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EFFECTS OF AGE AND SPEED ON COACTIVATION, VARIABILITY, AND JOINT KINETICS DURING WALKING

A Thesis in

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ABSTRACT

Metabolic cost of walking (C_w), defined as the rate of oxygen consumed during walking per unit distance traveled, has been shown to be greater for the elderly than for young adults across a range of walking speeds (Ortega & Farley, 2007; Mian et al., 2006). Several potential causes of increased C_w in older adults have been highlighted in current literature, including elevated antagonistic muscle coactivation, a proximal redistribution of lower limb work, and increased step width variability. The purpose of this study was to investigate age and speed effects on these three variables as well as C_w, and to quantify the relationship between these variables and C_w. Joint kinematics, joint kinetics, lower extremity electromyography, and C_w were measured for fourteen physically active older adults, and fourteen active young adults at four walking speeds (0.89, 1.12, 1.34, and 1.57 $m \cdot s^{-1}$). Older participants exhibited 23% higher C_w than young adults when averaged across walking speeds. Coactivation was significantly higher about the thigh but not the shank in older adults across all speeds tested. Also, a proximal redistribution of joint work was noted in older adults, as this group exhibited less knee work and more hip work than young. Ankle work, however, was similar between age groups. Step width, step width variability, stride length, and stride length variability were all similar for the two age groups. C_w was positively correlated with coactivation only. Due to the elevated coactivation in older adults, as well as the positive correlations exhibited between this variable and C_w, it seems as if coactivation may be a factor in the increased C_w seen in older adults.

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INTRODUCTION

Metabolic cost of walking (C_w), defined as the rate of oxygen consumed during walking per unit distance traveled, is higher for the elderly than for young adults across a range of walking speeds (Ortega & Farley, 2007; Malatesta, Simar, Dauvilliers, Candau, Borrani, Prefaut, & Caillaud, 2003; Martin, Rothstein, & Larish, 1992; Mian, Thom, Ardigo, Narici, & Minetti, 2006; McCann & Adams, 2002). In addition, age is associated with a reduction in maximum metabolic and force producing capacities. These two factors lead to many older adults completing activities of daily living at a higher percentage of their maximum ability than young, thereby increasing the likelihood of fatigue during these tasks (Hortobagyi, Mizelle, Beam & DeVita, 2000). Though it has been shown that age associated reductions in metabolic and force producing capacities may be slowed through training (for reviews see Kell, Bell, & Quinney, 2001; Faulkner, Larkin, Claflin, & Brooks, 2007), there are no definitive explanations regarding the causes of higher C_w in older adults. Nevertheless, a number of changes associated with age have been identified as potential causes of this phenomenon, including higher step variability (Owings & Grabiner, 2004a), a distal-to-proximal redistribution of lower limb work during walking (DeVita & Hortobagyi, 2000), and higher coactivation of agonist and antagonistic muscles (Mian et al., 2006).

Step Variability

Step width variability, described as deviations from mean step width during a walking bout, has been shown to be higher for older adults with respect to younger adults (Grabiner, Biswas, & Grabiner, 2001; Stolze, Friedrich, Steinauer, & Vieregge, 2000; Owings & Grabiner, 2004b). In addition, several recent studies have indicated that changes in step width variability may have an effect on C_w (Dean, Alexander, & Kuo, 2007; Donelan, Kram, & Kuo, 2001; Donelan, Shipman, Kram, & Kuo, 2004). Donelan et al. (2001) experimentally manipulated step width above and below the average preferred step width during walking, causing subjects to experience elevated metabolic expenditure. In a follow up study, Donelan et al. (2004) laterally stabilized subjects during gait, which resulted in reductions in both step width variability and metabolic expenditure with respect to normal walking. It is possible, therefore, that normal deviations from preferred step width during gait, quantified by step width variability, could be a factor that influences C_w.

To our knowledge, only one investigation (Malatesta et al., 2003) has directly examined the relationship between variability of multiple descriptors of step mechanics and C_w. Malatesta and colleagues determined stride time variability and C_w in 25, 65, and 80 year olds to determine whether step time variability was a good predictor of C_w. In this investigation, stride time variability was defined as the coefficient of variation of all step times collected during a 6 minute treadmill bout. Averaged across five walking speeds, stride time variability was significantly larger in the 80 and 65 year old groups compared to the 25 year old group. Average C_w was 22% higher for the 80 year old group with respect to 25 year olds when averaged across all five speeds, and was significantly larger in the 80 year old group than young at each walking speed. The C_w of the 65 year old group was also significantly larger than the 25 year old group, but only at the two highest speeds examined (~14% higher at 1.33 $m \cdot s^{-1}$ and ~18% higher at 1.57 $m \cdot s^{-1}$). Interestingly, when stride time variability was correlated with C_w for all subjects at their preferred walking speed, no significant relationship was shown (r=0.19). Malatesta and colleagues speculated higher stride time variability requires more stabilization activity from musculature, but suggested the metabolic cost associated with stabilization effort may be negligible when compared to the muscular actions associated with propulsion and support.

More recently, Owings and Grabiner (2004a) suggested step width variability is a better discriminator of old and young walkers than step time variability. They measured kinematics of old and young individuals while walking at a preferred speed. Specifically, they measured step width, length, and time, as well as the variability of each measure. Step width variability was significantly larger in the older adults with respect to young (approximately 20% larger), whereas step length and step time variabilities were not significantly different between groups. Although previous studies have indicated a possible connection between step width variability and $C_{w,}$ to our knowledge no studies have looked directly at the relationship between these variables.

Joint Moment Redistribution

Joint kinetics, particularly net joint moments and the mechanical work and power associated with them, have often been used as indirect indicators of muscular effort about individual joints (DeVita & Hortobagyi, 2000; Graf, Judge, Ounpuu, & Thelen, 2005; Kerrigan, Todd, Della Croce, Lipsitz, & Collins 1998; Winter, Patla, Frank, & Walt, 1990). Assessments of joint kinetics have shown older adults exhibit different joint moment profiles and work distribution than young. Specifically, the distribution of lower body joint moments during walking shifts proximally with age, such that more proximal moments account for a greater proportion of the total work or power generated in the lower extremities while more distal moments produce a smaller proportion of the total power of the lower limb (DeVita & Hortobagyi, 2000; Savelberg, Verdijk, Willems, & Meijer K, 2007; Silder, Heiderscheit, & Thelen, 2008). DeVita and Hortobagyi (2000) showed that when healthy older adults walked at

1.48 $m \cdot s^{-1}$, they did substantially more (279%) muscular work about the hip, but less work about the knee and ankle (39 and 29%, respectively) than young adults walking at the same speed. Savelberg et al. (2007) further substantiated these findings by measuring joint moments of old and young runners and non-runners at similar walking speeds. Though older adults in this study walked at a slower velocity with respect to young, they were shown to have smaller knee extensor and plantar flexor impulses and higher hip extensor impulses than young, regardless of physical activity status. While these changes could be related to higher $C_{w,}$ to our knowledge the relationship between the distribution of joint kinetics and C_w has not been examined.

Coactivation of Antagonistic Musculature

Coactivation, defined as concurrent excitation of antagonistic muscles, elicits notable beneficial effects, including increased stability of joints and a more even distribution of loads within the joint (Baratta, Solomonow, Zhou, Letson, Chuinard, & D'Ambrosia, 1988; Hortobagyi & DeVita, 2000; Seidler-Dobrin, He, & Stelmach, 1998). Despite these advantages, coactivation may also result in excessive muscular excitation for a resulting net joint moment, thereby contributing to unnecessary energy expenditure.

The effect of age on coactivation has been studied by a number of investigators. Results have been equivocal, in part because of differences in the movement tasks and musculature being considered. Several studies have contrasted antagonistic coactivation displayed by young and older adults during controlled isokinetic and isometric movements of the lower extremity (e.g., Izquierdo, Ibanez, Gorostiaga, Garrues, Zuniga, Anton, Larrion, & Hakkinen, 1999; Tracy & Enoka, 2002; Macaluso, Nimmo, Foster, Cockburn, McMillan, & DeVito, 2002; Simoneau, Martin, & Hoecke, 2005; Klass, Baudry, & Duchateau, 2005). Results of these studies showed age effects on coactivation differ between tasks. For example, active older adults exhibited similar amounts of antagonistic muscular activity, expressed as a percent maximal excitation, as young adults during maximal knee flexion (Macaluso et al., 2002; Izquierdo et al., 1999). Interestingly, during isometric and isokinetic knee extension, older adults had higher excitation of knee flexors (Izquierdo et al., 1999; Macaluso et al., 2002; Tracy & Enoka, 2002). Age-related changes in coactivation about the ankle were also inconsistent across studies. During maximal isometric plantar flexion, Simoneau et al. (2005) found lower levels of coactivation in older adults, whereas Morse et al. (2005) found similar coactivation between young and old individuals. Coactivation during dorsiflexion, however, has consistently been shown to be similar between young and old (Simoneau et al., 2005; Klass et al., 2004).

The effects of age on coactivation of antagonistic musculature during multi-joint dynamic motions are more consistent. Older adults typically have larger coactivation levels than young, especially about the thigh. Hortobagyi, Mizelle, Beam, and DeVita (2003) reported older adults had 60% higher coactivation levels and 80% higher overall muscular excitation than young adults about the thigh during downward stepping tasks. Hortobagyi and DeVita (2000) also showed old individuals had higher excitation of knee flexors and ankle dorsiflexors during downward stepping. More recently, Larsen, Puggaard, Hamalainen, and Aagaard (2008) quantified coactivation about the thigh and shank while ascending and descending stairs at preferred, controlled (35 steps per minute), and maximal speeds. In their analysis, the step cycle was broken into loading, stance and unloading phases. While ascending and descending stairs at all speeds, old showed approximately 20% higher coactivation about the thigh during the stance portion of the step cycle. Interestingly, coactivation about the shank showed quite different results. Shank coactivation was 43% lower in old during the unloading phase while ascending stairs at a standardized speed. When ascending stairs at a freely chosen speed, older adults showed 12% less coactivation about the shank than young, though they also ascended at a slower speed (Old: 50.4 steps per minute; Young: 60.3 steps per minute). At maximal ascending speed, as well as all descending speeds, shank coactivation levels were similar for the older and young adults. Overall, Larsen and colleagues reported that "Generally, no differences in Calf coactivation were observed between age groups" (p.11) when ascending or descending stairs.

To our knowledge, only one study has examined the effect of age on coactivation during level walking. In a broad investigation of walking mechanics and energy cost, Mian et al. (2006) hypothesized that antagonist muscle coactivation is higher in older adults and is thereby a factor contributing to a higher cost of walking in older adults compared to young adults. Pooling data across speeds, Mian et al. showed that both the cost of walking and coactivation about the thigh were 31% higher for the older adults with respect to young. In addition, authors found that age moderated the effects of speed on coactivation. For the two lower speeds of walking (0.83 and 1.11 $m \cdot s^{-1}$), the level of coactivation was similar for old and young subjects and was not affected by walking speed. Both old and young subjects showed significantly higher levels of coactivation at the two higher walking speeds (1.39 and 1.67 $m \cdot s^{-1}$), but coactivation levels were greater for the older subjects. This age effect was most pronounced at the fastest walking speed. Mian et al. subsequently pooled data for young and old subjects and examined the relationship between cost of walking and coactivation index at each walking speed. They found modest positive correlations ($r = 0.28 – 0.52$) indicating a higher C_w was associated with higher levels of coactivation. When these relationships were computed for just the older subjects, a significant linear relationship was found at just one walking speed (1.39 $m \cdot s^{-1}$, $r=0.52$). Mian et al. speculated that the higher coactivation levels observed about the thigh in older adults may be "a compensatory mechanism to increase joint stiffness and thereby enhance stability" (p.135). The higher coactivation would result in a larger metabolic cost with no change in mechanical work.

Coactivation was quantified by Mian et al. (2006) by measuring the time in which antagonistic thigh muscles were active. Though this method of quantification provides one indication of coactivation occurring about the segment, the magnitude of excitation of antagonists is not considered in this approach. An alternative and perhaps improved expression of coactivation quantifies the area of overlap of excitation profiles of two concurrently active antagonistic muscles. This method considers both the time of overlap and the magnitude of muscular excitation. Also, Mian and colleagues limited their analysis to coactivation about the thigh. As previously noted, Larsen et al. (2008) showed age effects on coactivation during stair stepping differed between the thigh and shank. A multi-segment assessment of coactivation in the lower extremity has not been conducted for overground walking.

Purpose and Hypotheses

The overall purpose of this study was to examine several potential mechanisms underlying higher metabolic costs of walking in older adults. Three age-related changes in walking mechanics - step variability, proximal joint moment redistribution, and antagonist muscle coactivation - have been identified in previous research efforts as potential contributors to the higher metabolic cost in old. Interestingly, few studies have looked specifically at the relationship between these variables and C_w. The purpose of this study, therefore, was to determine the effects of age and walking speed on step width variability, joint kinetics, and coactivation, and to quantify the relationship between these variables and C_w. Four hypotheses were assessed:

1. Healthy active older adults have a higher net C_w than young adults across a range of walking speeds.

2. Healthy active older adults have higher levels of antagonist muscle coactivation about the thigh and shank than young adults across a range of walking speeds.

3. Healthy active older adults exhibit a proximal redistribution of net joint moments and the work attributed to them, generating higher hip extensor moments and work but lower ankle plantar flexor and knee extensor moments and work than young adults across a range of walking speeds.

4. Healthy active older adults have higher step width variability than young adults across a range of walking speeds.

METHODS

Participants

Fourteen young and fourteen individuals were recruited to participate in this study (Table 1). Young adults were limited to an age range of 20-30 years, whereas older adults were limited to 65-80 years of age. These age ranges were chosen because healthy individuals 65 years and older have been shown to have differences in walking kinematics (Owings & Grabiner, 2004a; Grabiner et al., 2001), antagonist muscle coactivation (Hortobagyi & DeVita, 2000, Larsen et al., 2008; Mian et al., 2006), and economy (Malatesta et al., 2003; Martin et al., 1992; Mian et al., 2006) with respect to young adults. A concerted effort was made to match the height, body mass, body mass index, and leg length characteristics of the two age groups. These efforts were generally successful, although the older participants had modestly higher BMI values than the young subjects (Table 1). Medical history and activity levels were assessed by a questionnaire administered over the phone to potential participants. All participants chosen were physically active. Ten of the 14 older subjects regularly exercised four or more times per week at moderate to high intensity, with activities usually consisting of walking, tennis, squash, and resistance training. The four remaining older participants were vocationally active, with daily activity consisting of walking up and down stairs and hiking. Younger adults were also physically active, participating in several bouts of exercise weekly. Generally, exercise in the younger group consisted of a mix of weight and aerobic training. All participants were free of lower limb injuries to bones, joints, and musculature and had no serious lower limb operations or injuries in the previous six months. No participants had any known neurological disorders. A physician reviewed each subject's medical history to ensure they were healthy enough for participation. We specifically chose to recruit healthy, active older adults in an effort to examine effects of age, rather than pathology, on our dependant variables. In addition, physically active participants were chosen to minimize the effects of fatigue during the walking trials.

Table 1: Subject characteristics

	Old (n=14)		Young (n=14)	
	Male (n=6)	Female (n=8)	Male (n=6)	Female (n=8)
Age (yrs)	72.7 (5.3)	69.9 (2.3)	24.7 (2.9)	26.4 (2.8)
Mass (kg)	70.1 (10.3)	60.4 (7.5)	70.9 (7.2)	54.9 (4.7)
Height (m)	1.64 (0.07)	1.62 (0.07)	1.74 (0.05)	1.62 (0.04)
BMI ($kg \cdot m^{-2}$) *	26.1 (4.1)	23.0 (1.8)	23.5 (1.8)	21.0 (1.4)
Leg Length (m)	0.88 (0.29)	0.86 (0.05)	0.89 (0.04)	0.85 (0.04)
Activity Level *	5.0 (0.8)	4.7 (1.5)	3.4 (0.9)	3.0 (1.5)

There were no statistically significant differences between older and younger participants for body mass, height, and leg length. Older participants, however, show modestly higher levels of physical activity engagement (number of times subjects participated in 30 minutes of moderate to high intensity activity per week; $F_{1,21}$=9.49, p=0.006), and BMI ($F_{1,24}$=6.23, p=0.02) with respect to young. Significant age effects noted by (*).

Data Collection

Participants completed two sessions. In the first session, overground walking trials were collected at each of four walking speeds. These data formed the basis of a lower extremity inverse dynamics analysis. To conclude session 1, participants were familiarized to treadmill walking. In session two, participants walked on a treadmill at the same four speeds used in the overground trials. Data were collected for the calculation of muscle coactivation, step kinematics, and metabolic cost of walking.

Session 1.

Participants wore spandex bike shorts, t-shirt, and athletic shoes during the first session. Initially, participants were familiarized with the experimental protocol, testing procedures, and testing equipment before providing informed consent. In addition, subject screening information that was obtained during the telephone screening process was reviewed and confirmed. Body mass, height, and several anthropometric characteristics of the left leg and foot were measured using a standard balance scale, anthropometer, and tape measure. Reflective markers were then placed over estimates of joint centers of each subject's left ankle, knee, and hip, as well as on the lateral aspect of the fifth metatarsal head.

Participants then proceeded to the gait laboratory to complete overground walking trials at four walking speeds: 0.89, 1.12, 1.34, and 1.57 $m \cdot s^{-1}$. Speed conditions were randomly ordered for each subject. For each walking speed, participants first completed a number of practice trials so that they were able to match the targeted walking speed while contacting a force platform (Kistler Instrument Corporation, Amherst, NY) in the center of the walkway. A photocell-based timing system was used to measure the time required to traverse a 4.6 m distance in the middle of the walkway and to compute the average walking speed for each trial. Once comfortable with the speed condition, the subject walked through the gait lab while marker position and ground reaction force (GRF) data were sampled using the Motion Analysis system (Motion Analysis Corporation, Santa Rosa, CA). Calibration of the Motion Analysis system included static and dynamic components to establish the spatial reference frame and motion capture volume. Marker position and GRF data were collected synchronously at 100 and 1000Hz, respectively, for five acceptable trials. Acceptable trials were those in which the subject's average walking speed were within 3% of the target speed and the left foot contacted the force platform with no visual evidence that the subject adapted his or her stride pattern to ensure force platform contact.

After completion of the overground walking trials, reflective markers were removed and the subject was relocated to the treadmill area for treadmill walking accommodation. Previous studies have required between 10 and 30 minutes of treadmill walking prior to data collection to ensure full accommodation (Malatesta et al., 2003; Martin et al., 1992; Mian et al., 2006; Ortega

& Farley, 2006). Therefore, all participants in the current study were required to walk on a motor-driven treadmill (Quinton Instrumentation, Bothell, WA, USA) for 30 minutes during the initial test session to ensure they were well accommodated to treadmill walking prior to the treadmill-based data collection in session two. During this habituation period, participants started walking slowly with hands on rails. As they became more comfortable, hand rail support was removed, and participants increased walking speed. Participants practiced walking at all speeds used in the experimental protocol. A short rest period of approximately 5 minutes concluded the treadmill accommodation procedure. Participants then returned to the walkway for determination of preferred overground walking speed. Participants were asked to walk naturally and comfortably along a 10-meter walkway. The time required to traverse a 3 meter section in the middle of the walkway was recorded and converted to average speed. Preferred walking speed was defined as the average speed of 10 trials.

Session 2.

Participants returned to the laboratory 1 to 5 days following test session one. Upon arrival, electrodes used to capture surface electromyography (EMG) data were placed on subject's right leg. The area of skin where electrodes were attached was prepared by shaving, abrading, and cleaning the area with isopropyl alcohol to reduce surface impedance. Two bipolar silver-silver chloride electrodes (Vermed, Bellows Falls, VT, USA) were positioned approximately 2 cm apart over the tibialis anterior, the lateral aspect of the soleus, medial gastrocnemius, vastus medialis, biceps femoris, and semitendinosus muscles. Electrode placement was determined by identifying an area along the long axis of each muscle just distal of the center of the muscle belly. A common reference electrode was placed over the fibular head. EMG data were collected using bipolar silver-silver chloride electrodes. Preamplification with a fixed gain of 500 was provided 10 cm from the electrode attachment site. Amplifier gain was selectable from 500 to 15000 with a bandwidth of 10 to 1000 Hz. The common mode rejection ratio was 115dB at 60 Hz and the input impedance was approximately 10 GOhms. A custom program created in Labview (National Instruments Corporation, Austin, TX, USA) was used to sample EMG data at 1000 Hz.

After electrode attachment, the subject's resting $\dot{V}O_2$ was determined. All $\dot{V}O_2$ measurements were collected using a computerized metabolic cart (Parvo-Medics, Sandy, Utah, USA). The system was calibrated at the beginning of each session using standard calibration gases. Ventilation measurements were calibrated using a 3-liter syringe. Metabolic data were collected continuously and reported every breath. To determine resting $\dot{V}O_2$, each subject first sat quietly for five minutes. The subject then stood quietly for five minutes while metabolic measures were collected (Martin et al., 1992). Resting $\dot{V}O_2$ was computed as the average value over the last three minutes of the 5-minute collection.

