CONTROL OF WALKING SPEED IN YOUNG AND OLD ADULTS

by

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May 2012

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Walking is a basic form of locomotion and walking velocity is a good predictor of human health, with faster velocities indicating better health. While faster walking velocities have been attributed to overall increases in lower extremity joint torques and powers, the precise relationships of torque and power outputs at each joint to walking velocity remain elusive. We proposed that these relationships are fundamental in developing effective training programs aimed at increasing walking velocity in mobility-challenged populations such as older adults. Based on the previously established argument that aging induces asymmetric changes in joint torques and powers, we hypothesized that walking velocity is more strongly related to hip torque and power than ankle torque and power in old adults than young adults. The purpose of this study was to identify the relationships among maximum, sagittal plane hip, knee, and ankle joint torques and powers and walking velocity in young and old, healthy adults in order to identify biomechanical correlates associated with modulating walking velocity. Gait biomechanics were collected using 8 camera 3D motion capture and force platform systems. Twenty-two young healthy

adults and twenty-two old healthy adults each walked at 20 speeds ranging from relatively slow to relatively fast velocities. Maximum sagittal plane joint torques and powers derived through inverse dynamics were correlated to walking velocity ($p<0.05$). The data showed that all peak torques and powers were significantly ($p<0.05$) and directly related to walking velocity. Hip and ankle relations were curvilinear upward and downward; knee were linear. Peak joint torques showed distal to proximal decrease in the strength of their relationships. Peak joint powers were similarly related to walking velocity at each joint. In old adults, torques were most strongly related to walking velocity at the proximal hip (R^2=0.761) and decreased distally to the ankle (R^2 = 0.275); powers were similar to walking velocity across the hip (R^2=0.467) and the ankle (R^2=0.445). In young adults, torques were also most strongly related to the walking velocity at the proximal hip (R^2=0.747) and decreased distally to the ankle (R^2=0.289); powers were similarly related to walking velocity across the hip (R^2=0.601) than to the ankle (R^2= 0.528). Overall, mechanical output at the hip was the primary biomechanical correlate of walking velocity, whereas mechanical output at the ankle was most weakly correlated to walking velocity. This pattern of modulating walking velocity is used by both young and old adults similarly. Therefore, the hypothesis was refuted, as aging does not cause a mechanical plasticity in relation to walking velocity.

Control of Walking Speed in Young and Old Adults

A Thesis Presented to

The Faculty of the Department of Exercise and Sport Science

East Carolina University

In Partial Fulfillment of the Requirements for

The Masters of Science in Exercise and Sport Science

Biomechanics Option

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May 2012

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CHAPTER 1: INTRODUCTION

Walking is the fundamental form of human locomotion. It is the action of transporting our body masses through reciprocal lower extremity movements in which at least one foot is in contact with the supporting surface. Walking is relatively safe and has moderately low metabolic and biomechanical demands (Mian et al., 2006). Despite these qualities however, walking is a complex motor task involving a high number of degrees of freedom often moving in asynchronous patterns (Minetti et al., 2011). Thus, walking can be challenging to populations with motor deficits (Wert et al., 2010). Variations in the most basic biomechanical characteristics of walking can be indicative of such deficits (Richards et al., 2010).

One such characteristic is walking velocity, which is a good predictor of human health. Gait speed is a strong indicator of mortality and morbidity in old adults (Studenski et al., 2010). Not only can death and hospitalization events be foreseen by gait speed, but severe lower extremity limitation can also be predicted. Those adults with slower gait speeds have been seen to have higher rates of these three events, with the highest rate associated with hospitalization (Cesari et al., 2005). Recently, the evaluation of walking speed has been used as an incremental predictor of which hospital patients will experience a major adverse post-operative event after undergoing cardiac surgery and which will not. Slow gait speed is widespread among patients with cardiovascular disease and is foretelling of poor results following cardiac surgery (Afilalo et al., 2010). Not only is walking velocity related to physical health, but mental health as well (Heyn et al., 2004). Cognitive health predictions have been illustrated by levels of anxiety about falling in old adults. Those adults with higher concerns of falling have slower overall walking velocity (Delbare et al., 2009). Atherosclerosis can also be used as an example of altered health conditions due to faster walking velocity. Compared with adults that have a lower coronary calcium score, those adults with higher levels of atherosclerosis have slower walking velocities (Hamer et al., 2009).

Not only can gait speed be used as a predictor of human health, but by manipulating gait speed, health can also be manipulated. The connection between alterations of gait speed and changes in brain functionality in old adults demonstrates this idea. The introduction of exercise training has been associated with functional plasticity in large-scale brain systems in the aging brain (Voss et al., 2010). By adding walking as an exercise to their daily activities, the aging adults increased their brain connectivity. This rise in connectivity is viewed as an increase in cognitive health.

Understanding the biomechanical factors that produce walking velocity may enable the creation of more effective training programs in order to increase preferred walking speed in old adults. This increase in walking velocity will likely increase health status as well (Malatesta et al., 2010). It is already known that walking velocity is determined by muscle function (Neptune et al., 2008). In fact, walking velocity is directly associated with the magnitude of muscle activation (Hortobagyi et al., 2009). This activation spreads throughout all of the lower extremity joints including the hip, the knee and the ankle (Neptune et al., 2008). It is the modulation in muscle activation that leads to altering the production of torques and powers in the joints, which collectively produce human movement and which would modulate walking velocity. Torques and powers are positively correlated with muscle activations; as muscle activations increase, joint torques and powers increase as well. This relationship has been seen in all three of the lower extremity joints (Stoquart et al., 2008). All three joints show a positive relationship between torque and walking velocity, meaning that as torque increases, walking velocity increases as well (Neptune et al., 2002). Not only does torque increase, but power increases with increased walking velocity also (Lelas et al., 2003). The torques of each lower extremity joints are also correlated with walking velocity.

Many studies have shown the contributions each muscle makes towards modulating walking velocity in young adults. These contributions are not as well understood in old adults. However, it is known that aging reduces muscle properties, which results in old adults having slower walking velocities as compared to young adults (Kerrigan et al., 1998). In addition to reduced walking velocity, a reduction in both muscle power and strength (Metter et al., 1997). It is interesting, however, that this overall reduction caused by the aging process is not symmetrical throughout the lower extremity joints and muscles; there is an asymmetric aging process in both muscle properties and motor performance (DeVita et al., 2000). There is a greater loss of muscle mass in the distal muscles, such as the soleus, as opposed to the more proximal muscles such as the adductor longus (Bua et al., 2002). This loss of muscle mass may be related to old adults relying more on their proximal hip muscles to walk than young adults, which causes a distal to proximal shift in muscle function with age (DeVita et al., 2000). There is a greater joint torque used in the hip during walking in old adults than young adults. The knee and ankle are the opposite however, with young adults having greater torques in those joints when compared to old adults. This larger hip torque in old adults is counterbalanced by a smaller ankle ground reaction force than seen in young adults (Karamanidis et al., 2007). Age causes a redistribution of torques at the lower extremity, showing that mechanical plasticity exists with age (Kerrigan et al. 1998). This redistribution alludes to the idea that there are differences in the manipulation and control of gait velocity between old and young adults.

Hypothesis

Old adults have a stronger relationship between hip torque and power, and walking velocity, and a weaker relationship between ankle torque and power, and walking velocity than young adults. In short, age creates a redistribution of joint torques and powers in the control of walking velocity.

Purpose

The purpose of this study is to identify the relationships among lower extremity joint torques and powers and walking velocity in young and old adults. This study will also compare these relationships between the age groups.

Significance

Previous literature shows support for the idea that a faster walking velocity is associated with greater human health, both physiological and psychological. Therefore, a faster walking velocity is associated with not only overall improvement in health, but also in the maintenance of health in old adults. By introducing appropriate walking programs to the old population with the desired outcome of increase preferred walking velocity, one may increase health and survival rate, and decrease that of morbidity. The clinical application of this study is that old adults will be able to become more mobile and therefore more physically active, which leads to a healthier life. In doing so, this study will lead to maintaining both walking ability and faster walking speeds. Knowledge and data from this study will contribute to an overall higher physical capacity in old adults.

Delimitations

1. All subjects will be healthy mobile old and young adults with no previous history of any musculoskeletal or neuromuscular diseases.
2. Young adult subjects will be males and females between the ages of 18-25 and old adult subjects will be males and females between the ages of 68-85 years.
3. Subjects do not need assistance or have difficulty with performing activities of daily living (ADLs).
4. All subjects will have a Body Mass Index of less than 30 kg/m^2.
5. The subjects will be walking on a level surface only.
6. There will be measurements of the lower extremity, including the hip, knee and ankle, only.

Limitations

1. The accuracy of the 3-D Motion Capture position data.
2. Resolution of force data accuracy from force plate.

CHAPTER 2: REVIEW OF LITERATURE

The purpose of this study is to identify the relationships among lower extremity joint torques and powers and walking velocity in young and old adults. This study will also compare these relationships between the age groups. This chapter will review literature of previous research examining 1) current methods in biomechanics 2) biomechanical characteristics influencing walking velocity 3) a distal to proximal shift in muscle function with age.

Current Methods in Biomechanics

Similar methods have been used in multiple studies regarding human gait analysis. The main components of a gait analysis study include the use of three-dimensional motion capture, force plates, and inverse dynamic analysis. Three-dimensional analysis tracks reflective markers placed on the subject and allows further analysis after data collection. Force plates are used to measure ground reaction forces when stepped on by the subject. One of the first uses of motion capture analysis was shown by E.J. Marey in 1873. Marey is known for contributing chronophotography to the field of research. He was able to create geometric photography by dressing his subject in all black, placing shiny metal rods and buttons on his body to represent joints and segments, and asking the subject to walk in front of a black screen in the sunlight while being photographed. The pictures that were produced showed only the points and lines, which were then used in subsequent calculations (Michaelis, 1966).

Inverse dynamic analysis is a large component of human gait analysis. This analysis is used to calculate the joint torques by multiplying mass, usually of a single limb segment, and acceleration. Herbert Elftman was the originator of inverse dynamics as he used it to calculate the joint torques of the lower extremity in 1938. Elftman used inverse dynamics to calculate not only one-joint muscles, but also to create a model of three-joint muscles (Elftman et al., 1938). This study aided in the popularity of inverse dynamics and may be the reason the analysis is still used today. In 1950, Bresler and Frankel advanced inverse dynamic analysis with their publication entitled “The Forces and Moments in the Leg during Level Walking”. Bresel and Frankel focused on the effects of mass distribution, power supply and mannerisms of motion on displacement and the forces involved in the locomotion process. They stated that these were a few of the factors that introduced complications into the analysis of the experimental data. In this study, the researchers stated inverse dynamic analysis as the mass distribution of the leg as defined in terms of the weight (or mass), location of the center of mass, and the mass moment of inertia about the center of mass. Both force plates and a type of motion capture were also used in this study. A main emphasis of this study was the fact that Bresel and Frankel computed all of the calculations necessary by hand. They completed 14,000 numerical calculations with 72 curves plotted for each segment and 24 curves were subjected to graphic differentiation. The entire process of completing the calculations for one subject’s limb segment took 500 man hours. However, they later explained that with practice they were able to complete the calculations in only 250 hours (Bresler and Frankel et al., 1950).

Inverse dynamic analysis became even more popularized by David Winter. Winter published a study in 1980 in which he stated that the principle of the lower limb support was that the algebraic summation of all the extensor torques at the hip, knee and ankle must be positive during the stance phase. He also stated that the support of the lower body requires net extensor activity at the hip, knee and ankle and that a net torque at any of the three lower extremity joints which prevent collapse will contribute to lower limb support. Winter examined the joint torques for the hip, knee and ankle in old and young males and females and used the standard link segment kinematic program suggested by Bresler and Frankel to calculate the vertical and horizontal reaction forces plus the net joint torques at each joint for one complete stride. One of Winter's main principles was that kinematic assessments of gait should examine the total limb, not just a single joint (Winter et al., 1980). Winter opened the door for other researchers to use these equations and practice biomechanics. These publications have expressed the idea that though the overall motor patterns may be visible, these patterns need to be interpreted by the net product of the joint. While inverse dynamic analysis may still have some limitations, it gives researchers the ability to identify the biomechanics of joint torques, the fundamental cause of animal movement patterns. Examination of the literature shows joint torques and powers analysis has identified these variables as critical in walking. Therefore, these factors will be studied.

Biomechanical Characteristics Influencing Walking Velocity

There are many kinematic factors contributing to human movement including joint torques, joint powers and muscle activation. In regards to the relationship between joint torques and walking velocity in young adults, previous literature has shown a positive correlation between these factors. As walking velocity increases, the torques of the hip, knee and ankle increase as well. The peaks for both the hip and knee torques also increase, however the ankle torque does not (Stoquart et al., 2008). Observation of the lack of increase in the ankle torque creates the need for more research to be completed in this area of interest.

The hip torque of young adults has the strongest correlation with walking velocity, with the knee being slightly less positive, and the ankle showing the least positive relationship (Lelas et al. 2003). Joint torque is only one of the factors that relates to walking velocity. Both power and strength are affected during the aging process as well, which are both directly related to walking velocity. Aging reduces both muscle power and strength in men and women (Metter et al., 1997). In addition to these factors, a feature of walking as simple as step width varies between young and old adults. Old adults have a greater variability in step width than do young adults (Owings et al., 2003). Total work is also an important factor of walking velocity and gait preference. There are two types of work; positive and negative. Though there may be greater positive work done across all three lower extremity joints during level walking, the work done at each joint varies. All three joints have both positive and negative work, but the hip and the ankle have more positive work than negative causing the net work done at both of those joints to be positive. The opposite is true for that of the knee. Also, while evaluating the lower extremity as a whole, there is an overall net positive work done during level walking (DeVita et al., 2007). Research has also shown that an increase in muscle activation corresponds to a faster walking speed (Neptune et al., 2008). Muscles within each lower extremity including the gastrocnemius (an ankle muscle), the vastus lateralis (a knee muscle) and the biceps femoris (a hip muscle) show increased excitation during the gait cycle when walking speed is increased. Muscle activation initiates the process of muscle contraction leading to muscle force, torque, and power production. Larger muscle activations produce greater torques and powers.

