

Timothy J. Copple. STRIDE LENGTH MANIPULATION IN YOUNG AND OLD ADULTS DURING LEVEL WALKING. (Under the direction of Dr. Paul DeVita) Department of Exercise and Sport Science, April 2010.

Physiological and neurological changes with healthy aging cause old adults to alter biomechanical gait strategies. Mechanical plasticity is an ambulatory strategy in which old adults rely on proximal musculature in compensation for decreased distal muscle functioning. Since stride length has been shown to decrease with age, mechanical plasticity may be directly related to the control of stride length. It was hypothesized that old adults rely on hip joint torque and power more than knee or ankle torques and powers when manipulating stride length. It was also hypothesized that young adults rely on even distribution of lower-extremity joint torques and powers when manipulating stride length. The purpose of this study was to identify the relationship between lower extremity joint torques and powers and stride length in old and young adults while walking at an identical velocity.

Healthy young (ages 18-27) and old adults (ages 70-85) were instructed to walk across a level walkway at $1.50 \text{ m/sec} \pm 5\%$. Twenty strides ranging from each subject's shortest to longest strides were collected per subject. Stride length was manipulated from trial to trial to ensure that each subject had a relatively even distribution of stride lengths from shortest to longest strides. Ground reaction force and joint kinematics were collected and analyzed with inverse dynamics. Pearson product correlation analyses were used to identify relationships among individual joint torque and power variables and stride length. Stepwise regression analyses were used for a comprehensive view of all lower-extremity joint torques and powers.

Means of preferred and maximal stride lengths were shorter for old adults than young adults. Correlations provided from averaging individual subject correlations within each group resulted in strong predictability of stride length. This method of evaluating how old and young adults manipulate stride length more accurately identified how young and old subjects manipulated stride length. These results indicated that knee and ankle torques and powers were stronger predictors of stride length than hip torque and power. Also, all young adult correlations were stronger than corresponding old adult correlations. For example, young adult knee impulse ($r=0.864$, $r^2=0.746$, <0.05) had a stronger relationship with stride length than old adult knee impulse ($r=0.837$, $r^2=0.701$, <0.05). Stepwise regression analyses similarly suggested high predictive power of distal joint function. According to these regressions, hip variables were not predictive of young adult stride lengths while hip impulse, following ankle and knee impulse, was predictive of old adult stride lengths.

This study suggests young and old adults manipulate stride by altering knee and ankle muscle functioning more than by altering hip muscle function. These data did not support the proposed hypotheses. Stronger correlations for young adults suggest these individuals can more accurately control stride length with knee and ankle torques than old adults.

**STRIDE LENGTH MANIPULATION IN YOUNG AND OLD
ADULTS DURING LEVEL WALKING**

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CHAPTER 1: INTRODUCTION

Changes in gait mechanics, possibly caused by physiological and neurological adaptations, have been observed with healthy aging. Compared with young adults, old adults walk with a higher cadence, shorter stride length, and increased double support phase while walking at self-selected velocities (15; 40). It has been questioned whether these kinematic differences are consciously selected to provide more stable ambulation or if they are direct outcomes of ageing-related neuromuscular adaptations. Healthy old adults also exhibit decreased ability to produce ankle torque and power compared to young adults (17; 37). Because of these decreases in ankle kinetics, one would suspect old adults must compensate with the more proximal lower-extremity joint musculature in order to maintain functionality. Numerous investigators have observed this compensation in old subjects producing greater hip power and less ankle power than young subjects (3; 10; 15; 17; 26). Diminished ankle joint functioning can be explained by normal physiological alterations during healthy aging. Leg extensor strength and power decreases 1-2% each year in individuals over the age of 65 (33) while significant declines in isometric plantar flexion and dorsiflexion strength occur between early and late adulthood (37). Primarily associated with type II fibers, these decreases in lower-extremity strength are caused by loss of functioning motor units and skeletal muscle atrophy(4; 8). Age-related physiological and biomechanical changes in gait suggest that old adults have a distal to proximal shift in lower-extremity joint torques and powers.

Many kinematic characteristics, including stride length, are commonly used as descriptive variables during gait analysis. For example, reduced stride length is often one of the early observable gait alterations associated with many disorders such as Alzheimer's and Parkinson's disease (22; 41). Also, maximum step length, a measurement of one's ability to

maximally step out and back, declines with age and is negatively correlated with fall risk and mobility impairment in healthy and balance-impaired old adults (6; 19). Increases in lower-extremity strength have been shown to improve gait parameters, including stride length (25). Maximal step length has been reported to be predictive of hip and knee extensor strength, speed, and power (30). Surprisingly, despite stride length being used extensively as a descriptive factor to evaluate gait across multiple populations, little is known of the particular biomechanical factors that directly influence stride length.

Hypothesis

Age-related physiological and biomechanical reductions in lower extremity joint musculature appear to be related to the observed distal to proximal shifts of joint torques and powers during ambulation. This mechanical plasticity in locomotion with age can be used as the basis for a hypothesis describing how old and young adults control stride length. It was hypothesized that old adults rely on hip joint torque and power more than knee or ankle torques and powers when manipulating stride length. It was also hypothesized that young adults rely on an even distribution of lower-extremity joint torques and powers when manipulating stride length.

Purpose

The purpose of this study was to identify the relationships between lower extremity joint torques and powers and stride length in old and young adults while walking at the standard lab velocity of $1.50 \text{ m/s} \pm 5\%$.

Significance

The literature shows support for the theory of a distal to proximal shift in lower extremity joint torques and powers with age. No studies, however, have considered the influence of this

mechanical plasticity on stride length. This study is the first to control gait speed while manipulating stride length in order to observe the relationship between stride length and mechanical plasticity. Despite stride length being a fundamental variable of gait, presently, little is known of the underlying mechanics employed to manipulate it. This investigation will explore and help identify these underlying factors.

Delimitations

- 1.) All subjects will be healthy, without a history of lower extremity pain or injury, neuromuscular or musculoskeletal diseases, or other orthopedic problems.
- 2.) Young subjects will be males and females ages 18-27 years and old adult subjects will be males and females ages 70-85 years.
- 3.) Subjects do not need assistance or have difficulty performing activities of daily living (ADLs).
- 4.) All subjects will have a Body Mass Index of less than 30 kg/m^2 .
- 5.) Biomechanical analysis will focus on gait characteristics and calculations of joint torque and joint power.
- 6.) This study design only examines level walking in young and old adults.
- 7.) All subjects will walk at an accepted velocity of $1.50 \text{ m/sec} \pm 5\%$ (10).

Limitations

- 1.) The analysis is limited to the accuracy of the force plate and camera system.
- 2.) Kinematic and kinetic data may be affected by reflective marker placement or movement artifact.
- 3.) Fatigue may occur because of the testing procedures, especially in the older adults.
- 4.) Inverse dynamics calculations do not identify subtle, individual muscle function.
- 5.) Kinematic and kinetic data were collected on the right leg, assuming symmetry between legs.

Assumptions

- 1.) The equipment placed on the subjects' limb will not interfere with natural gait.
- 2.) Kinematic analysis assumes a Newton-Euler Rigid Link Model.
 - a. Each segment has a fixed mass located at the segmental center of mass.
 - b. The location of the segmental center of mass remains fixed.
 - c. Joints are considered frictionless, hinge joints.
- 3.) Net torque and joint forces are assumed to be applied from a sum of muscle forces, passive tissue forces, and frictional force joints.
- 4.) Anthropometry is an appropriate method of calculating segment dimensions, segment center of mass, and moment of inertia.

CHAPTER 2: REVIEW OF LITERATURE

The purpose of this study is to identify the relationship between lower extremity joint torques and powers and stride length in old and young adults. This chapter will review literature of previous research examining, 1) neuromuscular and physiological changes with age, 2) changes in gait kinematics and kinetics with age, and 3) influences on stride length.

Neuromuscular & Physiological Changes with Aging

Aging causes physiological declines that result in less efficient physical functioning. These changes do not occur equally in all cells or all locations in the body. Decreases in functioning with age result from losses of total motor units (MU), greater losses of fast MU, decreases in whole muscle and fiber size, losses in number of muscle fibers, and declines in muscular strength.

Adults age 60 and older, rapidly lose functioning MUs, while surviving units are typically enlarged and in a slow twitch nature (4). This size increase possibly occurred from MUs reinnervating muscle fibers (35). These authors also found conduction velocities significantly slower at more distal regions. Wang, de Pasqua, & Delwaide (39) found a gradual decrease in total motor units throughout adulthood. Specifically, it was predicted that from ages 20 to 90, 80% of total MUs would be lost. These authors also found fast twitch MUs to decrease much faster than slower MUs, as adulthood progresses.

Decreases in fast twitch MUs may lead to changes in muscle fiber properties. Total muscle area decreases during aging, specifically in regards to fast twitch fibers. Despite muscle area being smaller in old adults compared to young adults, there may not be percentile differences of type I, type IIa, and type IIb muscle fibers between old and young (8). However, type I fibers occupied a larger percentage of total muscle area in old compared to young. This

increase in percentage of type I fiber area was caused by a reduction in area of type II fibers. Reinnervation may be a reason for the decreases in type II fiber area (38).

Healthy aging affects upper and lower extremities as well as distal and proximal muscles differently. Frontera, Hughes, Fielding, Fiatarone, Evans, & Roubenoff (13) tested isokinetic strength in men around the age of 65. Twelve years later, the same individuals were tested again. Knee flexors and extensors lost significantly more strength than elbow flexors and extensors. Similarly, old versus young rats have significantly less muscle mass in the distal soleus, but do not have different muscle mass in the more proximal adductor longus (2). As individuals age, maximal voluntary contraction declines at a faster rate in more distal than proximal muscle groups (1).

Changes in Gait Kinematics & Kinetics with Aging

Gait differences between old and young adults at self-selected walking velocities may be attributable to adaptations related to a fear of falling in old. Old adults, walking with a similar cadence and slower velocity, walk with shorter strides than young adults (40). During this apparently safer gait, heel-strike occurred in a more flat-footed position and push-off was less vigorous in old adults, resulting in differences in ankle power between old and young (40).

While walking at a normal pace, old adults tend to walk at a slower velocity, have a faster cadence, and take shorter strides than young adults. Kerrigan, Todd, Della Croce, Lipsitz, & Collins (15) found hip and ankle powers may be related to stride length. Old adults took shorter strides and were shown to have lower plantarflexor power and greater hip flexor power late in the stance phase, compared to young. However, the investigators were unable to determine if shorter strides were a product of joint powers or stability mechanisms.

When all participants walked at the same speed, DeVita & Hortobagyi (10) observed old adults walking with a higher cadence, longer support phase, and greater range of motion in the hip joint while range of motion in the ankle joint was less compared to young. Support torque was found to be nearly identical between groups. However, hip extensors were used more and hip flexors were used less in old than young adults. In turn, knee extensors and ankle plantarflexors contributed less to this total output in the old. Specifically, plantarflexor function was reduced by 25% in the old participants during stance (10). In summary, torque distribution in old adults was higher in the hip and lower at the knee and ankle compared to young adults.

Savelberg, Verdijk, Willems, & Meijer (28) studied gait differences between both old and young active and inactive individuals. As with most studies, old adults walked slower and with a shorter stride lengths when asked to walk comfortably. While walking at identical velocities, active old adults produced support torque similar to those of both young groups. Inactive adults produced significantly less total torque than all other groups. Like the results reported by DeVita & Hortobagyi (10), specific joint torques were different between young and old. Old adults relied more on hip function and less on knee and ankle function to ambulate. This redistribution of function was more prominent in active old adults.

Influences on Stride Length

Numerous studies have reported that old adults take shorter stride lengths than young adults during normal, level walking (12; 24). Despite such wide use of stride length in research and clinical settings, little is known on how it is manipulated. In fact, the function of the plantarflexors is not completely understood. During two separate studies (32; 36), a tibial-nerve block of one leg caused reductions in step length of the intact leg, leading the investigators to conclude the primary function of plantarflexors to be resistance of forward momentum.

