FLEX CEUs

Muscle Recruitment in Cycling in Older Adults

Aging and Muscle Activity Patterns during Cycling

Abstract

Objective: The purpose of this study was to evaluate the muscle activation patterns of lower limb muscles during 90-second trials at randomly assigned workloads (0 and 100 W) with a constant cadence (60 rpm) in young and older healthy adults.

Methods: Twelve healthy, male, novice cyclists classified by age into two groups of young and older. Electromyographic (EMG) data were recorded from rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA), and gastrocnemius medialis (GT). Joints kinematics was also recorded simultaneously. A performance index (*PI*) was developed to evaluate the characteristics of muscle recruitment between workloads.

Results: EMG duration and peak magnitude increased significantly with increased workload in RF and BF in both groups. *PI* Values indicated that BF and RF had similar increases with increased workload for both groups, while younger groups had higher activation of TA (52% v/s 28%) and older group had higher activation of GT (17% v/s 1%). Both groups exhibited significant increases ($p<0.05$) in the co-activation of upper leg agonist and antagonist muscles as workload increased. Duration of co-activation between the upper and lower leg muscles of the young group significantly increased ($p<0.05$) with workload. The ROM of the knee splay angle in the older group showed a significant difference ($p<0.05$) compared to the young.

Conclusion: The alteration in limb muscle activation and coordination is not parallel between young and older adults. Older adults appear to use a different strategy in recruiting more muscle fibre to generate the same work. These findings are an indication that the threshold for muscle adaptation may differ between young and old adults. It is recommended that rehabilitation professionals consider these differences and recognize that the stimulus required to promote positive change in older skeletal muscles might be different from what may require for young adults.

Keywords: Cycling; Aging; Electromyography; Muscle activity patterns; Muscle co-activation; Kinematics; Lower-extremity

Background

The human musculoskeletal system experiences certain structural and functional changes due to the aging process. Aging is associated with sarcopenia, a process of decline in muscular mass, strength, and power. These changes may alter muscle activation patterns and, consequently, result in neuromuscular adjustments during activities [1,2]. Sarcopenia also causes reductions in the number of muscle fibers and reductions in muscle cross sectional area. Previous studies have reported a 30%-50% decrease in skeletal muscle mass between 40 to 80 years of age in the general population [3,4]. Even the muscle performance of athletes, who maintain a high level of fitness, decreases after approximately 40 years of age [5]. The decrease in muscle mass, along with reductions in muscle power and strength, may have a significant impact on the functional mobility of the elderly [1,6,7]. For example, older individuals have increased co-activation of knee muscles during walking, and isometric contractions [8,9]. The joint stiffness at the knee and ankle, as well as increased co-activation of involved muscles, may generate a compensatory mechanism from the hip joint, causing further damage to the musculoskeletal system. The same level of co-activation is possibly present during other complex activities, such as cycling. This phenomenon may cause atypical neural and muscle excitations and contribute to early occurrence of fatigue during these complex tasks and may ultimately encourage a sedentary lifestyle. Consequently, the age-related changes in skeletal muscles are accompanied by increased susceptibility to muscular damage, decelerated recovery, and decline in the neural drive when compared to younger individuals [1,5,10].

A long-term sedentary lifestyle in older individuals may lead to physical frailty. Frailty is defined as a clinical syndrome in which the person has atrophied muscles that show significant impairments in strength and endurance. This phenomenon may lead to impaired balance and mobility increased risk of falling declining health and quality of life. Yet, previous studies have shown that muscle integrity and function could be maintained and improved with age-appropriate physical activity and exercise interventions as the person ages [1-3,11]. Macaluso et al. [9] used cycling as a novel approach to resistance training and found it to be an effective approach in improving muscle strength and power in a group of healthy, elderly women [9]. They postulated that understanding muscle groups' activation, co-activation, and synchronization patterns involved in producing certain types of movements are crucial in developing an effective exercise program for older individuals.