Distal to Proximal Shift in Muscle Function with Age

Skeletal muscle is significantly degraded in the elderly population (Gallagher et al., 1997). In addition, there is an asymmetric aging process in both muscle properties and motor performance. This asymmetry comes from the observation that there is a distal to proximal shift in muscle function with age (DeVita et al., 2000). Old adults have more muscle degradation in their distal extremities, such as the ankle, than they do in their more proximal extremities, such as the hip. Bua completed a study in which the muscle mass in older rats were compared to those of younger rats. The results of the study showed that there was a greater loss of muscle mass in the distal soleus muscle than the more proximal adductor longus muscle. Therefore, aging has a greater degradation effect on distal muscles as opposed to those located more proximally (Bua et al., 2002).

This shift in muscle function causes old adults to rely more on their proximal muscles. Shimada demonstrated this idea by using a PET scan to identify glucose metabolism in both young and old adults after a walking exercise. The PET scan shows fluorodeoxyglucose (^{18}F) more visibly in the hip muscles of the old adults, as compared to other lower extremity muscles. The increased coloration in the hip region demonstrates that more glucose metabolism is occurring, which is an indicator of muscle activation (Shimada et al., 2009).

DeVita and Hortobagyi clearly observed a distal to proximal shift caused by aging. Results showed the individual torques of each joint of the lower extremity and also the sum of these torques compared between old and young adults during walking. It is interesting that the sum of the torques was the same for the young and old adults, meaning that the total output of the joints were the same, however each individual joint torque differed between the two age groups. This overall product was the combination of different contributions from each joint. When studying each individual joint, the old adults used had greater torques at their hip and less at the knee and ankle, meaning that old adults rely more on their hip torques than young adults. Plasticity exists when there is a change in walking speed and old adults adapt to a faster walking velocity by manipulating their muscle function, using more proximal muscles as opposed to distal. This shows that there is an overall change in motor strategy to produce walking locomotion in old adults (DeVita et al., 2000).

Summary

Biomechanical factors involved in walking velocity are themselves affected by age. Joint and muscle torques, power, and muscle activation all collectively contribute to an overall decrease in walking velocity in old adults. Joint and muscle torques decrease with an increase in age and seem to cause a greater degradation in distal muscle mass, a greater usage of the more proximal hip muscles, and less of a reliance on the knee and ankle joints. However, power has more of a symmetrical decrease across all lower extremity joints as one ages. Muscle activation is also affected by age, causing a change in the timing of muscle activity. Although previous literature discusses the idea that old adults walk more slowly than young adults, little is known about the specific biomechanical factors that influence this decrease and what happens when walking velocity is manipulated.

CHAPTER 3: METHODOLOGY

Subjects Characteristics

The following subject inclusion and exclusion criteria were used:

Inclusion Criteria:

1. Healthy and mobile with no previous musculoskeletal injuries or conditions of the lower extremities.
2. Free of pain or difficulty performing activities of daily living.
3. Body mass index of less than 30.0 kg/m^2.
4. Provide written informed consent.

Exclusion Criteria:

1. Difficulty performing activities of daily living.
2. Smoking cigarettes currently or within the past five years.
3. Cardiovascular problems including heart attack and uncontrolled high blood pressure.
4. Musculoskeletal problems including arthritis, osteoporosis, joint replacement, lower extremity or back surgery.
5. Neurological problems including stroke and Parkinson’s disease.

	N	Age	Mass	Height	BMI
Young	22	20.3 ± 1.5	69.1 ± 12.1	1.7 ± 0.01	23.3 ± 2.1
Old	22	73.5 ± 4.8	69.4 ± 12.6	1.69 ± 0.09	24.1 ± 3.5

Table 1. Subject characteristics

There were an equal number of old and young adults used in this study, with 22 participants in each group. The mean age of the young group was 20.3 years of age with a standard deviation of 1.5 years. The mean age of the old group was 73.5 years of age with a standard deviated of 4.8 years. Though the age difference between the groups was great, the differences between mass, height, and BMI between old and young adults was not. The mean mass of the young group was 69.1 kg with a standard deviation of 12.1 kgs. The mean mass of the old group was 69.4 kg with a standard deviation of 12.6 kgs. The mean height of the young group was 1.7 meters with a standard deviation of 0.01 meters. The mean height of the old group was 1.69 meters with a standard deviation of 0.09 meters. Due to the exclusion and inclusion criteria, each individual had a Body Mass Index below 30 kg/m^2. This caused the mean Body Mass Index of each group lower than 30 kg/m^2 as well. The mean Body Mass Index of the young group was 23.3 kg/m^2 with a standard deviation of 2.1 kg/m^2. The mean Body Mass Index for the young group was 24.1 kg/m^2 with a standard deviation of 3.5 kg/m^2.

Equipment

Walking kinematic and kinetic data were collected with eight QualisysProReflex MCU 240 cameras (Qualisys Medical AB, Gothenburg, Sweden) at 120 Hz. A force platform (AMTI Model LG-6, Newton, MA) located in the center of the walkway measured the ground reaction forces at a frequency of 960 Hz and a gain of 4000. Gait speed of each subject was collected using an infrared timing system (Brower timing system, model IRD-T175, Salt Lake City, Utah). Cadence, stride length and velocity were collected for each subject using the GaitRite system (CIR Systems Inc., Havertown, PA). All of the data was collected using Qualisys Track Manager Software (Innovision Systems Inc., Columbiaville, MI) and then analyzed by Visual 3D (C-Motion Inc., Rockwille, MD). The height and weight for all subjects were measured and recorded in meters and kilograms using a Seca 703 scale (Seca gmbn & Co.kg, Hamburg, Germany). Electromyography was collected using an EMG Myopac (Konigsberg Instruments, Pacedena, CA).

Procedures

Young subjects were recruited from the ECU campus via fliers and classroom announcements. Old subjects were recruited via newspaper advertisements and contacting participants from previous studies. Individuals interested in participating in the study were contacted via telephone and an interview was administered by a research associate. The telephone interview screened prospective subjects in order to determine if they were eligible to participate in the study based on the inclusion and exclusion criteria. Interview questions included a spread of past and present medical history, such as cigarette smoking and previous surgeries. If the subject was eligible to participate, he or she was scheduled for future data collection.

Data collection was completed in one session usually lasting about 90 minutes. All of the testing was conducted in the Biomechanics Laboratory, located in room 332 of Ward Sports Medicine Building, East Carolina University, Greenville, NC. Before testing began, subjects were asked to read and sign a consent form. The height and weight of each subject were also taken before data collection started. Subjects were asked to wear tight fitting shorts and their own athletic, non-reflective shoes.

Subjects were instructed to walk along the GaitRite mat in order to obtain their preferred cadence, walking velocity, and stride length for both their right and left feet. Before walking, subjects were instructed to “walk at their normal walking pace”. Electromyography electrodes were placed on four muscles including the lateral gastrocnemius, vastus lateralis, biceps femoris, and tibialis anterior. In preparation of the skin, the area where the electrode was to be placed was shaved, swabbed with alcohol, scrubbed with an abrasive exfoliate and swabbed with alcohol again. In order to locate the muscle belly of the muscle of interest, subjects were asked to show slight resistance in different means for each of the four muscles. The electromyography pack was then placed on the back of the subject and tied into place. Each electrode was connected to a corresponding wire from the pack, and the pack was connected to the computer via another lengthy wire. Once the electrodes were in place and the pack turned on, the subject conducted maximum isometric voluntary contractions for each of the four muscles of interest with a research assistant applying resistance. This data was collected while recording the muscle activation using the Qualisys Track Manager Software program. The electrodes were then secured with small strips of medical tape.

Reflective markers were placed on the subject on specific body joints and segments which included the greater trochanters (left and right), metatarsal heads, knees, ankles, iliac crests (left and right), heel, shank, thigh, anterior superior iliac spine (left and right), and the lumbosacral joint (L5S1). After recording a five second static trial in the anatomical position with arms crossed over the chest, calibration markers were removed and a second static trial was recorded. Calibration markers included the greater trochanters (left and right), metatarsal heads, knees (medial and lateral), ankles (medial and lateral) and iliac crests (left and right).

Subjects received instruction as to which speed to walk down the walkway and the appropriate starting position for each individual trial. For the first walking trial, subjects were told to walk at their preferred walking speed. Subsequent trials included the subjects being told to either walk slower or faster, with the instructor being sure to choose a random order. Data for trials where the subject did not step on the force plate or in which the subject adjusted his or her stride to contact the force platform were excluded and re-recorded. A successful trial included the subject walking down the walkway, stepping on the force plate with the right foot, and maintaining both a constant speed and normal walking gait throughout. During data collection an excel spreadsheet in which speed, normalized speed, velocity of each trial, and basic biomechanical walking information taken from the GaitRite system were calculated and recorded.

Data Reduction

The data collected from each subject was processed using Qualisys Track Manager Software. This produced position data for each subject and their respective trials in the global coordinate system. Visual 3D used inverse dynamics to calculate joint torque and power at the hip, knee and ankle. The lower extremity was built as a model using a linked rigid-segment system. The first recorded calibration trial was used to create an individualized model for each subject. This model enabled location of the joint centers, location of the segment center of mass, definition of the local coordinate system of each segment, and calculation of a transformation matrix to determine the location of the reflective markers in the global coordinate system. Joint centers were located by calculating fifty percent of the distance between the medial and lateral calibration markers for each joint. The hip joint was located twenty five percent of the distance between the markers identifying the right and left greater trochanters. Each segment's long axis was defined by a line from the proximal joint center to the distal joint center. Anthropometrics were used to locate each segment's center of mass from the proximal joint center.

Joint kinetics were calculated by shifting ground reaction forces and torques, center of pressure, force on the segment due to gravity, segment center of mass accelerations, proximal and distal moment arms, and proximal and distal joint center locations into the local coordinate system of each segment. Ground reaction force in Newtons were normalized to body mass, which was recorded in kilograms. Joint torques in Newton-meters and joint angular impulses in Newton-meters*second were normalized to percent body weight multiplied by height. Joint powers in Watts and joint work in Joules were normalized to body mass.

In order to calculate both the vertical and horizontal joint reaction forces, and the joint torque for each lower extremity joint, a free body diagram was initially drawn. The free body diagram is demonstrated as:

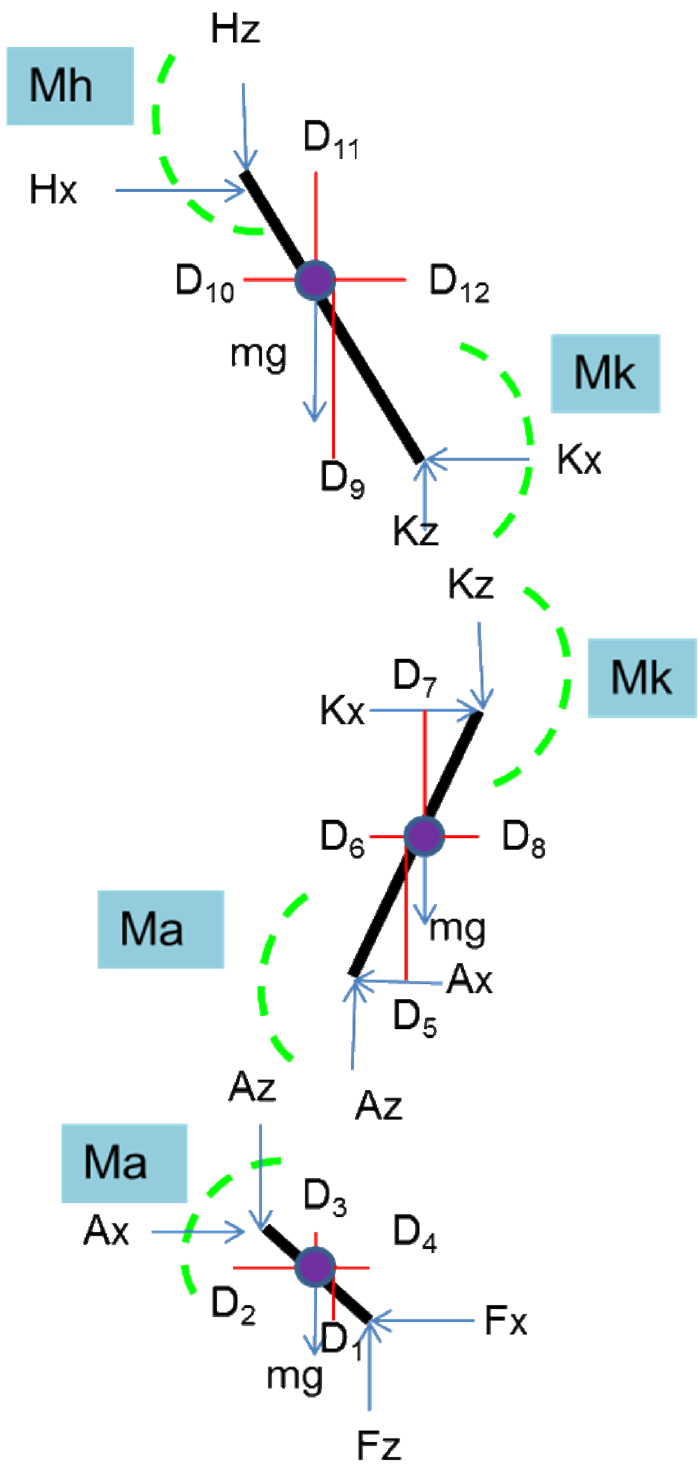


Figure 1: Free body diagrams of each lower extremity segment.