Neptune, Kautz, & Zajac (23) stated that making these conclusions is difficult because the nerve block alters more than just plantarflexor function, resulting in numerous adaptations in gait mechanics (velocity, step time, and joint angles).

Old adults tend to produce shorter, more stable strides by plantarflexing less and therefore producing less ankle work at terminal stance phase, compared to young adults (40). Because these individuals took shorter strides and heel contact was achieved at a reduced angle at the ankle, energy absorption was less than the young adults. Similarly, Donelan, Kram, & Kuo (11) investigated mechanical and metabolic work during the step-to-step transition. During single support phase, center of mass motion resembles that of an inverted pendulum and theoretically should not require mechanical work to continue the motion. However, work is needed to transfer COM from one arc to the next as an individual begins single support with the other leg.

Decreased lower-extremity strength, observed in old adults (33; 37), may limit the ability to produce greater torques and forces, and in turn restrict manipulation of stride length. While controlling for velocity, Kang & Dingwell (16) concluded that lower strength values in old adults were attributed to differences in stride length between old and young adults. Quadriceps strength has been shown to have a moderately high correlation ($r=0.558$) to stride length (21). Improving lower-extremity strength is also linked to improvements in the gait of pathological individuals. After eight weeks of resistance training, adults with mild-to-moderate Parkinson's disease significantly improved lower-extremity strength, stride length, and walking velocity (29). Individuals at least one year post stroke also increased stride length following four weeks of resistance training (42).

Summary

Physiological and neuromuscular changes occurring with age cause decreases in physical functioning of healthy old adults. These changes include losses of MUs and muscle fiber size. Losses appear to affect fast twitch MUs to a greater extent than their slow twitch counterparts. Also, these changes appear to affect distal muscles more extensively than proximal muscles.

These decreases of function in old adults lead to adaptations during walking. Old adults generally prefer to walk at a slower velocity, take shorter stride lengths, have a higher cadence, and increased double support phase than young adults. A tendency to rely more on hip joint torque and power than knee or ankle torques and powers is observed in old populations.

Despite great amounts of literature comparing stride lengths between groups, little is known about biomechanical factors that influence these stride variations. Some individuals may select shorter, more stable strides requiring less energy production and absorption than longer strides. More stable strides may also be selected because greater forces may not be able to be produced by individuals with decreased lower-extremity strength.

CHAPTER 3: METHODOLOGY

Subject Characteristics

Table 1 lists subject characteristics. The following subject inclusion and exclusion criteria were used (see Appendix D):

Inclusion Criteria:

1. Apparently 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 (ADLs).
3. BMI less than or equal to 30.0 kg/m².
4. Provide written informed consent.

Exclusion Criteria:

1. Difficulty performing ADLs including the use of an ambulatory device or experiencing more than one fall within the past year.
2. Smoking cigarettes within the last year.
3. Neurological problems including stroke, dementia, epilepsy, Parkinson's disease, etc.
4. Musculoskeletal problems including arthritis, osteoporosis, joint replacement, lower extremity or back surgery.
5. Cardiovascular problems including heart attack, high cholesterol, uncontrolled high blood pressure, pace maker, coronary artery disease, and peripheral artery disease.
6. Other health problems including cancer, diabetes, vision problems, etc.

Table 1: Subject Characteristics. Values are mean (SD).

	n	Age (yr)	Mass (kg)	Height (m)	BMI (kg/m²)
Young	16	21.9 (2.6)	75.4 (17.0)	1.74 (0.11)	24.5 (3.6)
Old	21	76.1 (3.5)	66.2 (12.7)	1.68 (0.09)	23.4 (3.0)

The young adult group was comprised of 10 males and 6 females while the old adult group had 6 males and 15 females. Chumanov, Wall-Scheffler, & Heiderscheit (7) reported no significant differences in sagittal plane mechanics between genders. While Kerrigan, Todd, & Croce (17) found differences in sagittal plane kinetics, these occurred only during the swing phase. Due to the investigation of only sagittal mechanics during stance phase, the difference in number of males and females will not affect the results of this study.

Equipment

Level walking kinematic and kinetic data was collected with 8 Qualisys ProReflex MCU 240 cameras (Qualisys Medical AB, Gothenburg, Sweden) at 120 Hz. The cameras were placed in a circular arrangement, focused on a common collection volume. Ground reaction forces were measured by force platform (AMTI Model LG-6, Newton, MA) located approximately in the center of the volume at a collection frequency of 960 Hz and gain of 4000. Gait speed was controlled using an infrared timing system (Brower timing system, model IRD-T175, Salt Lake City, Utah). All data was collected with Qualisys Track Manager Software (Innovision Systems Inc., Columbiaville, MI) and analyzed by Visual 3D (C-Motion Inc., Rockville, MD). Data from all subjects was compiled and group means were calculated using proprietary laboratory software. Height in meters and weight in kilograms was measured for all subjects using a Seca 703 scale (Seca gmbn & Co.kg, Hamburg, Germany).

Procedures

Subjects were recruited by various announcements on the ECU campus, in local newspapers, on a local television station and from previous studies. Potential subjects contacted the lab to conduct a telephone interview with a research associate (Appendix C). The interview

determined if the subjects were eligible to participate based on specific inclusion/exclusion criteria. Subject who passed this interview were scheduled for data collection.

Testing was conducted in one session lasting approximately 90-120 minutes. All testing procedures were conducted in the Biomechanics Laboratory, located in room 332 of Ward Sports Medicine Building, East Carolina University, Greenville, NC. Subjects were instructed to read and sign the informed consent form (Appendix A). Subjects' blood pressure, height, and weight were taken before testing began. Blood pressure was obtained for health-safety reasons. Subjects wore form fitting shorts and personal athletic shoes.

Subjects were timed and observed as they completed four functional tasks including level walking for 17.5 m, ascend and descend 12 standard steps, and a "get up and go" (rise from a chair, walk 2 m, pick up a small object, and return to the seated position). All functional tasks were performed in random order and at subjects' normal pace. For the purpose of this thesis, only the level walking time will be reported. Self-selected walking speed is a valid method of determining functionality (34). At 12.74 (± 0.04) s, young adults walked 8% faster than old adults at 13.79 (± 2.05) s. The similarity between these data suggests that the tested old adults were highly mobile and functional, had no walking related disability, and represented the population of healthy old adults.

Reflective markers were placed on several anatomical landmarks of subjects' pelvis and right leg (Appendix E). The anatomical landmarks included: right and left anterior superior iliac spine, lumbosacral joint (L5-S1), right and left iliac crest, right and left greater trochanter, medial and lateral joint line of the knee, medial and lateral malleoli, right and left heels, and the 1st and 5th metatarsal heads. Rigid rectangular plates with a marker attached to each corner were also placed on the lateral thigh and lateral shank. A three-marker triangular plate was placed over the

midfoot. A five second static trial of each subject standing in the anatomical position with the arms crossed over the chest was collected. The calibration markers (medial and lateral joint markers) were removed and a second static trial was recorded.

Subjects received instruction regarding the appropriate walking speed, starting location, and starting foot. Subjects were asked to speed up or slow down if they walked more than +/-5% of 1.50 m/s. The subjects were instructed to walk through and past the collection volume to ensure a constant gait velocity. The experimenter modified the starting point in order for the right foot to naturally step on the center of the force plate. A successful trial consisted of a subject stepping on the force plate with the entire right foot, walking at the appropriate velocity, and all reflective markers being recorded. The first two trials were collected at subjects' preferred stride lengths. Subsequent trials consisted of only one stride length between 1.20 m and subjects' longest stride lengths. A wide range of stride lengths were collected by having the investigator give instructions as to which stride lengths were to be produced by the subjects on a trial by trial basis. Trials were not collected if stride lengths varied within trials, if subjects appeared to unnaturally lunge during long strides, or if a particular stride length had already been recorded. Data collection ended when an even distribution of short to long stride lengths were recorded, with a maximum of twenty successful trials. Because joint torques and powers are used to control gait characteristics, stride length was used as a dependent variable.

Data Processing

Data were processed with Qualysis Track Manager software, which produced position data for all subjects and trials in the global coordinate system (GCS). Visual 3D was used to calculate joint torque and power at the hip, knee and ankle, through inverse dynamics. The model of the lower extremity was built as a rigid, linked segment system. The first standing

calibration trial was used to create individualized models for each subject as well as locate virtual joint centers, locate the segment center of mass, define the local coordinate system of each segment (LCS), and calculate a transformation matrix to determine the location of all markers in the GCS. The joint centers were found by calculating 50% of the distance between the medial and lateral joint calibration markers. The hip joint center was located 25% of the distance between the right and left greater trochanters. The long axis of each segment was defined by a line from the proximal joint center to the distal joint center. The segment COM was located using anthropometrics (9) and was measured from the proximal joint center. The LCS of each segment was located within the long axis of the segment.

Calculating the joint kinetics required transforming GRFs and torques, COP, force on the segment due to gravity, segment COM accelerations, proximal and distal moment arms, and proximal and distal joint center locations into the LCS. The calculations are performed on the foot segment first because the unknowns of the equations were the forces and moments at the proximal joint of the segment. Once found, the process was repeated for the leg, then thigh, and hip. Joint reaction forces (JRFs) and then joint torques were calculated for each frame of data. The JRF vector for the ankle is found by Equation 1:

$$\mathbf{Jrf}_{\text{ankle}} = m\mathbf{a}_{\text{cm}} - m\mathbf{g} - \mathbf{f}_{\text{grf}} \quad (1)$$

where m is the segment mass, \mathbf{a}_{cm} is the linear acceleration of the segment COM, $m\mathbf{g}$ is the gravity vector, and \mathbf{f}_{grf} is the GRF vector. The vector describing the ankle joint torque was expressed by the moments about the segment COM (Equation 2):

$$\mathbf{jm}_{\text{ankle}} = \mathbf{I}\alpha - (\mathbf{d}_1 \times \mathbf{jrf}_{\text{ankle}}) - (\mathbf{d}_2 \times \mathbf{f}_{\text{GRF}}) \quad (2)$$

where \mathbf{I} is the moment of inertia matrix, α is the angular acceleration matrix, $\mathbf{d}_1 \times \mathbf{jrf}_{\text{ankle}}$ is the vector describing the torque resulting from the JRF, $\mathbf{d}_2 \times \mathbf{f}_{\text{GRF}}$ is the vector describing the torque

resulting from the GRF. All force and torque calculations were performed in the LCS of the specific segment. Transformation of the joint force and moments were required and completed by the following matrices (Equation 3 and 4):

$$[\mathbf{JRF}_{\text{Ankle}}] = [\mathbf{T}_{\text{local2global}}] [\mathbf{jrf}_{\text{Ankle}}] \quad (3)$$

$$[\mathbf{JM}_{\text{Ankle}}] = [\mathbf{T}_{\text{local2global}}] [\mathbf{jm}_{\text{Ankle}}] \quad (4)$$

In order to continue inverse dynamics analysis for the other segments, the GRF component was replaced by the components of the distal JRF:

$$\mathbf{Jrf}_{\text{Prox}} = m\mathbf{a}_{\text{CM}} - m\mathbf{g} - \mathbf{jrf}_{\text{Distal}} \quad (5)$$

$$\mathbf{Jm}_{\text{Prox}} = \mathbf{I}\alpha - (\mathbf{d}_1 \times \mathbf{f}_{\text{jrf_Prox}}) - (\mathbf{d}_2 \times \mathbf{f}_{\text{jrf_Distal}}) - \mathbf{jm}_{\text{Distal}} \quad (6)$$

Joint power is the product of the joint torque and the joint angular velocity (27). It was calculated by the following equation:

$$\mathbf{P} = \mathbf{JM} \times (\mathbf{w}_{\text{Proximal}} - \mathbf{w}_{\text{Distal}}) \quad (7)$$

where \mathbf{P} is the joint power vector, \mathbf{JM} is vector representing the XYZ components of the joint torque and $\mathbf{w}_{\text{Proximal}}$ and $\mathbf{w}_{\text{Distal}}$ are the vectors representing the XYZ proximal and distal segment angular velocities.

