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Effects of Age, Power Output, and Cadence on Energy Cost and Lower Limb Antagonist Muscle Co-Activation during Cycling

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Abstract

It is unknown if higher antagonist muscle co-activation is a factor contributing to higher energy cost of cycling in older adults. We determined how age, power output, and cadence affect metabolic cost and lower extremity antagonist muscle co-activation during submaximal cycling. Thirteen young and 12 older male cyclists completed 6-minute trials at four power output-cadence conditions (75W-60rpm, 75W-90rpm, 125W-60rpm, and 125W-90rpm) while electromyography (EMG) and oxygen consumption were measured. Knee and ankle co-activation indices were calculated using vastus lateralis, biceps femoris, gastrocnemius, and tibialis anterior EMG data. Net rate of energy cost of cycling was higher in older compared to young cyclists at 125W (p=0.002) and at 90rpm (p=0.026). No age-related differences were observed in the magnitude or duration of co-activation about the knee or ankle (p>0.05). Our results indicated knee and ankle co-activation is not a substantive factor contributing to higher energy cost of cycling in older adults.

Keywords

submaximal, metabolic cost, ergometer

Disciplines

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Comments

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Keywords: submaximal, metabolic cost, ergometer

Higher aerobic demand or energy cost of locomotion is a commonly reported change associated with advancing age (Hortobágyi, Finch, Solnik, Rider, & DeVita, 2011; Mian, Thom, Ardigo, Narici, & Minetti, 2006; Peterson & Martin, 2010). In addition, higher co-activation of antagonist muscles has been suggested as a cause of inefficient movement (Winter, 2009) and a contributing factor to higher energy cost of locomotion in older adults (Hortobágyi et al., 2011; Mian et al., 2006; Peterson & Martin, 2010). When contrasting the responses of young and older adults during walking, Peterson and Martin (2010) reported a 23% higher rate of energy cost and a 47% higher total co-activation of flexor and extensor muscles about the ankle and knee joints for older participants. They concluded co-activation of antagonists helps explain the higher cost of walking typically observed in older adults.

With respect to cycling, most research on energy cost and lower extremity electromyography (EMG) has focused on young adults. Numerous studies have examined the effects of external power output and cadence on aerobic demand or energy cost (Belli & Hintzy, 2002; Bigland-Ritchie & Woods, 1974; Chavarren & Calbet, 1999; Gaesser & Brooks, 1975; Marsh & Martin, 1993; Samozino, Horvais, & Hintzy, 2006). Power output and cadence influences on lower extremity muscle contributions during submaximal cycling have also been studied in young adults by several investigators (Baum & Li, 2003; MacIntosh, Neptune, & Horton, 2000; Marsh & Martin, 1995). The effects of power output on energy cost and muscular effort are reasonably straightforward. Both increase systematically as power output increases (Baum & Li, 2003; Bigland-Ritchie & Woods, 1974; Chavarren & Calbet, 1999; Gaesser & Brooks, 1975). Energy cost was approximately 31% higher (Samozino et al., 2006) and lower extremity EMG was 53% higher (Bigland-Ritchie & Woods, 1974) when young adults cycled at an external power output of 125 W compared to 75 W. Similarly, multiple investigators have shown both aerobic demand (Belli & Hintzy, 2002; Chavarren & Calbet, 1999; Gaesser & Brooks, 1975; Marsh & Martin, 1993, 1995; Samozino et al., 2006) and lower extremity EMG (Baum & Li, 2003; MacIntosh et al., 2000; Marsh & Martin, 1995) increase as a function of cadence. For example, aerobic demand was 15% higher (Belli & Hintzy, 2002; Chavarren & Calbet, 1999) and lower extremity EMG was 10% higher (Marsh & Martin, 1995) when young adults cycled at 90 rpm compared to 60 rpm.

Comparisons of energy cost of cycling for older and young adults as a function of external power output and cadence have received limited attention. Moreover, it is not known whether older and young individuals respond similarly to changes in power output and cadence. Bell and Ferguson (2009) compared aerobic demand between older and young adults but did so using relative intensity (i.e. 75% of ventilatory threshold), which was 53 W lower for older compared to young adults. Hopker and colleagues (2013) also examined age-related differences in cycling cost, but they allowed participants to self-select pedaling cadences. Preferred cycling cadences have been reported to be 10-15 rpm lower for older compared to young adults across a wide range of power outputs (Sacchetti, Lenti, Di Palumbo, & De Vito, 2010). Since external power output and cadence are both confounding variables that independently affect metabolic cost and lower extremity muscular excitation (Belli & Hintzy, 2002; Bigland-Ritchie & Woods, 1974; Chavarren & Calbet, 1999; Gaesser & Brooks, 1975; Marsh & Martin, 1993; Samozino et al., 2006), we saw the need for further research using absolute power outputs and cadences. Age effects on energy cost and co-activation of antagonist muscles during walking have been explored reasonably thoroughly, but there is a need to explore whether these effects are similar for other submaximal tasks such as cycling, specifically when external power output and cadence conditions are matched.

Our purpose was to determine the effects of age, power output, and cadence on energy cost and lower extremity muscular excitation during submaximal, steady-state cycling. We hypothesized that: 1) older adults have higher net energy costs and higher levels of co-activation of lower extremity antagonists compared to young adults under matched power output and cadence conditions; 2) age-related differences in energy cost and co-activation of lower extremity antagonists are greater at higher cadences and; 3) age-related differences in energy cost and lower extremity co-activation are also greater at higher power outputs.

Methods

Thirteen young and 13 older healthy, community dwelling men who incorporate cycling into their recreational or exercise activities were recruited. Our study was limited to male participants for two primary reasons. First, pilot testing revealed that older men could achieve aerobic steady-state while completing our targeted power output (75 and 125 W) and cadence (60 and 90 rpm) conditions whereas older women were less likely to sustain steady-state. Second, by choosing to recruit only male participants, we anticipated a reduction in variability in our outcome variables and an increase in our statistical power to detect age effects. Based on a statistical power analysis, a total sample size of 20 participants (10 per group) was needed to have a statistical power of 0.8 to detect a medium effect size for two-way interactions among age, power output, and cadence at an alpha level of 0.05. Individuals who cycle regularly were specifically recruited because of the desire for participants to be familiar with the cycling task and have the capacity to ride at moderate intensity power output-cadence combinations. Moreover, EMG activity of lower limb muscles have been reported to be different for experienced and non-experienced cyclists (Marsh & Martin, 1995). To be classified as a cyclist, a person needed to have ridden a bicycle for transportation, recreation, and/or exercise (indoors and/or outdoors) during most weeks in the past year. Exclusion criteria for the study included any conditions that caused pain and/or discomfort during cycling such as rheumatoid arthritis, osteoarthritis, gout, swollen and painful joints, backache, and recent injury to lower limb muscles, bones, and/or joints. In addition, individuals with neurological conditions such as stroke, cerebral palsy, Parkinson's disease, and any other conditions that affect muscle tone and range of motion of the lower limb joints were also excluded. The University Institutional Review Board approved the study design and procedures, and all participants provided written informed consent.