The values were calculated using the foot segment first. Once found, the process was repeated for the leg, thigh and hip. The equation for the vertical ankle joint reaction force (JRF) is as follows:

$$Fz - m_f g + Az = m_f az \tag{1}$$

where Fz is the vertical ground reaction force, m_f is the mass of the foot, g is the acceleration due to gravity, Az is the ankle vertical JRF, and az is the vertical acceleration of the foot. The equation for the horizontal ankle joint reaction force (JRF) is:

$$Fx + Ax = m_f ax \tag{2}$$

where Fx is the horizontal ground reaction force, Ax is the ankle horizontal JRF, m is the mass of the foot, and ax is the horizontal acceleration of the foot. In order to find the ankle joint torque, the generalized equation of angular motion ($\sum T = I\alpha$) was applied to the free body diagram of the foot:

$$Fz(D_1) - Fx_x(D_2) + Az(D_3) - Ax(D_4) + Ma = I\alpha \tag{3}$$

where D1 – D4 are the moment arms for the applied forces onto the foot, Ma is the ankle joint torque, I is the moment of inertia of the foot, and α is the angular acceleration of the foot.

After the calculations for the foot segment were completed, the knee segment was then calculated. First, the vertical knee joint reaction force (JRF) was calculated:

$$-Az - m_l g + Kz = m_l az \tag{4}$$

where -Az is the ankle JRF reversed onto the leg, m_l is the mass of the leg, g is the acceleration due to gravity, Kz is the knee vertical JRF, and az is vertical acceleration of the leg. The horizontal knee joint reaction force (JRF) then followed as:

$$-Ax + Kx = m_l ax \tag{5}$$

where -Az is the ankle horizontal JRF reversed onto the leg, Kx is the knee horizontal JRF, m_l is the mass of the leg, and ax is the horizontal acceleration of the leg. Again, the generalized equation of angular motion was applied to the free body diagram of the leg to calculated the knee torque

$$-Az(D_5) - Ax(D_6) - Kz(D_8) - Kx(D_7) + Mk = I\alpha \tag{6}$$

where D5 – D8 are the moment arms for the applied forces onto the leg, Mk is the knee joint torque, I is the moment of inertia of the leg, and α is the angular acceleration of the leg.

Lastly, the thigh segment was processed to calculate the hip forces and torque. The vertical hip joint reaction force (JRF) was calculated as:

$$-Kz - m_t g + Hz = m_t az \tag{7}$$

where -Kz is the knee JRF reversed onto the thigh, m_t is the mass of the thigh, g acceleration due to gravity, Hz is the hip vertical JRF, and az is the vertical acceleration of the thigh. The hip horizontal JRF was calculated as:

$$-Kx + Hx = m_t ax \tag{8}$$

where -Kx is the knee JRF reversed onto the thigh, Hx is the hip horizontal JRF, m_t is the mass of the thigh, and ah is the horizontal acceleration of the thigh. Lastly, the hip joint torque was calculated with the generalized angular equation of motion applied to the thigh FBD:

$$Kz(D_{12}) - Kx(D_9) + Hz(D_{10}) - Hx(D_{11}) + Mh = I\alpha$$

D9 – D12 are the moment arms for the applied forces onto the thigh, Mh is the hip joint torque, I is the moment of inertia of the thigh, and α is the angular acceleration of the thigh.

Proprietary Laboratory software was used to identify the peak hip extensor torque and positive power in early stance, the peak knee extensor torque and negative power also in early stance and the peak ankle plantarflexor torque and positive power in late stance.

Data Analysis

Pearson Product Moment Correlation analyses and linear regressions were used to identify relationships among both subject group and individual subject peak joint torques and peak joint powers with walking velocity. Significance was tested at the 0.05 level.

CHAPTER 4: RESULTS

It was hypothesized that old adults have a stronger relationship between hip torque and power, and walking velocity, and a weaker relationship between ankle torque and power, and walking velocity than young adults. The purpose of this study was to identify the relationships among lower extremity joint torques and powers and walking velocity in young and old adults. This study also compared these relationships between the age groups. This chapter is separated into the following sections: 1) Preferred, Minimum, and Maximum Velocities, and Range of Speeds, 2) Joint Torques and Powers Investigated, 3) Group and Individual Joint Torques and Powers Observed, 4) Summary.

Preferred, Minimum, and Maximum Velocities, and Range of Speeds

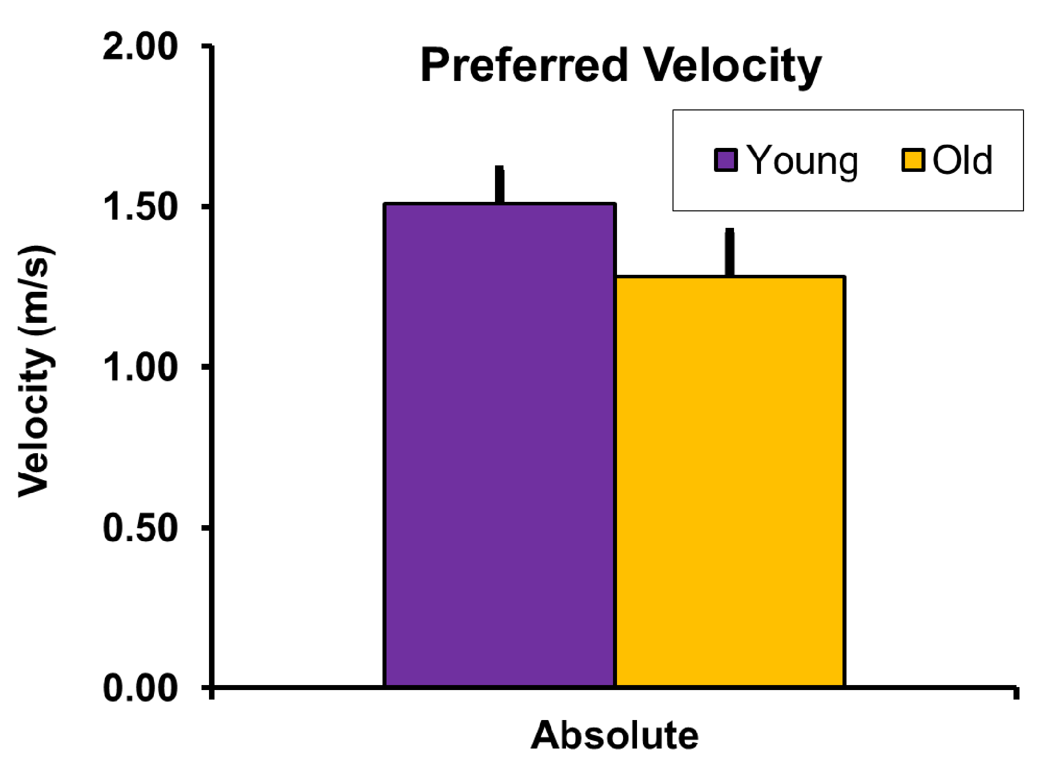


Figure 2: Preferred velocity of young and old subject groups.

Old adults had a slower preferred walking velocity than young adults, though these values were not tested for significance. The mean preferred walking velocity for old adults was 1.28 (±.014) m/s (Figure 2). The young adults had a mean preferred walking velocity of 1.51 (±0.11) m/s.

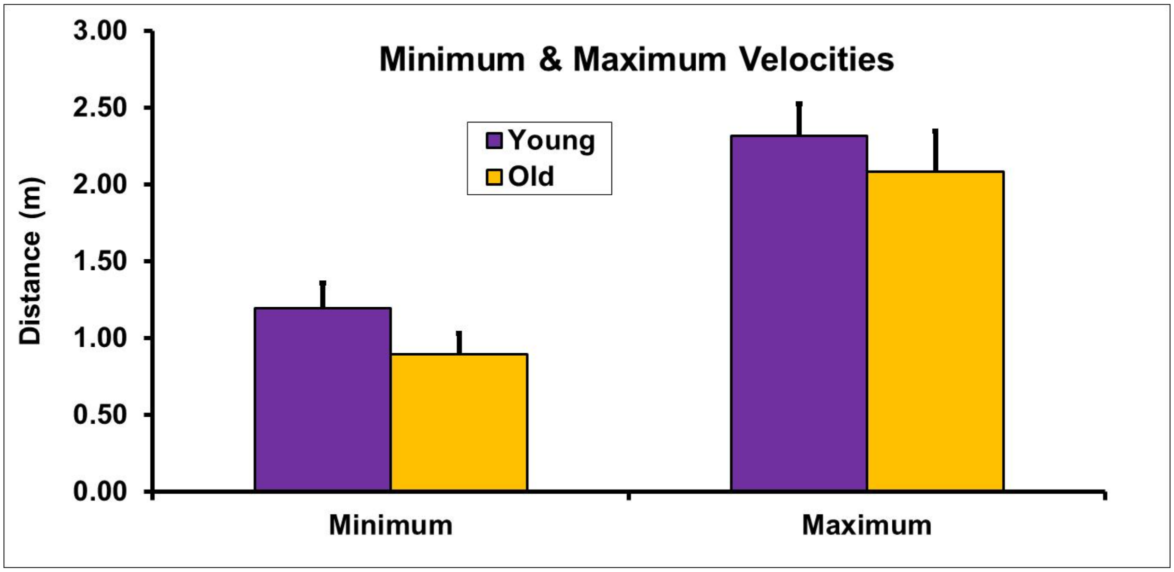


Figure 3: Mean data for minimum and maximum velocities in young and old adults subject groups.

Old adults had a 25% slower minimum walking speed compared to young adults ($p<0.001$). The mean minimum velocity for old adults was 0.89 (±0.14) m/s whereas the minimum velocity for young adults was 1.19 (±0.16) m/s. Old adults had a 10% slower maximum walking velocity compared to young adults ($p<0.001$). The mean maximum velocity for old adults was 2.08 (±0.26) whereas the mean maximum velocity for young adults was 2.32 (±0.21) m/s.

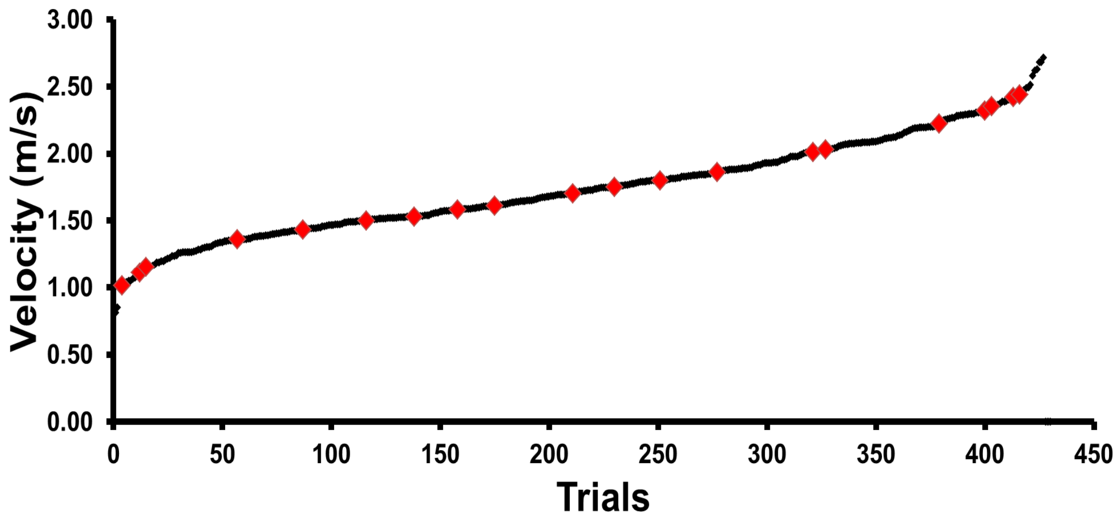


Figure 4: Velocity values from all trials for young subjects.

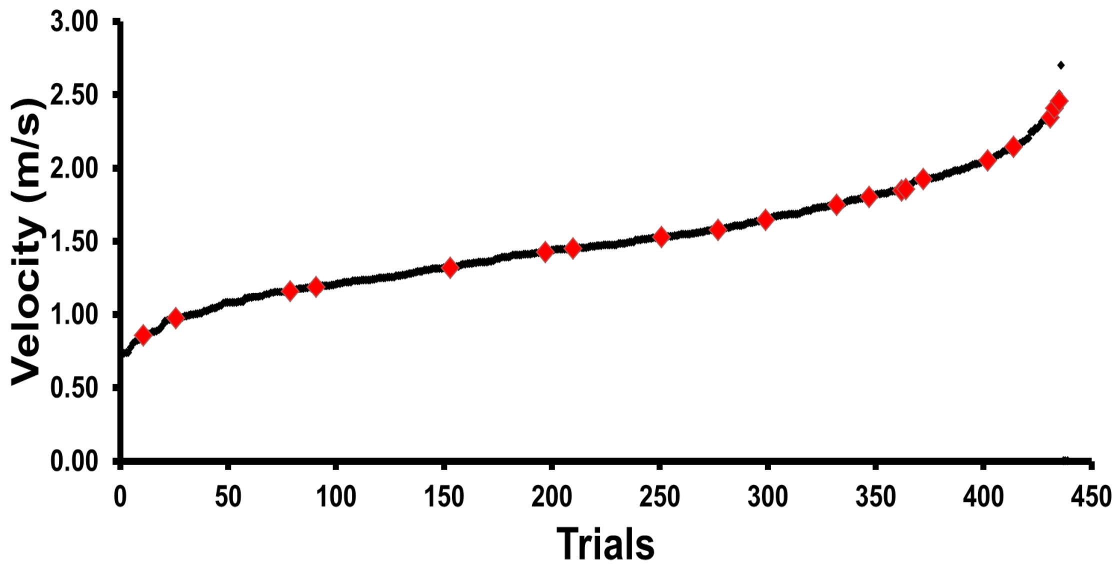


Figure 5: Velocity values from all trials for old subjects.

