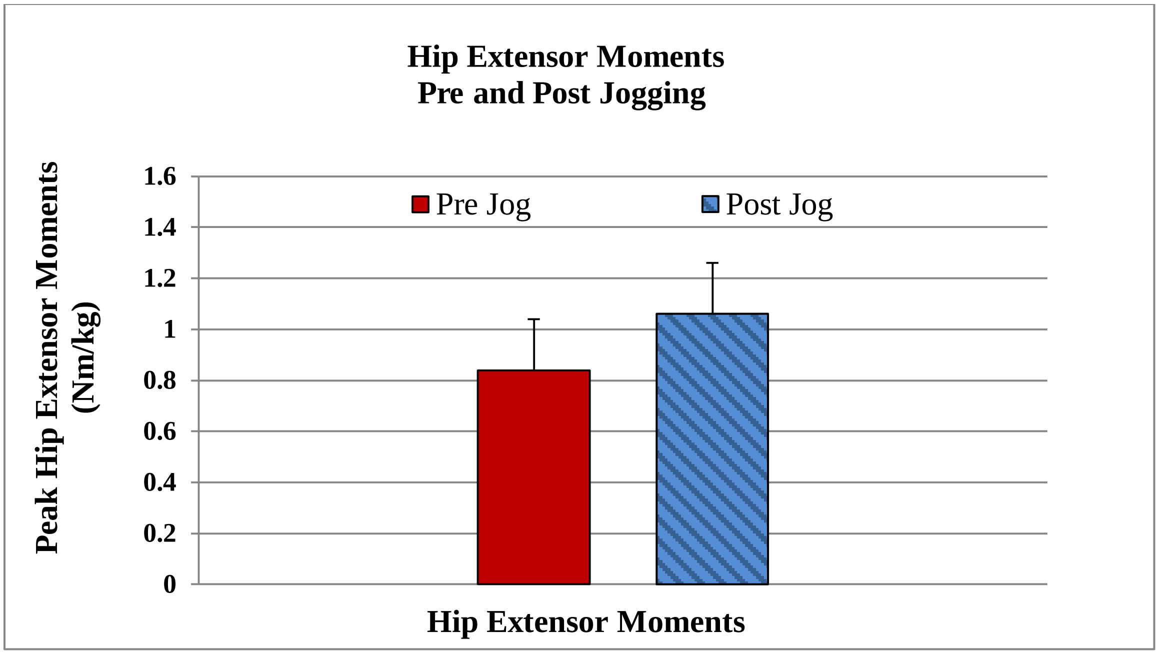

Figure 4-5: Mean and standard deviation for peak hip extensor moments for jogging trials. Only hip extensor moments showed an increase after the PACER protocol.

	WALKING				JOGGING			
Moments	**Hip Add**	**Hip Ex**	**Knee Add**	**Knee Ex**	**Hip Add**	**Hip Ex**	**Knee Add**	**Knee Ex**
Pre Pacer	0.60 (0.2)	0.77 (0.3)	0.35 (0.2)	0.57 (0.2)	0.98 (0.3)	0.84 (0.2)	0.39 (0.2)	1.24 (0.3)
Post Pacer	0.64 (0.3)	0.97 (0.4)	0.43 (0.2)	0.66 (0.2)	0.93 (0.4)	1.06 (0.4)	0.44 (0.2)	1.27 (0.4)
p-value	0.66	0.02	0.01	0.02	0.44	0.03	0.33	0.50

Table 4-3: Represents the mean and standard deviation of peak hip and knee moments for pre and post fatigue (PACER protocol). Significant p-values < 0.05 are highlighted.

Change in Moments after Fatigue:

The third hypothesis explored the association between changes in lower limb moments, from pre- to post-fatigue, and fitness levels. The change in moments for the walking trials did not show any strong association with fitness levels. The strongest relationship was seen for knee extensor moments (Figure 4-6). Peak knee adduction and peak hip adduction and extensor moments did not show any relationship with the fitness levels. The R-square values for all the lower limb moments are shown in Table 4-6.

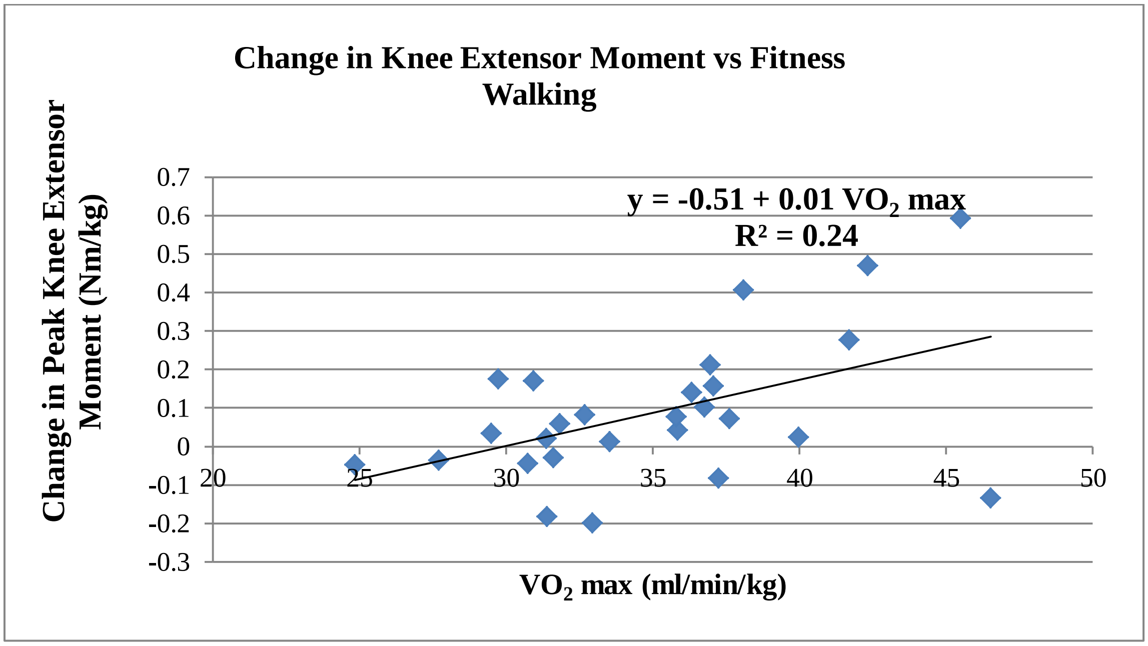

Figure 4-6: Shows association between change in peak knee extensor moments and fitness levels, as measured by estimated VO_2 max, during walking for 28 subjects.