After the determination of resting $\dot{V}O_2$, reflective markers were placed on the heel and third metatarsal head of both left and right feet. In addition, a marker cluster was placed at the level of the fifth lumbar vertebra. Participants were then asked to walk on a motorized treadmill for seven minutes at each of the four speeds used in the initial overground session. The order of speeds for each subject was the same as that used in session one for overground walking trials. All participants were encouraged to walk without using handrail support. Six older participants, however, still found it necessary to rest one hand lightly on a side handrail. Between each walking bout participants rested for a minimum of five minutes. Additional rest was given as necessary. During each walking trial, EMG, $\dot{V}O_2$, and marker position data were collected. Treadmill belt speed was checked at the beginning of each trial by timing 10 belt revolutions with a handheld stopwatch and was adjusted as necessary to match the nominal speed condition. After 2 minutes of walking, a 5-minute Motion Analysis capture began. Between minutes 2 and 5 of this capture, EMG data were also collected for 30 seconds. Metabolic data were collected for the entirety of the trial. Each subject's respiratory exchange ratio (RER) was continuously monitored to ensure RER remained less than 1.0 throughout the trials, thereby indicating metabolism was primarily in the aerobic pathway. After completion of the four treadmill bouts, participants sat with the right leg comfortably extended, supported, and in a relaxed state while five seconds of resting EMG data were collected. Data were visually inspected to ensure each muscle was inactive during this collection.

Data Analysis

Metabolic Cost of Walking (C_w).

$\dot{V}O_2$ and $\dot{V}CO_2$ (ml·min^{-1}) were collected for each expired breath and then averaged over the final 3 minutes of each metabolic collection. An approximation of the rate of energy cost (J·s^{-1}) was calculated using both $\dot{V}O_2$ and $\dot{V}CO_2$ (Brockway, 1987), and subsequently normalized to body mass. Resting metabolic rate was subtracted from metabolic rate during each walking trial to determine net rate of energy cost. To determine the net cost of walking (C_w), the net metabolic rate (J·kg^{-1}·s^{-1}) at each speed was divided by the walking speed (m·s^{-1}), which resulted in an expression of energy cost per unit distance traveled (J·kg^{-1}·m^{-1}).

Gait Kinematics Analyses.

During treadmill trials, selected step kinematics were determined by analysis of heel and toe markers of both feet during the first 400 consecutive steps. First, marker position data were low pass filtered at 2 Hz. Linear velocity, acceleration, and jerk were then calculated for these markers by differentiation using finite difference equations. Heel strike and toe off events were determined by a method described by Hreljac and Marshall (2000). Toe off was defined as the instant in time when the toe marker showed a local maximum in the horizontal component of acceleration (jerk equal to zero). Heel strike was defined as the instant when the heel marker has a local maximum for vertical acceleration. For step width determination, the instantaneous mediolateral position of each foot was defined as the average position of the heel and toe markers. Foot position was then averaged over the period of time when the contralateral leg was in its swing phase. Step width was computed as the mediolateral distance between right and left average foot positions during consecutive stance phases. Step width variability was subsequently quantified as the standard deviation of the first 400 consecutive step width values.

Although our principle focus was on step width variability, previous literature has shown other factors as potential contributors to C_w (Donelan, Kram, & Kuo, 2002). Therefore, several other measures- step width, stride length, and stride length variability- were also analyzed for speed and age effects. Stride length was determined by multiplying walking velocity by the time between consecutive heel strikes for the same foot (Figure 1). Stride length variability was defined as the standard deviation of the first 400 consecutive stride length values.

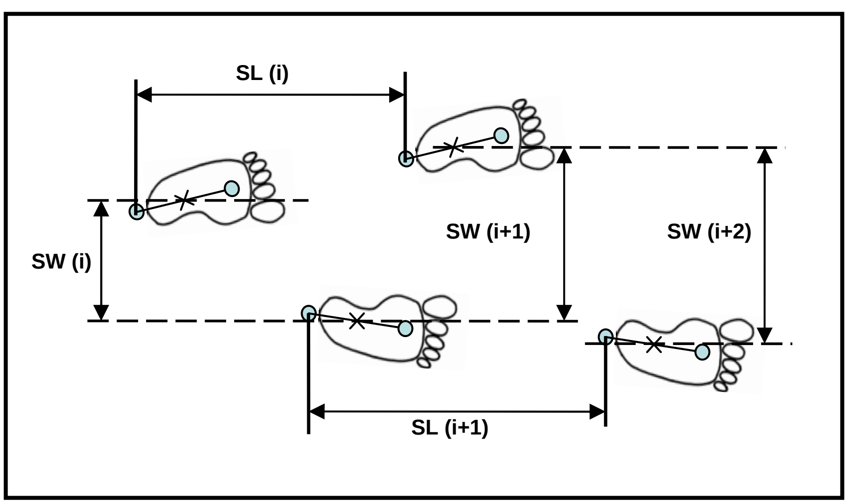
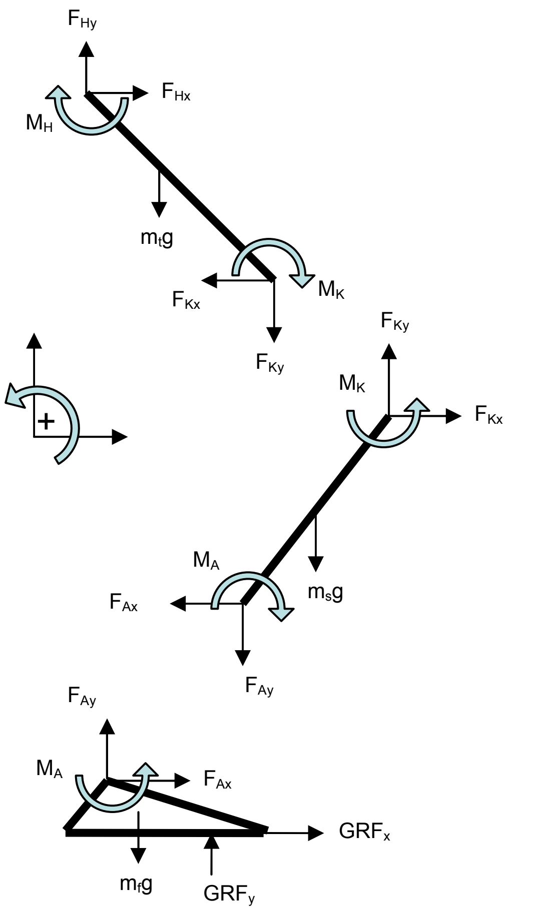


Figure 1: Schematic characterizing step width (SW) and stride length (SL) determination.

Gait Kinetics Analyses.

Marker and force plate data from overground trials were analyzed using a custom Matlab program which calculated sagittal plane net joint moments and powers for the hip, knee, and ankle. Initially, the force platform reference frame was translated to the reference frame established for the Motion Analysis system during calibration procedures. Marker position data were low-pass filtered at 6 Hz (Winter et al., 1990). Segment centers of mass (COM) positions were determined using percentages of segment lengths reported by de Leva (1996). Linear velocities and accelerations of joints and segment centers of mass and segment angular velocities and accelerations were determined through differentiation of smoothed position data using finite difference equations. In order to match the sample rate of marker position and ground reaction force data, every 10^{th} value in the ground reaction force data set was used.

Net joint moments were determined by using a sagittal plane inverse dynamics analysis with the lower body modeled as a rigid, linked segment system (Figure 2; Winter, 1990). Segment masses were estimated using percentages of total body mass (de Leva, 1996). Segment moments of inertia were determined by inputting individual subject anthropometric data into equations from Challis and Kerwin (1992). Net joint moments were subsequently normalized to percentage of stance phase in 1% increments. Finally, ensemble averages were computed for three trials within each speed for each subject. The support moment was calculated as the sum of the moments at the hip, knee, and ankle, such that extensor moments made positive contributions to the support moment. For each subject, joint moments were normalized to body mass. Positive angular impulses were determined during the stance phase for ankle, knee, hip and support moments by integrating net joint moments with respect to time. Each positive angular impulse was then normalized to stance time yielding units of N·m·kg^{-1}.



Variables:

m: segment mass
a: acceleration
α: segment angular velocity
I: segment moment of inertia
GRF: ground reaction force
x, y: horizontal (x) and vertical (y) position data
F_x, F_y: horizontal and vertical net joint forces
M: net joint moment

Subscripts:

A: ankle
K: knee
H: hip
f: foot
s: shank
t: thigh
cop: center of pressure

Ankle net joint moment determination (M_a)

$\sum F_x$: $F_{Ax} + GRF_x = m_f \cdot a_{fx}$
$\sum F_y$: $F_{Ay} + GRF_y - m_f \cdot g = m_f \cdot a_{fy}$
$\sum M$: $M_A + [\,F_{Ay} \cdot (x_A - x_f)\,] - [\,F_{Ax} \cdot (y_A - y_f)\,] - [\,GRF_x \cdot (y_{cop} - y_f)\,] + [\,GRF_y \cdot (x_{cop} - x_f)\,] = I_f \cdot \alpha_f$

Knee net joint moment determination (M_k)

$\sum F_x$: $-F_{Ax} + F_{Kx} = m_s \cdot a_{sx}$
$\sum F_y$: $-F_{Ay} + F_{Ky} - m_s \cdot g = m_f \cdot a_{sx}$
$\sum M$: $-M_A + M_K + [\,F_{ax} \cdot (y_A - y_s)\,] - [\,F_{ay} \cdot (x_A - x_s)\,] - [\,F_{Kx} \cdot (y_K - y_s)\,] + [\,F_{Ky} \cdot (x_K - x_s)\,] = I_s \cdot \alpha_s$

Hip net joint moment determination (M_h)

$\sum F_x$: $-F_{Kx} + F_{hx} = m_t \cdot a_{tx}$
$\sum F_y$: $-F_{Ky} + F_{hy} - m_t \cdot g = m_t \cdot a_{ty}$
$\sum M$: $-M_K + M_H + [\,F_{Kx} \cdot (y_K - y_t)\,] - [\,F_{Ky} \cdot (x_K - x_t)\,] - [\,F_{Hx} \cdot (y_H - y_t)\,] + [\,F_{Hy} \cdot (x_H - x_t)\,] = I_t \cdot \alpha_t$

Figure 2: Sagittal plane free body diagrams of the foot, shank, and thigh and the inverse dynamics equations used for net joint moment determination.

Joint powers were calculated as the product of the joint moments and joint angular velocities at each instant in time:

$$P_j = M_j \cdot \omega_j \quad \text{(Equation 1)}$$

where P_j is power in watts, M_j is the joint moment in N·m, and ω_j is the joint angular velocity in rad·s^{-1} (Winter, 1990). Positive power, which was assumed to be indicative of concentric muscle contributions, was generated when the joint moment and angular velocity had the same polarity.

Similarly, negative power, reflecting eccentric contributions, was generated when the joint moment and velocity had opposite polarities (Winter, 1990). Positive and negative work at each joint were calculated as the area under positive and negative portions of joint power curves. Work was also calculated across specific phases of stance for each joint. Ankle work was broken into negative (0 to ~75% stance) and positive (~75 to 100% stance) phases. As described by Winter et al. (1990), knee and hip work were broken into several periods of positive and negative power bursts. Specifically, knee work was calculated over the energy absorption phase of weight acceptance (K1, ~10 to 30%), the energy generation period as knee extends during mid stance (K2, ~30 to 50% stance), the energy generation period of knee flexion after midstance (K3, ~70 to 90% stance) and finally energy absorption period just before toe-off (K4, ~90 to 100% stride). Hip work was broken into three periods, the energy generated as hip extends during the middle portion of stance (H1, ~15 to ~70% stance), the work done on the hip flexors in late stance (H2, ~70 to ~90% stance), and energy generated by hip flexors in late stance (H3, ~90 to 100% stance) (Figure 3).

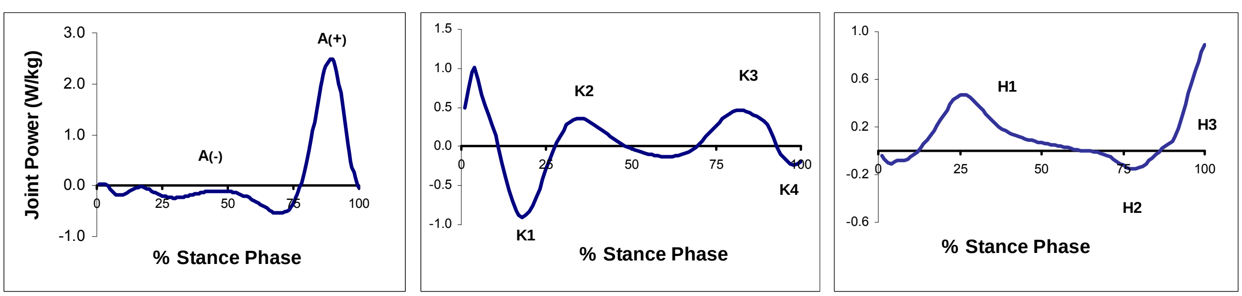


Figure 3: Exemplar joint power profiles during the stance phase of walking and multiple phases of positive and negative power at the ankle, knee, and hip (Winter et al., 1990).

The primary dependent variables of interest for assessing the mechanical work attributed to the net joint moments were ankle positive work, knee positive and negative work during the first half of stance (K1 and K2, respectively) and positive hip work during the middle portion of stance (H1). In order to show the distribution of these four variables in young and old adults, absolute joint work values were compared between young and old.

In addition to these primary dependent variables, age effects were also determined on each of the remaining work measures (K3, K4, H2, H3, and ankle negative work) and more global measures of work, including the summation of positive work across joints, summed negative work across joints, and total work (positive work plus the absolute value of negative work summed across all joints). Joint work measures were normalized by the distance traveled during stance yielding work results expressed in $J \cdot kg^{-1} \cdot m^{-1}$.

Electromyographical Analysis.

Raw EMG data were first full wave rectified and then low pass filtered at 10 Hz to create a linear envelope (Mian et al., 2006). Instantaneous EMG amplitudes for each muscle were then expressed as a fraction of the mean EMG amplitude over the first 20 strides at the 1.12 $m \cdot s^{-1}$ condition. In other words, during the 1.12 $m \cdot s^{-1}$ trial, EMG amplitude was averaged over 20 consecutive strides for each muscle. Instantaneous EMG amplitudes at all speeds were normalized to this average value for a given muscle. Onset and offset of muscular excitation was determined relative to resting EMG levels. The 5-second resting EMG trial was full wave rectified and low pass filtered at 10 Hz. Mean amplitude and its standard deviation were computed for the entire trial. Muscular onset and offset threshold was then defined as the mean resting EMG amplitude plus five standard deviations.

Coactivation was determined by describing the period in which antagonistic muscle pairs were simultaneously active. Four antagonistic pairs of muscles were identified about both the thigh and shank, including 1) tibialis anterior – soleus, 2) tibialis anterior – gastrocnemius, 3) vastus medialis – biceps femoris, and 4) vastus medialis – semitendinosus. Both soleus and gastrocnemius were included because of their differences in fiber type and uni/multiarticular structure. Both the biceps femoris and semitendinosus were included in the coactivation analysis because pilot work indicated these two muscles have different levels of coactivation with respect to the vastus medialis in young individuals.

For our third and final hypothesis, coactivation was quantified in a way which reflected the time when both antagonistic muscles were active and their amplitude of activation (Figure 4). For this determination, a coactivation index (CI) was defined as two times the area of overlap between the EMG signals divided by the summed area of each muscle in the pair (Winter, 1990):

$$CI = 2 \cdot \left(\frac{\int \min(EMG_{ag}, EMG_{antag})}{\int EMG_{ag} + \int EMG_{antag}} \right) * 100 \qquad \text{(Equation 2)}$$

CI was calculated for each of 20 consecutive strides and then averaged over these 20 strides within each speed. CI about the thigh and shank (CI_{TH}, CI_{SH}) were determined by averaging pairs of coactivation indices about each segment. Total CI (CI_{TOT}) was also determined as the sum of coactivation about the shank and thigh.

For comparison to previous studies, the time during each stride that both muscles in the agonist/antagonist pair were concurrently active was also computed and expressed as a percentage of stride time. This time of coactivation (TCA) was calculated for twenty consecutive strides and then averaged across strides for each pair of antagonist muscles. Time of coactivation about the shank (TCA_{SH}) was described as the average of time of coactivation between each pair of shank antagonistic muscles. Time of coactivation about the thigh (TCA_{TH}) was determined similarly by averaging the time of coactivation of both thigh antagonistic muscle pairs.

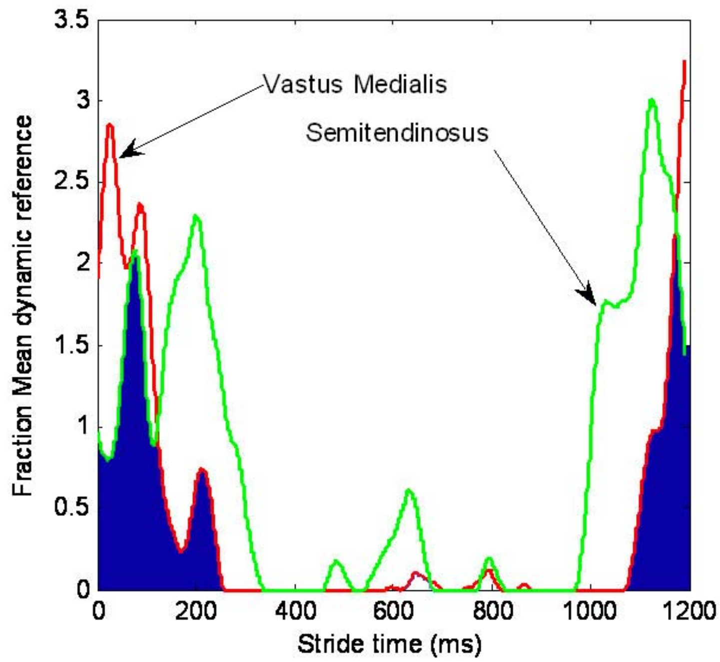


Figure 4: Example of coactivation index (CI) quantification over one stride at 1.34 m·s^{-1}. The shaded area represents the coactivation occurring between this antagonistic muscle pair. EMG data were normalized to the mean EMG amplitude while walking at 1.12 m·s^{-1}.

Statistical Analysis

A univariate two-factor (age x speed) analysis of variance (ANOVA) design with repeated measures on speed was used to examine the effects of age and speed and their interaction on step kinematic, joint kinetic, and coactivation dependent variables. To examine the relationships between C_w and the multiple dependent variables for step kinematics, joint kinetics, and muscle coactivation, linear correlation analyses were conducted. Correlations were first run with data collapsed across age at each speed. Additional correlation analyses were run within age groups at each speed. All statistical procedures were performed using Minitab for Windows with an alpha level of significance equal to 0.05.

Table 2: Primary and secondary dependant variables. Analyses of variance were run on each dependant variable, and correlations were run with each variable and C_w.

	Kinematics	Kinetics (Jt moment)	Kinetics (Jt work)	Coactivation
Primary	Step width variability	Ankle angular impulse Knee angular impulse Hip angular impulse	A(+), K1, K2, H1	CI_{TH}, CI_{SH}, CI_{TOT}
Secondary	Step width Stride length Stride length variability	Support angular impulse	A(-), K3, K4, H2, H3 (+) Work, all joints (–) Work; all joints Total work; all joints	TCA_{TH}, TCA_{SH}

RESULTS

Cost of Walking

In support of our first hypothesis, the net energy cost of walking (C_w; gross cost minus resting cost) was 23% higher in older adults with respect to young when averaged across the four walking speeds ($F_{1,26}$=29.86, p<0.001; Figure 5) and was systematically higher at every speed. C_w was also significantly affected by walking speed ($F_{3,78}$=36.79, p<0.001). C_w for both young and older adults displayed curvilinear responses across speed. In addition, resting metabolic cost was significantly higher for the young adults (M_{old}=1.37±0.19 $J \cdot kg^{-1} \cdot s^{-1}$; M_{young}=1.59±0.14 $J \cdot kg^{-1} \cdot s^{-1}$; $F_{1,26}$=12.64, p=0.001)

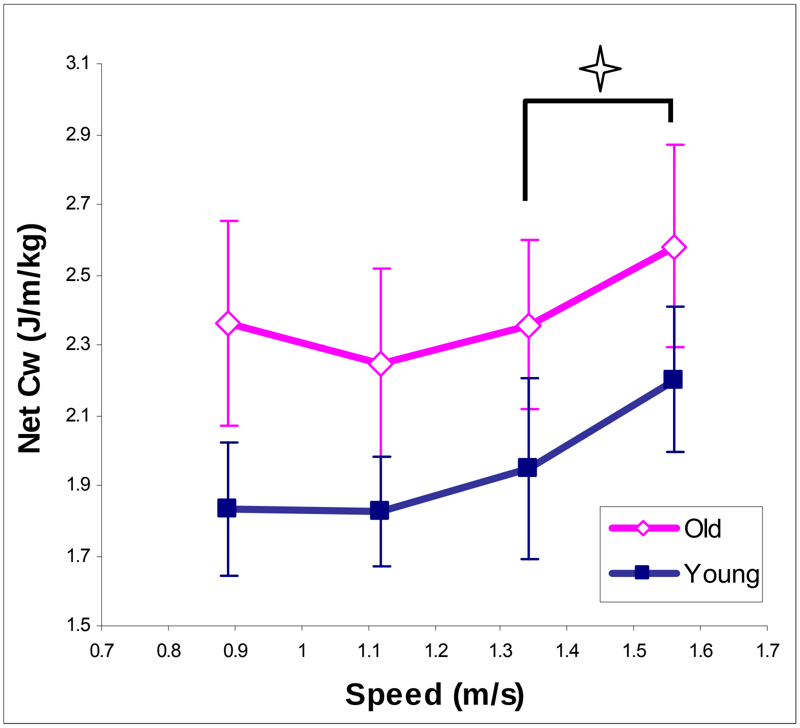


Figure 5: Net cost of walking (C_w) in old and young individuls as a function of walking speed. The star indicates a significant increase in C_w for both young ($F_{1,26}$=8.44, p=0.007) and old ($F_{1,26}$=4.96, p=0.04) between speeds 1.34 and 1.57 $m \cdot s^{-1}$.