Older adults with age-related muscle weakness typically receive rehabilitation treatment with the expectation of improvement in muscle strength and function. Low- to moderate-intensity exercises, such as cycling, are commonly utilized for rehabilitation, as they are easier to perform and can be implemented based on older adults' physical capabilities to lower the risk of injury. Semi-reclined cycling is the preferred type of cycling exercise andrehabilitation in older individuals due to its back support and comfortable seating. Walking is a complex form of locomotion and is hard to evaluate during different levels of exertion. Therefore, cycling might be an ideal exercise to evaluate muscles and biomechanics of older individuals [12,13]. With old age, variations in motor unit activation and discharge rate may result in unsteady force production and different neural strategy in skill full movements, such as cycling [14]. Older adults may activate the agonist and antagonist muscles disproportionately in an effort to compensate for reduced muscle strength and stability. The effects of different cycling cadences on lower extremity muscle activities and joint kinematics have been previously reported [15-18]. However, a thorough examination of muscle recruitment patterns during semi-reclined cycling between young and older adults has not been performed. Consequently, in this study we investigated the alterations in lower-extremity muscle activations during semi-reclined cycling at two workloads a constant cadence, in young and older healthy adults. We hypothesized that: 1) with an increased workload, older adults will manifest significantly different muscle activation patterns during each crank rotation compared to young adults, and 2) older adults will exhibit increased co-activation of lower limb muscles, in order to produce the required work.

Methods

Participants

Twelve healthy, male, novice cyclists based on their age, they were divided into two groups of six: young (mean ± SD: 22 ± 2 yrs, 1.8 ± 0.1 m, 74.8 ± 5.8 kg) and older (mean ± SD: 53 ± 3 yrs, 1.8 ± 0.1 m, 85.5 ± 10.3 kg). Individuals who had a known history of joint problems, movement limitations, or those who responded positively to any of the questions on the Physical Activity Readiness Questionnaire (PAR-Q) were excluded from participating in the study [19]. All participants signed an informed consent, which was approved by the university's Institutional Review Board. The study was conducted in the Functional Performance Laboratory, equipped with all necessary equipment and measurement systems for the study.

Preparation

After signing the consent form, each participant's height and weight were measured and recorded along with their exact age. Participants rode a standard stationary semi-reclined ergometer (SciFit ISO 7000R, Tulsa, OK) at a constant cadence of 60 rotations-per-minute (rpm) throughout the study [16].

The activities of rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA), and gastrocnemius medialis (GT) of the right leg were monitored using surface electromyography (sEMG). Electrode sites were identified, shaved, and lightly abraded based on the recommendations of the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) [20,21]. Bipolar pre-gelled silver/silver-chloride (Ag-AgCl) surface electrodes with an inter-electrode distance of 2 cm (Noraxon USA Inc., Scottsdale, AZ) were placed on each muscle, along the muscle fibers. A common reference electrode was positioned on a bony site, at the distal end of the left ulna. Although electrodes were self-adhesive, they were further fixed on the skin by using adhesive tapes to avoid possible artifacts. A 10-channel EMG system (Nexus-10, MindMedia B.V., Netherlands) was used to record sEMG signals.

A camera-based motion capture system (VICON Motion Systems Ltd., United States) was used to evaluate shoulder, hip, knee, ankle, and crank position during each trial. Retroreflective markers were placed on the right acromion, greater trochanter, lateral epicondyle of femur, lateral malleolus, lateral aspect of fifth metatarsal head, calcaneus, pedal spindle, and crank axle, as well as two reference markers on the bike. Since the fifth metatarsal head and calcaneus are covered with the sneaker, markers were placed directly on the sneaker approximate to their anatomical landmarks.

Experiment

Participants were asked to rest on a chair for 10 minutes before their resting heart rate and blood pressure were measured. Heart rate and blood pressure were measured by using a Polar T34 chest-belt (Polar Electro Oy, Kempele, Finland) and an Omron HEM-712C blood pressure monitor (Omron Healthcare Inc., Kyoto, Japan), respectively.

