Abstract

This study aimed to establish the effect of cycling mode and cadence on torque, external power output, and lower limb muscle activation during maximal, recumbent, isokinetic cycling. After familiarisation, twelve healthy males completed 6x10 s of maximal eccentric (ECC) and concentric (CON) cycling at 20, 40, 60, 80, 100, and 120 rpm with five minutes recovery. Vastus lateralis, medial gastrocnemius, rectus femoris, and biceps femoris surface electromyography was recorded throughout. As cadence increased, peak torque linearly decreased during ECC (350 to 248 N·m) and CON (239 to 117 N·m) and peak power increased in a parabolic manner. Crank angle at peak torque increased with cadence in CON (+13°) and decreased in ECC (-9.0°). At all cadences, peak torque (mean +129 N·m, range 111 – 143 N·m), and power (mean +871 W, range 181 – 1406 W), were greater during ECC compared to CON. For all recorded muscles the crank angle at peak muscle activation was greater during ECC compared to CON. This difference increased with cadence in all muscles except the *vastus lateralis*. Additionally, peak *vastus laterallis* and *biceps femoris* activation was greater during CON compared to ECC. Eccentric cycling offers a greater mechanical stimulus compared to concentric cycling but the effect of cadence is similar between modalities. Markers of technique (muscle activation, crank angle at peak activation and torque) were different between eccentric and concentric cycling and respond differently to changes in cadence. Such data should be considered when comparing between, and selecting cadences for, recumbent, isokinetic, eccentric and concentric cycling.

Torque, power and muscle activation of eccentric and concentric isokinetic cycling

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Introduction

An eccentric muscle action occurs when the external load exceeds that produced by the muscle and causes the muscle to lengthen whilst under tension. In recent years, there has been growing interest in the benefits of eccentric training for improving sports performance [Brughelli et al. 2007; Isner-Horobeti et al. 2013; Vogt et al. 2014; Douglas et al. 2016], as well as the quality of life for individuals with neuromuscular diseases (LaStayo et al., 2014;). A muscle acting eccentrically can produce between 1 – 2 times more force than when acting concentrically, depending on contraction velocity [Westing et al. 1988; Crenshaw et al. 1995; Kellis et al. 1998; Drury et al. 2006]. Additionally, per unit of force produced, eccentric muscle contractions can display between 5 - 50 % lower surface electromyographical (sEMG) activity [Bigland-Ritchie et al. 1976; Kellis et al. 1998; Penailillo et al. 2013] and up to 50% lower metabolic cost compared to concentric contractions [Abbott et al. 1952; Bigland-Ritchie et al. 1976; Penailillo et al. 2013].

A large number of studies have used isokinetic dynamometers for eccentric exercise prescription in order to isolate specific joints and limit extraneous movements [Higbie et al. 1996; Blazevich et al. 2007; Cadore et al. 2014]. However, isokinetic dynamometry does not necessarily represent cycling or weight bearing locomotion given that multiple joints and muscle groups are activated concurrently in these human movements. The prescription of dedicated eccentric training using traditional resistance training exercises poses logistical challenges; for example, the loads prescribed could exceed what can be lifted concentrically and thus require external assistance to complete the concentric phase. In addition, there is typically a concentric component that can elicit greater metabolic stress which might be undesirable for patients with poor cardiorespiratory fitness. Additionally, for athletes specifically attempting to mechanically overload their musculoskeletal system a substantial metabolic stress may only serve to compromise existing cardiovascular training. The advent of the eccentric cycling ergometer allows repeated eccentric muscle actions to be performed with minimal concentric contractions [Abbott et al. 1952; Elmer et al. 2010]. This makes eccentric cycling a potentially valuable tool to prescribe high volume, multi-joint, eccentric exercise and to understand the potential benefits of eccentric muscle training. Typically eccentric cycling is performed in a recumbent position in order to increase torso stability via the use of a back rest [Elmer et al., 2012; Leong et al., 2013].

The benefits of eccentric cycling have become the focus of a small, but growing number of research articles. After 6-8 weeks of submaximal eccentric cycling, increases in *vastus lateralis* (VL) muscle fibre cross sectional area, leg stiffness during sub-maximal hopping, vertical jump power (external power), isometric knee extensor strength, and pennation angle of the VL and *rectus femoris* (RF) have been observed [Lastayo et al. 2000; Gross et al. 2010; Elmer et al. 2012; Leong et al. 2013]. Although these studies highlight the potency of eccentric cycling as a training stimulus, little is yet known about the characteristics of the eccentric task and the effect of manipulating cadence on torque (pedal torque unless stated otherwise) and power, and muscle activity. The majority of eccentric cycling studies have utilised a narrow range of cadences between 50 – 70 rpm; which is likely due to the desire for high volumes of eccentric contractions whilst avoiding the greater technical proficiency required for faster cadences [Green et al. 2017]. However, evidence from maximal eccentric cycling suggests that greater power outputs can be attained at higher cadences [Brughelli et al. 2013] which, after the appropriate familiarisation, could be advantageous for athletes seeking to overload the musculoskeletal system.

The underpinning torque-cadence relationship and muscle activation characteristics of eccentric cycling remain unknown. During concentric cycling, maximal torque production decreases with increasing cadence in a linear manner and the corresponding external power output is a parabolic function of cadence [McCartney et al. 1983]. However, given the fundamental force-velocity differences between eccentric and concentric contractions of individual muscle fibres *in-vitro,* it is reasonable to suggest that differences might also exist with a complex multi-joint task such as cycling [Katz, 1939]. Non-cycling *in-vivo* observations suggest that as angular velocity increases single-joint eccentric torque production remains stable or marginally increases in the knee and elbow extensors/flexors [Westing et al. 1988; Ghena et al. 1991; Kramer et al. 1993; Chapman et al. 2005; Carney et al. 2012]. The expectation during eccentric cycling is that the rate of torque decline at higher cadences will be reduced compared to concentric cycling.

At submaximal intensities, eccentric cycling elicits lower electromyographical activity than concentric cycling [Bigland-Ritchie et al 1976; Penailillo et al. 2013]. Furthermore, during submaximal eccentric cycling peak RF and VL activation occur at similar knee angles to peak torque (~ 70°) whereas during concentric cycling, at a similar workload, peak RF and VL activation occur at different knee angles compared to peak torque (~40° and ~100° respectively) [Penailillo et al., 2017]. How these differences in the magnitude and timing of lower limb muscle activation contribute to torque production at maximal intensities of eccentric cycling is not currently well understood. Furthermore it is unknown what effect altering cadence might have on these parameters during maximal recumbent isokinetic eccentric and concentric cycling. A greater understanding of these muscle activation patterns, and the corresponding torque and power production, would facilitate interpretation of any ensuing neuromuscular adaptation following a period of training and inform future research in which eccentric and concentric cycling are typically used concurrently. Consequently, the aim of this study was to establish the effect of cycling mode (concentric and eccentric) and cadence on torque, power, and lower limb muscle activation during maximal, recumbent, isokinetic cycling.