There were 436 individual trials for the young adults and 427 individual trials for the old adults (Figures 4 and 5). Each groups' trials are represented by the continuous black line and the 20 trials of a representative individual within each group are represented by the red dots along that line. The total series within each age group demonstrates that the velocity values were evenly distributed from lowest to highest values. The individual data demonstrate that each subject's trials were generally well distributed throughout the range of total values and not clustered within a smaller range.

The minimum individual trial velocities for both age groups were similar with values of 0.62 and 0.81 m/s for old and young subjects, respectively. The maximum individual trial velocities for both groups were nearly identical and were 2.70 and 2.71 m/s for old and young subjects, respectively. The total range of velocities was also similar between groups and was 2.08 and 1.90 m/s for old and young subjects, respectively.

Joint Torques and Powers Investigated

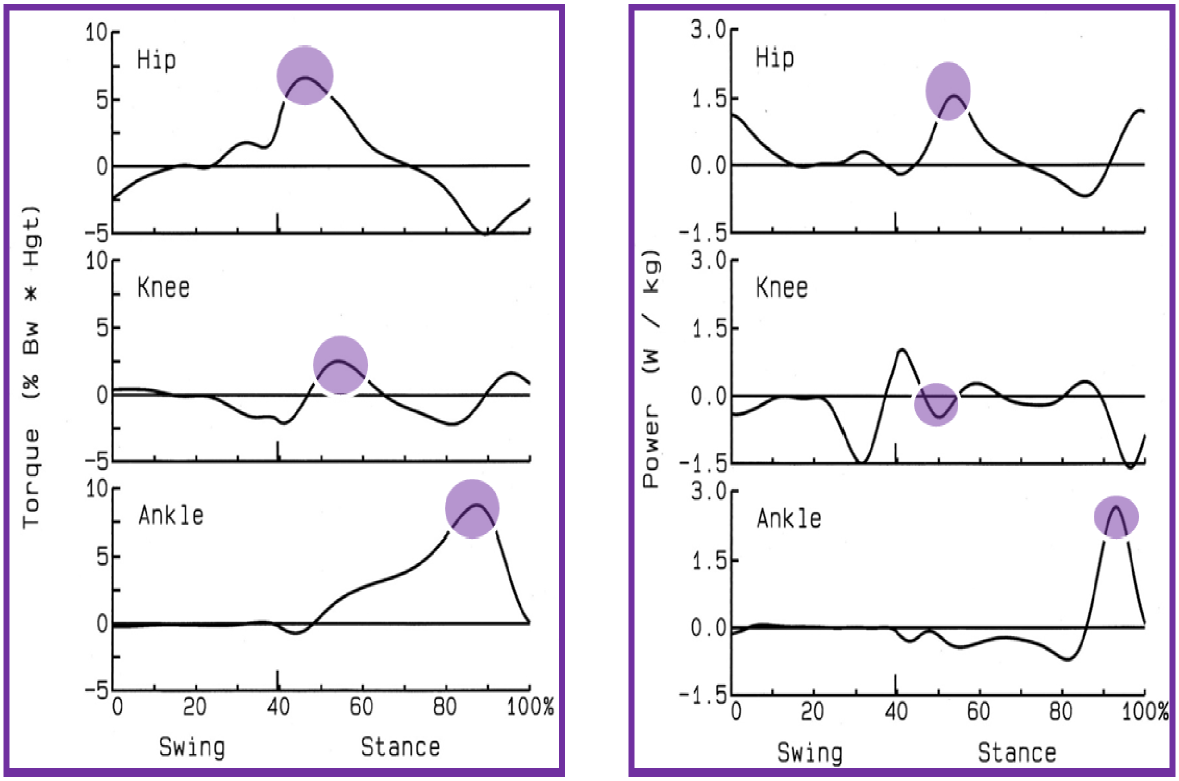


Figure 6: Joint torques and joint powers investigated at the hip, knee and ankle.

All three lower extremity joints were examined for their torques and powers (Figure 6). Torques describe which muscle group at each joint are contributing to the movement. A positive torque corresponds to a positive extensor torque, while a negative torque value refers to a negative flexor torque. Powers, which are the product of joint torques and joint angular velocities, describe the type of movement created. A positive power value corresponds to power generation, which is associated with a concentric muscle action. A negative power value corresponds to power dissipation, which is associated with an eccentric muscle action. At the hip, the peak positive extensor torque and the peak positive power generation was examined at 50% of the gait cycle, i.e. in the early stance phase. At the knee, the peak positive knee extensor torque and the negative peak power dissipation was examined at about 50% of the gait cycle, i.e. again in early stance. As for the ankle, peak positive plantarflexor torque and peak positive power generation were studied at about 90-95% of the gait cycle, i.e. in late stance.

Group and Individual Joint Torques and Powers Observed

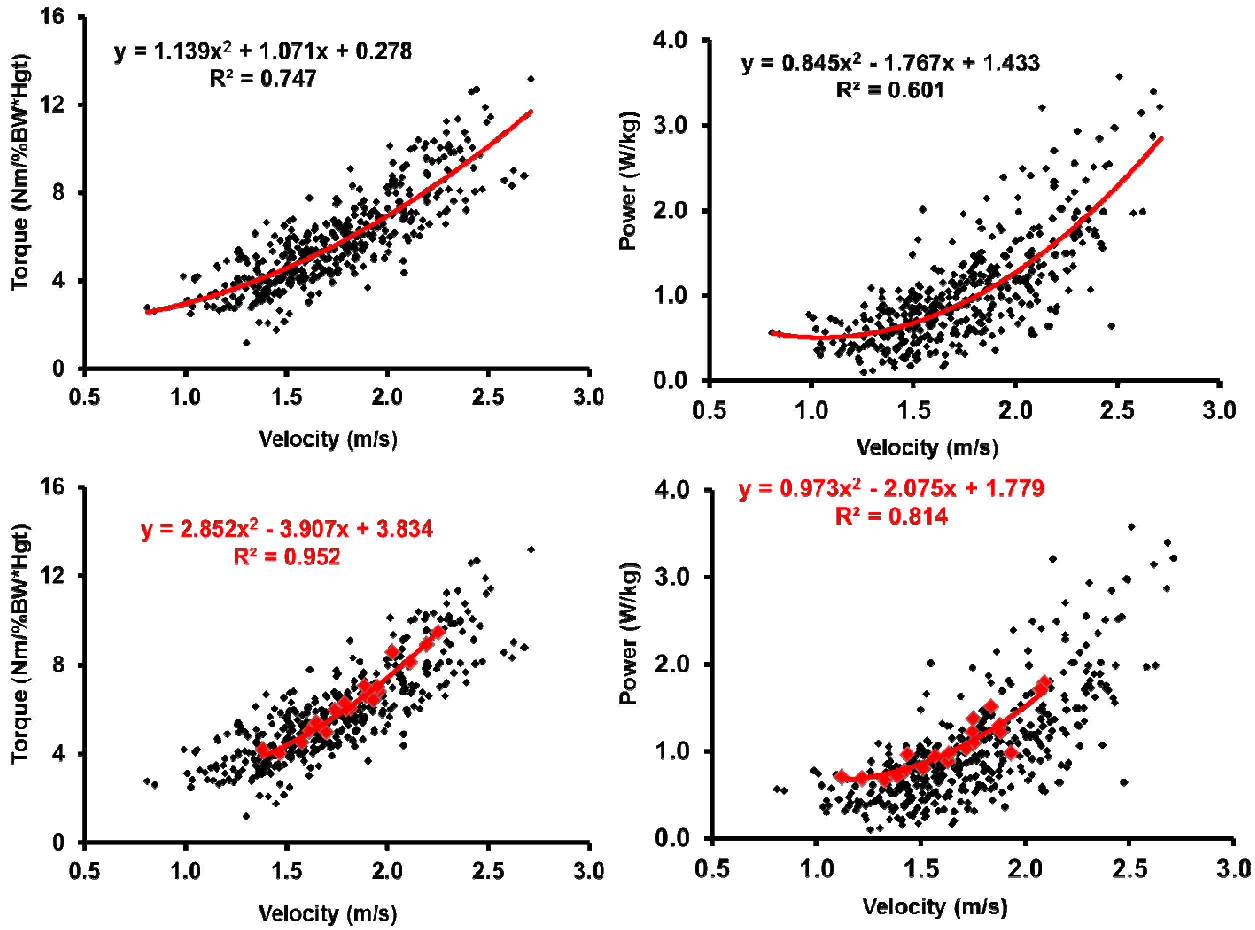


Figure 7: Young hip torques and powers for the groups and the most representative individual subject. The group values are displayed in the top graphs and the most representative individual values are displayed in the bottom graphs.

Curvilinear relationships were observed for the maximum hip torque and power in young adults (Figure 7). These relationships had increased slopes as velocity increased; i.e. the increase in hip torque and power per unit increase in velocity was larger at the faster compared to slower velocities. The squared regression coefficients were high for both maximum torque (R^2=0.747) and power (R^2=0.601) and indicated strong relationships between these variables and walking velocity. The individual subjects had the same curvilinear-upward relationship as the group results for both hip torque and power. The strengths of the relationships were stronger, however, for the individual subjects compared to the entire group of young adults. This is shown by the red line in the graph which displays the most representative individual's trials. The red points on the line indicate the individual's specific trials. Squared regression coefficients were 0.952 and 0.814 for maximum hip torque and power, respectively. All 22 individual subjects had stronger relationships between maximum hip torques and velocity than the group results, and 20 of the 22 individual subjects had stronger relationships between maximum power and velocity than the group results.

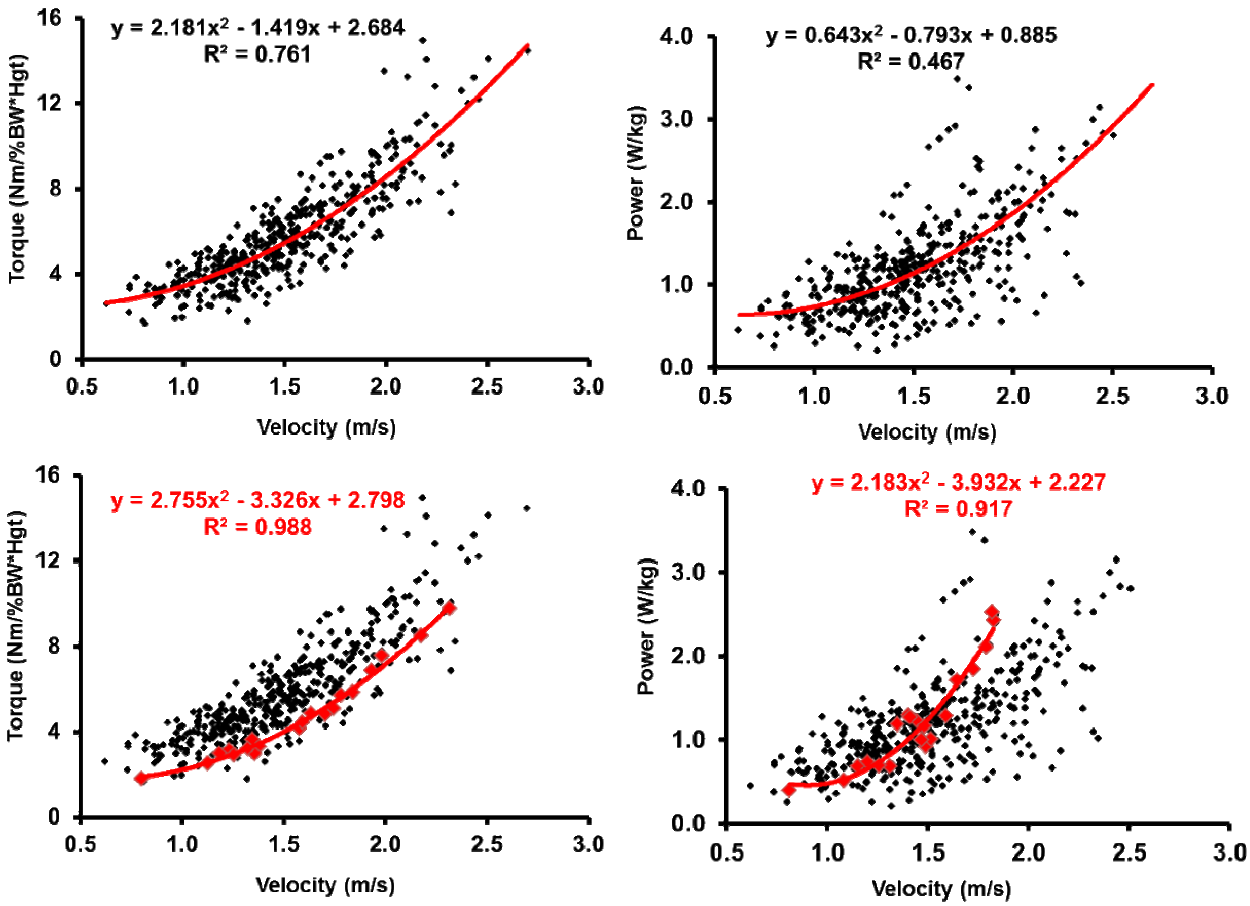


Figure 8: Old hip torques and powers for the groups and the most representative individual subject.