Gait Kinematics

Contrary to our fourth hypothesis, there were no significant age or speed effects on step width variability, the primary kinematic variable of interest. Further, step width, stride length, and stride length variability were not affected by age (Figure 6). When each of the four variables was correlated to C_w, few statistically significant relationships were observed across speeds (see Appendix B). In combination, these results indicate step width, step width variability, stride length, and stride length variability contributed little to the higher C_w observed for older adults.

Stride length increased systematically with speed ($F_{3,78}$=559.46, p<0.001), whereas step width decreased modestly with speed ($F_{3,78}$=3.30, p=0.03). Stride length variability ($F_{3,78}$=11.87, p<0.001) showed a curvilinear response to speed. Preferred walking speed was nearly identical ($F_{1,26}$=0.13, p=0.73) for old (1.32±0.12 $m \cdot s^{-1}$) and young (1.34±0.17 $m \cdot s^{-1}$) participants. Hip and ankle ranges of motion during stance were also shown to be similar between old and young adults (Hip ROM: $F_{1,26}$=0.15, p=0.71; Ankle ROM: $F_{1,26}$=0.67, p=0.42).

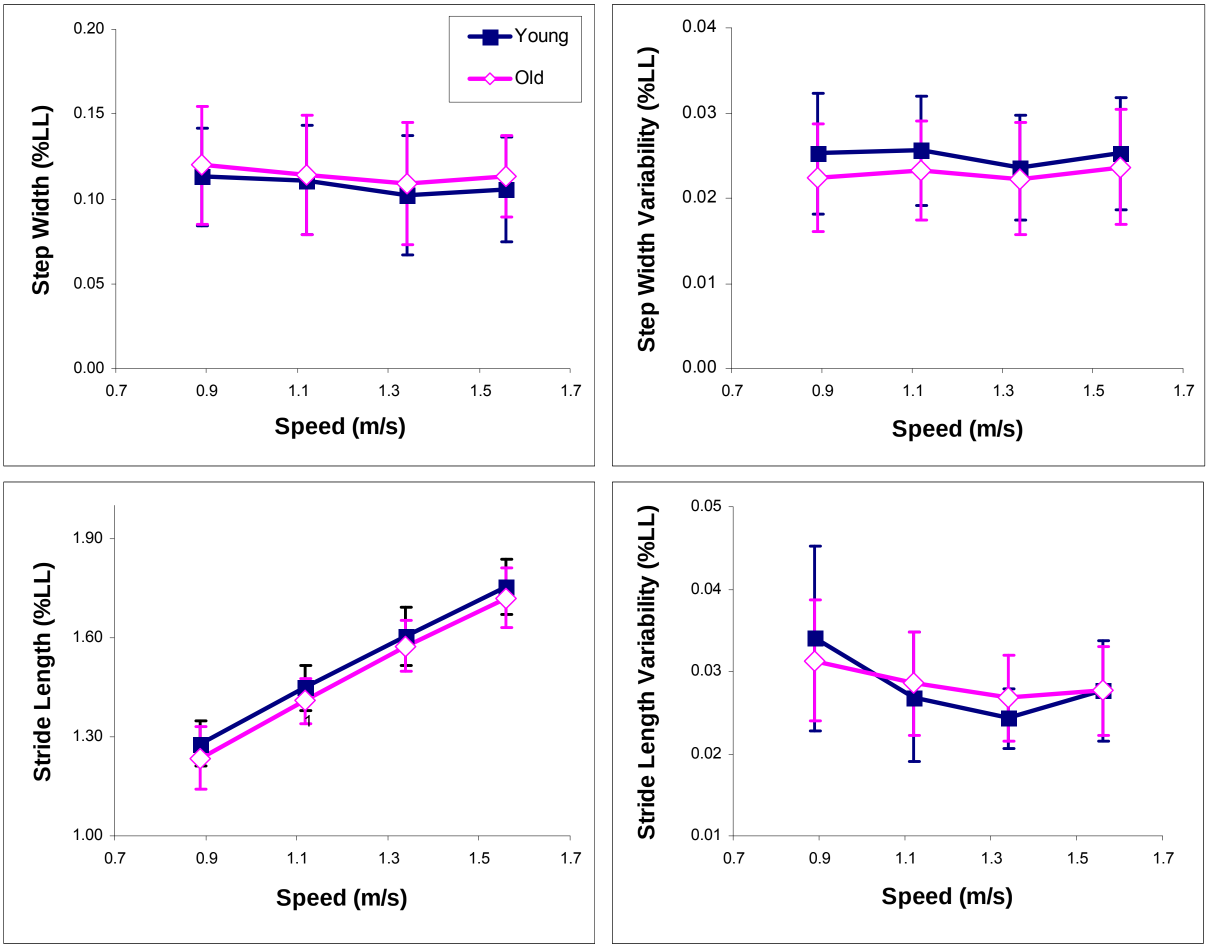


Figure 6: Step width, step width variability, stride length, and stride length variability as a function of speed for older and young adults. All four dependent variables were normalized to leg length and are reported as unitless values.

Gait Kinetics

Joint Angular Impulses.

Results for net joint moment angular impulses generally supported our third hypothesis which predicted older adults rely more on hip extensor moments and less on ankle and knee extensor moments (i.e., a proximal redistribution of lower extremity moments). Although there was no significant difference between older and young adults for ankle plantar flexor angular impulse ($F_{1,26}$=0.32, p=0.57), knee extensor angular impulse was significantly less ($F_{1,26}$=9.02, p=0.006) and hip extensor angular impulse was significantly higher ($F_{1,26}$=6.67, p=0.02) in older participants (Figure 7). Ankle, knee and hip angular impulses increased with speed ($F_{3,78}$=503.74, p<0.001; $F_{3,78}$=123.09, p<0.001; $F_{3,78}$=209.13, p<0.001, respectively). Further, age by speed interactions were noted at the knee and hip ($F_{3,78}$=6.99, p<0.001 and $F_{3,78}$=6.27, p=0.001, respectively), and indicated age-related differences were greater at higher speeds. Support moment angular impulse (the sum of impulses for ankle plantar flexor, knee extensor, and hip extensor moments) was not different between ages (Figure 7; $F_{1,26}$=1.29, p=0.26), but increased with speed ($F_{3,78}$=261.94, p<0.001).

Correlations between C_w and angular impulses at each joint were generally non-significant. The few significant relationships which were noted occurred primarily within the young group (Appendix B). Specifically, within the young group, knee extensor impulse was positively correlated with C_w at 0.89 and 1.57 $m \cdot s^{-1}$ (r=0.63, p=0.02; r=0.56, p=0.04, respectively), and hip extensor angular impulse was negatively correlated with C_w at 1.12 $m \cdot s^{-1}$ (r=-0.57, p=0.03). In the old group, ankle positive angular impulse was moderately correlated (r=0.61, p=0.02) to C_w at 1.34 $m \cdot s^{-1}$. Due to the overall lack of significance between angular impulse variables and C_w, it seems unlikely that the changes seen in joint moments are a major cause of increased C_w seen in older adults.

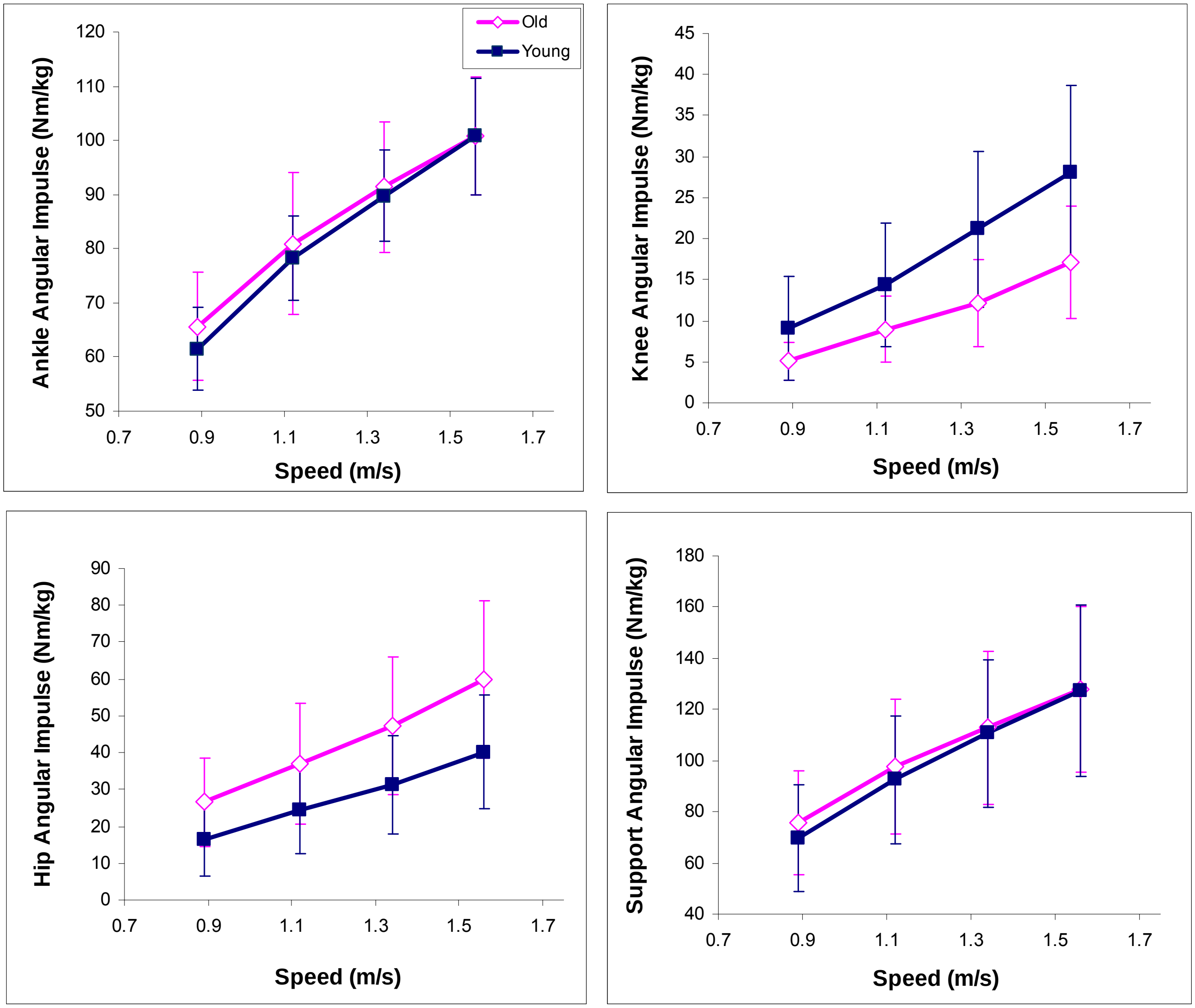


Figure 7: Ankle, knee, hip, and support moment angular impulses for older and young adults as a function of walking speed. Consistent with our hypothesis, older adults showed significantly lower knee angular impulses and higher hip angular impulses relative to young. Ankle plantar flexor impulses, however, were not different between the old and young age groups.

Joint Work.

Figure 8 contrasts ensemble average profiles for joint angular velocity, net joint moment, and instantaneous mechanical power for older and young adults at the fastest walking speed (1.57 $m{\cdot}s^{-1}$). A careful inspection of these profiles suggests joint angular kinematics for the ankle, knee, and hip were quite similar for the two age groups, whereas net joint moment profiles at the hip and knee differed between age groups. The knee moment average profile for older adults reflected a lower extensor function in the first half of stance and stronger flexor moment in the latter half of stance. The hip moment for older adults showed a stronger extensor function for much of the stance phase. As a result of the moment differences, instantaneous joint powers at the hip and knee also differed between groups.

Figure 10 summarizes mechanical work results for briefer periods of positive (energy generation) and negative (energy absorption) work reflected at the ankle, knee, and hip (e.g., K1, H1; Winter et al., 1990). As expected, these results are generally consistent those just reported for net moment angular impulses and with our second hypothesis which predicted a proximal redistribution of effort for older adults. Several observations are notable. First, statistically significant age effects were limited to results for K1, K2, and H1. At the knee joint, positive (K2) and negative (K1) work were lower for older adults than the young subjects (K2: $F_{1,26}$=7.31, p=0.01; K1: $F_{1,26}$=7.62, p=0.01). In contrast, the positive work done at the hip (H1) was higher for the older adults ($F_{1,26}$=8.88, p=0.006). In all three cases, work values for old and young differed by a factor of two. For example, work done at the hip during the H1 phase for older adults was approximately twice that for young adults. Second, there was little difference between older and young adults in positive and negative work done about the ankle during the stance phase ($F_{1,26}$=0.26, p=0.62 and $F_{1,26}$=0.20; p=0.66, respectively). We had predicted ankle work would be lower for older adults. Third, there were several statistically significant age x speed interaction effects (ankle positive work, K1 and H1). The age x speed interactions for K1 and H1 indicate the differences between old and young were accentuated at higher speeds (K1: $F_{3,78}$=7.72, $p<0.001$; H1: $F_{3,78}$=3.78, p=0.01). The interaction for ankle positive work is more subtle. Whereas positive work at the ankle increased as a function of walking speed for both age groups, the rate of increase in work done was lower for older adults at the higher walking speeds ($F_{3,78}$=6.17, p=0.001).

When summed across joints, older adults did significantly more positive work ($F_{1,26}$=11.28, p=0.002) and total work ($F_{1,26}$=6.31, p=0.02) than young (Table 2, Figure 9). Positive and total work were also shown to increase systematically with speed ($F_{3,78}$=157.24, p<0.001 and $F_{3,78}$=78.06, p<0.001, respectively). In contrast, results for negative work were not different between age groups. A complete summary of ANOVA results for work variables is included in Appendix A.

As with joint torques, correlations between joint work and C_w were sparse, and those relationships which did arise were focused generally in the young group (Appendix B). Due to the overall lack of significant findings, specifically in the old group, it seems as if the changes in work may not be major contributors to increased C_w in older adults.

Those work variables which did show significant relationships occurred only in young subjects, and exhibited negative correlations with C_w. Specifically, negative correlations were shown to exist at multiple speeds for total positive work (speeds 0.89 and 1.12$m \cdot s^{-1}$), total negative work (speeds 1.12, 1.34, and 1.57 $m \cdot s^{-1}$), K3 (speeds 0.89 and 1.12 $m \cdot s^{-1}$), K4 (all speeds), and H2 (all speeds). All correlation statistics are presented in Appendix B.

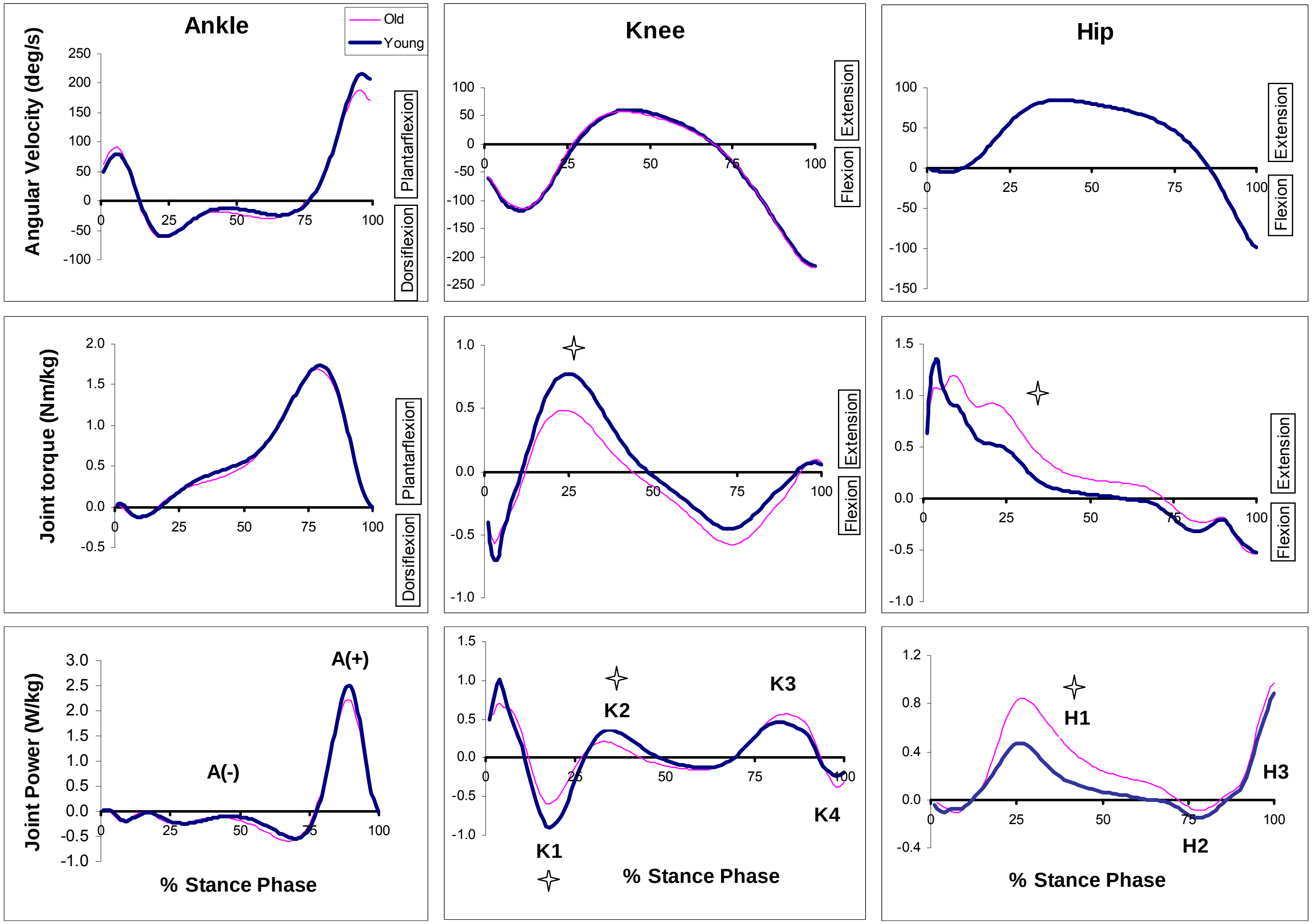


Figure 8: Joint angular velocities, net moments, and powers for young and old individuals at the ankle, knee, and hip during the stance phase of walking at 1.57 m·s^{-1}. K1-K4 and H1-H3 refer to specific periods of work absorption or generation about the knee and hip. Stars indicate significant differences between old and young. Because age effects on work were similar across speeds, angular velocity, joint moment, and joint power graphs are presented for one speed only.

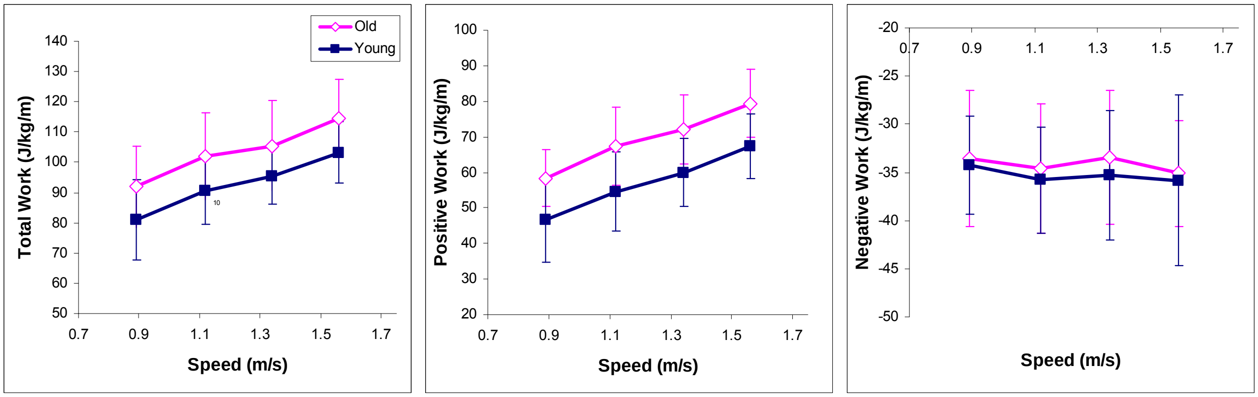


Figure 9: Total work (positive work plus the absolute value of negative work), positive work, and negative work in young and old across speeds.

