Michael P. McNally. POSITIVE VS. NEGATIVE MUSCLE WORK OF NON-LEVEL WALKING IN LEAN AND OBESE ADULTS. (Under the direction of Dr. Paul DeVita) Department of Exercise and Sport Science, April 2010.

When walking on non-level surfaces at a constant speed, an individual's total mechanical energy will increase when walking up an incline, and will decrease an equal amount going down a decline. Total muscle work performed however, has been previously shown to be greater during inclined gaits when compared to decline gaits. This can be explained by the fact that during incline gaits muscles will generate energy solely through the contraction of skeletal muscle, and during decline gaits muscles will dissipate energy through both skeletal muscle contraction and other possible mechanisms. One of the proposed mechanisms of energy dissipation during decline gaits is the vibration of soft tissues, which can include muscles, tendons, ligaments, and adipose tissue. The global hypothesis of this study is that skeletal muscles will generate more mechanical energy in gait tasks that raise the center of mass compared to the mechanical energy they dissipate in gait tasks that lower the center of mass, despite equivalent changes in total mechanical energy. Because obese adults have a greater amount of adipose tissue which is available for vibration and dissipation of energy, the sub-hypothesis of this study is that obese individuals will show a larger bias towards net positive muscle work in incline vs. decline walking compared to lean individuals.

Healthy lean adults (BMI <25) and healthy obese adults (BMI >35) were tested walking up an incline surface and down a decline surface at 1.5 m/s. Three dimensional kinematics and ground reaction forces were collected and used to calculate joint kinetics through standard biomechanical motion analysis and inverse dynamics. Selected gait variables were analyzed using a two way ANOVA with repeated measures, with $p<.05$.

A significantly greater amount of total muscle work was performed during incline walking compared to decline walking, with obese performing significantly more muscle work overall than lean adults. There was no significant interaction for total muscle work during incline and decline gaits for lean and obese adults.

This study is in agreement with the global hypothesis that skeletal muscles generate more mechanical energy during inclined gaits than they dissipate during declined gait, despite equivalent changes in total mechanical energy. The sub-hypothesis was not supported, as obese adults had a similar bias towards total muscle work in inclined gaits. This suggests that other mechanisms may be responsible for the bias towards positive muscle work, including more erect gaits during decline walking, increased positive muscle work from the forward propulsion of the body, or the dissipation of energy through the shoe element.

POSITIVE VS. NEGATIVE MUSCLE WORK OF NON-LEVEL WALKING IN LEAN AND OBESE ADULTS

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Michael P. McNally

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By

Michael Paul McNally

APPROVED BY:

DIRECTOR OF THESIS:

Paul DeVita, Ph.D.

COMMITTEE MEMBER:

Tibor Hortobágyi, Ph.D.

COMMITTEE MEMBER:

Anthony Kulas, Ph.D., ATC

COMMITTEE MEMBER:

Mr. Mike McCammon

CHAIR OF THE DEPARTMENT OF EXERCISE AND SPORT SCIENCE:

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DEAN OF THE COLLEGE OF HEALTH AND HUMAN PERFORMANCE:

Glen G. Gilbert, Ph.D.

DEAN OF THE GRADUATE SCHOOL:

Paul J. Gemperline, Ph.D.

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CHAPTER 1 – INTRODUCTION

Human locomotion is a process involving both the generation and dissipation of mechanical energy. This mechanical energy is generated by performing positive muscle work using concentric muscle contractions, and dissipated by performing negative muscle work using eccentric muscle contractions. The amount of total external mechanical work performed during locomotion is equivalent to the change in kinetic energy (related to changes in velocity) and the change in potential energy (related to changes in vertical displacement).   For example, during gait tasks in which a person changes speeds, an increase in kinetic energy while speeding up, and likewise a decrease in kinetic energy while slowing down would be observed. Also, gait tasks which raise the center of mass would increase the body's potential energy, and likewise the potential energy would decrease while lowering the center of mass. Understanding how our muscles function mechanically to contribute to these changes in energy is fundamentally important in the understanding of the biomechanics of locomotion.

Previous studies investigating muscle work during non-level gaits has reported positive and negative muscle work across individual joints [4, 7, 10, 12, 14, 18, 21]. In the present study we are investigating the total amount of mechanical energy that is generated and dissipated during non-level (i.e. incline and decline) walking throughout the lower extremity. As previously stated, gait tasks that raise the center of mass will increase the potential energy of the body, and while keeping a constant velocity, lowering the center of mass the same distance will cause an equivalent decrease in the body's potential energy. The change in potential energy, while keeping a constant velocity, will be reflected in the amount of net external mechanical work. During level walking (no change in potential energy) the amount of positive work and negative work will be the same. On non-level surfaces, as the angle of incline increases, the fraction of positive work to negative work increases while going up the incline. Going down the incline will result in a similar fraction of negative work to positive work [13]. However, when muscle work across joints is calculated, there exists a bias towards performing more positive work during inclined gaits than negative work during declined gaits despite having identical vertical displacements [3, 18]. The reason for this positive bias is not yet known, however a few theories exist in an attempt to explain it. Sasaki et. al. (2009) recently showed through musculoskeletal modeling, that the bias seen towards positive muscle work in level walking was negated by the loss in energy due to the negative shoe-element work [19]. In work by DeVita et al in 2007, a few kinematic observations were observed that could also produce the positive bias of muscle work. One observation was that despite larger ground reaction forces occurring during declined gaits, the body is more erect reducing the ground reaction force moment arm at each joint, thus reducing the muscle joint torque and work needed in declined gaits. Another observation was an increased single-leg stance phase time during inclined gaits, which could lead to an increase in the total amount of muscle work [1].

The focus of this thesis is the idea that while positive muscle work is the sole generator of mechanical energy during inclined gaits, negative muscle work along with the vibration of the muscle itself and other soft tissues dissipates mechanical energy. Therefore, an increase in soft tissue mass, such as that seen in obese individuals, would dissipate a greater amount of energy through the vibration of soft tissue.

Hypothesis

Our previous knowledge of muscle work shows that during non-level gaits, the total amount of net positive muscle work during incline walking is greater than the total amount of net negative muscle work during decline walking despite equivalent changes in total body mechanical energy. Some of this discrepancy can be attributed to a longer duration single stance phase in incline compared to decline gaits, increased moment arms at the hip and ankle during in incline compared to decline gait, negative shoe-element work in decline gait, and the dissipation of energy by both muscle work and muscle and other soft tissue vibration compared to only muscle contractile work in energy generating incline gaits. Obese individuals have a greater ratio of fat to fat free mass and therefore a lower relative amount of muscle tissue available for negative work. Our present global hypothesis is that skeletal muscles generate more mechanical energy in gait tasks that raise the center of mass compared to the mechanical energy they dissipate in gait tasks that lower the center of mass, despite equivalent changes in total mechanical energy. Our sub-hypothesis specific to this thesis is that obese individuals will show a larger bias towards net positive muscle work in incline vs. decline walking compared to lean individuals.

Purpose

The purpose of this study was to compare the amount of net positive muscle work during incline gait and net negative muscle work in decline gait, and to compare the difference in positive and negative muscle work between obese and lean individuals. We expected that the difference in total muscle work during inclined gaits versus total muscle work during declined gaits would be greater in obese than in lean individuals.

Delimitations

Delimitations of this study were as follows:

All subjects were healthy with no history of injury in the lower extremity

Subjects were excluded if they had previous surgeries or any neuromuscular and musculoskeletal diseases.

Lean subjects had a BMI of less than 25 kg/m².

Obese subjects had a BMI of greater than 35 kg/m².

Testing was limited to ramp ascent and descent at a 10 degree angle.

Analysis focused on the gait cycle of the right lower extremity on each subject.

Limitations

Analysis of data was limited by the accuracy of force platforms, video capture, and computer analysis systems, as well as collection of data by these systems.

All interview information collected was assumed to be correct.

Joint kinematics and kinetics were limited by correct placement of markers on obese adults.

CHAPTER 2 - REVIEW OF LITERATURE

This thesis' focus is on the previously hypothesized biomechanical principle that muscles of the lower extremity perform more total muscle work during gait tasks that raise the center of mass than they perform during gait tasks that lower the center of mass. In particular, it is focused upon the effects that a greater soft tissue mass associated with obesity will have on energy dissipation due to the vibration of these soft tissues. With this in mind, this review of literature will focus on: 1) Muscle work in non-level gaits, 2) Potential explanations for the bias of muscle function, and 3) The biomechanics of obese gait.

Muscle Work in Non-Level Gaits

Muscular work can be defined as the product of the torque applied by the muscle and the distance that the limb being acted upon has moved. When the direction of force and displacement of the limb are in the same direction, positive muscle work (W_{pos}) is performed and mechanical energy is generated. Conversely, when the direction of force and displacement of the limb are in opposite directions, the muscle is lengthening and negative muscle work ($W_{neg.}$) is performed, which absorbs mechanical energy. During the gait cycle, the energy generated and dissipated by the lower extremities is equal to the sum of the change in kinetic energy and the change in potential energy. Kinetic energy (KE) is the energy of motion concerning velocity (½ x mass x velocity^2) whereas potential energy (PE) is the energy of motion concerning vertical displacement (mass x acceleration of gravity x vertical displacement). Therefore, when velocity is kept constant on a level surface, the total change in mechanical energy is zero, reflecting an equal percentage of positive and negative muscle work in contributing to gait [13].

This study is focused on the contribution of muscle mechanical energy to gait tasks that raise the body center of mass and lower the body center of mass. The change in energy in these gait tasks is therefore going to reflect the change in vertical displacement (PE) when velocity (KE) is kept constant. Gait tasks that raise the center of mass will then involve more positive muscle work due to a positive change in PE, and gait tasks that lower the center of mass will involve more negative muscle work due to a negative change in PE [13]. This understanding was first shown in the analysis of individual joint muscle power in ascent and descent stair gait [11]. While muscle work itself was not referred to, it can be determined by the area underneath the power versus time curve, which showed greater amounts of positive work during incline, and greater amounts of negative work during decline.

Further work in inclined and declined stair gaits has shown the same pattern of positive and negative muscle work. Muscle work at individual lower extremity joints during incline stair gait was shown to generate energy across the hip, knee, and ankle joints, while decline stair gait showed the dissipation of energy across all three joints [18]. The only substantial amount of positive muscle work during decline walking was in the hip, although this was still only about 30% of that seen during incline walking. This was also one of the first times that a bias was shown in muscle function during stair gait to produce more positive muscle work during incline than negative muscle work during decline, despite equivalent changes in potential energy. This was shown through the summing of the positive muscle work during ascent (2.33 J) and negative muscle work during decline (-2.01 J) at all three joints, to get total lower extremity muscle work.

Incline and decline ramps have also shown the same pattern of muscle work through the gait cycle [10, 12]. In ramp gait however, energy generation during incline occurs mainly at the hip and ankle joints, with the knee contributing a negligible amount of work. During decline, the knee and ankle contribute to energy dissipation while the power about the hip remains negligible. In these studies, only joint work was measured and total work was not calculated.

As is shown, most of the previous work has measured muscle work at individual joints during the gait cycle, and did not measure total lower extremity muscle work. More recent research has summed lower extremity muscle work, and found similar to the stairs, that positive muscle work during walking was greater during incline walking (89 J/m) than negative work was during decline (-71 J/m) [3]. Individual joint work agreed with that of previous studies, with the hip and ankle contributing 86% of the positive work during incline, and the knee joint being the main contributor to negative muscle work in decline, contributing 56% of the negative muscle work. A similar study performed examined positive and negative muscle work during non-level running and also showed 25% greater positive muscle work during incline walking compared to negative muscle work during decline walking [6].