Curvilinear relationships were observed for the maximum hip torque and power in old adults as well (Figure 8). These relationships had increased slopes as velocity increased. The squared regression coefficients were high for both maximum torque (R^2=0.761) and power (R^2=0.467) and indicated strong relationships between these variables and walking velocity. The individual subjects had the same curvilinear-upward relationship as the group results for both hip torque and power, however the strengths of the relationships were stronger for the individual subjects compared to the entire group of old adults. Squared regression coefficients were 0.988 and 0.917 for maximum hip torque and power, respectively. All 22 individual subjects had stronger relationships between maximum hip torques and velocity than the group results, and all 22 individual subjects had stronger relationships between maximum power and velocity than the group results.

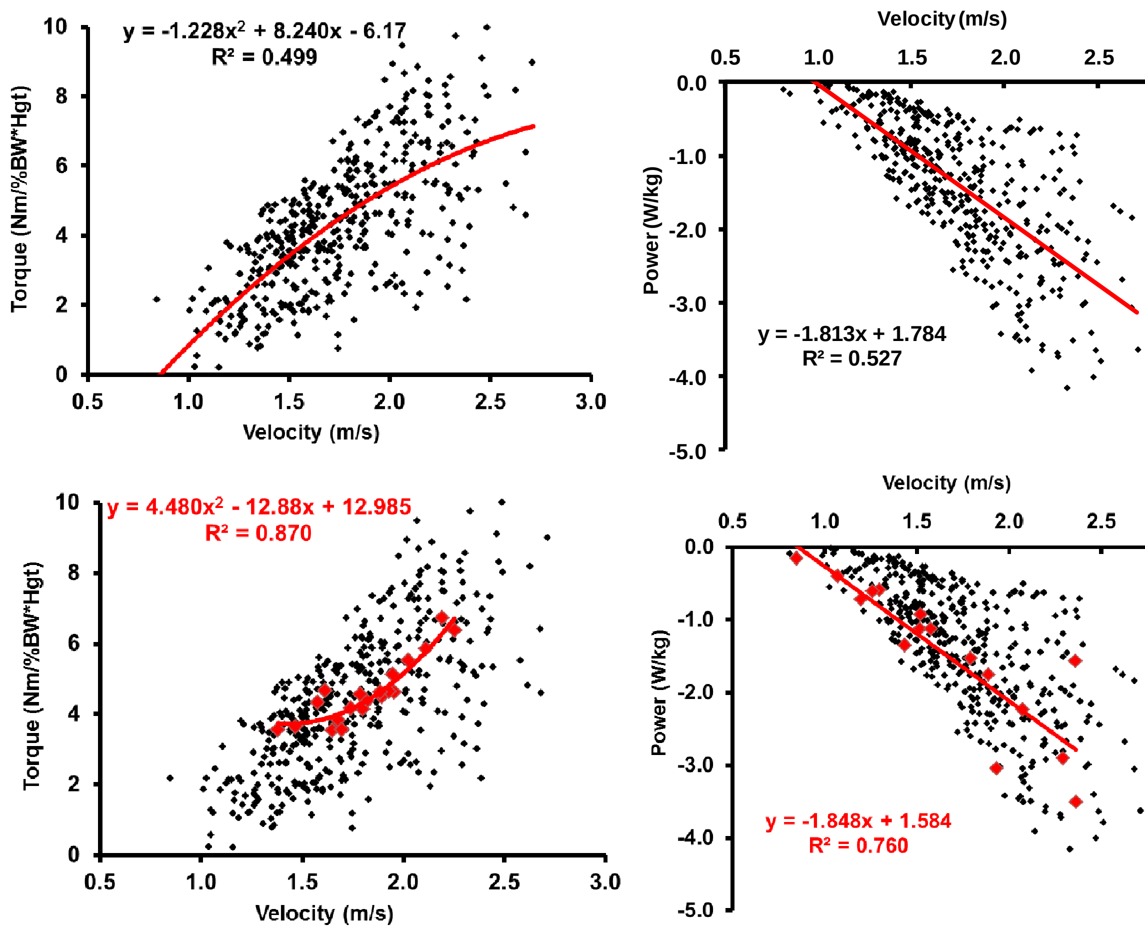


Figure 9: Young knee torques and powers for the groups and the most representative individual subject.

Curvilinear relationships were observed for the maximum knee torque and positive linear relationships were observed for the maximum knee power in young adults (Figure 9). These relationships had increased slopes as velocity increased. The squared regression coefficients for both maximum torque (R^2=0.499) and power (R^2=0.527) which indicated moderate relationships between these variables and walking velocity. The individual subjects had the same curvilinear relationship as the group results for hip torque and the same positive linear relationship as the group results for power. The strengths of the relationships were stronger, however, for the individual subjects as compared to the entire group of young adults. Squared regression coefficients were 0.870 and 0.760 for maximum knee torque and power, respectively. Twenty-one of the 22 individual subjects had stronger relationships between maximum knee torques and velocity than the group results, and all 22 individual subjects had stronger relationships between maximum power and velocity than the group results.

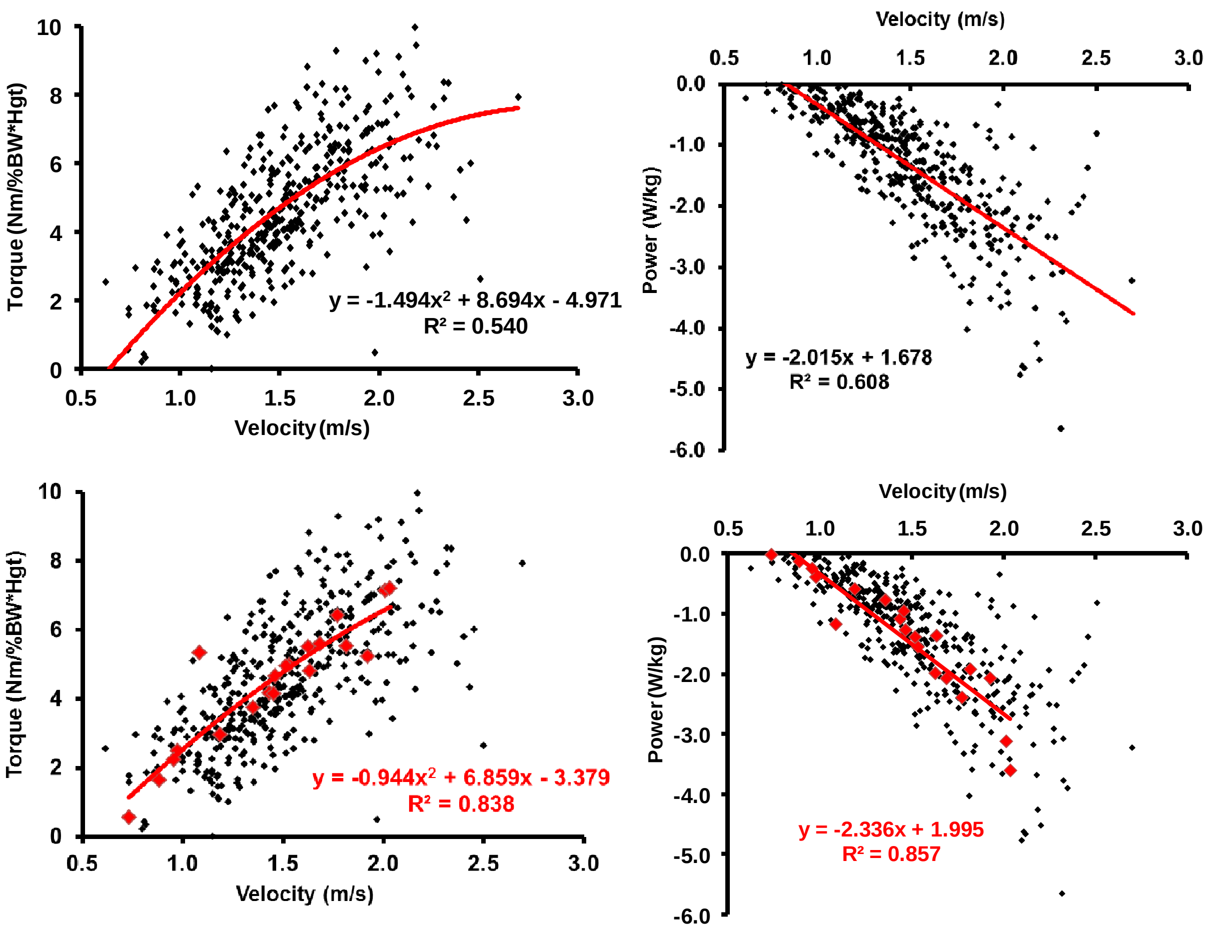


Figure 10: Old knee torques and powers for the groups and the most representative individual subject.

Curvilinear relationships were observed for the maximum knee torque and positive linear relationships were observed for the maximum knee power in old adults (Figure 10). The increase in knee torque and power per unit increase in velocity was larger at the faster compared to slower velocities. The squared regression coefficients for both maximum torque (R^2=0.540) and power (R^2=0.608) and indicated moderate relationships between these variables and walking velocity. The individual subjects had the same curvilinear relationship as the group results for knee torque and the same positive linear relationship as the group results for power. The strengths of the relationships were stronger, however, for the individual subjects compared to that of the entire group of old adults. Squared regression coefficients were 0.838 and 0.857 for maximum knee torque and power, respectively. Nineteen of the 22 individual subjects had stronger relationships between maximum knee torques and velocity than the group results, and 21 of the 22 individual subjects had stronger relationships between maximum power and velocity than the group results.

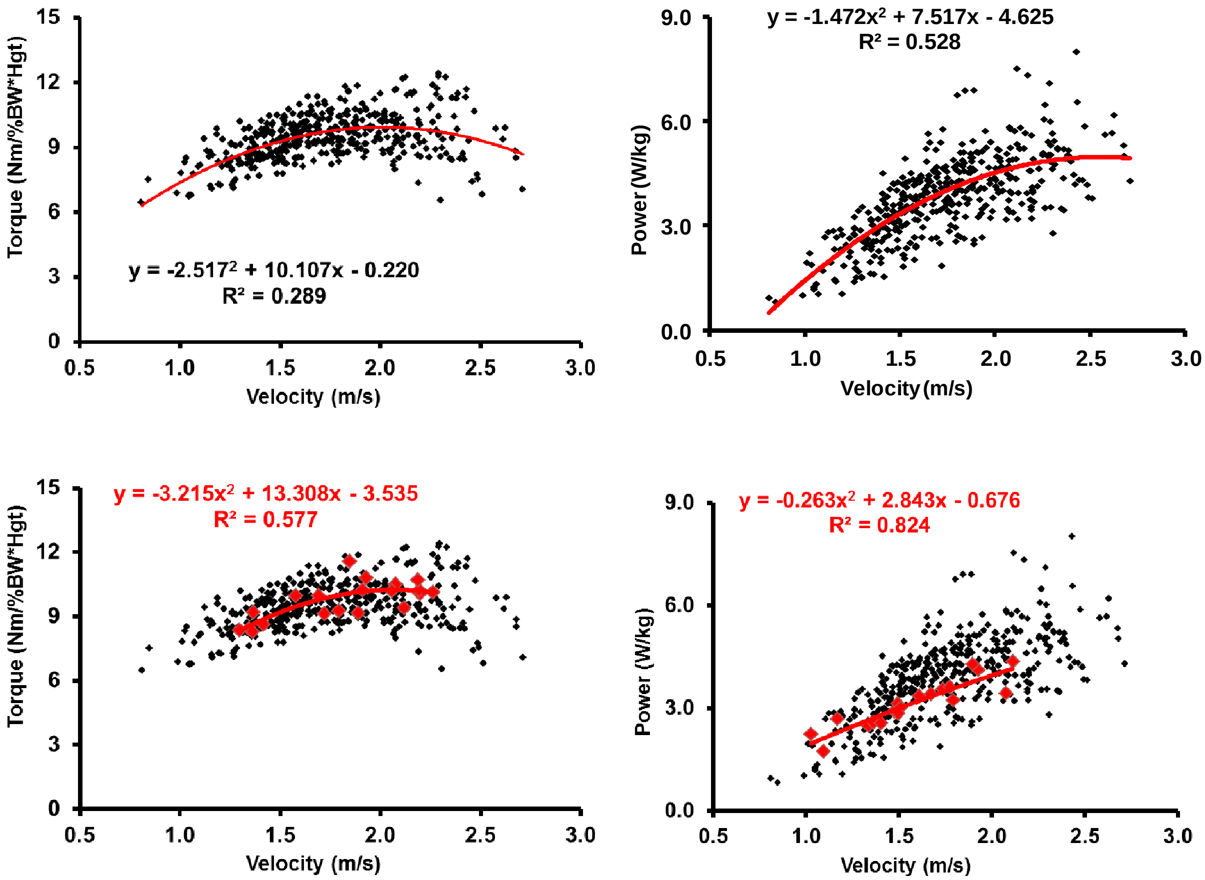


Figure 11: Young ankle torques and powers for the groups and the most representative individual subject.