Table 3: Total work (positive work plus absolute value of negative work) at each joint in old and young subjects across speeds, and the summation of those joint works giving the total work completed at the lower limb during stance. All values are expressed in J·kg^{-1}·m^{-1}.

Young	**Speed (m·s^{-1})**			
	0.89	1.12	1.34	1.57
Ankle	46.9 (5.5)	49.3 (5.1)	47.6 (5.2)	47.6 (6.0)
Knee	16.2 (5.9)	20.8 (4.4)	26.8 (3.7)	33.1 (4.5)
Hip	18.0 (6.7)	20.3 (6.3)	20.8 (5.9)	22.5 (6.4)
Total	81.0 (13.2)	91.5 (10.8)	95.1 (9.2)	103.2 (10.1)

Old				
	0.89	1.12	1.34	1.57
Ankle	49.7 (9.3)	51.8 (8.1)	49.7(8.0)	48.6 (8.2)
Knee	16.0 (4.9)	20.8 (5.2)	24.9(5.6)	31.2 (6.3)
Hip	26.3 (7.9)	29.5 (9.2)	31.0(9.5)	34.7 (9.9)
Total	92.0 (13.5)	102.1 (14.5)	105.5(14.86)	114.5 (13.1)

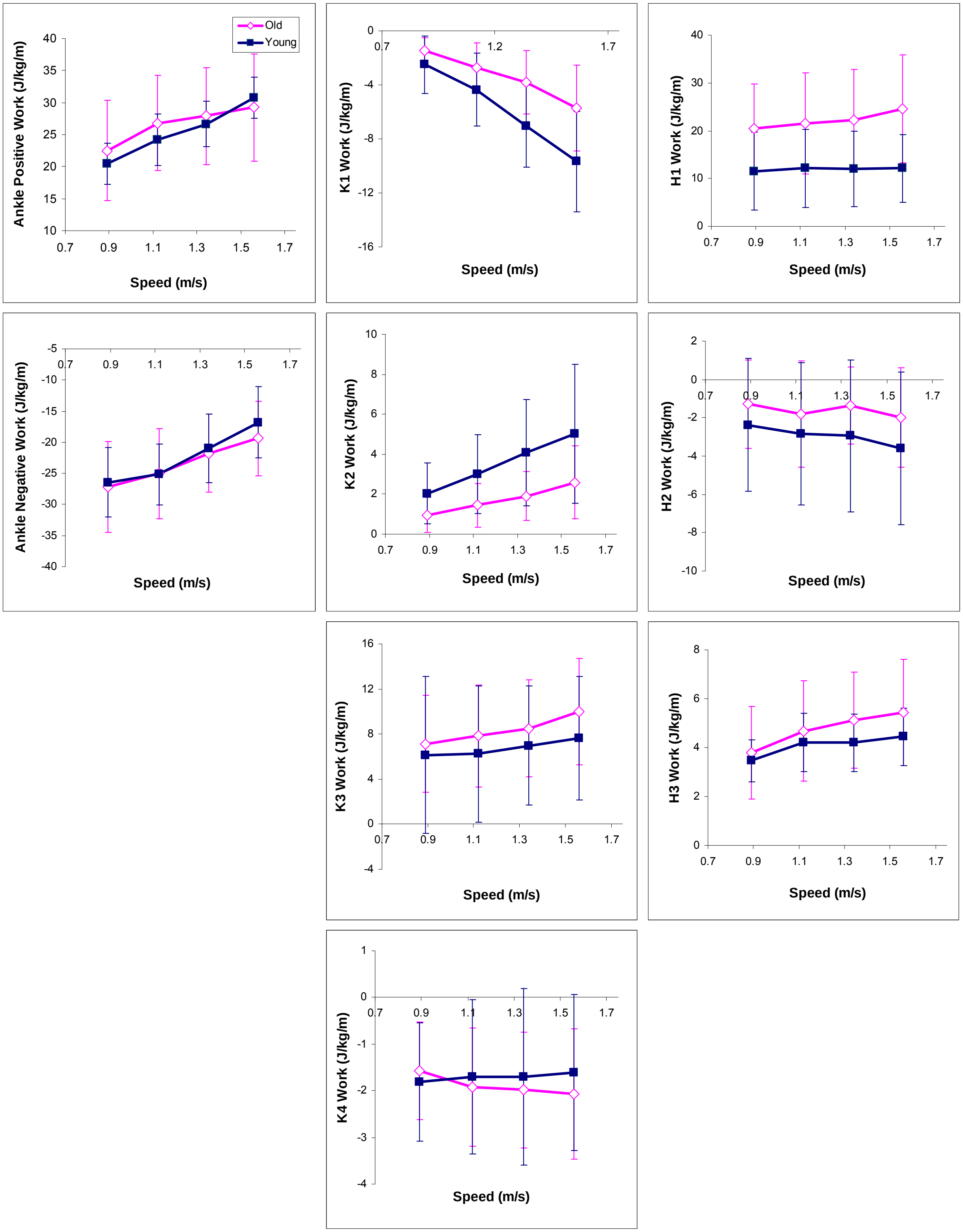


Figure 10: Work across speeds for the ankle, knee (K1-4), and hip (H1-3). Significant age effects were noted for K1, K2, and H1.

Antagonistic Muscle Coactivation

Consistent with our second hypothesis, older adults reflected higher coactivation of antagonistic musculature than young adults (Figure 11), although statistically significant differences were limited to the coactivation index about the thigh (CI_{TH}: $F_{1,26}=16.36$, $p<0.001$). In addition, significant positive correlations were shown to exist between C_w and the sum of thigh and shank coactivation indices (CI_{TOT}), indicating coactivation may indeed play a roll in the increase in C_w often reported in older adults.

Though coactivation index about the thigh was elevated, thigh time of coactivation (CI_{TH}) was not different between older and young participants ($F_{1,26}=0.37$, $p=0.34$). The increase in coactivation index with age, but not time of coactivation, indicates age primarily affected the amplitude of overlap between antagonistic musculature. For the shank, there was no difference between age groups for the coactivation index (CI_{SH}; $F_{1,26}=1.00$, $p=0.32$), though an insignificant trend for increased shank time of coactivation (TCA_{SH}) was noted ($F_{1,26}=2.91$, $p=0.10$). CI_{TOT} was larger in old ($F_{1,26}=13.57$, $p<0.001$), driven primarily by the increase of coactivation about the thigh.

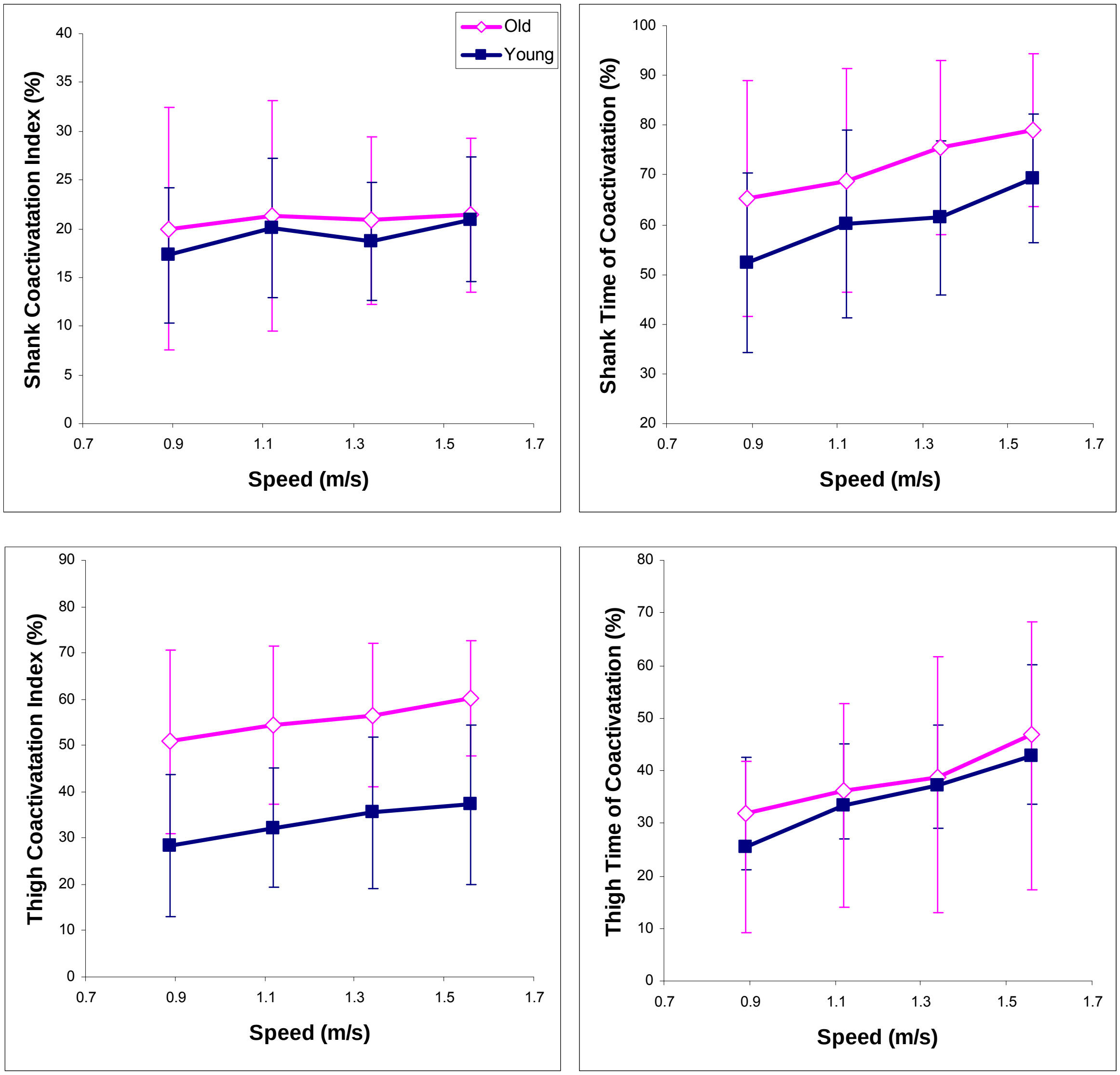


Figure 11: Coactivation in young and old across speeds. A significant age effect was seen for CI_{TH}, ($F_{1,26}$=16.36, p<0.001).

Correlations between coactivation and C_w yielded positive relationships, most prominent when data were collapsed across age. These relationships were also shown to persist in older subjects to a lesser degree in CI_{TH}, CI_{SH}, and TCA_{SH}.

When data was collapsed across age, significant positive relationships were shown to exist between C_w and CI_{TH} (all speeds), CI_{SH} (speeds 0.89, and 1.12 $m \cdot s^{-1}$) TCA_{SH} (all speeds), and CI_{TOT} (all speeds; see Appendix B). Since increases in both coactivation and C_w could be related to increases in age, rather than independently related to each other, separate correlation analyses were carried out for each age group. Results showed correlations persisted primarily in older individuals. Specifically, older adults showed positive correlations between C_w and CI_{SH} (speeds 0.89, and 1.12 $m \cdot s^{-1}$), TCA_{SH} (all speeds), and CI_{TOT} (speed 1.12 and 1.57 $m \cdot s^{-1}$; Figure 12).

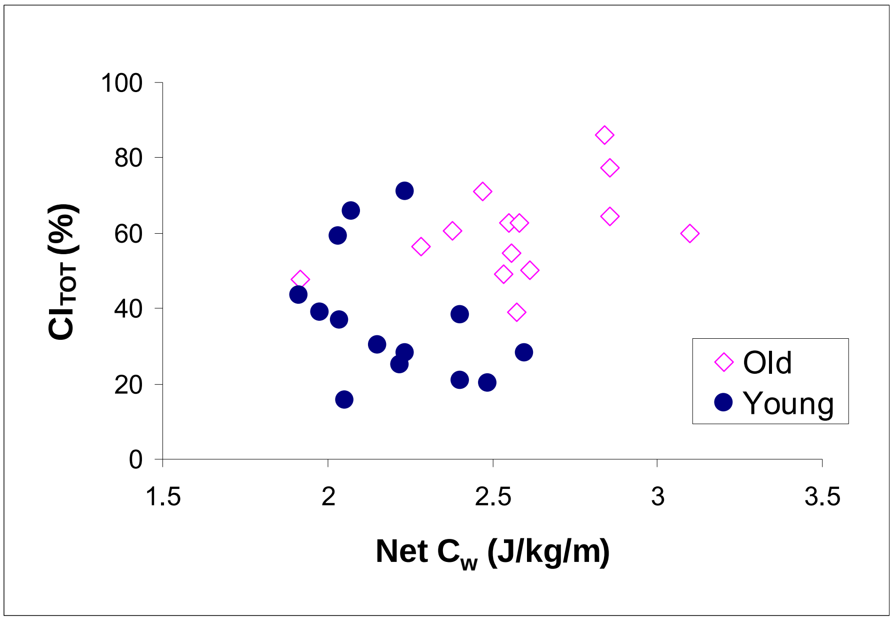


Figure 12: Scatterplot of CI_{TOT} versus net C_w at 1.57 $m \cdot s^{-1}$. When collapsed across age, CI_{TOT} and C_w showed significant positive relationship (r=0.51, p=0.01).

DISCUSSION

The overall purpose of this study was to determine the effects of age and speed on step width variability, gait kinetics, and coactivation, factors which have been highlighted as potential causes of higher net energy cost of walking (C_w) observed for older individuals. Additionally, we determined the relationship between these variables and C_w. Results indicated muscle coactivation about the thigh contributes to a higher cost of walking in older adults. Further, support for a proximal redistribution of lower extremity net joint moments and the mechanical work attributed to those moments was observed for the older subjects. Step width variability, however, did not discriminate between the age groups.

As our goal was to determine the effects of aging on C_w as opposed to pathological conditions that commonly afflict older adults, we recruited individuals, both old and young, who were recreationally physically active; free of lower limb musculoskeletal injuries and neurological disorders, and had a body mass index less than 26. The similarity between kinematic profiles of our older and young participants, including similar preferred walking speeds, hip and ankle ranges of motion, step widths, and stride lengths, suggest we were successful in identifying healthy older adults who had no limitations in their walking ability. Despite the apparent good health of our older participants, age-related adaptations were observed for several dependent variables including a higher C_w, higher levels of antagonistic muscle coactivation in the lower extremity, and a proximal redistribution of lower extremity net joint moment contributions and mechanical work attributed to the moments.

Cost of Walking

Across speeds older participants had an average net C_w that was 23% higher than young adult participants. C_w for both groups showed curvilinear relationships across speed. Other studies have also reported net C_w to be between 20 and 26% higher in older adults (Dean et al., 2007; Martin et al., 1992; Ortega & Farley, 2007). For example, Ortega and Farley (2007) showed older adults to exhibit 20% higher net C_w. In the current study, gross C_w was 7% higher in older adults when data were collapsed across walking speeds. Martin et al. (1992) reported similar age-related differences in the gross aerobic demand of walking. Older adults in their study reflected an 8% higher demand than young adults. In the following sections, three mechanisms which have been highlighted as potential contributors to the increased C_w seen in these older adults (coactivation, joint kinetics, and joint kinematics) are discussed.

Antagonistic Muscle Coactivation

In partial support of our second hypothesis, older adults showed 65% higher antagonistic muscle coactivation about the thigh (CI_{TH}). Coactivation about the shank (CI_{SH}), however, was not different for older and young participants. When CI was summed for the thigh and shank, the resulting CI_{TOT} was also elevated in old. Because the time of coactivation for the thigh was not different for older and young subjects, the higher CI_{TH} can be attributed primarily to elevated amplitude of concurrent antagonistic muscle activation.

When C_w was correlated to coactivation indices, several significant relationships emerged. Pooling the data for older and young participants, C_w values were positively related to CI_{TH} and CI_{TOT} at all speeds (CI_{TH}: r=0.46-0.57 across speeds; CI_{TOT}: r=0.39-0.46 across speeds), indicating higher C_w was associated with higher levels of antagonistic muscle coactivation. When correlations were run separately on old and young subjects, no relationships were shown to exist in young subjects. However, a positive relationship persisted with CI_{TOT} in the older adults at 1.12 $m \cdot s^{-1}$ (r=0.55) and 1.57 $m \cdot s^{-1}$ (r=0.55). These positive relationships, in combination with the higher CI_{TH} and CI_{TOT} exhibited by the older participants, imply coactivation may contribute to higher C_w frequently observed for older adults.

To our knowledge, only one other study has used coactivation indices to determine antagonistic activation in old and young individuals. Larsen et al. (2008) described coactivation about both the thigh and shank for old and young individuals during stair ascent and descent. In agreement with the current study, Larsen et al. reported that older adults exhibited higher coactivation indices about the thigh but similar coactivation indices about the shank when compared to young adults. The fact that age-related differences in coactivation were observed for the thigh but not the shank may be due in part to the need for active muscular stabilization at the knee, but not at the ankle. Hortobagyi and DeVita (2000) measured coactivation and leg stiffness, defined as the linear shortening of the lower limb divided by ground reaction force under the foot, for old individuals during downward stepping. Their study showed coactivation at both the ankle and knee to be directly related to leg stiffness. Stated differently, increases in coactivation about the ankle and knee joints were associated with increases in leg stiffness. They further noted that decrements in strength and neurological function may cause older adults to increase coactivation to stabilize and stiffen joints during dynamic tasks. Therefore, the higher coactivation we observed for the older adults about the thigh may have been an adaptation to actively stiffen the knee joint in the sagittal plane, particularly during the time surrounding heel contact. The lack of difference in coactivation between old and young about the shank may be due in part to a reduced need for active stabilization in the sagittal plane during stance. During stance phase, the two main forces acting to stiffen and stabilize the ankle are the ground reaction force (GRF) and the muscular force acting against the GRF. During this period, additional muscular force in the direction of GRF (i.e., effort from antagonistic muscles that would result in coactivation) seems unnecessary to further stabilize the joint. For example, immediately after heel strike, the dorsiflexors act eccentrically to oppose the plantar flexion influence of the GRF. In this scenario, the dorsiflexors and GRF act in combination to stiffen and stabilize the ankle joint. Thus, plantar flexor muscle contributions for the purposes of ankle stabilization is unnecessary.

Time of coactivation about the thigh was not significantly affected by age. In contrast to these results, a recent study by Mian et al. (2006) reported longer times of coactivation (TCA) in older individuals about the thigh during level walking and a modest positive relationship between TCA_{TH} and C_w. A potential reason for the discrepancy between Mian et al. and the current study is a difference in the muscles investigated. Mian et al. calculated time of coactivation for two muscle pairs about the thigh, vastus medialis – biceps femoris and vastus lateralis – biceps femoris. The current investigation, however, determined coactivation between vastus lateralis and two hamstring muscles (biceps femoris and semitendinosus). Alternatively, the health and physical activity status of subjects in the current study may have contributed to the discrepancy between study results. Indeed, previous research (Carolan & Cafarelli, 1992; Hakkinen, Kallinen, Izquierdo, Jokelainen, Lassila, Malkia, Kraemer, Newton, & Alen, 1998) has shown resistance training leads to reductions in coactivation about the knee during knee extension. Mian et al. (2006) studied individuals who were "free from frailty, or signs of gait impairment" (p.128), although no mention was made of physical activity levels of their subjects. In the current study, a concerted effort was made to recruit healthy, physically active older adults free of disease complications. Most of the older participants engaged in four or more bouts of moderate to high intensity exercise per week. Consequently, our subjects may have had lesser needs for antagonistic coactivation.

Though time of coactivation about the thigh was similar for the two age groups, there was a non-significant trend ($F_{1,26}$=2.91, p=0.10) for older adults to exhibit longer times of coactivation about the shank. Due to the trend toward higher TCA_{SH} in the absence of higher CI_{SH} for older adults, it is possible that both TCA_{SH} and the relationship between TCA_{SH} and C_w were dictated to some degree by an age-related increase in low level activation of antagonistic musculature. Due to the nature of TCA and CI calculations, prolonged periods in which antagonistic muscles were active at a low level could cause TCA to be elevated, while leaving CI relatively unaffected. Mian et al. (2006) reported higher TCA_{TH} in older adults during walking. They recognized the potential influence of low levels of coactivation on TCA, stating "It is possible that the increase in time overlap was mainly due to an increase in relatively low activation levels" (p.136). If TCA_{SH} in the current study was influenced to some degree by low level antagonistic activation, the relationships between C_w and TCA_{SH} may also have been affected. The positive relationships seen in the current study between C_w and TCA_{SH} are likely due to low level activation, as opposed to changes in the amplitude of antagonistic activation.

