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Effects of Relative Inertial Load on Quadriceps Electromyography during Maximal Effort Flywheel-based Iso-Inertial Training (FIT) Squats

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ABSTRACT

The effects of inertial load on quadriceps electromyography (EMG) during flywheel-based iso-inertial training (FIT) squats is unclear. Healthy, resistance-trained males (n=8) and females (n=8) completed five sets of five maximal-effort squats with varying inertial loads in a randomized order. Sagittal plane knee joint angles and surface EMG activity of the vastus lateralis (VL) were measured. Average knee angular velocity, but not peak knee flexion or total excursion, decreased during both the concentric and eccentric phases with increasing inertial load ($p<0.001$) in both sexes. Integrated EMG (iEMG) activity increased with increasing inertial load for all muscles ($p<0.001$). VL iEMG and peak amplitude were significantly greater in females. However, peak EMG amplitude was not significantly different among inertial loads for either sex. Mean EMG amplitude during the concentric phase and eccentric phases of the VL tended to be greater in females. Increasing relative inertial load increases quadriceps iEMG activity due to greater exercise duration, but has minimal effects on EMG amplitude.

Keywords: iso-inertial training, eccentric, resistance training, neuromuscular, motor unit activity

Abbreviations: 1RM: 1 repetition maximum; ANOVA: Analysis of variance; DCER: Dynamic constant external resistance; EMG: Electromyography; FIT: Flywheel-based Iso-inertial Training; iEMG: Integrated electromyography; MVIC: Maximal voluntary isometric contraction; PCP: Peak concentric power; PEP: Peak eccentric power; RF: Rectus femoris; VL: Vastus lateralis; VM: Vastus medialis.

INTRODUCTION

The squat is a ubiquitous motion among sporting disciplines for able-bodied participants, which requires vertical displacement of body weight and overcoming external resistance through coordinated flexion/extension of the hips, knees, and ankles. Squat strength and power output are associated with high levels of athletic performance such as vertical jump and sprinting[1,2]. Quadriceps electromyography

(EMG) amplitude is directly related to concentric and eccentric force during the squat[3], suggesting that motor unit activation is a major determinant of squat performance.

Flywheel-based Iso-inertial Training (FIT) is becoming an increasingly popular strength training modality that provides external resistance primarily by the moment of inertia of cylindrical disks. During FIT, a maximal effort concentric movement puts the flywheel into rotational motion and this energy is returned to the user during the eccentric phase. Some[4], but not all[5], data suggest that FIT may elicit greater increases in muscular strength, power, and hypertrophy compared to gravity-dependent forms of resistance training, which is often attributed to greater eccentric overload during FIT. During FIT squats, external resistance is primarily provided by flywheel moment of inertia ($I=\frac{1}{2}\ mr^2$). To date, little is known concerning the acute neuromuscular responses to FIT, particularly during multi-joint exercises such as the squat. Developing a better understanding of how inertial load affects motor unit recruitment will help practitioners develop more effective FIT exercise programs targeting specific adaptations.

Quantifying intensity during FIT has been problematic in previous studies[6,7]. Due to the inertial nature of FIT, exercise can be successfully completed as long as force is greater than zero since any non-zero force will be sufficient to rotate the flywheel[7]. Thus, unlike gravity-dependent exercise, there is no maximal load or one repetition maximum (1RM) during FIT which can be used to prescribe intensity. Several studies have implemented FIT based on absolute inertial load, but this does not account for variation in strength between subjects[11,24–26]. Vertical velocity has recently been proposed as a better index of FIT intensity than absolute inertial load[8–10]. However, establishing inertia-velocity relationships can be a difficult and time-consuming process, thus limiting the utility for practitioners. Increasing inertial load results in a decreased maximal movement speed[11], which consequently increases exercise duration and time under load. Developing a mechanism of quantifying relative inertial load may provide practitioners with a utilitarian method to prescribe FIT moment of inertia while accounting for inter-individual differences in strength.

