

Assessing, Understanding and Improving the Limits of Neuromuscular Function on a Stationary Cycle Ergometer

A thesis submitted in fulfilment of the requirements of the degree of

DOCTOR OF PHILOSOPHY

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Abstract

Adequate neuromuscular function (i.e. the combined work of the central nervous system and skeletal muscle to permit movement) over the life span is essential for the effective execution of functional tasks. Tasks performed can range from those required as part of daily life (e.g. rising from a chair and climbing stairs) to those completed in the sporting arena (e.g. jumping, running and cycling). Stationary cycle ergometers can be used to make an ecologically valid, safe and accurate assessment of the limits of the neuromuscular function of the lower limbs, for a wide range of populations. The force and power transferred to the cranks of the ergometer are determined by various physiological, biomechanical and motor control factors. Physiological factors affecting neuromuscular function encompass the mechanical properties (i.e. force-velocity, length-tension and force-frequency relationships) and active state of the various lower limb muscles. Biomechanical factors include the magnitude and orientation of the forces transmitted to the crank and kinematics of the lower limb joints. Finally, motor control factors include the coordination between muscles and joints and movement variability, which reflects how the central nervous system manages the abundance of motor solutions offered by the human body to produce the pedalling movement.

Within this thesis, a series of three studies were conducted, first to assess the limits of lower limb neuromuscular function, secondly to improve the limits of neuromuscular function using two 4-week interventions and thirdly to investigate how ankle taping affects the limits of neuromuscular function. Force-velocity (F-V) tests were performed on stationary cycle ergometers for all studies. Variables assessed in the first study included torque-cadence (T-C) and power-cadence (P-C) relationships; values predicted from these relationships to quantify the limits of NMF (i.e. maximal power, P_{max} ; optimal cadence, C_{opt} ; maximal torque, T_0 ; maximal cadence, C_0); crank torque profiles; EMG and co-activation profiles of the lower limb muscles. Also, the variability of torque, EMG and co-activation profiles was investigated. The same variables listed above were assessed in studies two and three with the addition of lower limb joint kinematics.

More specifically, the first study of this thesis aimed to measure variations in torque and EMG between maximal and non-maximal pedal cycles obtained during a F-V test performed on a stationary cycle ergometer, then to compare the ability of two modelling procedures to predict T-C and P-C relationships and quantify the limits of neuromuscular function. T-C and P-C relationships, the associated crank torque, and EMG of the lower limb muscles were assessed during the F-V test in 17 non-cyclist males. Selection of pedal cycles corresponding to maximal values of torque at regular intervals (every 5 rpm) over a wide range of cadences (40-180 rpm) resulted in average torque $5 \pm 5\%$ greater than that calculated from non-maximal pedal cycles.

The greater average torque was associated with higher values of peak crank torque ($+6 \pm 9\%$), peak EMG of the lower limb muscles ($+2 \pm 9\%$) and co-activation of all muscle pairs ($+12 \pm 10\%$). Less between-cycle variability was also observed for crank torque and EMG profiles for maximal pedal cycles. Higher order polynomials provided a better fit for T-C and P-C relationships, evidenced by higher r^2 and SEE and lower torque and power residuals, indicating that the shapes of these relationships are not linear nor symmetrical parabolas as previously reported. Further, low order polynomials resulted in an overestimation of torque and power values at low (<50 rpm, including T_0) and high (>170 rpm, including C_0) cadences. This study showed that participants were not able to maximally and optimally activate their lower limb muscles during each pedal cycle, which affected their ability to produce maximal levels of torque and power. Further, T-C relationships are not always perfectly linear and P-C relationships do not exhibit a symmetrical parabola as it has been commonly assumed. As such the collection of a large number of data points, the implementation of maximal data selection procedures and higher order polynomials used in this study provided a better reflection of the torque and power producing capabilities of the lower limb muscles on a stationary cycle ergometer.

Study two aimed to investigate the effect of two 4-week ballistic training interventions on a stationary cycle ergometer on the limits of neuromuscular function. Training consisted of brief all-out efforts performed against high resistances (RES; $n = 9$) or at high cadences (VEL; $n=8$) on a stationary cycle ergometer. Power production at training-specific cadences, P_{\max} , C_{opt} , C_0 and T_0 and variability in crank torque, EMG, co-activation and kinematic profiles were assessed before and after training. Lower limb volumes was also assessed before and after training. To enable the effect of training to be assessed at cadences for which the different interventions would have the greatest influence (i.e. at low to moderate cadences for RES and moderate to high cadences for VEL), variables were compared pre and post-training at intervals of 60-90 rpm and 160-190 rpm. Participants in RES trained at cadences ranging from 0 to 122 ± 15 rpm while those in VEL trained at cadences ranging from 131 ± 5 to 211 ± 10 rpm. A moderate $7 \pm 6\%$ improvement in power at cadences ranging from 60 to 90rpm was observed following the RES intervention. There was a moderate increase in T_0 ($+25 \pm 19\%$) for RES, while a small increase in P_{\max} ($+4 \pm 5\%$) and small reduction in corresponding C_{opt} (-3 ± 5 rpm) was observed. The increase in power observed following RES intervention was associated with an $11 \pm 13\%$ increase in peak crank torque, a reduction in ankle joint range of motion ($-6 \pm 4^\circ$), an increase in hip joint range of motion and an increased co-activation of the VAS-HAM, GAS-TA and GMAX-RF muscle pairs. Inter-cycle variability was also reduced for all joints and all muscle pairs following RES training, while inter-participant variability increased for crank torque and co-activation of all muscle pairs. Following VEL training, a possible $11 \pm 20\%$ increase in power was observed at cadences ranging from 160 to 190rpm. Trivial changes were seen for P_{\max} and T_0

in this group though there was a small increase of 3 ± 5 rpm in C_{opt} . The average response to VEL training was associated with reductions in minimum ($-13 \pm 15\%$) and peak ($-5 \pm 14\%$) crank torque, increased co-activation of GMAX-GAS and GAS-TA, as well as reductions in GMAX-RF. All joints and most muscles exhibited an increase in inter-cycle variability following VEL training. Inter-participant variability also increased for crank torque, all joints, all muscles and all muscle pairs. These findings show that 4-weeks of ballistic cycling training improved the limits of the lower limb neuromuscular function in the absence of changes in lower limb volume. The improvements in the limits of neuromuscular function were linked to increased magnitude of force applied to the crank at effective sections of the pedal cycle, increased co-activation of some agonist-antagonist muscle pairs providing joint stability and a reduction in ankle range of motion, simplifying the pedalling movement and/or improving power transfer across the joint. Additionally, it appears that each individual developed a more optimised movement strategy from cycle to cycle, but as a group did not implement a more cohesive strategy after RES training. VEL training at high cadences did improve power, although the responses were highly variable. The use of high resistance training on a stationary cycle ergometer may be useful for improving the level of power produced during movements or tasks performed at slow velocities which may be beneficial for not only healthy un-trained individuals but also in clinical and sporting populations.

The last study of this thesis aimed to investigate the effect of ankle taping on the limits of neuromuscular function on a stationary cycle ergometer and also to assess how ankle taping modified application of torque to the crank, lower limb kinematics, inter-muscular coordination and movement variability. Within the same testing session, the limits of neuromuscular function were assessed from P_{max} , C_{opt} , C_0 , T_0 and power produced at low (40-60 rpm), moderate (100-120 rpm) and high (160-180 rpm) cadences. A total of 13 participants (8 males and 5 females) were tested on a stationary cycle ergometer with their ankle joints bilaterally taped (TAPE) or not (CTRL). First, the results showed that T_0 values calculated in the downstroke were $7 \pm 8\%$ lower in TAPE than CTRL, while P_{max} and C_{opt} were unchanged. T_0 calculated in the upstroke was also lower in TAPE ($-14 \pm 14\%$), while C_{opt} was higher ($+4 \pm 5$ rpm). At 40-60 rpm ankle taping caused likely and possible reductions of power production during the downstroke ($-5 \pm 7\%$) and upstroke ($-10 \pm 18\%$) phases of the pedal cycle. The reduction in power observed in the downstroke at 40-60 rpm was concomitant with a $5 \pm 5\%$ decrease in peak crank torque occurring during the first quarter of the pedal cycle (0-25%). TAPE caused the largest reduction in ankle range of motion at 40-60 rpm ($-15 \pm 6^\circ$), while concomitant reductions in the peak EMG of the ankle muscles (GAS, SOL and TA) and less co-activation of agonist-antagonist (GAS-TA, SOL-TA) and proximal-distal muscle pairs (GMAX-GAS, GMAX-SOL) were seen in the downstroke phase for TAPE. Inter-cycle variability was higher for the ankle joint and most of the lower limb muscles in TAPE at 40-60 rpm. Inter-participant variability was higher for ankle joint, EMG of

most muscles and co-activation of all muscle pairs in TAPE at 40-60 rpm. Trivial differences in power produced at 100-120 rpm and 160-180 rpm were observed between conditions, even though small reductions were observed in minimum ($-11 \pm 15\%$) and peak ($-4 \pm 14\%$) crank torque values at 160-180 rpm. Ankle range of motion was still substantially reduced in TAPE by $8 \pm 6^\circ$ and $5 \pm 7^\circ$ respectively at 100-120 rpm and 160-180 rpm. Differences were more variable for peak EMG and average co-activation values at the higher cadence intervals and the variability between cycles and between participants between conditions were not cohesive. Bi-lateral ankle taping substantially reduced power produced during the downstroke phase of the pedal cycle at low cadences when cycling against high resistances, but had trivial effects at moderate and high cadences. The substantial reduction in ankle range of motion and the decrease in co-activation of the main muscle pairs are likely to have affected the transfer of force/power from the proximal muscles to the cranks. Greater between-participants variability in ankle kinematics and inter-muscular coordination shows that participants adopted different movement strategies in response to ankle taping. These findings indicate that a large range of motion at the ankle joint is essential to produce large levels of power when cycling at low cadences, whereas a limited range of motion at the ankle joint did not affect power production at moderate and high cadences.