These previous studies on non-level gaits show more energy generating positive work being performed during gait tasks that raise the center of mass, and more energy dissipating negative work being performed during gait tasks that lower the center of mass. Energy generation during incline walking can be attributed mostly to the hip and ankle, with the knee being the primary dissipater of energy. During decline walking, the knee contributes the most to energy dissipation with the hip performing very little work and the ankle performing both negative work during early and mid-stance, and positive work during toe-off. Finally, a bias in muscle function was shown towards positive muscle work in non-level gaits, explained by more energy being generated during inclined gait than energy being dissipated during declined gait. This bias occurs despite a constant kinetic energy and equivalent changes in potential energy.

This information leads us to believe that another mechanism of energy dissipation is present, which must be overcome by energy generating positive muscle work.

Potential Explanations for the Bias of Muscle Function

Various mechanisms attempt to explain this bias towards positive muscle work in incline over negative muscle work during decline. Through observation in current laboratory research, it has been seen that while the ground reaction force (GRF) vector is greater during decline walking, it is also farther from the joint centers during incline walking, while passing much closer to the joint centers in descent. This occurs because as humans walk up an incline, greater hip and knee flexion are required to raise the foot. This more flexed position moves the knee and ankle joint centers farther from the GRF vector which increases the amount of work which needs to be done at these joints to overcome the work being done on them by the ground. During decline gait however, humans will walk more upright and land more straight-legged. This moves the knee and ankle joint centers closer to the GRF vector, decreasing the amount of muscle work needed to overcome the ground reaction force. Therefore, due to just the direction of the GRF vector and its effect on joint moments, more positive muscle will be performed during inclined gaits than negative muscle work during declined gaits.

While the larger GRF does not greatly affect the work of muscle, it does have an impact on the soft tissues of the body. The vibration of these soft tissues, termed the “wobbling” mass, in the lower extremity reduce the amount of force absorbed by the rigid bodies, which consist of bone and muscle [2, 8, 16, 17]. When measuring GRF forces during a vertical drop in rigid body models and comparing them to a “wobbling” mass model, the rigid body models had vertical GRF of 40.5 body weights, compared to only 16.2 body weights measured in the “wobbling” mass model [17]. Most of this change in vertical GRF was found to occur within the first 10-30 ms within landing, suggesting that the impact of landing is what caused the vibration of soft tissue to absorb force [8]. The results of these studies showed that the "wobbling" mass model more closely mimicked experimental data, showing that soft tissue does account for some absorption of force during the impact incurred at heel strike. Less GRF during these impacts would then suggest a smaller amount of negative work, and therefore energy dissipation. Theoretically then, the vibration of the soft tissues account for some dissipation of energy during the gait cycle.

The Biomechanics of Obese Gait

Obesity in adults leads to a number of medical problems, including cardiovascular disease, insulin resistance, and an increase in the occurrence of joint and muscle pathologies. Also associated with obesity is a change in biomechanics during gait, and most of the literature has focused on those changes during level walking. The most obvious difference between lean and obese adults during gait, is an increase in vertical GRF, which increases the amount of work needed to lift and lower the center of mass [1]. The increased vertical GRF seen in obese individuals could increase the amount of energy dissipated during descent by accelerating the tissues more and leading to greater vibration.

Mechanically, the increased girth of the legs have been shown to lead to greater step width as well as a greater lateral leg swing [1, 20]. Obese adults have also been shown to walk at a slower preferred speed and with shorter step lengths [5, 9, 20]. By walking at slower speeds obese adults significantly reduce vertical GRF and joint torques [1], suggesting that the slower walking speed is preferred to enhance the safety of lower extremity joints. While walking at a set speed of 1.5 m/s, obese adults walked with reduced knee flexion, aligning the GRF vector more closely with the joint. This resulted in similar knee torque and work compared to lean adults despite significantly greater GRF [5]. While these data have been reported from level walking, they suggest that obese adults will reorganize their neuromuscular patterns on non-level surfaces as well to perhaps walk even more upright and align the joints to perform even less negative muscular work despite an equivalent changes in potential energy.

Summary

Much of the literature associated with this study involves individual joint contributions to muscle work on inclined and declined gaits. From these we can see a positive bias towards muscle work, shown in the fact that we have more positive muscle work going up the incline, and more negative muscle work going down the incline, despite having equal changes in total external mechanical energy. Positive muscle work is the only method of generating energy, however because of this bias we can see that negative muscle work is the primary dissipater of energy, but there are other methods of energy dissipation that are not yet accounted for. Ideas exist as to why we show this positive bias towards muscle function, but little is known as to the exact reasons. This study will investigate the idea that muscle and soft tissue vibration contribute to the dissipation of energy, and lead to the positive bias of muscle work, despite equal changes in total mechanical energy.

CHAPTER 3 – METHODOLOGY

This study included an experiment that tested the hypothesis that positive muscle work is greater in gait tasks that raise the center of mass compared to negative muscle work in gait tasks that lower the center of mass. This chapter describes the procedures used in this study. This chapter is divided into several sections: 1) Subjects, 2) Instrumentation, 3) Testing protocol, 4) Data Reduction.

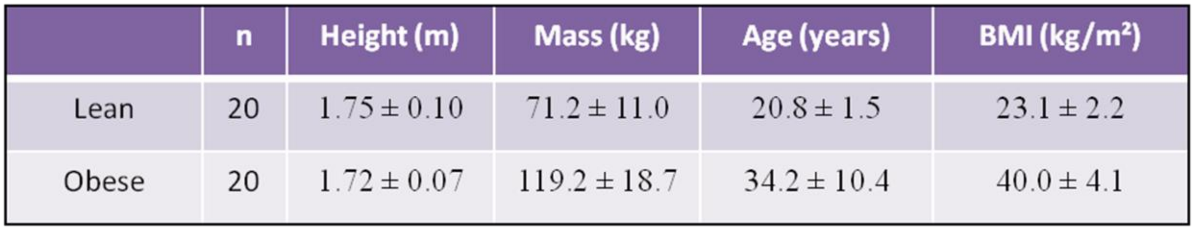
Subjects

The subject characteristics recorded from this study are reported in Table 1. This study involved two groups of adult participants between the ages of 18-45. A lean adult group was selected and an obese adult group was selected. Sixteen lean adults, classified as having a BMI of <25 kg/m^2, were taken from previously recorded data. These lean adults were compared to the obese adult group, classified as having a BMI of >35 kg/m^2, which were recruited with assistance from the Brody School of Medicine in Greenville, NC, as well as from a database of local volunteers. All subjects were selected based on a set of inclusion and exclusion criteria.

Inclusion Criteria:

Subjects were healthy as determined by our criteria

Table 1 – Subject demographics



	n	Height (m)	Mass (kg)	Age (years)	BMI (kg/m^2)
Lean	20	1.75 ± 0.10	71.2 ± 11.0	20.8 ± 1.5	23.1 ± 2.2
Obese	20	1.72 ± 0.07	119.2 ± 18.7	34.2 ± 10.4	40.0 ± 4.1

Subjects were within the designated ranges of age and BMI for the selected population

Subjects were informed of all testing procedures prior to participation, and completed the appropriate informed consent form.

Exclusion Criteria:

Subject had any diseases that may affect gait patterns, such as neuromuscular and musculoskeletal diseases.

Subject had previous surgery in the lower extremity.

Subjects had a level of pain that influences gait, as determined by a pre-testing physical function questionnaire.

Subjects walked with abnormal gait pattern such as a limp during incline and decline walking trials.

Testing Protocol

Each subject completed a physical ability questionnaire and informed consent document before testing. Height and weight were then measured in meters (m) and kilograms (kg). Subjects then changed into black form fitting shorts and a black tight fitting t-shirt, in order to minimize movement artifact of the markers.

Reflective markers were placed on the subject, with placement defined by a previously set arrangement. Fifteen tracking markers were used to track joint and limb movement, and eleven calibration markers were used to define joint centers of the lower extremity. A static standing trial was collected for 5 seconds to record joint centers. Calibration markers were removed and another standing static trial was collected for 5 seconds to assess joint positions.

Subjects then began ramp trials on a 4 m long ramp, set at an angle of 10 degrees. A starting point was selected along the walkway and subjects were instructed to walk up or down the ramp at 1.5 m/s, which was measured using a Brower timing gait system. Enough practice was allowed for the subject to become comfortable with the speed and for the researcher to adjust the starting position so that the right foot is striking the force plate in a normal gait stride. Once comfortable, the subject performed 5 successful gait trials walking up the incline, and 5 successful gait trials walking down the incline. A successful gait trial was defined at one in which the velocity was held constant at 1.5 m/s through the testing area, the entire right foot contacted the force plate, and there were no visual changes in the gait cycle to contact the force plate.

Kinematic data were collected using an eight camera Pro Reflex camera system, which uses infrared light to track the reflective markers placed on the subject. Each trial was collected at 120 Hz and stored in Qualisys Track Manager (QTM) software. Trials were tracked using appropriate protocols which labeled the markers, and filled in any gaps from brief marker fallout. A Panasonic video camcorder was also used to observe if any changes in the gait cycle may have occurred through the testing range. Ground reaction forces were collected using an AMTI Model LG-6 force platform, located in the middle of the ramp. Voltage signals were stored in QTM, amplified and sampled at a frequency of 1000 Hz.

Data Reduction

Following data collection, all trials were exported into Visual 3D software to be analyzed. This program created a lower extremity rigid model from segmental and joint positions defined in the static calibration trial, and applied this model to all gait trials. Segmental masses, their moments of inertia, and location of mass centers were estimated using anthropometric measurements. Inverse dynamics were then performed to calculate joint forces and torques. This analysis was performed on the foot first, because the foot contacted the force plate which measured the GRF during the stance phase. Joint reaction forces (JRF) at the ankle for each frame of data were found using the equation:

$$JRF_{ankle} = ma_{cm} - mg - f_{grf}$$

where m is the segment mass, a_{cm} is the acceleration of the center of mass, mg is the force vector of gravity, and f_{grf} is the GRF. Joint torques (JM) were found using the equation:

$$JM_{ankle} = I\alpha - (d_1 \text{ x } JRF_{ankle}) - (d_2 \text{ x } F_{GRF}) - t$$

where I is the moment of inertia, α is the angular acceleration, d_1 x JRF_{ankle} is the vector which describes the moment as a result of the JRF, d_2 x F_{GRF} is the vector described from the GRF, and t is the ground reaction torque vector. All calculations were performed in the specific segments local coordinate system. GRF was replaced by the distal JRF of the adjacent segment for the knee and hip JM and JRF calculations. These calculations were represented using the following equations:

$$JRF_{Prox} = ma_{cm} - mg - JRF_{Distal}$$

$$JM_{Prox} = I\alpha - (d_1 \text{ x } F_{JRF\_Prox}) - (d_2 \text{ x } F_{JRF\_Distal}) - JM_{Distal}$$

Joint power was calculated as the product of joint moment and joint angular velocity using the following equation:

$$P = JM \text{ x } (\omega_{Proximal} - \omega_{Distal})$$

where P is the joint power, JM is the joint torque, and $\omega_{Proximal}$ and ω_{Distal} are vectors representing the proximal and distal segment angular velocities. Positive and negative work variables were then calculated as the areas under selected portions of the joint power curves. Total positive and negative work for each gait task was calculated as was the difference between these values for total net work in each gait.

Statistical Analysis

Selected gait variables were analyzed with a repeated measures two way Analysis of Variance (ANOVA). The two factors were body composition (lean vs. obese) and gait direction (incline vs. decline). Repeated measures were used on the gait direction independent variable. The alpha level was set to 0.05 for all tests.

CHAPTER 4 - RESULTS

It was hypothesized that adults will generate more mechanical energy during gait tasks that raise the center of mass, compared to the amount of mechanical energy they dissipate in gait tasks which lower the center of mass. Our sub-hypothesis stated that obese adults will show a greater bias towards energy generation during incline vs. energy generation vs. decline compared to lean adults. Lean and obese adults were tested while performing an identical gait task by walking up and down an inclined surface at 1.5 m/s. This chapter presents the energetics and lower extremity biomechanics of lean and obese adults as they perform these gait tasks.