Increasing inertial load decreases power output and increases average force[12] but does not affect peak sagittal plane joint angles during FIT-squats[11]. Alkner et al.[13] recently described greater quadriceps EMG amplitude during both concentric and eccentric phases of FIT-leg press compared to a dynamic constant external resistance (DCER) front squat. Additionally, Norrbrand et al.[14] described greater EMG amplitude of the rectus femoris, but not the vastus lateralis or vastus medialis during the eccentric (but not concentric) phase of a FIT-based leg press compared to a DCER back squat. However, these studies only assessed EMG activity at a single inertial load. Carrol et al[8] reported increasing quadriceps EMG activity with increasing inertial loads. However, this was performed with small moments of inertia (0.010-0.050 kg·m^2), which may not be suitable depending on training goals. Thus, it remains unclear how increasing relative moment of inertia affects quadriceps EMG activity during FIT squats.

This study aimed to: 1) establish a method of quantifying relative inertial load for FIT exercise and 2) determine how increasing relative inertial load affects quadriceps EMG activity during FIT-based squats in resistance-trained subjects. We hypothesized that increasing inertial load would increase integrated EMG activity and EMG amplitude during FIT squats particularly in the eccentric phase.

METHODS

Subjects

Sixteen (8M, 8F) recreationally resistance-trained (> 6 months), but FIT-naïve, subjects participated in this study (Table 1). Participants were excluded from participation if they had a history of cardiovascular, metabolic, pulmonary, renal, or joint disease. Those with lower extremity or low back injury in the previous six months and those who were pregnant were also excluded. Participants were screened for participation by use of a health history questionnaire, a Physical Activity Readiness Questionnaire (PARQ), and a resting electrocardiogram. Those with an abnormal electrocardiogram (interpreted by a study physician) were excluded from participation. All subjects provided written informed consent prior to participation. All research procedures were approved by the University's institutional review board. All mandatory laboratory health and safety procedures were followed throughout the study.

Procedures

Participants visited the Exercise Physiology Laboratory on three separate occasions, separated

Table 1. Subject Characteristics. Body composition was determined by bioelectrical impedance analysis and muscle mass calculated as previously described (15). Strength ratio is expressed as one repetition maximum (1RM) per kg body weight. Data are presented as mean ± SD.

	Males (n = 8)	Females (n = 8)
Age (years)	25.5±4.8	22.6±2.7
Height (m)	1.78±0.11	1.66±0.05
Body mass (kg)	80.7±10.8	68.3±10.7
Body fat (%)	13.2±2.7	20.0±7.9
Muscle mass (%)	42.1±3.3	32.9±4.9
1RM (kg)	129.3±22.1	87.1±20.2
Strength Ratio	1.6±0.3	1.3±0.3

by at least three days for the first two sessions and seven days between the second and third visits. Anthropometric measures, bioelectrical impedance analysis (BIA), and one-repetition maximum (1RM) for the back squat were completed on the first visit to better characterize the resistance-trained nature of our subjects. Muscle mass was calculated from BIA data using the formula of Janssen et al 15 and was used to further characterize the resistance-trained status of participants. Visit two served as a familiarization session for FIT. The third visit was used to collect exercise performance data and quadriceps surface EMG during FIT-based squats with varying inertial loads.

Squat strength was assessed by 1RM of the barbell back squat. Subjects performed a 3 min warm-up on a cycle ergometer at a self-selected intensity, followed by lower body dynamic stretching (front and side lunges and standing quad and hamstring stretches). Subjects then performed sets of 5, 3, and 3 repetitions using with increasing external load prior to the first 1RM attempt. Subjects then performed a 1RM effort. The external load was increased by 5-10% for subsequent sets until the weight could no longer be successfully lifted. Five minutes of recovery were provided between sets. The highest load successfully lifted was recorded as the 1RM.