Finally, the body of work in this thesis provides: 1) a strong methodological contribution for a more accurate assessment of the limits of lower limb neuromuscular function on a stationary cycle ergometer, 2) evidence for the potential offered by power training interventions to be developed on stationary cycle ergometers to improve the limits of lower limb neuromuscular function and 3) an understanding of the effect of ankle taping on the limits of the lower limb neuromuscular function on a stationary cycle ergometer.

Declaration

Doctor of Philosophy Declaration

I, Briar Louise Rudsits declare that the PhD thesis entitled “Assessing, understanding and improving the limits of neuromuscular function on a stationary cycle ergometer” is no more than 100,000 words in length including quotes and exclusive of tables, figures, appendices, bibliography, references and footnotes. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. Except where otherwise indicated, this thesis is my own work.

Signature _____

Date _____

Dedication

In loving memory of my Grandparents

Dell Bonney (1929-2017), Alven Bonney (1925-2014) and Peter Rudsits (1926-1996)

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List of Publications and Awards

Conference Presentations

- **Rudsits, B. L.**, Taylor, S. B. and Rouffet, D. M. How fast can we really move our legs? Sensorimotor Control Conference, 2015, Brisbane, Australia.
- **Rudsits, B. L.** and Rouffet, D. M. EMG activity of the lower limb muscles during sprint cycling at maximal cadence. European College of Sport Science Conference, 2015, Malmo, Sweden.
- **Rudsits, B. L.**, Taylor, S. B., Stewart, A. M. and Rouffet, D. M. Effect of cadence-specific training on the maximal power-cadence relationships of non-cyclists. Exercise and Sport Science Australia, 2016, Melbourne, Australia.

Awards

- Australian Postgraduate Award - 2013-2015

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List of Abbreviations

°	degrees
°.s ⁻¹	degrees per second
π	pi
2D	two dimensional
3D	three dimensional
APF	ankle plantar-flexors
ATP	adenosine 5'-triphosphate
BDC	bottom dead centre
BF	<i>biceps femoris</i>
CAI	co-activation index
CI	confidence interval
CL	confidence limit
cm	centimetres
C _{max}	measured maximal cadence
CNS	central nervous system
C ₀	estimated maximal cadence
C _{opt}	estimated optimal cadence
CTRL	no ankle tape condition
EMD	electromechanical delay
EMG	electromyography
EXT	extension
F	force
F ₀	maximal force
F-V	force-velocity
FLX	flexion
GAS	<i>gastrocnemius</i>

GMAX	<i>gluteus maximus</i>
HAM	hamstrings
Hz	hertz
KEXT	knee extensors
KFLX	knee flexors
kg	kilogram
L	litre
LBDC	left bottom dead centre
LLV	lean leg volume
L-T	length-tension
LTDC	left-top-dead centre
LGAS	lateral <i>gastrocnemius</i>
max	maximum
MGAS	medial <i>gastrocnemius</i>
min	minimum
mm	millimetre
ms	millisecond
N	newton
N•m	newton metre
N•m.kg ⁻¹	newton metre per kilo of body mass
NMF	neuromuscular function
P-C	power-cadence
P _{max}	estimated maximal power
P-V	power-velocity
RBDC	right bottom dead centre
RER	rate of EMG rise
RES	high-resistance training
RF	<i>rectus femoris</i>

RFD	rate of force development
RM	repetition maximum
RMS	root mean square
ROM	range of motion
rpm	revolutions per minute
RTD	rate of torque development
RTDC	right-top-dead centre
s	seconds
SD	standard deviation
SEE	standard error of the estimate
SOL	<i>soleus</i>
ST	<i>semitendinosus</i>
Stand. Effect	standardised effect
T_0	estimated maximal torque
T_{opt}	estimated optimal torque
TA	<i>tibialis anterior</i>
TAPE	ankle tape condition
T-C	torque-cadence
TDC	top dead centre
TLV	total leg volume
T-V	torque-velocity
V_0	maximal velocity
V_{opt}	optimal velocity
VAS	<i>vastii</i>
VEL	high-cadence training
VM	<i>vastus medialis</i>
VL	<i>vastus lateralis</i>
VR	variance ratio

W	watt
W.kg ⁻¹	watt per kilo of body mass
y	year

Preface

Data collection, analysis and interpretations presented in this thesis are my own. Significant contributions include:

- In Chapter 3, David Rouffet designed the study; Rhiannon Patten assisted with data collection; Robert Stokes and Rhett Stephen provided assistance with technical design and support; Will Hopkins and Andrew Stewart provided assistance with statistical analysis.
- In Chapter 4, David Rouffet and myself designed the study; Simon Taylor provided support with the kinematics component, assisting with data collection and analysis; Rhiannon Patten assisted with data collection and helped supervise training sessions; Robert Stokes and Rhett Stephen provided assistance with technical design and support; Will Hopkins and Andrew Stewart provided assistance with statistical analysis.
- In Chapter 5, David Rouffet and myself designed the study; Simon Taylor provided support with the kinematics component, assisting with data collection and analysis; Robert Stokes provided assistance with technical design and support; Will Hopkins and Andrew Stewart provided assistance with statistical analysis.

Chapter 1 Introduction

Our ability to successfully execute a functional task requires adequate neuromuscular function (NMF) (i.e. the combined work of the central nervous system and skeletal muscle) to permit the movement. Tasks can range from those performed as part of daily life (e.g. rising from a chair and ascending stairs) to those required in the sporting arena (e.g. jumping, running and cycling) and most often require a large contribution from the lower limb muscles (Dorel et al., 2005; Gardner et al., 2007; Reid et al., 2008; Vandewalle et al., 1987). As such the investigation of NMF is important in research, clinical and sport science settings for a wide range of populations (e.g. healthy individuals, athletes, patients, and the elderly). A range of force-velocity (F-V) tests performed on stationary cycle ergometers have been well used in the literature as the method permits a safe, accurate and reproducible assessment of the capacity of the muscles involved in the movement to generate force and power (Arsac et al., 1996; Dorel et al., 2005; Driss & Vandewalle, 2013; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007). Further, due to the design of the stationary cycle ergometer, and the circular trajectory of the pedalling movement, the external resistance and kinematics of the movement can be well controlled making it an ideal exercise to investigate NMF of the lower limbs in different populations. Just as the relationships between force/power vs velocity of single muscle fibers/single muscles have been described previously by muscle physiologists (Hill, 1938; Wilkie, 1950), the data collected from a F-V test on a stationary cycle ergometer can be used to describe the relationships between torque vs cadence and power vs cadence (Arsac et al., 1996; Dorel et al., 2005; Driss et al., 2002; Hautier et al., 1996; Martin et al., 1997; Samozino et al., 2007; Sargeant et al., 1981). Variables commonly calculated from these relationships, such as maximal power, optimal cadence, maximal torque and maximal cadence can then provide an estimate of an individual's limits of NMF.

Unlike the force/power vs velocity relationship at the muscle fiber level, maximal cycling is a complex movement with physiological, biomechanical and motor control factors all affecting the limits of lower limb NMF (Dorel et al., 2010; Gordon et al., 1966; Hill, 1938; Latash, 2012; Muller & Sternad, 2009; Neptune & Kautz, 2001). Physiological or neuromuscular factors affecting these limits include muscle active state of the lower limb muscles and the primary mechanical properties of muscle such as force-velocity, length-tension and force-frequency relationships. Those factors considered to be biomechanical include the magnitude and orientation of the forces transferred to the crank and kinematics of the lower limb joints. Motor control factors include the coordination between muscles and joints and variability of the movement, reflecting how the central nervous system (CNS) manages the abundance of motor solutions offered by the human body to execute the pedalling movement. In isolation the effect of these different factors on power and torque have been observed using simulation studies or *in vitro*. Although, during

multi-joint, dynamic movements such as cycling, these physiological, biomechanical and motor control factors have different effects on the level of force that can be produced and transferred by the working muscles to the crank of the cycle ergometer, depending on the level of resistance or velocity at which the movement is performed. Due to the importance of the force and power producing capacity of the lower limb muscles, it is necessary to implement robust methods for their assessment. However, the approaches used to obtain experimental data and quantify the limits of NMF using a F-V test on a stationary cycle ergometer are equivocal in the literature (Arsac et al., 1996; Dorel et al., 2005; Martin et al., 1997), as such the most accurate method for its evaluation is unknown and warrants investigation.