Preliminary Results

Figure 1 shows the average acceleration of the right ASIS marker for lean and obese adults during decline walking. Measurements were obtained from the ASIS because of the large difference in soft tissue mass between lean and obese adults around the abdomen, and they were taken during decline walking because this is where larger ground reaction forces will cause greater vibration of soft tissue which may lead to an increase in energy dissipation. Raw accelerations of the ASIS marker showed a significant difference (determined by an independent t-test) between obese and lean adults, with obese having 16% greater acceleration than lean (7.4 ± 1.3 vs. 6.4 ± 1.2 m/s², p =0.01).

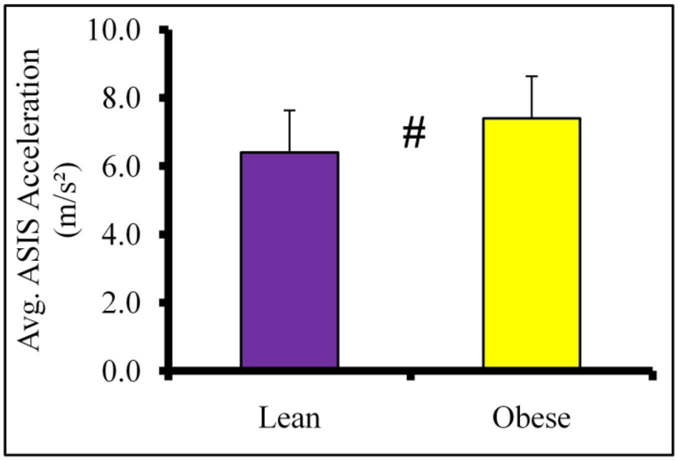


Figure 1 – Average ASIS acceleration for lean and obese adults during decline walking

= Obese > Lean, p = 0.01 (t-test)

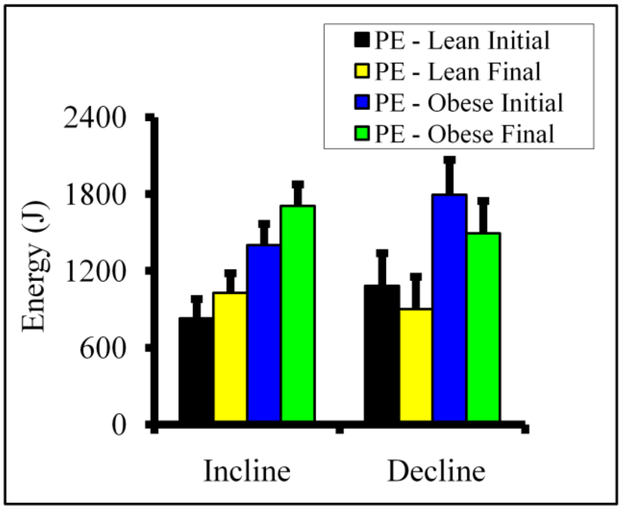


Figure 2 – Initial and Final potential energy during incline and decline walking

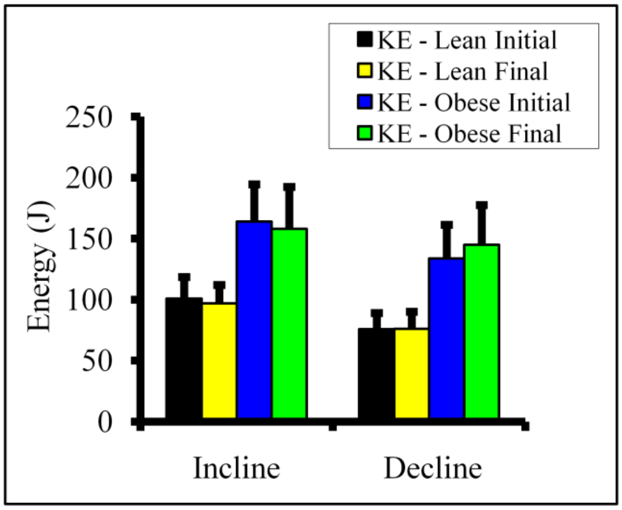


Figure 3 – Initial and final kinetic energy during incline and decline walking

Preliminary measurements were taken to check that the change in total mechanical energy was due to a change in potential energy, accomplished by keeping a constant average velocity which will result in no change of kinetic energy. Figure 2 showed the changes in potential energy for lean and obese groups during incline and decline walking. These data showed that during incline walking lean individuals increased their potential energy 200 J or 24%, and during decline walking they decreased potential energy 184 J, or 17%. Obese individuals during incline increased potential energy 308 J, 22% and decreased potential energy 302 J, 17% during decline. Kinetic energy data (Figure 3) showed that the lean group had a decrease of 4 J, 4% during

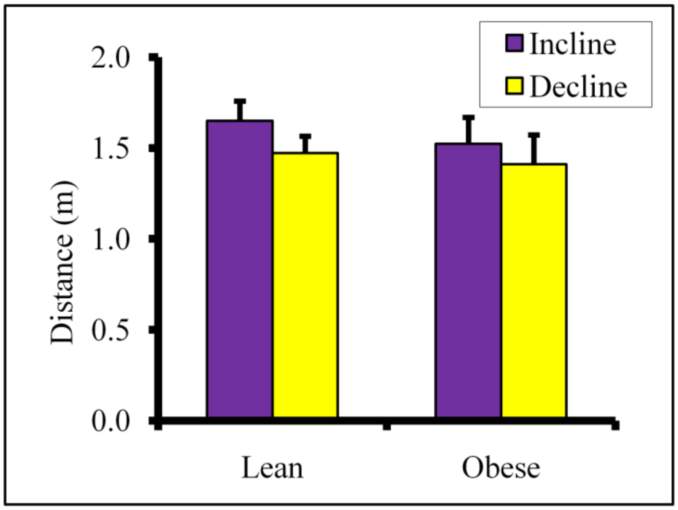


Figure 4 – Stride length for lean and obese adults during incline and decline walking

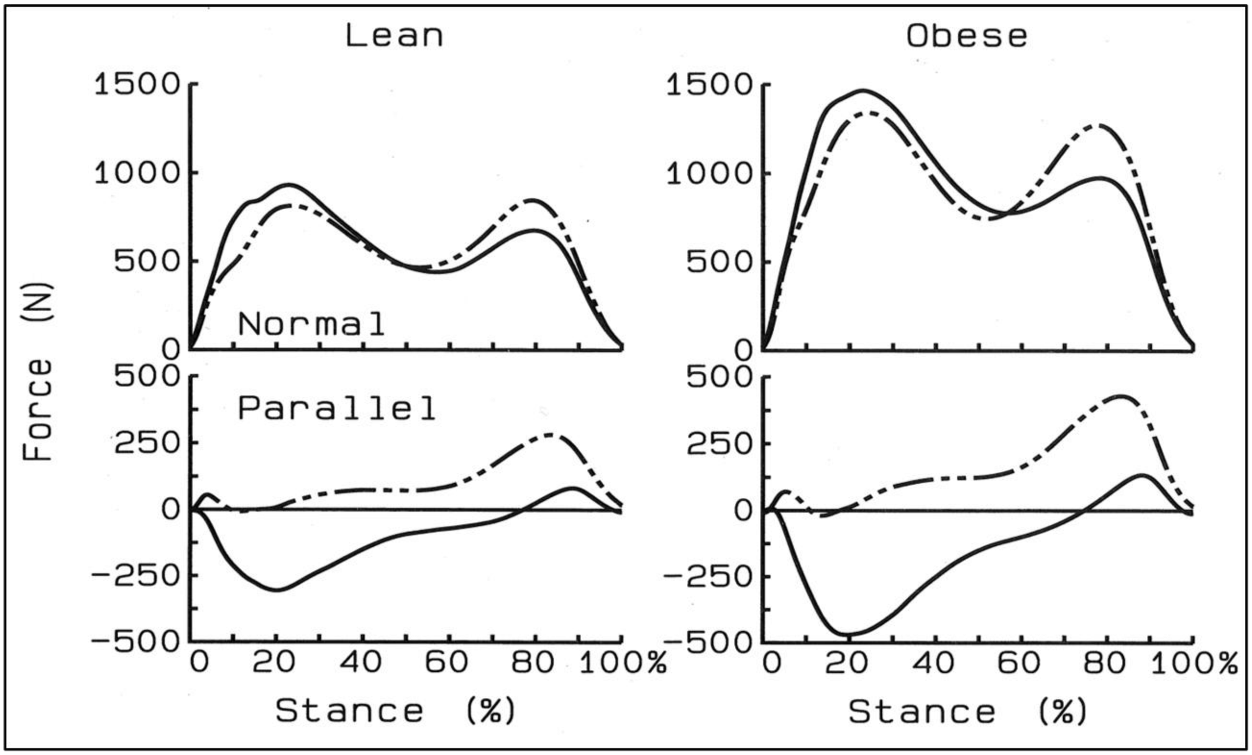


Figure 5 – Mean normal and parallel ground reaction forces from all trials for lean and obese adults during incline and decline walking

incline and no change or 0% during decline. Obese adults had a decrease in kinetic energy of 6 J, 4% during incline compared to an increase of 11 J, 6% during decline.

Stride length data showed changes within groups and differences between groups (Figure 4). Stride length was 0.18 m, 11% shorter in decline vs. incline walking in lean adults and 0.11 m, 7% shorter in decline vs. incline walking in obese adults. Lean adults had longer stride lengths than obese adults in both incline and decline walking. Stride length for lean vs. obese adults was 0.13 m, 8% longer in incline (1.65 vs. 1.52 m) and was 0.06 m, 4% longer in decline (1.47 vs. 1.41 m). Changes in stride length affect the amount of work done by affecting the vertical displacement of the center of mass during each stride. Therefore, total work variables that are being used o test the hypothesis were normalized to stride length which provides a per unit distance comparison instead of a per unit step comparison.

Incline and Decline Locomotion Biomechanics

Figure 5 shows the normal to surface slope GRF and the parallel to surface slope GRF of obese and lean adults, directly comparing incline and decline gaits. The normal impulse (from normal GRF; Figure 6) and the absolute propelling and braking impulses (from parallel GRF; Figure 7) are shown comparing obese, lean, incline, and decline groups. For both normal and parallel GRF impulses, the mass x direction interaction was not significant. There was a significant difference in normal impulse with incline walking producing an 8% greater impulse compared to decline walking (482 ± 155 Ns vs. 447 ± 146 Ns; $p<0.001$). Parallel impulse was not significantly different between incline and decline walking. A group effector also showed significant difference between obese and lean adults, with obese showing a 58% greater normal impulse when compared to lean (568 ± 135 Ns vs. 361 ± 73 Ns; $p<0.001$) and a 50% greater

