THESIS

EFFECTS OF SPEED AND GRADE ON THE BIOMECHANICS AND ENERGETICS OF WALKING IN OBESE ADULTS

Submitted by

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WE HEREBY RECOMMEND THAT THE THESIS PREPARED UNDER OUR SUPERVISION BY KELLIE EHLEN ENTITILED EFFECTS OF SPEED AND GRADE ON THE BIOMECHANICS AND ENERGETICS OF WALKING IN OBESE ADULTS BE ACCEPTED A FULFILLING IN PART REQUIREMENTS FOR THE DEGREE OF MASTER OF SCIENCE.

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ABSTRACT OF THESIS

EFFECTS OF SPEED AND GRADE ON THE BIOMECHANICS AND ENERGETICS OF WALKING IN OBESE ADULTS

Brisk walking is a recommended form of exercise for obese individuals. However, lower extremity joint loads and the associated risk of musculoskeletal injury or pathology increase with walking speed. Walking uphill at a slower speed may offer an alternative form of moderate intensity exercise that reduces joint loading. The purpose of this study was to quantify the biomechanics and energetics of level and uphill walking in obese adults. We hypothesized that compared to brisk level walking, walking slower up a moderate incline would reduce lower extremity joint loading while providing appropriate physiologic stimulus. Twelve obese adult volunteers, age = 27 (5.5) years, mass = 100.5 (15.7) kg, BMI = 33.4 (2.6) kg/m^2, (mean (S.D.)), participated in this study. We measured ground reaction forces, three-dimensional lower extremity kinematics and oxygen consumption while subjects walked on a dual-belt force measuring treadmill at several speed (0.50-1.75m/s)/grade (0-9°) combinations. We calculated net muscle moments at the hip, knee and ankle and metabolic rate for each condition. Walking slower uphill significantly reduced net muscle moments at the knee compared to faster level walking ($p<0.05$). Peak knee extension and adduction moments were reduced by ~19% and 26%, respectively, when subjects walked at 0.75m/s, 6° vs. 1.50m/s, 0°. The greater knee moments during level walking suggests subjects had greater medial compartment knee joint loads. All walking trials were moderate intensity (48.5-59.8% of VO_2max). A slower walking speed combined with a moderate incline appears to be an effective strategy for reducing knee joint loads while providing appropriate cardiovascular stimulus in obese adults.

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CHAPTER I

INTRODUCTION

The prevalence of obese adults (BMI > 30 kg/m^2) in America continues to exceed 30% across most age and sex groups [1]. Obesity is associated with many diseases including heart disease, certain cancers, and diabetes, and is the main preventable risk factor for large-joint osteoarthritis (OA) [2-4]. Obesity is due, in part, to an imbalance of energy intake and expenditure, and both diet and physical activity are considered essential tools in weight management [5].

Obese individuals are advised to participate in moderate intensity physical activity (40% - 60% of VO_2max) for a minimum of 30 minutes at least 5 times a week [6]. Walking is often the recommended form of physical activity for obese persons because it is convenient and there is a low injury rate among lean individuals [7]. However, when walking at their preferred speed (~1.0-1.4m/s), the intensity may not meet the moderate threshold [8-9]. As a result, obese individuals must walk faster (i.e. "brisk" pace, ≥1.5 m/s) to meet the physical activity guidelines, achieve the physiological benefits of exercise, and maximize energy expenditure.

A strong positive relationship exists between level walking speed and lower extremity joint loading (estimated via net muscle moments, joint reaction forces and joint loading rates) [10-11]. During level walking, the magnitude and medio-lateral distribution of knee joint loading can be estimated via extension and abduction (external adduction) net muscle moments, respectively. According to Lelas et al., peak early stance sagittal knee moments increase nearly 2.5 fold as walking speed increases from 0.75m/s to 1.5m/s [11]. External knee adduction moments also increase with walking speed, although the increase is more modest [12]. In addition, recent research suggests that absolute net muscle moments (Nm) may be greater in moderately obese vs. non-obese persons [13-14]. Therefore, the combination of brisk level walking and obesity is likely to result in relatively large loads across the lower extremity joints, particularly within the medial compartment of knee. Faster walking speeds also increase the rate of loading during early stance as peak vertical ground reaction force increases while early stance duration decreases. These greater and more rapidly applied loads may increase the risk of acute musculoskeletal injury or pathology such as OA [3, 15]. Walking slower would presumably reduce these risks but would also reduce the cardiovascular benefits due to the relatively low aerobic demand.

A potential strategy to maintain adequate exercise intensity would be to have obese individuals walk uphill at a relatively slow speed. Walking up a moderate incline (<10°) increases metabolic rate compared to walking on a level surface at the same speed [16]. American College of Sports Medicine (ACSM) equations are commonly used to estimate oxygen consumption and can therefore be used to predict speed/grade combinations that elicit similar metabolic responses [17]. Using standard prediction equations, walking at 0.75m/s up a 6° incline would require approximately the same oxygen consumption as level walking at 2.1m/s [17].

The biomechanics of incline walking has yet to be determined in the obese population. However, both McIntosh et al. and Lay et al., found that in non-obese persons, lower extremity hip and ankle net muscle moments increased with incline [18-19]. Lay et al., reported that peak extensor knee moments during early stance where not significantly different when subjects walked up an 8.5° ramp vs. level walking at ~1.25m/s [18], while McIntosh et al. reported that peak knee extensor moments increased ~50% during incline (8°) vs. level walking at ~1.65m/s [19]. Thus, incline vs. level walking at typical level walking speeds results in much smaller increases in knee extensor moments than observed when level walking speed is increased from 0.75m/s to 1.5 m/s. In addition, loading rates (slope of vertical ground reaction force during early stance) are reduced at slower speeds due to the smaller peak vertical ground reaction forces and longer early stance times [13]. The combination of reduced knee net muscle moments (extension and external adduction) and lower loading rates provide a rationale for investigating this form of exercise for obese adults. We recognize that net muscle moments may not reflect lower extremity joint loading during uphill walking if there is an increase in co-contraction of agonist/antagonist muscles [20]. However, muscle activation decreases as walking speed decreases [21] and the combined effects of slow speeds and incline on muscle activation are unknown. As a result, this investigation focused on the effects of speed and grade using net muscle moments as a proxy measure for joint loading.

The purpose of this study was to quantify the biomechanics and energetics of uphill vs. level walking in moderately obese adults. We hypothesized that: 1) decreased walking speeds combined with moderate inclines would reduce lower extremity net muscle moments vs. faster, level walking; and 2) slower walking up moderate inclines would provide similar physiologic stimulus and caloric expenditure vs. faster, level walking for moderately obese persons.

Chapter II

LITERATURE REVIEW

Obesity Epidemic

The population of obese adults in America has doubled since 1976 and in 2006, more than 34% of U.S. adults aged 20 years or older were obese [22]. Obesity is most commonly classified as having a Body Mass Index (BMI) > 30 kg/m^2 and is linked to many health issues including diabetes, heart disease, and musculoskeletal pathologies [23]. It has been estimated that approximately 20-30% of cardiovascular disease (CVD) mortality may be attributable to excess body weight [24]. This is significant because heart disease remains the most common cause of death in the U.S., contributing to more than 700,000 deaths a year [25].

Type 2 diabetes is another major health concern related to obesity. Compared to non obese persons, obese individuals have almost 10 times the risk of developing type 2 diabetes [26]. This supports a strong linear relationship between BMI and risk of developing type 2 diabetes. In the Nurses Health Study, many of the type 2 diabetes cases (80%) could be attributed to the combined effect of sedentary behavior and obesity. Additionally, type 2 diabetes is the leading cause of non-injury related amputations, and is associated with blindness, heart disease, kidney disease, neuropathy, and stroke [23].

Obesity is also correlated with a variety of other conditions such as sleep apnea, pregnancy complications, asthma, depression, and musculoskeletal injury [27].

Obesity is a result of many factors: genetic, behavioral, metabolic, and environmental influences [23]. Due to the rapid increase in obesity rates within the past few decades, it is unlikely that biological factors are the primary contributor to this pandemic. Instead, changes in behavioral and environmental factors are the thought to be most influential [23]. These changes are associated with increased caloric intake and decreased energy expenditure, which lead to both positive energy balance and weight gain.

According to the National Health and Nutrition Examination Survey data, the average energy consumption increased from 1971 to 2000, and the obesity rates doubled [28]. However, additional studies did not find a large difference in caloric intake between the mid 1960's to the mid 1990's [29]. This suggests that decreased physical activity levels may be a larger contributor to the rising levels of obesity.

Physical Activity and Obesity

As a result of improved technology, individuals living in industrialized countries have lower levels of physical activity. Although the number of recreational facilities and health clubs has increased, sedentary activities (such as watching television and playing videogames) have increased as well. For instance, there is evidence to support an increased risk of weight gain and obesity with increased hours of television viewing [30]. In addition, the level of activity required for daily living (household chores and transportation), has decreased [31]. Advances in technology have lead to decreased dependence on walking and bicycling for transportation, as well as decreased levels of physical activity required to do regular household chores. As a result, sedentary behavior has increased substantially [32-33].

Population studies indicate that there may be a correlation between decreased levels of physical activity and obesity rates. For instance, the prevalence of obesity in adults significantly increases with age (from 20-60 years of age) [34]. During this same time period, the percentage of adults meeting the minimal physical activity recommendations decreased [35]. It is important to note that although the decreased physical activity levels seen with older age may be partially responsible for the weight gain, it may be that older age is responsible for decreased physical activity. According to Obesity: The Report of the British Foundation Task Force, 28.3% of obese males report low physical activity levels [36]. This is nearly double the percentage of obese male adults who were physically active. Similarly, 29.2% of obese women reported low levels of physical activity, whereas only 16% reported high levels of physical activity[36].

Due to low levels of physical activity reported in obese persons, physical activity is often used in interventions as a way to prevent weight gain and manage weight loss. A Meta-Analysis completed by Richardson and colleagues reported that the average participant in a pedometer based walking program (independent of diet) lost approximately 1 kg during the intervention [37]. Although such modest amounts of weight loss may be discouraging to participants interested in losing weight, it is important to note the numerous health benefits associated with increased physical activity.

Independent of weight loss, increased physical activity has been shown to reduce the risk of cardiovascular events, lower blood pressure, and improve glucose tolerance in individuals with either impaired glucose or type 2 diabetes [38-39].

In order to prevent and treat obesity, it is imperative that physical activity levels increase and sedentary behaviors decrease. Walking is a frequently prescribed form of exercise for obese individuals and is often used in weight loss interventions. It has been shown that walking briskly for five or more hours per week significantly lowers the risk of developing obesity related diseases by 40% [40]. Therefore, walking is a common modality used to increase physical activity levels in an effort to regain energy balance and prevent further weight gain.

Despite well known beneficial effects of being physically active, 50% of U.S. adults do not get enough physical activity [22]. Walking provides a low impact and familiar form of exercise, and is therefore often used in exercise prescription for previously sedentary individuals. It has been shown that walking briskly for either long or short bouts will result in weight loss in overweight and obese persons [41]. In 2007, the ACSM updated their physical activity recommendations. These recommendations include participating in moderate intensity activity for 30 minutes, 5 days per week, or vigorous intensity activity for 20 minutes, 3 days per week [42]. The recommendations also allow for combinations of both moderate and vigorous activities which can be completed in increments of 10 minutes [42]. However, very few adults are meeting even the minimum hours of physical activity per week [43]. The most frequently prescribed moderate intensity activity for adults to participate in is walking at a brisk pace (1.5-1.6 m/s) [7, 44]. However, faster walking speeds may increase loading within the lower extremity joints by increasing net muscle moments about the knee [10-11].

Musculoskeletal Injury and OA

In the United States, the economic cost associated with musculoskeletal disorders is approximately $149 billion [45]. Recent research suggests that modest increases in weight bearing activities (i.e. walking) can induce musculoskeletal discomfort and pain in obese persons, which may lead to injury [46]. The increased risk of musculoskeletal injury is most arguably due to the lack of muscle mass and excess adiposity found in obese persons [47]. Decreased muscle to fat mass ratio is most commonly due to physical inactivity. Thus, it would seem that individuals with low levels of physical activity maybe more likely to develop musculoskeletal injuries during activity.