Data Collection

Participants completed two testing sessions. The first session was used to fully accommodate participants to all testing conditions. Data collection for the experimental conditions was completed in the second session. Participants were instructed not to perform any exhaustive lower body exercises 48 hours prior to each session.

Session 1. Participants completed a questionnaire that summarized their general health status, cycling experience, and muscle, orthopedic, and neurological health histories. Body height and weight were measured using a stadiometer and balance scale, respectively. A Lode bicycle ergometer (Lode Excalibur Sport Ergometer, Groningen, Netherlands), which can control power output as cadence varies, was used for all cycling tests. An ergometer set-up procedure was performed to standardize the cycling position of participants because posture is known to affect energy cost (Nordeen-Snyder, 1977), joint ranges of motion, and muscle activation patterns (Sanderson & Amoroso, 2009). Seat height, defined as the distance from the top of the saddle to the top pedal surface along the line of the seat tube, was set at 100% of trochanteric height (Nordeen-Snyder, 1977). Handlebar height was matched to seat height. Seat fore-aft position was adjusted such that when the crank arms were horizontal, a plumb line dropped from the inferior pole of the patella of the more forward leg hung directly over the pedal axle (Silberman, Webner, Collina, & Shiple, 2005). Handlebar fore-aft position was modified to achieve a forward-leaning trunk angle of 20 to 30° from the vertical (Korff, Newstead, van Zandwijk, & Jensen, 2014). Pedal toe clips were used to control foot position on the pedals during the cycling trials.

Participants then performed a 5-minute warm-up ride at low exercise intensity (50 W) using a comfortable cadence. The participant's preferred or self-selected cadences at 75 W and 125 W were then determined in random order. For each power output, participants were instructed to pedal at a cadence they would find comfortable for an extended ride while the cadence monitor remained covered. After two minutes of pedaling, participants were asked to confirm that the pedaling rate was comfortable, at which time the cadence was noted and recorded (Marsh & Martin, 1995).

Participants then practiced each of the four power output-cadence conditions (75 W at 60 rpm, 75 W at 90 rpm, 125 W at 60 rpm, and 125 W at 90 rpm) by riding under each condition for 6 minutes. A rest interval of 3 to 5 minutes separated the conditions. These power outputs were selected based partly on conditions used by Sacchetti et al. (2010) for older competitive cyclists riding at moderate intensity. When riding at 40% and 60% of maximal power output at 60 and 90 rpm, their older cyclists pedaled at power outputs of approximately 100 W and 150 W (Sacchetti et al., 2010). Since participants in our study were recreational and not competitive cyclists we chose moderately lower power outputs of 75 and 125 W. Absolute rather than relative power outputs were utilized in the current study because several previous studies have suggested that power output is a confounding factor that independently affects metabolic cost and lower extremity muscular excitation during cycling (Baum & Li, 2003; Bigland-Ritchie & Woods, 1974; Chavarren & Calbet, 1999; Gaesser & Brooks, 1975; Samozino et al., 2006). Cadences of 60 and

90 rpm were selected because these cadences closely approximate most economical and preferred cadences reported for young adults, respectively (Marsh & Martin, 1993, 1995; Umberger, Gerritsen, & Martin, 2006). These chosen cadences also fell within the range of cadences (i.e. 40-120 rpm) used in previous research investigating mechanical and physiological efficiency in older cyclists (Bell & Ferguson, 2009; Hopker et al., 2013; Sacchetti et al., 2010). Pilot testing of these power output-cadence conditions revealed these conditions were challenging and yet achievable for recreationally active older men.

Session 2. Participants returned to the lab within 3 to 7 days of session 1. The second session was used to capture EMG, oxygen consumption ($\dot{V}O_2$), carbon dioxide production ($\dot{V}CO_2$), and pedal position data while the participant completed 6-minute trials under each of the four randomly ordered power output-cadence conditions.

Prior to participant arrival, the Lode ergometer configuration was adjusted to setup conditions determined for each participant in session 1. Gas analyzers and respiratory flow meters of a MAX-II metabolic cart (Physiodyne Instrument Corporation, Quogue, NY) were calibrated with known gas concentrations (20.99% ± 0.03% O_2 and 5.03% ± 0.03% CO_2) and a 3-liter calibration syringe. Two passive reflective markers were attached to the lateral face of the left pedal and were subsequently used to distinguish individual crank revolutions using a Vicon motion analysis system (Centennial, CO, USA).

Participants were first prepped for EMG assessments. Skin preparation and identification of EMG electrode placement sites over the vastus lateralis (VL), biceps femoris (BF), medial gastrocnemius (GAST), and tibialis anterior (TA) of the left leg were performed according to standards derived from surface EMG for non-invasive assessment of muscles (SENIAM) guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). Participants sat quietly for 20-

minutes as skin areas over each muscle were shaved, gently abraded, and cleaned with alcohol to reduce surface impedance and ensure good electrode contact and stable attachment. A Delsys wireless EMG system (Delsys Inc., Boston, MA, USA) was used to capture EMG data. Differential surface EMG electrodes with parallel bar arrangement (contact area 10 x 1 mm, 10 mm inter-electrode distance) were aligned in parallel with muscle fibers (Winter, 2009). The electrodes were secured to the leg with athletic tape underwrap to minimize motion artifacts. The electrodes had a built-in gain of 1000, a common mode rejection ratio > 80 decibels at 60 Hz, and band pass filtering of 20-450 Hz. An additional amplification factor of 1.25 was also applied to the pre-amplified signals using the Vicon Nexus Software.

Participants were then fitted with the headgear, mouthpiece, and nose clip and sat quietly on the ergometer for 10-minutes. Metabolic data were collected during last 5-minutes of quiet sitting to obtain resting $\dot{V}O_2$ and $\dot{V}CO_2$ for 30-second windows. Resting values were computed as the averages for the last 3 minutes of the 5-minute collection.

Participants completed a 5-minute warm-up ride at 50 W using a self-selected cadence, followed by 6-minute trials under each of our four randomly ordered power output-cadence conditions. During each trial, metabolic data were sampled at 100 Hz and were collected and averaged over 30-second windows throughout the trial and were later averaged over the last 3 minutes of each 6-minute collection. The Vicon motion analysis system was used to sample EMG and pedal marker position data synchronously at 2000 Hz and 100 Hz, respectively, for 60 seconds during minute 5 of each trial. Participants sat quietly on the ergometer between power output-cadence conditions, usually for 3-5 minutes, until $\dot{V}O_2$ and $\dot{V}CO_2$ returned to near resting levels before beginning the next condition.