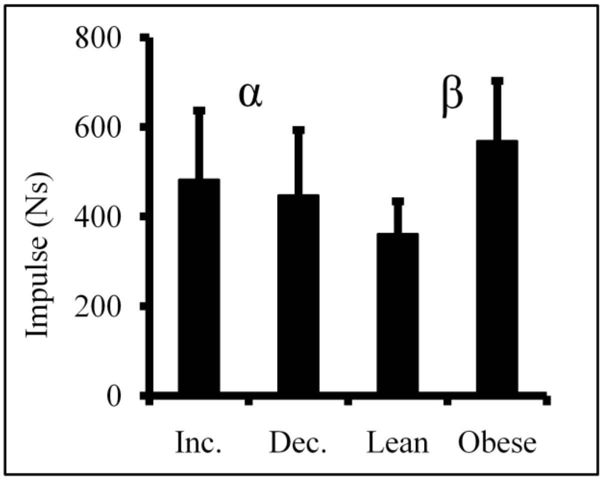


Figure 6 – Impulse from normal GRF

α = Incline > Decline, $p < 0.001$; β = Obese > Lean, $p < 0.001$

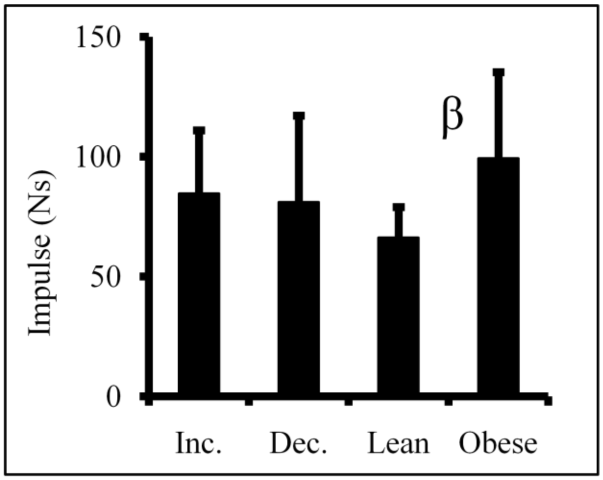


Figure 7 – Impulse from parallel GRF

β = Obese > Lean, $p < 0.001$

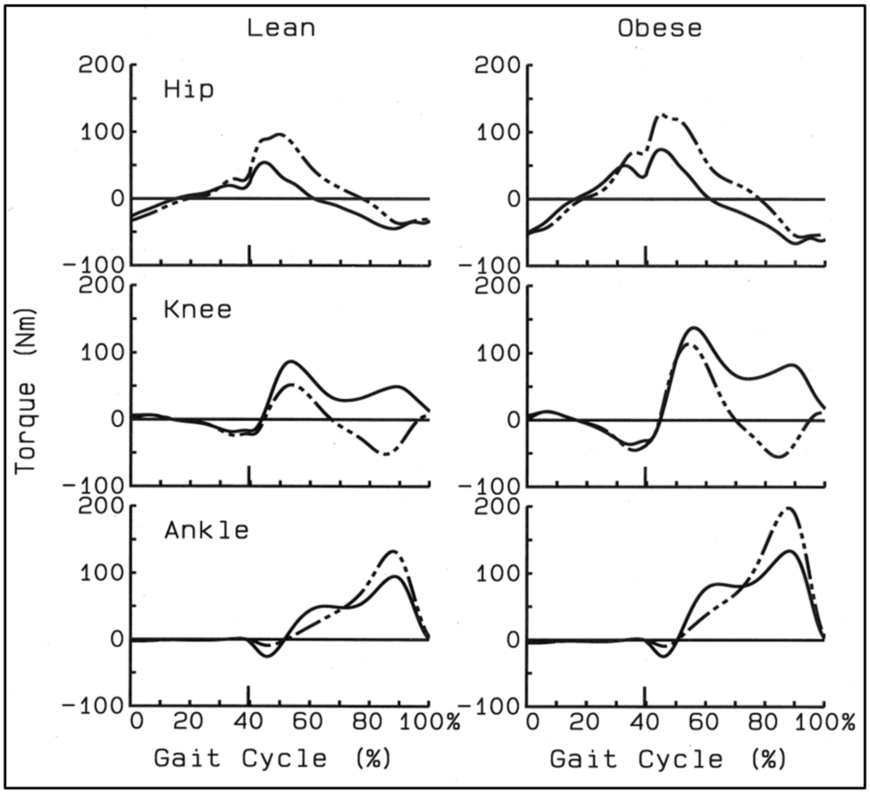


Figure 8 – Hip, knee, and ankle torques for lean and obese adults during incline and decline walking

parallel impulse than lean (99.3 ± 35.9 Ns vs. 66.2 ± 12.8 Ns; p<0.001).

Sagittal plane torque curves for the hip, knee, and ankle for lean and obese adults, and comparing incline and decline walking are shown in Figure 8. Extensor angular impulses of the hip, knee, and ankle during the stance phase are shown in Figure 9 and were used to compare groups. No significant mass x direction interaction occurred at the hip (Figure 9a). Extensor angular impulse during incline walking was 169% greater than during decline walking (25.3 ± 11.1 Nms vs. 9.4 ± 5.9 Nms; p<0.001), and obese adults had a 39% greater extensor angular impulse than lean individuals (20.2 ± 13.1 Nms vs. 14.5 ± 10.0 Nms; p = 0.03). Figure 9b shows the extensor angular impulse at the knee for lean and obese groups during incline and decline walking. There was a significant mass x direction interaction for the extensor angular impulse of

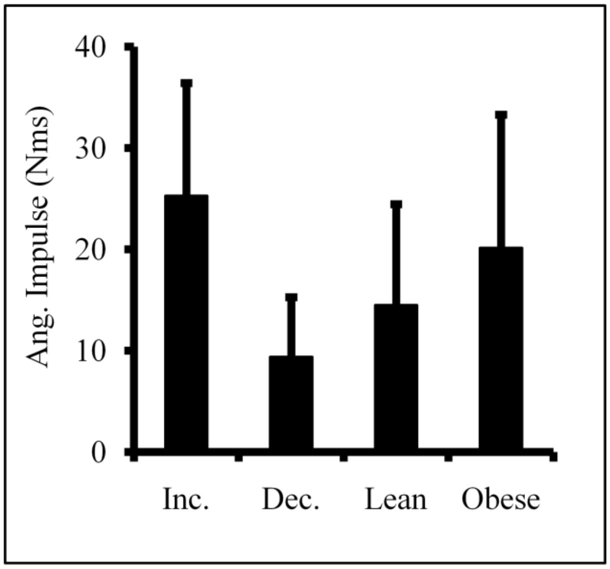


Figure 9a – Extensor angular impulse at the hip during the stance phase

α = Incline > Decline, p < 0.001; β = Obese > Lean, p = 0.03

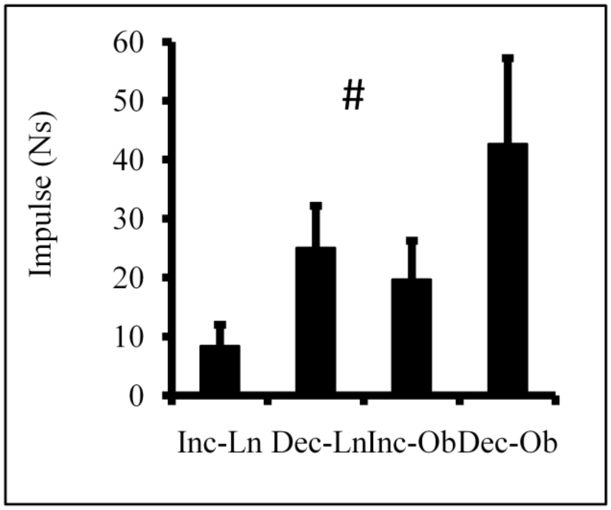


Figure 9b – Extensor angular impulse at the knee during the stance phase

= sig. mass x direction interaction, p = 0.025

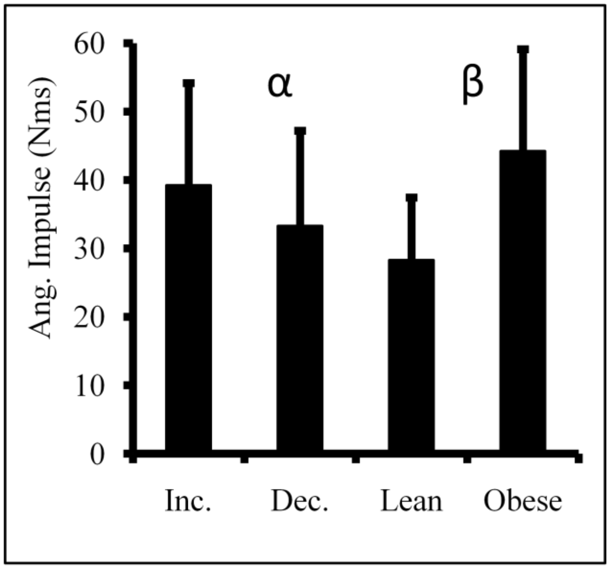


Figure 9c – Extensor angular impulse at the ankle during the stance phase

α = Incline > Decline, p < 0.001; β = Obese > Lean, p < 0.001

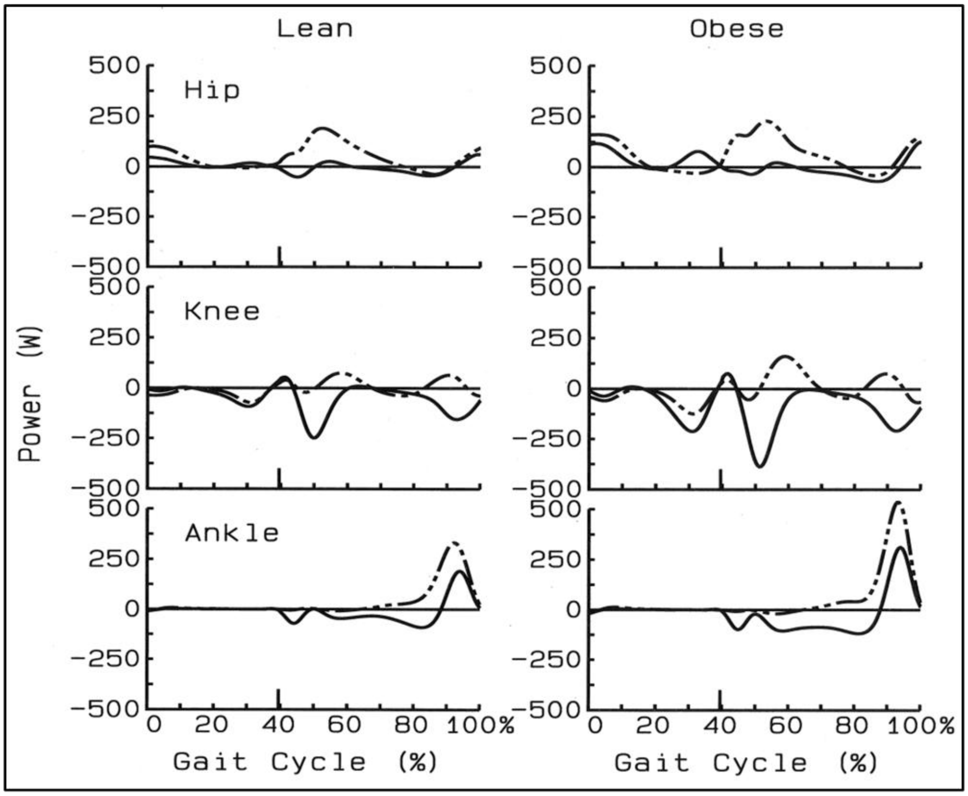


Figure 10 – Hip, knee, and ankle powers for lean and obese adults during incline and decline walking

the knee (p=0.025). Decline walking was performed with greater extensor angular impulse compared to incline walking in both groups (25.0 vs. 8.4 Nms for lean and 42.6 vs. 19.7 for obese). The difference however was significantly greater in obese adults than in lean adults (22.9 vs. 16.7 Nms). There was no significant mass x direction interaction effect at the ankle (Figure 9c). Plantarflexor angular impulse was 17% greater in incline when compared to decline (39.2 ± 14.9 Nms vs. 33.3 ± 13.9 Nms, $p<0.001$), and obese adults had a 56% greater plantarflexor angular impulse than lean (44.2 ± 14.9 Nms vs. 28.3 ± 9.2 Nms, $p<0.001$).

Figure 10 shows sagittal plane power curves throughout the gait cycle. Average power during the stance phase was calculated and shown for the hip, knee, and ankle in Figure 11. Average power at the hip did not show a significant mass x direction interaction (Figure 11a).