Maintaining and improving NMF is necessary for sustaining healthy movement across the lifespan. Accordingly, the improvements of the limits of NMF are a major focus in traditional resistance and ballistic training programs (Cormie et al., 2007; McBride et al., 2002). However, ballistic training is commonly recommended when improvements in power are sought, due to their specificity to many sports, allowing better transfer of adaptations to performance (Cady et al., 1989; Cronin et al., 2001; Kraemer & Newton, 2000; Kyröläinen et al., 2005; Newton et al., 1996). Ballistic sprint training on a stationary cycle ergometer may be effective for improving the limits of NMF as it offers the opportunity to maximally activate muscles over a larger part of the movement, facilitating greater adaptations. Sprint cycling interventions on stationary cycle ergometers have been shown to improve power production within two days to four weeks of training, attributed to motor learning and neural adaptations, although the improvements were not cadence specific (Creer et al., 2004; Martin et al., 2000a). Indeed, the use of exercises performed at high resistances and high velocities have been shown to elicit intervention specific improvements in power in other exercises (Coyle et al., 1981; Kaneko et al., 1983; Lesmes et al., 1978). As such, power training interventions implemented on a stationary cycle ergometer may be useful for improving the limits of lower limb NMF at specific sections of the T-C and P-C relationships, although this is unclear and warrants further investigation.

Maximal cycling requires large contributions from muscles spanning the hip and knee joints, but the ankle joint plays an important role in the transfer and orientation of force from these muscles to the pedal (Zajac, 2002). Previously it has been shown that when the motor system is perturbed (e.g. with changing cadence or in the presence of fatigue) motion at the ankle is reduced in response, attributed to a motor control strategy to reduce the degrees of freedom of the movement and thus its complexity (Martin & Brown, 2009; McDaniel et al., 2014). Ankle taping procedures are often employed in ballistic exercises to reduce the range of motion achieved by the joint, providing greater support. However, the effect of ankle taping on the limits of lower limb NMF during sprint cycling has not been previously investigated and would be useful to better understand the role of the ankle during this maximal task. In light of the observations outlined

above, the overall goal of this thesis was to better assess, understand and improve the limits of NMF on a stationary cycle ergometer.

Following a review of literature, this thesis is comprised of three chapters outlining the experimental studies undertaken:

- I. **Chapter 3 (Study one)** – Assessing the limits of neuromuscular function on a stationary cycle ergometer
- II. **Chapter 4 (Study two)** – The effect of high resistance and high velocity training on a stationary cycle ergometer
- III. **Chapter 5 (Study three)** – The effect of ankle taping on the limits of neuromuscular function on a stationary cycle ergometer

The main findings of the three study chapters are then discussed and conclusions made in Chapter 6. Limitations of the studies and suggested directions for future research are also included in the last chapter of this thesis.

Chapter 2 Review of Literature

2.1 Chapter Overview

This review of literature begins with an explanation of the importance of evaluating the limits of NMF or more specifically the ability to produce torque and power in both sport science and clinical settings. Further, this section details the use of stationary cycle ergometers to assess the NMF of the lower limbs. Section two outlines the physiological, biomechanical and motor control factors affecting torque and power production with specific reference to stationary cycle ergometry, while section three delves into methodological considerations for the assessment of the limits of NMF including the type of test protocol and modelling procedures implemented. A fourth section reviews the use of ballistic training interventions to improve NMF and the accompanying neural and morphological adaptations. Lastly, this review documents the role of the ankle joint during ballistic exercises, in particular sprint cycling and the effects of ankle taping on the limits of NMF on a stationary cycle ergometer.

2.2 The importance of understanding, assessing and improving the limits of NMF of the lower limbs

The human neuromuscular system encompasses the nervous system and all the muscles of the body. Assessment of the mechanical capabilities of the lower limb muscles allows the mechanical limits of the neuromuscular system to be characterized and has been previously assessed during ballistic movements in both animals (James et al., 2007) and humans (Cormie et al., 2011; Samozino et al., 2012). These mechanical limits include the maximal amount of force that can be produced, the highest velocity at which the limbs can move, the highest level of maximal power output and the optimal velocity it corresponds to. The assessment of NMF, particularly maximal power and torque generation is of importance for a multitude of purposes including the assessment of individual performance, the efficacy of training and rehabilitation programs and talent identification (Abernethy et al., 1995). The assessment of maximal power and torque is standard practice in athletic populations but is also important for older populations, those suffering from movement disorders which degenerate over time and normally healthy individuals recovering from injury to the lower limbs. Traditionally, an understanding of NMF was provided by values of maximal torque and power produced by a given muscle group during strength testing protocols using isometric and isokinetic exercises (Wilson & Murphy, 1996). However, given that most functional movement tasks are characterized by the rapid, forceful actions of many muscle groups simultaneously (e.g. running, jumping, rising from a chair, ascending stairs), the importance of

ballistic exercises to assess NMF is emerging in the literature (Hoffrén et al., 2007; Millet & Lepers, 2004; Sarre & Lepers, 2005). With this in mind, in both sport science and clinical settings there is a need to assess NMF using exercises (e.g. cycling) that encompass the muscles largely used in functional tasks.

2.2.1 Limits of lower limb NMF in sport science

The ability to produce a high level of power is considered to be fundamental in a successful sporting performance (Martin et al., 2007; Morin et al., 2002; Vandewalle et al., 1987), with many studies showing that high force and power outputs are well correlated with athletic performance (Baker, 2001; Kraemer & Newton, 2000; Sleivert & Taingahue, 2004). With regards to sprint cycling, a high maximal power output and the ability to maintain a high level of power output over a wide range of cadences is favorable to a successful sporting performance, especially as the velocity of the movement is continually changing over the duration of an event (e.g. a flying 200-m sprint) (Gardner et al., 2007; Martin et al., 2007; Morin et al., 2002; Vandewalle et al., 1987). Indeed, Dorel and colleagues (2005), found that when corrected for frontal area, maximal power was found to be a significant predictor of 200-m sprint performance in their cohort of world class athletes. Similarly, in other ballistic exercises maximal power has been positively correlated with jump height (Vandewalle et al., 1987) and sprint running speed (Morin et al., 2002). Further, during sprint cycling events that require a stationary start (e.g. 1000-m time trial, 500-m time trial, team sprint) a high torque generating capability is required at the start of the event to get the bike into motion as fast as possible, to allow the cyclist to reach velocities that maximise their power output.

The assessment of lower limb NMF can be used to define the level and training status of an athlete, via the reporting of maximal torque (i.e. strength) and velocity (i.e. speed) generating capabilities of an individual's neuromuscular system. Previously, Samozino and colleagues (2012) reported that both maximal power output and force-velocity profiles provided information regarding the NMF of the lower limbs. In particular, they suggested that an optimal force-velocity profile exists for each individual, for which performance is maximized. Quantifying these limits of NMF can also be used for the programming of athletic training, assessment of training program efficacy (Cormie et al., 2011; Cronin & Sleivert, 2005) and has implication for the identification and development of talent (Tofari et al., 2016).

2.2.2 Limits of lower limb NMF in clinical exercise science

An adequate level of NMF is required by all humans to perform activities of daily living. Muscle power has been strongly linked to the performance of activities of daily living (e.g. sit to stand, climbing stairs), with a reduction in muscle power leading to an inability to perform these activities (Bassey et al., 1992; Clark et al., 2006; Ferretti et al., 1994; Foldvari et al., 2000; Martin et al., 2000c). The maintenance of NMF over the life span improves the ability of an individual to move without assistance which is necessary for maintaining independent functioning and is of great importance to lessen the burden on public health systems. With these findings in mind it appears essential to have testing procedures that can be implemented with older and frail individuals, those recovering from injury and for those with motor impairment disorders (e.g. stroke, cerebral palsy) to monitor their limits of NMF.

Often, lower limb functionality is assessed using single-joint exercises (e.g. knee extension and flexion), evaluating the force and power producing capabilities of a small number of muscles during isometric contractions (Bassey et al., 1992; Clark et al., 2010). However, the results from isometric exercise tests have been previously shown to correlate poorly with dynamic performances (Baker et al., 1994). Although single-joint and isometric exercises are often deemed to be 'safer' for clinical populations to perform, they do not appear to provide an ecological evaluation of the power and torque producing capabilities of the lower limb muscles, therefore do not represent the requirements of the tasks and activities performed on a daily basis.

2.2.3 Assessing the limits of lower limb NMF on a stationary cycle ergometer

As maximal cycling is a ballistic, dynamic, multi-joint movement requiring the production of power from the lower limb muscles (the largest muscle mass of the body) it is well suited to provide an overall assessment of NMF. Like other ballistic running and jumping exercises, most of the external force and power is produced by the lower limb muscles during cycling (Nagano et al., 2005; van Ingen Schenau, 1989; Zajac, 2002). Further, as cycling involves repetitive alternating flexion and extension of the lower limb joints and alternating contraction of agonist and antagonist muscles similar to exercises such as running, it is ideal to evaluate the limits of lower limb NMF in a range of different populations and sports.