Data Analysis

Position data for the markers on the left pedal were digitally filtered at 4 Hz and used to compute coordinates of a virtual marker representing the pedal axle. A crank cycle was defined as the period between sequential top dead center positions of the left crank arm (i.e., peak vertical position of the pedal axle). Data for 15 crank cycles collected during minute 5 of each trial were included in subsequent co-activation analyses.

The raw EMG data for the cycling conditions were checked and corrected for DC bias, bandpass filtered at 30-300 Hz using a 6^{th} order Butterworth filter, full wave rectified, and then low pass filtered at 50 Hz using a 2^{nd} order Butterworth filter to create a linear envelope. For the most challenging experimental condition (i.e. 125W-90 rpm), peak EMG value was identified for each of 15 crank cycles for each muscle group. These peak EMG values were subsequently averaged to get an overall mean peak EMG value for each muscle group, which were then used to normalize EMG data for the other three cycling conditions (75W-60 rpm, 75W-90 rpm, and 125W-60 rpm). Our EMG normalization method is similar to that of Candotti et al. (2009) who normalized their EMG data to the peak value from one of their submaximal conditions when they examined differences in energy cost and lower extremity antagonist co-activation between triathletes and cyclists at different cadences. We also considered using maximal voluntary isometric contractions for EMG normalization, but this method has been criticized for not being suitable for EMG normalization during cycling (Hug & Dorel, 2009). Furthermore, recent studies showed that normalizing EMG amplitudes using peak EMG values from a submaximal condition during cycling demonstrated higher reliability and sensitivity in EMG outcomes for lower extremity muscles compared to normalization using a maximal isometric voluntary contraction (Albertus-Kajee, Tucker, Derman, & Lambert, 2010; Sinclair et al., 2015).

The onsets and offsets of normalized EMG activity of each muscle during cycling trials were determined using a Teager-Kaiser energy operator (TKEO), which reduces background noise and enhances signal-to-noise ratio (Solnik, Rider, Steinweg, DeVita, & Hortobagyi, 2010). The DC bias-corrected EMG data were processed using the TKEO to obtain the energy level of the signal, which was then full wave rectified and low pass filtered at 50 Hz using a 2nd order Butterworth filter to create a linear envelope. From the linear envelope, a baseline TKEO energy level for each muscle was computed as the minimum 30-sample moving average taken over the entire duration of a trial. The standard deviation for the minimum moving average was then computed. A threshold equal to the baseline TKEO energy level plus 60 standard deviations for each muscle group was computed and used to establish when each muscle was "on" or "off" during each crank cycle.

Average EMG amplitude was computed over 15 crank cycles for each muscle. Co-activation occurred when both muscles of each antagonist pair (i.e., VL-BF and GAST-TA) were "on." The area of overlap between the muscles during co-activation was calculated in 0.5 ms increments over each crank cycle. A co-activation index (CI) (Peterson & Martin, 2010; Winter, 2009) was defined as follows:

$$CI(\%) = 2 \times \left(\frac{\int \min(EMG_{M1}, EMG_{M2})}{\int EMG_{M1} + \int EMG_{M2}}\right) \times 100 \quad (1)$$

where $\int \min(EMG_{M1}, EMG_{M2})$ represents area of overlap of the antagonist muscle pair (M1 and M2), and $\int EMG_{M1}$ and $\int EMG_{M2}$ are activation areas for each individual muscle of the pair. Co-activation indices were averaged over 15 crank cycles for each power output-cadence condition for each participant. The duration of co-activation of the antagonist pair were also computed as a percentage of the crank cycle.

$\dot{V}O_2$ and $\dot{V}CO_2$ (ml·min^{-1}) collected during each of the 6-minute test conditions were averaged over the final 3 minutes. Each participant's respiratory exchange ratio (RER) was monitored to ensure RER remained less than 1.0 (Kenney, Wilmore, & Costill, 2011). In addition, variation in $\dot{V}O_2$ was examined for the final 3 minutes for each trial. Results showed $\dot{V}O_2$ varied by 2 ml·kg^{-1}·min^{-1} or less for all participants and conditions. These results indicated the cycling trials closely approximated aerobic, steady state conditions. Gross rate of energy cost (*Gross* $\dot{E}$, J·s^{-1}) for the last three minutes of each condition was estimated using average $\dot{V}O_2$ and $\dot{V}CO_2$ as described by Adamczyk and colleagues (2006):

$$Gross\ \dot{E} = 16.48\,\dot{V}O_2 + 4.48\,\dot{V}CO_2 \qquad (2)$$

where oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) are in ml·s^{-1}. Resting rate of energy cost was computed from the $\dot{V}O_2$ and $\dot{V}CO_2$ averaged over the final 3 minutes of the 5-minute resting condition. The net rate of energy cost was then determined by subtracting resting rate from gross rate for each condition.

Statistical Assessment

Three-factor mixed model ANOVAs with repeated measures on power output and cadence were used to test the effects of age, power output, and cadence and their interactions on net rate of energy cost, average muscle excitation, and knee and ankle co-activation durations and indices. For main effect comparisons, the alpha level was set a 0.05 and for post-hoc comparisons a Bonferroni adjustment was made to guard against type-1 error. Effect sizes (Cohen's f) were computed for primary dependent variables. Small, medium, and large effect sizes correspond to Cohen's f-values of 0.10, 0.25, and 0.40, respectively (Cohen, 1988). All statistical procedures were performed using SPSS (Version 23).

Results

One older participant was excluded from the study because his RER values exceeded 1.0 during cycling for the 125-60 rpm and 125 W-90 rpm conditions. Thus, outcomes are reported for 12 older and 13 young participants. The participant selection process successfully produced two groups that did not differ significantly on mass, height, and average duration of weekly cycling. Older adults had lower preferred cycling cadences than young adults at both 75 W (p=0.150, Cohen's f=0.31) and 125 W (p=0.212, Cohen's f=0.27). Although these differences reflected medium effect sizes, they failed to reach statistical significance (Table 1). For the experimental trials, both older and young adults closely adhered to the assigned cadences. Group mean cadences deviated from nominal values by no more than 1.6 rpm and standard deviations were less than 1.4 rpm for all conditions.

Net rate of Energy Cost

Magnitudes and trends for $\dot{V}O_2$ and $Gross\ \dot{E}$ of our participants were comparable with those previously reported (Chavarren & Calbet, 1999; Gaesser & Brooks, 1975; Samozino et al., 2006). Older adults' net rate of energy cost was about 8% higher than that of young adults when averaged across power output and cadence conditions (age main effect, p=0.002, Cohen's f=0.74, figure 1). Increases in power output and cadence also had significant and strong effects on average net energy cost (power output main effect, p<0.001; cadence main effect, p<0.001), but these effects were greater for our older compared to young participants. As power output increased from 75 W to 125 W, the average increase in net rate of energy cost for older adults (198 $J \cdot s^{-1}$) was greater than that for young participants (174 $J \cdot s^{-1}$; age x power output interaction, p=0.002, Cohen's f=0.74). Similarly, older adults showed a greater increase in net energy cost as cadence increased from 60 to 90 rpm (92 $J \cdot s^{-1}$) compared to young participants (66 $J \cdot s^{-1}$; age x cadence interaction, p=0.026, Cohen's f=0.50).