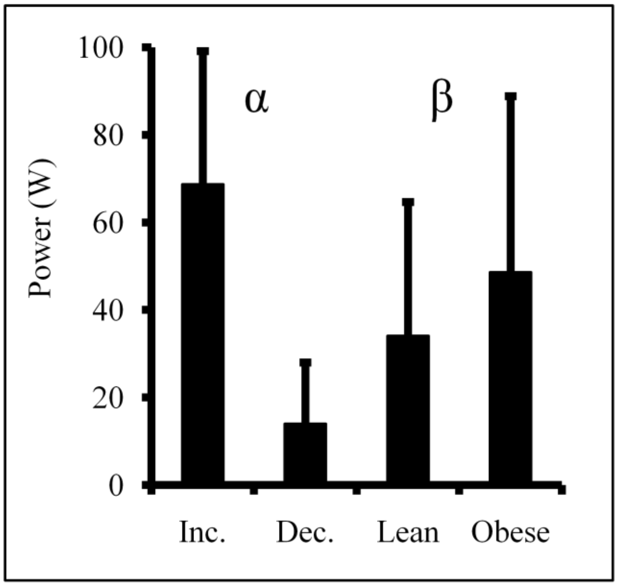


Figure 11a – Average power at the hip during the stance phase

α = Incline > Decline, $p < 0.001$; β = Obese > Lean, $p = 0.001$

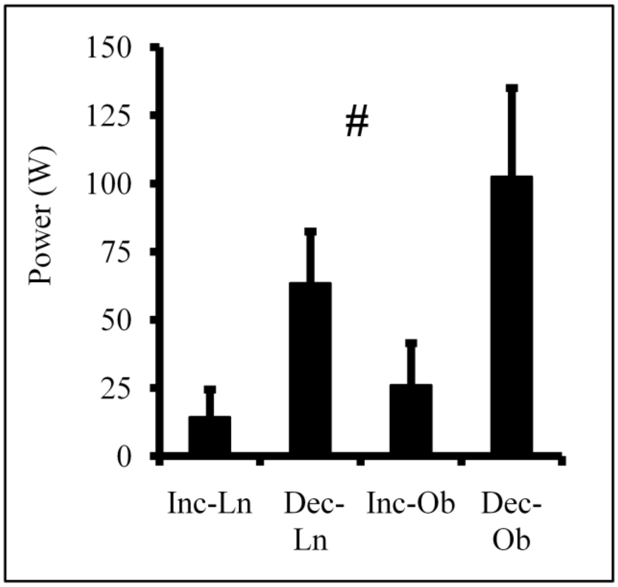


Figure 11b – Average power at the knee during the stance phase

= sig. mass x direction interaction, $p = 0.006$

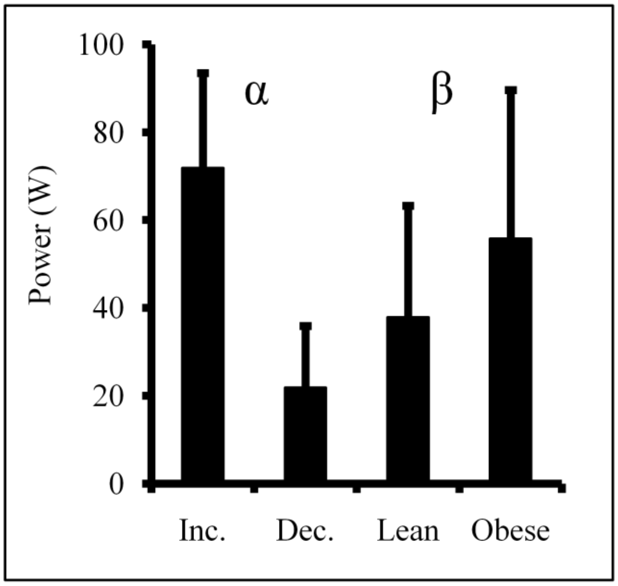


Figure 11c – Average power at the ankle during the stance phase

α = Incline > Decline, $p < 0.001$; β = Obese > Lean, $p < 0.001$

Greater average power was observed during incline vs. decline walking (69 ± 30 W vs. 14 ± 14 W, p<0.001), and obese had a higher average power than lean individuals (49 ± 40 W vs. 34 ± 31 W, p = 0.001). Average power at the knee for lean and obese groups during incline and decline walking are shown in Figure 11b. A significant mass x direction interaction was present for average power at the knee. Average power was greater during decline walking for both lean (63.4 vs. 14.5 W) and obese (102.5 vs. 26.1 W) adults, however obese adults had a greater increase in average power at the knee during decline walking vs. incline walking compared to lean adults (76.5 vs. 48.9 W, p = 0.006). There was not a significant mass x direction interaction at the ankle (Figure 11c). During incline walking the average power was significantly greater than during decline walking (72 ± 22 W vs. 22 ± 14 W, p<0.001), and obese adults had a greater average power compared to lean adults (56 ± 34 W vs. 38 ± 25 W, p<0.001).

Work in Incline and Decline Gaits

Figure 12 shows the amount of total muscle work performed by lean and obese adults during incline and decline walking. Muscle work calculations were normalized to stride length because the varied stride length among groups affected the amount of work performed. Contrary to expectation, no significant mass x direction interaction was found for the total amount of muscle work. More total muscle work was performed during incline vs. decline walking (73.29 ± 20.58 J vs. 60.62 ± 23.00 J; p=0.003), and obese adults performed more total muscle work than lean adults (81.63 ± 20.40 J vs. 52.28 ± 13.29 J; p<0.001).Positive muscle work for lean and obese during incline and decline walking, and negative muscle work for lean and obese during incline and decline walking, are shown in Figures 13 and 14. A significant mass x direction interaction was observed for both positive and negative muscle work in lean and obese adults

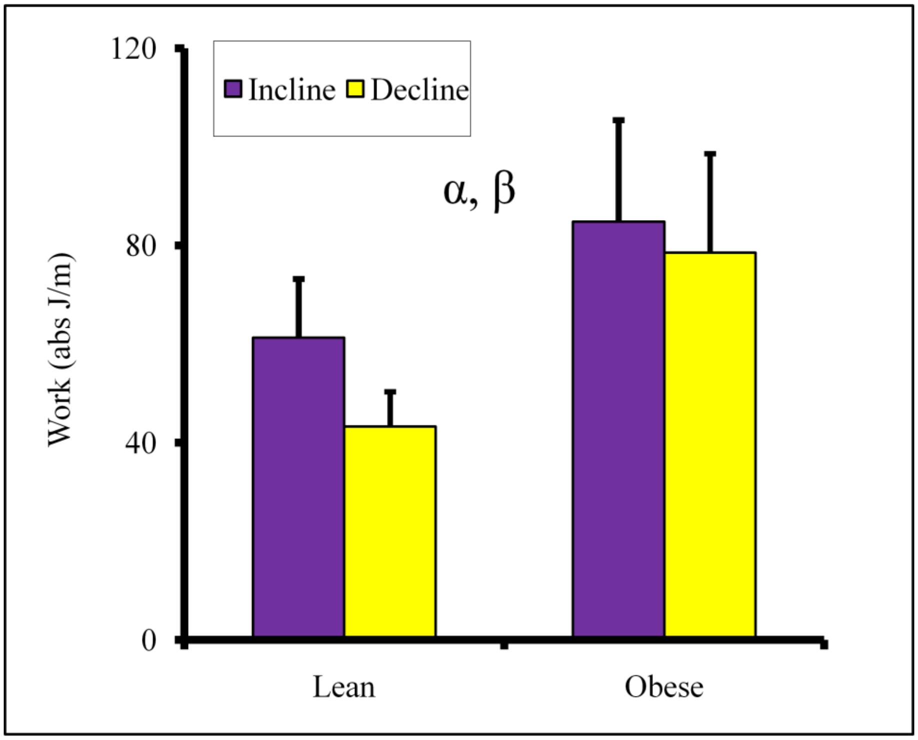


Figure 12 – Total muscle work performed by lean and obese adults during incline and decline walking (normalized to stride length)

α = Incline > Decline, p = 0.003; β = Obese > Lean, p < 0.001

during incline and decline walking. During incline walking lean adults performed a greater amount of positive muscle work compared to decline walking (82.79 vs. 28.79 J). Obese adults also performed a greater amount of positive muscle work during incline vs. decline walking (129.85 vs. 56.25 J). A significant mass x direction interaction in positive muscle work however showed that obese adults had a greater difference than lean adults (73.60 vs. 54.00 J, $p<0.001$) between incline and decline walking. Negative muscle work during decline vs. incline walking was higher in both lean (72.05 vs. 21.50 J) and obese groups (134.25 vs. 44.58 J). The greater increase in negative muscle work during decline seen by obese adults compared to lean adults (89.67 vs. 50.55 J) showed a significant mass x direction interaction ($p<0.001$).

Joint muscle work for the hip, knee, and ankle during incline and decline walking was calculated to show individual joint contributions to muscle work. There was no significant mass x direction interaction at the hip (Figure 15a). During incline walking there was significantly greater muscle work at the hip compared to decline walking (41.18 ± 14.10 J vs. 5.83 ± 6.09 J; p<0.001). Obese adults performed significantly more muscle work at the hip than lean adults (27.35 ± 22.06 J vs. 19.66 ± 18.96 J; p=0.001). The total amount of muscle work at the knee showed a significant mass x direction interaction (Figure 15b), as both lean and obese performed more total muscle work during decline vs. incline walking (obese 60.30 vs. 7.44 J, lean 35.30 vs. 4.12 J), but the difference in decline vs. incline walking was greater in obese than in lean (52.86 vs. 31.18 J; p<0.001). Total muscle work at the ankle showed no significant mass x direction interaction (Figure 15c). Incline walking produced a greater amount of total muscle work at the

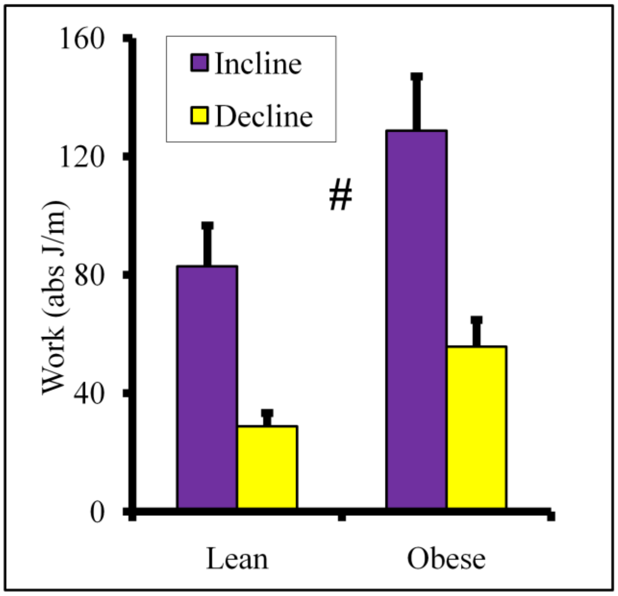


Figure 13 – Positive muscle worked performed by lean and obese adults during incline and decline walking (normalized to stride length)

= sig. mass x direction interaction, p < 0.001

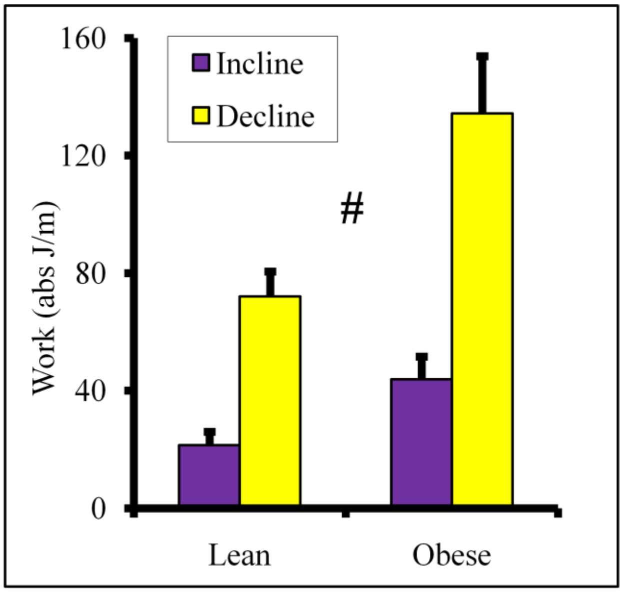


Figure 14 – Negative muscle worked performed by lean and obese adults during incline and decline walking (normalized to stride length)