Hootman and colleagues have reported that physical activity is correlated to musculoskeletal injuries in both sedentary and physically active adults. Greater than 80% of total all-cause injuries in men and women were related to physical activity; of which, 19-23% occurred at the knee joint [7]. Physical activity, particularly brisk walking, is recommended to prevent/treat obesity [48]. However, even in normal weight persons, the knee joint is exposed to compressive loads in excess of three times their body weight during brisk walking [49]. Therefore, although physical activity is widely accepted as an avenue for disease and injury prevention, it may also be responsible for musculoskeletal injuries. Over 75% and 65% of men and women, respectively, reported that their injuries temporarily stopped them from participating in their exercise program, and ~25% reported that the injuries they developed permanently stopped them from participating in their exercise program [7]. Intuitively, it would seem that obese individuals may be more likely to quit an exercise program if they get injured. This highlights the importance of improving our knowledge of musculoskeletal injury prevention and exercise prescription for the obese population.

Walking for exercise has also been shown to increase the risk of developing acute and chronic musculoskeletal injuries in obese individuals, and these injuries are highly associated with OA development [3, 50-51]. OA is traditionally defined as inflammation in the joints caused by the degradation of cartilage within the joint [4]. Older adults as well as obese individuals commonly develop OA, with the knee being the primary site of development [52]. Traditionally, the explanation of obesity-induced OA is that increased axial loads on the joints accelerate normal degeneration that occurs with aging [53]. An animal study by Radin et al. reported that increased mechanical loads within the knee resulted in bone remodeling followed by cartilage degeneration and stress (tearing and splitting of the deep and intermediate layers of cartilage) [54]. Obesity may also induce knee OA by shifting the placement of loads to less frequently loaded regions of the cartilage. For instance, obesity is associated with varus malalignment of the knee which increases medial loading within the joint and is a risk factor for the development and progression of OA [55]. Additionally, faster walking speeds increase the rate of loading during early stance as peak vertical ground reaction force increases while early stance duration decreases [56]. Therefore, these greater and more rapidly applied loads may increase the risk of acute musculoskeletal injury or the development of OA[3, 57].

Recent data suggests that there are other factors, besides mechanical loading, which result from obesity and may cause OA development. For instance, obesity is associated with numerous metabolic disorders which may increase inflammatory mediators (e.g. C-reactive protein and TNF- α) known to further the development of OA [58]. In *vitro* studies have reported that these inflammatory mediators produce catabolic changes within cartilage [59]. Thus, it seems the relationship between obesity and OA development may be due to increased mechanical loads and/or increased inflammatory mediators.

Biomechanics of Walking

In order to better understand the relationship between walking and the development of musculoskeletal injury and pathology, it is imperative that we understand the biomechanics of walking in obese individuals. Although there is a large body of literature regarding the biomechanics of walking in lean persons, few have studied the gait of obese persons [60-63]. In addition, even less information is known about incline walking and its effects on the biomechanics of walking in obese persons.

Joint Kinematics-Level

Kinematics is the study of the spatio-temporal apsects of motion and is frequently studied within the biomechanics field [64]. It is of particular importance in quantifying gait characteristics, including temporal-spatial joint kinematics, body segment and joint positions, angles, velocities, and accelerations.

Temporal stride kinematics include stride frequency, stride length, percent stance of gait cycle, percent swing of gait cycle, percent double support of gait cycle, and step width . In normal weight persons, stride frequency (Hz) and length (m) increase with walking speed. While walking at 1.50 m/s, the average stride frequency and stride length in lean persons is approximately 1.00 Hz and 1.50 m, respectively [19, 56]. At most walking speeds, the majority of the gait cycle is spent in stance, and with increasing speed, percent stance decreases and percent swing increases. At 1.50m/s, approximately 62% and 38% of the gait cycle in lean persons is spent in stance and swing, respectively [56, 60, 62]. Lastly, as speed increases, the time spent in double support falls while step width does not change [56, 62, 65].

Recent literature has compared joint kinematics in lean and obese adults. Although significant differences were observed in the temporal-spatial measurements, no significant differences were found in stride frequency and stride length between the two cohorts when walking at similar speeds [56, 60]. For obese participants, walking at 1.50 m/s results in a stride frequency and stride length of ~1.00 Hz and 1.50 m, respectively. This is in good agreement with stride frequencies and lengths found in lean persons listed previously.

Interestingly, obese persons tend to spend more time in stance and have a greater % double support compared to non obese persons. Browning and colleagues reported that at 1.50 m/s, obese subjects spent ~64% of their gait cycle in stance, which was significantly greater than the 61.5% stance reported in lean subjects [56]. Correspondingly, at 1.50 m/s DeVita et al. reported a 5% shorter swing time and a 3% longer stance time in obese compared to lean subjects [60]. In addition, time spent in double support is greater in obese compared to lean persons. It has been reported that at

1.50m/s, obese persons spend ~27.3% of their gait cycle in double support, whereas lean persons only spend ~21.8% of their gait cycle in double support [56]. Another difference between lean and obese kinematics is their step widths. Browning et al. reported a step width of 0.15 m and Spyropolous et al. reported a step width of 0.16 m in obese persons, which is double the step width reported in lean persons (0.8 m) [62, 66]. Due to their larger mass, particularly within the upper leg, obese persons walk with a much wider stance than lean persons. Adopting a wider stance may also be a result of attempting to maintain balance while walking.

In addition to the discrepancies in spatial-temporal characteristics between obese and lean persons, joint kinematic differences have been proposed. Research suggests that obese persons adopt a slower self-selected walking speed compared to their lean counterparts [14, 60, 62, 66]. In 2003, DeVita and colleagues found that obese subjects walked at a 16% slower preferred walking speed (1.29 m/s) compared to the speed (1.52 m/s) adopted by lean subjects. Similar results have been reported by Lai et al. and Spyropoulos et al., who found that obese have self selected walking speeds of 1.09 m/s-1.12 m/s vs. lean self selected walking speeds of 1.27 m/s-1.64 m/s [14, 62]. However, a recent study by Browning et al. challenges these findings. In this study, moderately obese persons preferred to walk at similar speeds as their lean counterparts (1.40m/s vs. 1.47m/s)[67]. The faster speeds reported in this study may be a result of methodological differences. All previous studies have either used a treadmill or indoor walkway to measure preferred walking speed, yet Browning and colleagues chose to measure the preferred speed of their subjects outdoors. Therefore, it is reasonable to assume that obese persons adopt a slower walking speed compared to their lean counterparts.

Angular Kinematics

Lower extremity angular joint kinematics during walking in lean individuals have been well established. The hip, knee, and ankle joints are denoted as 0° when the body is in anatomical position. The hip is flexed (~30°) at heel strike, extends into mid stance, and flexes again just prior to toe off and during swing [10, 19, 68-69]. Similarly, the knee is flexed at heel strike (5-10°) and experiences a peak flexion angle during loading response (~15-20°). The knee then extends through late stance and flexes (~40°) again prior to toe off [10, 62, 68-69]. During swing, the knee initially flexes (~60°) to aid in foot clearance and then extends prior to heel strike [69]. There is very little dorsiflexion (~5-10°) in the ankle at heel strike, and the ankle reaches peak planterflexion during early stance [10, 56, 60]. The ankle dorsiflexes until late stance, which is followed by a rapid plantarflexion before toe off [10, 19, 68]. During swing, the ankle is dorsiflexed to ensure foot clearance [69]. Slower walking speeds result in reduced range of motion at the hip and knee and an increased range of motion at the ankle (greater degree of plantarflexion) [56]. Browning et al. reported ~30° of hip flexion at 0.50m/s and ~33° of hip flexion at 1.75m/s in lean persons. Similarly, he reported ~13° of knee flexion at 0.50m/s vs. 20° at 1.75m/s [56]. The reduced range of motion at the hip and knee is most likely due to shorter stride lengths adopted during slower speeds.

It has been suggested that during level walking, obese individuals adopt similar joint angular kinematics as non-obese individuals [56, 62]. In 2007, Browning et al. reported no significant differences in midstance hip, knee and ankle joint angles between obese and normal weight subjects across a variety of walking speeds (0.50-1.75m/s) [70].

These results are consistent with those published in Spyropoulos et al., who reported no significant difference between mean hip and knee flexion angles in obese and non-obese subjects [62]. However, there is conflicting evidence that suggests obese persons may have different angular kinematics during walking compared to lean persons [60, 62].

Although Spyropoulos reported a trend for a lesser magnitude of hip flexion at heel strike, no significant differences existed between their lean and obese subjects [62]. However, DeVita and colleagues found that obese persons walking at 1.50m/s experienced ~5° more hip extension throughout stance [60]. DeVita et al. theorized that this greater degree of hip extension may lead to a more erect posture unique to obese persons. There is evidence to suggest that obese persons walk with a more erect and upright pattern than lean persons [60]. This gait pattern is known as quadriceps avoidance, which is achieved by reducing the early-midstance knee flexion angle. A more extended knee results in a reduced net demand on the quadriceps to support body weight, which helps to maintain knee stability [71]. DeVita et al. found that obese persons walk with significantly less knee flexion when walking at the same speed as lean persons, suggesting possible neuromuscular adaptations to walking [60]. However, these findings are unique to DeVita and colleagues, and more recent literature supports no changes in knee angular kinematics.

Browning and colleagues reported similar joint kinematics between moderately obese and lean subjects across a variety of speeds. The difference between Browning's and DeVita's findings may be attributable to a larger range of body mass and BMI used in DeVita's methodology. DeVita's subjects had ~16% greater body mass compared to

Browning's subjects (123.4 kg vs. 110.6 kg, respectively) [56, 60]. Furthermore, the range of BMIs used in DeVita's study (32.4-58.7 kg/m^2) was much greater than that used in Browning's study (30-43 kg/m^2) [56, 60]. Thus, the larger range of BMIs used by DeVita and colleagues may have influenced the results of his study. Although it has been hypothesized that a threshold may exist (where gait characteristics change above a certain level of adiposity) further research using subjects of varying levels of obesity is necessary to confirm this theory [56].

Knee adduction, commonly referred to as varus or bowlegged, occurs when the distal portion of the tibia and the foot tilts medially from the center of the knee in the frontal plane [69]. The differences in knee adduction angles between lean and obese persons are not well established. There is only one study published that has analyzed obese knee adduction angles compared to lean adduction angles. This is primarily because most other studies only measure joint kinematics in the sagittal plane. Lai et al. reported that obese persons walk with greater maximum knee adduction angles than their lean counterparts [14]. Greater knee adduction angles are associated with greater medial compartment loading of the knee [72]. Thus, Lai's results suggest obese persons adopt a gait pattern which increases their risk of developing musculoskeletal injures and pathologies.

Spyropoulos et al. found that at 10% of the walking cycle, obese subjects walked with less ankle plantarflexion [62]. The authors argue that this altered foot position seen in obese subjects may be a mechanism to bring the body weight over the foot as quickly as possible in order to aid in balance. However, DeVita and colleagues reported that compared to lean subjects, obese subjects walked with ~6° more plantarflexion throughout stance, and 7° more plantarflexion at toe off [60]. This discrepancy can most likely be explained by the differences in speeds reported in Spyropoulos. In Spyropoulos' study, obese subjects walked at 1.09m/s and lean subjects walked at 1.64m/s [62]. The 34% slower walking speed adopted by obese subjects resulted in more time spent in swing phase and a greater degree of ankle dorsiflexion.

Joint Kinematics-Incline

Most literature regarding gait kinematics is conducted on level surfaces. The limited literature available suggests that in lean individuals, the hip, knee, and ankle are more flexed during early to midstance while walking up an incline, as required to raise the limb for toe clearance and lift the body [19, 68]. For instance, Lay et al. reported an approximate 62% increase in hip flexion at heel strike while walking at 8.5° compared to level walking [68]. Similarly, McIntosh et al. reported an ~67% increase in hip flexion between level and 8° at heel strike [19]. Intuitively, adopting a more forward flexed trunk would aid in maintaining balance during uphill walking.