Lower extremity EMG response

Table 2 summarizes average normalized EMG responses of older and young participants for the VL, BF, GAST, and TA as a function of power output and cadence. Both VL and BF showed significantly higher average excitation for 125 W compared to 75 W cycling (VL: +39%, p<0.001, Cohen's f=2.47; BF: +41%, p<0.001, Cohen's f=1.49). There were no significant differences between older and young adults for either the VL or BF, and cadence had no significant effect on VL and BF excitation levels. An age x cadence interaction was observed for the BF (p=0.004, Cohen's f=0.68); older adults showed an increase in BF excitation with the shift from 60 to 90 rpm, whereas young adults showed a modest decline in excitation with the cadence increase.

GAST excitation was 47% higher for older compared to young adults (p<0.001, Cohen's f=0.93) and 63% higher for the 90 rpm compared to 60 rpm (p<0.001, Cohen's f=1.75), but was not significantly affected by power output. The GAST for older adults showed a greater sensitivity to the increase in cadence than that for young participants (age x cadence interaction, p=0.030, Cohen's f=0.68). For the TA, both power output and cadence affected TA response, but there was no systematic difference between older and young adults. TA excitation was 11% higher for the 125 W condition compared to 75 W (p=0.001, Cohen's f=0.77). This increase was attributed more to the increase in excitation shown by the young participants with the shift in power output from 75 to 125 W (age x power output interaction, p=0.020, Cohen's f=0.52). Compared to power output, cadence had a much greater effect on the TA; excitation was 89% greater when participants pedaled at 90 rpm compared to 60 rpm (p<0.001, Cohen's f=4.55).

Figure 2 shows ensemble averages of VL, BF, GAST, and TA EMG excitation profiles and periods of co-activation (shaded area) for the 125 W-90 rpm condition. These EMG profiles were comparable with those reported previously (Baum & Li, 2003; Marsh & Martin, 1995). Results for knee and ankle co-activation indices and durations of co-activation as a percent of the crank cycle are highlighted in figures 3 and 4, respectively. Across the experimental conditions, no statistically significant differences were observed between older and young participants for the knee or ankle co-activation indices or duration of co-activation, even though these indicators of co-activation systematically increased with the shift in power output from 75 to 125 W. For example, knee co-activation index and duration of co-activation about the knee were 8% and 15% higher, respectively, when pedaling at 125 W compared to 75 W (power output main effect; knee co-activation index: $p<0.001$, Cohen's f=0.96; knee co-activation duration: $p<0.001$, Cohen's f=0.91). Similarly at 125 W, ankle co-activation index was 12% higher (power output main effect, $p = 0.023$, Cohen's f=0.51) and duration of ankle co-activation was 16% higher (power main effect, $p = 0.041$, Cohen's f=0.45) compared to 75W conditions. Of all the co-activation-related dependent variables, changes in cadence affected only the duration of ankle co-activation. Duration of co-activation of antagonists about the ankle increased by 31% in both young and older individuals when pedaling at 90 compared to 60 rpm (cadence main effect, $p<0.001$, Cohen's f=0.77).

Discussion

We predicted older adults would have both higher rates of energy consumption and higher levels of co-activation of lower extremity antagonists compared to young adults when cycling under the same power output and cadence conditions. Our results for rate of energy cost partially supported our hypothesis. In our study, older adults exhibited 8% higher metabolic cost of cycling compared to young adults (figure 1). If our data were to be expressed as net efficiencies, older adults had 8% lower efficiency than young participants across the experimental conditions (21.9% vs. 23.9% for older and young, respectively). Other researchers (Bell & Ferguson, 2009; Hopker et al., 2013; Peiffer, Abbiss, Sultana, Bernard, & Brisswalter, 2016; Sacchetti et al., 2010), who did not control either power output or cadence, reported similar trends indicating older adults had 4-18% poorer efficiency compared to young adults during submaximal cycling. Our age-related differences in metabolic cost of cycling are lower compared to those reported for walking. For example, older adults had 19-31% higher metabolic cost of walking over a range of moderate to fast walking speeds compared to young adults (Hortobágyi et al., 2011; Mian et al., 2006; Peterson & Martin, 2010).

Older participants in our study did not show the hypothesized higher levels of co-activation of lower extremity antagonist muscles across the experimental conditions compared to young subjects (figures 3 and 4). We examined both the duration and magnitude of co-activation using antagonist pairs of muscles about the knee and ankle. None of our co-activation dependent variables showed higher levels of co-activation for our older participants. We have compared our outcomes for lower limb antagonist co-activation with analogous research in walking since we are aware of no other research that has investigated these contrasts in cycling. A wide range of age-related differences have been reported during walking (Hortobágyi et al., 2011; Hortobágyi et al., 2009; Mian et al., 2006; Peterson & Martin, 2010), all showing older adults with higher levels of co-activation (31-150% higher). These differences in lower limb co-activation between cycling and walking are perhaps explained by several factors. Ergometer cycling requires little weight bearing and involves limited mechanical and physiological demands for maintaining balance compared to walking. It is also a highly constrained motor task. In addition, to examine age effects *per se* on lower limb muscular excitation and antagonist muscle co-activation, we controlled external power output, cadence, cycling posture, and participant cycling experience in our study. Previous research has shown that each of these factors directly and substantially affect lower extremity muscular excitation during cycling (Baum & Li, 2003; Bigland-Ritchie & Woods, 1974; MacIntosh et al., 2000; Marsh & Martin, 1995; Sanderson & Amoroso, 2009). Thus, with the degree of control that we built into our experimental design, it is not surprising that we failed to observe age-related differences in antagonist muscle co-activation during ergometer cycling.

Age-related lower extremity antagonist co-activation differences have been limited to walking and stair negotiation. For example, Hortobágyi and DeVita (2006), whose research on lower extremity co-activation has primarily focused on walking and stair descent, indicated higher co-activation observed in older adults may represent a neural strategy "to stiffen their joints and stabilize motor output by reducing movement variability in an effort to compensate for reduced muscle strength and increased joint laxity" (p. 29). Schmitz et al. (2009) suggested similar implications for higher antagonist co-activation when they noted higher co-activation reflects a strategy of stiffening the limb during single support and thereby contributes to reduced push-off power during walking. Finally, higher co-activation levels reflected by older adults have been associated with higher rates of energy cost during locomotion activities such as walking on a level surface (Mian et al., 2006; Peterson & Martin, 2010), walking on an inclined surface (Hortobágyi et al., 2011), and stair negotiation (Hortobágyi & DeVita, 2000). In summary, for weight-bearing activities, higher co-activation could be viewed as an age-related compensatory strategy that comes with a cost. Higher antagonist co-activation increases joint stiffness and thereby may enhance stability during these weight bearing activities, but contributes to higher energy cost (Hortobágyi et al., 2011; Mian et al., 2006; Peterson & Martin, 2010) and poorer efficiency (Winter, 2009).