GRF data in Newtons (N) was normalized to body mass (kg), joint torques in Newton-meters (Nm) and joint angular impulses in Newton-meters*second (Nms) were normalized to percent body weight multiplied by height (%BW*HGT), and joint powers in Watts (W) and joint work in Joules (J) were normalized to body mass (kg).

Statistical Analysis

Pearson product correlation analyses were used to identify relationships among individual joint torque and power variables and stride length. Significance was tested a level of 0.05. Based on the sample sizes in the study, r values greater than 0.444 for individuals, greater than=0.112

for the young group, and greater than 0.101 for the old group were statistically significant. For the purpose of this study, results were used for qualitative comparison. Stepwise regression analyses were used for a comprehensive view of all lower-extremity joint torques and powers with an alpha level of $p \leq 0.05$ indicated statistical significance.

CHAPTER 4: RESULTS

It was hypothesized that old adults rely on hip joint torque and power more than knee or ankle torques and powers when manipulating stride length. It was also hypothesized that young adults rely on even distribution of lower-extremity joint torques and powers when manipulating stride length. The purpose of this study was to identify the relationship between lower extremity joint torques and powers and stride length in old and young adults while walking at a velocity of 1.50 m/sec. This chapter is separated into the following sections 1) Walking Characteristics, 2) Joint Angular Impulse, 3) Joint Work, 4) Stepwise Regressions, and 5) Summary.

Walking Characteristics

Successful trials were collected if walking velocity was within $\pm 5\%$ of 1.50 m/s. Mean velocity for young was 1.50 (SD 0.07) m/s while old walked at 1.53 (0.06) m/s (Figure 1). This was a difference of only 2% between groups.

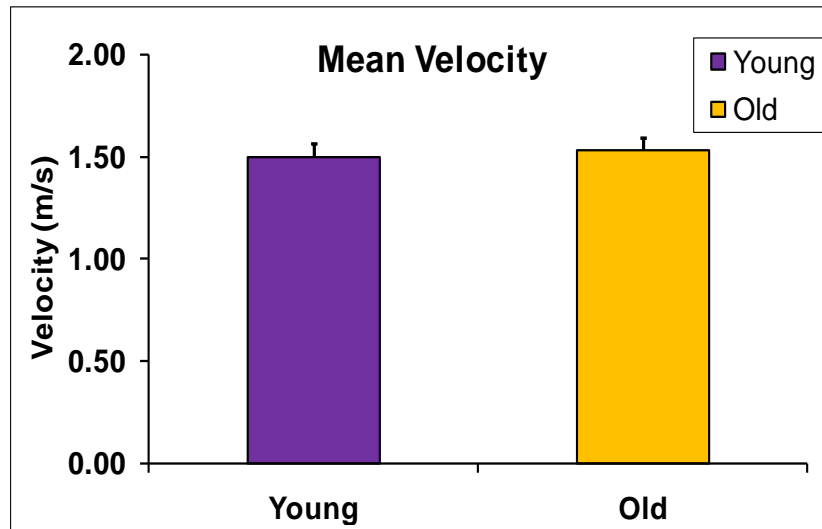


Figure 1: Mean velocity for young and old adults. Error bars are +SD.

Mean absolute stride length was 1.60 (0.10) m for young and 1.52 (0.08) m for old (Figure 2). Mean stride length normalized to height was 0.92 (0.06) m for young and 0.90 (0.04) m for old (Figure 2). All of the following stride length characteristics are presented in Figure 3.

Mean minimum stride length was nearly identical between groups, with young at 1.25 (0.11) m and old at 1.24 (0.10) m. However, mean maximum stride length was 15% longer in young at 2.17 (0.19) m and while old were at 1.89 (0.12) m. Because old could not reach the maximum strides of the young, mean stride length range was 1.37 m for young and 1.00 m for old.

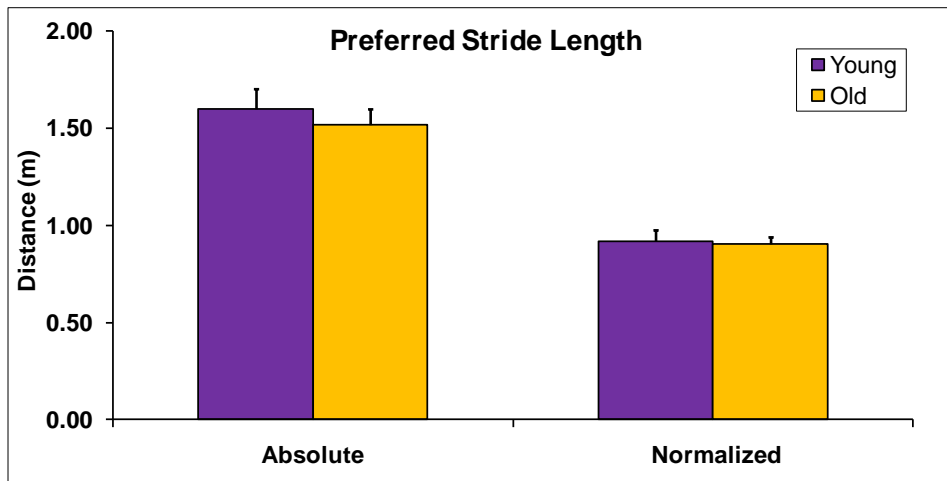


Figure 2: Mean preferred stride length for young and old adults. Error bars are +SD.

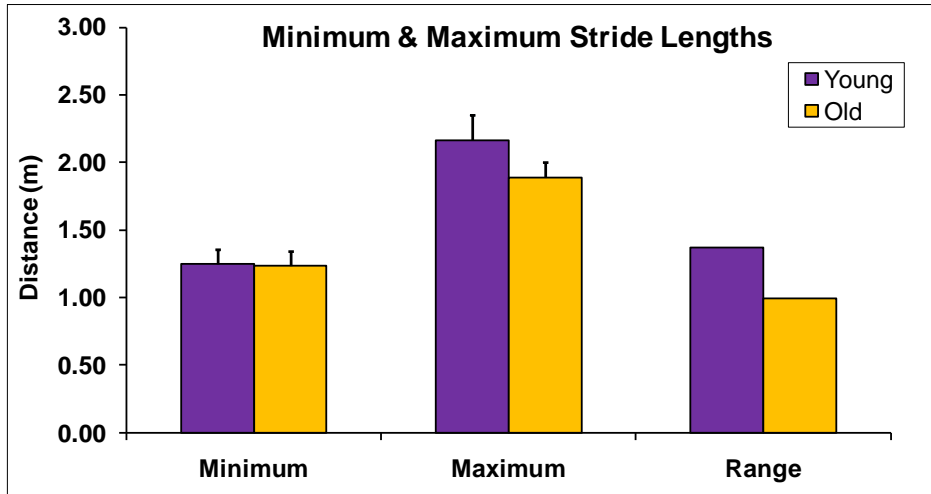


Figure 3: Mean absolute stride lengths for young and old adults. Error bars are +SD.

All young stride lengths produced a gradually increasing curve (Figure 4A) from group minimum (1.08 m) to maximum (2.45 m). All old stride lengths showed a similarly gradual curve between minimum (1.04 m) and maximum stride lengths (2.04 m) (Figure 4B). Individual

subjects in both young (Figure 5A) and old (Figure 5B) groups showed the same gradual increase of stride lengths.

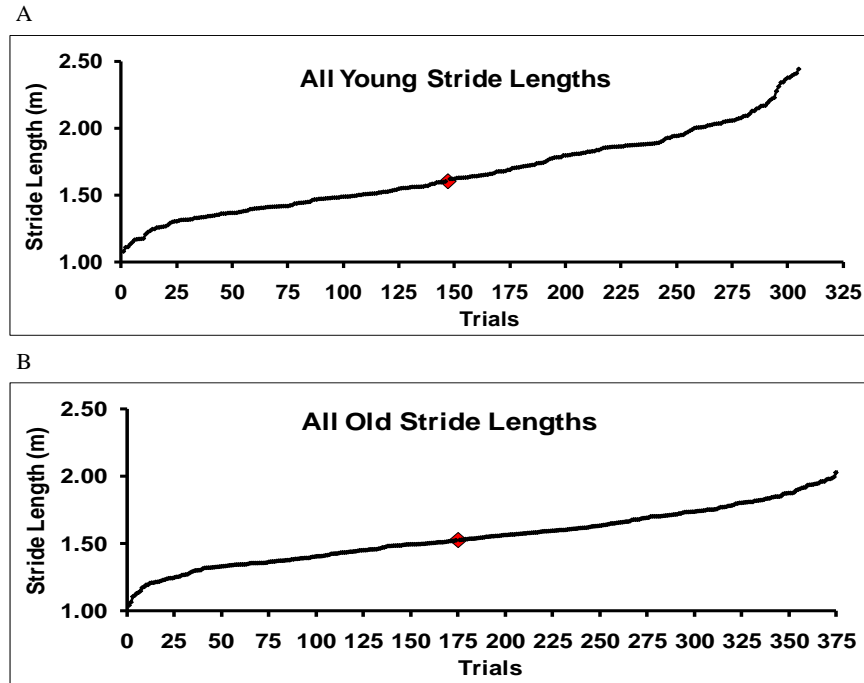


Figure 4: All stride lengths for (A) young and (B) old groups. Red data points represent mean preferred stride length.

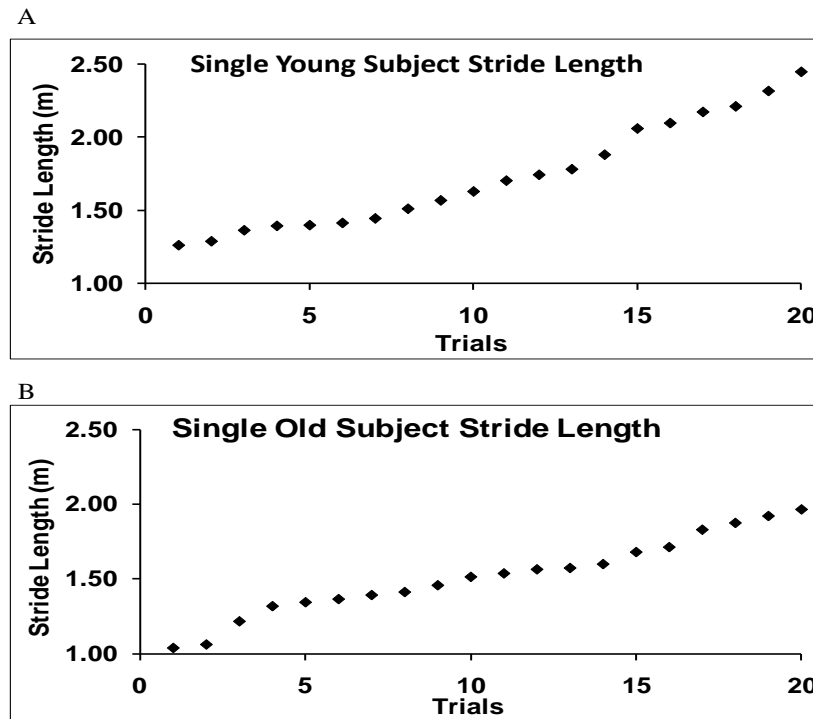


Figure 5: Single (A) young and (B) old subject stride length

Joint Angular Impulse

Joint torques were assessed by examining selected areas under the torque curves, positive extensor and plantarflexor angular impulses, during the stance phase. The following figures compare young group (A) and old group (B) correlations between joint angular impulses and stride length. Within these figures, the single young (C) and the single old (D) subject, whom most closely represented means of the mean of individual subject correlations between angular impulse and stride length, are also shown. These correlations were qualitatively and not statistically compared between groups. Relationships between hip extensor angular impulse and stride length were relatively weak in both young ($r=0.494$, $r^2 = 0.244$, $p<005$, Figure 6A) and old ($r=0.381$, $r^2 = 0.145$, $p<005$, Figure 6B) groups. Individually, however, young ($r=0.755$, $r^2 = 0.570$, $p<005$) and old ($r=0.708$, $r^2 = 0.501$, $p<005$) subjects had moderately strong relationships between hip extensor torque and stride length with the relationship being stronger in the young subjects.