However, it is important to note that neither study controlled for speed. Although the differences in speed reported in Lay's paper were small, those reported in McIntosh and colleagues were more substantial. For instance, McIntosh's subjects walked faster (1.57, 1.68, 1.76, and 1.73m/s) at greater inclines (0, 5, 8, and 10°), respectively[19]. Slower walking speeds result in shorter stride lengths, which limit the range of motion in all lower extremity joints. The faster walking speeds reported in McIntosh's study most likely contributed to greater stride lengths and range of motion. Thus, it is unclear as to whether the larger hip range of motion was a result of faster walking speeds or incline.

Further research is necessary to clarify these findings and to expand the knowledge of slope kinematics to include obese persons.

Kinetics

Joint kinetics describe the motion of the body and the forces it experiences during movement, which are represented via Ground reaction forces (GRF), net muscle moments (NMM), torque, joint reaction forces, and joint work and power. Similar to joint kinematics, very little data exists regarding joint kinetics in obese individuals. The following sections compare joint kinetics in lean persons compared to obese persons, and the implications of these differences.

Ground Reaction Forces-Level

During walking, weight shifts from one leg to the other, and upon heel strike, GRFs are generated between the foot and the ground. These forces act in three fundamental directions: vertical, medial/lateral, and anterior/posterior. GRFs are recorded using force plates containing piezoelectric or strain gauge transducers. These forces are important to understand as they, along with the moment arm data (defined as the perpendicular distance of force application to the axis of rotation), inertial components and angular and linear kinematics, are needed to calculate NMM [69]. NMM, also referred to as joint torque, are a proxy measure (as it can not isolate out for co-contraction of the musculature within the joint) of loads across a joint [73], and will be described in greater detail later in this chapter.

Vertical forces are a reflection of the vertical mass-acceleration product of all body segments and therefore represent the total of all net muscle and gravitational forces acting at each instant of time over the stance period [69]. At higher speeds, the vertical

GRF pattern consists of an initial peak at early stance, followed by a decrease during midstance. After a second peak during late stance, the GRF returns to zero at toe off.

There is evidence to suggest that obese persons experience greater vertical GRF than their lean counterparts [14, 56]. In 2007, Browning et al. reported significantly greater vertical GRF in their obese subjects compared to their lean subjects during level walking [56]. This study compared a variety of walking speeds, each of which resulted in ~60% greater GRF in obese compared to their normal weight counterparts. This increase was in direct proportion to the 64% greater body mass of the obese subjects, thereby highlighting the importance of body mass as a determinant of the vertical GRF. They also found that while walking at slower speeds, the first peak of the vertical GRF was much smaller compared to that of faster speeds [56].

Furthermore, Messier et al. reported that when adjusted for age, all vertical force parameters (vertical force, vertical impulse, and loading rate) were significantly associated with BMI [63]. As subjects' BMI increased, the vertical load placed on their lower extremity increased as well. The maximal vertical force was 28% greater for obese subjects than non obese subjects (968.5N vs. 756.2N) [63]. It is important to note that Messier's subjects were older with radiographic evidence of knee OA; therefore, the mass/GRF relationship seems to exist among persons with OA as well. Thus, all research supports the argument that obese persons experience greater vertical GRF while walking at similar speeds as their lean counterparts.

Anterior/posterior (AP) GRFs consist of a braking force during the first half of stance that decelerates the body as a person moves their weight forward over their base of support and a propulsive force during the second half of stance necessary for forward progression [63]. In 2008, Lai et al. found that the antero-posterior propulsive force of their obese subjects was significantly greater than their lean subjects [14]. For instance, their obese subjects experienced ~29% greater AP propulsive force compared to their lean subjects [14]. Further research supports the argument that obese persons experience greater AP GRF while walking compared to lean persons. Messier and colleagues found a significant positive correlation between BMI and AP GRF, and their obese subjects experienced 28% and 25% greater braking and propulsive forces, respectively. Similarly, Browning et al. reported that faster speeds (1.0-1.75m/s) result in ~63% greater peak AP GRF in their obese subjects compared to their normal weight subjects [13]. There was an even greater percentage difference between the two cohorts at slower walking speeds. For instance, while walking at 0.75m/s and 0.50m/s, obese subjects experienced ~71% and 85% greater AP GRF, respectively [13]. However, the within group comparisons suggest that slower walking speeds result in lower AP GRF. For instance, the lean subjects experienced a 72% decrease in AP GRF by slowing their speed from 1.5m/s to 1.0m/s [56]. This same 5% reduction in walking speed resulted in a 67% decrease in obese AP GRF.

Similar to vertical and AP GRF, the medial/lateral (ML) GRF are reportedly greater in obese compared to lean persons [56, 63]. Browning and colleagues found that obese subjects experience 85% greater peak ML GRF than lean subjects across a variety of walking speeds [56]. Slower walking speeds resulted in dramatically smaller first peak ML GRF. For example, decreasing the walking speed from 1.5m/s to 1.0m/s resulted in 29% reduction in ML GRF in obese persons [56]. In accordance with Browning's findings, Messier et al. reported a significant correlation between body mass and ML

GRF [63]. For instance, the medial and lateral GRF were 25% and 38% greater in obese compared to lean subjects, respectively [63]. It is important to note that although Messier et al. controlled for age when reporting vertical GRF, they did not do so for AP and ML GRF. To my knowledge there are no studies that report increased AP and ML GRF as a result of age. Therefore, although age may act as a confounding variable, it remains unlikely.

At heel strike, the maximum lateral force decelerates subtalor eversion and during midstance, the maximum medial force accelerates foot eversion. Greater foot eversion is highly associated with over pronation of the foot and there is some evidence that severely obese individuals walk with excessive subtalor pronation [61]. Messier et al. suggest that obese individuals experience larger ML forces because these forces are necessary to decelerate and accelerate excessive subtalor pronation [63]. It may also be possible that the greater ML forces absorbed by obese persons may translate into greater shear forces at the knee. It has been postulated that when ML and AP forces are combined, the shear forces across the tibial plateau increase as well, causing excess degradation of articular cartilage [63].

Ground Reaction Forces-Incline

Incline walking (both uphill and downhill) is a rather unstudied topic in gait biomechanics, particularly in obese persons. However, the possible implications of incline walking are intriguing as it provides a challenge in our environment which may require a unique neural control of locomotion. Thus, understanding the biomechanics of incline walking is important.

To date, the only literature that has discussed incline locomotion has used lean subjects. During early stance, Lay et al. reported no significant differences in normal GRF between 21°, 8.5°, and level surfaces [18]. In addition, the general shape of the normal GRF pattern was similar between level and incline walking. However, Lay et al. did report significant differences in the 2nd peak/late stance normal GRF. At 8.5°, the 2nd peak of the normal GRF was significantly greater than level, and at 21°, the 2nd peak of the normal GRF was significantly lower than 8.5° and less than that of level walking [18].

McIntosh et al. reported very similar results, however, they found a significant difference in early stance normal GRF as well [19]. Normal GRF increased with grade, with the biggest difference being between 5° and 8°[19]. As noted earlier, McIntosh et al. did not control for speed and subjects walked at faster speeds during the incline trials. Again the exact threshold at which differences in speeds will affect GRF is not known, yet it is still important to recognize the 5% increase in walking speed as a possible contributor to the increased normal GRF during incline walking.

McIntosh also found that as the incline increased, so did the corresponding AP GRF [19]. There was not much change between 5°, 8°, and 10°, but the AP GRF were noticeably larger than the level walking trials. One may expect to see a reduction in braking force with incline, corresponding with the increase in propulsive force necessary to propel the body forward; however, McIntosh et al. did not report this. Lay et al., on the other hand, did find that incline walking resulted in less breaking force. They also reported an increase in propulsive force with incline walking, which was not reported by McIntosh [19, 68]. These discrepancies within the literature and the lack of research discussing ML GRF during incline walking supports the need for more research focused on incline kinetics.

Net Muscle Moments-Level

Just as the sum of the forces produce linear acceleration in an object (∑Force= mass x acceleration), the sum of the moments produce an angular acceleration in an object around an axis of rotation (∑Moment = moment of inertia x angular acceleration). The net muscle moment (NMM) is calculated using inverse dynamics by adding all the moments within a joint together. This process is described in greater detail in Appendix A. It is important to note that biomechanists often report internal and external moments. External moments are the moment due to forces and accelerations about a joint. For example, during knee flexion, the external moment (flexor moment) describes the torque caused by GRFs. The internal moment (extensor/resultant moment) describes the torque required by the quadriceps to stabilize the knee [69].

The following are descriptions of the internal NMM within the hip, knee, and ankle joints during a normal gait cycle. In lean persons, a typical hip net muscle moment pattern has an extensor moment for the first half of stance, which assists in preventing the knee from collapsing and decelerates the forward rotating trunk. During the later part of stance, the hip experiences a flexor moment, followed by and extensor burst to decelerate the forward leg [10]. The knee NMM produces a relatively consistent pattern in lean persons as well. During early stance, an initial knee flexor moment occurs, followed by an extensor moment necessary to resist knee flexion as a person’s body weight shifts over the leg. A flexor moment occurs just before and after toe-off, and at about 65% of stride (right after toe-off), a knee extensor moment occurs in an effort to decelerate the backward leg and minimize heel rise. The knee also experiences a flexor moment just prior to heel strike in order to decelerate the swinging leg prior to the next heel contact [10]. It is important to note that the NMM experienced during the swing phase of walking are small and most studies focus on those occurring during stance. Lastly, during the % stance of level walking, the ankle has a small dorsiflexor moment followed by an increasing plantarflexor moment [10]. Although the NMM experienced at the hip and ankle are important in gait pathologies, I will focus primarily on knee NMM, as they are most pertinent to musculoskeletal injury, particularly in obese persons.

NMM have been hypothesized to be much greater in Obese persons as a direct result of their increased mass [56]. This has clinical importance as NMM are associated with increased joint loading and risk of musculoskeletal injury [74]. Browning et al. reported similar kinematics and greater GRF in obese vs. lean subjects, which resulted in significantly greater absolute NMM [56]. The hip, knee, and ankle NMM decreased significantly at slower walking speeds in both cohorts [56]. This suggests that slower walking speeds reduce the NMM and thus, most likely the loads across the joint.

These findings are not consistent with those of DeVita et al. and Lai et al., who reported similar joint moments between lean and obese persons [14, 60]. For instance, while walking at 1.52 m/s, obese and lean subjects experienced peak knee torques of 64N and 66N, respectively. This discrepancy is most likely due to differences in knee kinematics reported in DeVita's subjects. DeVita et al. reported that obese subjects walked with less knee flexion during stance, adopting a "quadriceps avoidance" type gait. Browning, on the other hand, reported no differences in knee joint kinematics between lean and obese subjects. This may be a result of differences in body mass between the obese subjects used in Browning's vs. DeVita's studies. As discussed earlier, DeVita and colleagues included a much larger range of BMIs than Browning. As a result, DeVita's subjects had ~16% greater body mass than the subjects in Browning's study [56, 60]. There may be a threshold, where above a certain level of adiposity, gait characteristics change. For instance, ~50% of DeVita's obese subjects had a BMI > 40 kg/m^2 [60]. These same subjects also experienced the lowest average absolute knee NMM during stance. Therefore, there may be a critical point (body mass) where individuals may begin to alter their gait in an effort to reduce knee-joint loads.

Additionally, obese participants in DeVita's study experienced smaller knee extensor torque while walking at their self selected speed (1.29m/s) compared to the standard speed of ~1.50m/s [60]. This is consistent with other studies confirming that slower speeds result in lower NMM, particularly within the knee joint [11, 75]. At slower speeds, less quadriceps recruitment occurs because the knee joint is more stable. Therefore, DeVita's findings support the use of slower speeds in exercise prescription for obese individuals.

It has previously been suggested that slower walking speeds are positively correlated with decreased external knee adduction moments [65]. However, in 2004 Mundermann et al. reported that walking speed only explained 8.9% of the variation in maximum knee adduction moment [57]. Both the subjects with varying levels of OA and the control group experienced a wide range of maximum knee adduction moments, and the relationship between maximum knee adduction moment and walking speed ranged from positive to negative associations. This suggests that slower walking speeds may not necessarily lead to smaller loads on the medial compartment of the knee [57]. However, it has been recently suggested that speed is in fact a contributing factor to knee adduction moments [56].