Moments	Ankle Extensor	Knee Adductor	Knee Extensor	Hip Adductot	Hip Extensor	Hip Flexor
Change in moments	0.05	0.04	0.24	0.00	0.05	0.05

Table 4-4: R-Square values for association between change in moment and $V0_2$ max for walking.

The change in moments from pre- to post-fatigue jogging did not show any strong association with fitness levels. The strongest relationship was seen for hip extensor moments (Figure 4-7). Peak knee adduction and extensor moments and peak hip adduction moments did not show any relationship with the fitness levels. The R-square values for all the lower limb moments are shown in Table 4-5.

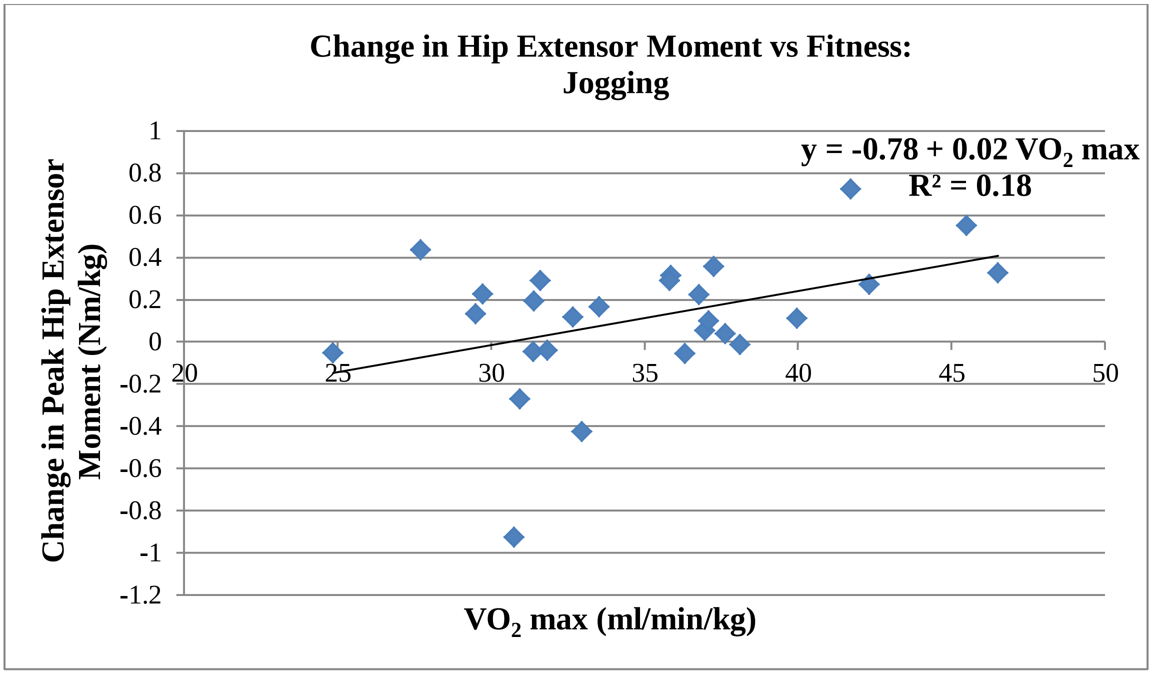

Figure 4-7: Shows association between change in peak hip extensor moments and fitness levels, as measured by estimated VO_2 max during jogging.

Moments	Ankle Extensor	Knee Adductor	Knee Extensor	Hip Adductor	Hip Extensor	Hip Flexor
Change in moment	0.03	0.06	0.07	0.11	0.18	0.007

Table 4-5: R-Square values for association between change in moments and $V0_2$ max for jogging.

<u>Multiple regression model results:</u>

The fifth hypothesis looked at the effect of adding measures of adiposity and strength to predict the relationship between cardiorespiratory fitness and gait biomechanics. VO_2 max, right leg strength and adiposity were used as the three predictor variables to predict the lower limb moments both pre-fatigue and post-fatigue. The stepwise model building approach for knee extensor moment for walking at pre-fatigue identified both strength and adiposity as predictors with an overall adjusted R-square value=0.35. The models for the other lower limb moments selected only one predictor variable (Tables 4-6, 4-7, 4-8 and 4-9). The following tables show the step wise regression for significant model results, for pre and post-fatigue walking and jogging trials.

Pre PACER walk

Moments	Variable Added	R-Square value	Adjusted	P-value
Knee Extensor	Strength and Adiposity	0.407	.358	0.002
Knee Adductor	Adiposity	0.263	0.233	0.006
Hip Adductor	Adiposity	0.331	0.303	0.002

Table 4-6: Showing the variable added into the step wise regression model for individual moments with r-square and p-value at each step. For ankle and hip extensor moments, no variables entered the predictive model equation.

Post PACER walk

Moments	Variable Added	R-Square value	Adjusted	p-value
Knee Extensor	Strength	0.321	0.294	0.002
Hip Adductor	Fitness	0.145	0.111	0.050

Table 4-7: Showing the variable added into the step wise regression model for individual moments with r-square and p-value at each step. For ankle extensor, knee adductor and hip extensor moments, no variables entered the predictive model equation.

Jogging Results: For Jogging, only hip adductor moments showed significant model for both pre and post-PACER trials.

Pre PACER Jogging

Moments	Variable Added	R-Square value	Adjusted	p-value
Hip Adductor	Adiposity	0.148	0.114	0.048

Table 4-8: Showing the variable added into the step wise regression model for individual moments with r-square and p-value at each step.

Post PACER Jogging

Moments	Variable Added	R-Square value	Adjusted	p-value
Hip Adductor	Strength	0.180	0.147	0.028

Table 4-9: Showing the variable added into the step wise regression model for individual moments with r-square and p-value at each step.

Discussion

The purpose of this project is to determine how cardiorespiratory fitness and fatigue influence gait biomechanics in overweight and obese children (aged 8-11 years). The knee and hip adduction moments showed weak to moderate inverse relationship with fitness as estimated by VO_2 max in the non-fatigue state. Introduction of cardiorespiratory fatigue, induced by the PACER protocol, caused an increase in hip extensor, knee adduction and knee extensor moments for walking, but the change in moments from pre to post fatigue was only associated with the knee extensor moment. Jogging did not influence the effects of fitness and fatigue as compared to walking. The addition of adiposity to the model helped improve the association between cardiorespiratory fitness and lower limb moments. The outcomes of each hypothesis has potential implications for obese children performing physical activity, as well as clinicians attempting to intervene in the cycle of obesity expected in these obese children.