Materials and Methods

Participants

Following institutional ethical approval, twelve recreationally active males (mean ± SD; age = 27 ± 3 years; body mass = 77.3 ± 10.1 kg; stature = 1.771 ± 0.054 m) with no history of lower limb injuries or neurological disorders participated in the study. Sample size was estimated using data collected in a pilot study. Based on an expected difference of 50% between eccentric and concentric peak torque, an alpha level of 0.05, and a power (1- β) of 0.95, it was shown that a minimum of 6 subjects were required (G*Power 3.1.9.2, Faul et al., 2007). All participants provided written informed consent and completed a pre-exercise physical activity readiness questionnaire and were asked to refrain from caffeine, alcohol and exercise in the 24 hours preceding each trial. The study adhered to the guidelines set out by the World Medical Association Declaration of Helsinki.

Design

Participants reported to the laboratory on five separate occasions, separated by at least 7 days but no more than 11, to perform maximal effort cycling on a custom-built recumbent eccentric cycling ergometer (BAE systems, London, UK; Figure 1). A single exercise bout consisted of 6 x 10 s efforts, presented in a randomised, counterbalanced order (Latin squares method) across a range of cadences (20, 40, 60, 80, 100 and 120 rpm), interspersed with 5 min recovery between efforts. In order to familiarise participants with the novelty of the eccentric task a single eccentric practice bout (6 x 10 s) was conducted during each of the first three visits [Green et al. 2017]. Visit 4 comprised an eccentric experimental bout followed by 10 minutes rest and a concentric familiarisation bout. Visit five consisted of a single concentric experimental bout (6 x 10 s).

Eccentric Ergometry

All ergometry was conducted on a custom built recumbent isokinetic cycling ergometer (Figure 1). A 2200 W motor drives the cranks at a pre-set cadence and participants either pushed with, or resisted against, the direction of crank movement in order to perform concentric or eccentric cycling respectively (Figure 1). In order to prevent the possibility of knee hyper-extension the seat position was adjusted and a goniometer was used to ensure participants could not, at any point of the pedal revolution, extend their knee beyond 160° (full extension = 180°). Additionally the ergometer only functioned if the participant constantly held buttons located on each handlebar (Figure 1), should the participant release either set of buttons, the ergometer stopped immediately. Rigid, carbon fibre soled, cycling shoes (Bontrager Riot RR-45, Trek, USA) and Look Keo pedals (Look Cycle, France) were used to achieve a consistent participant-ergometer interface. Torque data were obtained from a calibrated strain gauge located on the crank arm via a wireless telemetry system (Mantracourt Electronics, UK). Torque data was digitised (200 Hz; Power 1401, Cambridge Electronic Design, UK) and acquired for off-line analysis (Spike 2 version 8.02, Cambridge Electronic Design, UK). To establish the temporal relationship between torque and surface electromyography activity (sEMG), respective values from the left-hand crank and left-side limb were used for analysis. The left side was selected because the motor of the dynamometer was situated on the right side and pilot testing revealed interference with the sEMG signal. The effect of possible leg asymmetries was considered minimal as all participants were injury free and dependent variable comparisons were made within participants. A crank angle of zero represented top dead centre and crank angle increased with a counterclockwise movement (Figure 1). Due to the isokinetic nature of the ergometer crank angles were calculated as the product of angular velocity and elapsed time from pedal top dead centre, which was detected by a reed switch and magnet attached to the ergometer and left crank, respectively.

Instantaneous power values within the pedal revolution were calculated from torque data using the following equation:

$$\text{Power (W)} = \text{Torque (N·m)} \cdot \text{Cadence (rad·s}^{-1}\text{)}$$

$$\textit{where} \quad \text{Cadence (rad·s}^{-1}\text{)} = \text{Cadence (rpm)} \cdot \frac{2\pi}{60}$$

Experimental protocol

Each session commenced with a 5-min self-selected sub-maximal warm up in the modality to be utilised for testing e.g. eccentric or concentric. This warm-up was monitored and replicated prior to each session. Prior to each 10-s maximal effort, participants were given 30-s to familiarise themselves with the up-coming cadence, by undertaking a passive (i.e. no resistance) movement, driven by the ergometer. After this, a 60-s rest was observed before commencing the 10-s maximal effort. Prior to the start of each 10-s effort, participants were instructed to relax and have their legs passively turned by the ergometer (~3 s) to ensure the correct cadence was attained. During the maximal effort participants stabilised themselves with the aid of the backrest and side handles on the ergometer (Figure 1). The elapsed time of the effort was concealed from the participant, but the participant was informed when their effort should be terminated and the ergometer was subsequently stopped. Strong, verbal encouragement was given throughout each maximal effort by the experimenter. Throughout all test trials, torque, cadence, and sEMG of selected lower limb muscles, were recorded.

Surface electromyography (sEMG)

Two, 20 mm diameter, electrodes (Ag/AgCl; Kendall 1041PTS, Covidien, Mansfield, MA, USA) with an inter-electrode distance of 20 mm were placed on the left leg according to the SENIAM guidelines for EMG placement [Hermens et al. 2000]. The muscles used for analysis were the *rectus femoris* (RF), *vastus lateralis* (VL), *biceps femoris* (BF), and medial *gastrocnemius* (MG). The skin was prepared by shaving, and abraded with an alcohol swab. The positions of the electrodes were marked with indelible ink to ensure a consistent placement between trials; a reference electrode was placed on the patella. Surface EMG signals, recorded concurrently with torque data, were sampled at 4 kHz (Power 1401; Cambridge Electronic Design, UK), then amplified (×1000; 1902, Cambridge Electronic Design, Cambridge, UK), band-pass filtered (20-2000 Hz), and rectified (Spike 2, version 8.02; Cambridge Electronic Design, UK) according to ISEK standards [Merletti, 1999]. Surface EMG signals were also notch filtered (50 Hz). Data was smoothed using a 24 Hz fourth-order Butterworth low-pass filter [Gazendam et al. 2007]. For the normalization of sEMG values participants completed three 8 s maximal voluntary concentric contractions at the start of each trial. These contractions were conducted on the aforementioned ergometer (Figure 1) at 1 rpm, starting at a crank angle of 0^{o} (top dead centre) with 5 mins rest. Using a 0.2 s root-mean–square (RMS) window, the maximum sEMG activity from the three MVC efforts for each muscle was used to obtain a reference value for normalization purposes. For temporal normalisation the filtered muscle activation data for all revolutions in the experimental sessions were plotted separately for each cadence and modality before being fitted with a 3^{rd} order sum of sines model to determine muscle activation patterns (Matlab R2015b, Mathworks, USA).