Browning and colleagues in 2007 were the first to study knee adduction moments in obese persons during treadmill walking. They calculated the peak external knee adduction moment attributable to the ML GRF by using the first peak of the ML GRF and the corresponding moment arm distance from the knee joint marker to the treadmill belt surface [56]. Similar to their knee flexion/extension results, the peak external knee adduction moment increased with walking speed and was significantly greater in the obese versus normal weight subjects [56]. This suggests that obese persons have larger compressive loads on the medial compartment of the knee joint and therefore may have a greater risk of developing a musculoskeletal injury.

A recent study by Segal and colleagues aimed to determine whether the distribution of adiposity effected medial joint loading [76]. After controlling for weight, neither body shape (excess abdominal adiposity vs. excess thigh girth) was an independent predictor of the peak external knee adduction moment [76]. Interestingly, they found that regardless of the thigh or abdominal fat distribution, weight alone seemed to be the strongest predictor of external knee adduction moments. In other words, the presence of obesity alone seems to increase loading within the medial compartment of the knee, independent of body fat distribution.

Additionally, Karamanidis completed a study using older and younger adults (~64 years and ~28years, respectively). Both cohorts ascended a three step ramp that was 11.9° at selected speeds, although the authors report similar velocities between groups

[77]. Compared to the younger group, the older subjects showed higher knee adduction and knee internal rotation moments during stance phase [77]. These results suggests that obese persons experience greater knee adduction moments, similar to that reported in older adults. As discussed earlier, greater external knee adduction moments indicate higher mechanical loads at the medial compartment of the knee, which has been correlated with an increased risk of medial compartment knee OA [78-79]. This indicates that obesity prevention is imperative to avoiding the development of greater medial loads within the knee often seen in older adults.

Net Muscle Moments-Incline

Very little data exists on the kinetics of walking on an incline in lean persons. Generally, at heel strike a hip external extensor moment occurs, followed by an external flexor moment during late stance. Although Lay et al. also reported that the hip moment pattern was considerably different during incline walking compared to level walking, both McIntosh et al. and Lay et al. report that the magnitude of peak hip extensor and flexor moments were larger with increased incline [19, 68]. Yet Lay and colleagues reported no differences in knee NMM patterns between conditions [19, 68]. McIntosh and colleagues, on the other hand, found significant differences in the initial knee extensor moment between level and 10° [19, 68]. Clearly more research is necessary to clarify why Lay et al. reported no significant differences in knee flexor moments at an incline twice the degree reported in McIntosh's study. Lastly, both authors reported a progressive and significant increase in peak magnitudes of ankle planterflexor moments with steeper inclines, although the general pattern remained similar to that of level walking [19, 68]. McIntosh et al. reported a unique finding, where walking at a 5° incline resulted in a much lower ankle plantarflexion moment compared to level, 8°, and 10° [19]. This is interesting because it suggests that a positive relationship between incline and NMM does not exist.

An interesting study by Schwamender et al. in 2005 aimed to quantify the effects of walking speed on knee joint forces in lean persons. Using an 18° incline, subjects were asked to walked at set step lengths (0.46, 0.56, and 0.69 m) and cadences (80,100,120 steps per min) [80]. Walking speed (i.e. increased velocity) affected lower extremity joint loading significantly. Schwamender and colleagues calculated the patellofemoral compressive forces using a two-dimensional, quasistatic, rigid-body knee model refered to as 'Plakmos'. The patellofemoral compressive forces increased with both faster cadences and larger step lengths. However, the increase in patellofemoral compressive force with step length was much greater than that of increased cadence (130% vs. 30%, respectively) [80]. Thus, it may be better for individuals to walk uphill with shorter step lengths and faster cadences. It is important to note that a consequence of increasing the cadence and decreasing the step length is an increased number of loading cycles. It is unclear whether the unloading benefit of shorter step lengths outweigh the disadvantage of increased number of loading cycles. Clearly future studies are necessary to clarify and expand the knowledge available on the biomechanics of walking in both lean and obese persons.

Energetics

As discussed earlier, obesity is due, in part, to an imbalance of energy intake and expenditure. Physical activity is considered an essential tool in weight management and walking is often recommended as a form of exercise for obese persons [7]. Thus, understanding the energetics of level and slope walking is necessary for proper exercise prescription, particularly in the obese population. The energetics of walking is commonly calculated via energy expenditure (EE) and % VO_2max, and measured via net metabolic rate (NMR).

EE is a measurement frequently used in exercise prescription, as it is highly relatable to most individuals. In lean persons, the EE for women and men averages 2118 and 2626 k/cals/day, respectively. The average EE for obese women and men is substantially greater than lean persons (2505 and 3152 kcals per day, respectively) [81]. Additionally it has been reported in lean subjects that EE increases in a curvilinear manner (relationship does not follow a straight line) with speed. A similar relationship exists between EE and walking on an incline. In 1960 Bobbert et al. found that while walking at 0.75m/s, EE nearly doubled after increasing the gradient from 0° to 8°. Similarly, one study found that subjects who carried an extra 30 kg expended ~37% more kilocalories (91 vs 124 kcals) than those not carrying the 30kg [82]. This further supports the notion that increased mass is associated with increased EE during walking.

Another way to interpret the energetics of walking is through %VO_2max. Obese persons generally have lower maximal oxygen consumption levels, indicating that they are less able to transport and utilized the oxygen within the body during exercise. The VO_2max for obese and lean persons averages around 26 ml/kg/min and 38 ml/kg/min, respectively [67]. Similar to EE, %VO_2max increases significantly with incline walking. Lafortuna et al. reported a ~24% greater VO_2 in obese subjects while walking at 1.0m/s and 2.3° vs. 1.0m/s and 0° [83]. Lean subjects, on the other hand, experienced no significant differences in VO_2 at these same speed grade combinations. This suggests that obese persons may react differently to incline walking compared to lean persons and that modest inclines may be a useful tool in exercise prescription for weight loss and management.

Typically, the NMR (W/kg) is calculated by subtracting the standing metabolic rate from the exercise metabolic rate. It has been suggested that obese persons have ~20% greater gross NMR compared to lean individuals [84]. However, more recent literature argues that obese persons only experience ~10% greater NMR during walking compared to lean persons [85]. This is similar to Browning and colleagues, who found that obese subjects had a ~11% higher NMR (2.81 vs. 2.54 W/kg) compared to lean subjects across all walking speeds [67]. Additionally, NMR increased with walking speed (0.75-1.75m/s) in both obese and lean subjects [67].

Walking on an incline provides an additional way to increasing caloric expenditure. In 2008, LaFortuna et al. aimed to compare the energetics of walking in obese and lean women using varying speed and grade combinations. With speed held constant and a 4% increase in incline, NMR increased by 39% in obese subjects (2.34 vs. 3.26 W/kg) and 35% in lean subjects (2.21 vs. 2.99 W/kg) [83]. Similar to Browning et al., LaFortuna reported that the NMR increased by 63% in obese subjects during faster level walking trials (0.6 vs. 1.0m/s) [83]. Although increases in speed seem to have a greater effect on NMR in both lean and obese persons, it is important to remember the larger loads associated with increased speed [10-11]. Therefore, a safer exercise prescription may include moderate inclines with slower speeds.

Additionally, it is important to note the relationship between muscle work and energetics. Many models have been made in an effort to calculate the energetics of muscle contractions. For instance, many mathematical models have been developed in order to examine contraction energetics at the level of individual cross bridges. However, due to some limitations with those mathematical models (many unknown parameters), phenomenological models are more recently used. Phenomenological models express mathematically the results of an observed phenomena without paying detailed attention to its fundamental significance [86]. To estimate the total energy used by each muscle during contraction, these models include the rate of heat production and mechanical power [87]. The equation used to estimate metabolic cost (M) is as follows: M = activation and maintenance heat rate x basal metabolic rate x shortening heat rate x mechanical work rate [87]. Recently there has been a push for using simulation models to estimate the energetics of walking. However, obtaining values for the parameters listed in the previous equation is difficult, as few of them are measured directly in humans [87].

The inverted pendulum model is more traditionally used to describe the energetics of walking. Similar to a swinging pendulum, the swinging leg is thought to be metabolically inexpensive, as the motions are mostly passive and require little muscle activation. Results from Griffin et al. indicate that for walking at moderate speeds, the net metabolic cost of walking can be largely explained by the cost of generating muscular force during the stance phase, which is thought to prevent the leg from collapsing [88].

For instance, they found that across a variety of speeds, the net metabolic rate increased in direct proportion to the active muscle required to generate force and the rate of generating this force [88].

Conclusion

With the continual rise of adult obesity, it is imperative that we further understand how to properly prescribe exercise to these individuals. Brisk walking is a commonly recommended mode of physical activity for obese adults [7]. However, recent literature indicates that obese adults experience greater NMM, within all three lower extremity joints, while walking at faster speeds [56]. We also know that walking at an incline increases energy expenditure, although it too has been associated with increased NMM [19, 68]. It is important to remember, however, that these studies only used lean individuals who walked at similar speeds, regardless of the degree of incline. Thus, there may be a trade off effect that exists when walking at slower speeds up a moderate incline. In other words, it may be that slower speeds counteract the increased NMM reported during incline walking.

CHAPTER III

METHODS AND PROCEDURES

Subjects

Subjects were recruited using electronic sources in the Fort Collins area. In order to be included in the study, subjects must have been between the ages of 18-45 years old and had no known cardiovascular disease, pulmonary or metabolic disease signs and/or symptoms and no known disease or joint problem that would be exacerbated by exercise. Twelve obese adult volunteers (7 female and 5 male) participated in this experiment. Subjects were in good health, sedentary to lightly active (< 3 hours of physical activity per week), not taking any medications known to alter metabolism and body mass stable (< 2.5 kg net change during the previous 3 months). Physical characteristics of the subjects are shown in Table 1. Subjects gave written informed consent that followed the guidelines of the Colorado State University human research institutional review board.

Table 1: Physical characteristics of participants. Values are mean (SD).

Subject Characteristics	
Age (years)	27 (5.51)
Height (m)	1.73 (0.13)
Body Mass (kg)	100.53 (15.72)
BMI (kg/m^2)	33.43 (2.35)
Percent body fat (%)	37.99 (7.45)
Lean body mass (kg)	63.15 (16.70)
VO_2max (ml/kg/min) VO_2max (L/min)	29.63 (5.43) 3.03 (1.01)
Standing metabolic rate (W/kg)	1.21 (0.13)

Experimental protocol

Each subject completed three experimental sessions. During the first session, which followed a 12-hour fast, subjects underwent a physical examination and body composition was measured. We also recorded anthropometric characteristics required to determine lower extremity body segment parameters [89]. Finally, subjects completed a standard graded exercise stress test to determine maximal oxygen uptake (VO_2max). During the second and third sessions, which each followed a 4-hour fast, we collected metabolic and biomechanics data as subjects stood and walked (with shoes) at 16 speed/grade combinations (Table 2). Trials were 2 or 6 minutes in duration and subjects were allowed 5 minutes rest between trials. Prior to data collection during the second session, we familiarized the subjects to the treadmill by having them walk (0° grade) at a self-selected speed for ~10 minutes.

Table 2: Speed/Grade Combination. B denotes biomechanics data was collected and B+M denote biomechanics and metabolic data were collected. B and B+M trials were collected for 2 and 6 minutes, respectively. The plus (+) and minus (-) signs denote uphill and downhill walking trials, respectively.

Angle (degrees)	Speed (m/sec)					
	0.50	0.75	1.00	1.25	1.50	1.75
- 3				B+M		
0	B		B	B+M	B+M	B+M
+ 3			B+M	B+M	B+M	
+ 6		B+M	B+M	B		
+ 9	B+M	B+M		B		

Assessments

Physical Health and Activity

Each subject completed a health history form and were interviewed and assessed by a physician. Blood was drawn to test for normal metabolic function. Resting levels of thyroid-stimulating hormone and blood cell count were measured and confirmed to be within normal ranges. Physical activity levels were assessed via a questionnaire and only individuals with < 3 hours of moderate-vigorous physical activity a week were invited to participate.