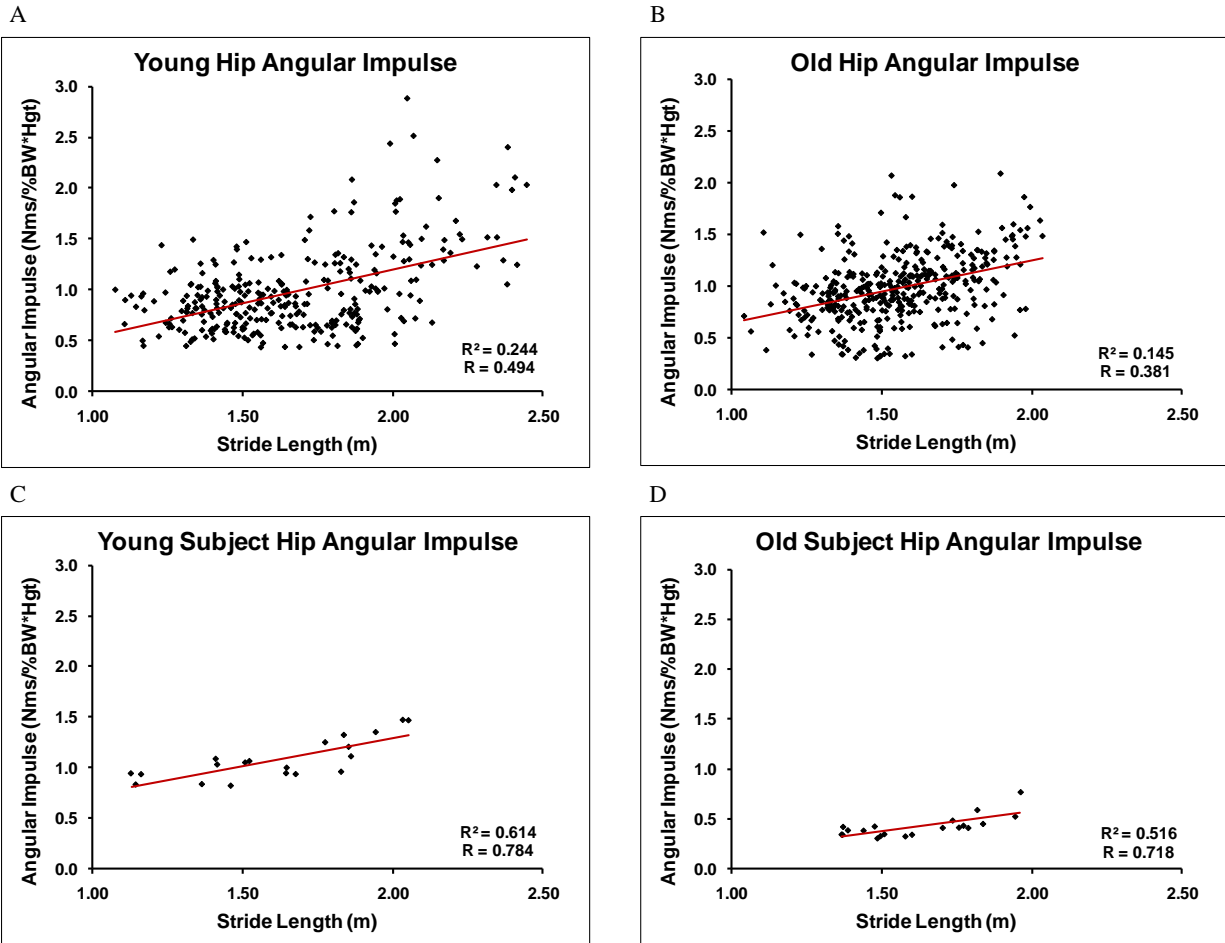


Figure 6: Relationships between hip angular impulse and stride length for (A) young group, (B) old group, (C) single representative young subject, and (D) single representative old subject.

Young group ($r=0.687$, $r^2 = 0.472$, $p<005$, Figure 7A) knee impulse correlation with stride length was slightly higher than old group ($r=0.617$, $r^2 = 0.381$, $p<005$, Figure 7B) knee correlation; both knee relationships were slightly higher than hip correlations (Figure 6). Individual knee impulse and stride length correlations were very strong and quite similar between young ($r=0.864$, $r^2 = 0.746$, $p<005$) and old ($r=0.837$, $r^2 = 0.701$, $p<005$) subjects.

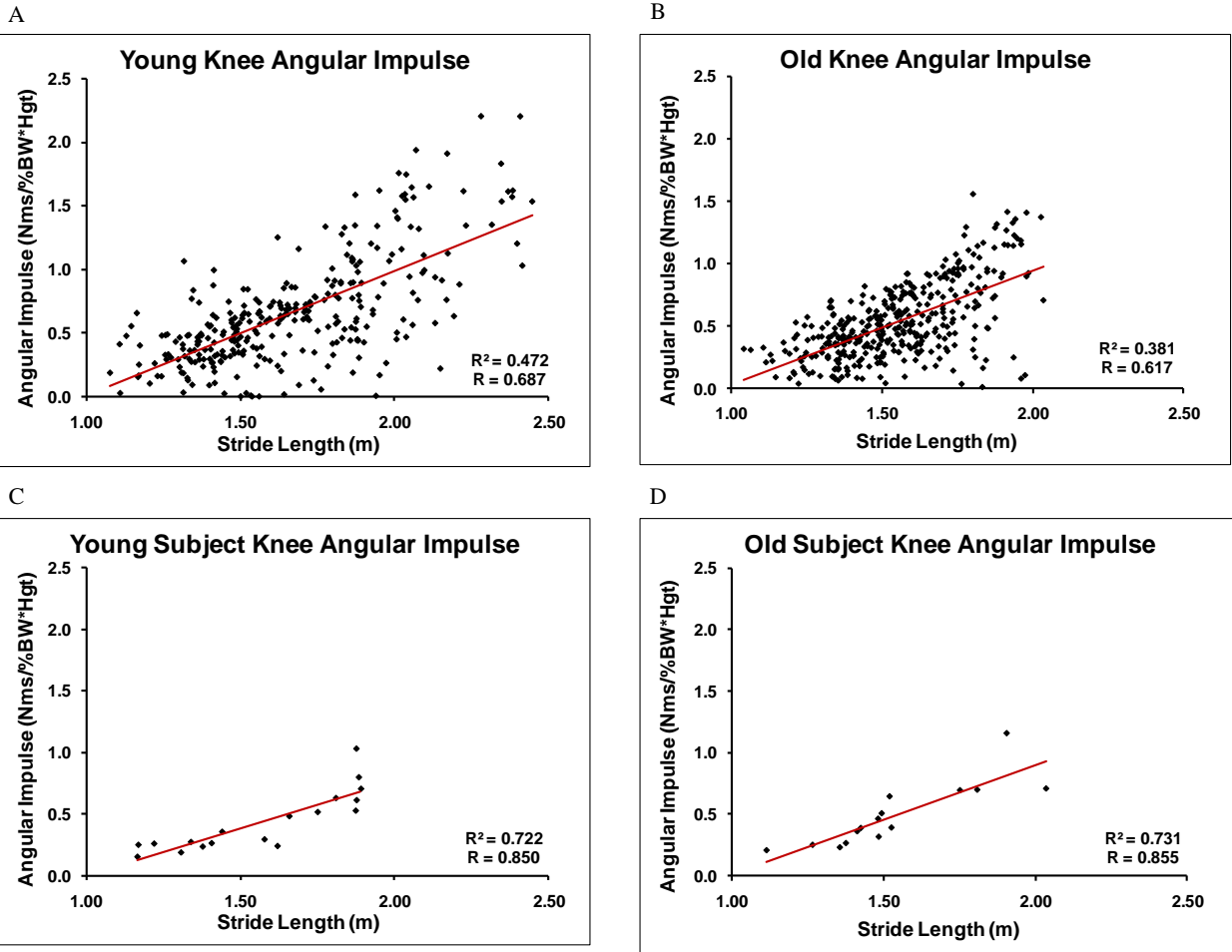


Figure 7: Relationships between knee angular impulse and stride length for (A) young group, (B) old group, (C) single representative young subject, and (D) single representative old subject.

Ankle impulse-stride length relationship was moderately strong for both young ($r=0.763$, $r^2 = 0.582$, Figure 8A) and old ($r=0.650$, $r^2 = 0.422$, Figure 8B) groups. Young ($r=0.860$, $r^2 = 0.740$, $p<005$) and old ($r=0.796$, $r^2 = 0.634$, $p<005$) individual correlation means also showed moderately high relationship between ankle impulse and stride length.

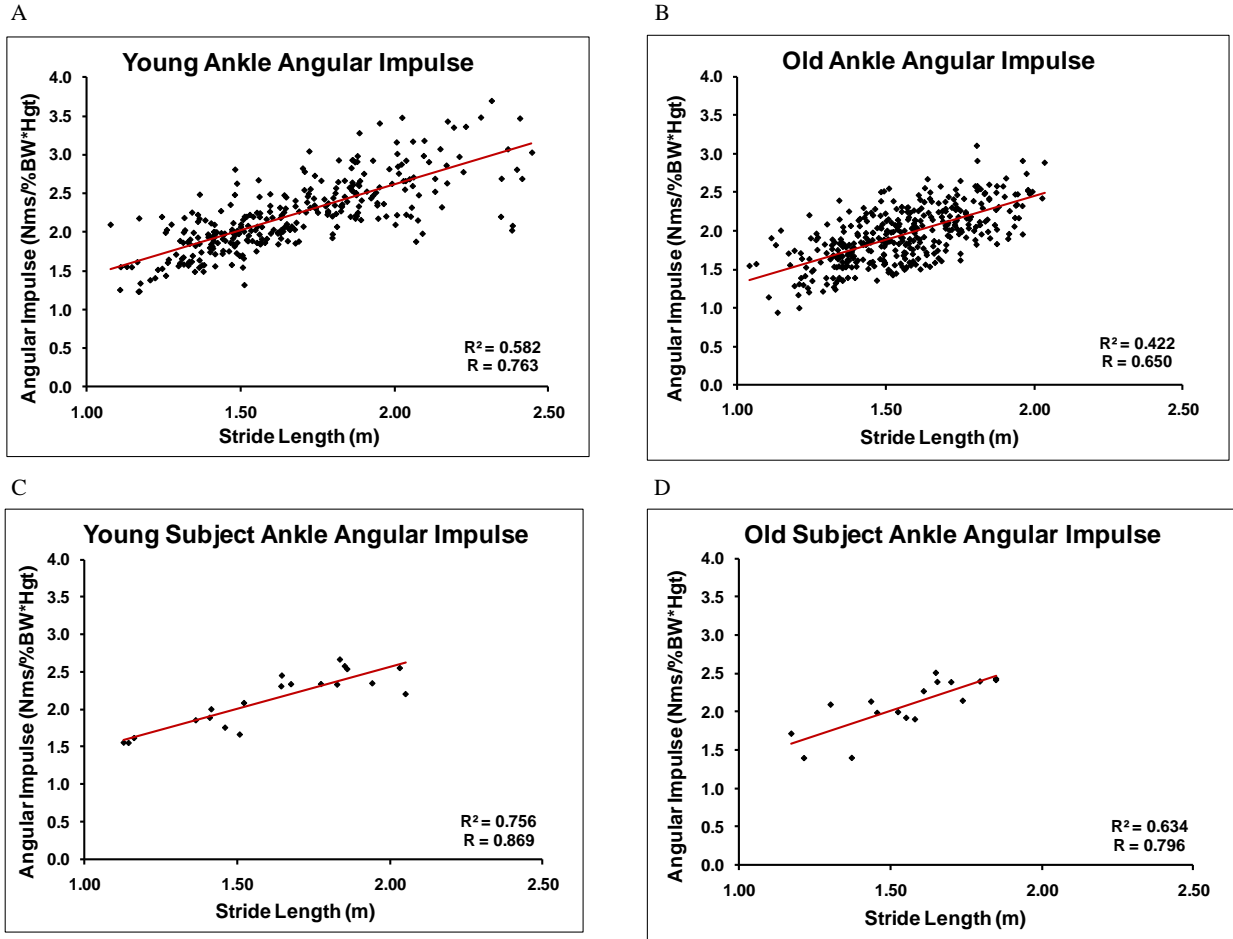


Figure 8: Relationships between ankle angular impulse and stride length for (A) young group, (B) old group, (C) single representative young subject, and (D) single representative old subject.