During the second and third visits, subjects performed a similar self-regulated warm-up. Subjects then completed five sets of five maximal effort FIT squats, with varying inertial loads in random order with at least 5 min of rest between sets. The inertial loads were based on subjects' back squat 1RM (%1RM·m^2) rather than absolute (kg·m^2) inertial load to account for differences in muscle strength among subjects. Absolute moments of inertia needed for each set were calculated by multiplying 1RM by 0.03, 0.07, 0.10, 0.12, and 0.15%. Thus, the relative inertial loads used in the present study were: 0.03±0.01, 0.07±0.01, 0.10±0.01, 0.12±0.01, and 0.15±0.02 %1RM·m^2 Absolute inertial loads used ranged from 0.025 to 0.200 kg·m^2. Subjects completed squats with varying inertial loads in a randomized order. Each set consisted of six maximal-effort repetitions (one warm-up repetition followed by five maximal concentric and eccentric movements).

It has previously been shown that lower extremity EMG activity varies with squat depth[19–21]. Therefore, subjects were instructed to descend until the thighs were parallel to the floor and to move as quickly as possible during both the concentric and eccentric phases. In addition to verbal cues from a member of the research team, we examined knee kinematics to ensure differences in knee flexion angle did not contribute to results. Sagittal plane knee joint angles were measured using a wireless electrogoniometer (Biometrics Ltd, Ladysmith, VA) placed on the right leg. The proximal end of the goniometer was placed in line with the greater trochanter and lateral epicondyle of the femur and the distal end was placed in line with the head and lateral malleolus of the fibula. Goniometer data were time synchronized with EMG data, and used to define concentric and eccentric phases. Calibration for the goniometer was completed at 0, 30, 60, and 90 degrees immediately prior to each data collection session. Data were only recorded during the final five repetitions of each set.

During the third session, surface electromyography (EMG) of the superficial quadriceps was measured (Delsys Trigno Avanti, Natick, MA). Following shaving, minor skin abrasion, and alcohol swabbing, electrodes were placed over the muscle belly of the superficial vastus lateralis (VL) of the subject's right leg according to SENIAM guidelines[16]. EMG data were recorded at 1000 Hz and band-pass filtered at 10 and 400 Hz (4th order Butterworth). The linear envelope was developed with a RMS of 125 ms and normalized to maximal voluntary isometric contraction (MVIC) at 120° knee flexion (similar to peak knee flexion during FIT squats) using a knee extension machine with a fixed arm. All sensors were secured to the body with double-sided adhesive as well as elastic wrap for the duration of data collection.

Data processing was performed using EMGworks (Delsys, Natick, MA). Data analysis was performed on five consecutive squats. Mean power frequency of the EMG signal was not different among sets 1-5 suggesting that progressive fatigue did not affect our results[17]. Additionally, mean angular velocity of the knee in the concentric and eccentric phases was determined at each inertial load as the difference in knee joint angle divided by time. Integrated EMG (iEMG) activity was determined following normalization to MVIC as an index of total muscle activity during each set. Since peak EMG activity occurred during the brief isometric period during the eccentric-concentric transition, mean EMG amplitude of the concentric and eccentric phases were analyzed independently. Spudic et al.[18] have recently reported high intra-session reliability (ICC > 0.95) for peak and mean EMG activity of the concentric and eccentric phases of five consecutive repetitions of the FIT squat. In our dataset, EMG activity was highly reproducible between repetitions (CV = 7.4, 8.0, and 9.2% for mean CON, mean ECC, and peak EMG activity, respectively).

Statistical Analyses

Statistical comparisons were made by two-way repeated measures analysis of variance with between-subjects (sex, 2 levels) and within-subjects (inertial load, 5 levels) factors with α=0.05 and Bonferroni's post-test when appropriate using SPSS v27. A Greenhouse-Geisser correction was used in cases where sphericity was violated. Data are presented as mean ± SD unless otherwise noted.

RESULTS

Effects of inertial load on sagittal plane knee kinematics

As shown in Figure 1A-B, peak knee flexion angle and total knee excursion were not significantly different among sexes or inertial loads. There was a significant main effect of load ($p < 0.001$), but not sex ($p = 0.525$) or sex x load interaction ($p=0.578$)