Ergometer's seat was adjusted for each participant's anthropometrics such that the minimal knee flexion angle of the right leg was measured at 40°, while the pedal was at 110° clockwise from its highest vertical position (i.e., top dead center or TDC). Straps were used to fix the position of the foot on the pedal without causing any limitations to the movement of the ankle. Participants were asked to keep their arms on the sides of the bike, and do not hold the handlebars. At the beginning of the session, each participant was acclimated to the bike by pedaling for five minutes with no resistance.

The cycling conditions included pedalling at two workloads of 0 Watts (i.e., no resistance) and 100 Watts (W), while the cadence was maintained at a constant rate of 60 rotations-per-minute (rpm). Testing order of these workloads was randomized for each individual to minimize any possible order effects. Two trials were executed for each participant following a 2-minute warm-up and a 2-minute cool-down at the completion of both trials. Each testing trial consisted of 90 seconds of cycling at a specific, predetermined workload. After a steady state of cycling (i.e., constant cadence) was reached, data was collected for 30 seconds within the 90-second testing window. A five-minute rest period was allotted between the two trials to minimize possible fatigue effects. To ensure participants' safety, heart rate was monitored continuously and blood pressure was measured before and after each trial.

Data collection and processing

Kinematic and sEMG data was collected from all participants as they pedalled during both trials. During the experiment, data was only collected from the right limb of participants. Previous research observed no significant difference between the knee extensors of the right and left limbs [22]. All recorded data was synchronized using a custom-written Labview (National Instruments Corp., Austin, TX) program and imported into Matlab (MathWorks Inc., Natick, MA) for further processing and statistical analyses.

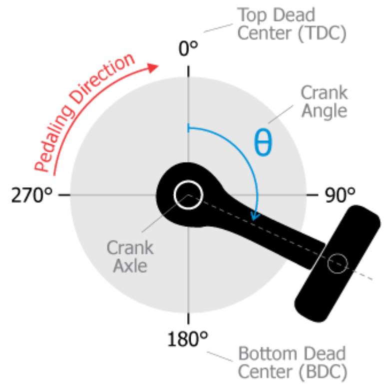

Figure 1: Schematic illustration of the pedal, crank angle, and pedalling direction.

Kinematic data

Raw kinematic data was collected at 128 frames per second and filtered using a low-pass, 4th order, zero-lag Butterworth filter with a cut-off frequency at 10 Hz [23]. Recorded coordinates of each marker were used to calculate crank and joint angles for each participant at every given workload. Hip (ΘH), knee (Θk), and ankle (ΘA) angles, on the sagittal plane, and the thigh abduction/adduction angle (ΘS), in the transverse plane, were calculated (Figure 1). The data was then ensemble averaged over 20 consecutive full revolutions of the crank (0°-360°) (Figure 2). Mean joint angles (MA) and ranges of motion (ROM) were calculated to quantify the extent of change in movement kinematics during cycling at different workloads.

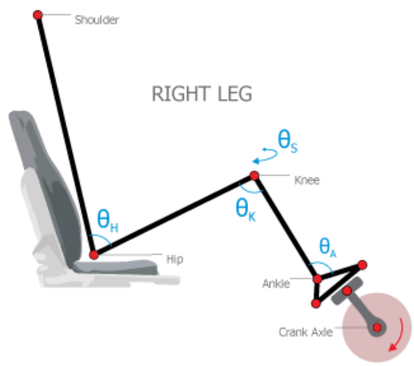

Figure 2: Schematic illustration of a participant's right leg, with joint angles defined as: hip (ΘH); knee (Θk); knee splay (ΘS); and ankle (ΘA). Retroreflective markers are marked by red circles.