Statistical analysis

All statistical testing was performed using SPSS 22 (IBM, New York, USA). To detect any effect of cadence and/or muscle contraction type on peak torque, peak power, sEMG peak amplitude, angle of peak torque, and angle of peak sEMG, data were analysed using a two-way repeated measures analysis of variance (ANOVA). Peak sEMG data from the sum of sines model was used for analysis. Where appropriate, pairwise differences were located using multiple t-tests corrected by the Ryan-Holm Bonferroni adjustment. Effect sizes (Cohen's *d*) were calculated for all pairwise comparisons. Pearson Correlation Coefficients were used to assess the strength of association between ECC and CON peak torque at each cadence. The magnitude of correlation was interpreted as follows; small (r = 0.10 – 0.29), moderate (r = 0.30 – 0.49), large (r = 0.5 – 0.69), very large (r = 0.70 – 0.89), and extremely large (r >= 0.90) [Hopkins et al., 2009]. Significance was set at an alpha level of 0.05. Greenhouse-Geisser corrections were applied to significant F-ratios that did not meet Mauchly's assumption of sphericity. All data is presented as mean ± standard deviation unless stated otherwise. For statistical testing when crank angles spanned 360°, i.e. differences in crank angles were geometrically minimal but numerically large, crank angles were uniformly adjusted prior to analysis. All crank angles were anchored to a functionally redundant part of pedal cycle i.e. the section of the pedal cycle that clearly dissociated the end of one cycle to the start of the next for the variable in question. This ensured that greater and lesser crank angles influenced the group mean in manner that was functionally accurate during statistical testing. Subsequent to statistical analysis crank angles were converted back to geometrically correct values for reporting.

Results

Torque

Peak torque was consistently higher in ECC compared to CON (average difference, 123 N·m, range 110 – 143 N·m, $F_{(1, 11)} = 86.5$, $p < 0.05$) at all cadences ($p < 0.05$; $d = 1.7 - 3.2$). There was a significant main effect of cadence on peak torque ($F_{(5, 55)} = 35.6$, $p < 0.05$; Table 1). As cadence increased, peak torque reduced in both ECC ($F_{(5, 55)} = 10.6$, $p < 0.05$) and CON ($F_{(1.8, 19.8)} = 122.7$, $p < 0.05$). This decrease in torque was linear for both ECC ($r^2 = 0.99$) and CON ($r^2 = 0.99$; Figure 2). However, there was no significant modality*cadence interaction effect on peak torque ($F_{(5, 55)} = 1.1$, $p > 0.05$). There was a very large relationship between ECC and CON peak torque at low cadences (20 and 40 rpm, $p < 0.05$). At faster cadences this relationship was only moderate (60 rpm), large (80 rpm), and small (100 and 120 rpm) ($p > 0.05$, Table 1).

Crank angle at peak torque

Crank angle at peak torque was significantly greater during CON compared to ECC ($F_{(1, 11)} = 134.7$, $p < 0.05$; Table 1), at all cadences ($p < 0.05$; $d = 1.8 – 4.2$). There was no main effect of cadence on crank angle at peak torque ($F_{(5, 55)} = 0.6$, $p > 0.05$). However, there was a modality*cadence interaction effect on crank angle at peak torque ($F_{(5, 55)} = 13.4$, $p < 0.05$). As cadence increased crank angle at peak torque decreased in ECC ($F_{(2.7, 29.4)} = 4.3$, $p < 0.05$) and increased in CON ($F_{(2.7, 30.1)} = 9.9$, $p < 0.05$).

Power

Peak power was greater during ECC compared to CON ($F_{(1, 11)} = 94.2$, $p < 0.05$), across all cadences ($p < 0.05$; $d = 1.4 – 3.3$). Furthermore, peak power increased with cadence ($F_{(5, 55)} = 143.9$, $p < 0.05$; Figure 3, Table 1) for both ECC ($F_{(2.2, 24.7)} = 83.0$, $p < 0.05$) and CON ($F_{(5, 55)} = 250.0$, $p < 0.05$). There was a significant modality*cadence interaction effect as peak power increased with cadence to a greater extent during ECC compared to CON ($F_{(2.5, 27.8)} = 28.6$, $p < 0.05$; Figure 3). The shape of this increase was parabolic for both ECC and CON. This is illustrated by the conversion of the linear torque-cadence relationship to the concomitant power-cadence relationship and displayed in Figure 3.

Electromyography - rectus femoris

Surface EMG data over a pedal revolution at each tested cadence is displayed in Figure 4. Peak RF activation occurred at significantly greater crank angles in ECC compared to CON ($F_{(1,11)} = 7.1$, $p < 0.05$). Pairwise comparisons located this difference at 60 rpm, 100 rpm, and 120 rpm ($p < 0.05$; $d = 1.5$, 1.3, and 1.5 respectively; Table 2). There was no main effect of cadence on the crank angle at peak RF activation ($F_{(1.8,20.2)} = 2.0$, $p > 0.05$). Although the crank angle at which peak RF activation occurred tended to increase at higher cadences in CON whilst decreasing in ECC, as evidenced by a significant modality*cadence interaction effect ($F_{(1.6, 17.2)} = 5.7$, $p < 0.05$). There was no main effect of cycling modality on peak RF amplitude ($F_{(1,11)} = 0.9$, $p > 0.05$). However, peak RF amplitude did increase at higher cadences ($F_{(2.1,22.9)} = 4.1$, $p < 0.05$). This increase was similar in ECC and CON as highlighted by a non-significant modality*cadence interaction effect ($F_{(2.9,32.4)} = 2.8$, $p > 0.05$).