= sig. mass x direction interaction, p < 0.001

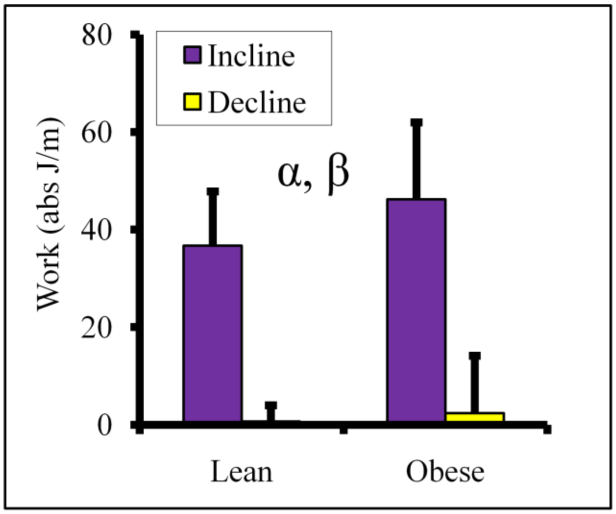


Figure 15a – Total muscle work at the hip performed by lean and obese adults during incline and decline walking (normalized to stride length)

α = Incline > Decline, $p < 0.001$; β = Obese > Lean, $p = 0.001$

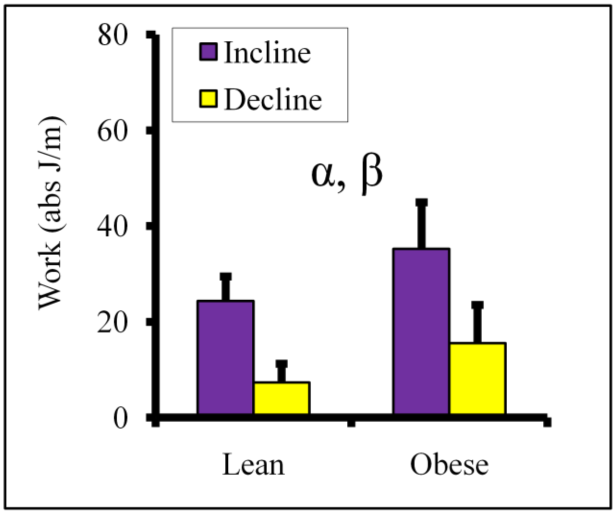


Figure 15c – Total muscle work at the ankle performed by lean and obese adults during incline and decline walking (normalized to stride length)

α = Incline > Decline, $p < 0.001$; β = Obese > Lean, $p < 0.001$

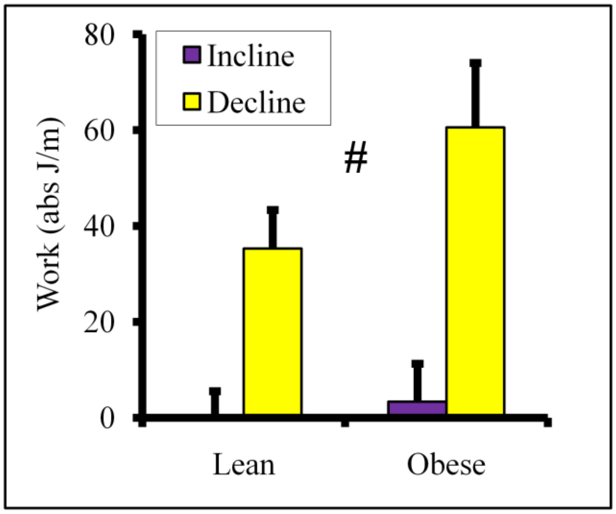


Figure 15c – Total muscle work at the ankle performed by lean and obese adults during incline and decline walking (normalized to stride length)

= sig. mass x direction interaction, $p < 0.001$

ankle when compared to decline walking (29.96 ± 9.38 J vs. 11.55 ± 7.34 J; $p<0.001$), and obese adults performed a greater amount of total muscle work at the ankle compared to lean adults (25.61 ± 13.21 J vs. 15.90 ± 9.63 J; $p<0.001$).

Specific to the hypothesis of this study that lower extremity muscles will perform more total muscle work during incline walking than during decline walking, these results showed that there were larger values for total net positive muscle work during incline walking for both lean and obese adults, than there were for total net negative muscle work during decline walking for lean and obese adults. There was no significant interaction for total muscle work however, related to the sub-hypothesis that obese adults will have a greater bias towards total muscle work during incline lean adults. The results of this study also showed that there were significant mass x interaction direction interactions as extensor angular impulse at the knee and average power at the knee showed a greater increase for obese than lean in decline vs. incline walking. A greater amount of total muscle work was performed by obese adults as compared to lean adults. Obese adults also showed a greater increase in positive muscle work during incline vs. decline walking as compared to the increase seen by lean adults, and obese had a greater increase in negative muscle work during decline vs. incline walking as compared to lean, both of which were shown to have a significant mass x direction interaction. The only significant mass x direction interaction at an individual joint was seen in the knee, as the obese group showed a greater increase in total muscle work during decline vs. incline walking compared to the increase in total muscle work seen by the lean group.

CHAPTER 5 - DISCUSSION

This study was conducted to investigate the biomechanical principle that muscles will perform more total net positive work during incline walking vs. total net negative work during decline walking, despite an equal change in total mechanical energy, and to compare the bias towards greater muscle work in incline between obese and lean individuals. We hypothesized that both lean and obese adults would have to perform a greater amount of total muscle work during incline walking than the total amount of muscle work during decline walking. Our sub-hypothesis was that obese compared to lean adults would show a larger bias towards more total muscle work during incline because of the increased vibration of soft tissue.

This study was designed to test the kinematic and kinetic differences between lean and obese adults while walking on an inclined surface and a declined surface at 1.5 m/s. This chapter will discuss the results to the literature and the hypothesis, and is organized in the following manner: 1) Development of the Hypothesis, 2) Quality of Data, 3) Incline and Decline Locomotion Biomechanics, 4) Work in Incline and Decline Gaits, and 5) Conclusions.

Development of the Hypothesis

During non-level gaits, muscles need to generate energy to lift the body up an incline, and during decline we need to dissipate energy to control the body as it lowers. When maintaining a constant velocity, this change in energy will reflect the change in potential energy that is brought by raising or lowering the body's center of mass. This change in potential energy is equivalent in incline and decline gaits of equal vertical displacement. Despite equivalent changes in total mechanical energy during incline and decline gaits, recent studies have shown that there is a bias towards muscles generating more mechanical energy during incline gait than they dissipate during decline gait [3, 10, 12].

One possible explanation for this bias is that while energy is generated solely by the concentric contraction of muscle, energy is dissipated both through the eccentric contraction of muscle and the vibrations and compression of soft tissues. When a soft tissue component was included in the biomechanical model, joint forces and joint torques were significantly reduced, suggesting that the soft tissues were involved in the dissipation of energy and the reduction of the joint forces and torques[2, 8, 15, 16]. In this study we assume that the greater body composition of adults with a body mass index greater than 35 kg/m^2 reflects a greater amount of adipose tissue that would be available for vibration, and consequently the dissipation of mechanical energy. To confirm that obese adults do have greater soft tissue vibrations compared to lean adults, we measured the average magnitude of acceleration for the right ASIS marker vibration throughout swing-stance cycle of declined gaits. The right ASIS marker was chosen as this is the area with the greatest discrepancy in adipose tissue between lean and obese adults. Obese adults had an average vibration magnitude of 7.56 m/s^2 compared to lean adults with an average vibration of 5.79 m/s^2 ($p<0.05$, t-test). Only decline gaits were chosen for this analysis because the impact of descending gaits is proposed to cause greater vibration and dissipate a greater amount of energy. The larger magnitude of acceleration seen in obese adults shows that overall they have more vibration of soft tissue compared to lean adults.

This previous review of literature led to the current hypotheses. The global hypothesis guiding this study was the generalized biomechanical principle that lower extremity muscles generate more energy during incline gaits than they dissipate during decline gaits, despite equivalent changes in total mechanical energy. The specific hypothesis tested in this study was that because of the increased soft tissue available for the dissipation of energy, the bias towards energy generation would be greater in obese adults than in lean adults.

Quality of Data

Both lean and obese groups similarly increased and decreased potential energy during incline and decline walking. Obese had 57% greater changes in potential energy, primarily due to the 67% larger body mass in obese and a 6% shorter stride length. Changes in kinetic energy for both groups were close to zero, demonstrating that subjects were able to maintain a constant velocity. These preliminary results verify that the change in mechanical energy reflected the change in potential energy, and this change was similar but not quite equal during both incline and decline gaits for both groups.

The observed differences in stride length between groups and gait directions led directly to the differences in potential energy changes. Stride length was larger in lean compared to obese adults and during incline compared to decline walking. These differences affect the amount of work by the body. Total work variables were therefore normalized by stride length to account for the changes in work that were caused by differences in stride length. The results therefore should be considered as based on unit distance a person walks and not per unit stride.

Biomechanics of Non-Level Gait

The larger mass of obese adults resulted in a 58% greater normal linear impulse and a 50% greater parallel linear impulse exerted by the ground reaction force compared to lean adults. However, there were no significant interactions suggesting that this difference between lean and obese adults was due to the greater body mass of obese and not a change in gait biomechanics.

Incline walking showed a slightly greater, and significant difference in normal impulse compared to decline walking. While decline walking had a greater peak GRF compared to incline walking, previous work has shown that the duration of stance phase is greater in inclined vs. declined gaits[3]. This increase in stance duration allows the force to be applied over a greater amount of time, which led to a greater impulse during incline walking.

Extensor angular impulse at the hip and ankle for each group was greater during incline walking. Obese adults had greater extensor angular impulses at the hip and ankle compared to lean adults. No significant mass x direction interactions existed at the hip and ankle though for extensor angular impulse. Similar trends existed for average power, with greater average powers at the hip and ankle during incline walking compared to decline walking, and obese adults had greater average power at the hip and ankle compared to lean adults, with no significant mass x direction interactions. The only significant mass x direction interactions occurred at the knee for extensor angular impulse and average knee power. Both lean and obese adults had a greater extensor angular impulse and average power at the knee during decline walking compared to incline walking. Obese adults however, showed a greater increase in the extensor angular impulse and joint power at the knee during decline walking compared to lean adults. These results indicated that obese adults used the knee to a greater extent than lean individuals to lower their center of mass in a more controlled manner. DeVita et. al. reported joint torque and power curves with values and patterns similar to those presented in this study for lean adults [3]. Lay et. al. reported hip, knee, and ankle joint moments in lean adults during incline, level, and decline walking [10]. Peak hip torque and ankle torque were both greater in incline walking compared to decline walking (hip: 1.93 Nm/kg vs. 0.75 Nm/kg, ankle: 1.95 Nm/kg vs. 0.92 Nm/kg), and peak knee torque was greater during decline compared to incline walking (1.18 Nm/kg vs. 0.81

Nm/kg). McIntosh also reported similar patterns of joint torque for lean adults, however their values were much lower than the lean adults in this and previous studies [12]. The difference with that study was possibly due to subjects walking barefooted at a self-selected speed, as opposed to this study which has subjects walking at a standardized speed wearing normal walking shoes. Each one of these studies reported different calculations that reflect joint torque and power, making them difficult to compare. By visual comparison and evaluation of the data reported however, we can see that these studies agree with results for the lean adults of this study, showing greater torque and power at the hip and ankle during incline walking, and greater torque and power at the knee during decline walking. Currently there are no studies that have investigated the biomechanics of obese adults on non-level surfaces, so comparisons with previous literature are limited to that of lean adults.