Electromyography data

Raw EMG data was sampled at 2048 Hz. According to the manufacturer's recommendations, raw EMG signals were initially bandpass filtered from 20 to 500 Hz using a 5th-order Butterworth filter to reduce low-frequency motion artifacts and high-frequency noise. To form the linear envelope, signals were then full-wave rectified and smoothed by a low-pass, zero-lag, 4th-order Butterworth filter with a cut-off frequency at 5 Hz. EMG ensemble average (EEA) curves were then calculated over 20 consecutive full revolutions of the crank, at each workload, and normalized to the maximum EMG magnitude found across both workloads for each participant [13]. The muscle was considered active when EEA values were greater than a threshold of 10% of each participant's maximum EMG value across both workloads [24]. The regions of muscle activity were indexed between the starting (EMG_{on}) and ending (EMG_{off}) crank angles. The EMG_{on} (onset) and EMG_{off} (offset) were measured in units of degrees. The duration of muscle activity (EMG_d) was defined as the number of degrees between the onset and offset of each muscle activity during one crank rotation:

$$EMG_d = EMG_{off} - EMG_{on}$$

EMG peak magnitude (EMG_{peak}) for each participant at each workload was identified as the maximum EEA value (µV), after signals were normalized over the maximum EMG value found across both workloads [24].

The change in muscle recruitment between the workloads of 0 and 100 W workloads (i.e., performance index) was calculated and compared between the young and older groups. The performance index (*PI*) was calculated for each individual muscle as the symmetrized percent change by using the following formula [25].

$$PI = \frac{\bar{E}^{100}_{peak} - \bar{E}^{0}_{peak}}{\bar{E}^{100}_{peak} + \bar{E}^{0}_{peak}} \times 100$$

Where E_{peak} is the mean peak EMG of 20 consecutive rotations at each workload and *PI* is the performance index, which is bounded between -100 and +100.

Co-activation of agonist and antagonist muscles were examined using two lower limb muscle pairs: RF-BF and TA-GT. Additionally, we looked at the synchronized activity of upper and lower leg muscle pairs (i.e., RF-TA, RF-GT, BF-TA, and BF-GT). Duration of co-activation (EMG_c) was identified and calculated for each muscle pair at each workload.

Statistical analysis

Circular statistics [26-28] were utilized to examine within-subject differences for three variables of EMG_{on} and EMG_{off} across the two workloads. Mean crank angle and circular standard deviation were calculated by means of circular statistics, which has been previously implemented and published on cycling data [26-28]. EMG_d, EMG_{peak}, duration of co-activation, as well as mean angles and mean ranges of motion (ROM) of both groups across the two workloads were examined using a two-way (age × workload) analysis of variance (ANOVA). For multiple comparisons, Bonferroni post-hoc correction was applied. A t-test was used to identify any significant difference in the performance index (*PI*) between the two groups. Significance level was set at p-value less than 0.05 for all analyses.

Results

Participants maintained a relatively constant cadence of 60 rpm throughout trials. Activity patterns of four lower limb muscles are displayed as EEA curves of one young and one older participant

(Figure 3). The shaded areas depict the SE. Activations (EMG_{on}) and deactivations (EMG_{off}) of all muscles were evaluated during 360° of pedal rotation. In both groups, as workload increased from 0 to 100 W, average onsets of muscle activation (EMG_{on}) for all muscles occurred sooner in the pedal cycle, during earlier crank angles (Figure 4). At the same time, the average offsets of muscle activation (EMG_{off}) in all muscles exhibited delays and occurred later in the pedal cycle (Figure 4) as workload increased in both groups.