Joint Work

Joint powers were assessed by examining selected areas under the power curves, positive extensor and plantarflexor work, during the stance phase. The following figures compare young group (A) and old group (B) correlations between positive joint work and stride length. Within these figures, a single young (C) and old (D) subject, both of whom most closely represented means of the individual subject correlations between joint work and stride length, are also shown. Again, these correlations were qualitatively and not statistically compared between groups. Hip joint work was weakly correlated with stride length for both young ($r=0.422$, $r^2 = 0.178$, Figure 9A) and old ($r=0.157$, $r^2 = 0.025$, Figure 9B) groups. Individual subject hip and

stride length relationships were slightly higher than group correlations and, young subjects ($r=0.702$, $r^2 = 0.493$, $p<005$) still exhibited a higher correlation than old subjects ($r=0.417$, $r^2 = 0.174$, $p<005$).

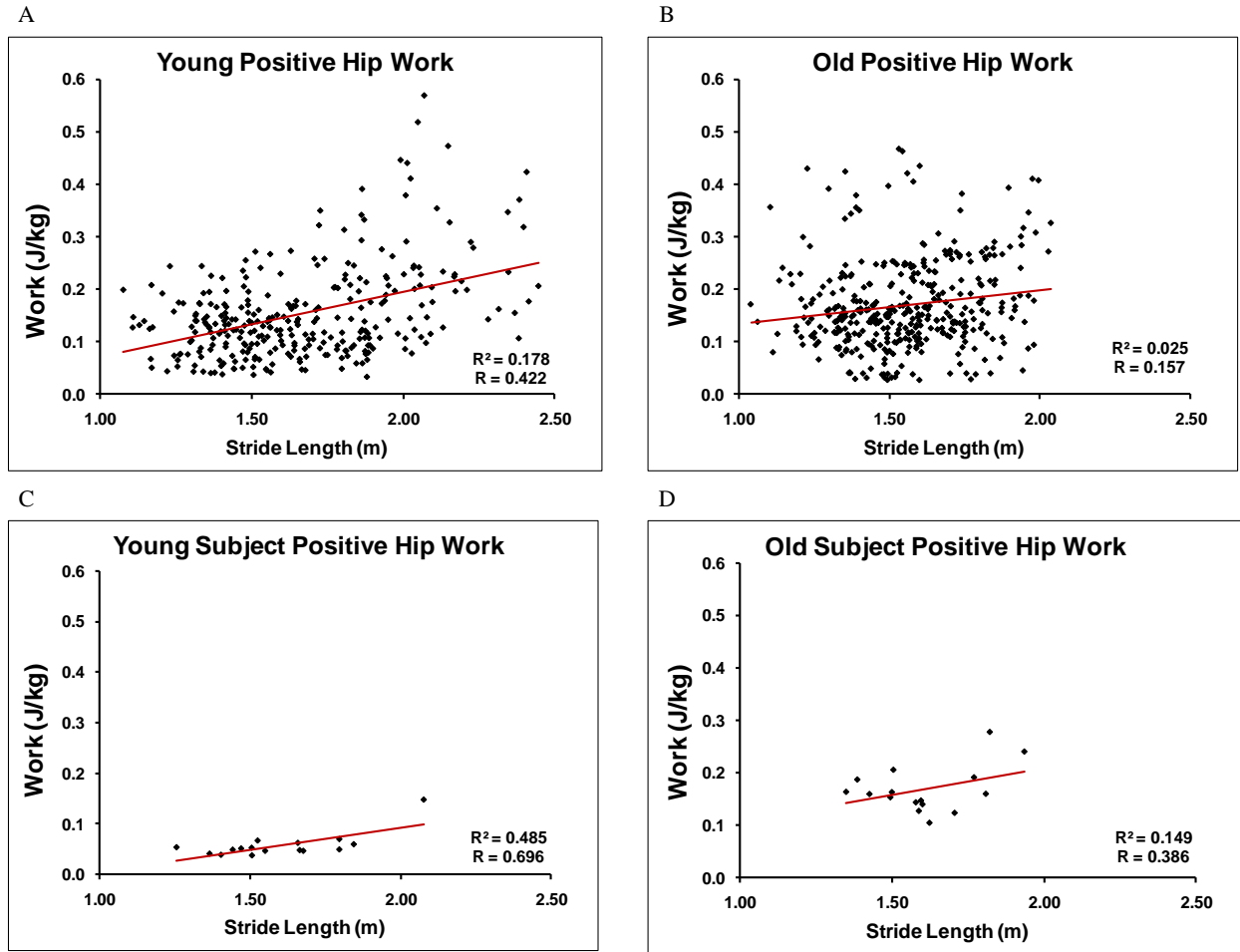


Figure 9: Relationships between positive hip work and stride length for (A) young group, (B) old group, (C) single representative young subject, and (D) single representative old subject.

Young group ($r=0.730$, $r^2 = 0.532$, Figure 10A) knee joint work was moderately predictive of stride length, while old group ($r=0.533$, $r^2 = 0.285$, Figure 10B) knee work was weakly correlated. As seen in the hip results, individual knee joint correlations were higher than whole group correlations. Young subjects ($r=0.844$, $r^2 = 0.712$, $p<005$) had a strong knee work correlation to stride length and old subjects' ($r=0.714$, $r^2 = 0.510$, $p<005$) correlations showed only moderate relationship.

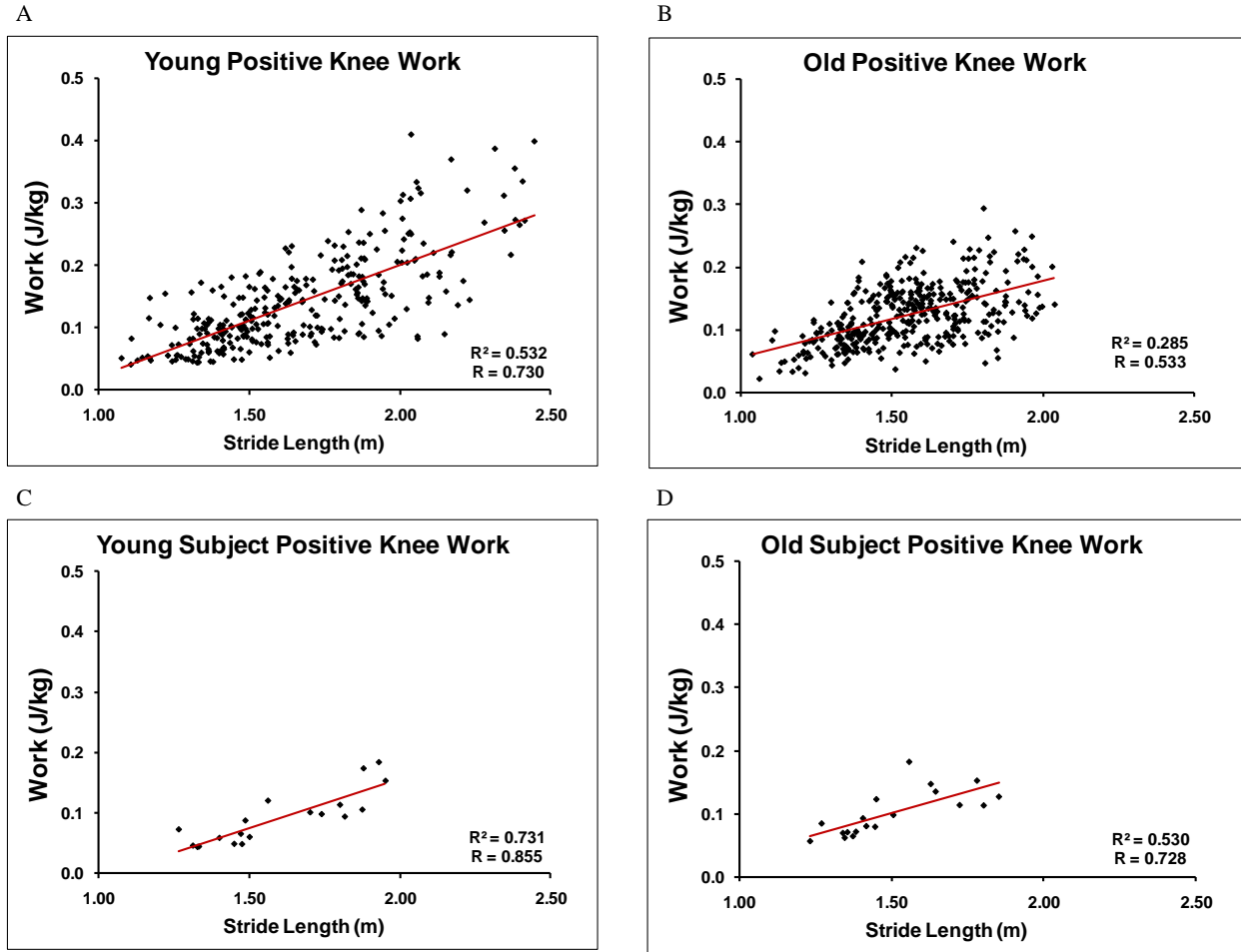


Figure 10: Relationships between positive knee work and stride length for (A) young group, (B) old group, (C) single representative young subject, and (D) single representative old subject.

Young group ($r=0.701$, $r^2 = 0.491$, Figure 11A) ankle work was a weaker predictor of stride length than knee work, but was a better predictor than hip work. Ankle work was the strongest predictor of stride length for the old group ($r=0.621$, $r^2 = 0.386$, Figure 11B). When individual subject correlations were considered, ankle work showed the greatest relationship to stride length for both young ($r=0.878$, $r^2 = 0.771$, $p<005$) and old ($r=0.826$, $r^2 = 0.682$, $p<005$) subjects.