Biceps femoris

Overall there was no main effect of cycling modality ($F_{(1, 11)} = 4.1$, $p > 0.05$) or cadence ($F_{(1.9, 20.9)} = 2.9$, $p > 0.05$) on the crank angle at which peak BF activation occurred. However, there was a significant modality*cadence interaction effect on the crank angle of peak BF activation ($F_{(1.6, 17.6)} = 9.2$, $p < 0.05$). At higher cadences the crank angle of peak BF activation increased in ECC ($F_{(1.7,18.5)} = 4.1$, $p < 0.05$) and decreased in CON ($F_{(1.3, 14.1)} = 6.4$, $p < 0.05$). Pairwise comparisons located this difference at 100 rpm ($p < 0.05$; $d = 1.5$, Table 2). Peak BF amplitude was greater during CON compared to ECC ($F_{(1,11)} = 17.9$, $p < 0.05$), at all cadences ($p < 0.05$; $d = 1.0 – 1.8$). There was no main effect of cadence on peak BF amplitude ($F_{(1.3,14)} = 3.2$, $p > 0.05$) and there was no modality*cadence interaction effect on peak BF amplitude ($F_{(1.5,17.2)} = 2.5$, $p > 0.05$).

Vastus lateralis

The crank angle at which peak VL activation occurred was significantly greater in ECC compared to CON ($F_{(1, 11)} = 10.8$, $p < 0.05$) at 20 rpm ($p < 0.05$; $d = 2.8$, Table 2). There was no main effect of cadence on the crank angle of peak VL activation ($F_{(2.3, 25.6)} = 0.6$, $p > 0.05$). Additionally, there was no modality*cadence interaction effect on the crank angle of peak VL activation ($F_{(2.5, 27.6)} = 0.5$, $p > 0.05$). Peak VL amplitude was greater during CON compared to ECC ($F_{(1,11)} = 52.3$, $p < 0.05$) at all cadences ($p < 0.05$; $d = 1.3 – 2.5$). There was no main effect of cadence on peak VL amplitude ($F_{(2, 22)} = 1.1$, $p > 0.05$), however, there was a significant modality*cadence interaction effect ($F_{(5,55)} = 3.9$, $p < 0.05$). As cadence increased VL activation increased in CON ($F_{(5,55)} = 2.9$, $p < 0.05$) between 20 – 40 rpm and 40 – 60 rpm but remained similar across all cadences in ECC ($F_{(5, 55)} = 1.4$, $p > 0.05$).

Medial gastrocnemius

Crank angle at peak MG activation was greater during ECC compared to CON ($F_{(1, 11)} = 102.4$, $p < 0.05$). This was significant at all cadences between 40 – 120 rpm ($p = < 0.05$; $d = 1.2 – 4.3$, Table 2). There was a significant main effect of cadence on the crank angle of peak MG activation ($F_{(5,55)} = 22.2$, $p < 0.05$) which increased with cadence in both ECC ($F_{(5,55)} = 24.0$, $p < 0.05$) and CON ($F_{(3,33.2)} = 3.2$, $p < 0.05$). However, this increase was greater in ECC as evident by a significant modality*cadence interaction effect ($F_{(5,55)} = 10.3$, $p < 0.05$). There was no main effect of cycling modality on peak MG amplitude ($F_{(1,11)} = 3.6$, $p > 0.05$). However, there was a main effect of cadence on peak MG amplitude ($F_{(5,55)} = 10.5$, $p < 0.05$). Peak MG amplitude increased with cadence in ECC ($F_{(2.5,27.8)} = 7.5$, $p < 0.05$) and CON ($F_{(1.8,19.5)} = 6.5$, $p < 0.05$). This increase was similar between ECC and CON as evidenced by a non-significant modality*cadence interaction effect ($F_{(5,55)} = 0.8$, $p > 0.05$).

Discussion

This investigation examined the differences in torque production, power output, and lower limb muscle activation during maximal eccentric and concentric isokinetic cycling and their changes over a range of cadences. For the first time, we present data showing 1) the relationship between pedal cadence, torque, and power output, which was similar between eccentric and concentric isokinetic cycling; 2) torque decreased linearly with cadence, and power increased in a parabolic fashion; 3) at equivalent cadences, the absolute peak torque was 1.4 – 2.1 times greater during ECC compared to CON; 4) peak torque occurred at smaller crank angles during ECC compared to CON whereas peak muscle activation (RF, VL, MG, BF) occurred at greater crank angles in ECC compared to CON; and 5) concentric cycling elicited greater peak muscle activation in the VL and BF.

As cadence increased, peak torque decreased in both ECC and CON. The gradient of this trend line was similar between groups (ECC, -1.0246; CON -1.2486) and is similar to the rate of torque decline previously described in concentric cycling (-1.016) [McCartney et al. 1983]. Additionally, in further agreement with our findings, multiple studies have observed a linear decline in torque with increasing cadences during concentric cycling [McCartney et al. 1983; Vandewalle et al. 1987; Seck et al. 1995; Capmal et al. 1997; Dorel et al. 2010]. Importantly, this is the first study to observe a similar relationship during eccentric cycling. Our observed eccentric torque-cadence relationship deviates from the classic *in-vitro* force-velocity, and the single joint *in-vivo* torque-velocity relationships. As contraction velocity increases, in-vitro force increases [Katz 1939] and individual joint torque marginally increases or remains constant [Westing et al. 1988; Ghena et al. 1991; Kramer et al. 1993; Chapman et al. 2005; Carney et al. 2012]. Evidence of the opposite, i.e. decreasing joint torque as muscle lengthening velocity increases, is limited, although it has been observed in the elbow flexors [Colson et al. 1999]. Although it is important to note that the range of lengthening is not uniform across these studies. Our findings clearly demonstrate a linear decrease in eccentric torque from slow cadences (20 rpm) to fast cadences (120 rpm), which is comparable to concentric cycling [McCartney et al. 1983], i.e. the torque-velocity relationship is inverse, linear and does not mirror the *in vitro* or isolated muscle *in vivo* torque-velocity relationship. This similarity between the ECC and CON torque-cadence relationships, combined with their distinctly different *in vitro* curves, suggests that this relationship is shaped by a technique dependant cycling factor rather than an intrinsic muscle characteristic associated specifically with either eccentric or concentric muscle actions [McDaniel et al. 2014; Bobbert et al. 2016].

Similar eccentric - concentric torque ratios to the current study have been observed during isolated knee extension; at 30, 150, and 270 deg·s^{-1} maximal knee extensor eccentric torque can exceed concentric torque by 1.2, 2.0 and 2.3 times, respectively [Westing et al. 1988; Kellis et al. 1998]. In absolute terms the torque observed in the current study exceeds that previously observed with eccentric (up to 299 N·m) and concentric (up to 237 N·m) muscle actions of the knee extensors [Westing et al. 1988; Pain et al. 2013]. This is likely due to the cumulative contribution of multiple leg extensor muscles in the current study, compared to isolated knee extensors. However, when considered as a tangential force (crank length = 175 mm), peak ECC torque in the 20 rpm condition equates to ~2000 N which, given the body mass of the cohort, is approximately 2.6 times body weight, and similar to the force observed during maximal vertical jumping [Cuk et al. 2014]. Although the contribution of the stretch shortening cycle to this force will differ between jumps and eccentric cycling. This highlights the potency of eccentric cycling as a potential training stimulus – the participant can achieve high levels of peak torque/force that are seen during maximal jumps, but in a more repetitive manner and a closed kinetic chain movement pattern.