In our second hypothesis, we predicted age-related differences in energy cost and co-activation of antagonists are greater at higher cadences, which was partially supported by our results. Differences in energy cost between our older and young participants were greater at 90 rpm (figures 1). A large effect size was observed for this age x cadence interaction. Our metabolic cost results are in agreement with Sacchetti and colleagues (2010), who reported gross efficiency was lower at higher cadences for both older and young cyclists. This cadence effect, however, was more pronounced in older participants. Bell and Ferguson (2009) also reported lower gross efficiency in older compared to young participants but they failed to find an age x cadence interaction as Sacchetti et al. and we found. We feel the inconsistency may be attributed to differences in experimental control of power output; Bell and Ferguson examined the effect of cadence (45, 60, 75, and 90 rpm) at a relative power output (75% of the ventilatory threshold) rather than absolute power output in our design.

In contrast to our energy cost results, no age x cadence interaction for lower extremity antagonist co-activation was observed, indicating our older participants did not show a greater sensitivity to the increase in cadence from 60 to 90 rpm (figures 3 and 4). Our results are consistent with Candotti et al (2009), who also did not observe an increase in knee co-activation with increases in cadence in young adults. In summary, for our older participants, cycling at the higher cadence was metabolically more demanding but it was not accompanied by greater co-activation of antagonists about the knee and ankle compared to young individuals. Unlike results of studies on walking and stair negotiation, our results indicate that older adults may not be relying on increased antagonist co-activation to provide joint stabilization to compensate for increased joint laxity and reduced muscular strength when cycling at higher cadences (Hagood, Solomonow, Baratta, Zhou, & D'Ambrosia, 1990; Hortobágyi & DeVita, 2006).

The physiological and mechanical demands of the cycling task increase with power output. As expected, we found that net rate of energy cost and excitation of the VL, BF, GAST, and TA were greater in both young and older adults when pedaling at a higher power output (125 W vs. 75 W). We also observed that the ankle and knee co-activation indices and duration of co-activation as a percent of crank cycle were higher at 125 W in both young and older adults. Our results are consistent with those of Momeni and Faghri (2015) who reported greater activation of the rectus femoris and biceps femoris and longer duration of antagonist co-activation about the knee in both young and middle-aged individuals at higher power output (0 W vs. 100 W) during semi-reclined cycling.

We also hypothesized that age-related differences in energy cost and co-activation of antagonists are greater at higher power outputs (i.e., an age x power output interaction). Our net energy cost results supported our hypothesis. Older adults experienced a greater increase in net energy cost than young participants for the 50 W increase in power output (198 vs. 174 $J \cdot s^{-1}$). Our results are partially consistent with Hopker et al. (2013) who found cycling efficiency was 9% lower in older compared to young trained cyclists at 150 W but not different at 100 W. Our lower extremity co-activation results did not show an age x power output interaction as we had predicted. Momeni and Faghri (2015) also failed to find an age x power output interaction for duration of antagonist co-activation about the knee in young and middle-aged individuals during semi-reclined cycling. In summary, our results show that cycling at higher power outputs is metabolically more demanding for older adults. In addition, knee and ankle co-activation increased with power output, but this systematic power output manipulation did not exaggerate age-related differences in lower limb co-activation (figure 3 and 4).

Other age-related adaptations in muscular properties and function have also been suggested to affect energy cost and muscular effort of submaximal tasks. With advancing age these adaptations may include lower muscle mass and strength (Malatesta et al., 2003; Martin, Rothstein, & Larish, 1992), reduction in size, number, and discharge rates of motor units (Kallio et al., 2012), and reduction in number and size of muscles fibers, especially fast-twitch type-II fibers (Nilwik et al., 2013). These changes likely make pedaling at a given power output and cadence more challenging for older adults, particularly at higher power outputs and cadences, and contribute to poorer efficiency when generating a given external power output. .

In conclusion, older adults had higher rates of energy cost than young adults for ergometer cycling. These age-related differences in energy cost are exaggerated by systematic increases in power output and/or cadence. Older adults did not have higher antagonist muscle co-activation in the lower extremity compared to young adults. Neither cadence nor power output further affected differences in lower extremity antagonist co-activation between young and older adults. These findings indicate that higher levels of co-activation of antagonist muscles do not appear to be a contributing factor to higher rates of net energy cost of cycling in older adults.

“Effects of Age, Power Output, and Cadence on Energy Cost and Lower Limb Antagonist Muscle Co-Activation during Cycling”
by Buddhadev HH, Martin PE
Journal of Aging and Physical Activity

References

Adamczyk, P. G., Collins, S. H., & Kuo, A. D. (2006). The advantages of a rolling foot in human walking. *Journal of Experimental Biology, 209*(20), 3953-3963.

Albertus-Kajee, Y., Tucker, R., Derman, W., & Lambert, M. (2010). Alternative methods of normalising EMG during cycling. *Journal of Electromyography and Kinesiology, 20*(6), 1036-1043.

Baum, B. S., & Li, L. (2003). Lower extremity muscle activities during cycling are influenced by load and frequency. *Journal of Electromyography and Kinesiology, 13*(2), 181-190.

Bell, M. P., & Ferguson, R. A. (2009). Interaction between muscle temperature and contraction velocity affects mechanical efficiency during moderate-intensity cycling exercise in young and older women. *Journal of Applied Physiology, 107*(3), 763-769.

Belli, A., & Hintzy, F. (2002). Influence of pedalling rate on the energy cost of cycling in humans. *European Journal of Applied Physiology, 88*(1-2), 158-162.

Bigland-Ritchie, B., & Woods, J. J. (1974). Integrated EMG and oxygen uptake during dynamic contractions of human muscles. *Journal of Applied Physiology, 36*(4), 475-479.

Candotti, C. T., Loss, J. F., Bagatini, D., Soares, D. P., da Rocha, E. K., de Oliveira, A. R., & Guimaraes, A. C. (2009). Cocontraction and economy of triathletes and cyclists at different cadences during cycling motion. *Journal of Electromyography and Kinesiology, 19*(5), 915-921.

Chavarren, J., & Calbet, J. A. L. (1999). Cycling efficiency and pedalling frequency in road cyclists. *European Journal of Applied Physiology, 80*(6), 555-563.

Cohen, J. (1988). *Statistical Power Analysis for the Behavioral Sciences* (2nd ed.). Hillsdale, New Jersey: Lawrence Erlbaum Associates.

Gaesser, G. A., & Brooks, G. A. (1975). Muscular efficiency during steady-rate exercise: Effects of speed and work rate. *Journal of Applied Physiology, 38*(6), 1132-1139.