Curvilinear relationships were observed for the maximum ankle torque and power in young adults (Figure 11). These relationships had increased slopes as velocity increased. The squared regression coefficients for both maximum torque (R^2=0.289) and power (R^2=0.528) indicated moderate relationships between these variables and walking velocity. The individual subjects had the same curvilinear-downward relationship as the group results for both ankle torque and power. The strengths of the relationships were stronger, however, for the individual subjects compared to the entire group of young adults, with the relationship for power being much stronger than that of torque. Squared regression coefficients were 0.577 and 0.824 for maximum ankle torque and power, respectively. Sixteen of the 22 individual subjects had stronger relationships between maximum ankle torques and velocity than the group results, and 21 of the 22 individual subjects had stronger relationships between maximum power and velocity than the group results.

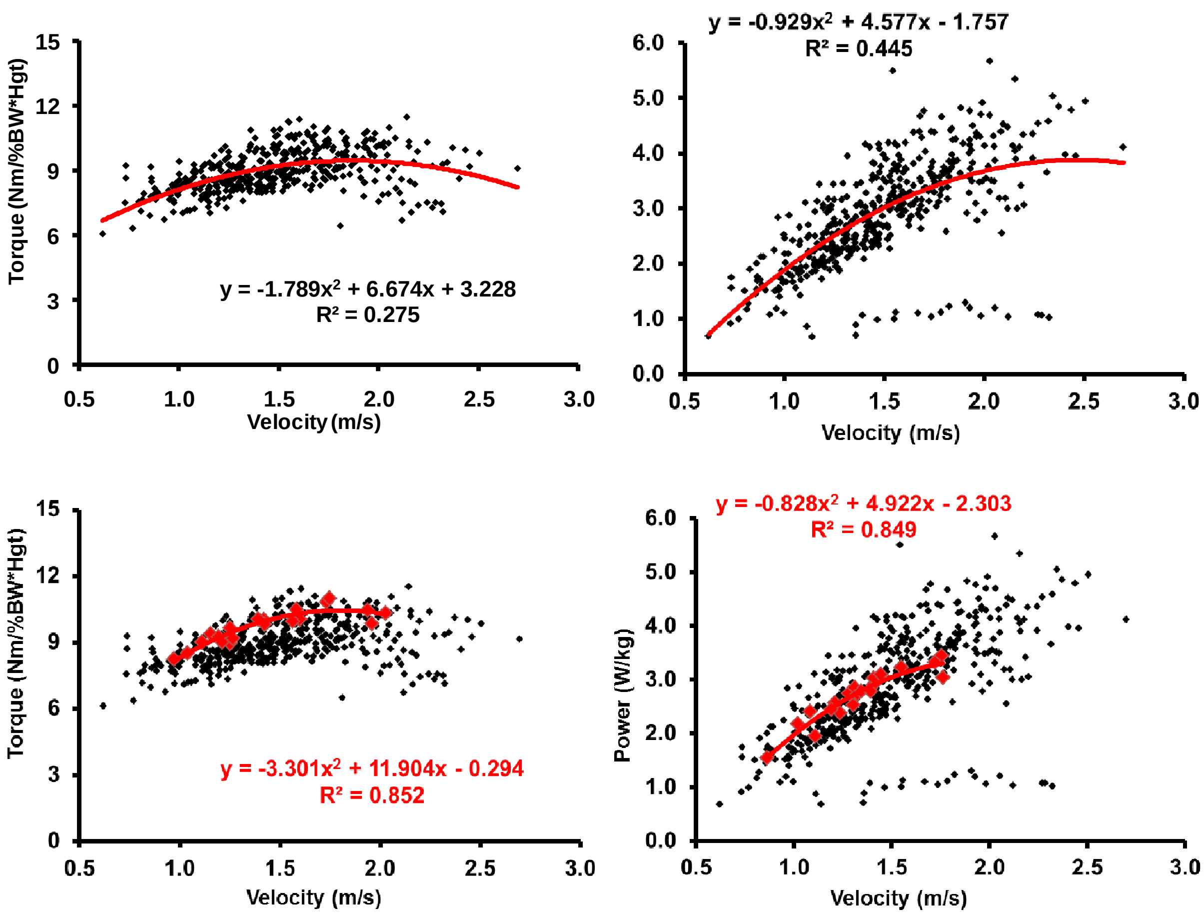


Figure 12: Old ankle torques and powers for the groups and the most representative individual subject.

Curvilinear relationships were observed for the maximum ankle torque and power in old adults (Figure 12). The increase in ankle torque and power per unit increase in velocity was larger at the faster compared to slower velocities. The squared regression coefficients for both maximum torque (R^2=0.275) and power (R^2=0.445) were moderate. The individual subjects had the same curvilinear-downward relationship as the group results for both ankle torque and power. The strengths of the relationships were much stronger for the individual subjects compared to the entire group of old adults. Squared regression coefficients were 0.852 and 0.849 for maximum ankle torque and power, respectively. Twenty-one out of the 22 individual subjects had stronger relationships between maximum ankle torques and velocity than the group results, and 19 of the 22 individual subjects had stronger relationships between maximum power and velocity than the group results.

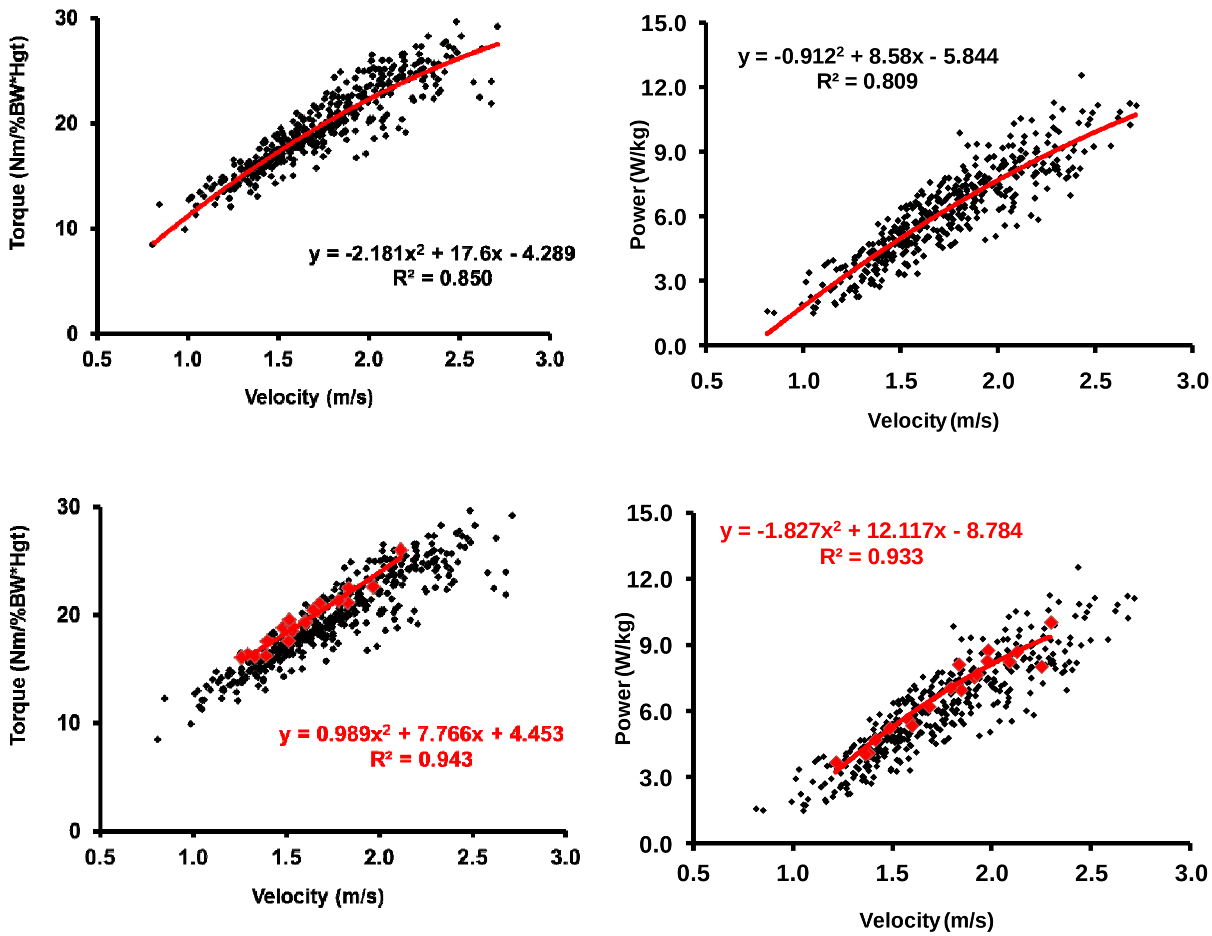


Figure 13: Sum of the young hip, knee, and ankle torques and powers for the groups and the most representative subject.

Curvilinear relationships were observed for the sum of the torque and power values in young adults (Figure 13). These relationships had increased slopes as velocity increased. The squared regression coefficients were high for both sum of the torque values (R^2=0.850) and the sum of the power values (R^2=0.809) and indicated strong relationships between these variables and walking velocity. The individual subjects had the same curvilinear-downward relationship as the group results for both the sum of the torque and power values. The strengths of the relationships were stronger for the individual subjects compared to the entire group of young adults. Squared regression coefficients were 0.943 and 0.933 for the sum of the values of torque and power, respectively. All 22 individual subjects had stronger relationships between the sum of the values for torque and velocity than the group results, and all 22 individual subjects had stronger relationships between the sum of the values of power and velocity than the group results.

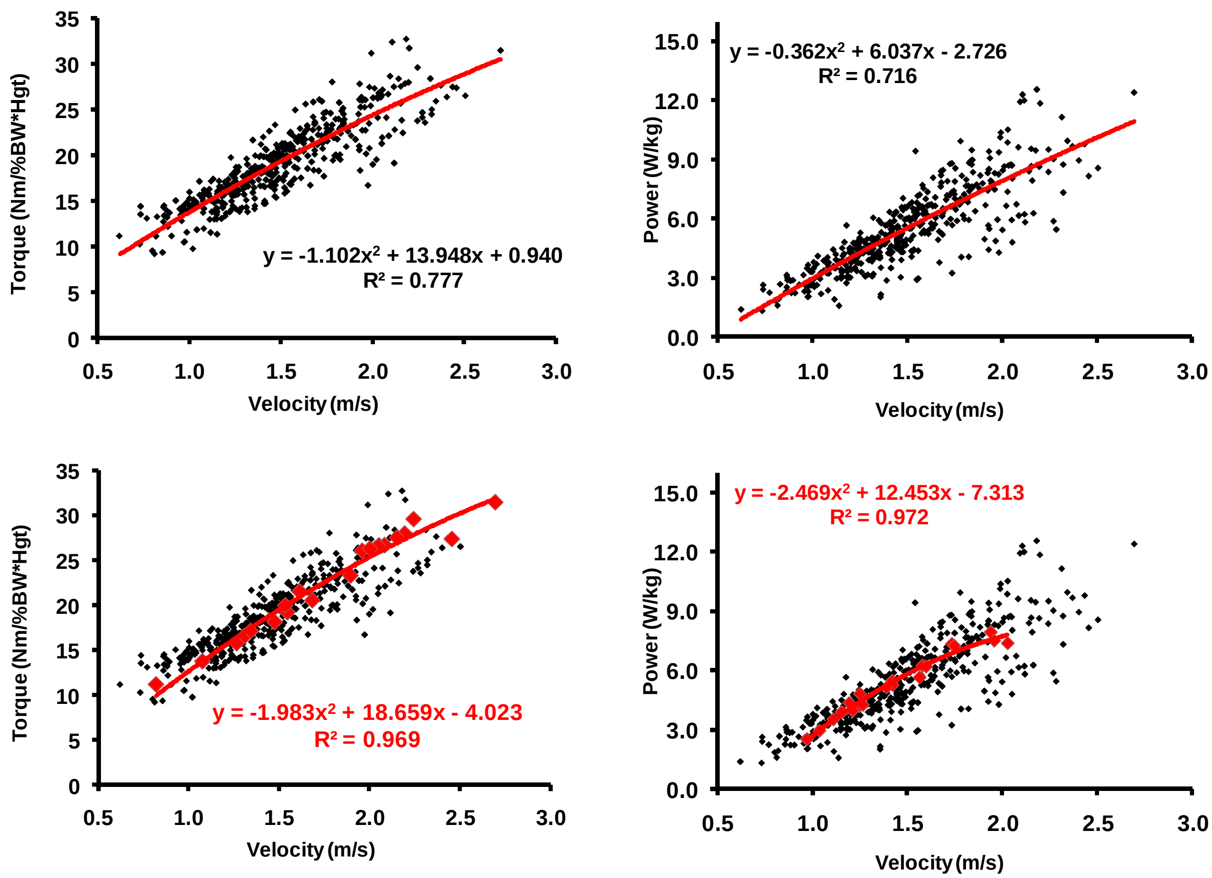


Figure 14: Sum of the old hip, knee, and ankle torques and powers for the groups and the most representative subject.

Curvilinear relationships were observed for the sum of the old hip, knee, and ankle torque and power in old adults (Figure 14). These relationships had increased slopes as velocity increased. The squared regression coefficients were high for both sum of the torque values (R^2=0.777) and sum of the power values (R^2=0.716) and indicated strong relationships between these variables and walking velocity. The individual subjects had the same curvilinear-downward relationship as the group results for both sum of the values for torque and power. The strengths of the relationships were stronger for the individual subjects compared to the entire group of old adults.