Previous studies on the biomechanics of gait in overweight and obese children have shown that obese children exhibit hip and knee joint stresses higher than their normal weight counterparts (Shultz, 2010; Strutzenberger, 2011). The magnitude of pre-fatigue hip and knee moments reported in the current study were similar to those reported by Shultz et al. For example, knee adductor moments were 0.35 Nm/kg in both studies and hip extensor moments were also similar (0.77 Nm/kg in the current study and 0.89 Nm/kg in the Shultz study).

One of the primary goals of this study was to better understand the relationship between fitness and gait biomechanics; the current study aimed at testing subjects with a wide spectrum of fitness levels. A recent study by Lee et al, explored the association between cardiorespiratory fitness and abdominal adiposity in youth (mean age 11 years) (Lee, 2007). Cardiorespiratory fitness was grouped into low (23.8 ± 3.8 ml/kg/min), and moderate/ high (37.3 ± 8.0 ml/kg/min) categories. The current study had subjects with a range of 22.6 to 46.5 ml/min/kg VO_2 max, as estimated by PACER protocol. This range occupied the full spectrum of Lee's definition of low and high cardiorespiratory fitness, fulfilling one of the study's primary recruitment goals.

The results showed a weak to moderate inverse relationship between estimated $V0_2$ max and hip and knee moments in a non-fatigue state, with the strongest relationship seen for hip and knee adduction moments. Correcting the moment values for speed marginally improved the relationship between moments and fitness levels. Although the relationships were not strong, there are trends suggesting that unfit people have higher biomechanical loads than their more fit counterparts. These trends may have implications on participation in their physical activity and long term effects on the musculoskeletal system.

The associations found for the frontal plane show some similarities with our previous work on obese adult women, however, the sagittal plane associations are in contrast to our previous results where the hip extensor moments showed a positive relationship with fitness levels (Chapter 3). Greater adduction moments in children, as observed in lesser fit children in the current study may result in the development of altered frontal plane alignment, further increasing the risk of OA development as adults (Taylor, 2006). It has been shown that the increased knee adductor moments unevenly distribute force across the medial compartment of the knee and create an increased risk of genu valgum, a condition common among obese children with BMI values above the 95th percentile for age and sex (Wearing, 2006).

The impact of cardiorespiratory fatigue on gait biomechanics was one of the primary foci of this study. All subjects, at the end of PACER protocol, had heart rates of more than 80 percent of their estimated maximum heart rate, affirming the presence of cardiorespiratory fatigue. While the cardiorespiratory fatigue recovery window may vary between subjects, certain studies have suggested that a 1 minute heart rate recovery is an effective measure to assess recovery (Singh, 2007). Children with higher BMI, and those with lower exercise endurance, have a slower heart rate recovery (Singh, 2010), suggesting the possibility of an even longer window for fatigue recovery in obese subjects. Additionally, studies on muscular fatigue have collected meaningful post-fatigue pre-recovery gait data within a ten minute window to avoid recovery (Cheng and Rice, 2005; Parijat and Lockhart, 2008). The current study had a mean time of less than one minute from the end of the PACER fatigue protocol to the start of jogging and walking trials data collection, during which time the children were typically not idle.

As hypothesized, there was an increase in the hip and knee moments after cardiorespiratory fatigue induced by the PACER protocol. Hip and knee extensor moments in addition to knee adduction moments increased after fatigue. Because lesser fit children have higher knee adduction moments at pre-fatigue, the introduction of fatigue further increased the moments, thus amplifying the effect observed in hypothesis 1. While no literature could be found on the effects of cardiorespiratory fatigue on gait biomechanics in obese children, several studies have noted the impact of muscular fatigue in obese adult subjects. Studies investigating running and drop-landing type activities in adults have reported that muscular fatigue leads to an increase of ground impact forces (Christina, 2001). The current study reported an increase in hip and knee extensor moments during the acceptance phase of gait, which could also be attributed to higher ground reaction forces, although the ground force data were not analyzed as part of this study. The results are in contrast with our previous work on adult women which showed an increase in knee extensor moments, but a decrease in the hip extensor moments (Chapter 3).

The increase in peak hip and knee extensor moments due to a cardiorespiratory fatigued state, suggests increased biomechanical stresses which can have long term implications on the musculoskeletal system in children. The greater hip extensor moments could contribute to increased compressive forces on the capital femoral growth plate during gait (Wills, 2004). This, combined with the increased shearing forces, due to larger adduction moments, could cause the femoral neck to exceed failure loads at the proximal femoral epiphysis, resulting in slipped capital femoral epiphysis (Pritchett, 1988). Increased absolute peak knee extensor moments can increase the force with which tibiofemoral contact is made during the screw home mechanism. This increase in intraarticular pressure may increase the risk of earlier progression of osteoarthritis in the overweight population (Loder, 1996).

There were no strong associations between changes in moments from pre to post-fatigue and fitness levels. The only notable relationship was for knee extensor moments, but the results were contrary to the hypothesis, as the change in moments was higher in subjects with higher fitness levels. The results are similar to our work on obese adults where change in knee extensor moments after 30 minutes of continuous walking showed good association with fitness levels (Chapter 3). One possible explanation for these findings is that some lesser fit subjects already have higher moments pre-fatigue (Figure 4-8) and they may not have the capacity, unlike fit subjects to change their gait pattern to counteract fatigue. However, hip moments did not show any associations with fitness levels, which implies that the higher magnitude of moments in lesser fit children stays high after a fatiguing protocol. These higher moments make them susceptible to higher joint stress.