Hagood, S., Solomonow, M., Baratta, R., Zhou, B. H., & D'Ambrosia, R. (1990). The effect of joint velocity on the contribution of the antagonist musculature to knee stiffness and laxity. *American Journal of Sports Medicine, 18*(2), 182-187.

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology, 10*(5), 361-374.

Hopker, J. G., Coleman, D. A., Gregson, H. C., Jobson, S. A., Von der Haar, T., Wiles, J., & Passfield, L. (2013). The influence of training status, age, and muscle fiber type on cycling efficiency and endurance performance. *Journal of Applied Physiology, 115*(5), 723-729.

Hortobágyi, T., & DeVita, P. (2000). Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging. *Journal of Electromyography and Kinesiology, 10*(2), 117-126.

Hortobágyi, T., & DeVita, P. (2006). Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. *Exercise and Sport Sciences Reviews, 34*(1), 29-35.

Hortobágyi, T., Finch, A., Solnik, S., Rider, P., & DeVita, P. (2011). Association between muscle activation and metabolic cost of walking in young and old adults. *Journals of Gerontology Series A Biological Sciences and Medical Sciences, 66*(5), 541-547.

Hortobágyi, T., Solnik, S., Gruber, A., Rider, P., Steinweg, K., Helseth, J., & DeVita, P. (2009). Interaction between age and gait velocity in the amplitude and timing of antagonist muscle coactivation. *Gait and Posture, 29*(4), 558-564.

Hug, F., & Dorel, S. (2009). Electromyographic analysis of pedaling: A review. *Journal of Electromyography and Kinesiology, 19*(2), 182-198.

Kallio, J., Sogaard, K., Avela, J., Komi, P., Selanne, H., & Linnamo, V. (2012). Age-related decreases in motor unit discharge rate and force control during isometric plantar flexion. *Journal of Electromyography and Kinesiology, 22*(6), 983-989.

Kenney, W. L., Wilmore, J., & Costill, D. (2011). *Physiology of Sport and Exercise* (5th ed.). Champaign, Illinois: Human Kinetics.

Korff, T., Newstead, A. H., van Zandwijk, R., & Jensen, J. L. (2014). Age- and activity-related differences in the mechanisms underlying maximal power production in young and older adults. *Journal of Applied Biomechanics, 30*(1), 12-20.

MacIntosh, B. R., Neptune, R. R., & Horton, J. F. (2000). Cadence, power, and muscle activation in cycle ergometry. *Medicine & Science in Sports & Exercise, 32*(7), 1281-1287.

Malatesta, D., Simar, D., Dauvilliers, Y., Candau, R., Borrani, F., Pre´faut, C., & Caillaud, C. (2003). Energy cost of walking and gait instability in healthy 65- and 80-yr-olds. *Journal of Applied Physiology, 95*(6), 2248-2256.

Marsh, A. P., & Martin, P. E. (1993). The association between cycling experience and preferred and most economical cadences. *Medicine & Science in Sports & Exercise, 25*(11), 1269-1274.

Marsh, A. P., & Martin, P. E. (1995). The relationship between cadence and lower extremity EMG in cyclists and noncyclists. *Medicine & Science in Sports & Exercise, 27*(2), 217-225.

Martin, P. E., Rothstein, D. E., & Larish, D. D. (1992). Effects of age and physical activity status on the speed-aerobic demand relationship of walking. *Journal of Applied Physiology, 73*(1), 200-206.

Mian, O. S., Thom, J. M., Ardigo, L. P., Narici, M. V., & Minetti, A. E. (2006). Metabolic cost, mechanical work, and efficiency during walking in young and older men. *Acta Physiologica, 186*(2), 127-139.

Momeni, K., & Faghri, P. D. (2015). Aging and muscle activity patterns during cycling. *International Journal of Physical Medicine and Rehabiliatation, 3*(5), 1-7.

Nilwik, R., Snijders, T., Leenders, M., Groen, B. B., van Kranenburg, J., Verdijk, L. B., & van Loon, L. J. (2013). The decline in skeletal muscle mass with aging is mainly attributed to a reduction in type II muscle fiber size. *Experimental Gerontology, 48*(5), 492-498.

Nordeen-Snyder, K. S. (1977). The effect of bicycle seat height variation upon oxygen consumption and lower limb kinematics. *Medicine and Science in Sports, 9*(2), 113-117.

Peiffer, J., Abbiss, C. R., Sultana, F., Bernard, T., & Brisswalter, J. (2016). Comparison of the influence of age on cycling efficiency and the energy cost of running in well-trained triathletes. *European Journal of Applied Physiology, 116*(1), 195-201.

Peterson, D. S., & Martin, P. E. (2010). Effects of age and walking speed on coactivation and cost of walking in healthy adults. *Gait and Posture, 31*(3), 355-359.

Sacchetti, M., Lenti, M., Di Palumbo, A. S., & De Vito, G. (2010). Different effect of cadence on cycling efficiency between young and older cyclists. *Medicine & Science in Sports & Exercise, 42*(11), 2128-2133.

Samozino, P., Horvais, N., & Hintzy, F. (2006). Interactions between cadence and power output effects on mechanical efficiency during sub maximal cycling exercises. *European Journal of Applied Physiology, 97*(1), 133-139.

Sanderson, D. J., & Amoroso, A. T. (2009). The influence of seat height on the mechanical function of the triceps surae muscles during steady-rate cycling. *Journal of Electromyography and Kinesiology, 19*(6), e465-471.

Schmitz, A., Silder, A., Heiderscheit, B., Mahoney, J., & Thelen, D. G. (2009). Differences in lower-extremity muscular activation during walking between healthy older and young adults. *Journal of Electromyography and Kinesiology, 19*(6), 1085-1091.

Silberman, M. R., Webner, D., Collina, S., & Shiple, B. J. (2005). Road bicycle fit. *Clinical Journal of Sport Medicine, 15*(4), 271-276. doi:

Sinclair, J., Taylor, P. J., Hebron, J., Brooks, D., Hurst, H. T., & Atkins, S. (2015). The reliability of electromyographic normalization methods for cycling analyses. *Journal of Human Kinetics, 46*, 19-27.

Solnik, S., Rider, P., Steinweg, K., DeVita, P., & Hortobagyi, T. (2010). Teager-Kaiser energy operator signal conditioning improves EMG onset detection. *European Journal of Applied Physiology, 110*(3), 489-498. doi:10.1007/s00421-010-1521-8

Umberger, B. R., Gerritsen, K. G. M., & Martin, P. E. (2006). Muscle fiber type effects on energetically optimal cadences in cycling. *Journal of Biomechanics*, *39*(8), 1472-1479.

Winter, D. A. (2009). *Biomechanics and Motor Control of Human Movement* (4th ed.). Hoboken, New Jersey: John Wiley & Sons.