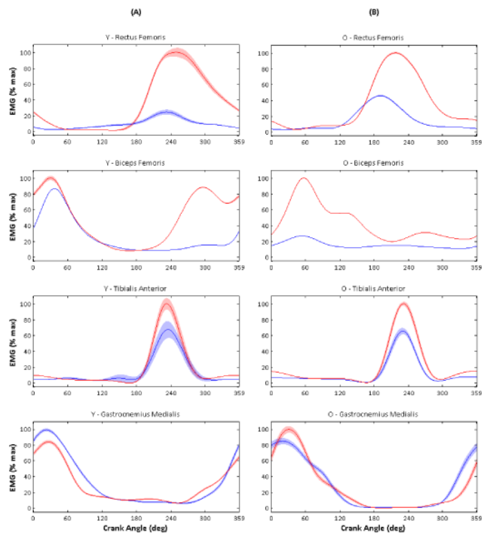

Figure 3: Normalized EMG ensemble average (EEA) curves (± SE) of four lower-extremity muscles for two participants: Young (column A) and older (column B).

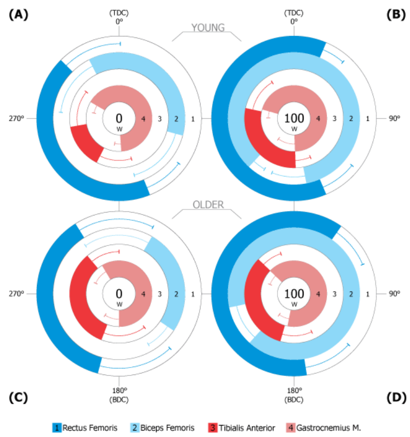

Figure 4: Activity patterns of four lower-extremity muscles (RF, BF, TA, and GT) in young (A and B) and older (C and D) groups, at both workloads (0 and 100 W).

Duration (EMG_d):

The duration of muscle activation (EMG_d) increased significantly (p<0.05) in upper leg muscles (i.e., RF and BF) as workload increased in both groups. The lower leg muscles also showed some trends towards longer durations of activity with increase in workload; however, the change was not significant in either group (Figure 4). Table 1 depicts the EMG duration, in units of degrees (0°-360°), for both groups during 0 and 100 W workloads.

	Young		Older	
Workload (Watts) ?	**0**	**100**	**0**	**100**
RF	**143 (24)***	**226 (18)***	**114 (33)$^\#$**	**227 (21)$^\#$**
BF	**99 (39)***	**314 (16)***	**147 (45)$^\#$**	**324 (8)$^\#$**
TA	76 (15)	143 (28)	142 (50)	228 (51)
GT	236 (14)	252 (24)	235 (30)	237 (16)

*Significant difference between workloads 0 and 100 W within the Young group (p-value<0.05)

$^\#$Significant difference between workloads 0 and 100 W within the Older group (p-value<0.05)

Table 1: Duration of activity (EMG_d) presented as mean (± SE) in units of degrees.

Peak magnitude (EMG_{peak})

EMG peak magnitude (EMG_{peak}) of upper leg muscles (i.e., RF and BF) showed significant increases (p<0.05) in both age groups, as workload increased. In lower leg muscles, however, the young group exhibited significant increases (p<0.05) in the EMG_{peak} of TA as workload increased. Furthermore, EMG_{peak} of GT significantly increased (p<0.05) with increase in workload in the older group. Except for GT in young individuals, the trend was for the peak magnitude to increase with workload (Table 2).

	Young		Older	
Workload (Watts) ?	**0**	**100**	**0**	**100**
RF	**32 (6)***	**100 (0)***	**36 (12)$^\#$**	**100 (0)$^\#$**
BF	**40 (13)***	**100 (0)***	**26 (3)$^\#$**	**100 (0)$^\#$**
TA	**32 (9)***	**100 (0)***	56 (14)	91 (9)
GT	85 (9)	83 (5)	**64 (8)$^\#$**	**95 (4)$^\#$**

*Significant difference between workloads 0 and 100 W within the Young group (p-value<0.05)

$^\#$Significant difference between workloads 0 and 100 W within the Older group (p-value<0.05)

Table 2: Normalized EMG peak magnitude (EMG_{peak}) across workloads expressed as mean (± SE).