Body Composition

We measured each subject's body composition using a dual X-ray absorptiometry (DEXA; Hologic Discovery, Bedford, MA). We determined percent body fat and percent lean mass for the entire body and 3 regions of interest: thigh, shank and foot. Regions of interest were manually identified using the DEXA software and similar techniques used in previous studies =. The thigh segment proximal end was defined as a line between the superior border of the iliac crest and the inferior border of the coccyx, excluding the pelvis. The thigh segment distal end and shank proximal end was a line between the femoral condyles and the tibial plateau. The shank segment distal end was a line between the inferior aspects of the medial and lateral malleolus. The foot segment was the remainder of the leg below the distal endpoint of the shank.

Maximal Oxygen Uptake

We used a modified Balke treadmill protocol to determine each subject's VO_2max [48]. Subjects were familiarized with the treadmill and the Borg Rating of Perceived Exertion scale (6-20 scale) [90]. A 12-lead electrocardiogram was used to monitor heart function. Each subject's heart rate and blood pressure was measured in the supine, sitting, and standing position to test for orthostatic intolerance. Subjects warmed up for ~5 minutes after which we slowly increased the speed of the treadmill until subjects reported an RPE indicative of moderate intensity exercise (~11). Treadmill speed was then held constant and the grade increased by 1% every minute. The subjects were encouraged to continue to exhaustion. During the test, normal physiological responses to exercise were measured by recording heart rate, blood pressure, and RPE every 3 minutes. Heart function was monitored by a physician. We determined maximal oxygen consumption via open circuit respirometry (Oxycon Mobile, Yorba Linda, CA), with expired gas data averaged over 30-second block intervals.

Biomechanical Measurements

To record biomechanics data we used a three-dimensional motion capture system (Motus 9.0, Vicon, Centennial, CO) and a dual-belt, inclinable, force-measuring treadmill (Bertec, Columbus, OH). We placed lightweight reflective spheres in accordance with the modified Helen Hayes marker set to identify anatomical landmarks and delineate lower extremity segments [91]. Markers were placed on the sacrum, anterior superior iliac spine, mid-thigh (femoral wand), femoral epicondyle, mid-shank (tibial wand), lateral malleolus, second metatarsal head, and calcaneous of each leg. Marker trajectories were recorded at 60 Hz using eight optoelectronic cameras. Ground reaction forces (GRF) were recorded at 1200 Hz by force platforms embedded under each treadmill belt. Kinematic and kinetic systems were synchronized. Data was collected for 30 seconds during the final minute of each trial.

Raw coordinate and kinetic data was smoothed using a fourth-order, zero lag recursive digital Butterworth filter with a cutoff frequency of 5Hz and 12Hz, respectively. We used vertical GRF data and a threshold of 15N to determine heel strike and toe off for each leg and computed temporal characteristics of each trial using custom software (Matlab, v12.0, Mathworks, Natick, MA). To determine the thigh and shank body segment parameters, we used the DEXA data to estimate thigh and shank mass and used the regression equations provided by Durkin and Dowling to determine radius of gyration [8, 92]. Segment mass and radius of gyration were used to calculate frontal plane moment of inertia. We used frontal plane values to represent sagittal plane moment of inertia of the thigh and shank [93]. Foot segment parameters were estimated using the Motus software (v9.0, Vicon/Peak, Denver, CO). Lower extremity kinematic and kinetic variables (joint angles and net muscle moments) were computed using Motus software and were normalized to represent a percentage of the stance or stride. Step width was determined as the distance between the mid-stance center of pressure of the right and left leg during consecutive steps. We calculated the mean of each variable of interest of the right leg over 5-25 strides at each speed/grade combination for each subject and the mean across subjects for each speed/grade combination.

We have elected to report the absolute (Nm) rather than normalized joint moments (Nm/kg/m). Although normalizing is useful when comparing across groups, it does not address the actual loads placed on the joints. Given that joint articulating surface area does not scale with body mass, we feel this approach is justified. In a recent study by Ding et al., body mass was 48% greater in the obese subjects versus normal-weight subjects, whereas tibial articulating surface area was only 8% larger [94]. Thus, obese adults have increased mechanical loads (e.g., stress) across the tibial/femoral articulating surfaces as well as increased rates of loading. Because the development of OA has been associated with increased magnitude and rate of mechanical loading, absolute values seem an appropriate comparison [4, 95].

Energetic Measurements

To determine metabolic rate during standing and walking, we measured the rates of oxygen consumption (VO_2) and carbon dioxide production (VCO_2) using a portable open circuit respirometry system (Oxycon Mobile, Yorba Linda, CA). Before the experimental trials, we calibrated the system and measured standing metabolic rate for 6 minutes. For each trial, we allowed 4 minutes for subjects to reach steady state (no significant increase in VO_2 during the final 2 minutes and a respiratory exchange ratio

<1.0) and calculated the average VO_2 and VCO_2 (ml/sec) for the final 2 minutes of each trial. We calculated net metabolic rate (W/kg) from VO_2 and VCO_2 using a standard equation [96]. We then subtracted standing metabolic rate from the walking values to derive net metabolic rate.

Statistical Analysis

A repeated-measures ANOVA determined how obesity affected temporal gait characteristics, mid-stance joint angles, peak net muscle moments and metabolic rate. If a significant main effect was observed, post-hoc comparisons using the Holm-Sidak method were performed. When data failed the normality test, a multiple comparison Tukey test on ranks was used. A criterion of $p<0.05$ defined significance.

CHAPTER IV

RESULTS

As the purpose of this study was to compare speed/grade combinations that met the criteria of moderate exercise intensity, we report results from five of the speed/grade trials. The trials included: 0.50m/s (9°), 0.75m/s (6°), 1.25m/s (3°), 1.50m/s (0°), and 1.75m/s (0°).

Kinematics

Temporal stride characteristics were significantly different across the speed/grade combinations (*p*< 0.005) (Table 3). As walking speed increased and grade decreased, stride frequency and stride length increased while duty factor (proportion of time per stride that each foot spends on the ground) and double support time decreased. Step width was significantly greater only at the slowest speed.

*Table 3: Temporal stride kinematics. Values are mean (SE). * significant difference between condition and 1.50m/s condition.*

Speed (m/s)	Grade (degrees)	Stride Frequency (Hz)	Stride Length (m)	Stance (% Cycle)	Swing (% Cycle)	Double Support (% Cycle)	Step Width (m)
0.50	9	0.58* (0.02)	0.87* (0.03)	72.6* (0.6)	27.4* (0.6)	44.0* (1.2)	0.188* (0.014)
0.75	6	0.71* (0.02)	1.08* (0.04)	70.3* (0.5)	29.7* (0.5)	40.0* (0.9)	0.170 (0.01)
1.25	3	0.89* (0.02)	1.42* (0.03)	66.9* (0.7)	33.1* (0.7)	34.7* (0.8)	0.168 (0.011)
1.50	0	0.99 (0.03)	1.54 (0.05)	64.5 (0.5)	35.5 (0.5)	30.2 (0.6)	0.161 (0.011)
1.75	0	1.07* (0.02)	1.64* (0.03)	63.6 (0.4)	36.4 (0.4)	27.4* (0.7)	0.167 (0.010)

Although we did not statistically analyze joint angles, it appears that the knee and ankle joint angles differed across all speed/grade combinations but hip joint angles did not (Figure 1). Increasing the incline and decreasing the walking speed resulted in more flexion at the knee during early stance. Mean peak knee flexion in early stance was ~70% greater (39° vs. 23°) at 0.50m/s (9°) vs. 1.75m/s (0°), respectively. A steeper incline resulted in a more dorsiflexed ankle throughout the gait cycle.

Kinetics

Similarly, we did not statistically analyze peak GRFs, as we were primarily interested in NMM. However, the following trends should still be reported. During early and late stance, peak normal GRFs were greater in trials with faster speeds and lower grades (Figure 2A). Peak anterior/posterior (AP) GRFs in early stance were greater in the faster, level speeds and there was no braking force at the steepest incline (Figure 2B). Faster speeds and lower grades had slightly greater medial/lateral (ML) ground reaction forces compared to slower speeds and moderate inclines (Figure 2C). We found that loading rates were ~3x greater during level walking at 1.50m/s vs. 0.75m/s (6°).

Peak net flexor and extensor muscle moments about the hip and knee and peak net plantarflexor and dorsiflexor muscle moments about the ankle were smaller during the slower speed/moderate incline trials and increased as the speed increased and incline decreased (Table 4 and Figure 3). Peak extensor net muscle moments about the knee during early stance were significantly lower when walking at 0.75m/s (6°) vs. both 1.50m/s (0°) and 1.75m/s (0°) (p=0.002). There was a 19% and 28% reduction in peak knee extensor net muscle moment between 0.75m/s (6°) vs. 1.50m/s (0°) and 1.75m/s

(0°), respectively (Figure 3B). The peak knee abduction/adduction muscle moments increased with speed (Figure 3D) (p =0.005). The peak knee abduction moment was reduced ~54% and ~26% when walking at 0.50m/s (9°) and 0.75m/s (6°) vs. 1.50m/s (0°) respectively.

*Table 4: Peak knee flexion/extension, knee abduction/adduction and ankle plantarflexion net muscle moments. Values are mean (SE). * significant difference between condition and 1.50m/s condition.*

Speed (m/s)	Grade (degrees)	Hip Extension (Nm)	Knee Extension (Nm)	Knee Abduction (Nm)	Ankle Plantarflexion (Nm)
0.50	9	44.8 (3.7)*	52.8 (4.2)	38.7 (5.3)*	125.4 (15.1)*
0.75	6	59.8 (3.5) *	49.0 (5.8) *	47.3 (3.9)*	140.0 (11.8)*
1.25	3	101.0 (8.0)	55.2 (6.1)	50.3 (6.1)	175.8 (17.4)
1.50	0	100.3 (8.4)	58.1 (6.8)	59.4 (7.2)	172.6 (18.6)
1.75	0	131.0 (6.9) *	62.8 (7.4)	63.0 (5.7)	179.4 (15.0)

Energetics

Mean VO_2max was 29.6 ml/kg/min and mean standing metabolic rate was 1.2 W/kg. Net metabolic rate was similar during level walking at 1.50m/s vs. walking uphill at 0.50m/s(9°) and 0.75ms(6°), but was 29.8% and 14.4% greater during level walking at 1.75m/s and 1.25m/s (3°), respectively (Figure 4). The mean net metabolic rate for 0.50m/s (9°), 0.75m/s (6°), and 1.50m/s (0°) was 3.67, 3.88 and 3.80 W/kg, respectively. All trials were moderate intensity and required between 49 – 60% of VO_2max. The mean relative aerobic effort for 0.50m/s (9°), 0.75m/s (6°), and 1.50m/s (0°) was ~49, 51, and 50 % VO_2max, respectively.

Figure 1

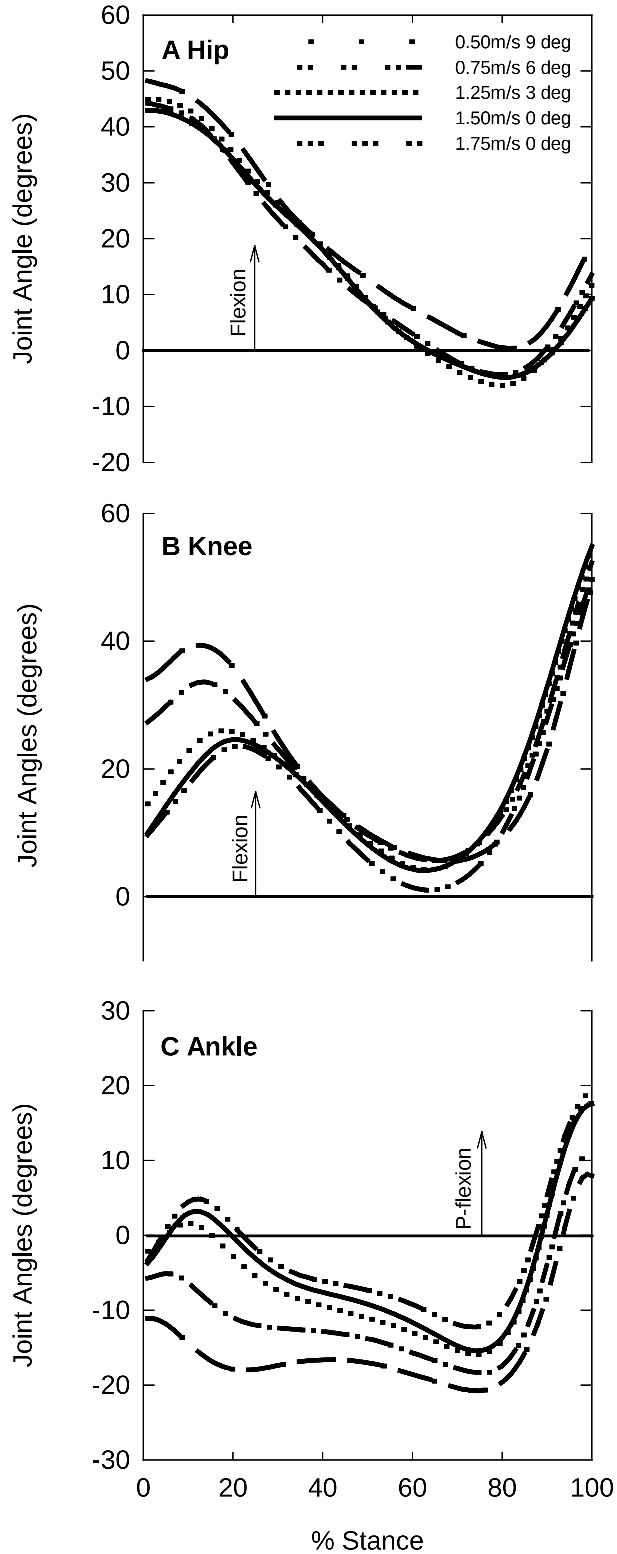

Figure 1. Mean hip (A), knee (B) and ankle (C) joint angles for each speed/grade. Stance begins at right heel strike.