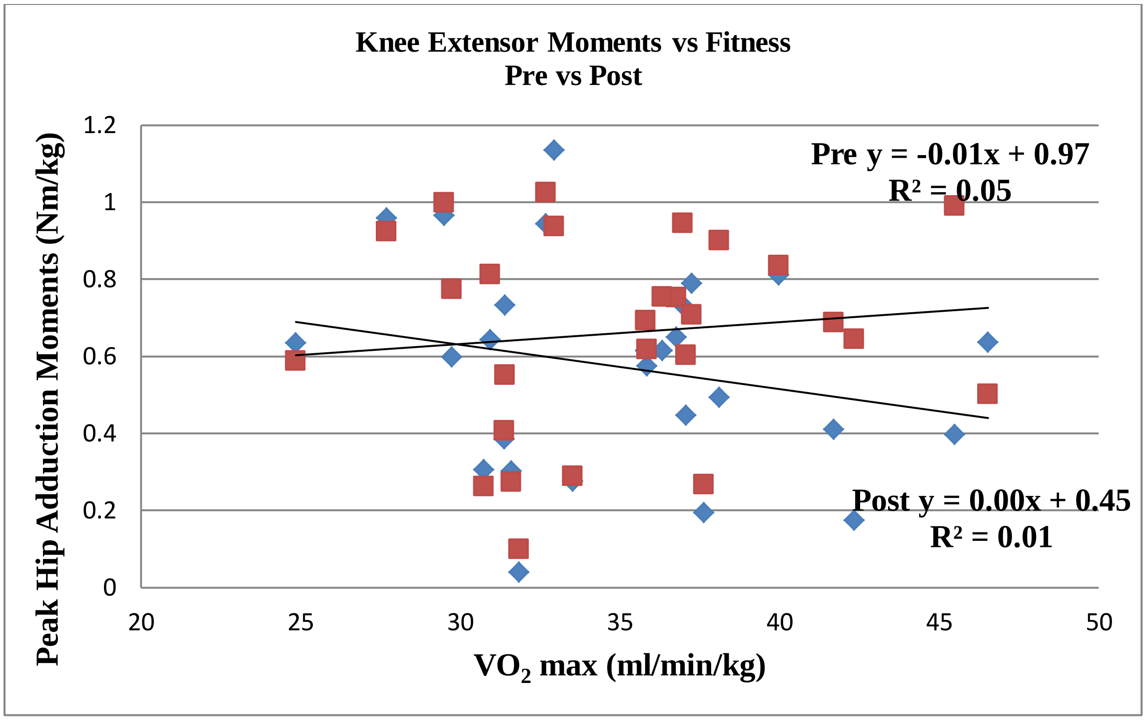

Figure 4-8: showing the association of knee extensor moments with fitness levels pre-fatigue (blue) and post-fatigue (red) walking trials. It can be observed that some subjects especially at the higher end of the fitness spectrum show a greater increase than some of the subjects with lower fitness.

There was no association between fitness and moments during pre-fatigue and post-fatigue jogging. Hip extensor moments showed an increase during post-fatigue jogging trials. The PACER protocol induced cardiorespiratory fatigue, and jogging trials were performed before the walking trials; therefore, it was expected that jogging would show greater change in moments post fatigue. Knee adduction or extensor moment did not show an increase. A recent study looked at sagittal plane running kinetics in 7-10 year normal developmental boys (Chia, 2012). The magnitude of knee extensor moments reported in the current study were lower than in their study, however, the hip extensor moments in the current study were higher which could be attributed to the difference in jogging and running mechanics. In the current study, knee extensor moments were approximately twice than those during walking, whereas hip extensor moments were almost similar. After fatigue, hip extensor moments increased significantly for both walking and jogging, in contrast knee extensor moments remained almost similar after jogging. This suggests that obese children may not have much in reserve to compensate for fatigue, however this relationship between fatigue and moments during jogging is unclear and is grounds for further study.

It was hypothesized that including measures of strength and adiposity will improve the relationship between cardiorespiratory fitness and moments. Linear regression models were developed using stepwise approaches to investigate the effects of adding adiposity and strength measures. Adiposity was the most common variable selected for pre PACER trials, which improved the R-square values. The model for hip adduction moment entered adiposity for both pre PACER walking and jogging trials. In addition the model for knee adduction also included adiposity, suggesting that the higher moments could be due to the presence of excess adipose tissue. The increase in knee adduction could be due to the increased amounts of adipose tissue between the thighs of overweight and obese children (Gushe, 2005). Previous studies have also shown gait deviations in the frontal plane by placing foam around the thighs of normal weight children to mimic the adipose tissue between the thighs (Davids, 2009). This finding suggests that adiposity is an important variable in predicting frontal plane moments in overweight and obese children.

Strength was tested using a custom made leg press machine. The knee extensor model was the only model to include both strength and adiposity in the step wise model. The model for knee extensor moments included knee extensor strength in both pre and post walking models. This finding suggests that strength is a meaningful predictor for moments, particularly when strength is measured for the specific joint being examined. It was interesting that the model for hip adductor included adiposity for pre-fatigue jogging trials whereas strength was included for post-jogging trials. This points to the possibility of an increased role of hip muscles during fatigue. It has been shown that during fatigued walking, hip extensor moments increase during the weight acceptance phase of the gait cycle to stabilize the pelvis and femur (Wang, 2010). The same phenomenon was seen for jogging trials in the current study as hip extensor moments showed a significant increase from pre to post-fatigue jogging trials. These findings suggest that strength is an important variable for predicting the joint moments, mainly in the sagittal plane and can have potential implications on the both walking and jogging.

Overall, adiposity was the main factor in models for adduction moments, whereas knee strength correlated with knee extensor moments. These results suggest that level of adiposity and strength might be important factors in predicting the relationship between fitness levels and gait biomechanics. While these findings should be explored further, clinicians should consider the role of adiposity and strength, in addition to BMI, while developing exercise prescriptions to respond to or prevent obesity in children.

The study did have some limitations. The subjects were asked to perform the PACER protocol with markers on, which might have affected their ability to perform at full capacity. As the subjects were not connected to the motion analysis system during the PACER, the current study did not capture the change in gait patterns during the fatigue protocol. This also meant that subjects needed to be reconnected to the wire immediately after they finished the PACER protocol, delaying the start of post-fatigue trials, and potentially allowing for some cardiorespiratory recovery. Finally, as the study only recruited obese and overweight children, there was no normal weight group for the purposes of comparison.

The results show a weak to moderate inverse relationship between cardiorespiratory fitness and gait biomechanics, as measured by hip and knee moments. This has demonstrated associations between level of fitness and adiposity and biomechanical loads, which may have implications for participation in activities and long-term effects on the musculoskeletal system. Furthermore, the hip and knee extensor and knee adduction moments show a significant increase after fatigue. This might have implications in the clinics, where gait patterns may not be present when obese children are briefly examined during an unfatigued state. This study provides information on how the level of fitness and fatigue might affect the response during clinical evaluations in overweight and obese children. The study also underscores the importance of adiposity in predicting the relationship between fitness and biomechanical loads.