Squared regression coefficients were 0.969 and 0.972 for sum of the values for torque and power, respectively. All 22 individual subjects had stronger relationships between the sum of the torques and velocity than the group results, and all 22 individual subjects had stronger relationships between the sum of power and velocity than the group results.

	Torque		Power	
Hip - Pop	$y = 1.139x^2 + 1.071x + 0.278$	$R^2 = 0.747$	$y = 0.845x2 - 1.767x + 1.433$	$R^2 = 0.601$
Ind	$y = 2.852x^2 - 3.907x + 3.834$	$R^2 = 0.952$	$y = 0.973x^2 - 2.075x + 1.778$	$R^2 = 0.814$
Knee - Pop	$y = -1.228x^2 + 8.241x - 6.170$	$R^2 = 0.499$	$y = -1.813x + 1.784$	$R^2 = 0.527$
Ind	$y = 4.480x^2 - 12.88x + 12.985$	$R^2 = 0.870$	$y = -1.848x + 1.584$	$R^2 = 0.760$
Ankle - Pop	$y = -2.517^2 + 10.107x - 0.220$	$R^2 = 0.289$	$y = -1.472x^2 + 7.517x - 4.625$	$R^2 = 0.528$
Ind	$y = -3.215x^2 + 13.308x - 3.535$	$R^2 = 0.577$	$y = -0.263x^2 + 2.843x - 0.676$	$R^2 = 0.824$
Sum - Pop	$y = -2.181x^2 + 17.6x - 4.289$	$R^2 = 0.850$	$y = -0.9122 + 8.58x - 5.844$	$R^2 = 0.809$
Ind	$y = 0.989x^2 + 7.766x + 4.453$	$R^2 = 0.943$	$y = -1.827x^2 + 12.117x - 8.7839$	$R^2 = 0.933$

Table 2: Torque and power equations and coefficient of determination values for the hip, knee, ankle, and sum of all three lower extremity joints for the young subjects.

	Torque		Power	
Hip - Pop	$y = 2.181x^2 - 1.419x + 2.684$	$R^2 = 0.761$	$y = 0.643x^2 - 0.793x + 0.885$	$R^2 = 0.467$
Ind	$y = 2.755x^2 - 3.326x + 2.798$	$R^2 = 0.988$	$y = 2.183x^2 - 3.932x + 2.227$	$R^2 = 0.917$
Knee - Pop	$y = -1.494x^2 + 8.694x - 4.971$	$R^2 = 0.540$	$y = -2.015x + 1.678$	$R^2 = 0.609$
Ind	$y = -0.944x^2 + 6.859x -3.379$	$R^2 = 0.838$	$y = -2.336x + 1.995$	$R^2 = 0.857$
Ankle - Pop	$y = -1.789x^2 + 6.674x + 3.228$	$R^2 = 0.275$	$y = -0.929x^2 + 4.577x - 1.757$	$R^2 = 0.445$
Ind	$y = -3.301x^2 + 11.904x - 0.294$	$R^2 = 0.852$	$y = -0.828x^2 + 4.922x - 2.303$	$R^2 = 0.849$
Sum - Pop	$y = -1.102x^2 + 13.948x + 0.940$	$R^2 = 0.777$	$y = -0.362x^2 + 6.037x - 2.726$	$R^2 = 0.716$
Ind	$y = -1.983x^2 + 18.659x - 4.023$	$R^2 = 0.969$	$y = -2.469x^2 + 12.453x - 7.313$	$R^2 = 0.972$

Table 3: Torque and power equations and coefficient of determination values for the hip, knee, ankle, and sum of all three lower extremity joints for the old subjects.

The above tables (Table2, Table 3) contain regression equations for group and individual subjects. The population includes all participants and trials, while the individual signifies the most representative individual subject. The coefficient of determination value of this individual was closest to the coefficient of determination values across all the individual participants. All of these equations symbolize a curvilinear line except for the knee power for both the population and the individual in each of the age groups.

Summary

Old adults walked at a slower preferred walking velocity than young adults. There was no noticeable difference between the maximum walking velocities; however there was a difference seen in the minimum walking velocities between the two age groups. The range of speeds were similar for both groups, however it was evident that the slowest speed for the young adults was faster than that of the old adults. It was evident that the individuals in each group followed a similar fashion when the data was plotted, as each individual followed the trend line of the entire group.

The positive peak extensor hip torques and peak positive power generations were observed along with the peak knee positive extensor torque and peak negative power dissipation. The ankle peak positive plantarflexor torque was also examined in addition to the peak positive power generation at the ankle. These observations gave the data indicated in the figures.

Young adult hip, knee, and ankle, torque and power group coefficient of determination values were lower than those seen by the most representative individuals. The old adult hip, knee, and ankle coefficient of determination values were also lower for the group than those of the most representative individuals. As for the sum of the lower extremity joint torque and power coefficient of determination values for the young adult group, the values were still lesser than that of the most representative individual, but the differences between the values are not as great as those seen when observing the joints individually. The same concept is applicable to that of the old adults; however there is a slightly greater difference between the coefficient of determination values for the old adults than for those of the young adults.

CHAPTER 5: DISCUSSION

The purpose of this study was to identify the relationships among lower extremity joint torques and powers and walking velocity in young and old adults. This study also compared these relationships between the age groups. This chapter is divided into the following sections: 1) Comparison of Gait Biomechanics with Previous Literature, 2) Comparison between Old and Young Torque and Power to Previous Literature, 3) Control of Walking Velocity in the Populations of Young and Old Adults, 4) How Humans Control Walking Velocity, 5) Group vs. Individual Analyses, and 6) Conclusion.

Comparison of Gait Biomechanics with Previous Literature

Results found in this study agree with those found in previous literature. The preferred walking velocity of old adults was slower than that of young adults. However, it is important to note that these results are not suggesting that old adults do not have the ability to walk as quickly as young adults; but rather suggests that these old adults preferred to walk at a slower speed than the young adults. In fact, the old adults that participated in this study were quite healthy and functional as they were able to obtain approximately 90% of the maximum walking velocity of the young adults. This study showed a mean preferred walking speed of 1.52m/s for young adults and a mean preferred speed of 1.28m/s for old adults. These are similar to the findings of Kerrigan et al. and Riley et al., in which the young adults had a preferred walking velocity of faster than that of the old adults (Kerrigan et al., 1998, Riley et al., 2001, Kim et al., 2005, Khandoker et al., 2010). Comparatively, the difference between young and old mean preferred walking velocity in this study was 0.24m/s, while the results from Kerrigan et al. showed a difference of 0.18m/s. Though these numbers are similar, a possible reason why the Kerrigan et al. study shows a greater difference between the young and old values may be due to the fact that the study was conducted in 1998, twelve years prior to this study.

Preferred walking velocity is not the only factor observed by researchers. The step length of old and young adults is also of particular interest (Table 4). Surprisingly, there is no true difference between the preferred step length of these two age groups (Thelen et al., 2007 and 2009). Additionally, the similarity between the step length values span all speeds from slow, medium, to fast walking velocities (Cofre et al., 2011).

Study	Step Length - Old vs. Young (m)	Preferred Walking Velocity - Old vs. Young (m/s)
Present	0.74 ± 0.87 / 0.78 ± 0.08	1.28 ± 0.14 / 1.58 ± 0.11
Riley et al., 2001	0.61 ± 0.01 / 0.71 ± 0.07	1.20 ± .01 / 1.40 ± 0.20
Kim et al., 2005	0.61 ± 0.06 / 0.64 ± 0.06	1.20 ± 0.10 / 1.33 ± 0.12
Thelen et al., 2007	0.82 ± 0.06 / 0.83 ± 0.07	1.32 ± 0.13 / 1.33 ± 0.13
Cofre et al., 2011	0.70 ± 0.05 / 0.71 ± 0.04	1.31 ± 0.03 / 1.30 ± 0.05

Table 4: Old adult data comparison between present and past studies. Joint torques and powers in old adults seen in this study were also similar to those found in past studies. Peak hip torque and power occurred at ~45 and ~55% of the gait cycle or shortly into the stance phase (figure 6, above). Peak knee torque and its associated negative power occurred at ~55 and 50% of the gait cycle, again early in the stance phase. Peak ankle torque and positive ankle power generation occurred ~88 and 92% through the gait cycle or in late stance phase. These curves are very similar to those found in previous studies (Kerrigan et al., 1998, Graf et al., 2005, Cofre et al., 2011).

Previous literature also showed that old adults preferentially increase the mechanical output at their hips relative to young adults but not at their knees and ankles (Kerrigan et al., 1998, DeVita et at., 2000, Thelen et al., 2007). This study shows that this fact is true for both young and old adults alike, meaning; there is a distal to proximal usage of lower extremity joint torques and powers in old adults. This usage demonstrates a distal to proximal mechanical plasticity in adults. We compared peak torques and powers and the areas under the torque and power curves at the time of the peak values between young and old adults (Table 5). Overall, we observed the distal to proximal shift in mechanical function in old adults. Relative to young adults, they had greater hip angular impulse and lower peak ankle torque and power (all $p<0.05$).

Joint Torque						
	Hip		Knee		Ankle	
	Maximum	Impulse	Maximum	Impulse	Maximum	Impulse
Young	5.850 (1.00)	0.723 (0.200)*	4.290 (1.520)	0.547 (0.206)*	9.410 (0.860)*	1.970 (0.240)
Old	5.580 (1.23)	0.897 (0.285)*	4.510 (1.630)	0.688 (0.246)*	8.820 (0.660)*	2.010 (0.300
	Joint Power					
	Hip		Knee		Ankle	
	Maximum	Work	Maximum	Work	Maximum	Work
Young	1.030 (0.370)	0.108 (0.041)*	-1.370 (0.610)	-0.080 (0.040)	3.800 (0.860)*	0.314 (0.092)*
Old	1.180 (0.480)	0.174 (0.067)*	-1.360 (0.470)	-0.080 (0.030)	2.790 (0.630)*	0.224 (0.059)*

Table 5: Comparison of lower extremity joint peak torques and powers between young and old adults. * Indicate significant differences.

Control of Walking Velocity in the Populations of Young and Old Adults

Mechanical plasticity is the element of the hypothesis that states that old adults have a stronger relationship between hip torque and power and walking velocity, and a weaker relationship between ankle torque and power and walking velocity than young adults. In short, the hypothesis states that age creates a redistribution of joint torques and powers in the control of walking velocity. However, the results of this study showed that old and young adults modulate walking velocity in the same way; both age groups increased hip torque when walking velocity was increased. Though this similarity exists, it is important to note that old adults begin with a higher hip torque naturally as compared to young adults. This means that the hip torque of old adults will simply continue to be greater than that of the young adults when walking velocity is manipulated. This observation did not support the hypothesis previously stated. The hypothesis was refuted specifically due to the fact that the relationship between ankle torque and walking velocity was not weaker in old compared to young adults. Therefore, there was no difference in mechanical plasticity between old and young adults in the modulation of walking velocity. There seems to be a consistent pattern between both age groups of joint torques and powers throughout the lower extremity. Therefore, manipulation of walking velocity is controlled similarly in old and young adults.

Although old adults rely more on their hip torque while walking, they do not rely on it to manipulate walking velocity. This observation shows that while there is a modulation of lower extremity joint torques during increasing and decreasing walking velocity. This modulation is independent of age and both the biomechanical and physiological adaptations that are associated with age. This result is shown in Figure 15 which graphs the correlation coefficients for peak torques and walking velocity. Note the nearly identical values in young and old adults.

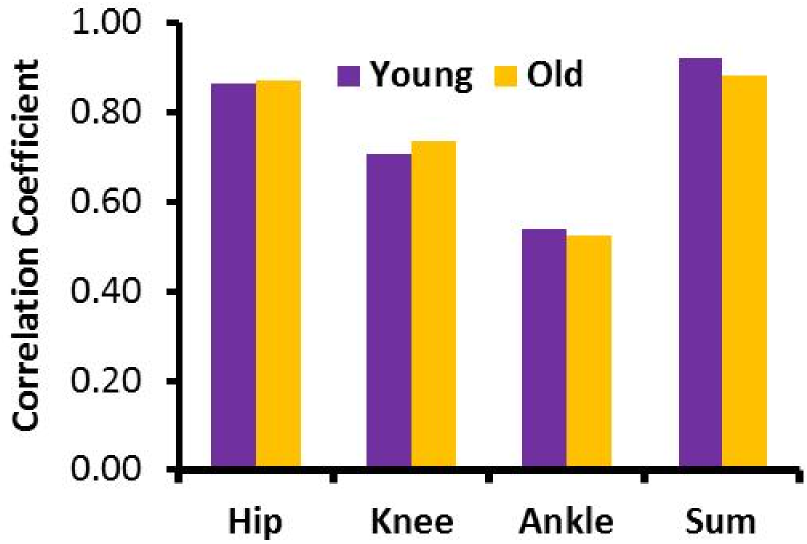


Figure 15: Population based correlation coefficients for peak torques in young and old subjects.

As seen in the figure above (Figure 15), the hip has the greatest correlation coefficient. However, what is not evident in this figure, but is shown in both figures 7 and 8, is that the curvilinear line representing the relationship between hip torque and power and walking velocity in young and old adults is curvilinear upwards. This means that at faster velocities, the relationship between hip torque and power and walking velocity increases at a faster rate. The opposite is true for the curvilinear downward line seen in both figures 11 and 12, which graph the relationship between both torque and power and walking velocity at the ankle in young and old adults. As the velocities increased, the regression lines plateaued. This means that ankle torque and power can only increase so much before an increase is no longer possible.