Figure 2

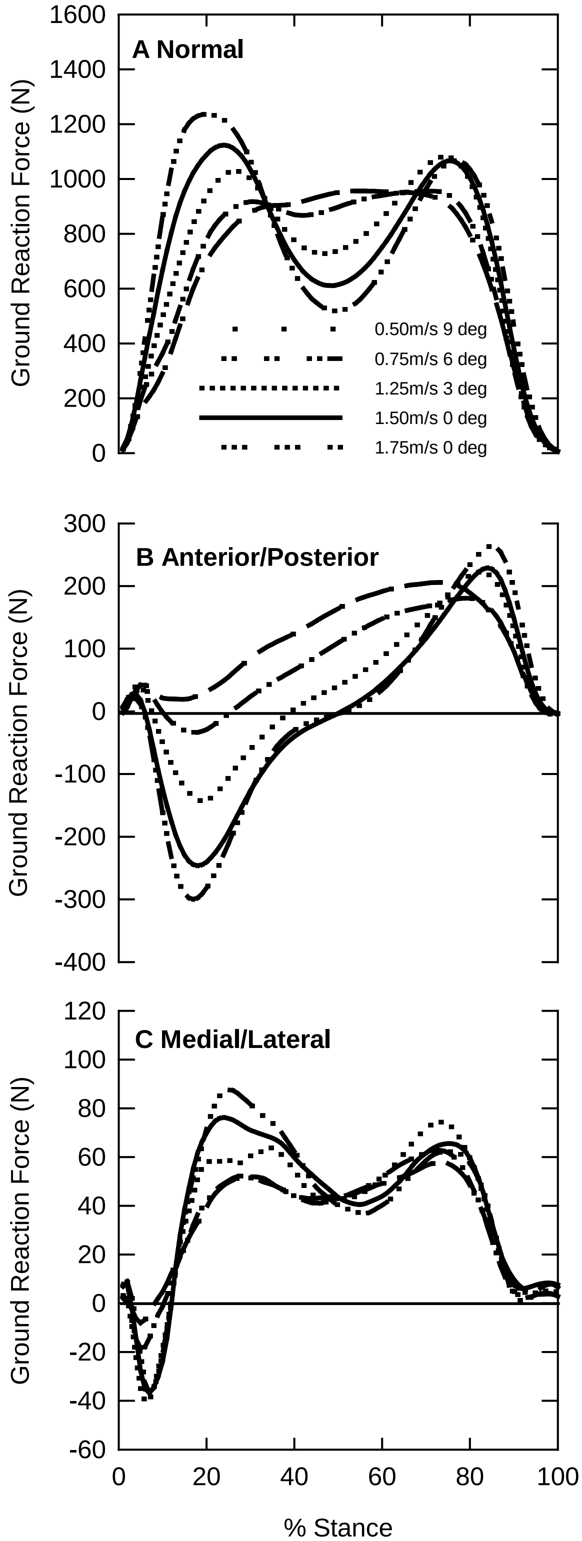

Figure 2. Mean normal (A), anterioposterior (B), and mediolateral (C) ground-reaction forces while walking at all speed/grade combinations.

Figure 3

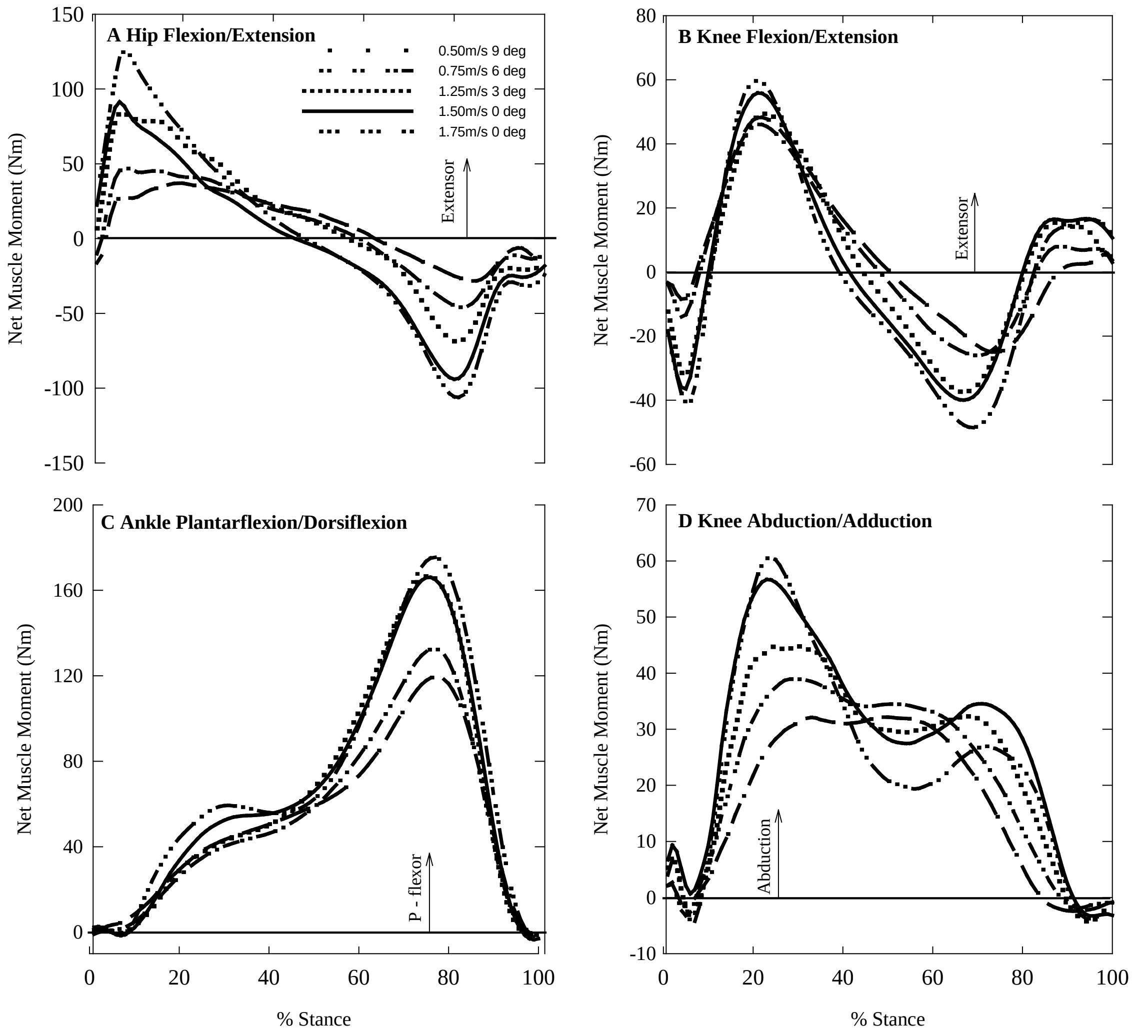

Figure 3. Mean flexion/extension hip (A), knee (B), and ankle (C) net muscle moments during stance while walking at all speed/grade combinations. Positive moments are extensor. Stance begins at right heel strike. Mean abduction/adduction moments at the knee (D) for all speed/grade combinations. Positive moments are abduction.

Figure 4

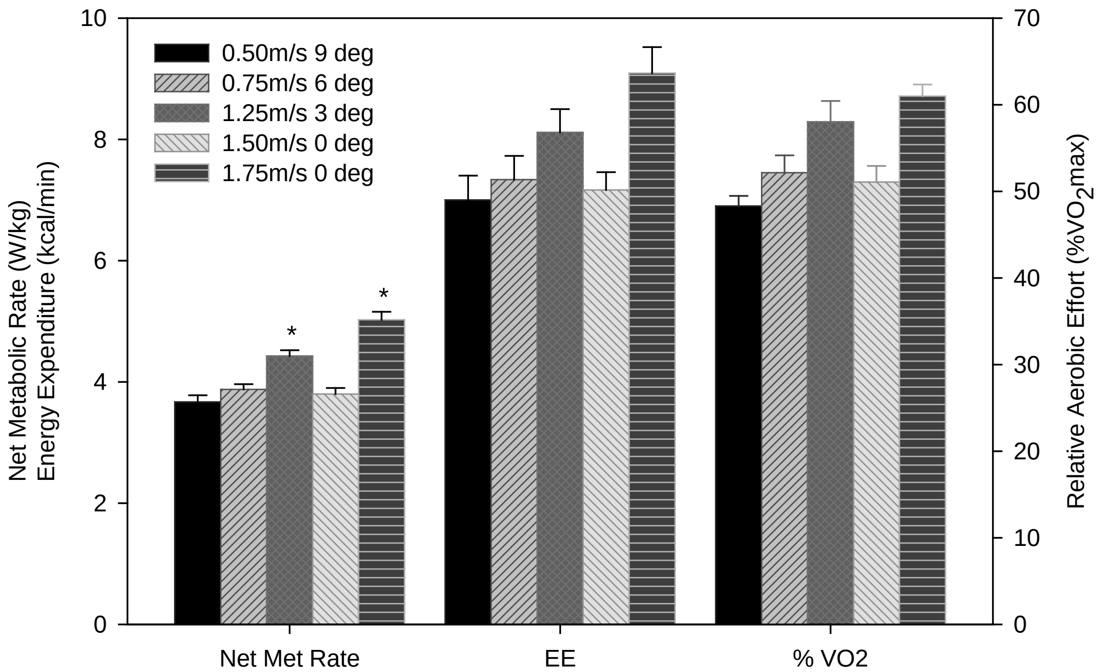

*Figure 4. Mean net metabolic rate (W/kg), energy expenditure (Kcal/min), and relative aerobic effort (%VO_2max) for all trials. * Significantly different compared to 1.50m/s at 9°. Values are means ± SE.*

CHAPTER V

DISCUSSION

Kinematics

Temporal-spatial stride characteristics changed with speed/grade. The temporal stride characteristics we report are similar to previous studies with obese subjects [13, 60]. In our recent study of moderately obese subjects during level walking, stride length and frequency increased with walking speed, while duty factor decreased [13]. A comparison of the results of this study with our earlier level walking results suggests that stride characteristics are dependent on speed and not grade, as has been reported for non-obese individuals [13]. Thus, moderate obesity does not appear to elicit changes in the stride length/frequency relationship during uphill vs. level walking. However, obese adults spend more time in stance and double support than their non-obese counterparts [14]. Although we did not have a non-obese control group, our level walking results support this finding as our values are similar to those previously reported [14].