Indeed, all-out cycling has been used largely in previous literature to evaluate the power and force producing capabilities of the lower limb muscles (Arsac et al., 1996; Dorel et al., 2005; Driss & Vandewalle, 2013; Hintzy et al., 1999; Sargeant et al., 1981). Although cycling is a complex movement requiring the successful coordination of three joints and more than 20 muscles by the CNS, it is a simple exercise task to implement, requiring little more than a commercial stationary cycle ergometer. Due to the accessibility of stationary cycle ergometers in most

exercise testing laboratories, community gyms and clubs, the ease and affordability of performing a maximal cycling test on an ergometer is high. Furthermore, due to its closed kinetic chain nature and ability for individuals to be seated during the movement it is a relatively safe exercise, with the ergometer modifiable (e.g. upright or dropped hand positioning, flat or clipless pedals, addition of a back rest to improve stability) to suit the population tested (e.g. athletes, elderly, the injured and those with movement disorders) (Janssen & Pringle, 2008). Indeed, several studies have been conducted whereby the stationary cycle ergometer was modified to suit the requirements of the research aim (Lopes et al., 2014; Reiser Ii et al., 2002; Sidhu et al., 2012). Also, unlike other ballistic movements such as jumping and sprint running the risks of falling and injury are very low in stationary cycle ergometry, even for those who are not accustomed to the movement.

2.3 Factors affecting the limits of lower limb NMF on a stationary cycle ergometer

It is often seen that the disciplines of biomechanics, physiology and motor control are somewhat compartmentalised with regards to the investigation of NMF. However, the limits of NMF (i.e. maximal power, optimal cadence, maximal torque and maximal cadence) are affected by a combination of these inter-related factors during stationary cycle ergometry. The physiological or perhaps more appropriately termed neuromuscular factors affecting NMF include the mechanical properties of muscle such as the force-velocity, length-tension and force-frequency relationships and muscle fiber type distribution, while neural factors include the active state of the muscles. Biomechanical factors include the magnitude and orientation of the forces transferred to the crank and kinematics of the lower limb joints; while motor control factors include the coordination between muscles and joints and variability of the movement, reflecting how the CNS manages the abundance of motor solutions offered by the human body to execute the pedalling movement. Few studies have tried to synthesise the collective knowledge and research methods designed to investigate these factors, particularly when cycling on a stationary ergometer. Although, a recent article by Latash (2016) explained how the fields of motor control and biomechanics are inseparable when describing motor function. Therefore, understanding the relative contribution and integration of these different, but integrated factors is important when assessing and challenging the limits of NMF. As such, the physiological, biomechanical and motor control factors affecting the limits of NMF on a stationary cycle ergometer are discussed in further detail in the sections below.

2.3.1 Physiological (neuromuscular) factors

2.3.1.1 Activation of the lower limb muscles

Human skeletal muscles function to produce force and motion by acting on the skeletal system causing bones to move about their joint axis of rotation and are primarily responsible for changing posture and locomotion. In order for movement to occur, muscles must produce a contraction that changes the length and shape of the muscle fibers. The activation of motor units is the first event in the sequence of the production of muscle force. The action of a muscle results from the individual or combined actions of motor units which consist of alpha motor neurons and the muscle fibers it innervates. A single muscle is innervated by a motor neuron pool consisting of a collection of alpha motor neurons. These motor neurons are comprised of a cell body, axon and dendrites, enabling transmission of nerve impulses or action potentials from the CNS to the muscle. Along the myelin sheath encased axon, nodes of Ranvier form uninsulated gaps between the myelin sheaths allowing nerve impulses to move toward the terminal branches at the neuromuscular junction. The neuromuscular junction serves as the crossing point between the end of the myelinated motor neuron and a muscle fiber and functions to transmit the nerve impulse to initiate a muscle action. Arrival of an impulse at the neuromuscular junction triggers a release of neurotransmitter acetylcholine, changing the electrical nerve impulse into a chemical stimulus. Within the postsynaptic membrane acetylcholine combines with a transmitter-receptor eliciting a wave of depolarization (action potential) that spreads along the sarcolemma, into the transverse-tubule system for initiation of muscle contraction. Excitation-contraction coupling serves as the mechanism whereby the electrical activity of the action potential initiates chemical events at the cell surface causing muscle contraction, with intracellular calcium ions responsible for regulating cross-bridge cycling and therefore muscle contraction (Klug & Tibbits, 1988).

The active state or level of muscle activation and therefore the amount of force a muscle can exert at a given length and velocity is dependent on the number of motor units recruited by the CNS and the frequency at which action potentials are discharged (Adrian & Bronk, 1929). Motor units are recruited systematically according to size (i.e. Henneman's size principle), with smaller motor units recruited first, followed by larger motor units, and consequently slow-twitch muscle fibers (type I) recruited before fast-twitch muscle fibers (type II) (Henneman, 1957). The order of which motor units are recruited appears to be the same for isometric and dynamic muscle contractions (Duchateau et al., 2006) and also during more rapid (ballistic) contractions (Desmedt & Godaux, 1978).

Using surface electromyography (EMG) the active state of a muscle (and the control operated by the CNS) can be non-invasively investigated. Surface EMG is used to detect the electrical potential generated by muscle cells between pairs of electrodes placed on the skin surface

allowing the extracellular recording of action potentials propagating along the muscle fibers (Merletti et al., 2001). Surface EMG has been used extensively to assess the neuromuscular control of the lower limb muscles during submaximal (Chapman et al., 2009; Chapman et al., 2008a; Chapman et al., 2008b; Chapman et al., 2006; Dorel et al., 2008; Hug, 2011; Hug et al., 2008; Hug et al., 2010) and maximal cycling (Dorel et al., 2012; O'Bryan et al., 2014). The main lower limb muscles involved in the pedalling movement include muscles surrounding the hip, knee and ankle joints. As such, the muscles most commonly assessed using EMG include: *gluteus maximus* (GMAX) that functions as a hip extensor; *vastus medialis* (VM) and *vastus lateralis* (VL) (when combined are referred to as the *vastii* (VAS)) that function as knee extensors; *rectus femoris* (RF) that functions as a hip flexor and knee extensor; *semimembranosus* (SM) and *biceps femoris* (BF) (when combined are referred to as the hamstrings (HAM)) that function as a hip extensor and knee flexor; *gastrocnemius lateralis* and *gastrocnemius medialis* (when combined are referred to as *gastrocnemii* (GAS)) that function as a knee flexor and ankle plantar-flexor; *soleus* (SOL) that functions as an ankle plantar-flexor and *tibialis anterior* (TA) that functions as an ankle dorsi-flexor (Dorel et al., 2012; Hug et al., 2008; Hug et al., 2010; Jorge & Hull, 1986; Rouffet & Hautier, 2008; Rouffet et al., 2009; Ryan & Gregor, 1992) (Figure 2.1). Although these muscles listed are typically assessed, other deeper muscles contributing to the pedalling movement (i.e. *psoas*, *vastus intermedius*, *tibialis posterior*, *iliacus*) cannot be discounted, but are practically difficult to measure. Consequently, literature regarding the activity patterns of these deep muscles during pedalling is limited (Chapman et al., 2006, 2010).

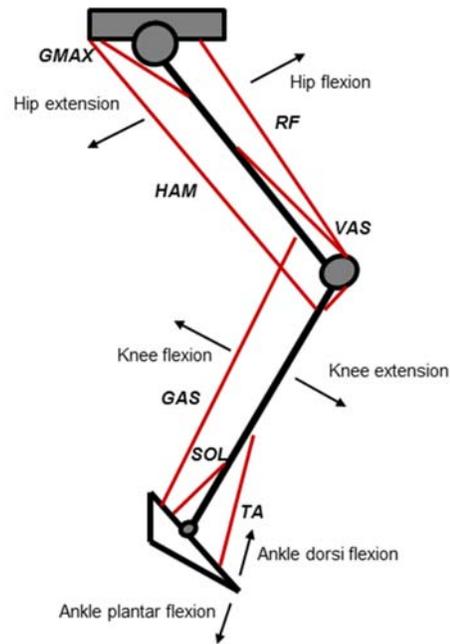


Figure 2.1. Schematic illustrating the phases of hip, knee and ankle joint movement and the location of the main muscles involved in the pedalling movement. GMAX (*gluteus maximus*), RF (*rectus femoris*), VAS (*vastus lateralis* and *vastus medialis*), HAM (*semimembranosus* and *biceps femoris*), GAS (*gastrocnemius*), SOL (*soleus*), TA (*tibialis anterior*).

Although surface EMG appears to be the most preferred method for assessing muscle active state, physiological (e.g. fiber membrane properties: conduction velocity and synchronisation of motor units, and motor unit properties) and non-physiological (e.g. cross-talk from adjacent muscles, impedance, subcutaneous fat thickness, size and distribution of motor unit territories and electrode placement) factors are known to affect the EMG signal (Farina et al., 2004). Where possible, these factors should be minimised. Accordingly, in an attempt to reduce the effect of electrode placement and standardise the methodology of this technique, recommendations have been produced by the Biomedical and Health and Research Program of the European Union (SENIAM project) (Hermens et al., 2000) and identified in previous research (Rainoldi et al., 2004).