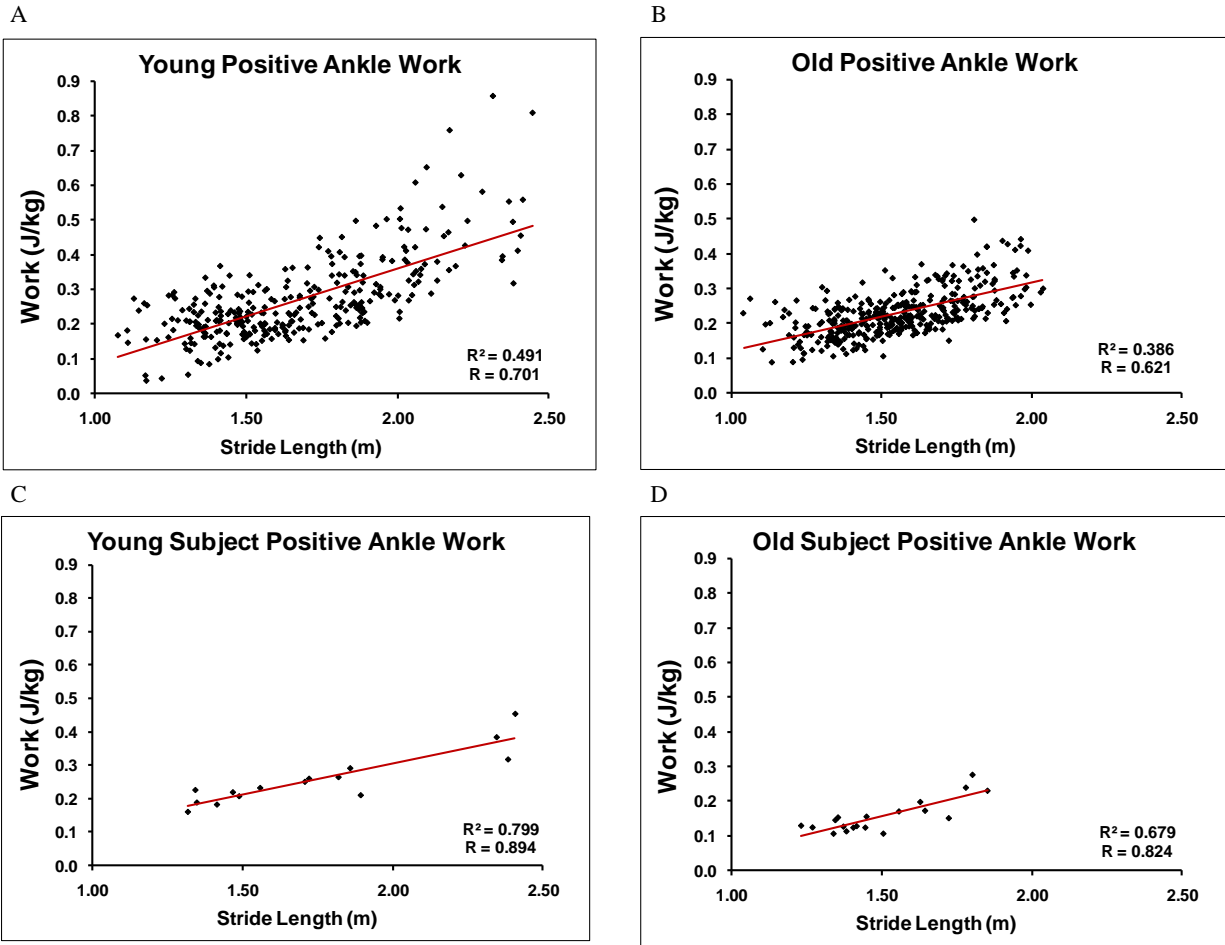


Figure 11: Relationships between positive ankle work and stride length for (A) young group, (B) old group, (C) single representative young subject, and (D) single representative old subject.

Stepwise Regressions

Tables 2 and 3 summarize stepwise regression results for the young and old groups, respectively. During this type of analysis, the strongest predictor of stride length forms model 1 and remains a constant. The analysis was run again to determine which new variable, along with the constant, would better predict stride length. Adding new variables is continued until additional variables no longer increase predictability.

For the young (Table 2), ankle impulse was the single greatest predictor of stride length, forming model 1. Model 2 added knee work, model 3 included ankle work, and model 4 introduced knee impulse. The complete model had a strong relationship with step length with a

correlation of $r=0.909$ and a coefficient of determination of $r^2=0.827$. Neither hip impulse nor hip work was included in the models.

Table 2: Young group stepwise results

Model	R	R Square	Std. Error of the Estimate
1	0.763	0.582	0.192
2	0.876	0.767	0.143
3	0.906	0.821	0.126
4	0.909	0.827	0.124

1. Ankle Impulse
2. Ankle Impulse, Knee Work
3. Ankle Impulse, Knee Work, Ankle Work
4. Ankle Impulse, Knee Work, Ankle Work, Knee Impulse

Table 3 shows stepwise results for the old group. As in the young group, ankle impulse was the greatest single predictor of stride length and formed model 1. Model 2 added knee impulse, model 3 added hip impulse, model 4 added knee work, and model 5 added ankle work. The complete model had a strong relationship with step length, with a correlation of $r=0.852$ and a coefficient of determination of $r^2=0.725$. Whereas the model for young adults did not include hip variables, hip impulse added significant predictability during model 3 in the old adults. The fact that ankle impulse explained 16% more variance in the young adults compared to the old adults may hint at a distal to proximal shift. Addition of hip impulse in the old adult regression, and not the young adult regression may also support this idea.

Table 3: Old group stepwise results

Model	R	R Square	Std. Error of the Estimate
1	0.650	0.422	0.156
2	0.800	0.640	0.124
3	0.822	0.676	0.117
4	0.836	0.699	0.113
5	0.852	0.725	0.108

1. Ankle Impulse
2. Ankle Impulse, Knee Impulse
3. Ankle Impulse, Knee Impulse, Hip Impulse
4. Ankle Impulse, Knee Impulse, Hip Impulse, Knee Work
5. Ankle Impulse, Knee Impulse, Hip Impulse, Knee Work, Ankle Work

Summary

The walking velocity and stride length data suggest the quality of the data was good. Young and old adult mean velocities were similar and within $\pm 5\%$ of the target value. Most importantly, the stride lengths for both young and old groups and young and old individuals were evenly distributed throughout the range of observed values. Preferred stride length for young adults was slightly longer than preferred stride length for old adults. Resulting in a greater range of strides, young took longer maximum and similar minimum lengths of strides, compared to old.

The relationships among the individual joint torques and powers were identified with Pearson product moment correlations. Individual subject correlations between joint angular impulses and stride length and joint work and stride length were stronger than group correlations. Young group and individual correlations were qualitatively higher than old group and individual correlations for every variable (Tables 4 and 5). Ankle and knee joint impulse and work appeared to be the best individual predictors of stride length for young and old adults. Surprisingly, old hip joint work was least correlated to stride length. As a whole, more distal knee and ankle joints were the best predictors of stride length.

Table 4: Joint angular impulse correlation r-values

		Young	Old
Hip	Group	0.494	0.381
	Ind.	0.755	0.708
Knee	Group	0.687	0.617
	Ind.	0.864	0.837
Ankle	Group	0.763	0.650
	Ind.	0.860	0.796

Note: All values statistically significant $p < 0.05$

Table 5: Positive joint work correlation r-values

		Young	Old
Hip	Group	0.422	0.157
	Ind.	0.702	0.417
Knee	Group	0.730	0.533
	Ind.	0.844	0.714
Ankle	Group	0.701	0.621
	Ind.	0.878	0.826

Note: All values statistically significant $p < 0.05$

CHAPTER 5: DISCUSSION

Lower-extremity joint torques are produced by muscular effort and joint powers illustrate how these torques are used. Altering stride length is an example of how these joint variables are used to modify and control gait and therefore represent valid and insightful variables to distinguish between gait biomechanics in young and old adults. The purpose of this study was to identify the relationship between lower extremity joint torques and powers and stride length in old and young adults while walking at a velocity of 1.50 m/sec. This chapter is separated into the following sections: 1) Development of Hypothesis, 2) Discussion of Results, 3) Summary, 4) Conclusions, and 5) Future Recommendations.

Development of Hypothesis

Physiological and neurological adaptations are part of normal, healthy aging. Total functioning motor units (MUs) decrease by as much as 80% from the age of twenty to the age of ninety (39). Surviving MUs become larger and slower, due to fast neurons being replaced by slower neurons (4). Accelerated loss of fast MUs results in reductions in size and number of fast twitch muscle fibers (8; 38). Because fast twitch muscle fibers produce greater forces at a faster rate than slower muscle fibers, strength and power are lost with age (13).

Interestingly, these age-related adaptations have been shown to occur disproportionately throughout the body. Specifically, distal tissues are more susceptible to changes than proximal tissues. Allowing for reinnervation by slow twitch motoneurons, distal tissues lose more fast twitch muscle fibers than proximal tissues (35). Ultimately, this reinnervation reduces total number of motor units (4). Old versus young rats have shown less muscle in the distal soleus, while muscle mass in the adductor longus remained unchanged (2). This decrease in distal muscle mass leads to decreased strength in distal tissues (1).

Possibly due to these age-related physiological changes, gait mechanics are dissimilar between young and old adults. It has been suggested that old adults select more stable gait, causing heel-strike to occur more flat-footed and push-off to be less dynamic than that of young adults. However, limitations accompanying aging suggest old adults are forced to have less dynamic gait than young adults. These differences in kinematics are associated with reduced ankle power in old adults (40). Kerrigan et al. (17) reported that along with reduced ankle power, old individuals had greater hip flexor power than young when walking at self-selected speeds. When speed is controlled, old compared to young still rely more on hip function than knee or ankle function (10). Savelberg et al. (28) reported this redistribution after comparing young and old runners, as well as comparing young and old sedentary adults. Like physiological changes, redistribution of lower-extremity joint function appears to be part of normal aging.

Stride length, as a descriptive kinematic variable, has been used extensively in the literature to evaluate many populations including healthy young and old adults (12; 24) and pathologic patients (29; 41). Maximal stride length decreases with age and is exaggerated in balance-impaired adults (6; 19). Use of maximal stride length as an assessment of fall risk has been demonstrated to be as effective as other clinical assessments (6; 19). In addition to clinical assessments, maximal stride length is positively correlated to hip, knee, and ankle strength and power (19; 30). With evidence that old adults have lower plantarflexor strength than young adults (37), lower-extremity strength training improves stride length (25), and the results of this study may help clinicians to assess the influences of each lower-extremity joint on stride length and to prescribe adequate exercise regimens to improve gait characteristics.

In an attempt to identify the role of plantarflexor muscles in affecting stride length in young adults, Sutherland et al., (36) applied a tibial-nerve block to elicit acute lower-leg

paralysis. These investigators found that step length was reduced because subjects could not transfer weight to the anterior half of the foot on the blocked leg. Conclusions were made that plantarflexors primarily function to control forward progression of the body's mass during late-stance. Neptune et al., (23) pointed out that making this claim is difficult because applying a tibial-nerve block alters more than just plantarflexor function, resulting in numerous adaptations in gait mechanics.

Rather than directly altering the mechanics of specific muscles and joints, manipulation of stride length is an alternative method to identify the main contributors to stride length variation. This study may be the first to investigate stride length in this way. Since redistribution of lower-extremity joint function occurs naturally in old adults during normal gait, manipulation of stride length may elicit these same patterns. Namely it is possible that old adults used a different pattern of muscular and joint contributions to control stride length compared to young adults. It was hypothesized that old adults rely on hip joint torque and power more than knee or ankle torques and powers when manipulating stride length. It was also hypothesized that young adults rely on even distribution of lower-extremity joint torques and powers when manipulating stride length.

Discussion of Results

Walking Characteristics: Walking velocity has been shown to alter joint kinematics and kinetics (5; 17). To properly investigate the relationships between joint torques and powers and stride length, it was imperative to regulate gait velocity. In conjunction with previous literature (17; 28), the healthy adults in this investigation could adequately walk at the required 1.50 m/s. Preliminary screening potentially insured elimination of adults who may not have been able to walk at this velocity (see appendices C & D). Young adult mean velocity was 1.50 m/s while old

adult mean velocity was 1.53 m/s. Both groups walked with similar velocities and were well within the target range of 1.50 ± 0.08 m/s ($\pm 5\%$). Old adults confirmed this velocity to be comfortable and, in some instances, slower than normal walking velocity. Mean absolute stride length for young was 1.60 (0.10) while old stride length was 5% shorter at 1.52 (0.08) m. These lengths are representative of those reported by other investigators (10; 28; 31). Perhaps the tested velocity of the current study has resulted in these mean stride length values. For example, DeVita and Hortobagyi (10) reported stride lengths of 1.50 (0.08) m and 1.44 (0.08) m for young and old adults walking at the same nominal velocity. Walking at slower speeds has been shown to produce shorter strides in both young and old adults (15; 24). In this study, mean maximum stride length was 15% longer in young at 2.17 (0.19) m while old were at 1.89 (0.12) m. Reduced range of motion (18), loss of total muscular strength (33), and decreased plantarflexor functioning (1; 28) with age may lead to old adults' inability to produce strides as long as those observed in young adults. A greater range of stride lengths has a tendency to increase correlations with tested lower-extremity variables and therefore the higher correlations in young compared to adult adults may have been due to the larger range of stride lengths in the young group. However, it is considered that the longer range of stride lengths in fact produced the most accurate correlations for young adults. Therefore the comparisons between age groups were considered accurate and valid. The collected stride lengths and resulting ranges provide true assessments of differences between young and old adults.