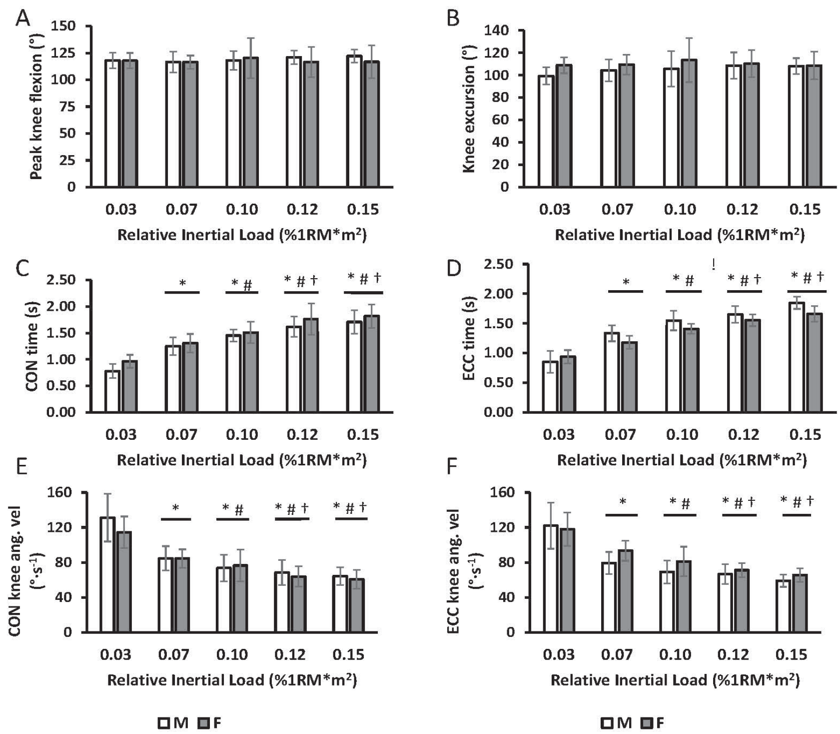

Figure 1. Sagittal plane knee kinematics during flywheel-based isoinertial squats with varying inertial loads. Sagittal plane knee joint angles were measured with electrogoniometers and time-synchronized with quadriceps electromyography. Concentric (CON) and eccentric (ECC) phase duration was defined by peak and minimal knee joint flexion. A. Peak knee joint flexion. B: Total knee excursion (range of motion). C: concentric phase duration. D: ECC phase duration. E: mean knee angular velocity in the CON phase. F: mean knee angular velocity during the ECC phase. Data are mean ± SD. n = 8 per group. * $p < 0.05$ v. inertial load of 0.03; # $p < 0.05$ v. inertial load of 0.07; † $p < 0.05$ v. inertial load of 0.10, ! significant sex x load interaction ($p < 0.05$).

on concentric phase duration (Figure 1C). Indeed, concentric phase duration was significantly ($p \leq 0.002$) greater with each increase in inertial load except for 0.15%1RM·m^2, which was not significantly different ($p = 0.072$) from inertial load of 0.12%1RM·m^2. There were significant main effects of load ($p < 0.001$) and sex x load interaction ($p = 0.016$) on eccentric phase duration (Figure 1D). Eccentric phase duration increased with increasing inertial load ($p < 0.001$) and this effect tended to be lesser in females as indicated by a significant load x sex interaction ($p < 0.05$). However, eccentric phase duration was not significantly different between sexes at any given inertial load. When collapsed across sexes, eccentric phase duration was significantly greater with each increase in inertial load ($p \leq 0.001$) with the exception of 0.15 %1RM·m^2 which was not significantly different from 0.12 %1RM·m^2 ($p = 0.161$).

There was a significant main effect of inertial load on mean knee angular velocity during the concentric and eccentric phases (both $p < 0.001$). However, there was no significant effect of sex or sex x load interaction in either the concentric ($p = 0.908$ and 0.436) or eccentric ($p = 0.210$ and 0.755) phases (Figure 1 E-F). Mean knee angular velocity in the concentric and eccentric phases was significantly lesser with each increase in inertial load ($p \leq 0.002$) with the exception of 0.15 %1RM·m^2 which was not significantly different from 0.12 %1RM·m^2 ($p = 0.080$).