CHAPTER V
CONCLUSIONS

The purpose of this thesis was to explore how segment biomechanics, in the form of joint stress and restricted range of motion, are influenced by obesity and fitness. The results from this work may better explain the biomechanical underpinnings of the reverse causation hypothesis in relation to obesity in both children and adult populations.

The first study (Chapter 2), titled 'Biomechanical Loads during Common Rehabilitation Exercises in Obese Individuals,' collected data on 20 adult female subjects performing lunge and squat rehabilitation exercises. Subjects were measured squatting down, feet shoulder width apart with right foot on force plate and held for 3 seconds at 3 different knee angles: 60, 70, and 80 degrees (full knee extension being 0 degree). Real time feedback was used to achieve the desired knee angle. Forward lunging was held for 3 seconds, with the right lead foot on the force plate at 3 different distances between feet-heel to toe: 1, 1.1, and 1.2 times subject's tibial length. Mean values, over 3 seconds while holding the positions, were calculated for lower limb joint moments and support moments (summation of the lower limb extensor moments). The moments were normalized to body mass. The results suggest that obese individuals experience higher biomechanical stress than normal weight subjects while performing squat and lunge exercises. Non-linear associations were uncovered between anthropometric measures and kinetic measures, which makes the assessment of how best to approach exercise in this population even more challenging. Thus, while this study advocates for the need to consider obesity as a factor in exercise prescription, it acknowledges the apparent complexity that inhibits the understanding of issues that bias the kinetic measures.

The second study (Chapter 3), titled "Changes in Gait over a 30 Minute Walking Session in Obese Females" assessed the biomechanical gait changes in obese subjects over a 30 minute walking session. Data were collected on 20 adult female subjects and peak hip and knee moments, normalized to body mass, were calculated for five gait cycles during both pre and post treadmill data collection. Results show increases in extensor moments at both the hip and knee joints, pointing to greater muscle work and the possibility of increased joint stress. Increased muscle and joint stress can cause discomfort and lead to non-compliance and attrition from the walking programs. Taking time dependent changes in hip and knee joint kinetics into consideration may improve compliance to walking programs.

The third study (Chapter 4), titled 'Do Fitness and Fatigue affect Gait Biomechanics in Overweight and Obese Children?' investigated obesity-related biomechanical issues in children. Three dimensional motion analyses and fatigue protocols were used to analyze the effect of cardiorespiratory fitness and fatigue on walking and jogging biomechanics in 8-11 years old overweight and obese children. A weak to moderate inverse relationship was seen between cardiorespiratory fitness and gait biomechanics, as measured by hip and knee moments. Peak knee adduction, knee extensor, and hip extensor moments during walking increased after the PACER fatigue protocol. This study provides information on how an individual's level of fitness and adiposity affects biomechanical loads, which may have implications for participation in activities and long-term effects on the musculoskeletal system.

Research Findings and Hypotheses

These three studies provide pieces of evidence towards the research hypotheses suggested in Chapter 1. The first hypothesis was that restricted joint mobility in obese subjects will be associated with decreased hip and increased knee joint moments and that these differences will be more evident as the level of difficulty of squat and lunge increases. The results from Chapter 2 on squat and lunge showed that this hypothesis was partially supported. The knee moments did increase as the depth of the squat increased, but did not show any changes for the lunge. Contrary to the hypothesis, hip moments increased during the lunge and were significantly higher in obese subjects.

The second hypothesis was that the hip and knee adduction and extensor moments, will increase in obese individuals, following a 30 minute walking period. The results from Chapter 3 again partially supported the hypothesis. Whereas knee extensor moment increased in obese subjects there was also an increase in moments for normal weight subjects. In addition, there was a decrease in hip extensor moments, proving the second part of this hypothesis incorrect.

The third hypothesis considered how gait biomechanics was associated with cardiorespiratory fitness and cardiorespiratory fatigue in overweight and obese children. This hypothesis, addressed in Chapter 4, included three sub-hypotheses, expecting higher moments in obese children with low fitness, which would be further amplified after the introduction of fatigue. A weak to moderate relationship was found between cardiorespiratory fitness and moments partially supporting the hypothesis, and the moments increased after fatigue, fully supporting the second hypothesis. In addition, measures of adiposity and lower limb strength seem to help in predicting moments.

This thesis concluded that the relationship between biomechanics and obesity, observed in adults and children, should be considered in the larger framework of the reverse causation hypothesis. This research suggests fairly new ideas on reverse causation hypothesis and the complex interaction of biomechanics and fitness levels in overweight and obese populations. Evidence to support the reverse causation hypothesis in both adults and children should continue to be accumulated, and future research is recommended to develop these ideas and provide further evidence to support them.