Joint Torques and Powers: It is accepted that old adults, compared to young, rely more on proximal than distal musculature at both self-selected and controlled walking velocities (10; 15; 17; 20; 28; 31). With exception of Monaco et al. (20), these studies all reported reduced stride length with age. Few studies have investigated a relationship between joint kinetics and

stride length. Kerrigan et al. (17), while publishing reduced stride length and increased double support time with age, acknowledged the difficulty of correlation studies. Those investigators raised the question of whether reduced ankle power limited stride length, or whether reduced ankle power and strides were products of balance maintaining strategies. After subjects walked at different self-selected speeds, Judge et al. (15) reported 52% of stride length variance could be explained by ankle power where as hip extensor and flexor power explained 6% and 4%, respectively. The correlations in the present study provide support that stride length is controlled more so by distal vs. proximal muscle function.

Individual vs. Group Results: Figure 12 demonstrates how individual subject correlations often differed from the group correlations. It shows the old group correlation (red line) compared to two individual subject correlations (blue & green lines) between hip angular impulse and stride length. After observing the stronger correlations within each subject, especially in proximal joints (Table 3), it was apparent that group analyses did not identify the nature of the relationships among stride length and joint torques and powers within each individual. Especially in the old, subjects manipulated stride length with quite consistent patterns, but the group as a whole was rather inconsistent. These individual strategies used to manipulate stride, needed to be considered. Means of the individual subject correlations became the preferred method of investigating individual joint contributions to stride length manipulation.

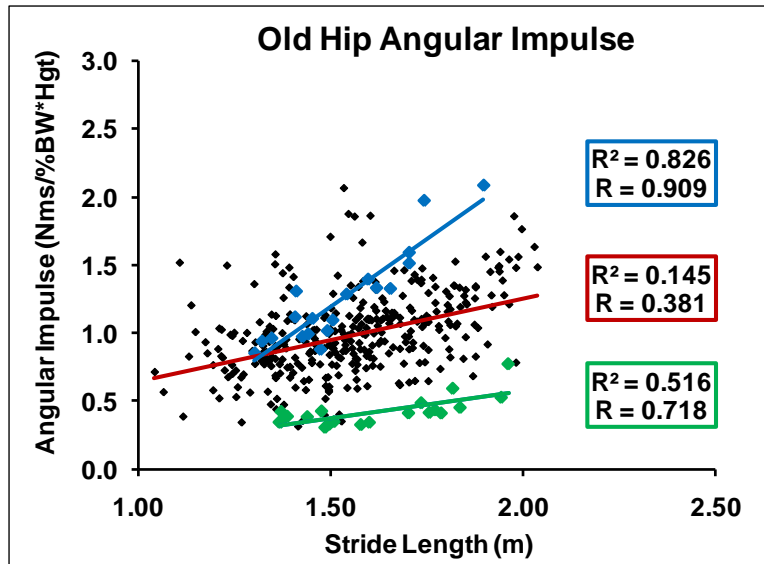


Figure 12: Group (red) and two single (blue & green) old subject relationships between hip angular impulse and stride length.

Using this type of individual correlation analysis resulted in higher relations for each variable. As groups, knee and ankle joints similarly contributed to stride length while the hip did so to a lesser degree. Old adults demonstrated less of a reliance on the hip than the young. Individual analysis improved hip correlations for both groups, but hip function was still more important in the young. With an r-value of 0.417, old hip work was least correlated with stride length. Ankle impulse and work for young ($r=0.860$ & $r=0.878$, respectively) and old ($r=0.796$ & $r=0.826$, respectively) were strong predictors of stride length. Despite flawed methods, Sutherland et al. (36) were correct that plantarflexors strongly influence stride length. Active, highly functional old adults used in this study may be a skewed representation of the entire old population, and the observed results should be limited to this subset of old adults. While old adults rely on proximal musculature to walk (10; 17; 28), this strategy does not seem to be used to control stride length, according to these correlation results. In young and old adults stride length appears to be driven by distal, rather than proximal, muscle function. Also, all young

correlations were higher than those of the old, demonstrating young having more control over stride length manipulation.

While correlations showed individual joint contributions, stepwise regression allowed for a comprehensive view of all lower-extremity joint variables. Stepwise regression is an analysis used to identify predictor variables and to also evaluate the order of importance of these variables for the entire young and old groups. After adding the most predictive variable, forward selection gradually adds one variable at a time until predictability no longer statistically increases (14). This type of analysis, similar to correlations, illustrated distal musculature to be the driving force behind stride length. Neither hip work nor impulse significantly added to the young regression and therefore was not added. In the old, hip impulse did significantly add to the regression, but was not entered until the third model. Despite low hip correlations, old adults coordinated hip function with knee and ankle function more than young adults. This suggests the old group may have used pelvic, trunk and upper extremities muscles more than the young group. Measuring these segments however was beyond the scope of this study. It is an understood limitation that stepwise regressions may have produced such strong results due to interdependent variables overestimating the importance of some joint variables. Uniformity of young adult results may suggest stride length manipulation requires a precise strategy that some old adults may not be able to execute. Due to kinetic variability in old adults, each subject may select individualized strategies to maintain normal functioning.

Summary

Because both groups of adults were highly functional, they were able to comfortably maintain the testing velocity of 1.50 m/s. Reduction of stride length with age was seen with old adults having shorter mean and maximal strides compared to young adults. When data were

analyzed for groups as a whole, correlations between each variable and stride length were lower compared to mean individual subject correlations. The mean individual subject correlations resulted in better predictions of how each group manipulated stride length. Knee and ankle torques and powers were more strongly correlated to stride length than hip torques and powers for both groups. These results were confirmed with the use of stepwise regression analyses. It appears that stride length is manipulated by distal muscle function more than proximal muscle function in healthy young and old adults.

Conclusion

It was hypothesized that old adults rely on hip joint torque and power more than knee or ankle torques and powers when manipulating stride length. It was also hypothesized that young adults rely on even distribution of lower-extremity joint torques and powers when manipulating stride length. Age related redistribution of torques and powers is well supported in current literature. However, this change in ambulatory strategy is not fully understood. This study attempted to relate age dependent redistributions of joint torques and powers to the fundamental characteristic of stride length.

In conclusion, the correlation results of this study do not support the hypothesis that old adults rely on hip joint torque and power more than knee or ankle torques and powers when manipulating stride length. However, group stepwise regression analysis may suggest a greater reliance on hip function in old compared to young adults. The results in the present study also refute the hypothesis that young adults rely on even distribution of lower-extremity joint torques and powers when manipulating stride length. On the contrary, stride length is more strongly correlated with knee and ankle torques and powers in both young and old adults. Finally, because all young subject correlations were similar in r-value and were stronger than old

correlations, young adults can apparently control stride almost equally with any joint and do so more accurately than old adults.

Future Recommendations

Possibly a source of unexpected results, strength levels were not measured for either of these groups. Highly functional old adults used in this study may not justly represent most old adults. Future studies could investigate old populations of all levels of functionality and pathologies. Because reductions in strength may influence stride length capacity, future studies could measure and correlate strength and stride length.

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APPENDIX A: INFORMED CONSENT

Consent Form
Mechanical Plasticity in Locomotion with Age

Investigator: Paul DeVita, Ph.D., Tibor Hortobágyi, Ph.D.
Address: 332 Sports Medicine Building
Biomechanics Laboratory
East Carolina University, Greenville, NC 27858
Telephone: (252) 737 - 4563, (252) 737 - 4564

I am being asked to voluntarily participate in this research project conducted by Paul DeVita, Ph.D. and Tibor Hortobágyi, Ph.D. The **purpose** is to examine the effects of exercise on muscle strength and mobility. Depending on my group assignment, the study involves up to 4 sessions of testing 1 hour each or these testing sessions plus supervised exercise training of the leg muscles 3 times per week for 10 weeks. There will be about 40 participants in this study.

I understand that my written consent is required before I can participate in this project. I may not participate in this project if: I had a falling accident in the past; I have had lower extremity surgeries or neurological conditions (i.e. stroke) or I have orthopedic conditions that substantially modify my walking pattern; I am afraid of ascending and descending stairs; I am on medication that causes dizziness; I am a smoker; I have a body mass/height ratio greater than 28; I have high blood pressure (140 systolic and 90 mm Hg diastolic) or I have a heart condition.

Procedures: I understand that I will be told which procedures apply to me:

1. Testing procedures: After 2-3 minutes of quiet sitting, my resting blood pressure and heart rate will be measured. In preparation, certain areas of the skin on my legs will be shaved and cleaned with alcohol. To determine the exact location of the muscle belly, water-soaked probe will be placed on the skin through which a very brief (100-millisecond) electrical stimulus will be applied to the skin. The stimulus may be applied 1-4 times per muscle site. Because the stimulation is very brief, I will feel no pain. EKG sensors will then be placed on 4 muscles of my right leg over the cleaned areas. These electrodes do not emit electricity but record muscle activity. The EKG electrodes will be connected to cables that lead to a small box secured on a waist-belt. As a warm-up for the subsequent tests, I will ride a stationary bicycle ergometer at a light resistance for about 5 minutes.

A. Position sense test. I will be seated on the seat of a computerized device. This is a precision test. For this experiment I will wear a blindfold. There will be no resistance applied to my leg throughout this test. I will be asked to move my leg from the starting position to a designated position, try to remember this position, return my leg to the starting position, then resume this position the best I can remember. After a few practice trials, I will be asked to repeat this test for 5 different positions using my right leg and my right ankle, respectively.

B. Steadiness test. This test will be done with eyes open. This is a matching test. I will be asked to extend my right knee and try to match a target force level that appears in front of me on a computer screen. The force I need to match will be just a few pounds. After some practice trials, I will perform 5 trials at a very low force and 5 trials at a somewhat greater force using my knee and my ankle, respectively.

C. Maximal leg strength test. This test will be done with eyes open. As a specific warm-up, I will perform 2 light, 2 medium, and 2 harder efforts. I will be tested for maximal knee extension and flexion and ankle extension and flexion strength of the right leg using a computerized strength-measuring device. Three repetitions of static effort (effort without actual movement) and 3 repetitions of dynamic efforts (effort with movement at the joint) with 1 minute of rest between conditions.

D. Walking test. Level walking: I will be asked to walk 10 yards as fast as I safely can. My movement will be video taped from the side. I will perform 5 trials. Then I will perform 5 trials at 1.5 m/s speed which is a comfortably slower speed. Walking on a Ramp: I will be asked to walk up a wooden ramp, stop at the end, turn around and walk down. The ramp is about 15 feet long and its slope is similar to a handicap ramp. Walking with ankle weights: I will be asked to walk with weights attached to my left or right ankle with Velcro. The weights are very light and correspond to only 5% or 10% of body weight.

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Mechanical Plasticity in Locomotion with Age
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E. Ascending and descending stairs. I will be asked to walk up at a self selected pace 4 steps, stop on the top of the stairwell, turn around, and descend. My movement will be video taped from the side. I will perform 5 trials.