As per the theory of Nyquist (1928), to accommodate the frequency content, EMG signals should be sampled at a rate twice that of the highest expected maximum frequency of the signal to ensure a true representation of the signal recorded. The frequency content of raw EMG signals ranges between approximately 6 and 500 Hz, with the majority of this frequency between 20 and 150 Hz. After collection of the EMG signal and prior to using it to assess muscle activation and timing, the signal is usually rectified (i.e. the negative component of the signal is made positive) and filtered to remove non-physiological noise or artefact. Briefly, following rectification the

signal is typically smoothed using filters (i.e. low-pass, high-pass, band-pass) in accordance with the characteristics of the movement (e.g. the frequency at which its performed) and purpose of EMG analysis in mind. To estimate the level of neural drive to the individual muscles the amplitude of an EMG signal can be assessed. A typical approach taken during voluntary movements to quantify EMG amplitude is the root mean square (RMS) value of the EMG, which reflects the mean power of the signal (Dorel et al., 2008; Laplaud et al., 2006). The timing and duration of muscle activation is also commonly assessed by defining the time of signal burst onset and offset that is often based upon a minimum threshold of three standard deviations of the baseline EMG signal (Neptune et al., 1997; Rouffet et al., 2009). Lastly, the reproducibility of EMG activity levels has been shown to be high during the pedalling movement (Dorel et al., 2008; Houtz & Fischer, 1959; Laplaud et al., 2006).

Due to the aforementioned physiological and non-physiological factors affecting the raw EMG signal, it is difficult to interpret the level of the processed signal without expressing it in relation to a reference value. The EMG signal must be 'normalised' to a meaningful and repeatable value, typically a mean or peak EMG to allow comparisons to be made between EMG results obtained from different muscles/subjects or within the same subject on different days. There are several methods which can be used for normalisation including referencing the signal to a peak or mean activation level during isometric and dynamic contractions (Burden, 2010; Burden & Bartlett, 1999; Hug & Dorel, 2009; Rouffet & Hautier, 2008). However, to date there appears to be no consensus as to the most appropriate approach. Using the peak EMG signal from a maximal cycling exercise bout (or more specifically from a F-V test) has been shown to be a valid and reliable way to study muscle activation of the lower limb muscles during cycling (Rouffet & Hautier, 2008). Using this approach the EMG signals of the different muscles recorded during a cycling bout can be expressed as a percentage of the peak muscle activity that occurred during the maximal intensity or reference exercise bout for a given muscle and for a given individual. This normalisation approach has been shown to decrease inter-individual variability in comparison to using a reference value from a maximal voluntary isometric contraction or using the raw EMG data (Chapman et al., 2010; Yang & Winter, 1984). Further, appropriate normalisation lessens the impact of non-physiological factors (e.g. cross-talk, impedance, subcutaneous fat thickness, electrode placement) that can influence the EMG signal (Rouffet & Hautier, 2008).

During cycling muscle activation changes throughout the pedal cycle, accordingly it is necessary to define the beginning (i.e. 0° or 0%) and end (i.e. 360° or 100%) of a pedal revolution to allow activation patterns to be referenced within the cycle. Typical patterns of muscle activation during the pedalling movement have been well described in the literature, but most pertain to submaximal cycling (Jorge & Hull, 1986; Li & Caldwell, 1998; Rouffet et al., 2009; Ryan &

Gregor, 1992). More recently, patterns of lower limb muscle activation during maximal intensity cycling have been illustrated for cadences corresponding to 80% of the participant's optimal cadence (Dorel et al., 2012). Specifically, as illustrated in

Figure 2.2 below, GMAX was shown to be active during the power producing downstroke portion of the cycle from 360° (just before top-dead-centre (TDC)) to 120°, while VAS (VL and VM) was also active before TDC at 305° until 100°. RF activity occurred earlier in the cycle (260°) than both GMAX and VAS because of its dual function as a bi-articular muscle and was active to 90°. Medial and lateral GAS appeared to exhibit similar activity patterns, active from TDC, to 220° (beyond bottom-dead-centre (BDC)), while SOL was not active for as long (350° to 140°). Those muscles primarily active during the upstroke (i.e. 180° to 0°) include the HAM group (SM, ST and BF) and TA. HAM was active from 260° to TDC, while TA became active just before BDC up until TDC. It is also important to note that the method for reporting activation patterns can vary between studies, typically for those muscles for which a secondary burst of activation within a pedal cycle can occur (e.g. the bi-articular muscles and TA) (Dorel et al., 2012).

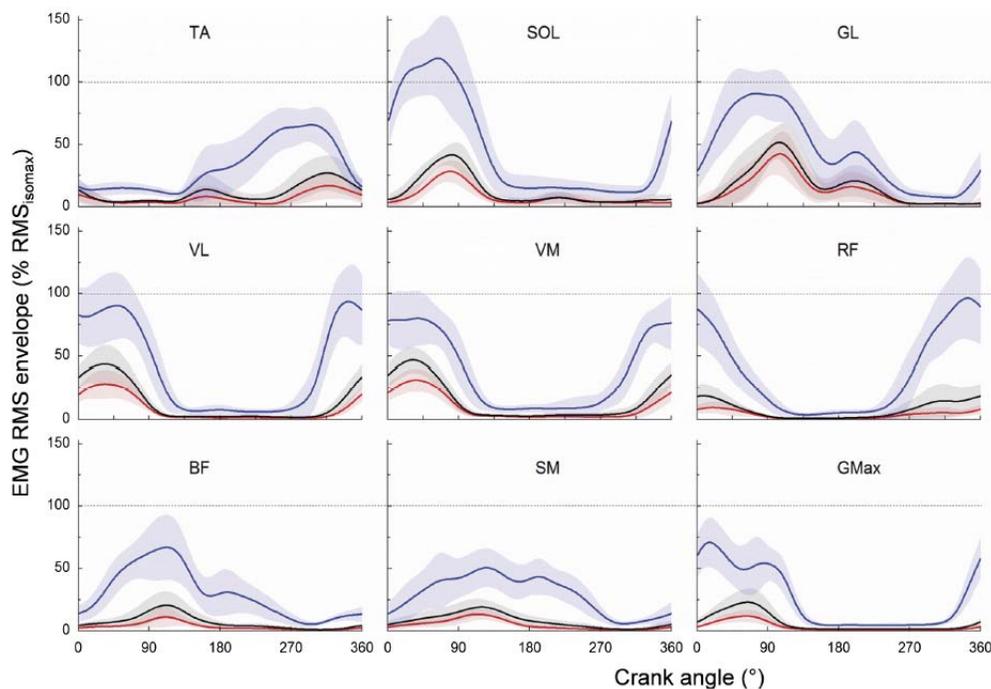


Figure 2.2. EMG profiles of six lower limb muscles during all-out cycling. Blue lines denote all-out sprint (blue line), red and black lines denote two submaximal conditions. TA (*tibialis anterior*), SOL (*soleus*), GL (*lateral gastrocnemius*), VL (*vastus lateralis*), VM (*vastus medialis*), RF (*rectus femoris*), BF (*biceps femoris*), SM (*semimembranosus*), GMax (*gluteus maximus*). Taken from Dorel et al. (2012).

Late in the 19th century, the notion that skeletal muscles have different functional roles which are largely dictated by the number (i.e. mono-articular or bi-articular) and type (i.e. ball-and-socket or hinge) of joints the muscle crosses was put forward by Cleland (1867). Since then, it is well accepted that during ballistic exercises such as jumping, sprint running and cycling, mono-articular muscles, those crossing only one joint are suggested to act as primary force producers while bi-articular muscles, those crossing two joints work to transfer the force from the mono-articular muscles and help to control external forces (i.e. the application of force to the crank/pedal in cycling) (Kautz & Neptune, 2002; van Ingen Schenau, 1989; Van Ingen Schenau et al., 1995). Although, it has also been argued that due to the redundant nature of the musculoskeletal system the task being executed will dictate the role a muscle plays regardless of the number of joints it spans (Kuo, 1994). A simulation of maximum speed pedalling has shown that the mono-articular hip (GMAX) and knee extensor (VAS) muscles provide the greatest amount of mechanical energy within a pedal cycle at ~20% and ~35% respectively, while energy produced by the muscles surrounding the ankle (GAS, SOL, TA) and other bi-articular muscles (RF, HAM) are considerably less (Raasch et al., 1997) (Figure 2.3). In agreement, during submaximal cycling Neptune et al. (1997) found that GMAX and VAS produced 80% of their activity during the extension region, while Ericson (1988) reported that muscle force produced during hip and knee extension provided ~70% of total positive work.

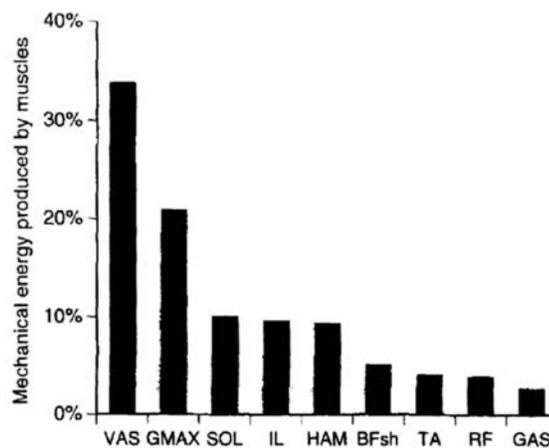


Figure 2.3. Mechanical energy produced by the leg muscles during simulated maximal cycling. VAS (*vastii*), GMAX (*gluteus maximus*), SOL (*soleus*), IL (*ilipsoas*), HAM (*semimembranosus*), BFsh (*biceps femoris* short head), TA (*tibialis anterior*), RF (*rectus femoris*), GAS (*gastrocnemii*). Taken from Raasch et al. (1997).