At each cadence, peak power was higher in ECC compared to CON and this difference increased as cadence increased. Our observed peak eccentric power values are approximately twice that previously described by Brughelli et al. [2013]. We speculate that such a discrepancy in torque could be due to the recumbent nature of the bike used in the current study which provides a fixed "backrest" to push against thus potentially augmenting torque production when compared with an upright bike. However, our observed concentric peak power values are similar to previous work in upright cycling [Martin et al. 2001]. It is possible that the effect of a recumbent cycling position on power output might be different between eccentric and concentric cycling. Given the discrepancy in absolute torque production between modalities it is possible that the greater stability offered by a backrest might be more beneficial during eccentric cycling, however, further investigation would be required to determine if such an effect exists.

In agreement with previous literature, peak power during CON was greatest between 100-120 rpm [McCartney et al. 1983] – the peak of the parabolic relationship between cadence and power. In contrast, the parabolic trend line between peak power and cadence during ECC was still increasing at 120 rpm (our highest cadence used), which suggests the optimum cadence for power production occurs at higher cadences in eccentric cycling, and beyond the range studied here. Due to safety features on our cycle ergometer it was not possible to investigate cadences greater than 120 rpm. At cadences above 60 rpm, peak power was greater during ECC (~1900 W) compared to that attained at any cadence during CON (~1400 to 1500 W at 100 to 120 rpm). Therefore, if achieving peak power is the primary aim of a training session, maximal eccentric cycling at cadences above 60 rpm would provide a more potent stimulus (in terms of mechanical load to the lower limb) than maximal concentric cycling at any cadence.

The weakening correlation between ECC and CON peak torque as cadence increases indicates a potential divergence in the mechanisms of torque production. Greater differences in the crank angle at peak torque between ECC and CON at higher cadences also support the notion that mechanisms of torque production diverge (n.b. Due to the isokinetic nature of the ergometer the angle at peak torque is equivalent to the angle at peak power). Additionally, eccentric torque production can display greater variability at higher cadences [Green et al., 2017], which suggests that the technical characteristics of eccentric cycling, such as muscle activation strategies, might limit torque production to a greater extent at faster cadences. Our data show that as cadence increases during ECC the crank angle of peak RF activation and peak torque converge. In contrast, as cadence increases during CON the crank angle of peak RF activation and peak torque diverge. This suggests the RF might play a more prominent role in torque production during eccentric, compared to concentric, cycling, especially at higher cadences. Furthermore, peak RF activation was similar between ECC and CON, whereas peak VL activation was greater in CON. This greater eccentric muscle activation in the RF (relative to the concentric equivalent) also indicates a greater role for the RF in eccentric cycling when compared with concentric cycling. Identification of lower limb kinematics would help to further elucidate muscle activation during eccentric cycling.

Although the crank angle of peak VL activation was greater during ECC compared to CON, when considered relative to the crank angle at peak torque it occurred earlier in the pedal cycle during both modalities. As cadence increased peak VL activation occurred progressively earlier than peak torque in both ECC and CON. This occurred due to peak torque occurring later in the pedal cycle as cadence increased in both ECC and CON (lesser and greater crank angles respectively due to the difference in pedal direction). This mirrors the findings of Bobbert et al. [2016] who observed that as cadence increased during concentric cycling muscle activation occurred earlier in the pedal cycle, relative to peak torque, to allow sufficient time for muscle de-activation to occur (muscle activation dynamics). Our observation of earlier VL activation relative to peak torque at higher cadences supports the theory that muscle activation dynamics might contribute to the decrease in torque at higher cadences during concentric cycling. Furthermore, our data suggests that similar muscle activation dynamics might also contribute to the decline in torque observed at greater cadences during eccentric cycling.

Also of note was the difference in the crank angle of peak activation between ECC and CON in the VL, RF, and MG. In the longer term these differences might affect the ability of the lower limb to express force at different crank angles, which should be considered when interpreting changes in torque or power after eccentric and concentric isokinetic cycling. With respect to the implications for training, it is possible that such differences in crank angles at peak activation might induce differing adaptations within the muscle. Improvements in strength after isometric knee extensor training can be specific to the angle utilised in training [Kitai et al. 1989]. Also, increases in squat performance can be specific to the depth of squat utilised during training [Rhea et al. 2016]. Therefore when using isometric tests within task specific ranges of motion to examine the efficacy of an eccentric or concentric isokinetic cycling program it might be prudent to utilise a range of knee angles.

To conclude, maximal recumbent eccentric cycling elicits power output and torque that is approximately two-fold greater than that observed during concentric cycling. The shape of the torque-cadence and power-cadence relationships is similar between eccentric and concentric cycling. In contrast to previous *in-vivo* observations of the eccentric force-velocity profile, a linear decrease in torque occurs during eccentric cycling as movement velocity (cadence) increases. A very similar decrease is seen during concentric cycling which suggests the shape of this relationship is not controlled by the type of muscle contraction, at least in this closed-chain cycling movement pattern. Peak torque was elicited at lesser crank angles during eccentric cycling compared to concentric cycling, a difference that increased with cadence. Additionally, peak muscle activation occurred at greater crank angles during eccentric, compared to concentric, cycling, a difference that also increased with cadence. These differences in muscle stimulation should be considered when comparing these two exercise modalities or when utilising them for training as they might impact subsequent adaptation.