These findings can be used as the foundation for exercise programs created for old adults. By working the ankle, one can increase the ankle torque and power, which will help increase the range of motion at the joint. This increase in the range of motion may increase the preferred walking velocity of the individual. This type of exercise will help those old adults who walk well below the average preferred walking velocity. After gaining a full range of motion in the ankle joint, one must move proximally to the hip joint in order to further increase walking velocity. By exercising the muscles of the hip joint and concurrently increasing the torque and power output at that joint, one can increase the range of motion and ultimately increase preferred walking velocity to much higher values. This increase in preferred walking velocity may then lead to an improvement in overall health.

How Humans Control Walking Velocity

As discussed earlier, it is thought that age does not affect the control of walking velocity. In both old and young adults, there was evidence of more hip control than any other lower extremity joint. The coefficient of determination for the hip torque of young adults was 0.747, while the old adults had a coefficient of determination very similar to that with a value of 0.761. When compared to the coefficient of determination values of the knee and ankle, the young adults had values of 0.499 and 0.289, respectively. A similar trend is seen in the old adults with knee and ankle coefficient of determination values of 0.540 and 0.275, respectively. In general, adults of all ages have stronger relationships between hip torques and velocity as compared to the more distal lower extremity joints.

Lelas et al. found similar patterns in their study (Figure 16). They focused only on young adults; therefore, the studies are alike in the sense that both investigated modulation in walking velocity in young adults. However, though the methods and materials may have been similar, there were differences in the results. The coefficients of determination for the peak hip flexion moment, peak knee flexion moment, and peak ankle dorsiflexion moment recorded by Lelas are 0.81, 0.73, and 0.48, respectively (Lelas et al., 2003). These values are higher than those observed in this study for the young adults. Collectively, both studies show that the relationship was strongest at the hip and then gradually decreased as one moves distally. However, the values observed by Lelas are greater than those observed in this study (Figure 17). This difference may be due to the fact that the velocities seen in this study were greater than those used by Lelas and as seen in both studies, variances were larger at the faster compared to the slower velocities.

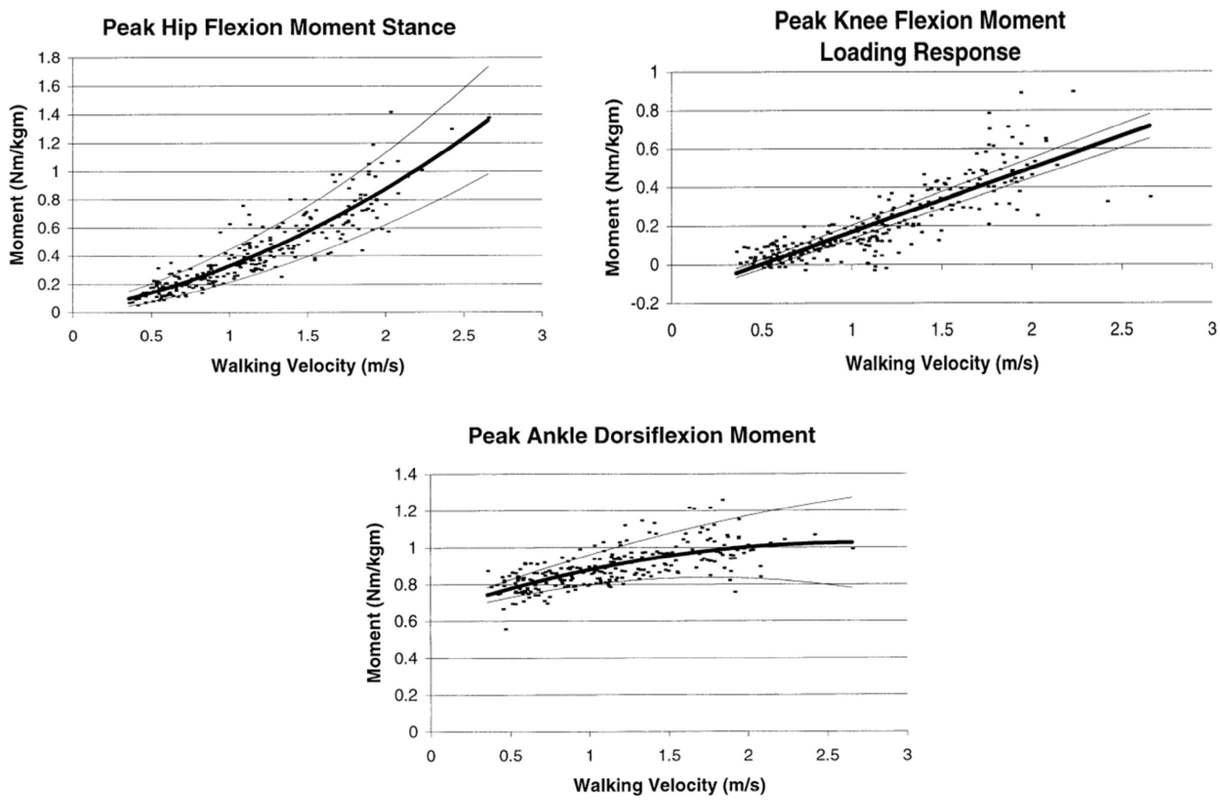


Figure 16: Peak hip flexion, knee flexion and ankle dorsiflexion graphs from Lelas at el., 2003.

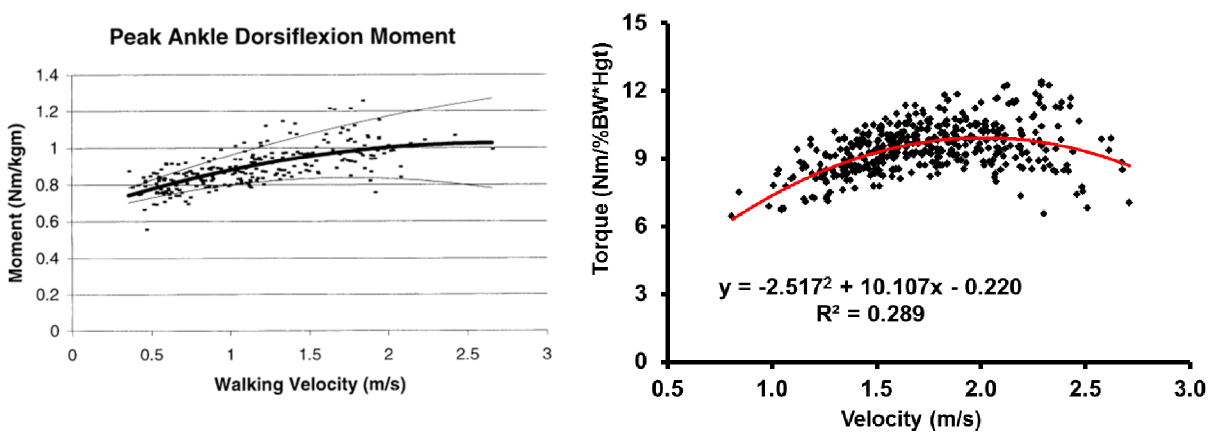


Figure 17: Comparison between peak ankle torque in Lelas et al. and this study.

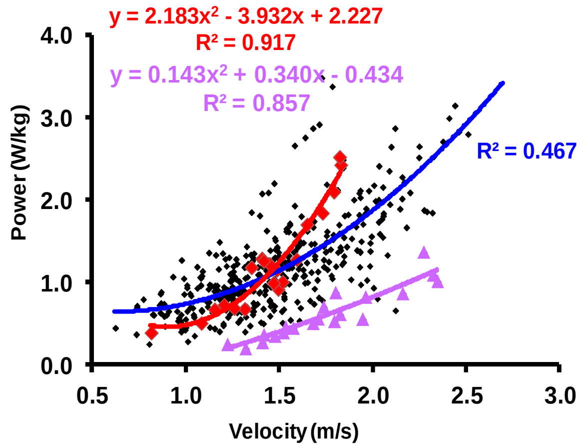
The range of velocity observed in the Lelas experiment starts at approximately 0.4 m/s and ends at approximately 2.2 m/s. However, the present study tested a velocity range starting at about 0.8 m/s and ending around 2.9 m/s (Figure 17). The higher velocities in the present study may be the reason for the tighter fit seen in the Lelas results as compared to those found in this study. As can be seen in both Lelas et al.'s data and the present data, the variability in the relationships increased at higher walking speeds.

Comparatively, power values were similar between these two studies. Joint power relationships were more similar across the lower extremity joints than joint torque relationships. While Lelas had stronger relationships with power, with a noted coefficient of determination values of 0.759 for the hip, 0.700 for the knee, and 0.852 for the ankle, this study had lower values of 0.601, 0.527 and 0.528 for the hip, knee and ankle, respectively. The most interesting point from these values is that they are all similar across all lower extremity joints, though values from Lelas are greater than those found in this study. This shows that while the relationship between torque and walking velocity for the lower extremity joints vary from distal to proximal when modulating walking velocity, the relationship between power and walking velocity across the joints is arguably even. However, the explanation behind the differences in coefficient of determination values may be due to the differences in walking velocity as mentioned earlier.

Group vs. Individual Analyses

An interesting finding of this study was the differences observed between the coefficient of determination values of the group versus the individual participant analyses. The most representative individual is the individual within the group that had a coefficient of determination closest to that of the mean individual participant value. After studying the coefficient of determination values for these individuals and the overall group, it has become clear that following a strict group observation approach fails to truly identify how humans modulate walking velocity. Instead, it is better to study the relationships between the variables at the individual subject level.

The graphs below (Figure 18) display peak hip power generation data for the group and two representative individuals in the old group on the left and the same type of graph is shown for the young adults on the right. It is evident that there is a lower coefficient of determination value for the old and young groups, R^2=0.467 and R^2=0.601, respectively, as compared to that of both the representative in each age group. The individual lines indicate that the group data points are comprised of twenty-three subjects who are highly associated with the two factors of power and velocity. This demonstrates the idea that, in truth, the variables are more tightly coupled than the group data shows. Thus, the magnitude between individuals differs. The individuals that start with a lower power value remain at a lower power value as compared to other individuals within the group.



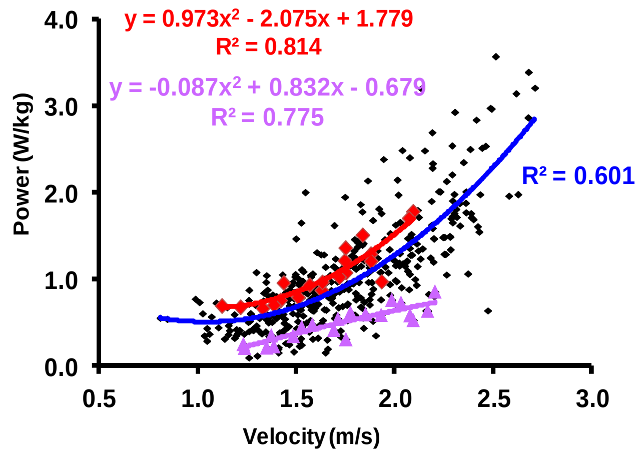


Figure 18: Peak hip power generation for the group and two representative individuals in the old adult group (left). Peak hip power generation for the group and two representative individuals in the young adult group (right). Purple and red trend lines, equations and data points indicate the two representative individuals. Blue correlation of determination values and trend lines represent data of the entire group.

The individual analyses are more insightful and truly show what people are doing in comparison to group analyses. Future research should focus more on the individual outputs as opposed to grouping many individuals together and analyzing the data as a whole population. This study gives a strong reasoning as to why this type of analysis should be utilized in upcoming research.

Conclusion

The previously stated hypothesis that old adults have a stronger relationship between hip torque and power, and walking velocity, and a weaker relationship between ankle torque and power, and walking velocity than young adults was refuted by the results found in this study. Though hip torque and power were greater than ankle torque and power observed in the old adults overall, both old and young individuals modulated joint torques and powers similarly when modulating walking velocity. Also, walking velocity in both young and old adults was more strongly related to hip torque and power than the more distal joints. It is also evident that focusing on individual outputs as opposed to group values gives more precise data and truly tells how people, old and young, manipulate walking velocity.

It seems that velocity is directly related to torque and power at the individual joints. However, the strongest relationships are visible when observing the sum of the torques and powers at all three lower extremities combined. Not only does this refute the hypothesis, but the results of this study also reveal that aging does not change the mechanics of velocity modulation during walking.

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APPENDIX

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
1L-09 Brody Medical Sciences Building• 600 Moye Boulevard • Greenville, NC 27834
Office **252-744-2914** • Fax **252-744-2284** • **www.ecu.edu/irb**

TO: Paul DeVita, PhD, Department of EXSS, ECU, Mailstop #158

FROM: UMCIRB

DATE: October 19, 2010

RE: Expedited Category Research Study

TITLE: "Walking and Running at Different Speeds"

UMCIRB #10-0548

This research study has undergone review and approval using expedited review on 10/18/10. This research study is eligible for review under an expedited category number 4 & 6. 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 **10/18/10** to **10/17/11.** The approval includes the following items:

- Internal Processing Form (dated 9/28/10)
- Informed consent (received 10/7/10)
- COI disclosure form (dated 9/28/10)
- Telephone interview for general medical and mobility status
- Medical survey

The Chairperson (or designee) does not have a potential for 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
IRB00004973 East Carolina U IRB #4 (Behavioral/SS Summer) IORG0000418
Version 3-5-07

UMCIRB #10-0548
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