It appears that maximal muscle activation (i.e. recruitment of all motor units, firing at maximal rates) during a voluntary effort is possible in humans; therefore active state shouldn't be a limiting factor for the maximal force generating capacity of a given muscle. However, during dynamic movements such as cycling which require the coordination of many muscles, maximal activation would be required by every muscle involved, for every pedal cycle to get a true level of maximal force. Additionally, activation levels are highly variable within and between muscles and individuals, with many repetitions of the movement task often required before a true maximal effort can be generated (Allen et al., 1995). There are a variety of other factors influencing the active state of the muscles involved in the pedalling movement (and subsequently the level of power they can produce) that include movement frequency and subsequent effect on activation-deactivation dynamics; rate of EMG rise; neural inhibitions and post-activation potentiation that are outlined below.

Cadence affects the amount of power (and force) that an individual can produce with increasing cadence imposing two constraints on the neuromuscular system: 1) an increase in joint angular velocity; and 2) decreased time for muscle activation and deactivation (Martin, 2007). Due to the fixed trajectory of the pedal, at a given cadence each muscle will only be active once every pedal cycle, therefore the effect of cadence (or more specifically cycle frequency) on the activity of individual muscles producing the pedalling movement can be easily examined using surface EMG. The effect of cadence on EMG activity level appears to be equivocal, but there is some general agreement that during submaximal cycling, linear increase in GAS, HAM and VAS activity occurred with increasing cadence, while GMAX and SOL exhibited inverted quadratic relationships with the lowest level of EMG occurring at 90 rpm (Ericson, 1986; Neptune et al., 1997). In contrast, reduced VAS and GMAX activity with increasing cadence has been observed by Lucia et al. (2004) in well-trained cyclists. However, less is known regarding the effect of cadence on EMG during maximal effort cycling. Hautier et al. (2000) did not see variations in EMG activity during a 5-s sprint for which cadence reached 150 rpm. Further, Samozino and colleagues (2007) found that average EMG activity did not differ between 70 and 160 rpm for the main muscles involved in the pedalling movement - GMAX, RF, BF, VL.

In order to maximise the force output of a muscle, the activation level of that muscle is required to be as high as possible during the phase for which the muscle shortens and as minimal as possible in its phase of lengthening (van Soest & Casius, 2000). The alteration in muscle active state with increasing cadence is partly due to the time requirements for muscle activation and relaxation. As eloquently described by Neptune and Kautz (2001) activation-deactivation dynamics 'are the processes that describe the delay between muscle force development (i.e. the delay between neural excitation arriving at the muscle and the muscle developing force) and relaxation (i.e. the delay between the neural excitation ceasing and the muscle force falling to

zero)'. During fast cyclical contractions such as pedalling, the effect of activation-deactivation dynamics becomes more influential on the amount of positive and negative work produced by a muscle. The short cycle duration accompanying high cadences starts to become problematic due to the physiological time requirements for the rise and decline of muscle active state and the delay between neural excitation and muscle force response (i.e. electromechanical delay; EMD) (Neptune & Kautz, 2001; van Soest & Casius, 2000). Factors attributed to causing the latency have been suggested to include: the time course of action potential propagation along the sarcolemma into the transverse tubules (i.e. axonal conduction velocity), the processes of excitation-contraction coupling and the time required to stretch the series elastic component of muscle (i.e. force transmission) (Muraoka et al., 2004; Norman & Komi, 1979). However, the contribution of each of these factors to overall EMD is undetermined. EMD has been documented between 30 and 100 ms in duration from onset of muscle active state to peak muscle force (Cavanagh & Komi, 1979; Corser, 1974; Inman et al., 1952; Winters & Stark, 1988) but approximately 90 ms in most of the leg muscles during cycling (Van Ingen Schenau et al., 1995; Vos et al., 1991). It has been suggested that EMD remains relatively constant regardless of movement complexity (Cavanagh & Komi, 1979), cadence (Li & Baum, 2004) and duration for which the movement is performed (Van Ingen Schenau et al., 1992). The functional role of the muscles involved does not appear to affect EMD, with no substantial differences in time reported between mono-articular (93 ± 30 ms) and bi-articular (95 ± 35 ms) muscles (Van Ingen Schenau et al., 1995). As such a blanket EMD of 100 ms has been used in cycling studies when shifting the EMG signal by a given time period or a given portion of the pedal cycle to enable associations to be made between muscle activation and crank torque patterns (Samozino et al., 2007). Using EMG analyses several authors have reported that peak muscle activation occurs earlier in the pedal cycle with increasing cadence, and have suggested that it is a strategy by the CNS to compensate for EMD, in an attempt to maintain a high level of pedal force occurring at the most effective section of the pedal cycle (Neptune et al., 1997; Samozino et al., 2007; Sarre & Lepers, 2007).

As illustrated in Figure 2.4 the time to complete a pedal cycle reduces as cadence increases and hence the time window available for muscles to activate and deactivate within a pedal cycle becomes narrower. In particular, deactivation corresponds to a greater portion of the pedal cycle as the process of muscle relaxation is slower than that of activation (Caiozzo & Baldwin, 1997; Neptune & Kautz, 2001). The time available is further reduced when taking into consideration that muscles must activate and deactivate within their respective phases of flexion and extension phases which takes place within half a pedal cycle (Figure 2.4). At relatively slow cadences, when cycle duration is adequate to accommodate the time requirements of muscle activation and relaxation the same challenges like those experienced at high cadences are not

imposed on the neuromuscular system (Askew & Marsh, 1998). For example at a cadence of 60 rpm each pedal revolution takes ~ 1 -s to complete, with the flexion and extension phases occurring within half that time (~ 0.5 -s) adequate time is available for muscles to reach and maintain a high active state and fully relax within a pedal cycle. As such the effect of activation-deactivation dynamics is minimal at this cadence, with force applied to precise sections of the pedal cycle, which enables power output to be maximised. Alternatively, at higher cadences, such as 180 rpm a pedal revolution takes ~ 333 ms to complete, with flexion and extension each having to take place within 167 ms. As the physiological time delays for activation and deactivation remain fairly constant, the time required for these processes represent a greater portion of the pedal cycle at higher cadences. Consequently the active state of a muscle is not maximal over the full period for which it shortens and is not zero during the phase at which it lengthens, reducing positive pedal force during the downstroke phase and increasing negative pedal force during the upstroke. Although it should not be forgotten that it is both the combination of muscle active state and increasing shortening velocity contributing to the reduction in pedal force and therefore power with increasing cadence (Martin, 2007; Samozino et al., 2007; van Soest & Casius, 2000).

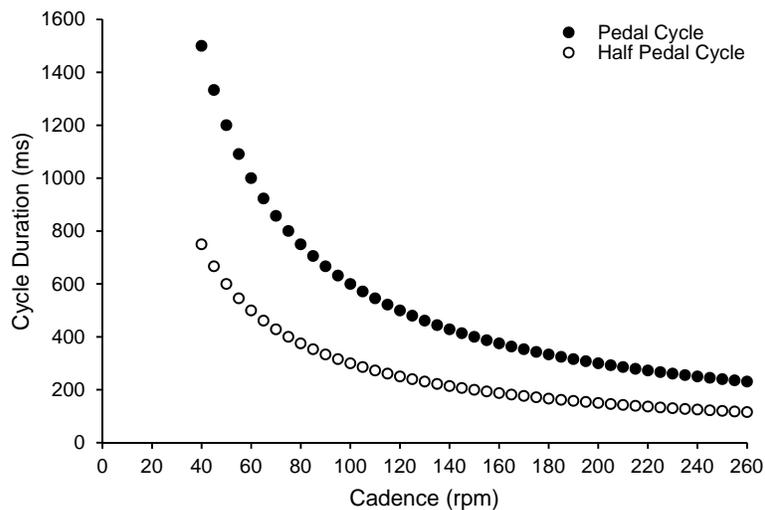


Figure 2.4. The relationship between pedal cycle duration and cadence.

The speed at which the CNS can maximally activate skeletal muscles at the beginning of a contraction or rate of EMG rise (RER) can also influence the active state of a muscle and corresponding level of power that can be produced. RER is closely linked to the rate of torque development (RTD), the ability to rapidly develop muscular force within the early phase of contraction (Andersen & Aagaard, 2006; Morel et al., 2015). As expected, a high level of

contractile RTD is necessary for a good performance in sports requiring high levels of power output, but also for the execution of daily activities and the prevention of injury in the elderly and diseased populations. As outlined above, during ballistic movements such as maximal cycling the time available for muscles to contract can be less than 167 ms (at very fast cadences), though the time required to reach maximal muscular force has been previously shown to be greater than 300 ms in human skeletal muscle (e.g. knee extensors) (Thorstensson et al., 1976b). Consequently, during fast limb movements, the accompanying short period of time available for contraction (e.g. 0-200 ms) may not allow maximal muscle force to be reached and reduce the level of external torque and power produced particularly at high cadences during maximal cycling exercise. RTD has been suggested to be influenced by muscle cross-sectional area, muscle fiber type (i.e. myosin heavy chain composition) and the neural drive to the muscles (i.e. the magnitude of neural drive and rate of motorneuron firing frequency) (Morel et al., 2015).