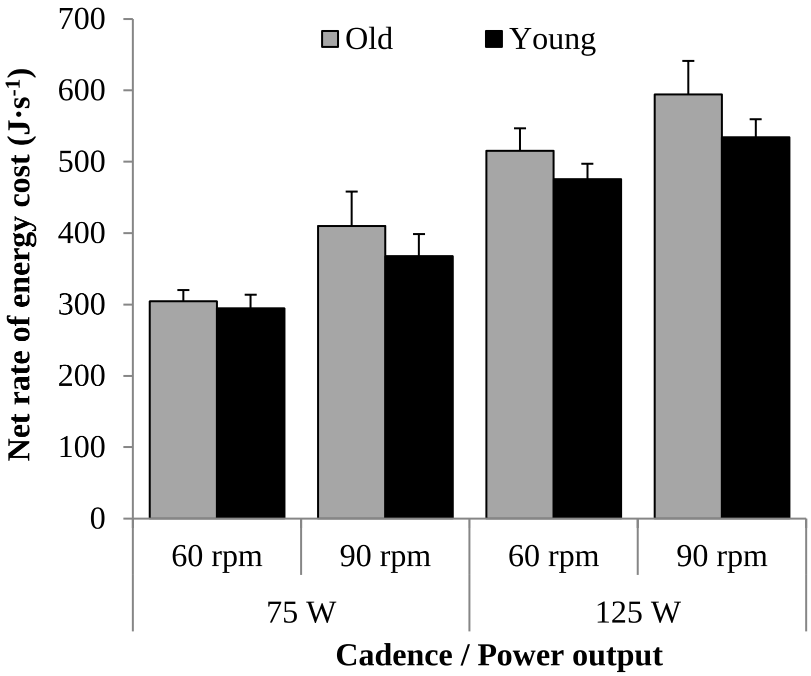

Figure 1: Results for net rate of energy cost reflected main effects for age (older adults had 8% greater cost than young participants, p=0.002), power output (125 W was 54% greater than 75 W, p<0.001), and cadence (90 rpm was 20% greater than 60 rpm, p<0.001). Energy cost differences between older and young participants were greater at 125 W (age x power output interaction, p=0.002) and at 90 rpm as predicted (age x cadence interaction, p=0.026). Values are mean ± 1 SD.

"Effects of Age, Power Output, and Cadence on Energy Cost and Lower Limb Antagonist Muscle Co-Activation during Cycling"
by Buddhadev HH, Martin PE
Journal of Aging and Physical Activity

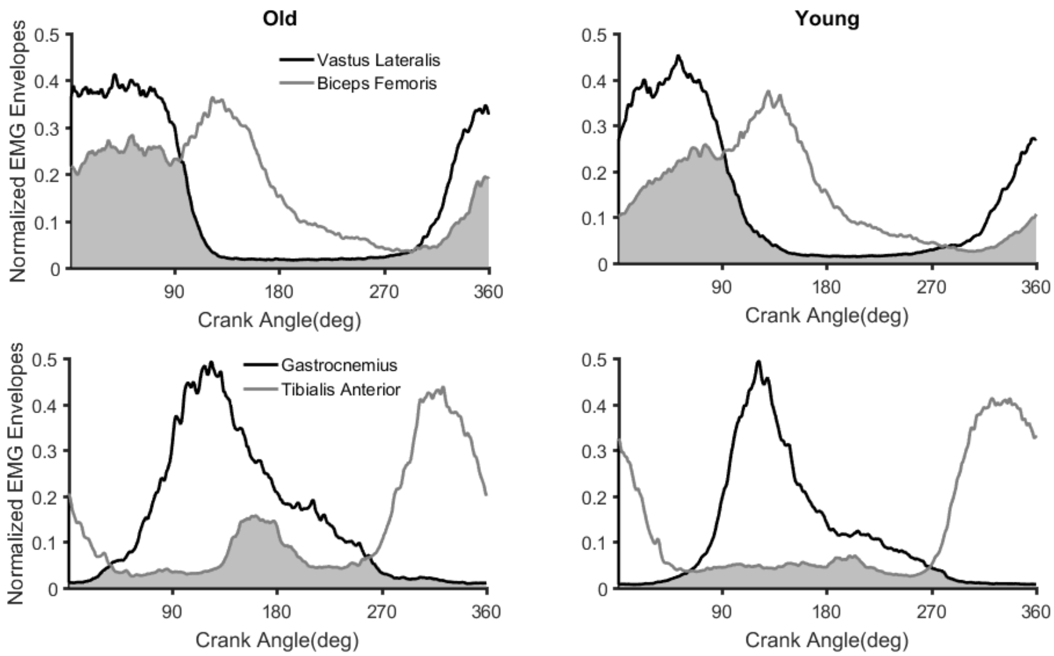

Figure 2: Ensemble averages of vastus lateralis, biceps femoris, medial gastrocnemius, and tibialis anterior excitation profiles for one full crank cycle (top dead center = 0 and 360 degrees) for the 125 W-90 rpm condition, EMG amplitudes have been normalized to peak EMG activity during the 125 W-90 rpm condition. The areas shaded in gray represent co-activiation between the antagonist pair of muscles.