Performance index (PI)

Higher values of the performance index (PI) indicate greater muscle recruitment with an increased workload. The older group showed higher PI values in BF and GT, compared to young participants, indicating more muscle fiber recruitment while performing the same task. However, lower PI levels in RF and TA were observed in the older group, compared to young, which translates into less muscle fiber recruitment (Table 3). Regardless of age, the average PI value for each muscle was positive in both groups, indicating that higher workloads require a higher level of muscle fiber recruitment. Figure 5 depicts all *PI* values for both groups, presented separately for each muscle.

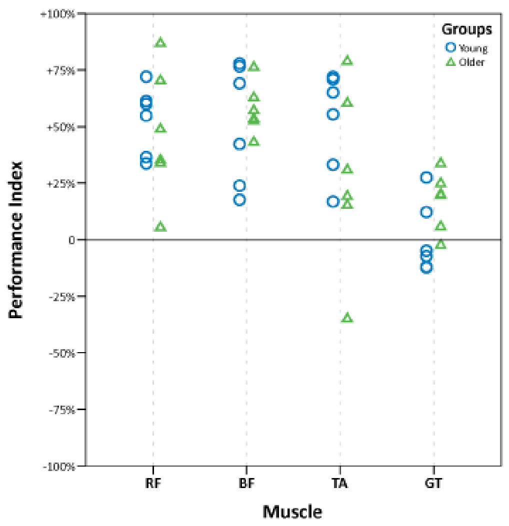

Figure 5: Performance Index (*PI*) of four lower limb muscles for both groups of young (circles) and older (triangles) individuals.

	Young	Older
RF	+53% (6)	+47% (12)
BF	+51% (11)	+58% (5)
TA	+52% (9)	+28% (16)
GT	+1% (7)	+17% (5)

Table 3: Performance index (*PI*) presented as percentage mean (± SE).

Co-Activation (EMG_c)

As workload increased, both young and older groups showed significant increases ($p<0.05$) in the co-activation duration (EMG_c) of their upper leg agonist-antagonist pair (i.e., RF-BF). However, none of the groups exhibited a statistically significant difference in co-activation duration of the lower leg agonist-antagonist pair (i.e., TA-GT) with increased workload (Table 4).

Furthermore, synchronized activities of four additional muscle pairs (i.e., RF-TA, RF-GT, BF-TA, and BF-GT) were evaluated. In the young group, with increased workload, EMG_c of the RF-TA, RF-GT, BF-TA, and BF-GT pairs significantly increased ($p<0.05$). However, in the older group, with increased workload, only the EMG_c of BF-TA increased significantly ($p<0.05$). Overall, the co-activation durations of all muscle pairs were observed to have a tendency to increase with workload (Table 4).

Kinematics

Statistical evaluation of joint kinematics has shown that there is a significant difference ($p<0.05$) in the ROM of the knee splay angle (Θ_s) in the older group as workload increased. No other significant differences were observed in joint kinematics (i.e., ROM and MA) with increased workload.

Discussion

Semi-reclined cycling is more acceptable for deconditioned elderly or obese individuals due to its comfortable seating and back supports [29]. Although all four lower limb muscles are active during cycling, the BF and RF are by far the most active muscles. Of the lower leg muscles, GT is relatively more active than TA. In the present study, we evaluated the muscle activation patterns of upper and lower leg muscles, as well as the effects of agonist and antagonist co-activations, under different workloads during semi-reclined cycling. Our study also provides a better understanding of the agonist and antagonist muscle activities as well as synergistic behaviour of the upper and lower leg muscles during cycling. Previous research has focused on various factors influencing the cycling movement and cyclists' performance, such as cadence [18,30], workload [13,30], body position [31,32] , and saddle height [33]. To the best of our knowledge, there are no published studies that have examined potential age-related changes in the activity patterns of the musculoskeletal system during semi-reclined cycling.