Acting at the opposite end of the F-V relationship to activation-deactivation dynamics, when the velocity of the movement performed is slow, the level of activation that can be achieved by a muscle or group of muscles can also be affected. Previously, it has been shown that during slow knee extension exercises (i.e. when muscle shortening velocity is slow) muscle activation and subsequently torque output were reduced (Babault et al., 2002; Westing et al., 1991). Babault et al. (2002) and Westing et al. (1991) showed that knee extensor muscle activation was reduced concomitantly with slowing muscle shortening velocities ($360^{\circ} \cdot s^{-1}$ to $45^{\circ} \cdot s^{-1}$) during concentric maximal knee extension exercise; although the corresponding absolute value of torque was not documented. Further, Caizzo and colleagues (1981) noted that the high force/slow velocity region ($\sim 95^{\circ} \cdot s^{-1}$) of the F-V relationship exhibited a levelling off in force output in subjects performing knee extension exercise. It was suggested that the decrease in neural drive reported may be an attempt to limit the generation of high levels of tension in the *vastii* muscles, a mechanism to protect the musculoskeletal system from injury. More specifically, the Golgi tendon organs sense the high tension levels in the working muscles, increasing inhibitory feedback accordingly to reduce alpha motoneuron excitability and subsequently force output (Solomonow et al., 1988). Although documented in single-joint movements, the occurrence of reduced neural drive in multi-joint movements such as maximal cycling at slow velocities (cadences) is currently unknown.

Another physiological factor which can affect NMF that has particular relevance to stationary cycle ergometry is muscle potentiation. Muscle potentiation is a phenomenon by which force exerted by a muscle is increased due to previous contractions (i.e. the contractile history of the muscle) influences the mechanical performance of subsequent muscle contractions via an enhanced neuromuscular state (Robbins, 2005; Sale, 2002). In particular, muscle potentiation increases the amount of force produced during concentric (in comparison to isometric) contractions like those experienced in cycling (Sale, 2002). Mechanisms proposed for muscle

potentiation include an increase in synaptic excitation within the spinal cord, leading to greater post-synaptic potentials and more force produced by the muscles involved (Rassier & Herzog, 2002) and an increased sensitivity of actin-myosin to calcium released from the sarcoplasmic reticulum following subsequent muscle contractions (Grange et al., 1993). It appears that muscle fiber type is the greatest muscle characteristic affecting muscle potentiation magnitude with muscles comprised of a greater proportion of type II fibers exhibiting the greatest potential for muscle potentiation (Hamada et al., 2000). Activities that require short bursts of maximal intensity exercise (such as sprints), adequate recovery between bouts is required to enable phosphocreatine stores to be replenished (McComas, 1996). Although, if recovery is too long the performance enhancing effects of muscle potentiation may be limited due to the lack of preceding muscular contractions before the start of the maximal effort, consequently affecting the level of power produced in the subsequent contractions (i.e. for recurring pedal cycles).

2.3.1.2 Muscle force vs velocity and length vs tension relationships

Early research showed that the force generated by a single muscle fiber was a function of the velocity at which it shortens. During concentric contractions the force vs velocity (F-V) relationship of in-vitro (Fenn & Marsh, 1935; Hill, 1938) and in-vivo (Perrine & Edgerton, 1978; Thorstensson et al., 1976a; Wilkie, 1950) muscle has been shown to be hyperbolic (Figure 2.5). Accordingly, the greatest amount of muscle force is produced at slow contraction velocities (i.e. maximal force; F_0) due to more time available for the generation of tension via increased cross-bridge attachment. However, as the speed of muscle shortening increases, myosin and actin filaments slide past each other at a faster rate, missing potential binding sites, resulting in fewer cross-bridge attachments and ultimately a reduction in force produced by the muscle (i.e. the sliding filament theory) (Huxley, 1957). As power is a function of force and shortening velocity, researchers have used the classic hyperbolic F-V relationship to calculate the power a muscle can produce at a given shortening velocity (Figure 2.5). As such, each muscle produces its maximal power (i.e. P_{max}) at an optimal shortening velocity (i.e. V_{opt}), occurring at the apex of the power vs velocity (P-V) relationship, estimated to occur at approximately one-third of its maximum shortening velocity (i.e. V_0). The limits of mechanical function (i.e. F_0 , V_0 , P_{max} and V_{opt}) of a single muscle fiber depends primarily on the details of its myosin heavy chain isoform composition or more simply put, muscle fiber type (Bottinelli et al., 1991). Muscle fibers are typically categorised into three types: slow-twitch (type I), fast-twitch oxidative (type IIa) or fast-twitch glycolytic (type IIb). The distinct characteristics of each of these fiber types cause them to exhibit different force-velocity relationships (Bottinelli et al., 1991; Greaser et al., 1988). Type I fibers are characterised by slower shortening speeds, related to slower calcium release and reuptake from the sarcoplasmic reticulum and low myosin ATPase activity than that of fast-twitch

fibers. These distinguishing features make these fibers highly resistant to fatigue. Unlike type I fibers, type II fibers can generate energy rapidly, contributing to fast, powerful actions due to speeds of shortening and tension development up to five times higher than type I fibers (Fitts et al., 1989). The characteristics of these muscle fibers include a high capacity for the electromechanical transmission of action potentials, rapid and efficient calcium release and reuptake by the sarcoplasmic reticulum and a high rate of cross-bridge turnover. Type IIb fibers exhibit the fastest shortening speeds of all the fibers, producing very high levels of force, power and speed. Type IIa fibers fall in between type I and type IIb fibers. While still exhibiting a fast shortening speed the capacity for energy transfer is well-developed from both aerobic and anaerobic systems for type IIa fibers making them unable to produce the same level of force as type IIb fibers but more resistant to fatigue. It has been shown that irrespective of conditioning level type IIa fibers can contract 10 times faster than type I fibers and twice as fast as type IIb fibers (Bottinelli et al., 1999; Larsson & Moss, 1993). Further, Sargeant (1994) displayed that the optimal shortening velocity and corresponding maximal power was different between type I and type IIa and IIb fibres.

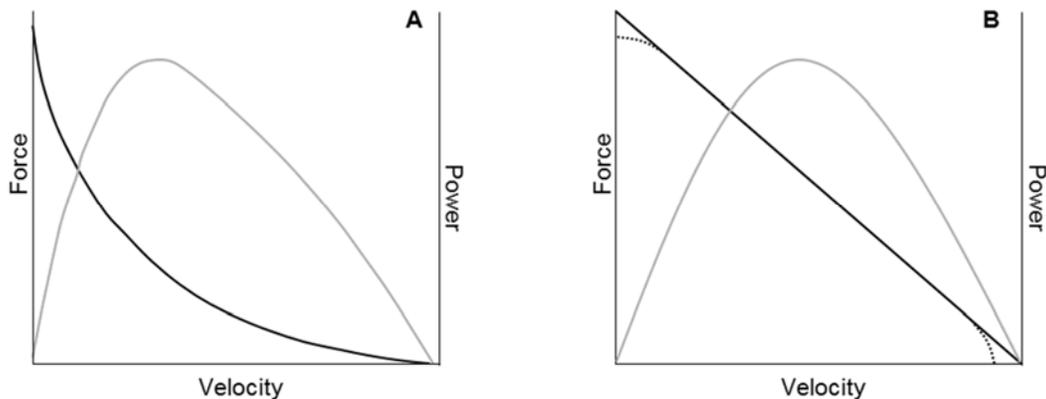


Figure 2.5. Force-velocity and power-velocity relationships for a single muscle/joint and for multi-joint movements. **A:** illustrates the force-velocity (black line) and power-velocity (grey line) relationships observed for single muscle and joints, **B:** illustrates these relationships observed for multi-joint movement. Dotted line denoting the ‘quasi’ linear relationship suggested by Bobbert (2012). Adapted from Hill (1938) and Wilkie (1950).

In concert with velocity, muscle fiber length (i.e. the length-tension relationship) also influences the amount of force produced by a muscle and thus the amount of power generated at the joint that it surrounds (Gordon et al., 1966). According to the sliding filament theory, the development of force depends on the attachment-detachment of cross-bridges. As the production

of force only occurs during the attachment phase, the myosin and actin filaments must be close enough to elicit it. As sarcomere length changes, the number of actin binding sites available for cross-bridge cycling changes, with the amount of overlap between the different filaments influencing the amount of the tension that can be generated by the sarcomere. Consequently, a muscle will produce its greatest force when operating close to its ideal length. As illustrated by Figure 2.6, adapted from Gordon and colleagues (1966), when a muscle fiber is shortened or lengthened beyond its ideal length the amount of force the muscle fiber can generate decreases.