Effect of inertial load on EMG activity

Figure 2 shows iEMG, peak EMG activity, and mean CON and ECC EMG activity of the VL. For iEMG activity, we noted a significant main effect of inertial load ($p<0.001$) and sex ($p=0.020$), but not a sex x load interaction ($p=0.751$). Compared to males, females demonstrated 13-47% greater iEMG, but were not significantly greater at any individual inertial load. When collapsed across sexes, iEMG was significantly lesser with inertial load of 0.03%1RM·m^2 compared to all other inertial loads ($p\leq0.001$). Similarly, iEMG was significantly lesser with inertial load of 0.07 %1RM·m^2 compared to inertial loads of 0.12 and 0.15 %1RM·m^2 ($p\leq0.014$). There was no significant difference in iEMG of the VL among the three greatest inertial loads ($p\geq0.057$).

Importantly, iEMG is a function of both EMG amplitude and duration. Due to the greater repetition duration with greater inertial loads, we sought to

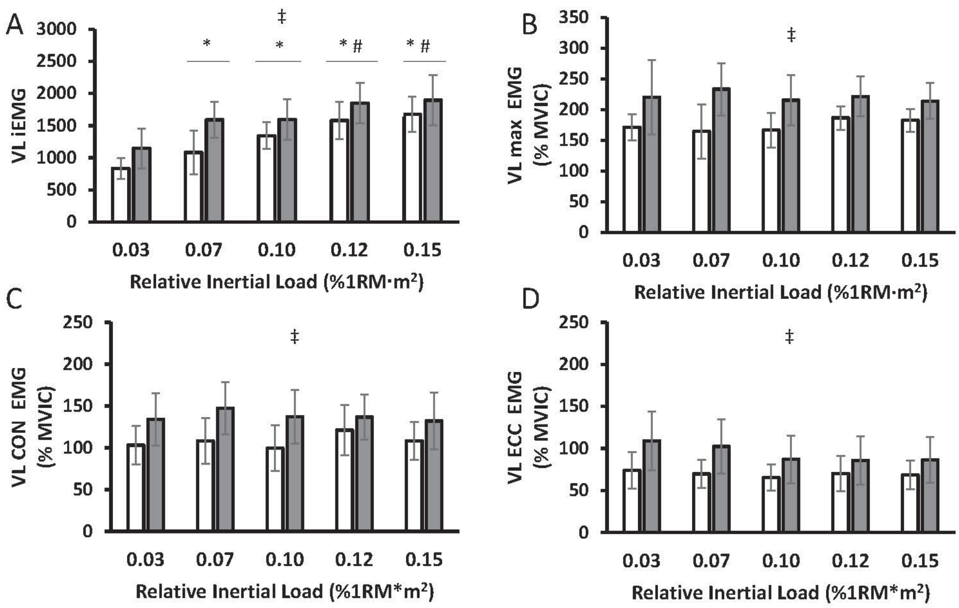

Figure 2. Electromyography activity of the vastus lateralis during flywheel-based inertial training (FIT) squats with varying inertial loads. A: Integrated electromyography (iEMG) was calculated over the course of five consecutive repetitions after normalizing to maximal voluntary isometric contraction at 120° knee flexion. B: Peak VL EMG activity at each inertial load. C: Mean concentric (CON) EMG activity of the VL with varying inertial loads. D: Mean eccentric (ECC) EMG activity of the VL with varying inertial loads. Data are mean ± SD. n = 8 per group. * $p < 0.05$ v. inertial load of 0.03; # $p < 0.05$ v. inertial load of 0.07; † $p < 0.05$ v. inertial load of 0.10; ‡ significant main effect of sex ($p < 0.05$).

determine how EMG amplitude responded to inertial load. As shown in Figure 2B, there was a significant main effect of sex (p=0.015), but not load (p=0.657) nor sex x load interaction (p=0.169) on peak EMG amplitude. Females tended to have greater peak EMG activity compared to males ($p < 0.05$).

Peak EMG amplitude occurred during the brief isometric transition from eccentric to concentric movement of each repetition. Therefore, we next assessed mean EMG amplitude during the concentric and eccentric phases of the squat. As shown in Figure 2C-D, mean EMG amplitude during the concentric (p=0.051) and eccentric (p=0.040) phases tended to be greater in females. There were no significant effects of load (p=0.123, 0.098) or sex x load interaction (p=0.173, 0.521) for mean amplitude in the concentric or eccentric phases.