Work in Incline and Decline Gaits

Obese performed a significantly greater amount of total muscle work normalized to stride distance compared to lean adults on both inclined and declined gaits. This is consistent with the increased torque and power previously shown in the biomechanics of obese gait [1]. The increase in total muscle work in obese adults was expected and necessary to raise and control the lowering of the larger mass in obese adults. This same logic was reported in a previous study investigating lean and obese adults walking at different speeds on a level surface. Obese adults had larger torques and powers when compared to the lean adults, which is associated with an increase in the muscle work performed [1]. There was a greater amount of total muscle work performed during incline gait tasks compared to decline, which supported the global hypothesis that lower extremity muscles will perform a greater amount of total net positive muscle work during incline gait tasks compared to total net negative muscle work during decline gait tasks, despite equivalent changes in total mechanical energy. The greater amount of muscle work performed during incline gait tasks is consistent with previous work from DeVita [3], which reported 89 J/m of net positive muscle work during incline and -71 J/m of net negative muscle work during decline. Other studies have also showed greater amounts of positive muscle work during inclined gaits vs. negative muscle work during declined gaits [10, 12], however these studies only examined individual joint work in lean adults and they did not report the total muscle work.

The sub-hypothesis that obese adults will show a greater bias towards net positive muscle work during incline gaits vs. net negative work during decline gaits when compared to lean adults was not supported. During incline walking both lean and obese adults performed a greater amount of total muscle work compared to the total muscle work performed during decline walking, however the interaction effect between groups and gait directions was not significant. Therefore, it cannot be concluded that the positive bias in muscle work across gaits is different between lean and obese adults. Surprisingly, the sample means for the difference between total muscle work performed during incline and decline was unexpectedly smaller in obese adults compared to lean adults. These results occurred despite the increased soft tissue vibration that was seen in obese adults. This would suggest that the vibration of soft tissue is not an explanation for the positive bias of muscle function in non-level gaits.

The absence of the interaction effect in total muscle work might partially be explained by the significant mass x direction interaction seen when the total muscle work is partitioned into positive and negative components. During incline walking both groups performed a greater amount of positive muscle work than negative muscle work, but the obese group had a greater increase in the amount of positive muscle work performed. A similar interaction occurred during decline walking, as both groups performed more negative muscle work, but obese adults had a greater increase in the amount of negative muscle work performed during decline compared to lean adults. Obese adults even showed a greater amount of negative muscle work during decline compared to incline (-134.25 J/m vs. 129.85 J/m), while lean adults performed a greater amount of positive muscle work during incline than negative muscle work during decline (82.79 J/m vs. -72.05 J/m). The greater amount of negative muscle work seen in obese would cancel out the positive muscle work and would reduce the bias of the lower extremities to produce a greater amount of total muscle work during incline gaits. The greater amount of positive muscle work in lean individuals however, would contribute to the increasing bias towards positive muscle work during incline.

Individual joint work throughout the swing and stance phases of walking showed similar patterns to that seen earlier in the joint torques and powers in non-level gait biomechanics. Obese adults performed significantly greater muscle work than lean adults at all three joints. Hip and ankle work was greater during incline walking when compared to decline walking, and knee work was greater during decline walking compared to incline walking. A significant mass x direction interaction occurred at the knee where the obese group had a greater increase in knee work performed during decline vs. incline, when compared with the lean group. This interaction suggests that obese adults would use a greater amount of knee work to lower the body in a more controlled manner by lowering the center of mass at a slower rate. This more controlled lowering of the body could serve as a protective mechanism to reduce the high forces at the joints that are associated with heel strike of declined walking. This could also explain the lack of bias in positive muscle work seen in obese gait. The larger GRF that is seen at heel strike during decline walking was proposed to cause a greater amount of vibration, and the larger GRF of obese along with a greater amount of soft tissue would lead to a greater amount of vibration in obese vs. lean adults. As obese adults lowered themselves with a greater amount of knee work they may reduce the vibration caused by larger GRF of impact and therefore the proportion of energy that is dissipated through the vibration of soft tissue. This dampened vibration could more closely match the vibrations seen during incline walking, and then a bias towards total muscle work during incline walking would become less apparent, like is seen in the results of this study. During this study we used compression shorts and wraps to eliminate any movement of markers being associated with loose clothing, and we also used tape wrapped lightly around the thigh to hold marker plates in place. The restrictive clothing and tape could have also reduced the amount of vibration of the soft tissues, and taken away some of the dissipation of energy through soft tissue vibration.

The difference that is seen in muscle work performed by lean and obese adults during non-level gaits is an area that requires further study. Currently, there is not any known literature which investigates the non-level gait biomechanics of obese adults. To validate the results of this study, more work in this area will be required for comparison. The design of this study may also affect the test of the hypothesis. In the development of the design of the study body composition was assumed to be reflected by a difference in body mass index. Body mass index does not consider differences between fat mass and lean mass, so BMI may not accurately predict a greater amount of adipose soft tissue mass that was required in the formation of this study. Differences in body mass were also not considered. To more accurately test the effect that soft tissue has on energy dissipation during non-level gaits, groups of equal body mass with a significant difference in body composition should be compared. This may also reduce some of the effects that different biomechanics between lean and obese adults could have on the overall muscle work of non-level gait.

Conclusions

It was hypothesized a generalized biomechanical principle that a greater amount of energy would be generated through positive muscle work as adults walked up an incline compared to the amount of energy dissipated through negative muscle work during decline walking, despite having equivalent changes in total mechanical energy. This study supported this principle, and expanded it to include both lean and obese adults showing a bias towards greater muscle work performed during incline gaits. Our related sub-hypothesis that obese adults would have a greater bias towards energy generation compared to lean adults was also refuted. Part of this decreased bias towards energy generation in obese could be related to the increased amount of muscle work performed at the knee during decline walking, possibly allowing obese adults to lower themselves in a more controlled manner, as opposed to the more dynamic gait of lean adults.

To better understand the function of muscles during non-level gaits, further work needs to be done on the bias of muscle function towards energy generation. This study suggests that this principle may not spread across all populations, however without more work done the results of this study cannot be validated. Similar designs in study may also be warranted for investigating the effects of soft tissue on energy dissipation, most ideally comparing two groups of similar body mass with differing body compositions. Alternative explanations for the energy generating bias of muscle function seen in lean adults should also be investigated. One possible explanation is the more upright gait observed during decline walking, which could reduce the work done by the joints despite larger GRF upon impact. The muscle work needed to propel the body in a forward motion could also affect the work done by lower extremity muscles, and research done to investigate the amount of work used for forward propulsion of the body could show some insight into the positive bias of muscle function. The shoe element has also recently been proposed as a possible mechanism for the dissipation of energy through the compression of the sole during the heel strike, which would lead to greater dissipation during decline walking as the larger GRF would compress the sole of the shoe more than during incline walking [19]. In general, a greater amount of research needs to be done to investigate the energy generation and dissipation during non-level gaits across more populations, and to determine possible mechanisms for the possible bias that muscle function has towards energy generation during incline gaits over energy dissipation during decline gaits.

BIBLIOGRAPHY

1. Browning, R. C. and R. Kram. Effects of obesity on the biomechanics of walking at different speeds. *Med Sci Sports Exerc*. 39:1632-1641, 2007.

2. Challis, J. H. and M. T. Pain. Soft tissue motion influences skeletal loads during impacts. *Exerc Sport Sci Rev*. 36:71-75, 2008.

3. DeVita, P., J. Helseth, and T. Hortobagyi. Muscles do more positive than negative work in human locomotion. *J Exp Biol*. 210:3361-3373, 2007.

4. DeVita, P. and T. Hortobagyi. Age causes a redistribution of joint torques and powers during gait. *J Appl Physiol*. 88:1804-1811, 2000.

5. DeVita, P. and T. Hortobagyi. Obesity is not associated with increased knee joint torque and power during level walking. *J Biomech*. 36:1355-1362, 2003.

6. Devita, P., L. Janshen, P. Rider, S. Solnik, and T. Hortobagyi. Muscle work is biased toward energy generation over dissipation in non-level running. *J Biomech*. 41:3354-3359, 2008.

7. Eng, J. J. and D. A. Winter. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? *J Biomech*. 28:753-758, 1995.

8. Gruber, K., H. Ruder, J. Denoth, and K. Schneider. A comparative study of impact dynamics: wobbling mass model versus rigid body models. *J Biomech*. 31:439-444, 1998.

9. Lai, P. P., A. K. Leung, A. N. Li, and M. Zhang. Three-dimensional gait analysis of obese adults. *Clin Biomech (Bristol, Avon)*. 23 Suppl 1:S2-6, 2008.

10. Lay, A. N., C. J. Hass, T. Richard Nichols, and R. J. Gregor. The effects of sloped surfaces on locomotion: an electromyographic analysis. *J Biomech*. 40:1276-1285, 2007.

11. McFadyen, B. J. and D. A. Winter. An integrated biomechanical analysis of normal stair ascent and descent. *J Biomech*. 21:733-744, 1988.

12. McIntosh, A. S., K. T. Beatty, L. N. Dwan, and D. R. Vickers. Gait dynamics on an inclined walkway. *J Biomech*. 39:2491-2502, 2006.

13. Minetti, A. E., L. P. Ardigo, and F. Saibene. Mechanical determinants of gradient walking energetics in man. *J Physiol*. 472:725-735, 1993.

14. Nadeau, S., B. J. McFadyen, and F. Malouin. Frontal and sagittal plane analyses of the stair climbing task in healthy adults aged over 40 years: what are the challenges compared to level walking? *Clin Biomech (Bristol, Avon)*. 18:950-959, 2003.

15. Nigg, B. M. and W. Liu. The effect of muscle stiffness and damping on simulated impact force peaks during running. *J Biomech*. 32:849-856, 1999.

16. Pain, M. T. and J. H. Challis. The influence of soft tissue movement on ground reaction forces, joint torques and joint reaction forces in drop landings. *J Biomech*. 39:119-124, 2006.

17. Pain, M. T. and J. H. Challis. The role of the heel pad and shank soft tissue during impacts: a further resolution of a paradox. *J Biomech*. 34:327-333, 2001.

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

19. Sasaki, K., R. R. Neptune, and S. A. Kautz. The relationships between muscle, external, internal and joint mechanical work during normal walking. *J Exp Biol*. 212:738-744, 2009.

20. Spyropoulos, P., J. C. Pisciotta, K. N. Pavlou, M. A. Cairns, and S. R. Simon. Biomechanical gait analysis in obese men. *Arch Phys Med Rehabil*. 72:1065-1070, 1991.

21. Swanson, S. C. and G. E. Caldwell. An integrated biomechanical analysis of high speed incline and level treadmill running. *Med Sci Sports Exerc*. 32:1146-1155, 2000.

APPENDIX A

Biomechanics Laboratory
Tibor Hortobágyi, PhD
252.737.4564

Paul DeVita, PhD
252.737.4563

332 Ward Sports Medicine Building
East Carolina University
Greenville, NC 27858

Consent To Participate in the Research Project:
Gait Analysis in Obesity - Adults

I am being asked to voluntarily participate in this research project conducted by Tibor Hortobágyi, Ph.D. and Paul DeVita, Ph.D. The **purpose** of this study is to determine how children and adults who are light or heavy walk. Depending on how much I am interested in the study, I have the opportunity to volunteer for one or more 2-hour long sessions. There will be approximately 100 participants in this study over several years.