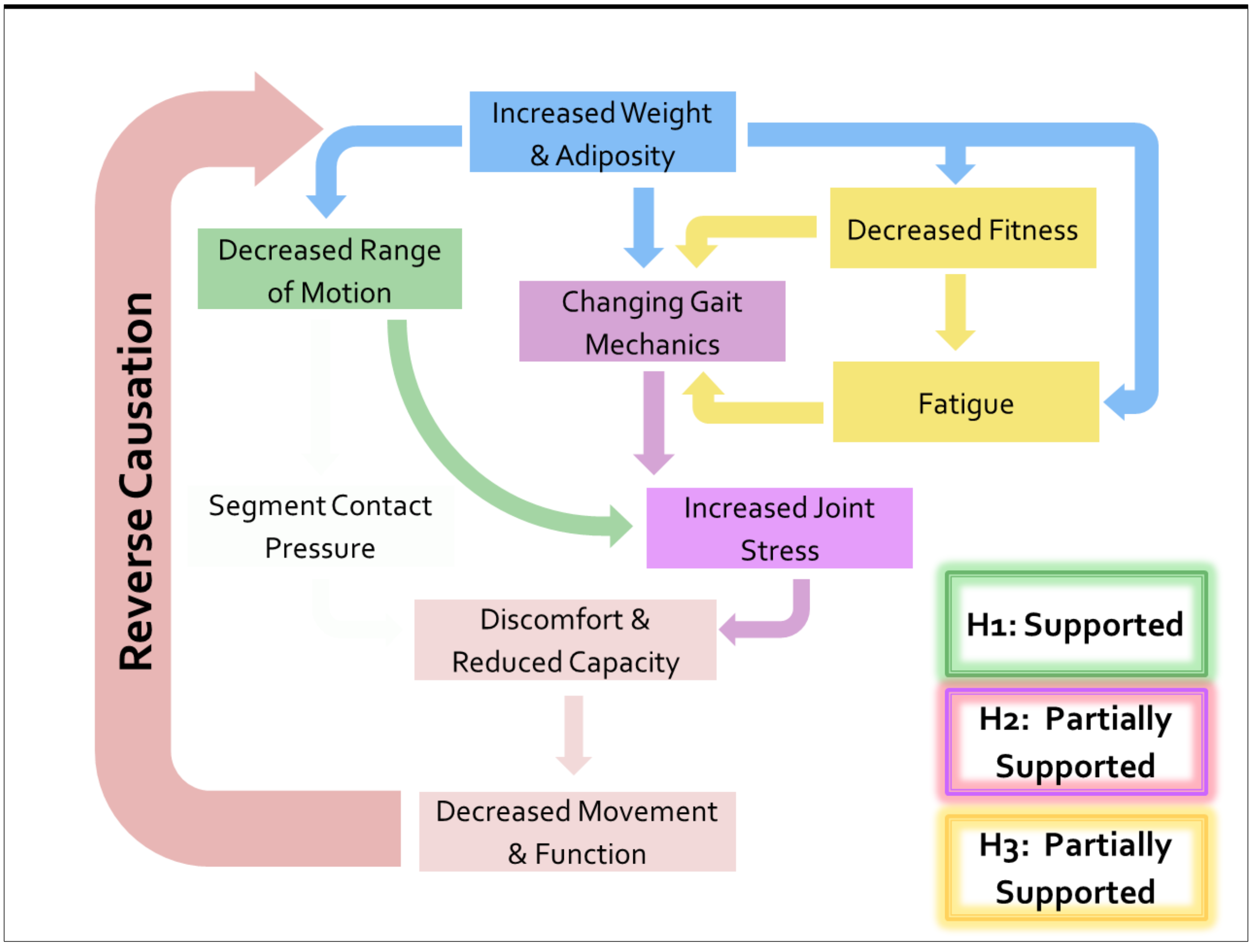

Figure 5-1: "Biomechanics and Reverse Causation" outlines the aspects of the reverse causation hypothesis feedback loop as explored by this thesis. The outside arrow, leading upwards from "Decreased Movement and Function" to "Increased Weight and Adiposity," summarizes the reverse causation hypothesis in action.

Public Health Implications

As recommendations from American College of Sports Medicine continue to increase the physical activity demands on obese children to lose weight, it becomes imperative to explore how their fitness levels along with their obesity affects their joint stresses. The fact that moments increase in presence of fatigue presents a challenging situation to the clinicians who want to keep increasing the physical activity recommendations. Given that change in moments from pre to post fatigue was not strongly related to fitness levels, but the pre fatigue moments were related to fitness levels might suggest that fitness is a more crucial factor. Clinicians can focus on trying to improve pre fitness levels and avoid fatigue but further research is needed in this direction.

Future Directions

Given the prevalence of childhood and adult obesity and the difficulty many obese individuals have to complete effective, long-term weight loss interventions, it is critical that we continue to improve our understanding of how obesity affects the biomechanics and energetics of human locomotion. A critical need is longitudinal data so that we may better understand the etiology of musculoskeletal disorders associated with obesity when gait becomes painful and then fractionalize or modify exercise to allow for health benefits. Very little is known about how increasing or decreasing body mass, cardiovascular fitness or musculoskeletal strength affects gait biomechanics in obese individuals. Such studies are necessary to implement successful weight management programs that promote physiological as well as musculoskeletal health.

APPENDIX A
THE JACKSON NON-EXERCISE TEST

PA-R Directions. Select the appropriate number (0 to 7) which best describes your general activity level for the **previous month**.

Category 1: Do not participate regularly in programmed recreational sport or heavy physical activity.

0 - Avoid walking or exertion, e.g., always use elevator, drive whenever possible instead of walking.

1 - Walk for pleasure, routinely use stairs, occasionally exercise sufficiently to cause heavy breathing or perspiration.

Category 2: Participated regularly in recreation or work requiring modest physical activity, such as horseback riding, calisthenics, gymnastics, table tennis, bowling, weight lifting, yard work.

2 - 10 to 60 minutes per week.

3 - Over one hour per week. Prediction Tests.

Category 3: Participate regularly in heavy physical exercise such as running or jogging, swimming, cycling, rowing, skipping rope, running in place or engaging in vigorous aerobic activity- type exercise such as tennis, basketball, or handball.

4 - Run less than one mile per week or spend less than 30 minutes per week in comparable physical activity.

5 - Run 1 to 5 miles per week or spend 30 to 60 minutes per week in comparable physical activity.

6 - Run 5 to 10 miles per week or spend 1 to 3 hours per week in comparable physical activity.

7 - Run over 10 miles per week or spend over 3 hours per week in comparable physical activity.

Modified Physical evaluation form/ Questionnaire

Name………………………………

Age……………………

D.O.B ___/___/19_____

Address…………………………………

Height………………

Weight………………

Please circle YES or No to the following:

Has your doctor ever said that you have a heart condition and recommended only medically supervised physical activity?

YES /NO

Do you frequently have pains in your chest when you perform physical activity?

YES / NO

Do you feel any discomfort or pain after during exercise?

YES /NO

Do you lose your balance due to dizziness or do you ever lose consciousness?

YES / NO

Do you have a bone or joint problem that could be made worse by a change in your physical activity?

YES /NO

Are you pregnant now or have given birth within the last 6 months?

YES /NO

Have you had a recent surgery?

YES / NO

Are you aware of any injury, past or present, which may be aggravated by any form of exercise?

YES /NO

Do you have previous treadmill walking experience?

YES /NO

Are you presently, or have you previously, played a specific sport?

YES /NO

Do you currently participate in any regular activity program designed to improve or maintain your physical fitness?

YES /NO

On a scale of 1-10, how would you rate your present fitness level (1=Worst 10=Best)?