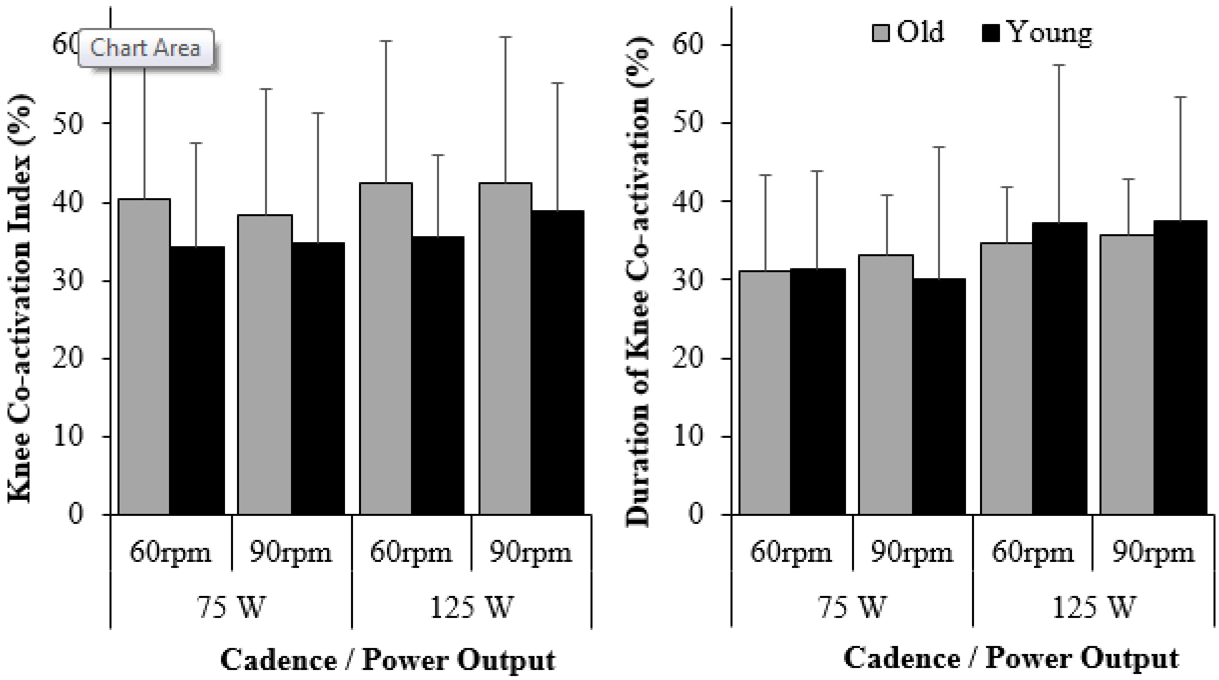

Figure 3: Knee co-activation indices and duration of knee co-activation as percent of crank cycle ($p<0.001$) were significantly greater for 125 W compared to 75 W. No age-related differences in co-activation were observed. Data represent mean ± 1 SD.

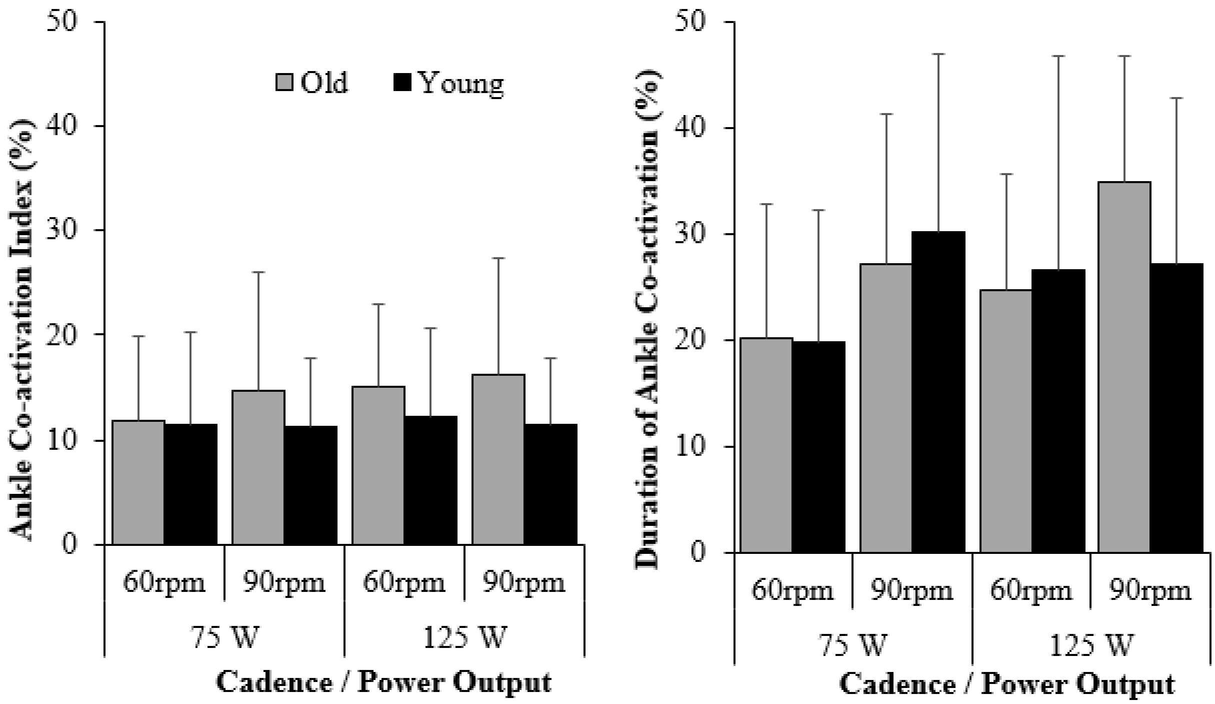

Figure 4: Ankle co-activation indices (p<0.023) and duration of ankle co-activation (p<0.041) as percent of crank cycle were also significantly greater for 125 W compared to 75 W. Duration of ankle co-activation also increased as function of cadence (*p*<0.001). No age-related differences in co-activation were observed. Data represent mean ± 1 SD.

"Effects of Age, Power Output, and Cadence on Energy Cost and Lower Limb Antagonist Muscle Co-Activation during Cycling"
by Buddhadev HH, Martin PE
Journal of Aging and Physical Activity

Table 1: Participant demographic characteristics, average weekly cycling, and preferred cadences at 75W and 125W. Older and young participants did not differ on mass, height, average duration of weekly cycling, and their preferred cadences a 75W and 125W, respectively.

	Old (*n*=12)	Young (*n*=13)
Age (years)	69.5±4.6	22.9±3.4
Mass (kg)	81.4±8.4	75.6±9.7
Height (cm)	177.0±5.3	178.6±6.2
Average weekly cycling (min)	131.8±69.3	148.5±89.3
Preferred cadence at 75W (rpm)	71.3±15.8	79.7±11.3
Preferred cadence at 125W (rpm)	71.5±14.5	78.62±12.2

Note. Values are mean ± one standard deviation

"Effects of Age, Power Output, and Cadence on Energy Cost and Lower Limb Antagonist Muscle Co-Activation during Cycling"
by Buddhadev HH, Martin PE
Journal of Aging and Physical Activity

Table 2: Average EMG activity per crank cycle expressed as a ratio of the peak EMG obtained in 125W-90 rpm condition for each muscle.

Condition	**75W-60 rpm**	**75W-90 rpm**	**125W-60 rpm**	**125W-90 rpm**
Vastus Lateralis *				
Old	0.115 ± 0.033	0.113 ± 0.027	0.164 ± 0.051	0.151 ± 0.017
Young	0.099 ± 0.021	0.100 ± 0.027	0.137 ± 0.039	0.142 ± 0.024
Biceps Femoris *, ‡				
Old	0.096 ± 0.029	0.137 ± 0.043	0.148 ± 0.060	0.171 ± 0.037
Young	0.115 ± 0.050	0.105 ± 0.040	0.168 ± 0.082	0.149 ± 0.034
Gastrocnemius ^, #, ‡				
Old	0.090 ± 0.024	0.158 ± 0.036	0.094 ± 0.031	0.153 ± 0.030
Young	0.069 ± 0.028	0.096 ± 0.022	0.064 ± 0.023	0.109 ± 0.021
Tibialis Anterior *, #,†				
Old	0.065 ± 0.026	0.138 ± 0.034	0.079 ± 0.036	0.132 ± 0.038
Young	0.053 ± 0.021	0.112 ± 0.019	0.073 ± 0.019	0.127 ± 0.016

Note. Values are mean ± SD. Statistically significant (p<0.05) age main effect (^), power main effect (*), cadence main effect (#), age x power interaction (†), and age x cadence interaction (‡).