F. Balance test. This test will be done under 7 conditions as listed below. A person will stand behind me to hold me, should I lose my balance. I will be asked to stand on a level platform that is even with the floor and try to maintain my balance for 30 seconds, 3 trials each of the following conditions:

1. Standing on the platform eyes open;
2. Standing on the platform eyes open head extended;
3. Standing on the platform eyes closed head extended;
4. Standing on the platform eyes open wearing the hat and head extended;
5. Standing on foam placed on the platform eyes open head extended;
6. Standing on foam placed on the platform eyes closed head extended, and
7. Standing on foam placed on the platform with hat on and head extended. The 'hat' refers to a dome that will block my view but lets light through.

G. Other mobility tests. The following tests will be administered by staff members who will rate my mobility for time and quality of movement. I will wear no electrodes and I will not be video taped.

1. Arising from a chair without an arm rest.
2. Immediate standing balance after arising from a chair.
3. Nudged on sternum 3 times at maximal standing height.
4. Pick up a pencil/pen from floor.
5. Turn 360°.
6. Arise from a chair and turn 360°.
7. Arise from a chair and walk 15 m in a carpeted hallway.
8. Ascend about 20 stairs.
9. Descend about 20 stairs.
10. Reaching for maximal length without losing balance or making a step.

H. Maximal leg press strength. In supine position, the weight I can move with my legs one time, will be measured in a leg press weight lifting machine. I will perform warm up trials at very light and medium loads to reach my maximum gradually.

I. Step length test. I will walk across the level walkway about 25 times using steps ranging from short to long in length. My movements will be recorded with a camera system and my muscles will be measured with electrodes placed on four muscles in the right leg.

2. Training procedures: If I am selected to be in the exercise training group, I will report to the Biomechanics Laboratory 3 times per week for 10 weeks to undergo a supervised weight lifting exercise program with an emphasis on leg strength. I will exercise by performing 4 to 6 bouts of 10 to 15 repetitions at 40-50% of my maximal weight or by performing 4 to 6 bouts 5 to 10 repetitions at 80-90% of my maximal weight. As I improve, the weights will be gradually adjusted every week. I will exercise my thigh and hip muscles and my calf muscles. A 'bout' refers to a unit of exercise during which I perform a given number of repetition. There will be 2 minutes of rest between each exercise bout. My blood pressure will be monitored before, during, and after exercise. Optional upper body exercises will be available but not required.

Risks: Because the forces produced during the Position and Steadiness tests are very low, there are minimal risks associated with these tests. In contrast, 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 tests are the Maximal leg strength and Maximal leg press tests. 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 elders in terms of falling. The Balance tests represent some risk of falling when my eyes are closed or wearing the vision-blocking hat. The Mobility tests represent some risks of losing balance, falling, rapid rise in blood pressure, and dizziness. The Training procedures represent risks of elevated blood pressure, dizziness, shortness of breath and my legs may become tired.

All these risks will be reduced by: allowing subjects to participate who have been previously cleared by their physicians; carefully screening for the various risk factors prior to testing; monitoring blood pressure and heart rate; having a spotter monitor the tests that are associated with risks for falling; carefully explaining and demonstrating the tasks, and by having subjects properly warmed up for and thoroughly familiarized with the tests.

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Benefits: All results will be explained to me. I will be entitled to \$50 if my assignment is a non-exercising control subjects. In this case I will be asked to perform the tests twice about 8-10 weeks apart. If my assignment places me in the exercise group, I will be entitled to \$150. To receive this benefit, I will be asked to perform the tests twice, once before and once after the exercise training program. 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 experiment is voluntary and refusal to participate will involve no penalty or loss of benefits to which I am otherwise entitled, and 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. The video tape footages will be seen only by the researchers involved in data analysis. The video tapes will be destroyed after all data had been extracted.

Contact person. If I have any questions about the research or possible research-related injury, I may contact Dr. DeVita at home ([252] 756 – 8070) or at work ([252] 737 – 4563) or 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 Policy and Review Committee on Human Research ([252] 816-2914).

I have read the above material and it has been explained to me by Dr. DeVita or Dr. Hortobágyi. I have been encouraged to ask questions about the study and all inquiries have been answered to my satisfaction. A copy of this consent form shall be given to the person signing as the subject or as the subjects authorized representative.

Patient's name (Print)

_____ Date _____
Patient's signature

Auditor witness: I confirm that the contents of this consent form were orally presented.

Auditor's name (Print)

_____ Date _____
Auditor's witness signature

Principal investigator's name (Print)

_____ Date _____
Principal investigator's signature

Family physician's name (Print)

_____ Date _____
Family physician's signature

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TO 7.14.10

APPENDIX B: INSTITUTIONAL REVIEW BOARD APPROVAL



University and Medical Center Institutional Review Board
East Carolina University • Brody School of Medicine
600 Moye Boulevard • Old Health Sciences Library, Room 11L-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: Paul DeVita, PhD, Dept of EXSS, ECU—332 Ward Sports Medicine Building
FROM: UMCIRB *KL*
DATE: July 15, 2009
RE: Expedited Continuing Review of a Research Study
TITLE: "Mechanical Plasticity in Locomotion with Age"

UMCIRB #98-044

The above referenced research study was initially reviewed and approved by the convened UMCIRB on 9.1.1998. This research study has undergone a subsequent continuing review using expedited review on 7.15.09. 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. It is also a research involving materials (data, documents, records, or specimens) that have been collected, or will be collected solely for nonresearch purposes (such as medical treatment or diagnosis). (NOTE: Some research in this category may be exempt from the HHS regulations for the protection of human subjects. 45 CFR 46.101(b)(4). This listing refers only to research that is not exempt.) The Chairperson (or designee) deemed this NIH/ECU grant funded 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 7.15.09 to 7.14.10. The approval includes the following items:

- Continuing Review Form (dated 7.9.09)
- Informed Consent (dated 11.11.04) (received 7.10.09)

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.

APPENDIX C: SUBJECT TELEPHONE QUESTIONNAIRE

*Telephone interview for general medical and mobility status
Mechanical plasticity in locomotion with age (DeVita, PI)*

Demographic data:

Date _____
Name _____ Phone number _____
Address _____
Birth date _____ Age _____

Height (ft/in) _____ Height (m) _____
Weight (lbs) _____ Mass (kg) _____
BMI (kg/m²) _____

Do you smoke? Yes _____ No _____
Have you smoked in the past? Yes _____ No _____
If yes, when did you stop smoking _____

Functional ability in ADLs:

How much difficulty do you have when you

Walk on level surface	None	Some	A lot
Walk up or down a ramp	None	Some	A lot
Climb stairs	None	Some	A lot
Descend stairs	None	Some	A lot

How much pain do you have in your knee or hip joints when you

Walk on level surface	None	Some	A lot
Walk up or down a ramp	None	Some	A lot
Walk up and down a ramp	None	Some	A lot
Ascend and descend stairs	None	Some	A lot

Can you do the following activities independently:

Dress	Yes _____	No _____
Bath	Yes _____	No _____
Continence	Yes _____	No _____
Eating	Yes _____	No _____

Do you use a walker or cane when walking? Yes _____ No _____

During the past year, did you fall down more than once while walking or climbing stairs?

Yes _____ No _____

Medical:

Do you have any neurological problems such as stroke or Parkinson's disease? Yes _____
No _____

Do you have any problems with your heart such as atrial fibrillation, pace maker, coronary artery disease, or congestive heart failure? Yes _____ No _____

Do you have any musculoskeletal problems such as arthritis, joint replacement, or other orthopaedic problems? Yes _____ No _____

Do you have any pulmonary disease such as difficulty in breathing or emphysema?
Yes _____ No _____

Do you have any peripheral artery disease? Yes _____ No _____

Do you have high blood pressure (>160/90 mm Hg)? Yes _____ No _____

Medical Information for Dr. Steinweg:

Do you take medication to control your blood pressure? Yes _____ No _____

List the medications you are currently taking

Do you have any loss of vision? Yes _____ No _____

If yes, do you have eye glasses or contact lenses that correct your vision? Yes _____
No _____

Do you have any other medical problems we did not talk about? Yes _____ No _____

If, "Yes," what is or are the conditions? _____

List any surgeries you have had.

APPENDIX D: INCLUSION & EXCLUSION CRITERIA

Inclusion and Exclusion Criteria Subjects will be Included if they:

Are of the designated age criteria
Are healthy as determined by our criteria
Can perform level, stair and ramp walking with no difficulty
Can participate in a strength testing protocol
Have a BMI $\leq 30.0 \text{ kg/m}^2$
Provide written informed consent

Subjects will be Excluded if they have any of the following:

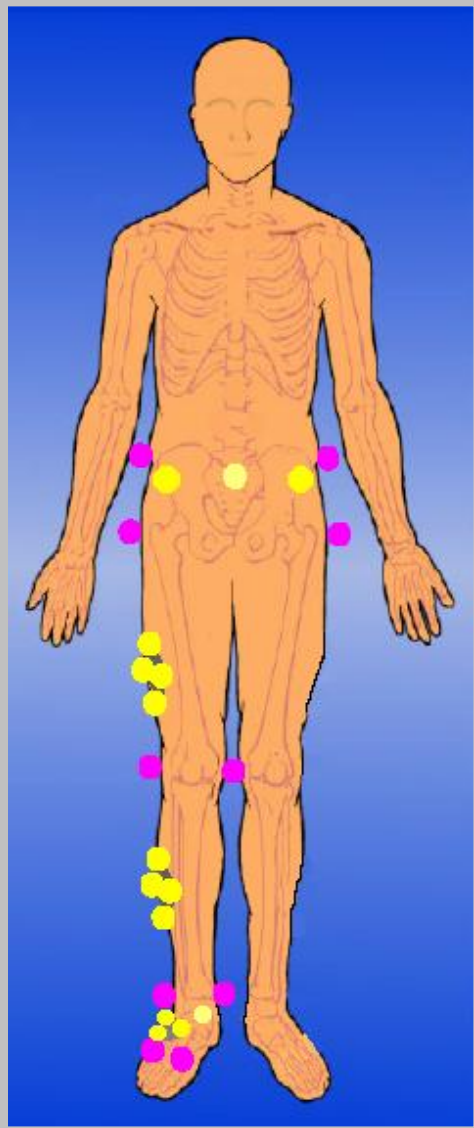
Active coronary artery disease (CAD) with symptoms BMI $> 30.0 \text{ kg/m}^2$
Cancer (with a past or present history of chemotherapy or radiation therapy)
Congestive Heart Failure (CHF)
Chronic Obstructed Pulmonary Disease (COPD)
Dementia
Diabetes
Heart rhythm other than sinus rhythm
High blood pressure ($> 160/100 \text{ mm Hg}$)
History of current anemia with Hgb below 11.0
History of multiple falls within the last year
History of renal failure or renal insufficiency with creatinine above 2.0
History of spinal surgery
Hypertension treated with beta blockers or calcium channel blockers with a resting heart rate $< 66 \text{ bpm}$
Joint replacement in the lower extremity

Joint surgery on the lower extremity
Osteoarthritis limiting movement capabilities
Osteoporosis with vertebral or hip fracture
Pain in the lower extremities from unknown cause
Parkinson's disease or history of stroke
Peripheral neuropathy
Peripheral vascular disease (PVD)
Presence of a pacemaker
Rheumatoid arthritis
Use of ambulatory walking aid (cane, etc)
Visual impairment that restricts independent ambulation

Or if they:

Presently smoke cigarettes
Quit smoking but still have smoking related health problems
Cannot perform the walking and strength tests

APPENDIX E: DIAGRAM OF REFLECTIVE MARKER PLACEMENT



Calibration Markers Location

- Right and left iliac crest
- Right and left greater trochanter
- Medial and lateral knee joint line
- Medial and lateral malleoli
- 1st Metatarsal head
- 5th Metatarsal head

Tracking Markers

- Right and left anterior superior iliac spine
- 5th Lumbar vertebrae/1st sacral vertebrae
- Thigh plate
- Shank plate
- Foot plate
- Heel