Gait Kinetics

In support of our third hypothesis, older adults exhibited a proximal redistribution of net joint moment angular impulses and mechanical work in the lower extremity compared to young participants. Age by speed interactions were also observed for positive work generated at the ankle, knee, and hip (see Figure 10, p.32). At the knee and hip, age-related differences became more prominent as speed increased. At the ankle, young participants systematically increased the positive work done as speed increased. Older adults, however, displayed more modest increases in positive work with increases in speed for the two higher speeds (1.34 and 1.57 $m \cdot s^{-1}$). These interactions indicate that older participants relied more on hip musculature and less on ankle plantar flexors and knee extensors than young to generate the higher walking speeds.

Several previous studies (DeVita & Hortobagyi, 2000; Savelberg et al., 2007; Silder et al., 2008) have reported lower levels of mechanical work about the ankle and knee and higher levels of work about the hip during walking for older individuals when compared with young adults. For example, DeVita and Hortobagyi (2000) determined the joint torques and powers of old and young adults at each group's preferred walking speed, which were nearly identical for their young and old groups (1.48 $m \cdot s^{-1}$). Despite walking at the same speed, older adults produced less ankle work, less knee work in the first half of stance, and more hip work. Savelberg et al. (2007) investigated joint kinetics for aerobically trained and inactive older and young adults. They showed lower ankle and knee joint powers and higher work at the hip for their older subjects. For their study, subjects were asked to walk at a controlled walking speed (1.50 $m \cdot s^{-1}$) and their preferred velocities. However, for both the controlled speed condition and preferred walking condition, older adults walked at significantly slower speeds than young. For example, in the controlled walking speed condition, young adults walked at an average of 1.69 $m \cdot s^{-1}$, while older adults walked at 1.47 $m \cdot s^{-1}$. In the current study, our older participants showed lower levels of work about the knee and higher levels of work about the hip than our young subjects, whereas work about the ankle was similar for the two groups. The discrepancy between our results for work done about the ankle and those of other investigators again is likely associated with the high physical activity levels and high quality of health of our older participants. Our older subjects participated in several bouts of aerobic or resistance training per week, including tennis, squash, hiking, and resistance training. Indeed, Silder et al. (2008)

reported peak positive plantar flexor power during walking to be directly related to maximum isokinetic ankle strength. Though lower extremity strength was not measured in the current investigation, it is possible our older participants have shown less decline in plantar flexor strength than is typically observed in older adults and were better able to maintain power production about the ankle during walking relative to older subjects in several other investigations.

Savelberg et al. (2007) specifically recruited aerobically trained and inactive young and older adults, and showed both active and inactive older adults to have substantially reduced ankle power during walking with respect to young. However, older adults in Savelberg's study also walked at a significantly slower speed (1.47 $m \cdot s^{-1}$) than their young subjects (1.69 $m \cdot s^{-1}$). As our study shows, net joint moments and power about the joints of the lower extremity are often directly influenced by walking speed (Appendix A). Thus the lower ankle work reported by Savelberg et al. may simply be due to the slower walking speeds used by their older subjects. In the current study, old and young individuals each walked at the same absolute speeds, and no differences in ankle work between age groups were shown. Results from Savelberg et al. are confounded by differences in walking speed between their young and old adults, which makes it difficult to compare directly their results with those from the current study.

Previous investigations have cited several mechanisms for the proximal redistribution of joint work in older adults. One potential cause is the disproportionate loss of distal leg muscle strength seen with age. Indeed, plantar flexor and knee extensor muscular strength have been shown to decline more quickly than hip extensor musculature (Candow & Chilibeck 2005; Kubo, Ishida Komuro, Tsunoda, Kanehisa, & Fukunaga 2007; Schulz, Ashton-Miller, & Alexander, 2007). For example, Kubo et al. (2007) measured the torque production of ankle plantar flexors and knee extensors in young and old adults. Older adults had knee extensor and plantar flexor decrements of 38 and 35%, respectively, relative to young adults. Candow and Chilibeck (2005) found similar results. Ankle plantar flexor torque for their older subjects was 25% lower and knee extensor torque was 28% lower than younger subjects. Schulz et al. focused on age-related changes in hip and knee extension strength. They reported substantially greater differences between older and young women for knee extensor strength (~40%) than hip extensor strength (~20%). These results are consistent with the hypothesis that reductions in maximal force production specifically at the knee and ankle may contribute to a proximal redistribution of net joint moments and work. In congruence with these findings, our results indicate our older subjects relied less on knee extensor contributions and more on hip extensor contributions than our young participants. Though there were no significant differences in ankle work between older and young participants, the age by speed interactions observed for positive work at both at the ankle and hip were consistent with the proximal redistribution hypothesis. Although the older participants were able to generate sufficient plantar flexor work at slower walking speeds, results demonstrated increasing importance of hip power and less reliance on ankle power at the higher walking speeds.

An alternative explanation for the reduction in plantar flexor contributions seen in older adults during walking is associated with age related changes in joint range of motion (ROM) during walking. DeVita and Hortobagyi (2000) showed older adults produce less plantar flexor and more hip extensor work during walking than younger individuals. In addition, older adults exhibited a larger hip ROM and a reduced ankle ROM during stance phase. DeVita and Hortobagyi, therefore, suggested that “the reduced ROM at the ankle used by the elderly adults may have contributed to the reduced torque and power produced by the ankle plantar flexors” (p.1810). In the current study, there were no differences between older and young participants in the observed ranges of motion of the hip or ankle during stance. This kinematic similarity may partly explain discrepancies between our study and those of DeVita and Hortobagyi related to ankle contributions to walking.

Gait Kinematics

Contrary to our fourth and final hypothesis, no age-related differences were seen in step width variability when older and young walked at similar speeds. Further, none of the other kinematic variables of interest, including stride length, stride length variability, preferred walking speed, and joint ranges of motion, showed significant differences between age groups. This lack of difference in kinematic variables is not entirely inconsistent with current literature. Results of studies looking at differences in step width variability between old and young adults have been equivocal. For example, Grabiner and colleagues (Owings & Grabiner, 2004a; Grabiner et al., 2001) observed higher step width variability for older participants, whereas others (e.g., Kang & Dingwell, 2008; Dean et al., 2007) reported no age-related differences in step width variability. Though these studies give limited information on the physical activity levels of older and young participants, it is possible that health and physical activity status of subjects played a role in the discrepancy of these results. In the current study, older subjects were particularly active, potentially allowing the older adults in this study to retain kinematic profiles similar to younger adults.

Though step width variability was similar for older and young participants in our study, C_w was significantly higher in older subjects. Due to the lack of covariance of these variables across age groups, step width variability does not reliably explain the higher C_w seen in our older adults.

Cost of Force Production

In this investigation, we tested three potential causes of higher C_w in older adults: step width variability, proximal redistribution of joint work, and coactivation about the shank and thigh. Another potential cause of higher C_w in older adults that was not addressed by the current study is an increase in the metabolic cost to produce force with age. To our knowledge, no research has been conducted to specifically address the relationship between age-related changes in C_w and muscle specific force. Nevertheless, there is evidence that muscle specific force, i.e., force per physiological cross-sectional area, declines as a function of age and that older adults may need to recruit more motor units than young to produce the same total force (Frontera, Suh, Krivickas, Hughes, Goldstein, & Roubenoff, 2000; Larsson, Li, & Frontera 1997; Macaluso, Nimmo, Foster,Cockburn, McMillan, & DeVito, 2002; Morse, Thom, Reeves, Birch, & Narici, 2005).

Macaluso et al. (2002) quantified the specific force of knee extensors and flexors in old and young adults. Since musculature in older adults has been shown to have larger components of connective tissue than young (Kent-Braun, Ng, & Young, 2000), Macaluso and colleagues limited their analysis of specific force to only contractile portions of musculature. They determined the volumes of both contractile components and connective tissue components of quadriceps and hamstring muscles for young and old adults. Specific force was then defined for knee extensors and flexors as maximal isometric torque divided by the volume of contractile components. Results showed maximal torque per unit contractile volume of knee extensors and flexors was 20-35% lower in older adults relative to young subjects. Macaluso and colleagues, however, also indicated the reduction in maximal torque with age, and thus specific force, may have been confounded by higher levels of activation of antagonist musculature. In a more recent study (Morse et al., 2005), reductions in the specific force of older adults were shown to persist despite similar levels of antagonistic activation. Morse et al. (2005) looked at effects of age on plantar flexor specific force and several additional aspects of musculature, including antagonistic activation, agonist activation, and pennation angle. Older adults exhibited lower agonist muscular activation during isometric maximal contraction, similar antagonist muscular activation levels, and smaller pennation angles with respect to young. After accounting for these differences, specific force was approximately 30% lower in older adults. Studies have also been carried out to test the force producing capacity of isolated aged and young muscle fibers (Frontera et al., 2000; Larsson et al., 1997). For example, Frontera et al. (2000) determined contractile properties of chemically skinned segments from single fibers of vastus lateralis obtained from muscle biopsies from old and young men. Results showed that specific force of type I and type IIa fibers were 39 and 45% higher, respectively, in younger adults.

The adaptations in specific force seen in both whole muscle and at the fiber level suggest older adults need to recruit more motor units than young to produce the same overall force during a particular task. As a consequence of recruiting more motor units, one would expect a higher cost for generating a given amount of force. If one extrapolates this argument to walking patterns for older and young adults that are kinematically-similar, one would predict a higher energetic cost for older adults.

If older adults needed to recruit more muscle units to complete each walking task than young, one would also expect to see differences in EMG responses between age groups. In the current study, a direct comparison of EMG responses of the older and young participants cannot be made because of the EMG normalization scheme used. EMG was normalized to the average amplitude of EMG during a reference speed ($1.12\ m{\cdot}s^{-1}$). Since we normalized data to a reference speed rather than a standardized contraction level (e.g., voluntary maximal isometric contraction), the relative EMG amplitudes cannot be compared across groups. Hortobagyi and DeVita (2000), however, showed older adults produced substantially larger EMG than young during similar tasks. They looked at EMG of the vastus medialis during several activities of daily living (stair ascent, descent, and standing from a chair) in both old and young adults. EMG data were normalized to maximal voluntary isometric contractions. Normalized EMG of the vastus medialis was approximately two times larger in older adults than young in each of the activities of daily living. These results support the idea that older adults need to recruit more motor units to complete a given task than young adults.

Limitations of the Current Study

Several limitations of the current study should be noted. First, measures of joint kinetics were taken during overground walking, whereas metabolic, kinematic, and electromyographic data were collected during treadmill walking. Therefore, correlations between C_w, collected during treadmill walking, and joint kinetics, collected during overground walking, may be affected. A recent study by Lee and Hidler (2008) showed that although kinematic profiles were similar between overground and treadmill walking in young healthy adults, joint powers were significantly different between the two modes of walking. Specifically, the magnitude of maximum negative knee power during the first half of stance was smaller for treadmill walking with respect to overground. In addition, maximal positive hip power was elevated for treadmill walking. In order to minimize these potential differences between overground and treadmill gait patterns, each subject in the current study underwent at least 30 minutes of treadmill walking over the range of speeds tested. Nevertheless, it remains possible that differences in overground and treadmill kinetics persisted. If kinetics during treadmill walking were different than overground in the current study, the relationship between kinetics, collected during overground walking, and Cw, collected during treadmill walking, may not provide an accurate assessment of the true relationship between these variables.

Although a substantial amount of treadmill practice was completed for both age groups, six of the older adults (two males and four females) felt the need to rest a hand on the handrail as a means of keeping their bearing on the treadmill belt. Previous literature has shown hand rail support produced modest decreases to mediolateral sway (variance about the mean mediolateral position of the center of mass), and to a lesser degree, aerobic demand during walking (Berling, Foster, Gibson, Doberstein, & Porcari, 2006; Dickstein & Laufer, 2004; Owings & Grabiner 2004a). In the current study, when data collected from subjects who used handrail support were removed from the analyses, mean stride width and stride width variability for the older group were 7 and 15% higher, respectively. Nevertheless, when statistical contrasts were performed on these data, there still were no significant differences in gait variability (SW: $F_{1,14}$=0.89, p=0.36; SWV: $F_{1,14}$=0.47, p=0.51). In addition, removing the data for these six older participants from the C_w analysis did not affect the mean C_w for the older subjects.

Finally, kinematic and metabolic data were collected simultaneously during treadmill walking. In other words, the kinematic data were captured when the subject was linked to the metabolic cart by the mouthpiece and hose used to capture expired air. It is possible that using the mouthpiece may have affected measures of walking mechanics. Specifically, variability in step width and stride length may have been attenuated. Simultaneous measurements of metabolic data and variability in stride width (Dean et al., 2007) and stride time (Malatesta et al., 2000) have been used in the past. Malatesta and colleagues reported higher stride time variability for older subjects despite use of the mouthpiece while temporal and kinematic characteristics were captured. The effect of this methodology on variability, however, was not investigated in these or the current study.

Conclusions

Several conclusions can be drawn from the results of this study:

1. Healthy active older adults exhibit higher net C_w than young at a range of walking speeds. Further, C_w shows a curvilinear relationship across speed in both young and older adults.

2. Healthy active older adults have higher levels of antagonist muscle coactivation about the thigh than young adults across a range of walking speeds, but they do not have higher levels of coactivation about the shank. Total coactivation index, the sum of CI_{TH} and CI_{SH} was positively correlated to C_w at speeds 1.12 and 1.57 $m \cdot s^{-1}$. The higher total coactivation index for older adults, along with the positive relationship between CI_{TH} and C_w implies CI_{TOT} may contribute to the higher C_w observed for older adults.

3. Healthy active older adults exhibit a proximal redistribution of net joint moments and the work attributed to them across a range of walking speeds. This conclusion is based on results for the knee and hip joints only. Inconsistent with our hypothesis, older adults did not display lower ankle plantar flexor moments and work than young adults. Though older adults saw both a shift in work from the knee to hip in older adults and higher C_w, positive relationships were not established between knee and hip work variables and C_w. Therefore, a direct relationship was not established between knee and hip work and C_w.

4. Healthy active older adults do not display higher step width variability than young adults across a range of walking speeds. The absence of age-related differences and the lack of significant correlations between C_w and gait variability suggests gait variability indicators do not distinguish C_w between older and young adults.

Future Directions

To fully understand cause and effect relationships between two variables, one must specifically manipulate one variable while measuring changes in the other. Therefore, an alternative approach to determining the causes of higher C_w in older adults is to manipulate factors such as joint kinetics and measure the effects on C_w. For example, determining how manipulations of plantar flexor or hip power affect C_w could give a clearer indication of how age-related adaptations in joint kinetics affect metabolic cost. Also, the relationship between muscle specific force and C_w has yet to be established in the literature, although several authors (e.g., Malatesta et al., 2002; Martin et al., 1992; Ortega & Farley, 2007) have identified it as a potential cause of higher C_w in older adults. One way in which this might be possible is to have older individuals undergo aerobic or resistance training, while variables such as coactivation, kinetics, kinematics and specific force are periodically quantified. A recent study by Thomas, DeVito, and Macaluso (2007) showed an overspeed training program with body weight unloading reduced C_w at several walking speeds in older adults. Unfortunately, variables such as specific force, gait kinematics, kinetics, and coactivation were not measured in these individuals. Future training studies may be able to establish covariance between these variables and C_w, shedding more light on factors affecting C_w. Finally, effects of physical activity level on coactivation in older adults should be carried out. Coactivation has been shown to be lower in isometric tasks after resistance training. To our knowledge, however, no studies have looked at the effects of physical activity status on coactivation during functional tasks such as walking. Determining whether C_w and coactivation are affected similarly by activity level could provide further insight as to how these variables are related.

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APPENDIX A

Means, Standard Deviations and Analysis of Variance Statistical Results

Cost of Walking

Table A1. Means (SD) and ANOVA results for net C_w ($J \cdot kg^{-1} \cdot m^{-1}$)

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	1.83(0.19)	1.83(0.16)	1.95(0.26)	2.20(0.21)	1.95
Old	2.36(0.29)	0.25(0.27)	2.36(0.24)	2.58(0.29)	2.39
Means	2.10	2.04	2.15	2.39	

	F value	p value
Age	29.86	<0.001
Speed	36.79	<0.001
Age x Speed	1.64	0.19

Table A2. Means (SD) and ANOVA results for gross C_w ($J \cdot kg^{-1} \cdot m^{-1}$)

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	3.62(0.26)	3.24(0.16)	3.13(0.25)	3.21(0.19)	3.30
Old	3.89(0.31)	3.47(0.35)	3.38(0.31)	3.46(0.37)	3.55
Means	3.76	3.36	3.25	3.34	

	F value	p value
Age	6.67	0.02
Speed	80.31	<0.001
Age x Speed	0.23	0.88

Kinematic Variables

Table A3. Means (SD) and ANOVA results for step width (normalized to leg length).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	0.113(0.029)	0.111(0.032)	0.102(0.035)	0.106(0.031)	0.108
Old	0.120(0.035)	0.114(0.036)	0.109(0.036)	0.114(0.024)	0.114
Means	0.116	0.113	0.106	0.110	

	F value	p value
Age	0.29	0.60
Speed	3.30	0.03
Age x Speed	0.92	0.16

Table A4. Means (SD) and ANOVA results for step width variability (normalized to leg length, unitless).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	0.025(0.007)	0.026(0.006)	0.024(0.006)	0.025(0.007)	0.025
Old	0.022(0.006)	0.023(0.006)	0.022(0.007)	0.024(0.007)	0.023
Means	0.024	0.024	0.023	0.024	

	F value	p value
Age	0.79	0.38
Speed	1.56	0.21
Age x Speed	0.35	0.80

Table A5. Means (SD) and ANOVA results for stride length (normalized to leg length, unitless).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	1.28(0.07)	1.45(0.07)	1.60(0.9)	1.75(0.08)	1.52
Old	1.23(0.10)	1.41(0.07)	1.57(0.07)	1.72(0.09)	1.48
Means	1.26	1.43	1.59	1.74	

	F value	p value
Age	0.39	0.54
Speed	559.46	<0.001
Age x Speed	0.25	0.86

Table A6. Means (SD) and ANOVA results for stride length variability (normalized to leg length, unitless).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	0.029(.011)	0.022(0.008)	0.019(.004)	0.023(0.006)	0.029
Old	0.026(0.007)	0.024(0.006)	0.022(0.005)	0.023(0.005)	0.024
Means	0.028	0.023	0.021	0.023	

	F value	p value
Age	0.03	0.87
Speed	11.8	<0.001
Age x Speed	1.69	0.18

Table A7. Means (SD) and ANOVA results for ankle range of motion during stance (º).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	23.1(3.7)	26.6(3.6)	26.3(2.8)	26.8(3.4)	26.5
Old	24.9(5.1)	25.6(4.6)	25.4(5.2)	24.9(5.1)	25.2
Means	25.5	26.1	25.8	25.9	

	F value	p value
Age	0.67	0.42
Speed	0.50	0.69
Age x Speed	0.44	0.73

Table A8. Means (SD) and ANOVA results for hip range of motion during stance (º).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	34.0(3.6)	36.2(2.9)	39.2(3.6)	43.0(4.5)	38.1
Old	34.5(2.9)	36.9(3.1)	39.4(3.2)	43.4(3.7)	38.5
Means	34.3	36.5	39.3	43.2	

	F value	p value
Age	0.15	0.71
Speed	183.74	<0.001
Age x Speed	0.12	0.95

Kinetic Variables: Joint Angular Impulse

Table A9. Means (SD) and ANOVA results for ankle angular impulse ($N \cdot m \cdot kg^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	61.5(7.7)	87.3(7.8)	89.8(8.4)	100.7(10.7)	82.6
Old	65.7(10.1)	80.9(13.1)	91.4(12.1)	100.8(10.8)	84.7
Means	63.6	79.3	90.6	100.8	

	F value	p value
Age	0.32	0.57
Speed	503.7	<0.001
Age x Speed	1.5	0.22

Table A10. Means (SD) and ANOVA results for knee angular impulse ($N{\cdot}m{\cdot}kg^{-1}$).

	Speed ($m{\cdot}s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	9.1(6.3)	14.3(7.5)	21.2(9.4)	28.0(10.7)	18.2
Old	6.1(2.3)	8.9(4.0)	12.1(5.3)	17.1(6.8)	10.8
Means	7.1	11.6	16.6	22.6	

	F value	p value
Age	9.02	0.006
Speed	124.09	<0.001
Age x Speed	6.99	<0.001

Table A11. Means (SD) and ANOVA results for hip angular impulse ($N{\cdot}m{\cdot}kg^{-1}$).

	Speed ($m{\cdot}s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	16.6(10.2)	24.4(11.8)	31.4(13.3)	40.1(15.5)	28.1
Old	26.5(12.1)	37.0(16.5)	47.2(18.7)	59.8(21.4)	42.6
Means	21.5	30.7	39.3	50.0	

	F value	p value
Age	6.67	0.02
Speed	209.0	<0.001
Age x Speed	6.27	0.001

Table A12. Means (SD) and ANOVA results for support angular impulse ($N{\cdot}m{\cdot}kg^{-1}$).