DISCUSSION

Here we describe, for the first time, the effects of inertial load on quadriceps EMG activity during FIT squats. We show that knee angular velocity, but not peak knee flexion, increases with greater relative inertial load. Contrary to previous reports of increased quadriceps EMG amplitude with greater load in DCER squats[22,23], we noted no significant differences among a wide range of relative inertial loads during FIT squats. Thus, FIT may present a novel neuromuscular training stimulus where quadriceps EMG amplitude is maximal even with low inertial loads. Furthermore, we noted a significant main effect of sex where females displayed greater VL EMG amplitude than males during FIT squats which conflicts with recent reports[27].

Importantly, we implemented a novel scheme for quantifying the relative intensity of FIT exercise. Previous work has used absolute inertial load as a means of quantifying FIT exercise intensity. However, those with greater strength will likely move at a faster velocity for any given moment of inertia than those with lower strength. Due to the inverse relationship between maximal movement velocity and inertial load, it has recently been proposed that the velocity-inertia relationship may be a more appropriate means of quantifying FIT exercise intensity than absolute moment of inertia[6–10]. Here, we used relative inertial load (%1RM·m^2) to quantify relative intensity. Importantly, we noted a positive relationship between relative inertial load and repetition duration as well as an inverse relationship between inertial load and knee angular velocity. It is possible that relative inertial load may provide a utilitarian method of quantifying exercise intensity during FIT exercise. However, the exact relationship between concentric vertical velocity and relative inertial load remains unclear. Establishing this relationship may allow practitioners to easily determine appropriate loads when implementing FIT.

Results showed that repetition duration increased, and average knee angular velocity decreased during both the concentric and eccentric phases with increasing inertial load. This finding is consistent with those of Worcester et al.[11] who reported that increasing inertial load reduces rotational velocity of the knees, hips, and ankles during FIT squats. Interestingly, despite this wider range of inertial loads used in the present study, maximal knee joint angles and total excursion were not affected by inertial load, which is consistent with previous data from our lab[11]. Therefore, it appears that increased repetition duration with increasing inertial load is driven by reduced angular velocity of the lower extremity joints, not by an increased range of motion.

We noted a significant linear trend for increasing total muscle activity (iEMG) with increasing relative inertial load. Furthermore, we noted a significant effect of sex where females displayed greater peak EMG activity than males. However, no sex x load interactions were noted suggesting that males and females increase total motor unit recruitment similarly to increasing inertial load. Due to the time-dependency of iEMG measures, we also explored EMG amplitude to better understand the magnitude of motor unit recruitment during this exercise.

Peak EMG activity was achieved during the eccentric-concentric transition and therefore, likely represents EMG activity during this brief isometric phase. We noted a significant effect of sex on peak EMG activity. However, no effect of inertial load or sex x load interactions were noted. Similarly, we noted a main effect for females to display greater mean EMG activity of the VL during both the concentric and eccentric phases than their male counterparts. These data indicate that FIT elicits a greater EMG amplitude of the VL in all phases (concentric, eccentric, and isometric) in females, but the lack of response in peak EMG activity was similar between sexes.

To better understand EMG activity during the concentric and eccentric phases, we also compared mean EMG activity during each phase independently. Many of the proposed benefits of

FIT such as rapid and robust hypertrophy[24,25] have been attributed to increased loading during the eccentric phase. It has recently been reported that eccentric overload increases with increasing levels of absolute inertial load during squats[12] and knee extension[26]. Therefore, we expected that EMG activity would increase with greater relative inertial loads. Mean EMG activity during the concentric and eccentric phases were greater in females, but no effect of inertial load and no sex x load interaction was noted. Therefore, it is likely that increased iEMG activity observed with increasing inertial load is due to primarily to greater time under load rather than the degree of motor unit activation.