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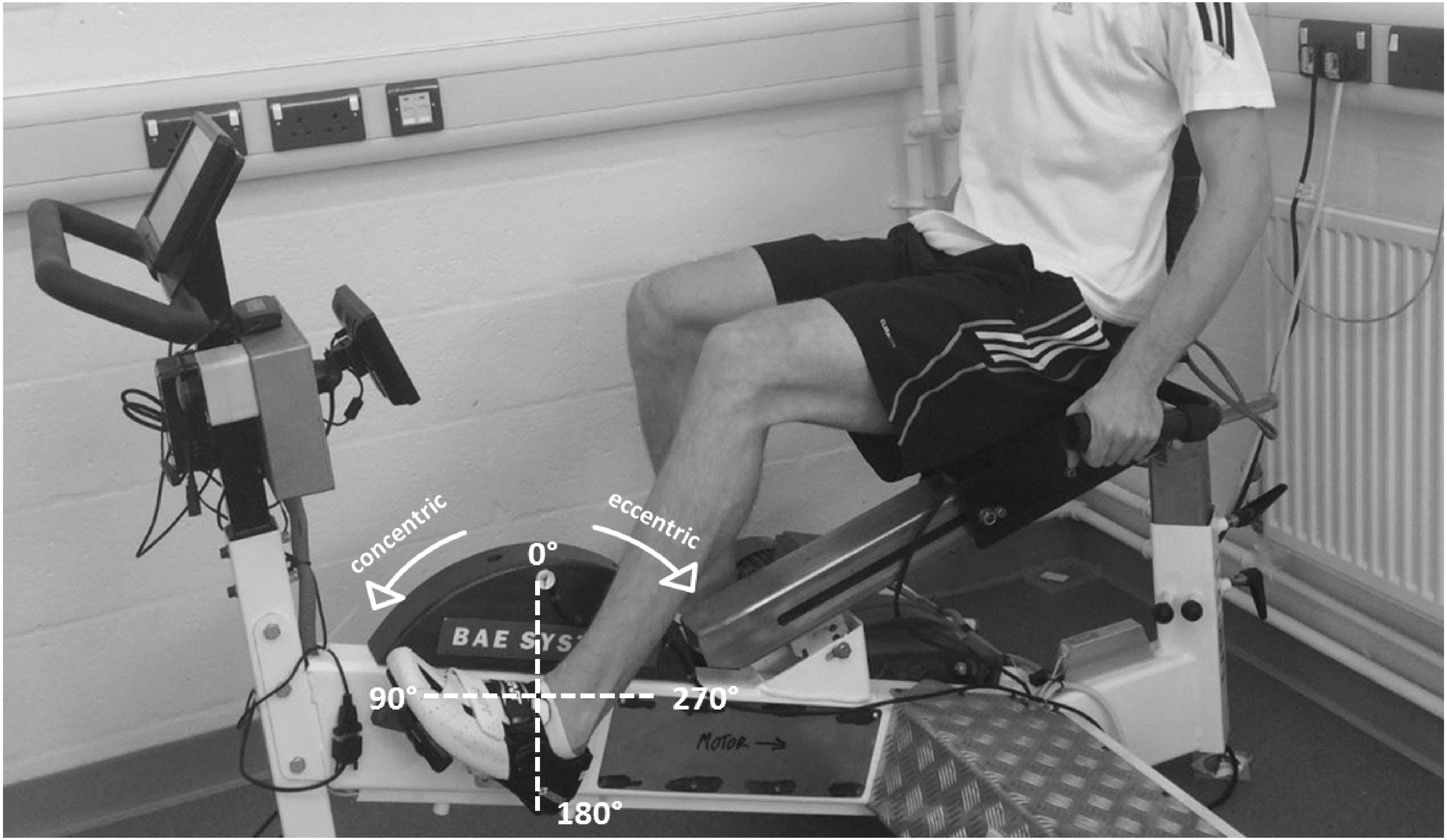
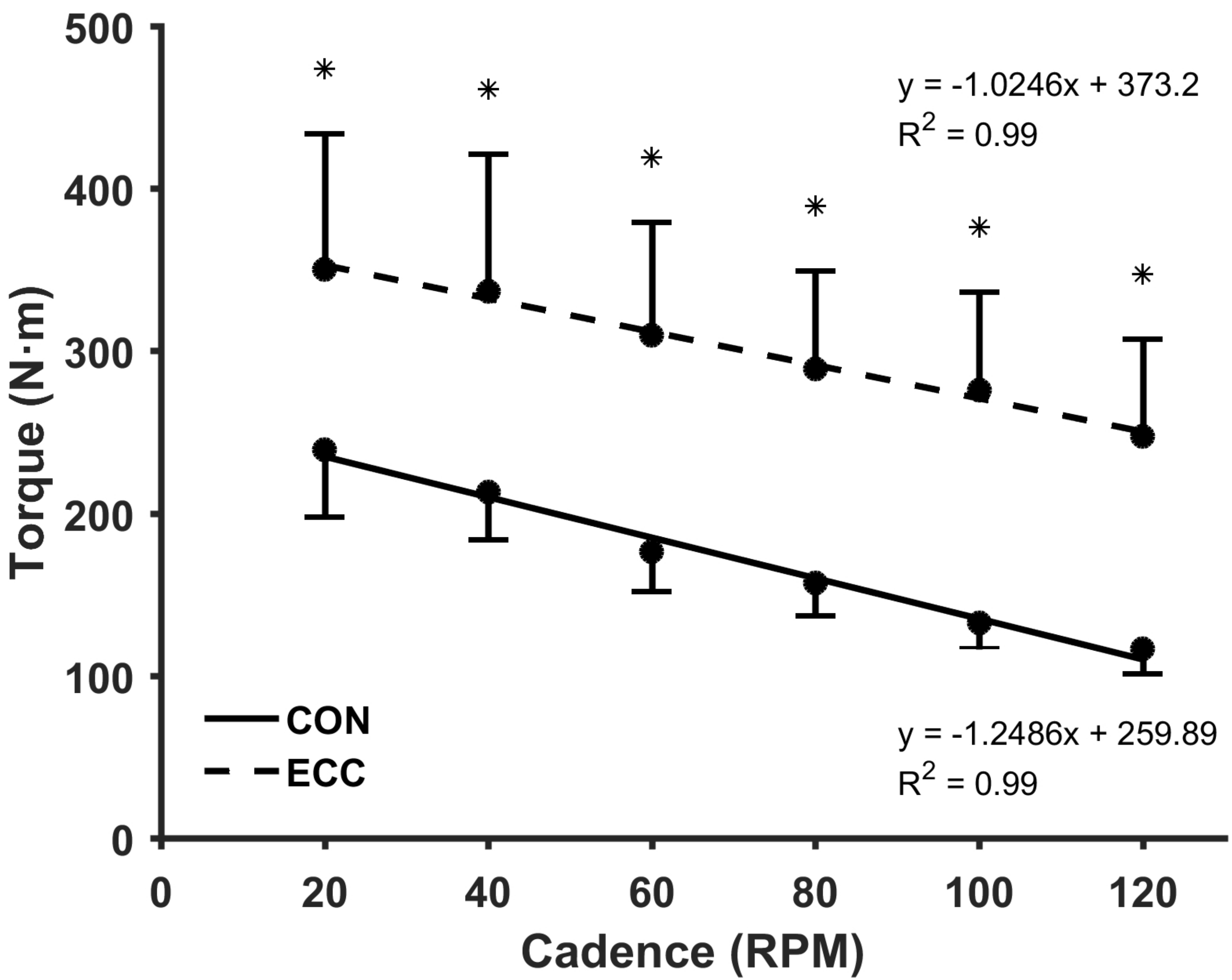
Fig.1 Isokinetic eccentric cycle ergometer. Depending on direction of crank rotation the participant either pushes with or resists against the pedals to conduct concentric or eccentric muscle actions respectively