	Young		Older	
Workload (Watts)?	**0**	**100**	**0**	**100**
RF-BF	**22 (12)***	**194 (24)***	**26 (16)#**	**193 (58)#**
TA-GT	21 (21)	56 (34)	38 (33)	86 (40)
RF-TA	**53 (14)***	**120 (10)***	52 (21)	141 (32)
RF-GT	**43 (13)***	**132 (28)***	37 (23)	106 (14)
BF-TA	**20 (14)***	**121 (26)***	**43 (43)#**	**152 (45)#**
BF-GT	**88 (36)***	**230 (20)***	140 (42)	235 (13)

*Significant difference between workloads 0 and 100 W within the Young group (p-value<0.05)

#Significant difference between workloads 0 and 100 W within the Older group (p-value<0.05)

Table 4: Duration of co-activation (EMG_c) presented as mean (± SE).

Overall activities of the lower-extremity muscles (i.e., RF, BF, TA, GT) were affected by workload in both age groups. More specifically, the duration of muscle activity, and its peak magnitude, increased by workload in both groups. This observation is in agreement with similar studies on upright cycling [34] and semi-reclined cycling [35]. Muscle activation patterns during upright and semi-reclined cycling positions were observed to be similar in a previous study, after adjusting for different orientation in the gravity field [16]. Previous research in semi-reclined cycling has shown that the onset of muscle activation typically occurs earlier in each cycle of the pedal as workload increases; moreover, the offset of muscle activation happens later in the cycle with an increased workload. As a result, the duration of muscle activity had increased proportionally with increasing cycling workload, during a constant cadence [13,35]. In the present study, we observed similar behaviours of the muscles during semi-reclined cycling while workload increased from 0 to 100 W.

The duration of activity (EMG_d) in upper leg muscles (i.e., RF and BF) significantly increased with workload, emphasizing the major role of these muscles during cycling. At 100 W, activity durations (EMG_d) of RF and BF were closely similar between the two groups; however, EMG_d values at 0 W were relatively different. More specifically, young group showed longer activity duration in RF at 0 W, while their BF activity duration is much shorter in comparison. This may imply a different cycling strategy acquired by the older adults in order to maintain the required cycling cadence, which entails minimal muscle power production. This implication becomes even more evident after examining the EMG_d of TA at 0 W between the two groups. EMG_d of TA in older group was relatively longer at 0 W than any of the workloads in the young group. It is apparent from the observed differences in the activity of lower limb muscles that older adults employ a different strategy when the task does not require a lot of force production, but still demands coordination and precise control over the movement (i.e., maintaining a constant cycling cadence). This is in agreement with previous findings that the amplitude of force fluctuations among young and old adults is greatest at low forces [14].

The mean peak magnitude EMG (EMG_{peak}) of all muscles increased except for GT in young as workload increased from 0 to 100 W. This finding is consistent with previous research [13] reporting that additional recruitment of muscle fibers may be essential for maintaining a constant pedalling cadence with increased workload. At 100 W, upper leg muscles (i.e., RF and BF) of both groups remained active for a significantly longer period during each cycle of the pedal. This change was reflected in the EMG peak magnitude (EMG_{peak}) of upper leg muscles as they increased significantly with workload in both age groups. Although there are differences between groups in the EMG_d values of upper leg muscles, the level of muscle recruitment seems to be relatively similar as workload increased.

In the lower leg muscles, however, EMG_{peak} of the two groups exhibited a different behaviour as workload increased. Young group maintained a relatively high EMG_{peak} in the GT muscle (i.e., over 80%) while recruited more muscle fibers in the TA muscle as workload increased. However, older group started with an over 50% recruitment of both lower leg muscles and increased recruitment of these muscles as workload increased. These observations of the lower leg muscle activity suggest that young individuals rely more on TA, which is a mono-articular, force-producing muscle, for a relatively demanding task; while older adults recruit both muscles to the same extent and for a longer duration within each cycle of the pedal.