The observed joint angle patterns are in good agreement with some of the previous studies reporting level walking kinematics of obese persons [13-14, 62]. DeVita and Hortobagyi report that obese adults walk with a more erect posture (i.e. less knee flexion and greater plantar flexion during stance) compared to lean adults [60], while other studies have not supported this finding [14, 62]. We found that obese adults had ~15° of early stance knee flexion during level walking at 1.50m/s, which is similar to the range of motion reported for both non-obese and obese adults [13, 18, 62]. The more extended posture adopted by the participants in the DeVita and Hortobagyi study may be attributable to fact that some of their participants were severely obese [60].

Our results show that uphill walking requires a greater degree of knee flexion during early stance and ankle dorsiflexion throughout stance compared to level walking. The combination of more flexed joints during early stance and greater extension range of motion is characteristic of uphill walking and is necessary to raise the body's center of mass and ensure foot clearance [19]. To our knowledge, no study has analyzed the kinematics of uphill walking in obese persons. We observed similar hip joint angles across the speed/grade combinations. Lay et al. and McIntosh et al. reported that hip flexion increased ~60% at heel strike between level and uphill (~9°) walking [18-19]. This difference is most likely a result of walking speed. At the 9° incline, our subjects walked much slower than the participants in these studies [18-19]. Slower walking speeds result in shorter stride lengths, which limit the range of motion of the hip joint. It is also possible that the obese subjects did not increase pelvic tilt during uphill walking, as has been observed in non-obese individuals [19]. A smaller change in pelvic tilt would reduce the increase in hip flexion during uphill vs. level walking and may reduce the muscle forces required to support the upper body. Future studies that quantify pelvic motion in obese adults during level and uphill walking are needed.

Kinetics

We accept our hypothesis that decreased walking speeds combined with moderate inclines would reduce lower extremity net muscle moments compared to faster, level walking. As walking speed decreased and grade increased, hip, knee and ankle net muscle moments decreased. During level walking, we found hip, knee and ankle net muscle moment patterns (NMM) to be consistent with previous studies of non-obese and obese adults [10, 13, 19]. DeVita and Hortobagyi reported a peak knee extensor moment during early stance of 0.52 Nm/kg when obese subjects walked at 1.50m/s, comparable to the 0.58 Nm/kg reported here [60]. Lower extremity joint moments decrease with speed during level walking [70] and increase with incline when speed is held nearly constant [18-19]. During uphill walking, the hip joint extensor moment increases dramatically, while early stance knee extensor moment and late stance ankle plantarflexor moments have a more modest increase [18-19]. For example, when non-obese adults walked up an 8.5° grade at ~1.2m/s, peak hip and knee extensor and ankle plantarflexor moments increased by 104, 46 and 18%, respectively [18]. Comparing our results with Lay et al. suggests that speed may influence lower extremity muscle moments more than grade, at least for the speed/grade combinations tested.

Net muscle moments provide a proxy measure for in vivo loads across the joints of the lower extremity and have been used by many investigators to estimate these loads [13]. A limitation of this approach is that the NMM does not reflect the role of groups/individual muscles (i.e. knee extensors) that cross a joint but is a resultant (net) moment at the joint based on all muscles that cross that joint. Thus, there may be many combinations of agonist/antagonist muscle force production that produce a similar NMM at a particular joint. When walking uphill vs. level at a similar speed there is an increase in co-contraction of the knee flexors and extensors [20]. Lay et al. report that when walking up a 21° slope vs. level walking, Vastus Medialis and Biceps Femoris muscle activity amplitude increased 451 and 347%, respectively and activity duration increased from ~20 to 40% of stride for both muscles [20]. Lay et al. state that the much larger hip extension moment required to walk uphill requires increased activation of the bi-articular hip extensor/knee flexor muscles which in turn requires an increase in knee extensor muscle activity to maintain an extensor moment at the knee [20]. Importantly, these increases in muscle activity did not result in a significant increase in knee flexion/extension NMM (although the peak was 85% greater) and suggests that the changes in NMM are not reflective of the changes in joint loading during incline walking.

While we acknowledge the possibility that the decrease in knee NMM we report during incline vs. level walking may not reflect a decrease in joint loading, our subjects were walking on a more gradual incline at much slower speeds than those reported by Lay [18] and muscle activity decreases with speed [21]. The finding that hip extension moments were much smaller during our incline vs. level walking trials suggests that the bi-articular hip extensor/knee flexor muscles may not be required to assist with hip extension, which would reduce the co-activation of knee flexion/extension muscles and reduce the need for increased knee extensor torque production. It is also important to note that an additional benefit of the slower speed/incline is that rates of loading are much slower, given the smaller peak normal ground reaction forces and longer duration of stance. Loading rates (slope of vertical ground reaction force during early stance) [97] may lead to the initiation or progression of cartilage damage and the development of musculoskeletal injury [98-99]. As mentioned previously, we found that loading rates were ~3x greater during level walking at 1.50m/s vs. 0.75m/s (6°). This suggests slower walking speeds may reduce the risk of musculoskeletal injury. Clearly, future studies that quantify muscle activity and use musculoskeletal models to estimate joint forces will lead to a better understanding of the relationship between NMM and joint loading during incline walking.

Slower walking up moderate inclines reduced the peak early stance knee abduction NMM compared to faster level walking. Greater abduction/adduction moments at the knee are associated with increased medial compartment knee loading, varus malalignment and a greater risk of OA progression [100]. Our abduction/adduction NMM pattern is consistent with that reported in Foroughi et al. [101]. By decreasing the speed from 1.75m/s to 0.50m/s and walking up an incline of 9°, the peak knee abduction/adduction moment decreased by 63%. This finding suggests that the distribution of load on the medial compartment is reduced during slower, incline walking versus faster, level walking.

Energetics

We accept our hypothesis that slower walking up moderate inclines provides similar physiological stimulus and energy expenditure vs. faster, level walking for moderately obese persons. Our results are in agreement with other studies which found that inclines increase the metabolic rate of walking [83, 102] while slower speeds decrease the metabolic rate [8, 83]. Our biomechanics results provide insights into the metabolic cost of walking. The primary determinant of the metabolic cost of walking is muscle work required to support and move the center of mass as well as swing the legs [87]. Although we were not able to quantify individual muscle work, our joint moment data allows the determination of joint work (integral of joint power) which is related to metabolic cost [103]. During incline walking, lower extremity muscles must perform positive mechanical work to increase the gravitational potential energy with each step and more of this work is performed by proximal muscles/joints (hip and knee) compared to level walking [87, 104]. Other factors that influence metabolic rate, including co-contraction and elastic energy storage/return must account for the difference between joint work and metabolic rate. As noted above, co-contraction of muscles that cross the hip and knee increase during incline walking and this would increase metabolic rate without changing measured joint work.

During level walking, considerable elastic energy is returned, primarily via the Achilles tendon [87]. Elastic energy storage/return is smaller at slower speeds [105] and the contribution of this energy to the total work required is less when walking on an incline vs. level walking at the same speed [106]. This would suggest that the contribution of elastic energy storage/return would be less during the slower speed/moderate incline trials. However, it is also possible that obese individuals have an altered ability to store/return elastic energy that would be affected by speed but not grade or vice versa. Muscle activity to control upper body posture may also play a role in the energetics/mechanics relationship. Individuals lean forward when walking uphill and this may incur a metabolic cost without changing lower extremity NMM. Future studies that are able to identify the individual muscle contributions to the metabolic cost of walking in obese and non-obese will assist us in understanding the influence of excess body mass and whether obese adopt a walking gait that conserves metabolic energy.

Relevance for Exercise Prescription

Our results suggest that slower, uphill walking may be a good form of exercise for moderately obese adults, particularly those who do not have lower extremity joint pain. For instance, a 100kg person walking on a 6° incline at 0.75m/s for 30 minutes will expend approximately 220 gross kilocalories, which is slightly more than what that person would expend walking at 1.50m/s on a level surface for the same amount of time. Walking at slower speeds also reduces the perceived exertion of the exercise [9], which may result in increased activity time and adherence. Finally, it is important to note this is a treadmill specific exercise prescription because walking uphill outdoors typically requires downhill walking as well, which is known to increase joint loading [18-19]. Thus, our findings support an exercise prescription specific to treadmill and not overground walking.

Limitations

There are a few limitations of our study that should be noted. Marker placement and movement was a limitation due to our subject's excess adiposity. We did not collect EMG data and therefore were unable to analyze the muscle activation pattern that occurs during level and incline walking in obese adults. Another limitation is the lack of a control (lean) group, and therefore our findings are limited to a single group comparison. Lastly, no measures were made to quantify fatigue. It may be possible that as subjects tired, they altered their gait. Knapik et al found that when subjects were fatigued, they walked with increased trunk flexion, shorter stride time, and reduced stride length [107]. Although we report differences in stride length, these are most likely due to changes in walking speed. Thus, because our kinematic results differed from those reported in fatigued subjects, it is probable that our subjects were not fatigued. Additionally, the intensity of our walking trials were relatively low, and therefore it is very unlikely that subjects experienced fatigue. Furthermore, it has been shown that individuals do not report feeling fatigued until walking for greater than 60 minutes at their self selected speed [108]. This is substantially longer than what our subjects walked at and thus, it is likely that fatigue did not affect our results.

Conclusion

We found that walking at slower speeds and moderate inclines resulted in smaller net muscle moments across the lower extremity joints. With the exception of the fastest walking trial, the energetics of slope walking was similar across most speed/grade combinations. The results suggest that walking at slower speeds and moderate inclines may reduce loads across the lower extremity joints while ensuring adequate aerobic stimulus for weight management.

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APPENDIX A

INVERSE DYNAMICS

Inverse Dynamics

Inverse dynamics is a technique commonly used within the biomechanics field that enables researchers to calculate the net muscle moments (NMM) and net joint reaction forces using available anthropometric measures, kinematics, and external forces. NMM are calculated using gravitational forces, ground reaction forces (or joint reaction forces if computed for a joint other than the most distal), and distal NMM. Using these forces, moment of inertia (I), and angular velocity (α), we can solve for the proximal joint moment. In order to calculate NMM and net joint reaction forces, a number of assumptions must be made.

1. The mass of each segment is fixed and is located at its center of mass
2. All joints are considered ball and socket joints
3. The mass moment of inertia about its center of mass is constant during movement
4. The segment length remains constant during movement
5. Assume a frictionless joint with no ligaments or other passive forces

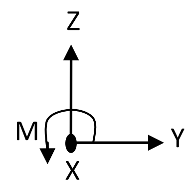
Below are equations necessary to perform inverse dynamics, followed by a free body diagram (following page) depicting the process.

$$\sum F = ma \qquad M = F \times D \qquad \sum M = I\alpha \qquad I = \sum_{i=1}^{N} m_i r_i^2$$

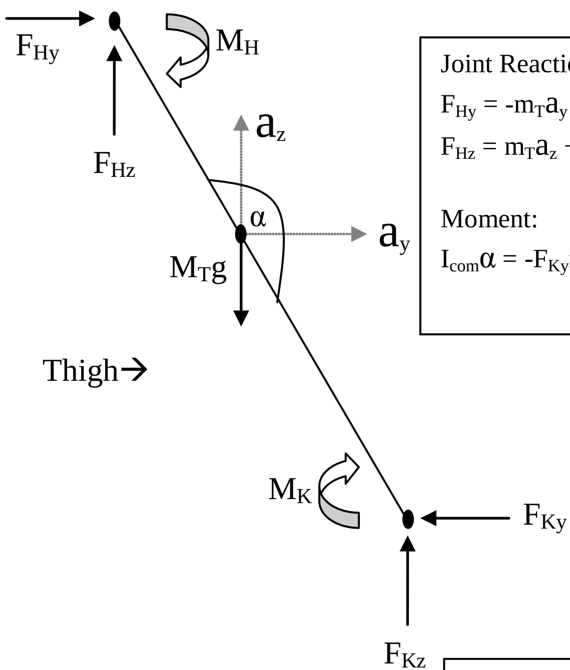
Where: F = force, m = mass, M = moment, D = moment arm, I = moment of inertia, α = angular acceleration, r = distance between axis and rotation mass , a = acceleration, N = number of mass segments

Free Body Diagram

Planar, lower extremity



Hip



Joint Reaction Forces:

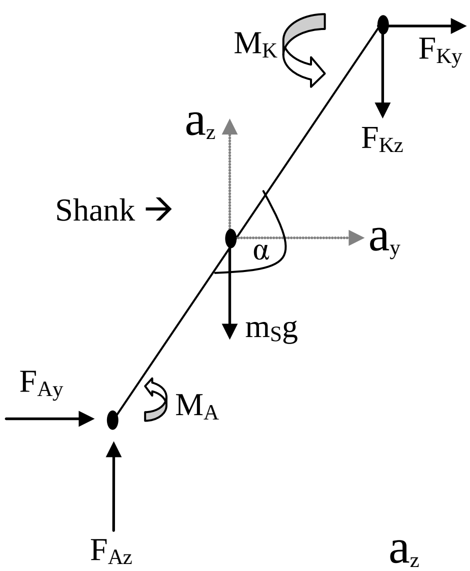
$F_{Hy} = -m_T a_y + F_{Ky}$

$F_{Hz} = m_T a_z + m_T g - F_{Kz}$

Moment:

$I_{com}\alpha = -F_{Ky} r_{comy} + F_{Kz} r_{comz} + M_K - F_{Hz} r_{Hz} - F_{Hy} r_{Hy} + M_H$

Knee



Joint Reaction Forces:

$F_{Ky} = m_s a_y - F_{Ay}$

$F_{Kz} = m_s a_z + m_S g - F_{Az}$

Moment:

$I_{com}\alpha = -F_{Ay} r_{comy} + F_{Az} r_{comz} - M_K - F_{Kz} r_{Kz} - F_{Ky} r_{Ky} + M_A$

Joint Reaction Forces:

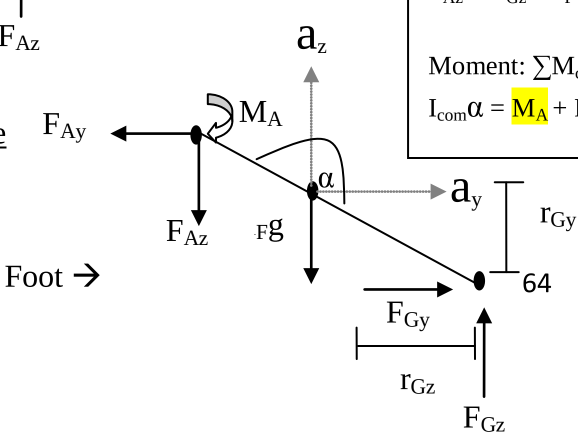
$F_{Ay} = m_F a_y - F_{Gy}$

$F_{Az} = F_{Gz} - m_F a_z - m_F g$

Moment: $\sum M_{com} = I_{com}\alpha$

$I_{com}\alpha = M_A + F_{Az} r_{Az} + F_{Ay} r_{Ay} - F_{Gy} r_{Gy} - F_{Gz} r_{Gz}$

Ankle



Note: Calculations flow from distal to proximal ends, using net forces and moments from a distal FBD to solve the more proximal FBD.

It is important to note that the inverse dynamics process described above is 2D, and therefore does not take into account the 3 dimensional world in which gait occurs. The motion capture system which we used in our study (Vicon Motus v9.0) calculated the joint reaction forces and moments of the lower extremity joints in a slightly different way (3D) than traditionally seen in 2D. In addition to the global reference system, 3D inverse dynamics includes a local reference system (LRS) for each individual segment. When studying movement in 3D, six independent coordinates need to be described. These coordinates include three Cartesian coordinates (X, Y, and Z) and three angles of rotation, often quantified as Euler angles. Euler angles are important because they are used to determine the angular velocities and accelerations of each segment, which is then used to calculate joint moments. Rather than starting with Newton's Laws of Motion equations, Vaughn and colleagues elected to use angular momentum prior to Newton's second law to calculate joint reaction forces and moments. An advantage of using 3D inverse dynamics is that it enables biomechanists to calculate internal and external rotation, thereby creating a more complete analysis of movement.

APPENDIX B
INFORMED CONSENT

Consent to Participate in a Research Study
Colorado State University

TITLE OF STUDY: Biomechanics and Energetics of Gradient Walking in Adults

PRINCIPAL INVESTIGATOR: Ray Browning, PhD. 970-491-5868

WHY AM I BEING INVITED TO TAKE PART IN THIS RESEARCH? You are a sedentary or moderately physically active man or woman between the ages of 18-45 years and you do not have any major health problems and are not pregnant. Our research study is designed to determine the effects of level, uphill and downhill walking on how many calories your burn and the loads on your leg joints.

WHERE IS THE STUDY GOING TO TAKE PLACE AND HOW LONG WILL IT LAST? The study will be performed in the Human Performance Clinical Research Laboratory (HPCRL) and the Physical Activity (PA) laboratory on Colorado State University's campus. This study will require you to visit the lab three times and each visit will take approximately 60-90 minutes of your time. The total amount of time required for this study will be approximately 3-4.5 hours.

WHAT WILL I BE ASKED TO DO? If you agree to participate in this study, you will be asked to schedule 1 visit to the HPCRL and 2 visits to the PA lab. Each visit will take about 1-1.5 hours of your time. We will try to make your testing appointments as convenient as possible for you.

<u>Visit 1: Pre-study screening and maximal exercise test</u>

- You will be asked about the medical history of you and your family and your level of physical activity.
- You will undergo a standard health and physical exam by a physician.
- We will draw a small amount of your blood from a vein in your arm.
- We will measure your body composition using a DXA machine. This machine is like a large X-ray machine. You will lay on the surface of the machine while a beam passes over your body. This procedure takes a few minutes.
- We will measure your resting EKG (heart function) and blood pressure.
- You will also complete a maximal exercise test (VO_2max test). You will be asked to walk on a motorized treadmill and we will increase the degree of incline every few minutes until you reach exhaustion. We will measure the rate at which you consume oxygen by analyzing the air that you breathe out. This will involve wearing a mask that covers your nose and mouth.

<u>Visit 2 and 3: Treadmill walking</u>

- The second and third visit will be very similar. You will be asked to walk at eight different speed/grade combinations for either 2 or 6 minutes.
- During each 2 minute trial, we will measure the forces you exert against the treadmill belt and your leg movements. We will attach reflective markers to your skin to record your leg movements.
- During each 6 minute trial, we will measure the rate at which you consume oxygen by analyzing the air that you breathe out. This will involve wearing a mask that covers your nose and mouth.
- After each trial, you will be given the option of a 5 minute rest.

ARE THERE REASONS WHY I SHOULD NOT TAKE PART IN THIS STUDY?
If you are not 18-45 years of age, are pregnant, are a regular smoker, use illicit drugs (cocaine or methamphetamine), have a condition that limits your ability to walk or run (e.g. knee pain during walking), or have any diseases that would affect our measurements, we will not be able to include you in the research.

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?
The potential risks associated with this study are similar to those involved in recreational athletics. None of the procedures should cause discomfort. However, if you do experience any discomfort, you may terminate the experiments at any time. There is a risk of falling from the treadmill and injuring yourself. To minimize this risk, you will be instructed in proper safety procedures before the treadmill is turned on. It is very important is to always grab the handrails when the treadmill is starting or stopping. You may feel some mild muscle discomfort or fatigue in your legs for a few days after participating in this study. Other specific risks include:

- <u>Maximal Exercise Testing</u>- There is a very slight chance of an irregular heartbeat during exercise (< 1% of all subjects). Other rare risks of a stress test are heart attack (< 5 in 10,000) and even death (<2 in 10,000). There is a small risk of fainting and fatigue. Finally, exercise can cause fatigue and minor discomfort.
- <u>Body Composition</u> - There is a small amount of radiation exposure associated with the DXA, which is less than 1/20 of a typical chest x-ray. The more radiation one receives over the course of one's life, the more risk of having cancerous tumors or of inducing changes in genes. The changes in genes possibly could cause abnormalities or disease in a subject's offspring. The radiation in this study is not expected to greatly increase these risks, but the exact increase in such risks is unclear.
- <u>Treadmill Testing</u> - You could injure yourself by falling while walking on a motorized treadmill. However, you will be instructed in proper safety procedures before the treadmill is turned on. The most important instruction is to always hold the handrails when the treadmill is starting or stopping. Dr. Browning has six years of experience conducting these sorts of experiments and has never had a subject experience a serious injury. If you sustain an injury that requires immediate medical attention, we will make sure you receive it.
- <u>Instrumentation</u> - the devices used to measure energy expenditure and biomechanics (i.e. forces and leg movements) are non-invasive and pose no known risk.
- It is not possible to identify all potential risks in research procedures, but the researcher(s) have taken reasonable safeguards to minimize any known and potential, but unknown, risks.

RETENTION OF BLOOD SAMPLES
You should understand that we plan to keep any extra blood samples that are not used in the analysis for this study. In other words, if we have any "extra" blood we will keep them in a freezer in our lab. <u>It is very possible that we will use all of the blood obtained in this study and will have none left, but in the event that we do, we would like your permission to keep the samples in the event that they can be used for further research.</u> We will use these samples in the future solely for additional research on obesity and metabolism; specifically, all future research will simply be an extension of what we hope to accomplish with the current study. We may simply analyze your blood for the presence of other hormones or metabolites. Your stored samples will be coded in such a way that your confidentiality will be maintained. Only the Principal Investigator (Professor Browning) will have access to the coding system for your samples. There is a possibility that your samples may be shipped to other departments on the CSU campus, or to colleagues at other Universities for assistance with analysis. Under such circumstances, the same coding system will be used, so researchers in other labs will not be able to identify you. We do not anticipate ANY commercial product development from your tissue, the samples will be used solely for research purposes. You should be advised that we do NOT have plans to re-contact you in the future regarding any additional analyses, but will seek full approval of the CSU Regulatory Compliance Office prior to initiating any further research on your samples.

By checking "Yes" below and signing on the accompanying line, you are agreeing to allow the investigators retain any blood samples obtained during this study. If you do not wish the investigators to retain any samples, please check the box marked "No" and also sign on the accompanying line.

The investigators may keep any blood samples obtained during the course of this study for future research on obesity and metabolism:

☐YES ☐NO

ARE THERE ANY BENEFITS FROM TAKING PART IN THIS STUDY? There are no direct benefits to you for participating in this study except knowing your level of fitness and how many calories you burn when walking at various speed/grade combinations.

DO I HAVE TO TAKE PART IN THE STUDY? Your participation in this research is voluntary. If you decide to participate in the study, you may withdraw your consent and stop participating at any time without penalty or loss of benefits to which you are otherwise entitled.

WHAT WILL IT COST ME TO PARTICIPATE? There is no cost to you for participating except that associated with your transportation to our testing location.

WHO WILL SEE THE INFORMATION THAT I GIVE?
We will keep private all research records that identify you, to the extent allowed by law.

Your information will be combined with information from other people taking part in the study. When we write about the study to share it with other researchers, we will write about the combined information we have gathered. You will not be identified in these written materials. We may publish the results of this study; however, we will keep you name and other identifying information private.

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information, or what that information is. For example, your name will be kept separate from your research records and these two things will be stored in different places under lock and key. You should know, however, that there are some circumstances in which we may have to show your information to other people. For example, the law may require us to show your information to a court.

CAN MY TAKING PART IN THE STUDY END EARLY? Your participation in the study could end in the rare event of an injury or if you become pregnant.

WILL I RECEIVE ANY COMPENSATION FOR TAKING PART IN THIS STUDY? You will not receive compensation for taking part in this study.

WHAT HAPPENS IF I AM INJURED BECAUSE OF THE RESEARCH? The Colorado Governmental Immunity Act determines and may limit Colorado State University's legal responsibility if an injury happens because of this study. Claims against the University must be filed within 180 days of the injury.

WHAT IF I HAVE QUESTIONS?
Before you decide whether to accept this invitation to take part in the study, please ask any questions that might come to mind now. Later, if you have questions about the study, you can contact the investigator, Ray Browning at 970-491-5868. If you have any questions about your rights as a volunteer in this research, contact Janell Barker, Human Research Administrator at 970-491-1655. We will give you a copy of this consent form to take with you.

Your signature acknowledges that you have read the information stated and willingly sign this consent form. Your signature also acknowledges that you have received, on the date signed, a copy of this document containing 3 pages.

____ ____________________
Signature of person agreeing to take part in the study Date

Printed name of person agreeing to take part in the study

Name of person providing information to participant Date

Signature of Research Staff