Fig.2 Peak instantaneuous eccentric and concentric torque during isokinetic eccentric (--) and concentric (-) cycling at candences between 20 – 120 rpm (n = 12). Values are mean ± SD. * denotes significant difference at $p < 0.05$. Data points have been fitted with a linear line of best fit

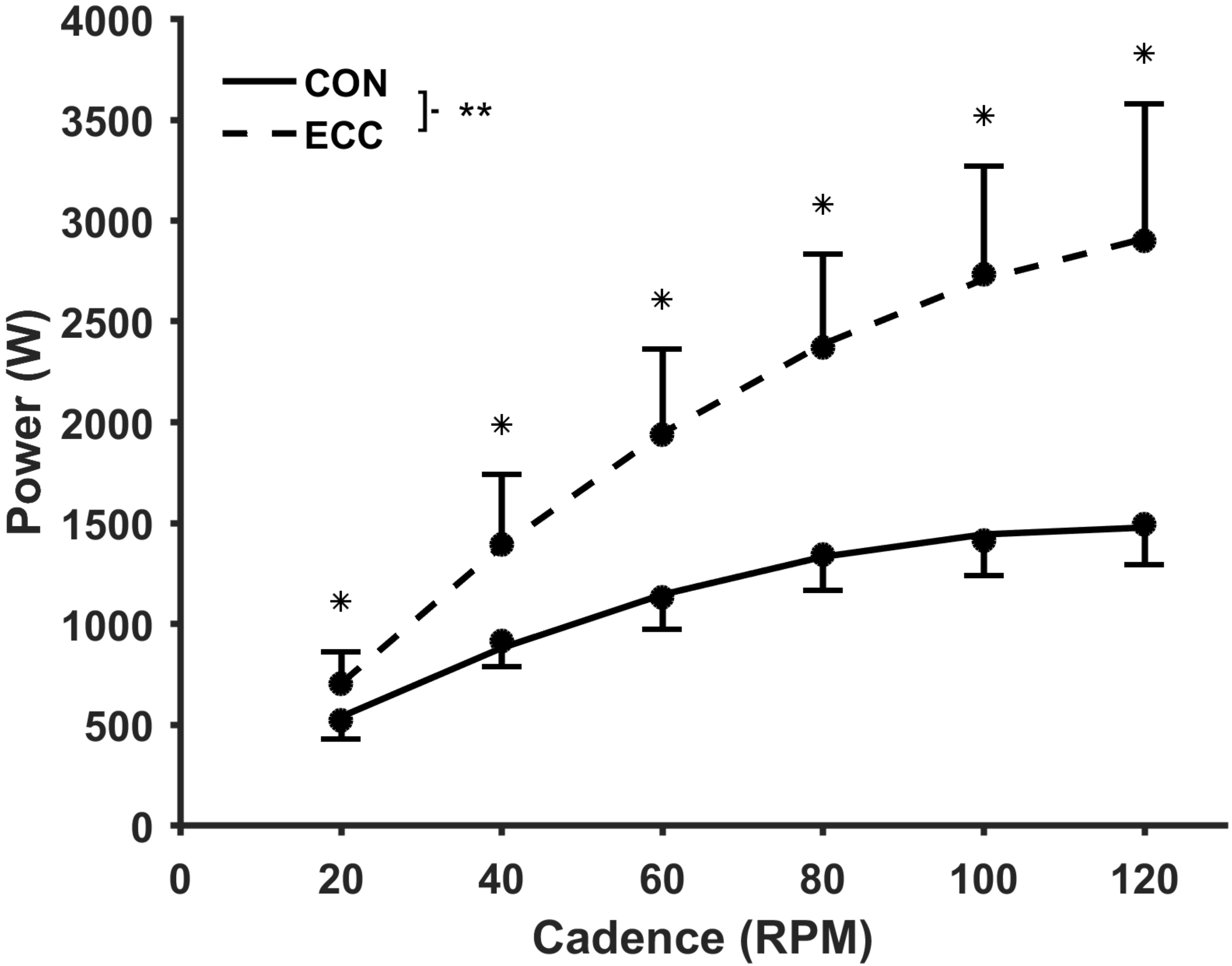
Fig.3 Peak instantaneuous eccentric and concentric power during isokinetic eccentric (--) and concentric (-) cycling at cadences between 20 – 120 rpm (n = 12). Values are mean ± SD. * denotes significant difference at $p < 0.05$. Data point have been fitted with a 2nd order polynomial line of best fit ** denotes significant interaction of cadence and contraction type at $p < 0.05$

Fig.4 Average sEMG activation of the *biceps femoris*, *vastus lateralis*, *rectus femoris* and *medial gastrocnemius* during a pedal revolution across increasing cadences (n = 11). The pedal revolution is defined as 360^{o} of rotation from top dead centre (0) to an identical position on the subsequent cycle (360). Horizontal dashed (--) and solid (-) lines represent the muscle activation of eccentric and concentric cycling respectively. Vertical dashed (--) and solid (-) lines represent the crank angle of peak torque during eccentric cycling and concentric cycling respectively at the relevant cadence