Our data demonstrate a different EMG amplitude response of the VL in males and females. This is in contrast to data from Mehls et al.[27] who found no differences between sexes during DCER back squats performed at 85% 1RM. The disparate findings may be due to sex-specific responses to inertial loading or biomechanical differences between back squat and FIT squats, but this remains to be explored.

It is commonly reported that EMG activity during the eccentric phase of DCER is lower than the concentric phase. Although we did not compare EMG activity between concentric and eccentric phases, the observed mean values were greater in the concentric phase for each muscle and at each inertial load. Previous work has shown greater eccentric overload and EMG activity during the eccentric phase of FIT compared to DCER exercise[4,13,28–31].

Our data appear to conflict with previous findings which described greater EMG amplitude with increasing inertial load during FIT squats[8]. We believe the disparate findings between these studies may be due to the external loads used. Due to differences in methods of quantifying inertial load between these studies, it is difficult to directly compare EMG responses in these datasets. Carrol et al. (2019) used absolute inertial loads ranging from 0.010 to 0.050 kg·m^2 and described greater VL EMG activity in the concentric phase. The minimal absolute inertial load used in the present study was 0.025 kg·m^2 and all subjects completed FIT with loads of at least 0.125 kg·m^2. It is possible that we may have observed differences in EMG amplitude among inertial loads if we had implemented lower inertial loads.

Despite increases in iEMG activity with increasing inertial load, we observed no effect of inertial load on mean EMG amplitude during the concentric phase suggesting motor unit activation is largely unchanged by inertial load. During FIT, an external force is dictated, in large part, by flywheel rotational velocity. Thus, by completing all repetitions at maximal effort during FIT, EMG amplitude may be maximal even with relatively small inertial loads. Using isokinetic knee extension exercise, Cramer et al.[32] previously demonstrated a velocity-dependence of quadriceps EMG activity with velocities ≥180°·s^{-1} (RF) and ≥ 240°·s^{-1} (VL and VM). Mechanomyography, but not EMG activity of the quadriceps, significantly increased during isokinetic knee extension suggesting that muscle contractile activity, but not motor unit recruitment is altered by increasing velocity within this range. For comparison, the average knee angular velocity in the concentric and eccentric phases in the present study was approximately 60-120°·s^{-1}. Therefore, differences in speed of muscle contraction are unlikely to contribute to our results.

Our data indicate that during maximal effort FIT, motor unit activation of the quadriceps is nearly maximal in isometric, concentric, and eccentric phases even with low relative inertial loads. In accordance with the force-velocity relationship of skeletal muscle, we hypothesize that during FIT with low relative inertial loads, there is a greater number of detached myosin cross-bridges, but not decreased neural drive, to support high movement velocity. As inertial load increases, a greater proportion of myosin heads exist in the attached, force-generating state to overcome flywheel inertia which limits movement velocity. Thus, a greater number of attached actin-myosin cross-bridges, not necessarily greater muscle recruitment, drives increased force output commonly described with higher inertial loads.

EMG amplitude is dictated both by motor unit recruitment and firing rate of individual motor units. A major limitation of the current study is that we were unable to measure the effects of inertial load on firing rates or recruitment of individual motor units. It has previously been demonstrated that those motor units recruited at lower forces demonstrate higher firing frequencies and that motor unit firing rate decreases with increasing force production[33]. Therefore, motor unit activation may be driven by greater firing rates at low inertial loads and by a greater number of motor units recruited with higher inertial loads. As in many EMG studies, we demonstrated considerable variability in EMG data which may be due, in part, to variability in relative inertial loads. Furthermore, it should be noted that we did not examine joint kinetics. It is possible that differences in movement acceleration or joint kinematics may result in similar knee joint moments (and thus similar EMG activity) with varying inertial loads.

Increasing relative inertial load decreases speed of movement during FIT squats and may be a utilitarian method for prescribing loads. EMG amplitude is nearly maximal even at very low inertial loads during FIT squats. Therefore, maximal effort FIT may provide a novel neuromuscular training stimulus. Altering relative inertial load may however be necessary to induce specific adaptations (i.e. hypertrophy, power, etc).

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DECLARATION OF INTEREST

The authors declare no conflict of interest and have no financial interests to disclose.

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