	Speed ($m{\cdot}s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	69.6(21.1)	92.6(25.0)	110.6(28.7)	127.2(33.3)	100.0
Old	75.7(20.1)	97.7(26.2)	112.8(29.8)	128.0(32.5)	103.6
Means	72.7	95.2	111.7	127.6	

	F value	p value
Age	0.13	0.73
Speed	261.94	<0.001
Age x Speed	0.73	0.54

Kinetic Variables: Joint Work

Table A13. Means (SD) and ANOVA results for total work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	81.0(13.2)	90.5(10.8)	95.3(9.2)	103.2(10.1)	92.5
Old	92.0(13.5)	102.1(14.5)	105.5(14.9)	114.5(13.1)	103.5
Means	86.5	96.3	100.4	108.8	

	F value	p value
Age	6.31	0.02
Speed	78.06	<0.001
Age x Speed	0.08	0.97

Table A14. Means (SD) and ANOVA results for positive work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	64.7(11.9)	54.7(11.3)	60.0(9.6)	67.3(9.0)	69.3
Old	58.4(8.0)	67.5(11.0)	72.1(10.0)	79.4(9.7)	57.2
Means	52.6	61.1	66.0	73.3	

	F value	p value
Age	11.28	0.002
Speed	157.2	<0.001
Age x Speed	0.12	0.95

Table A15. Means (SD) and ANOVA results for negative work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	-34.3(5.1)	-35.8(5.5)	-35.3(6.7)	-35.8(8.9)	-35.3
Old	-33.6(7.1)	-35.6(6.7)	-33.4(6.9)	-35.1(5.5)	-34.2
Means	-33.9	-35.2	-34.3	-35.5	

	F value	p value
Age	0.32	0.58
Speed	1.39	0.25
Age x Speed	0.46	0.71

Table A16. Means (SD) and ANOVA results for positive ankle work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	20.5(3.2)	24.2(4.1)	26.6(3.6)	30.8(3.2)	25.5
Old	22.5(7.8)	26.8(7.4)	27.9(7.6)	29.2(8.4)	26.6
Means	11.47	13.62	14.18	14..84	

	F value	p value
Age	0.26	0.62
Speed	82.18	<0.001
Age x Speed	6.17	0.001

Table A17. Means (SD) and ANOVA results for negative ankle work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	-26.4(5.6)	-25.2(4.9)	-21.0(5.5)	-16.8(5.7)	-22.4
Old	-27.2(7.3)	-25.0(7.3)	-21.7(6.3)	-19.4(6.0)	-23.3
Means	-26.8	-25.1	-21.4	-18.1	

	F value	p value
Age	0.20	0.66
Speed	74.62	<0.001
Age x Speed	1.58	0.20

Table A18. Means (SD) and ANOVA results for K1 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	-2.5(2.13)	-4.35(2.68)	-7.04(3.07)	-9.66(3.72)	-5.89
Old	-1.48(9.44)	-2.71(1.81)	-3.79(2.35)	-5.71(3.19)	-3.42
Means	-1.99	-3.53	-5.42	-7.69	

	F value	p value
Age	7.62	0.01
Speed	99.85	<0.001
Age x Speed	7.72	<0.001

Table A19. Means (SD) and ANOVA results for K2 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	22.03(1.51)	3.00(1.96)	4.07(2.65)	5.01(3.48)	3.53
Old	0.96(.87)	1.43(1.09)	1.90(1.24)	2.59(1.84)	1.72
Means	1.50	2.22	3.00	3.81	

	F value	p value
Age	7.31	0.01
Speed	23.13	<0.001
Age x Speed	2.18	0.10

Table A20. Means (SD) and ANOVA results for K3 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	6.15(7.00)	6.23(6.03)	6.96(5.27)	7.65(5.50)	6.75
Old	7.12(4.32)	7.83(4.5)	8.49(4.32)	9.97(4.72)	8.36
Means	6.63	7.03	7.73	8.81	

	F value	p value
Age	0.69	0.42
Speed	12.2	<0.001
Age x Speed	1.03	0.39

Table A21. Means (SD) and ANOVA results for K4 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	-1.81(1.27)	-1.70(1.66)	-1.70(1.89)	-1.60(1.67)	-1.70
Old	-1.57(1.04)	-1.82(1.26)	-2.00(1.23)	-2.06(1.40)	-1.89
Means	-1.69	-1.81	-1.84	-1.83	

	F value	p value
Age	0.13	0.72
Speed	0.32	0.81
Age x Speed	1.49	0.22

Table A22. Means (SD) and ANOVA results for H1 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	11.5(8.1)	12.1(9.2)	12.0(7.9)	12.1(7.1)	25.2
Old	20.4(9.4)	21.5(10.6)	22.2(10.7)	24.6(11.2)	22.2
Means	16.0	16.8	17.1	18.4	

	F value	p value
Age	8.88	0.006
Speed	5.93	0.001
Age x Speed	3.78	0.01

Table A23. Means (SD) and ANOVA results for H2 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	-2.37(3.47)	-2.84(3.71)	-2.94(3.97)	-3.60(3.97)	-2.94
Old	-1.29(2.30)	-1.80(2.79)	-1.37(2.02)	-1.98(2.61)	-1.61
Means	-1.83	-2.32	-2.16	-2.79	

	F value	p value
Age	1.32	0.26
Speed	4.32	0.007
Age x Speed	0.65	0.59

Table A24. Means (SD) and ANOVA results for H3 work ($J \cdot kg^{-1} \cdot m^{-1}$).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	3.46(0.87)	4.20(1.19)	4.21(1.18)	4.44(1.19)	4.08
Old	3.79(1.91)	4.68(2.04)	4.13(1.96)	5.43(2.18)	4.76
Means	3.63	4.44	4.67	4.93	

	F value	p value
Age	1.55	0.22
Speed	11.61	<0.001
Age x Speed	0.95	0.42

Antagonistic Muscle Coactivation

Table A25. Means (SD) and ANOVA results for Coactivation Index, Thigh (%).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	28.3(15.2)	32.3(12.9)	35.6(16.3)	37.3(17.2)	33.4
Old	50.9(19.8)	54.3(17.1)	56.5(15.5)	60.2(12.4)	55.5
Means	39.6	43.3	46.1	48.7	

	F value	p value
Age	16.36	<0.001
Speed	6.97	<0.001
Age x Speed	0.09	0.97

Table A26. Means (SD) and ANOVA results for Coactivation Index Shank (%).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	17.3(6.9)	20.1(7.1)	18.7(6.0)	20.9(6.4)	19.2
Old	20.0(12.5)	21.3(11.8)	20.9(8.6)	21.4(7.9)	20.9
Means	18.6	20.7	19.8	21.2	

	F value	p value
Age	0.29	0.60
Speed	2.65	0.06
Age x Speed	0.54	0.66

Table A27. Means (SD) and ANOVA results for Time of Coactivation, Thigh (%).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	25.4(16.2)	33.4(19.3)	37.3(24.4)	42.8(25.5)	34.7
Old	31.9(10.8)	36.1(9.0)	38.8(9.9)	46.9(13.2)	38.4
Means	28.7	34.7	38.1	44.9	

	F value	p value
Age	0.37	0.36
Speed	28.53	<0.001
Age x Speed	0.7	0.56

Table A28. Means (SD) and ANOVA results for Time of Coactivation, Shank (%).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	52.3(18.0)	60.2(18.9)	61.4(15.5)	69.3(12.9)	60.8
Old	65.2(23.7)	68.9(22.5)	75.5(17.6)	79.0(15.4)	72.2
Means	58.7	64.6	68.5	74.2	

	F value	p value
Age	2.91	0.10
Speed	31.91	<0.001
Age x Speed	0.26	0.29

Table A29. Means (SD) and ANOVA results for Total Coactivation Index (%).

	Speed ($m \cdot s^{-1}$)				
Age group	0.89	1.12	1.34	1.57	Means
Young	45.6(18.0)	52.4(14.3)	54.3(14.2)	58.2(15.2)	52.6
Old	70.9(27.4)	75.7(22.7)	77.3(18.6)	81.6(15.8)	76.4
Means	72.9	63.9	70.7	74.0	

	F value	p value
Age	13.57	<0.001
Speed	7.95	0.001
Age x Speed	0.09	0.97

APPENDIX B

Results for Correlation Analyses between Cost of Walking and Gait Variables

Gait Kinematics

Step width (SW)	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.52	-0.19	0.41	-0.24	0.80	-0.05
Speed 2	0.05	-0.53	0.42	-0.23	0.34	-0.19
Speed 3	0.81	0.07	0.73	-0.10	0.74	-0.06
Speed 4	0.14	-0.42	0.81	-0.07	0.68	-0.08

SW variability	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.46	0.21	0.28	0.31	0.89	0.03
Speed 2	0.90	-0.03	0.38	-0.25	0.24	-0.23
Speed 3	0.74	0.10	0.71	0.11	0.90	0.03
Speed 4	0.64	0.14	0.91	-0.03	0.84	0.04

Stride Length (SL)	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.30	-0.29	0.73	0.10	0.48	-0.14
Speed 2	0.37	-0.25	0.11	0.36	0.98	0.00
Speed 3	0.51	-0.18	0.01	0.65	0.65	0.09
Speed 4	0.45	-0.21	0.70	0.11	0.76	0.06

SL Variability	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.17	0.39	0.27	0.31	0.56	0.11
Speed 2	0.39	-0.25	0.11	0.45	0.34	0.19
Speed 3	0.81	0.07	0.07	0.49	0.03	0.41
Speed 4	0.04	0.55	0.21	0.36	0.06	0.36

Joint Angular Impulse

Ankle Impulse	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.48	-0.20	0.19	0.37	0.87	-0.03
Speed 2	0.86	-0.05	0.18	0.38	0.66	-0.09
Speed 3	0.86	0.05	0.02	0.61	0.54	-0.12
Speed 4	0.75	0.09	0.34	0.28	0.64	-0.09

Knee Impulse	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.02	0.63	0.55	0.18	0.71	-0.07
Speed 2	0.08	0.49	0.09	0.47	0.53	-0.12
Speed 3	0.91	0.03	0.38	0.26	0.12	-0.30
Speed 4	0.04	0.56	0.53	0.18	0.45	-0.15

Hip Impulse	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.09	-0.47	0.07	-0.51	0.94	0.00
Speed 2	0.03	-0.57	0.33	-0.28	0.80	0.05
Speed 3	0.21	-0.36	0.58	-0.16	0.55	0.12
Speed 4	0.22	-0.35	0.89	0.04	0.30	0.20

Joint Work

Work Summed for all joints

Total Work	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.06	-0.52	0.11	-0.45	0.98	0.00
Speed 2	0.22	-0.35	0.26	-0.32	0.65	0.09
Speed 3	0.76	-0.09	0.29	-0.30	0.58	0.11
Speed 4	0.69	0.12	0.71	-0.11	0.19	0.26

Positive Work	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.02	-0.61	0.14	-0.42	0.57	0.11
Speed 2	0.01	-0.65	0.54	-0.18	0.44	0.15
Speed 3	0.09	-0.47	0.35	-0.27	0.55	0.12
Speed 4	0.12	-0.43	0.86	-0.05	0.28	0.21

Negative Work	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.77	-0.08	0.17	0.39	0.31	0.20
Speed 2	0.01	-0.65	0.15	0.40	0.56	0.11
Speed 3	0.05	-0.54	0.36	0.26	0.93	0.00
Speed 4	0.03	-0.57	0.54	0.18	0.51	-0.13

WORK AT EACH JOINT

Work at ankle

Ankle Positive Work

	Young p	Young r	Old p	Old r	All p	All r
Speed 1	0.32	0.29	0.20	0.36	0.06	0.36
Speed 2	0.71	0.11	0.58	0.16	0.18	0.26
Speed 3	0.78	-0.08	0.72	0.10	0.60	0.10
Speed 4	0.58	0.16	0.98	0.00	0.78	-0.05

Ankle Negative Work

	Young p	Young r	Old p	Old r	All p	All r
Speed 1	0.12	0.43	0.73	0.49	0.17	0.27
Speed 2	0.65	0.13	0.11	0.45	0.17	0.26
Speed 3	0.99	0.00	0.30	0.30	0.70	0.08
Speed 4	0.84	-0.05	0.44	0.23	0.79	-0.05

Work at knee

K1

	Young p	Young r	Old p	Old r	All p	All r
Speed 1	0.35	-0.27	0.82	0.07	0.43	0.15
Speed 2	0.13	-0.42	0.30	0.30	0.86	0.03
Speed 3	0.94	0.00	0.45	0.22	0.16	0.28
Speed 4	0.22	-0.35	0.30	0.30	0.60	0.10

K2

	Young p	Young r	Old p	Old r	All p	All r
Speed 1	0.66	0.13	0.66	0.13	0.22	-0.24
Speed 2	0.39	0.25	0.38	0.25	0.34	-0.19
Speed 3	0.26	-0.32	0.65	0.13	0.02	-0.44
Speed 4	0.20	0.36	0.31	0.29	0.70	-0.08

K3

	Young p	Young r	Old p	Old r	All p	All r
Speed 1	0.05	0.54	0.03	-0.57	0.16	-0.27
Speed 2	0.03	0.57	0.13	-0.42	0.30	-0.20
Speed 3	0.67	0.13	0.09	-0.47	0.62	-0.10
Speed 4	0.07	0.50	0.05	-0.53	0.21	-0.24

K4

	Young p	Young r	Old p	Old r	All p	All r
Speed 1	0.03	-0.59	0.12	-0.43	0.22	-0.24
Speed 2	0.01	-0.65	0.34	-0.28	0.08	-0.34
Speed 3	0.01	-0.65	0.83	-0.06	0.05	-0.38
Speed 4	0.001	-0.77	0.64	0.14	0.11	-0.31

Work at Hip

H1

	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.08	-0.48	0.16	-0.39	0.62	0.09
Speed 2	0.01	-0.69	0.97	0.00	0.35	0.18
Speed 3	0.13	-0.42	0.86	-0.05	0.36	0.18
Speed 4	0.10	-0.46	0.34	0.28	0.05	0.38

H2

	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.02	-0.62	0.08	-0.49	0.33	-0.19
Speed 2	0.01	-0.67	0.66	-0.13	0.52	-0.13
Speed 3	0.01	-0.65	0.53	-0.18	0.34	-0.19
Speed 4	0.002	-0.75	0.84	0.05	0.59	-0.10

H3

	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.42	-0.23	0.35	0.27	0.39	0.17
Speed 2	0.59	-0.15	0.51	0.19	0.85	0.03
Speed 3	0.61	-0.15	0.89	0.04	0.82	-0.04
Speed 4	0.65	-0.13	0.63	0.14	0.59	-0.10

Antagonistic Muscle Coactivation

Total Coactivation (CI)

CI total	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.62	-0.14	0.31	0.29	0.01	0.46
Speed 2	0.24	-0.34	0.04	0.55	0.002	0.57
Speed 3	0.67	-0.13	0.18	0.38	0.01	0.48
Speed 4	0.34	-0.28	0.04	0.55	0.01	0.51

Thigh Coactivation

CI Thigh	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.39	-0.25	0.84	0.06	0.04	0.39
Speed 2	0.16	-0.40	0.36	0.26	0.01	0.46
Speed 3	0.62	-0.14	0.29	0.30	0.03	0.41
Speed 4	0.20	-0.37	0.10	0.45	0.03	0.41

TCA Thigh	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.43	-0.23	0.87	0.04	0.82	0.04
Speed 2	0.13	-0.43	0.55	0.18	0.83	0.04
Speed 3	0.65	-0.13	0.31	0.29	0.62	0.10
Speed 4	0.16	-0.40	0.41	0.24	0.46	0.14

Shank Coactivation

CI Shank	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.55	0.18	0.04	0.55	0.04	0.40
Speed 2	0.89	0.04	0.01	0.67	0.03	0.40
Speed 3	0.73	0.10	0.33	0.28	0.20	0.25
Speed 4	0.25	0.33	0.16	0.40	0.11	0.31

TCA Shank	Young		Old		All	
	p	r	p	r	p	r
Speed 1	0.34	0.27	0.02	0.60	0.003	0.54
Speed 2	0.95	0.00	0.01	0.69	0.01	0.47
Speed 3	0.51	0.19	0.05	0.53	0.01	0.51
Speed 4	0.05	0.54	0.04	0.55	0.001	0.61

APPENDIX C

Literature Review

Metabolic cost of walking (C_w), defined as the rate of oxygen consumed during walking per unit distance traveled, has been thoroughly evaluated, generally showing significant differences between populations. C_w in the elderly has been an area of special concern, as this group has been shown to use approximately 20% more oxygen per unit distance traveled during walking than young, regardless of walking speed (Larish, Martin, & Mungiole, 1988; Malatesta, Simar, Dauvilliers, Candau, Borrani, Prefaut & Caillaud, 2003; Martin, Rothstein & Larish, 1992; Mian Thom, Ardigo, Narici, & Minetti, 2006; Ortega & Farley, 2007). Though significant research has examined the causes of these differences, it remains unclear why older adults typically reflect higher C_w. This review will cover a number of factors that affect metabolic C_w with a particular focus on how they relate to age. This review will also examine research literature on muscle coactivation during walking, describing causes of coactivation, ways in which it may be altered, its effects on joints, and specifically how it changes with age.

Section 1: Causes of Increased Cw

Much research has been dedicated to determining why older individuals walk with higher C_w. The literature is generally focused on several factors, including: (1) the overall amount of work done during walking, (2) the distribution of this work across joints, (3) changes in muscular efficiency, (4) kinematic and kinetic variability, and (5) the amount of coactivation of antagonistic muscles about joints.

Work During Walking.

With advancing age come significant changes to the neuromuscular system. One way in which these adaptations are expressed is through changes in gait kinematics, including decreased stride length, increased cadence, decreased preferred cadence, and wider step width (Himann, Cunningham, Rechitzer, & Paterson, 1988; Grabiner, Biswas, & Grabiner, 2001; Rosengren, McAuley & Mihalko, 1998). Based on these changes in gait kinematics, it has been suggested that older individuals may perform larger amounts of work than young adults while walking. Using both segmental and center of mass-based energetic models, Mian et al. (2006) contrasted internal and external mechanical work done by older and young adults during treadmill walking. They found no differences between ages in either total work or external work (based on movements of subject's center of mass). Internal work (based on movement of segments with respect to the body's center of mass) of the upper body was higher in elderly (67%), however its small contribution to total work was not enough to cause significant differences in overall work completed. All other work measures showed no differences between age groups. Authors suggested that the increased internal work done by arms in the aged may have been an artificial effect due to unnaturally rigid arm movements of elderly subjects while walking on the treadmill. Some have argued that estimating work through joint power analyses may be superior to segmental approaches because it provides a more direct indication of muscular contributions to movement (Robertson & Winter, 1980; Umberger & Martin, 2007). Previous studies have been conducted to determine the joint moments associated with walking in young and old adults. Results show that although the distributions of joint torques are different between young and older adults, the total work done to support the body during stance, or support torque, was similar between young and old (Hortobagyi & DeVita, 2000; Winter, Patla, Frank, & Walt, 1990).

Simple inverted pendulum models have predicted that large amounts of work may be necessary in the period during double support when the body moves from one inverted pendulum cycle to the next. During this interval, defined as the transition period, two main processes occur. First, the front and back feet redirect the center of mass' vertical velocity from downwards to up. Secondly, the center of mass medio-lateral (M-L) velocity is directed medially, toward the opposite limb. During this period, negative work is done by the front foot slowing (horizontally) and lifting (vertically) the center of mass, and positive work is done by the back foot to increase center of mass horizontal velocity. These opposing forces from the front and back feet change the center of mass' horizontal velocity little, and thus go undetected with center of mass based work models.

As may be expected, transition work is highly regulated by kinematics such as stride length and width. Specifically, transition work in the anterior-posterior (A-P) direction increases with the 4^{th} power of step length, and M-L transition work increases with the square of step width. Previous studies have shown that older adults walk with increased step width, as well as increased time in double support. It was hypothesized by Ortega and Farley (2007) that due to these kinematic variances with age, and non-linear associations between step width/length with cost, old walkers may show increases in transition work. Investigators determined the individual limb transition work in young and old subjects by integrating individual limb power output with respect to time during periods when power was positive. Interestingly, transition work was lower in elderly. This was due to smaller fluctuations in anterior-posterior velocities in elderly during the double support phase. The older subjects in this study did not show greater stance widths or double support times, potentially due to their relatively high level of fitness. Menz, Lord, and Fitzpatrick (2003) also observed smaller changes in trunk velocity during walking in frail elderly in comparison to young adults. These reductions in center of mass acceleration may be due to the decreased plantar flexion push off experienced in elderly (DeVita & Hortobagyi, 2000). Alternatively, reductions in acceleration may be an adaptation adopted to increase stability during walking. These studies seem to show a consensus that though there is a redistribution of work within lower limb joints, the total work completed during walking is similar between young and old.

Redistribution of Work within Joints.