I understand that I must voluntarily agree to participate in this study. I understand that I may not participate in this study if I have certain medical conditions. I will not be able to take part in the study if one of my legs is longer than the other leg, I was born with cerebral palsy, my feet have grown abnormally (for example, club foot), have had a stroke, have: cancer, glucose levels higher than 126 mg/dL, blood pressure higher than 140/90, abnormal heart rhythms, a joint replaced in my leg, osteoporosis, pain in legs from unknown cause, peripheral neuropathy, peripheral vascular disease, low red blood cell count, a history of multiple falls, a history of spinal surgery, a pacemaker, a cane to walk, visual impairment that hinders my gait, and my body weight does not meet study requirements.

Procedures: I will be involved in the procedures that are indicated with a check (√) mark:

<u>Preparation:</u> The preparation lasts about 30 minutes. In the laboratory, I will be asked to put on shorts that will be provided, T shirt, and wear a comfortable pair of athletic shoes with white socks (tennis, running shoes are acceptable). (I can bring my own clothing as well.) I will be asked to kick a ball 3 times to determine which leg is the dominant leg. About 25 reflective markers will be attached to the clothing or taped on the skin. These markers look like tiny balls and brightly reflect light. Finally, 9, small self-adhesive electrodes will be placed on the muscles of the front and back of thigh and leg. These electrodes detect the activity of muscles.

<u>Walking tests:</u> These tests last about 1 hour and 20 minutes. I will have the opportunity to practice each task until I feel comfortable to perform them. I will practice to walk at a given speed and step with my dominant leg on a metal plate that measures the forces under the foot.

______ One task will be to simply walk at a pace I choose.

______ For the second task, one of the staff persons in the laboratory will tell me how fast to walk. This pace will be probably a little bit faster than the pace in the first task.

______ In the third task I will be asked to jog.

______ The fourth task is to climb a stairwell of 4 steps and then descent the same stairwell.

______ Finally, I may be asked to walk up and down on an incline surface (like a ramp).

I will be asked to perform each task 5 times, totaling about 30-40 trials. The distance I will walk is about 20 feet in each task. I can sit down and rest between tasks or trials.

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FROM 5-6-09
TO 5-5-10

Subject's Initials ________

______ Leg strength test. This test lasts about 10 minutes. I will be asked to perform this test either before or after the walking tests. Leg strength will be measured on machine that looks like a leg press in a gym. I will sit on the seat of this special leg press machine. Comfortable straps will be put around the trunk, hips, and dominant leg to help me give a good effort. The ankle of the dominant leg will be strapped to a padded cuff. I will practice a few times using mild, medium, and strong efforts to press with my leg into the footplate of the machine. After a few minutes of rest, I will be asked to exert as hard as I can for 5 seconds. I will rest for 1 minute and repeat the effort 2 more times with 1 minute of rest between efforts. Neither the machine nor my leg will move.

Risks: Any tests that require maximal effort represent risks in terms of high blood pressure, stroke, heart attack, temporary pain, and muscle strain or joint sprain. Such test is the Leg Strength Test. The Walking tests represent low risks although during rapid walking temporary breathlessness or dizziness may develop. Ascending and especially descending stairs can be hazardous for individuals who are heavy and cannot fully see the steps. All these risks will be reduced by: allowing me to participate only if I feel comfortable performing the tasks; by carefully explaining and demonstrating the tasks, and by having me properly warmed up for and thoroughly familiarized with the tests.

Benefits: All results will be explained to me. If I am interested in being counseled on the health benefits of physical activity, the laboratory staff will make every effort to freely refer me to an ongoing physical activity program on the ECU campus or elsewhere. I will be entitled to receive $25 per session. The payment will be available to me upon the completion of the study or will be prorated in proportion to the extent of participation.

Withdrawal, Injury, Confidentiality: The nature and purpose of the procedures, the known risks involved, and the possibility of complications have been explained to me and I understand them. I understand that not all risks and side effects of these procedures are foreseeable.

I understand that participation in this study is voluntary and refusal to participate will involve no penalty or loss of benefits to which I am otherwise entitled. I may discontinue participation at any time without penalty. The policy of East Carolina University does not provide for compensation or medical treatment for subjects because of physical or other injury resulting from this research activity. However, every effort will be made to make the facilities of the School of Medicine available for treatment in the event of such physical injury.

I understand that my personal data will be held in strict confidence by the investigators. I understand that if any publications result from this study my name or any identifiable codes will not be used. This experiment does not produce any "video" image of my face or body. The recording equipment is sensitive only to the reflective markers and does not record, digitally or in any other form, my face or body.

Contact person. If I have any questions about the research or possible research-related injury, I may contact Dr. Hortobágyi at home (252.355.7715) or work (252.737.4564). Also, if questions arise about my rights as a research subject, I may contact the Chair of the University and Medical Center Institutional Review Board (252.744.2914).

I have read the above material and Dr. Hortobágyi has it explained to me. I have been encouraged to ask questions about the study and all inquiries have been answered to my satisfaction. I will receive a copy of this consent form.

____________________	______________________________	_______
Subject's Name (Print)	Signature of Subject	Date
______________________________	______________________________	_______
Witness's Name (Print)	Signature of Witness	Date
______________________________	______________________________	_______
Principal Investigator's Name (Print)	Signature of PI	Date

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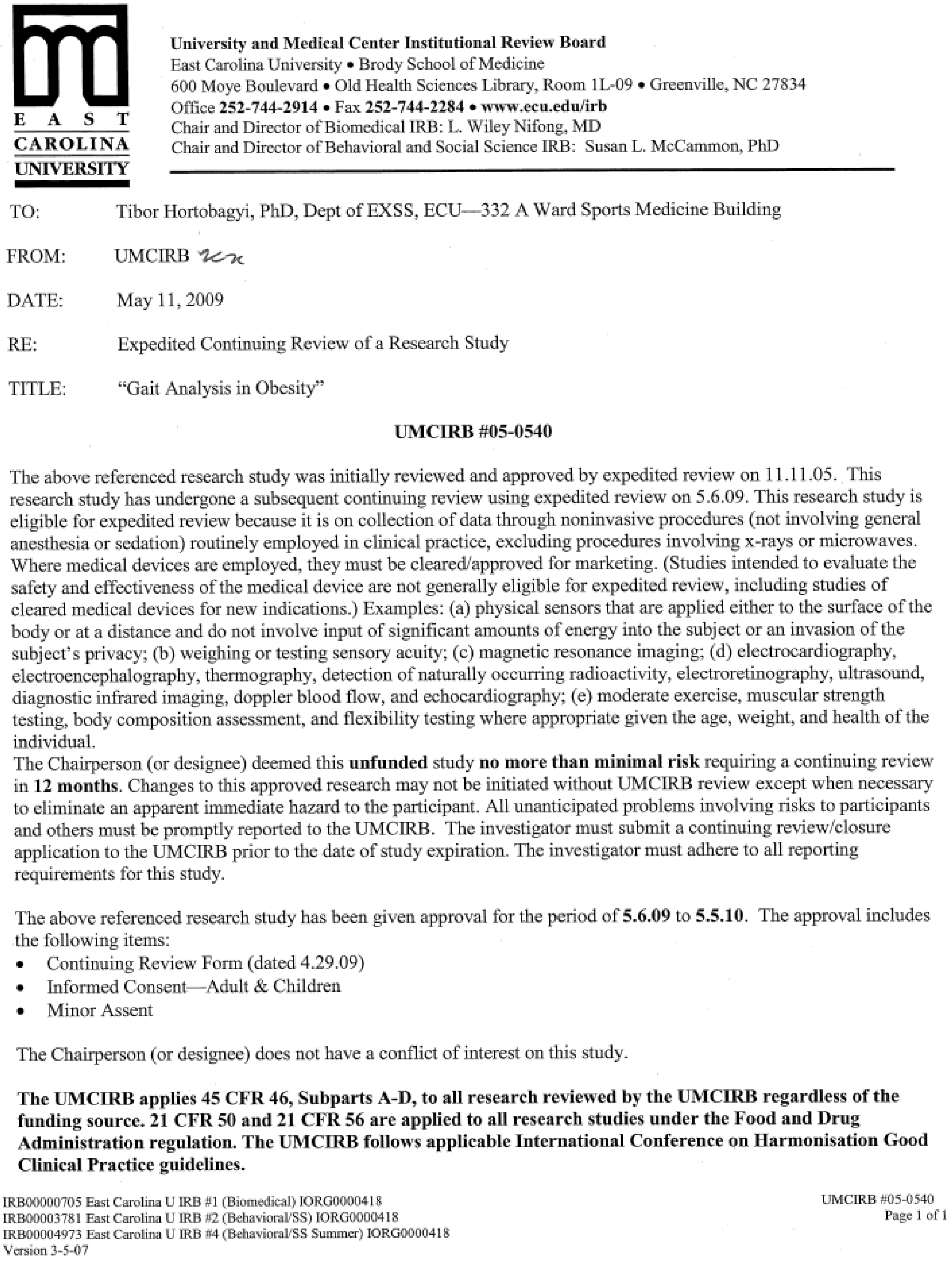
APPENDIX B



University and Medical Center Institutional Review Board
East Carolina University • Brody School of Medicine
600 Moye Boulevard • Old Health Sciences Library, Room 1L-09 • Greenville, NC 27834
Office **252-744-2914** • Fax **252-744-2284** • **www.ecu.edu/irb**
Chair and Director of Biomedical IRB: L. Wiley Nifong, MD
Chair and Director of Behavioral and Social Science IRB: Susan L. McCammon, PhD

TO: Tibor Hortobagyi, PhD, Dept of EXSS, ECU—332 A Ward Sports Medicine Building

FROM: UMCIRB

DATE: May 11, 2009

RE: Expedited Continuing Review of a Research Study

TITLE: "Gait Analysis in Obesity"

UMCIRB #05-0540

The above referenced research study was initially reviewed and approved by expedited review on 11.11.05. This research study has undergone a subsequent continuing review using expedited review on 5.6.09. This research study is eligible for expedited review because it is on collection of data through noninvasive procedures (not involving general anesthesia or sedation) routinely employed in clinical practice, excluding procedures involving x-rays or microwaves. Where medical devices are employed, they must be cleared/approved for marketing. (Studies intended to evaluate the safety and effectiveness of the medical device are not generally eligible for expedited review, including studies of cleared medical devices for new indications.) Examples: (a) physical sensors that are applied either to the surface of the body or at a distance and do not involve input of significant amounts of energy into the subject or an invasion of the subject's privacy; (b) weighing or testing sensory acuity; (c) magnetic resonance imaging; (d) electrocardiography, electroencephalography, thermography, detection of naturally occurring radioactivity, electroretinography, ultrasound, diagnostic infrared imaging, doppler blood flow, and echocardiography; (e) moderate exercise, muscular strength testing, body composition assessment, and flexibility testing where appropriate given the age, weight, and health of the individual.

The Chairperson (or designee) deemed this **unfunded** study **no more than minimal risk** requiring a continuing review in **12 months**. Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The investigator must adhere to all reporting requirements for this study.

The above referenced research study has been given approval for the period of **5.6.09** to **5.5.10**. The approval includes the following items:

- Continuing Review Form (dated 4.29.09)
- Informed Consent—Adult & Children
- Minor Assent

The Chairperson (or designee) does not have a conflict of interest on this study.

The UMCIRB applies 45 CFR 46, Subparts A-D, to all research reviewed by the UMCIRB regardless of the funding source. 21 CFR 50 and 21 CFR 56 are applied to all research studies under the Food and Drug Administration regulation. The UMCIRB follows applicable International Conference on Harmonisation Good Clinical Practice guidelines.

IRB00000705 East Carolina U IRB #1 (Biomedical) IORG0000418
IRB00003781 East Carolina U IRB #2 (Behavioral/SS) IORG0000418
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