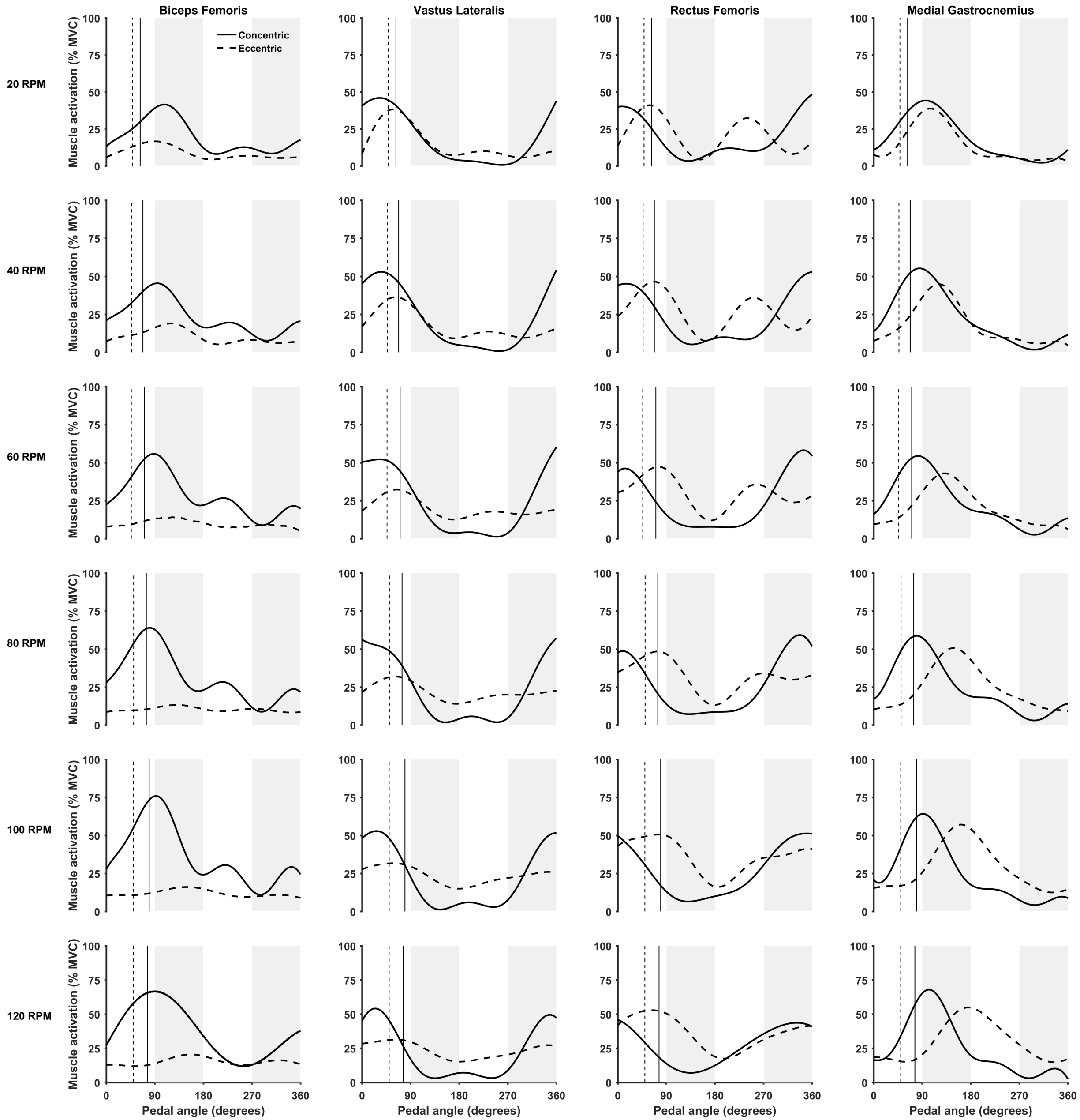

concentric
eccentric
0°
90°
270°
180°
BAE SYS
MOTOR →



500
400
300
200
100
0
Torque (N·m)
0
20
40
60
80
100
120
Cadence (RPM)
y = -1.0246x + 373.2
R^2 = 0.99
y = -1.2486x + 259.89
R^2 = 0.99
CON
ECC



CON
ECC
**
Power (W)
4000
3500
3000
2500
2000
1500
1000
500
0
0
20
40
60
80
100
120
Cadence (RPM)



Biceps Femoris
Vastus Lateralis
Rectus Femoris
Medial Gastrocnemius
Concentric
Eccentric
20 RPM
40 RPM
60 RPM
80 RPM
100 RPM
120 RPM
Muscle activation (% MVC)
Pedal angle (degrees)
0
90
180
270
360
0
25
50
75
100

Table 1. Peak power and peak torque data across all tested cadences.

Cadence (RPM)	Peak Power (W)		Peak Torque (N·m)		
	ECC	**CON**	**ECC**	**CON**	**Pearson Correlation**
20	700 ± 159*	519 ± 93	350 ± 83*	239 ± 42	0.89‡
40	1391 ± 346*	911 ± 127	337 ± 84*	213 ± 29	0.70‡
60	1935 ± 425*	1130 ± 160	310 ± 69*	176 ± 25	0.45
80	2370 ± 461*	1342 ± 177	289 ± 60*	157 ± 20	0.51
100	2733 ± 535*	1411 ± 173	276 ± 61*	132 ± 16	0.21
120	2898 ± 679*	1492 ± 201	248 ± 59*	117 ± 15	0.24

Mean (± 1 SD) eccentric and concentric peak instantaneous torque and peak instantaneous power output values and Pearsons correlation coefficients between ECC and CON peak torque measured over a range of cadences (20 – 120 rpm). * denotes significant difference to equivalent concentric value ($p < 0.05$). ‡ denotes significant Pearson correlation coefficient ($p < 0.05$).

Table 2. Pedal angle at peak sEMG amplitude and peak torque data.

Cadence (RPM)	Pedal angle at peak sEMG (°)								Pedal angle at peak torque (°)	
	Rectus Femoris		*Biceps Femoris*		*Vastus Lateralis*		*Medial Gastrocnemius*			
	ECC	**CON**	**ECC**	**CON**	**ECC**	**CON**	**ECC**	**CON**	**ECC**	**CON**
20	1 ± 82	8 ± 37	92 ± 22	112 ± 71	61 ± 11*	20 ± 17	108 ± 16	102 ± 25	50 ± 10**	64 ± 5
40	19 ± 75	3 ± 40	117 ± 20	94 ± 54	80 ± 57	34 ± 12	118 ± 19*	93 ± 23	48 ± 11**	69 ± 6
60	60 ± 51*	351 ± 38	116 ± 32	90 ± 52	68 ± 73	23 ± 21	135 ± 18*	85 ± 16	47 ± 10**	72 ± 6
80	59 ± 52	345 ± 35	140 ± 56	79 ± 51	83 ± 89	10 ± 22	146 ± 27*	84 ± 19	42 ± 14**	77 ± 6
100	51 ± 55*	348 ± 38	153 ± 48*	80 ± 51	62 ± 76	19 ± 20	162 ± 24*	97 ± 25	40 ± 12**	78 ± 5
120	46 ± 55*	328 ± 53	156 ± 71	91 ± 45	64 ± 80	17 ± 22	179 ± 17*	105 ± 17	41 ± 11**	77 ± 10

Mean (± 1 SD) eccentric and concentric pedal angles at peak sEMG amplitude and peak instantaneous torque measured over a range of cadences (20 – 120 rpm). * denotes significant difference to equivalent concentric value ($p < 0.05$). ** denotes significant difference to equivalent concentric value ($p < 0.001$).

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