Furthermore, the performance Index (*PI*) provided a better understanding of the relative difference in the extent of muscle recruitment between the two workloads. As expected, both groups increased their upper leg muscle recruitment at similar levels when workload was increased to 100 W. The *PI* value of TA was relatively low in the older group (i.e., +28%, as opposed to +52% in young). This might be due to TA's longer duration of activity and higher peak magnitude at both workloads, when compared to younger group. On the other hand, the calculated of GT (i.e., 17%) was considerably higher in the older group, compared to young (i.e., 1%). This finding suggests that older individuals increase the recruitment of both TA and GT muscles when a higher workload is applied. In contrast, young adults rely mostly on the TA and maintain the activity of the GT at almost the same level.

Co-activations of two agonist-antagonist muscle pairs (i.e., RF-BF and TA-GT) were examined in this study as well as four additional muscle pairs (i.e., RF-TA, RF-GT, BF-TA, and BF-GT). The reason behind choosing the latter four pairs was to evaluate the synchronization between the upper and lower leg muscles and compare them between the two age groups. The co-activation duration of all muscle pairs increased with workload in both groups. This observation is not entirely surprising since the average duration of activity in all muscles increased in both groups with workload. The upper leg agonist-antagonist pair showed significant increases of EMG_c in both groups, while the lower leg agonist-antagonist pair exhibited only slight increases in EMGc. Comparing the two groups, the increase in EMG_c of the older group was larger, which implies a longer duration of simultaneous activity in the lower leg muscles of older individuals. It is important to note that TA is a monoarticular muscle, while the other three are biarticular (i.e., RF, BF, and GT). As described in previous research, monoarticular muscles are mostly responsible for force production, while biarticular muscles distribute and transfer energy and contribute relatively little work [30]. This may explain the significant EMG_c increases in all muscle pairs, of the young group, that involve at least one of the biarticular upper leg muscles (i.e., RF and BF). RF, BF, and GT are responsible for distribution and transfer of energy and their increased EMG_c may indicate the individual's attempt to distribute the increased workload and maintain the constant cycling cadence. However, most of these EMG_c increases are not significant in the older group, which adds to our previous speculation on older adults adopting a different cycling strategy while cycling workload is increased.

Previous gait studies have proposed that older adults exhibit longer durations of co-activation during walking to increase joint stability [8,36]. In our study, semi-reclined cycling was used as the cyclical movement; therefore, maintaining balance did not play a role in the muscle activity patterns. On the other hand, it has been shown that maintaining a constant cadence, by optimizing pedal forces, modifies the activity patterns of the RF and BF muscles [37]. If we assume that maintaining a constant cadence in a controlled environment like semi-reclined cycling is equivalent to maintaining balance during gait, our findings do not support the theory of increased co-activation for better stability due to similar levels of muscle co-activation between the two age groups in all muscle pairs.

Previous research suggests that there are no joint kinematics differences as workload increases. Bini et al. [38] assessed the effect of cycling workload on joint kinematics among athletes and non-athletes and concluded that changes in submaximal workload do not substantially affect joint kinematics. Similarly, Gregor et al. [12] investigated age-related differences in generalized muscle moments as workload increased and reported no significant differences across age groups. Our result provided support for these findings, while no significant differences were observed in joint kinematics (i.e., ROM and MA of Θ_H, Θ_K, and Θ_A) as workload increased during constant cadence cycling. However, the ROM of the knee splay angle was significantly different between the two groups, further confirming our theory that there is a change in older adults' cycling strategy as workload increases.

Conclusion

In conclusion, our results indicate that changes in workload will affect the lower-extremity muscular activity and coordination during cycling. These differences are not always the same across different age groups. Older adults appear to adopt a different cycling strategy to maintain a constant cadence that includes higher activation and co-activation of lower leg muscles as well as increased ROM of the knee splay angle. We conclude that this change in strategy is adopted by older individuals to increase muscle fiber recruitment, maintain a constant pedaling cadence, and overcome more challenging, higher workloads. The outcome of this research provides the researchers and clinicians new insights in designing effective exercise programs and a new direction from which to move forward in reducing the conditions of physical frailty.