APPENDIX B
HIP MOMENT VS FITNESS CORRECTED FOR SPEED

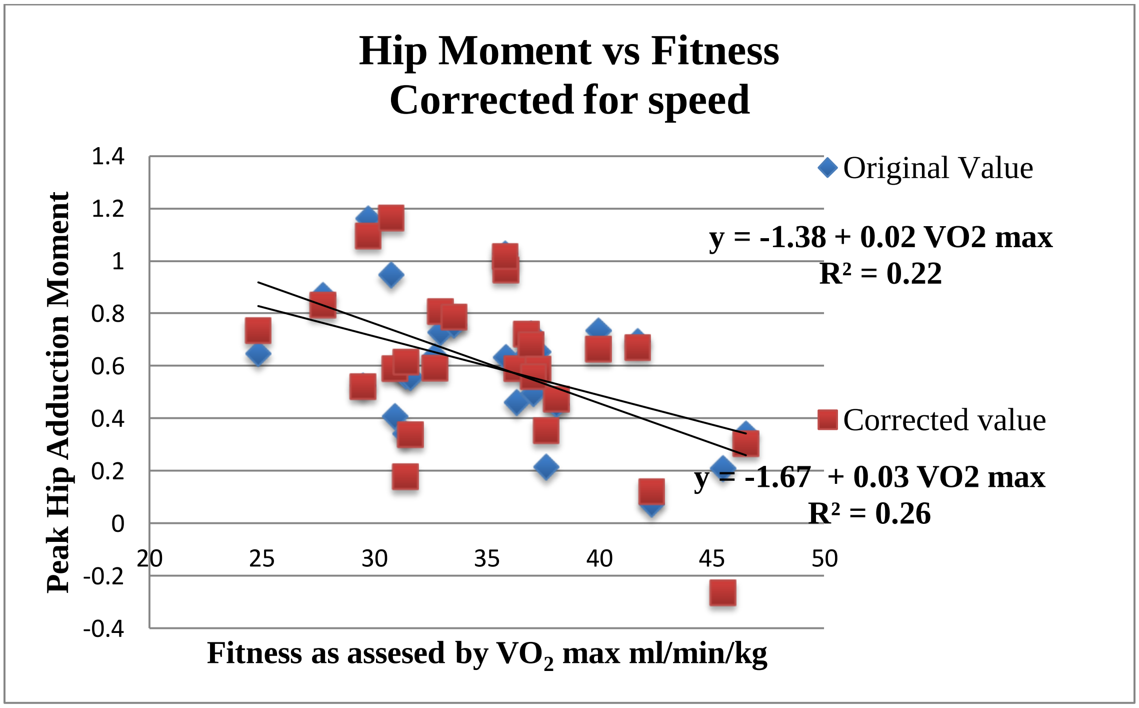

Figure B-1: Shows inverse relationship between peak hip adduction moments and fitness levels, as measured by estimated VO_2 max in a non-fatigue state during walking for 28 subjects.

APPENDIX C
WONG-BAKER PAIN RATING SCALE

Wong-Baker FACES Pain Rating Scale

0 = VERY HAPPY, NO HURT
1 = HURTS JUST A LITTLE BIT
2 = HURTS A LITTLE MORE
3 = HURTS EVEN MORE
4 = HURTS A WHOLE LOT
5 = HURTS AS MUCH AS YOU CAN IMAGINE
(Don't have to be crying to feel this much pain)

APPENDIX D
SUBJECT DEMOGRAPHIC AND ANTHROPOMORPHIC CHARACTERISTICS, TABLE D-1

OBESE										
ID #	Age	Height (m)	Weight (kg)	BMI	Waist Circumference (m)	Hip Circumference (m)	Non-Exercise Test	Self-Selec	Resting HI	V02 max
OB-1	33	1.75	117	38.20408	1.1	1.14	2	3	70	29.39416
OB-2	39	1.83	109.6	32.72716	1.205	1.238	1	3.1	86	30.6651
OB-3	45	1.74	112.4	37.12512	0.985	1.345	0	3.2	68	**35.38984**
OB-4	43	1.64	85.4	31.75193	0.875	1.145	1	3.1	74	31.52122
OB-6	32	1.7	98.5	34.08304	1.01	1.133	1	2.8	68	32.71408
OB-7	45	1.7	123	42.56055	1.2	1.412	2	2.5	62	26.9865
OB-8	34	1.68	88.2	31.25	1.04	1.14	2	3.7	74	38.70604
OB-9	39	1.65	102.3	37.57576	1.06	1.29	1	3	71	29.90212
OB-10	32	1.58	102	40.85884	1.105	1.328	1	3.18	66	33.37696
OB-12	41	1.655	133	48.55742	1.26	1.4	3	3.2	76	35.67048
Mean Obese	38.3	1.6925	107.14	37.46939	1.084	1.2571	1.4	3.078	71.5	32.43265
S.D.	5.229192	0.069412215	14.93342262	5.440063	0.116065116	0.112549693	0.843274043	0.3076	6.587024	3.477009
NORMAL WEIGHT										
C-1	33	1.57	45.5	18.45917	0.62	0.9	3	2.6	82	28.49284
C-2	44	1.595	54.5	21.42275	0.77	0.905	1	3.3	72	35.6974
C-3	32	1.58	57	22.83288	0.84	0.95	1	3.4	84	33.59324
C-4	42	1.745	82.5	27.09337	0.91	1	5	3.5	80	35.05752
C-5	36	1.79	71	22.15911	0.94	1	3	3.4	76	38.97936
C-6	34	1.615	59.5	22.81245	0.92	0.96	4	3	53	38.38768
C-7	40	1.66	56	20.32225	0.69	0.74	1	3.4	68	37.0246
C-8	36	1.73	71	23.72281	0.71	0.99	3	3.2	65	36.58168
C-9	44	1.655	56	20.44523	0.67	0.96	1	3.75	78	37.11156
C-10	40	1.665	57.5	20.74146	0.68	0.89	1	3.5	80	40.2844
Mean Normal	38.1	1.6605	61.05	22.00115	0.775	0.9295	2.3	3.305	73.8	36.12103
S.D.	4.483302	0.074403181	10.67304497	2.356466	0.11843892	0.07804735	1.494434118	0.316623	9.531235	3.308922

Table D-1: Subject demographic and anthropomorphic characteristics, along with the self-selected treadmill walking speed, resting heart rate and VO_2 max.