Though the total work during walking has been shown to be similar for old and young adults, recent research suggests there are some age-related differences in how that work is accomplished. DeVita and Hortobagyi (2000) showed that at 1.48 $m \cdot s^{-1}$, healthy older adults showed a 279% increase in average work at the hip, as well as 39 and 29% reductions in average work at the knee and ankle respectively. These changes in joint work occurred primarily during the stance phase of walking, and have been shown to be consistent across a range of walking speeds (Silder, Heiderscheit, & Thelen, 2008). Savelberg, Verdijk, Willems, and Meijer (2007) compared the joint moments of endurance trained old adults (65 years) and healthy untrained older adults (70 years). This study showed that the redistribution of joint torques during walking is present even among aerobically trained elderly. The trained group reflected significantly higher hip moments during walking than the untrained elderly subjects, and their ankle joint moments were similar to the untrained group. Several effects of proximal redistribution of lower extremity joint moments on C_w have been proposed. One potential effect relates to the differing fiber types of muscles surrounding lower limb joints. Previous studies have shown that musculature with a higher proportion of type I fibers is more efficient at producing force (Coyle, Sidossis, Horowitz, & Beltz, 1992; Horowitz, Sidossis, & Coyle, 1994). As the hip produces a higher percent of total joint torque, reduced muscular efficiency of hip extensors (with respect to ankle plantar flexors) may affect C_w. Indeed, hip extensor muscles (biceps femoris and gluteus maximus) have a smaller percent type I fibers (66.9 and 52.4% respectively) than plantar flexors (medial gastrocnemious – 50.8%; soleus – 89.0%) (Johnson, Polgar, Weightman, & Appleton, 1973). As older adults shift the distribution of work production from the ankle musculature to the less efficient hip muscles, an increase in cost may occur. A second possible explanation of the higher C_w observed for older adults relates to the crouched walking posture seen in some older adults. Exaggerated hip flexion through the gait cycle could put the hip extensors at a mechanical disadvantage, decreasing their effectiveness and thus increasing C_w (Biewener, 1989). Unfortunately, these speculations have not been thoroughly evaluated, and thus potential effects have not been determined.

The decrease in preferred walking speed commonly reported for the elderly has also been proposed as a potential factor in higher C_w. When plotted as a function of walking speed, C_w shows a U-shaped curve for both young and old (Larish et al., 1988). Martin et al. (1992) showed that both old and young individuals minimized their C_w at a walking speed near 1.34 $m \cdot s^{-1}$. Testing sedentary and active, young and old individuals, this study showed all four groups' preferred walking speed to deviate from most economical. In all groups, the C_w at preferred walking speed still lied along the minimum plateau of the U-shaped curve, and C_w at preferred speed was at most 3% larger than the C_w at the most economical speed. The C_w of old subjects was ~10% larger than young at all speeds below 1.57 $m \cdot s^{-1}$. In this study, both young and old preferred walking speeds deviated from most economical speed, however the increases in C_w due to preferred speeds deviating from economical walking speeds were consistent across groups, and much smaller than the differences in C_w between young and old. This implies the changes in C_w due to decreased preferred walking speed in some elderly probably have little or no influence on the higher C_w seen in old.

Muscular Efficiency.

Another component of aging in humans is the alteration of the structure and physiology of muscle (Conley, Esselman, Jubrias, Cress, Inglin, Mogadam, & Schoene, 2000; Kent-Braun, Ng, & Young (2000); Narici, Maganaris, Reeves, & Capodaglio, 2003; for reviews, see McGibbon, 2003; Hortobagyi & DeVita, 2006; Larsson, Yu, Hook, Ramamurthy, Marx, & Pircher, 2001). These changes, including decreases in the CSA (cross sectional area), maximal force production, and contractile tissue within muscles occur more in the lower extremities (Ly & Handelsman, 2002; Simoneau, Martin, & Hoecke, 2005). Further, changes in the internal structure of musculature have been reported with age, such as a decrease in specific tension (force production per CSA) (Larrson, Li, & Frontera, 1997; Ochala, Frontera, Dorer, Van Hoecke, & Krivickas, 2007; Frontera, Suh., Krivickas, Hughes, Goldstein, & Roubenoff, 2000). Larsson et al. (1997) studied the force production across age in single skeletal muscle fibers. The specific tension of old individuals (73-81 years) was 0.18 $N \cdot mm^{-2}$, whereas young subjects showed a significantly higher value, 0.25 $N \cdot mm^{-2}$ ($p<0.001$). Two of the four older individuals tested were highly physically active. These individuals showed an average specific tension of 0.21 $N \cdot mm^{-2}$, significantly lower than young, though not different than untrained elderly. More recently, Ochala et al. (2007) reported decreases in the specific tension of both type I and type IIa muscle fibers in old in relation to young (25 and 33% reductions, respectively). These decreases in specific tension at the cellular level have been contributed to a smaller number of strongly bound cross-bridges, as well as reduced force-generating capacity per cross-bridge (Ochala et al., 2007). In addition, decreases in the ratio of contractile to connective tissue have been shown on the whole muscle level. Using magnetic resonance imaging on the tibialis anterior, Kent-Braun et al. (2000) demonstrated larger amounts of connective tissue and fat in the muscle of old subjects. Both men and women showed 9.6 and 7.9% increases respectively in the non-contractile component of muscle with age. Research has also shown that both the fiber length and pennation angle are reduced with age (Narici et al., 2003), alterations which could affect the force production per sarcomere in old.

Reduced muscle quality (Kent-Braun et al., 2000) and force production per fiber (Larrson et al., 1997), as well as the changing architecture of muscle cells (Narici et al., 2003) may require the activation of more surrounding muscle fibers in order to produce the same desired force. This increase in recruitment of muscle fibers would also require more energy and thus increased metabolic cost for a given force output. In addition, elder individuals will need to activate a larger percent of their musculature and thus a larger portion of their total force producing capacity in order to maintain the same force. Indeed, elder subjects have been shown to require as much as two times as much relative force (percentage of maximal joint moments) about the lower limbs to complete activities of daily living such as stair climbing and standing from a seated position (Hortobagyi, Mizelle, Beam, & DeVita, 2003).

Training has been shown to correct the degradation of many of the effects age has on muscular properties, including force production, CSA, specific force, non-contractile content, and endurance (Reeves, Narici, & Maganaris, 2004; Korhonen, Cristea, Alen, Hakkinen, Sipila, Mero, Viitasalo, Larrson, & Suominen, 2006; Kent-Braun et al., 2000) (For a complete review on this subject, see Blazevich, 2006). With this in mind, Mian, Thom, Ardigo, Narici, and Minetti (2007) hypothesized that these increases in functionality that come with training might also help to bring elderly C_w nearer to young. Old subjects were put through a 12 month physical conditioning program including aerobic, balance, and resistance training in an attempt to counteract the age related physiologic decrements that may affect C_w. After the training period, subjects did show increases in functionality, including 21% increase in knee extension strength, 30% increase in single leg balance, and a 6% decrease in the 6 minute walking test. Interestingly, this intervention had no effect on the C_w of subjects. A recent study completed by Thomas, DeVito, and Macaluso (2007), did however show training to effect C_w. This study used body weight unloading (BWU) to allow subjects to walk at a maximal speed for extended periods of time. During the first week of training, subjects maximal walking speed with 40% BWU was determined to be 1.69 $m \cdot s^{-1}$. Over the following six weeks, walking speed was slowly increased while maintaining 40% BWU. The second six weeks held speed constant at the highest speed reached in the first six weeks (1.82 $m \cdot s^{-1}$), while BWU was reduced systematically to 10%. At the end of the 12 week period, subjects were walking at a speed significantly higher then their original maximum walking speed, with only 10% BWU. In addition to the increases in maximum walking speed, C_w was shown to be significantly reduced in the trained group at slow, preferred, and fast walking speeds. Though the factors potentially associated with this reduction in Cw were not directly measures, authors speculated that "gait stability offered by the harness system during speed walking on the treadmill might have helped the older participatnts to "relearn" to walk with improved gross motor efficiency" (pg. 1602). These results imply training studies may be able to reduce C_w, however more studies should be carried out to determine what interventions are best suited to produce changes in C_w.

Stride Variability.

Healthy elderly subjects have been shown to exhibit variability of step width that is approximately 20% higher than young subjects (Grabiner et al., 2001; Owings & Grabiner, 2004a). Other measures of kinematic variability during gait, such as step length and step time, have also been shown to be increased in older healthy populations (Malatesta, Simar, Dauvilliers, Candau, Borrani, Prefaut, & Caulliud, 2003; Grabiner et al., 2001), however other studies have shown no difference in variability with age (Gabell & Nayak, 1984; Owings & Grabiner, 2004a; Stolze, Friedrich, Steinauer, & Vieregge, 2000).

Stride width variability can be characterized as deviations medially and laterally from preferred stride width. Recent studies have shown preferred stride width to be most economical, with manipulations both medially and laterally to induce substantial increases in C_w (Allor, Pivarnik, Sam, & Perkins, 2000; Donelan, Kram, & Kuo, 2001). If one varies laterally, he or she will need to produce a larger muscular force in order to re-direct the center of mass back towards the mid-line. When one deviates medially, there will be extra muscular force necessary to move the leg laterally around the stance leg. For these reasons, even though subjects may show similar average step widths and mediolateral transition work, increased step width variability may still exact a metabolic cost during gait. By manipulating step width, Donelan et al. (2001) measured the relationship in young individuals between step width and metabolic cost, showing that C_w is related to the square of step width. Due to this non-linear association between variables, authors noted that increased step width variability should lead to increases in metabolic cost of walking even when average step widths are the same.

To our knowledge, only one study has compared gait variability and Cw in young and old. Malatesta et al. (2003) determined that though both stride time variability and Cw were approximately 40 and 22%, respectively, larger in elder adults with respect to young, these two factors were not correlated with each other. More recently, Owings and Grabiner (2004b) showed that step width variability may be a better measure when discriminating between young and old individuals. This study showed differences in variability in step width to be larger (approximately 20% larger in old) than variability in the sagittal plane such as step time or step length (approximately 6 and 11% larger in old, respectively) (Owings and Grabiner, 2004a). However, to our knowledge, no direct comparisons have been made between stride width variability and metabolic cost.

Though too much stride width variation in elderly may be detrimental to C_w, too little may also be problematic. Maki (1997) showed that elderly subjects with a history of falls demonstrated a slight but significant decrease in standard deviation of stride width with relation to non-fallers (0.022 vs. 0.0236, respectively). It was speculated that the decrease in step width variability may be a sign that there is a decrease in the adaptability of the control system, increasing the likelihood of a fall (Lipsitz, 2004). It is interesting to note, however, that increases in the variation of other kinematic variables including velocity, step length, double support time, lateral sway and hip/knee extension angles at push-off have all been shown to be associated with an increased risk of falls (Maki, 1997; Barak, Wagenaar, & Holt, 2006; Hausdorff, Edelberg, Cudkowicz, Singh, & Wei, 1997). Based on these results, it seems as if some variability may be a positive trait, however too much may result in elevated Cw.

Section 2: Coactivation of Antagonistic Muscles During Walking

Coactivation is defined as simultaneous activation of agonist and antagonistic muscles about a joint. Coactivation of opposing muscles is a common feature of many well learned movement patterns, including walking, and is a necessary part of stable locomotion. However, its presence represents wastefulness during movement. Due to the opposing torques generated by agonist and antagonistic muscles, individuals with increased coactivation would require more torque production from agonist muscles to produce the same net torque. In the following sections, we will discuss the causes of coactivation in the elderly, how it may be altered, its effects on joints and movement, and finally, how it is affected by age.

Causes of Elevated Coactivation in the Elderly.

Several potential causes of coactivation have been proposed, both voluntary and involuntary in nature. One mechanism described is the decrease in antagonistic inhibition. Kido, Tanaka, and Stein (2004) determined the amount of reciprocal inhibition occurring in subjects from 22 – 82 years of age. In muscles surrounding the ankle (soleus and tibialis anterior) there was an inverse relationship between inhibition of antagonistic activation and age in both standing in walking. Another potential cause of coactivation in old is the cortical and subcortical re-organization of the brain. With age comes neural degradation. The re-organization that occurs in response to this degradation includes the re-innervation of orphaned muscle fibers by remaining motor units, reducing the fine control of muscles. With this reduction of fine control, an overlap of agonist and antagonistic activation muscular activation is more likely to occur. Further, Sailer, Dichgans, and Gerloff (2000) showed older individuals to have higher overall brain activity during metronome paced finger movements, specifically in the brains areas of movement planning and initiation. It was hypothesized that the increased brain activation initiated agonistic musculature and additionally spread to nearby antagonistic muscles, causing elevated activation of both sets of muscles. For a full review of both neural inhibition and neural re-organization, see Hortobagyi and DeVita (2006).

Yet another potential cause of coactivation could be an increase in overall muscular activation. In Cerebral Palsy patients for example, involuntary increases in muscular activation due to overactive muscular signaling cause significantly larger coactivation levels (Unnithan, Dowling, Frost, & Bar-Or, 1996). Elderly have also been shown to exhibit much high muscular activity during daily tasks than young (~1.8 fold), leading to larger amounts of coactivation (Hortobagyi et al., 2003). Voluntary muscular contraction may also induce increases in coactivation as well. When subjects show uncertainty during a task, they may exhibit increased coactivation (De Luca & Mambrito, 1987; Carolan & Cafarelli, 1992). In elderly subjects, hesitation and uncertainty during a task may lead to a more tense posture, and thus increased coactivation. This adaptation, though wasteful in terms of net joint moment, may have several benefits to elderly populations, such as decreased variability in movement, as well as increased joint and whole-body stability.

Ways to Alter Coactivation.

Several mechanisms have been suggested which may alter coactivation. Specifically, speed of movement, practice of a task, and resistance training have all been shown to have significant effects on coactivation.

Speed Affects on Coactivation.

Several studies have been completed determining effects of speed of movement on coactivation (Table C-1). Studies focused specifically on single joint movements emphasize the importance of the type of contraction (eccentric or concentric) in question. Several studies showed that about the knee, increases in speed during *eccentric* contractions do not seem to produce increases in coactivation (Amiridis, Martin, Morlon, Martin, Cometti, Pousson, & Van Hoecke, 1996; Hubley-Kozey & Earl, 2000; Bassa, Patikas, & Kotzamanidis, 2005). In contrast, coactivation during *concentric* exercises about the knee increases substantially with speed in the majority of studies (Hagood, Solomonow, Baratta, Zhou, D'Ambrosia, 1990; Kellis & Baltzopoulos, 1998; Bassa et al., 2005). Coactivation during concentric exercises is also somewhat variable, however, as Amiridis et al. (1996) showed no change in coactivation during concentric contractions across speeds. The disparity between coactivation during eccentric and concentric actions may be a factor of joints needing a mechanism to reduce velocity towards the end of joint range of motion. When the task is concentric in nature, the antagonists are called upon to decrease speed during towards the end of movement. During concentric actions, increases in angular velocity would require larger antagonistic activation in order to slow movement. Indeed, when movement is broken into stages, coactivation has been shown to occur primarily at the end of the range of motion for most (Hortobagyi & DeVita, 2000; Aagaard, Simonsen, Andersen, Magnusson, Bojsen-Moller, & Dyhre-Poulsen, 2000) though not all (Hagood et al., 1990) studies. During eccentric contractions, activation of the antagonist would work to increase speed of movement towards the end of the range of motion. Though activation of antagonistic musculature may still be necessary for joint stabilization, it is not contributing to the reduced velocity towards the end of movement at any speed and therefore may not be as necessary during eccentric contractions.

The effect of speed of movement on coactivation during multi-joint movement show consistent increases with rising speed. Results from Mian et al. (2006) show trends of increased coactivation in both old and young subjects with increases in walking speed (0.8-1.6 $m \cdot s^{-1}$). During downward stair stepping, Larsen, Puggaard, Hamalainen, and Aagaard (2008) also showed larger coactivation at a maximal stepping speed than at preferred. Finally, unpublished data from Penn State University on coactivation changes with speed in young subjects also supports these results, incorporating coactivation about both knee and ankle joints.

It is important when determining coactivation during walking to consider the method in which coactivation is determined. Many coactivation analyses incorporate the amplitude of EMG (electromyography) as well as the time in which the two muscles (agonist and antagonist) are active. Analyses of this sort are determined first by plotting the linear envelopes of both agonists and antagonists over time. The coactivation is described as the area under the smaller EMG linear envelope. This value is often divided by the total area under both antagonist and antagonistic muscles, as described by Winter (1990). This normalization removes the effects of differing EMG amplitudes between subjects. This normalization should be cautioned when making adjustments to speed, as it may under represent the true changes in coactivation that are occurring with speed. To determine effects of speed on coactivation, a more appropriate analysis may involve only determining the time in which both agonist and antagonistic muscles are active, as this represents a measure of coactivation without the confounding variable of EMG amplitude. This measure was determined in Mian et al. (2006), while Larsen measured coactivation determined by both time of overlap and the amplitude of EMG. Analyses within a speed, across subjects should include amplitude, and be appropriately normalized as described by Winter (1990). It is important to include the amplitude of activation in these situations, as the reduction in net joint moment and thus “wastefulness” is affected by the strength of the antagonist’s activation. Based on these results, speed of movement does seem to have a significant effect on coactivation, and care should be taken to account for these changes if speed is to be manipulated.

Table C-1: Speed effects on coactivation during concentric (A), eccentric (B), and multi-joint (C) actions.

(A) Concentric actions

Author	Joint	Antagonistic musculature	Speed ($°·s^{-1}$)	Change in coactivation with increased speed
Hagood et al. (1990)	Knee	Extensors/Flexors	15 – 240	Increase / Increase
Kellis & Baltzopoulos (1998)	Knee	Extensors/Flexors	30 – 150	Increase / Increase
Bassa et al. (2005)	Knee	Extensors/Flexors	45, 90, 180	Increase / Increase
Amiridis et al. (1996)	Knee	Extensors	120 – 300	No Change

(B) Eccentric actions

Author	Joint	Antagonistic musculature	Speed ($°·s^{-1}$)	Change in coactivation with increased speed
Kellis & Baltzopoulos (1998)	Knee	Flexors	30 – 150	Increase
Kellis & Baltzopoulos (1998)	Knee	Extensors	30 – 150	No Change
Amiridis et al. (1996)	Knee	Extensors	120 – 300	No Change
Bassa et al. (2005)	Knee	Extensors/Flexors	45, 90, 180	No Change / No Change
Hubley-Kozey & Earl (2000)	Ankle	Plantar flexors	30 – 150	No Change

(C) Dynamic Multi-joint actions

Author	Joint	Task	Speed (m·s^{-1})	Change in coactivation with increased speed
Mian et al. (2006)	Knee	Walking	0.8-1.6	Increase
Larsen et al. (2008)	Knee	Stair stepping	"Maximum and Preferred speeds"	Increase

Practice Effects on Coactivation.

When individuals attempt novel tasks, they are often more tense, and may produce a movement which changes as one familiarizes themselves with the necessary muscular activation. More specifically, changes seem to be occurring in agonist and antagonistic activation with practice (Carolan & Cafarelli, 1992). Several studies have probed the issue of coactivation during practice, beginning with Person (1958). During filing tasks, coactivation was shown to decrease after a practice session. However during more forceful movements, such as hammering a nail, coactivation was not shown to decrease with practice. A more recent study by Moore and Marteniuk (1986) substantiated these results. Coactivation was measured during ballistic flexion and extension movements about the elbow. The movements were set at either 200 or 500 ms, and a learning effect was demonstrated by the variability and accuracy of arm placement at the end of the task. Over 400 forearm extension movements produced on four consecutive days, EMG amplitude of antagonistic musculature and coactivation were significantly reduced. Several other studies evaluated the coactivation of muscles about the elbow during ballistic movements of varying velocity, showing mixed results. Darling and Cooke (1987) demonstrated that in elbow flexion and extension tasks practice induced increased coactivation with earlier onset of antagonistic muscles. Similar results were also shown by Gabriel and Boucher (1998). Lagasse (1979) however described decreased coactivation and later onset of antagonistic musculature. These three studies did not hold speed constant. In fact, the measure of learning for these studies was a decrease in the time necessary to reach the target. This speed manipulation may have had an effect on coactivation, confounding inferences into changes in coactivation with practice. As speed was changing with practice, so was the necessary muscular activation. With an increase in speed, an earlier onset of antagonistic contraction (and potentially greater amplitude) must be present in order to slow the movement towards the target. Recently, a study by Calder and Gabriel (2007) looked the effect of practice on coactivation in maximal isometric contractions about the elbow. Fifteen elbow flexion contractions were produced over three days to avoid strength training. The coactivation over these trials alternated significantly above and below baseline levels. This interesting response implies no directional change in coactivation with practice.

Based on the three most applicable studies (Moore & Marteniuk, 1986; Person, 1958; Calder & Gabriel, 2007), it seems as if the effect of practice on coactivation is significant, however task specific, with sub-maximal tasks decreasing coactivation with practice, and actions nearer to maximal activation showing less response to practice. With this in mind, it is important to thoroughly familiarize subjects to novel tasks if coactivation is to be measured.

Training Effects on Coactivation.

Strength training may also cause changes in the coactivation seen at a joint. In addition to physical changes occurring with muscular training such as increased CSA, training induces neurological changes to the way in which muscles contract, such as increased discharge rates of motor units. In addition it is possible that the muscle timing becomes more effective by better coordinating the activity agonist and antagonistic muscles. Simoneau, Martin, Porter, and Van Hoecke (2006) reported training effects at the ankle in healthy elder subjects (78±3 years). Plantar flexors were trained with the use of therabands as well as a calf-raise machine. After the six month training period, coactivation was measured during both dorsi and plantar flexion.