APPENDIX E
SUBJECT DEMOGRAPHIC AND ANTHROPOMORPHIC CHARACTERISTICS, TABLE E-1

S.No	Age	Exact Age	Sex	Height (cn	Weight (k	Waist (cm	Hip (cm)	Waist:Hip	BMI	Seated He	Leg Lengtl	VO2 max (	VO2 max (	PACER	maturity
1	9	9.08	M	141	37.32	61	75	0.813333	18.77169	74	80	1850.133	49.57483	45.493281	-3.98489
2	9	9.50	M	145	67.78	100	102.5	0.97561	32.23781	74	68	2195.231	32.3876	30.941896	-3.86407
3	11	11.92	M	152	67.973	95	98	0.969388	29.42045	108	79	2416.58	35.55206	32.953397	0.8061313
4	11	11.00	M	155.5	52.79	84	91	0.923077	21.83187	74	70	2564.345	48.57633	36.771347	-3.232161
5	10	10.58	M	148	46.41	76	89	0.853933	21.18791	121	62	2122.535	45.73444	42.336379	0.952213
6	10	10.00	M	154	74.381	106	105.5	1.004739	31.36321	122	85	2358.663	31.71056	31.408524	0.9732362
7	11	11.33	M	156	79.19	110.5	115.5	0.95671	32.54027	108	68	2494.031	31.49427	29.736042	0.3151554
8	10	10.00	M	156	61	92	100	0.92	25.06575	90	72	2538.975	41.62254	38.107569	-2.175214
9	10	10.00	M	166	108.001	121.25	127	0.954724	39.19292	125	66	3096.4	28.67011	24.845951	0.9354319
10	11	11.00	M	158	54.84	82	96	0.854167	21.96763	122.5	98	2430.532	44.32042	39.978977	1.9537553
11	8	8.08	M	133	36.02	65	77.5	0.83871	20.36294	112	75.5	1687.431	46.84705	46.528272	-1.421938
12	10	10.67	M	143	55.029	87	99	0.878788	26.91036	118	83	2060.065	37.43598	36.971041	1.0352646
13	9	9.08	M	138	52.57	85	95	0.894737	27.60449	117	82	1805.831	34.35099	32.679354	-0.199067
14	9	9.42	M	147.5	67.53	97	107	0.906542	31.03936	107	92	2308.019	34.17769	27.703454	-0.729742
15	9	9.42	M	148	61.645	91	99	0.919192	28.14326	117.5	84	2046.647	33.20054	30.748162	0.115113
16	11	11.50	F	142.5	71.68	105	106	0.990566	35.29948	120	79	1920.278	26.7896	21.934885	0.3150706
17	9	9.42	F	138.5	43.97	86	85	1.011765	22.92223	114	76	1493.605	33.96874	36.329067	-0.976779
18	10	10.42	F	154.5	75.15	95	106.5	0.892019	31.4827	119.5	84	2405.409	32.0081	29.501843	-0.334317
19	10	10.42	F	148.5	54.91	89.5	91.5	0.978142	24.89995	119	82	1817.473	33.09912	35.852149	0.0881539
20	10	10.42	F	147.5	59.007	90	93	0.967742	27.12186	117.5	86	1884.97	31.94486	31.37748	0.0427543
21	11	11.00	F	161.5	63.53	86	100	0.86	24.35756	122.5	92	2237.866	35.22535	35.818425	1.0149337
22	11	11.67	F	153.5	59.74	90	99	0.909091	25.35412	122	93	2008.028	33.61278	41.712526	1.6374099
23	11	11.33	F	166	106.38	103	128	0.804688	38.60502	127	97	2567.34	24.13367	22.610291	0.6128241
24	9	9.08	F	134	49.308	80	92	0.869565	27.45767	113	83	1248.948	25.32953	33.534634	-1.119116
25	10	10.00	F	153	48.674	73	84	0.869048	20.79286	121	93	1861.212	38.23831	37.261535	0.5861435
26	9	9.17	F	136.5	41.374	76	82	0.926829	22.20344	114	84	1465.502	35.42085	31.83502	-0.760711
27	11	11.00	F	156	65.086	98	99.5	0.984925	26.74227	122.5	93	2133.841	32.78495	31.611355	1.0185281
28	10	10.50	F	149	48.88	76	89	0.853933	22.01703	116	90	1817.035	37.17339	37.628415	0.4431913
29	8	8.50	F	135	42.81	82	86.5	0.947977	23.48971	111	81	1315.363	30.7256	37.083155	-1.601021

Table D-2: Subject demographic and anthropomorphic characteristics, along with the VO_2 max and Maturity (peak height velocity).

APPENDIX F
NEMETH AND PACER

Nemeth and PACER

Fitness of subjects in form of VO2 max was estimated by two methods in the current study. The usual method used for measurement of maximum oxygen consumption (VO_2 max) ($mL \cdot kg^{-1} \cdot min^{-1}$) is by performing open circuit spirometry with a progressive treadmill walking protocol till volitional exhaustion. A variety of assessments like 20 min shuttle run and 1-mile run/walk have been developed that allow for the prediction of VO2max with an equation and data from a brief episode of exercise. Although only PACER was used for further analysis, both Nemeth and PACER methods are useful. While Nemeth has been validated in a large sample size normal weight subjects and PACER is commonly used in schools to test fitness. The unique finding on this study was to compare a group activity test in form of PACER to one done individually on the treadmill (Nemeth). We had a research team member running with the subjects to keep them motivated and given that the final HR at the end of PACER was close to 80% of their HR confirms the fact that the subjects were fatigued. Given the Nemeth and PACER are strongly related and give the same results is a positive finding gives valuable information which can be used by exercise physiologists as well as physical education teachers in schools.

APPENDIX G
GENDER AND MATURITY

Gender and Maturity

Cummings et al. (2010), in an attempt to explore the relationship between insulin resistance and fitness levels, divided children into groups based on fitness and adiposity levels, and found that gender and BMI differences significantly impacted the relationship (Cummings, 2010). Only overweight and obese children were recruited for the current study and no significant differences in age, height, weight, adiposity and fitness levels were found between boys and girls. This could be attributed to the pre-pubertal age range from 8-11 years of the subjects. There is some evidence that obese girls tend to achieve puberty earlier than their normal weight counterparts. It could be argued that there are abrupt or irregular changes in strength and stature during puberty, which might result in different gait patterns. There were no differences between maturity levels of boys and girls as reported by peak height velocity and no relationship was found between strength and the maturity levels suggesting maturity was not an issue in the current study.

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