During dorsiflexion, activation of the triceps surae complex (antagonists) was shown to significantly decrease with training. During plantar flexion, however, the tibialis anterior (antagonist) was shown to increase activation after the training period. Also important, the net maximal torque was shown to increase after training in both dorsi and plantar flexion. These results were attributed to the fact that the triceps surae complex is able to produce a much larger amount of force than the tibialis anterior. Small decreases in activation of this complex during dorsiflexion would cause significant changes in the overall dorsiflexion torque. Indeed, this mechanism seemed to be adopted in order to increase total net torque. Small increases in activation of dorsiflexors during plantar flexion, however, would have a minimal effect on the net plantar flexion torque. Increases in coactivation with training in this scenario would have little negative effect on net joint torque, while some beneficial effects such as joint stabilization and joint force distribution would be experienced. In contrast to these results, Ochala, Lambertz, Van Hoecke, and Pousson, (2005) conducted a 24 week exercise routine, where older adults participated in several exercises such as walking, marching, and resistance training (calf raises and leg press). Interestingly, authors demonstrated no change in coactivation during plantar flexion in elderly. Patten and Kamen (2000) described reductions in coactivation with training only in old. Intervention consisted of a two week period of "force modulation training" three times a week. Force modulation training was defined as a training program designed specifically to improve the accurate control of force during ankle dorsiflexion. After this two week training period, subjects participated in a four week retention period, training only one time per week. Elder individuals showed significant decreases in coactivation (plantar flexion activation) during dorsiflexion. Young subjects, however, showed no change in coactivation with training. Interestingly, the coactivation decreases in elder subjects over time persisted, and in fact continued to decrease through the four week retention phase. Other studies focused on the knee joint also showed decreases in coactivation with training. Carolan and Cafarelli (1992) described changes in coactivation in knee extensions of young healthy subjects. After an eight week training period, coactivation of the hamstrings was shown to decrease by approximately 20%. Measurements were made on a weekly basis, and the changes in coactivation were seen only after the first week of training. A more recent study looked at both middle aged and old individuals, with subjects in this study trained through leg press and bilateral knee extension exercises for a period of six months (Hakkinen, Kallinen, Izquierdo, Jokelainen, Lassila, Malkia, Kraemer, Newton, & Alen, 1998). Training protocol integrated high resistance bouts, as well as ballistic "explosive power" exercises (fast movements at approximately 20% MVC). After training, male and female elder individuals showed significant decreases in coactivation during isometric knee extension, though the reductions were very small in elder male subjects (24 - 21% over 6 months). Coactivation of middle aged individuals had shown no significant change. Coactivation was also measured during maximal isokinetic leg extension, and during static jumps. During isokinetic extension, only the elderly women expressed significant decreases in coactivation, though trends toward reduction of coactivation were seen in all groups. No differences in coactivation were seen after training in the static jump task. The lack of significance in coactivation during static jump may be due to the specificity of the training. Though there was an emphasis on explosive power during training sessions, lower body training consisted of leg press and bilateral knee extensions, potentially limiting carry-over into the jumping task. As with Carolan and Cafarelli (1992), the changes in coactivation that were observed occurred after the first period of training (two months) only. Finally, Amiridis et al. (1996) compared the coactivation at the ankle joint of highly skilled high jumpers to sedentary subjects. During isometric knee extension, the highly trained subjects showed significantly less coactivation, as well as increased force production.

Reeves et al. (2005) also measured coactivation in elderly subjects during both concentric and eccentric isokinetic knee extension. After a 14 week knee extension/leg press training program, there were no changes in coactivation. One potential rationale for the disparity between these results and Hakkinen et al. (1998) may be the training programs which were used. Reeves et al. (2005) used leg extension and leg-press exercise as training interventions. Hakkinen et al. (1998) utilized the same exercises, however an explosive power component was added to their training.

As noted previously, novel tasks may produce larger coactivation levels, and a reduction in coactivation may occur with practice. Though many of the studies mentioned went through substantial familiarization procedures in order to habituate subjects to the testing procedure, there could be a learning response that is occurring in addition to neurological adaptations seen with resistance training. In order to reduce the practice effects, Hakkinen et al. (1998) brought subjects in one month prior to the start of the training period to test strength and coactivation and familiarize them to the testing procedure. There was no change in the coactivation between this measurement and the coactivation which was measured at the start of the training program. Simoneau et al. (2006) trained subjects with different motions than were tested. Training procedure in this study consisted of therabands and dynamic exercise, while coactivation was measured while subjects performed maximal isometric contractions. This variation between the testing and training protocols may be beneficial to reducing the learning effect on coactivation. Carolan and Cafarelli (1992) included a control group which went through testing procedures, however did no strength training. These subjects did see reductions in antagonistic activity presumably due to a learning effect, however trained subjects showed coactivation levels which were significantly lower than controls.

The specific mechanisms responsible for reductions in coactivation with training are illusive, and to our knowledge, no research has been directly specifically at this issue. Though no conclusions have been made, decreases in coactivation with training could be due to neural organization, or more precise timing of muscular activation. Also, reciprocal inhibition has been shown to decrease with age (Kido et al., 2004), and may cause larger coactivation. Future studies should be conducted on the effect of training on pre-synaptic inhibition and antagonistic musculature. It should also be considered that there is a fine line between the “neural adaptations” occurring with training, and skill and practice effects, both of which could result in more effective activation of musculature. Based on the current body of research, it is difficult to say whether the changes in coactivation with training are due primarily to neurological adaptations or practice effects, or whether these causes are innately coupled. However, studies such as Carolan and Carafelli (1992), Simoneau et al. (2006), and Hakkinen et al. (1998) imply that there is at least some neurological effect of resistance training on coactivation.

Effects of Coactivation.

Coactivation has been shown to have significant effects on the way in which people move. For example, when coactivation is occurring, the opposing muscular forces increase joint pressure as well as stiffness of the joint (Hortobagyi & DeVita, 2000). During downward stepping, stiffness, computed as “the ratio of force under the foot and the linear shortening of the limb” (Hortobagyi & DeVita, 2000) was 64% greater in elderly subjects compared to young subjects. Coactivation accounted for 50% of the variance in this stiffness. In addition, two studies determined the amount of coactivation necessary to stabilize the wrist with a varying load applied. A strong linear relationship was found between wrist stiffness and the coactivation of muscles around that joint (Milner, Cloutier, Leger, Franklin, 1995; De Serres & Milner, 1991). Finally, with EMG input from healthy elder subjects, an IEMG-moment model designed by Kellis and Baltzopoulos (1999) showed that during eccentric and concentric motions, the compressive and posterior shear joint forces at the knee are significantly higher with inclusion of the antagonist muscle.

Another affect of coactivation is a decrease in movement variability. Several studies have shown that elderly have greater variability of movement and force production than young, due primarily to the neuromuscular changes that occur with age (Cooke, Brown, & Cunninghan, 1989; Darling, Cooke, & Brown, 1989; Seidler-Dobrin, He, & Stelmach, 1998; Seidler & Stelmach, 1995). As elderly age, motor neurons die, and the orphaned muscle fibers must then become re-innervated by other surrounding motor neurons. This decreases the fine motor control of each motor unit, potentially causing increased variability. However, Seidler-Dobrin et al. (1998) reported similar variability scores between young and old during a simple elbow flexion task. It was hypothesized that during this simple motion, elderly subjects were able to co-contract their muscles to a greater degree in order to maintain variability levels similar to young. To determine whether this decrease in variability could have been related to the coactivation, an EMG-based model was created. Using EMG data from normal young and old individuals, this model showed that the increased coactivation in old did in fact decrease both kinematic and kinetic variability. The model also showed that the increased coactivation prevented elderly subjects from accelerating their arm as rapidly as young (Seidler-Dobrin et al., 1998). In a more recent study, a destabilizing mechanism induced variability during a wrist flexion task, and coactivation was then measured. In young individuals, square-wave vibrations (at 10 and 5.5

Hz) were applied to the wrist during slow 20º and 30º flexion movements. It was shown that when destabilizing vibrations were implemented, coactivation was significantly increased both during movements and after movement cessation.

Finally, the activation of both agonists and antagonistic muscles may serve to better distribute intersegmental forces within joints. Baratta, Solomonow, Zhou, Letson, Chuinard, and D'Ambrosia (1988) reported that in addition to maintaining stability within the knee joint during extension, coactivation worked to better distribute forces between the femur and tibia. This distribution decreases the prevalence of point loads, which can damage cartilage and induce serious conditions such as osteoporosis.

Though there are many benefits associated with increased coactivation, it also represents wastefulness during movement and potentially increases C_w. In healthy young individuals, many kinematic variables associated with gait (stride width, length, velocity) are naturally optimized to produce a minimum overall metabolic expenditure (Donelan et al., 2001, Donelan, Kram, & Kuo, 2002; Umberger & Martin 2006; Martin et al., 1992). If it is assumed that healthy young individuals express near optimum coactivation levels, the higher levels of coactivation sometimes seen in old may be occurring in order to counteract some of the neurological and physical changes that occur with age.

Effects of Age on Coactivation.

As previously stated, age has a profound effect on the way in which we move and these neurological and physical changes may have an effect on coactivation level. Studies on this topic include single and multi-joint tasks, both isometric and dynamic in nature. Results of this literature have been somewhat variable (Table C-2).

Coactivation with Age in Single Joint Movements.

Adaptations in coactivation with age have been studied extensively, with the majority of research focused on single joint tests. Both static (isometric) and dynamic (concentric and eccentric) motions have been analyzed in this manner, showing a variety of results. Tracy and Enoka (2002) showed that during isometric, concentric, and eccentric muscular contractions about the knee, coactivation increased with age. They also reported that the timing of agonist and antagonist muscle activation was not different between age groups. Patten and Kamen (2000) looked at how often antagonistic musculature was activated in association with agonist musculature bursts during concentric dorsi and plantar flexion. It was shown that during agonist bursts, antagonist musculature was activated twice as often in active elderly subjects as in young. Hakkinen et al. (1998) described elderly women to have 12% higher activation of the hamstrings during concentric knee extension than young women. Men however, showed no age effect. Other studies showed old to have significantly higher coactivation than young during static and dynamic contractions about both the knee and elbow (Izquierdo, Ibanez, Gorostiaga, Garrues, Zuniga, Anton, & Larrion, 1999; Macaluso, Nimmo, Foster, Cockburn, McMillan, & DeVito, 2002; Klein, Rice, & Marsh, 2001).

Several other studies, similar in design, have shown no change in coactivation with age. Klass, Baudry, and Duchateau (2005) for example studied isometric, concentric, and eccentric ankle dorsiflexion, reporting no change in the level of coactivation with age. No change in coactivation with age has also been reported during isometric ankle plantar flexion (Morse et al., 2004) and isometric knee flexion (Macaluso et al., 2002). Pousson, Lepers, and Van Hoecke (2001) further described both static and dynamic flexion tasks about the elbow, with no change in the coactivation between young and old subjects. Simoneau et al. (2005) studied isometric ankle plantar and dorsiflexion. Though coactivation during dorsiflexion was similar between young and old subjects, coactivation during plantar flexion was actually *lower* in elder subjects, a difference that corresponded to the difference in voluntary maximal torque between groups. Authors proposed that coactivation levels may be closely tied to the overall torque output and how it relates to the amount of joint stability necessary. Since elder individual's plantar flexion torque was smaller than young individuals, they needed less stabilization at that joint, and thus less coactivation. For a further review on this topic, see Klass et al. (2007).

The variance in response of coactivation with age seems to contradict the hypothesis that the main cause of coactivation in older individuals is neural decrements. If neural adaptations occurring with age were a significant factor in coactivation levels, these results should be seen across activation types as well as joints. However this does not seem to be the case. Though the age and training level varied somewhat between elder subjects in the mentioned studies, a pattern describing the observed results is not apparent.

Coactivation and Age during Multi-joint Activities.

Coactivation levels in elderly during multi-joint activity have received less attention than single joint movements, however results have been much more consistent. Several studies have been completed describing coactivation about the knee during downward stepping tasks (Hortobagyi & DeVita, 2000; Hortobagyi et al., 2003, Larsen, Puggaard, Hamalainen, & Aagaard, 2008). Hortobagyi and DeVita (2000) noticed that for sedentary young and old individuals, activation of antagonistic musculature during downward stepping (knee flexors and ankle dorsiflexors) was 120% greater in elderly. Further, this study showed a direct relationship between the coactivation and stiffness about the knee joint. In this task, the agonistic musculature (knee extensors and ankle plantar flexors) were primarily performing eccentric motions, with antagonistic musculature performing concentric contraction. The cause of higher coactivation in old was not specifically determined in this study, however authors proposed that the higher antagonistic activation, and thus coactivation seen in old was at least partly voluntary in nature; due to the fears and risks associated with the task. Individuals compensated for decreases in muscular force production (Hortobagyi, Zheng, Weidner, Lambert, Westbrook, & Houmard, 1995), slower rate of tension development (Thelen, Schultz, Alexander, & Ashton-Miller, 1996), and miscued limb positioning (Skinner, Barrack, & Cook, 1984) by increasing the leg stiffness (and potentially stability) through voluntary coactivation of lower limbs (Hortobagyi & Devita, 2000). Hortobagyi et al. (2003) looked at the coactivation and relative effort (% EMG during MVC) present about the knee during three activities of daily living including stair ascent, stair decent, and standing from a seated position. This study determined that elderly individuals complete daily activities at approximately two times higher relative muscular activation than young. In addition, activation of antagonistic musculature was raised 96, 39, and 41% during stair ascent, decent, and standing from a chair, respectively. These increases in coactivation were proposed as major contributors to the increased relative effort during the tasks. Larsen et al. (2008) also studied the coactivation in elderly and young during stair stepping, determining coactivation about both the knee and ankle joints. Both young and old subjects participated in exercises at least once a week and were considered moderately trained. In stair ascent and decent at a standardized speed (cycle frequency of 35 min^{-1}), coactivation was shown to be elevated in elderly about the knee only, and further was only higher during the loading phase. When subjects ascended stairs at a freely chosen speed, elderly chose to step at a rate 20% lower than young. At these chosen speeds, stair ascent showed thigh and calf coactivation to be larger in old (16.8 and 12.3%, respectively), while decent showed higher coactivation about the knee only, during both stance and loading phases (19.2 and 35.2%). Only one study determined the coactivation present about the knee during level walking. Mian et al. (2006) looked at differences in coactivation between old and young individuals at four walking speeds. It was shown that coactivation was higher in the elderly group, and that these increases in coactivation were directly related to C_w at one speed (1.39 $m \cdot s^{-1}$). This study looked only at the coactivation present about the knee. Also, both young and old subjects habituated to the treadmill task for only 15-20 minutes. Recent studies have shown that treadmill accommodation may take longer for elderly subjects, and 20 minutes may not be enough to elicit "normal" walking patterns. Wass, Taylor, and Matsas (2005) conducted a treadmill habituation study looking at elder individuals over a 15 minute walking period. These subjects, average age 74, were in reasonable health and successfully completed the revised physical activity readiness questionnaire (PAR-Q) before participating in the study. Over the fifteen minute walking period, subjects showed no significant step to step differences after the fourth minute; however subjects never reached overground kinematic values. Further, authors stated that "Most of the older adults did not appear comfortable while walking on the treadmill" (p.77), and two thirds of the sixteen subjects needed to use handrail support for the duration of the test. Results from this study suggest that either some elderly subjects do not fully habituate to treadmill walking (achieve overground kinematic values), or a longer habituation time (>15 minutes) is necessary for this process to occur.

The inconsistent response of age on coactivation during single joint movements is intriguing, and implies neurological adaptations to be minor causes of increased coactivation in elderly. If elderly do exhibit neural degradation leading to coactivation, this response should be consistent during both single and multi-joint motions. The results of these studies are instead mixed, with several single joint studies showing no change in coactivation with age. Responses of coactivation to age during multi-joint movements are much more consistent, with old generally showing higher coactivation levels than young. These results, however, may be due to factors other than, or in addition to neurological adaptation. For example, the more complex multi-joint movements may necessitate additional coactivation to reduce instability. Alternatively, coactivation during multi-joint tasks could be partly voluntary in nature. Elderly subjects feeling uncomfortable during, for example, a stair stepping task, may voluntarily increase the stiffness and contraction of the lower limbs (Hortobagyi & Devita, 2000). It has been shown that descending stairs requires approximately 79% MVC in elderly subjects, whereas young individuals complete this activity at around 33% of max (Hortobagyi et al., 2003). In order to determine how coactivation is related to C_w, it may be beneficial to determine coactivation during multi-joint tasks which are more familiar and less taxing to older subjects. To our knowledge, Mian et al (2006) is the only study to look specifically at coactivation during such a task. Results showed higher coactivation in the elderly population than young during overground walking, however the lack of sufficient treadmill habituation could potentially have skewed these results.

Table C-2: Age effects on coactivation during single joint (A) and multi-joint (B) movements.

(A)

Author	Static / Dynamic	Joint	Action	Subjects	Activity level	Coactivation
Izquierdo et al. (1999)	Static	Knee	Flexion	O-65; Y-42 (M)	"habitually physically active"	O=Y
Izquierdo et al. (1999)	Static	Knee	Extension	O-65; Y-42 (M)	"habitually physically active"	**O>Y**
Macaluso A (2002)	Static	Knee	Flexion	O-69; Y-22 (F)	"Recreationally active"	O=Y
Macaluso A (2002)	Static	Knee	Extension	O-69; Y-22 (F)	"Recreationally active"	**O>Y**
Simoneau et al. (2005)	Static	Ankle	Dorsiflexion	O-77; Y-23 (M)	< 2x/wk	O=Y
Simoneau et al. (2005)	Static	Ankle	Plantar flexion	O-77; Y-23 (M)	< 2x/wk	O<Y
Morse CI et al. (2004)	Static	Ankle	Plantar flexion	O-73; Y-24 (M)	"physically active, no sports"	O=Y
Patten & Kamen C (2000)	Static	Ankle	Dorsiflexion	O-71; Y-20 (M/F)	"regular vigorous activity"	**O>Y**
Klein et al (2001)	Static	Elbow	Flexion	O-81; Y-22 (M)	"Sedentary"	**O>Y**
Klein et al (2001)	Static	Elbow	Extension	O-81; Y-22 (M)	"Sedentary"	**O>Y**
Pousson M et al. (2001)	Static	Elbow	Flexion	O-70; Y-21 (M/F)	Not defined	O=Y
Izquierdo et al. (1999)	Dynamic	Knee	Extension	O-65; Y-42 (M)	"habitually physically active"	**O>Y**
Tracy BL & Enoka (2002)	Dynamic	Knee	Extension	O-71; Y-22 (M/F)	< 3hr/wk moderate activity	**O>Y**
Klass et al. (2005)	Dynamic	Ankle	Dorsiflexion	O-78; Y-25 (M/F)	"moderate physical activity"	O=Y
Pousson M et al. (2001)	Dynamic	Elbow	Flexion	O-70; Y-21 (M/F)	Not defined	O=Y

(B)

Author	Task	Joint	Subjects	Activity level	Coactivation
Hakkinen et al (1998)	Jumping	Knee	O-70; Y-42 (M/F)	"habitually physically active", 2-3 times per week	**O>Y**
Hortobagyi & DeVita (2000)	Downward stepping	Knee	O-69; Y-20 (F)	Inactive; One or fewer times per week	**O>Y**
Hortobagyi & DeVita (2003)	Stair Stepping	Knee	O-74; Y-22 (M/F)	Inactive; One or fewer times per week	**O>Y**
Larsen et al (2008)	Stair Stepping	Knee	O-72; Y-25 (M/F)	One time per week "moderately trained"	**O>Y**
Larsen et al (2008)	Stair Stepping	Ankle	O-72; Y-25 (M/F)	One time per week "moderately trained'	O=Y
Mian et al. (2006)	Walking	Knee	O-74; Y-26 (M/F)	Functionally (standing/stair climbing) "free from frailty"	**O>Y**

Summary.

Elderly individuals use more energy than young to walk a certain distance. The causes of this phenomenon have been extensively researched; however, a complete answer has not been reached. Based on the literature, four potential mechanisms have been suggested to influence C_w in elderly, including the distribution of this work across joints, changes in muscular efficiency, kinematic variability, and the amount of coactivation of antagonistic muscles about joints. Future research is necessary to determine the relative contributions of these factors to the increased cost in elderly.

Coactivation is a necessary and beneficial mechanism occurring in order to maintain stability and distribute forces within joints. However, coactivation also represents wastefulness in movements, and excessive coactivation may contribute to increased metabolic cost of movement. Recent studies have shown that during multi-joint movements, old individuals show higher coactivation than young. These studies focused on difficult tasks such as stair climbing, potentially eliciting voluntary increases in coactivation. The high level of muscular activation necessary during these tasks may limit the meaningfulness of these data to level ground walking. Associations between coactivation and C_w during gait have been proposed, but not well established in the literature. Future studies should include both cost and coactivation measures to better determine the effects of coactivation on C_w, specifically in old.