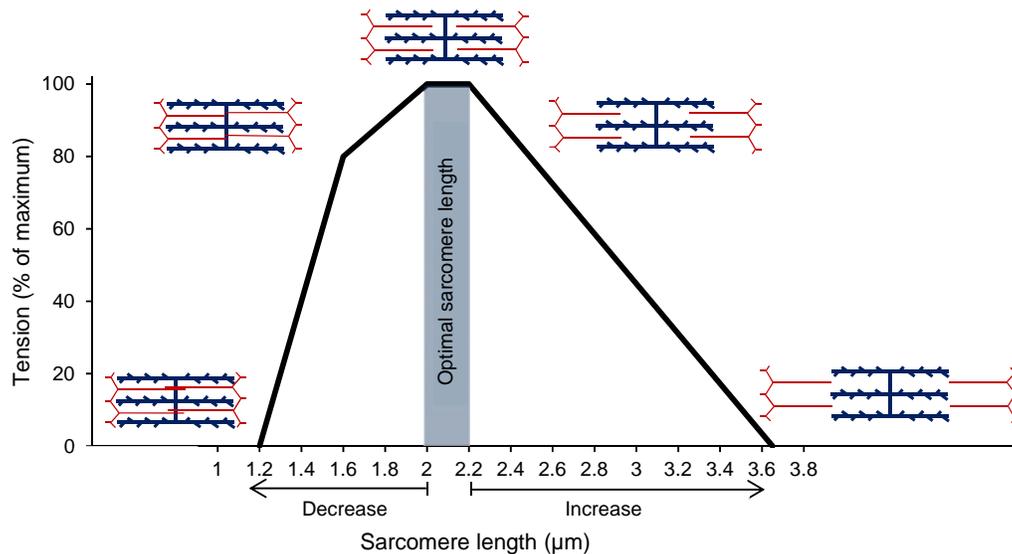


Figure 2.6. Relationship between tension and sarcomere length of skeletal muscle. Optimal sarcomere length occurs when the interaction between myosin (blue lines) and actin (red lines) filaments is greatest. Tension output decreases outside of this optimal range as a consequence of too little or too much overlap of the filaments, altering sarcomere length. Adapted from Gordon et al. (1966).

Although it is necessary to understand the mechanics by which a single muscle fiber can produce force, it is the whole muscle comprised of thousands of single muscle fibers and connective tissues positioned about a joint which provides the necessary force for movement. Consequently, the F-V and L-T relationships of whole muscle depends not only on the aforementioned active components of contractile properties (i.e. the active processes of cross-bridge cycling, actin-myosin filament overlap) of the individual muscle fibers but also on passive structures (i.e. Hills three-element muscle model (1938)) which include series (e.g. connective tissues- endomysium, epimysium, perimysium, tendon) and parallel (e.g. the passive force of the connective tissues) and the architecture of the muscle (e.g. fiber type distribution within the muscle, pennation angle

of the fibers, and arrangement of the muscle around the joint (Lieber & Fridén, 2000; Russell et al., 2000)). Based upon the F-V and L-T relationships, work loop techniques (i.e. length vs velocity) have been used to assess the mechanical work and power (area within the loop) produced by skeletal muscle during cyclical contractions in-vitro (Marsh, 1999). However, due to obvious limitations of measuring shortening velocity and muscle length in-vivo it is not possible to ascertain the amount of power that each muscle can generate individually.

The force generated by the lower limb muscles is transferred to the skeleton via the series elements of the musculo-tendinous unit. Indeed, a large portion of the change in muscle-tendon length that occurs during dynamic movements comes from the series elements (Biewener et al., 1998). Accordingly, force production is in part dependent on the stiffness of the series elements, i.e. the tendon (Hansen et al., 2006). Using ultrasonography, tendon stiffness is determined by both its architecture (i.e. cross-sectional area and length) and its relationship between force and tendon stretch (i.e. Young's modulus) (Waugh et al., 2013). As such, muscles with short tendons (e.g. the quadriceps muscle and patella tendon) are typically stiffer than those muscles with longer tendons (i.e. the ankle plantar-flexors and Achilles tendon). The stiffer the tendon, the faster force is transmitted through the muscle-tendon unit, influencing RFD. As the stiffness of the tendon increases with the length of the muscle-tendon unit, force transfer may be slower in longer units which have greater compliancy (Wilkie, 1950).

Mechanical loading of the tendons can have a large impact on their stiffness, therefore, an individual's training history can affect force transmission by the muscle-tendon unit (Waugh et al., 2013). Sex also appears to impact tendon stiffness and the responsiveness of tendon mechanical properties to repeated loading, with females exhibiting lower values than males. These differences have been attributable in some part to continual hormone changes in females (Magnusson et al., 2007). Further, substantial inter-individual differences have been observed within similar populations, with ~30% of the variance in RTD between trained male cyclists attributable to tendon stiffness (Bojsen-Moller et al., 2005). Based on theoretical cycling models (Zajac, 2002), it could be assumed that individuals with stiffer patella tendons could transfer more force from the knee extensors which may ultimately affect the level of power transmitted to the cranks. Although, consideration should be given to the notion that the performance of the pedalling movement requires multiple muscle-tendon units working simultaneously and therefore it is the combination of these units which dictates the amount of force delivered to the crank.

The influence of tendon stiffness on power production at different cadences appears to be unexplored. However, as cadence influences the time available for muscle contraction (Figure 2.4), the tendons of the lower limb muscles need to be capable of quickly transmitting the force produced by the contractile components to the pedal to avoid the production of negative muscle

work (Andersen & Aagaard, 2006). Therefore, the combined effect of cadence and tendon stiffness may impact the amount of force the agonist muscles can deliver to the crank.

A recent systematic review has shown that strength training can increase tendon stiffness by approximately 50% (Wiesinger et al., 2015). The time course for this increase in stiffness appears to occur with long-term resistance training (i.e. greater than 12 weeks) of the knee extensors and ankle plantar flexors. The training-induced changes in stiffness were similar between the knee extensor and ankle plantar-flexor tendons (Kubo et al., 2007; Reeves et al., 2003). However, shorter duration resistance training programs of eight weeks did not appear to elicit a change in the stiffness of the ankle plantar-flexor tendon (Kubo et al., 2002). It has been reported that traditional heavy load strength training is more beneficial for improving tendon stiffness compared to plyometric and ballistic exercise training (Kubo et al., 2007). Further, training against low resistances whereby low forces are produced (i.e. at high cadences in cycling) does not have the same positive effect on tendon adaptations as training against high resistances whereby high forces are produced (i.e. low cadences in cycling) (Bohm et al., 2014).

2.3.1.3 Muscle fiber type distribution

Individual skeletal muscles are comprised of thousands of muscle fibers, with the percentage of type I, type IIa, type IIb fibers varied from one skeletal muscle to another. Most muscles contain a mix of fiber types, however the proportion of each reported vary, with reports often conflicting. The hip extensor muscles (i.e. GMAX and HAM) are reportedly made up of a greater percentage of type I muscle fibers containing approximately 44 to 60%, dependent on the study examined (Dahmane et al., 2005; Evangelidis et al., 2016; Johnson et al., 1973). Muscles extending the knee have been reported to have different fiber type compositions dependent on their functional role with mono-articular VAS displaying more type I fibers (e.g. between 45-65%) and bi-articular RF displaying slightly more type II fibers (e.g. 50-70%) (Garrett et al., 1984; Gouzi et al., 2013; Johnson et al., 1973). Mono-articular SOL which plantar-flexors the ankle is largely comprised of type I fibers in the order of 80-90%, whereas bi-articular GAS tends to have a slightly greater proportion of type I fibers ranging between 50-75% (Dahmane et al., 2005; Johnson et al., 1973).

Just as different fiber types are characterised by different limits of mechanical function (i.e. F_0 , V_0 , P_{max} and V_{opt}) the distribution of different fiber types within a muscle and the combination of different muscles within a limb has been correlated with limits of NMF. The early work of Barany (1967) noted that the V_0 of a muscle was a function of its fibre type composition, while some years later, Thorstensson (1976) showed that force generation during mono-articular knee extension was highly related to the fiber-type composition of the muscles involved in the movement. With regards to multi-joint exercise such as maximal cycling, C_{opt} has been shown to

be highly correlated with the proportion of cross-sectional area occupied by type II fibres in the vastus lateralis, with higher C_{opt} and P_{max} values associated with a higher percentage of type II fibres (Hautier et al., 1996; McCartney et al., 1983c; Pearson et al., 2006). Accordingly, C_{opt} has been suggested by some as method of indicating the relative contributions of type I and type II muscle fibres in the lower limb muscles (Sargeant, 1994). Although, it should be noted that the C_{opt} at which P_{max} is maximised is not solely specified by the mechanical properties of the muscles involved in the movement, activation-deactivation dynamics appears to play a significant role too (Neptune & Kautz, 2001; van Soest & Casius, 2000).

Overall, it is well accepted that individuals presenting with a larger proportion of type I fibers are better at performing sustained repeated contractions (e.g. endurance running) (Costill et al., 1976; Foster et al., 1978), whereas those with more type II fibers perform better in activities requiring a short period of intense (i.e. maximal) activity such as sprinting (Bar-Or et al., 1980; Inbar et al., 1981). Genetics appears to play a substantial role in muscle fiber type distribution within an individual. Simoneau and Bouchard (1995) estimated that approximately 45% of the total variance in the proportion of type I fibers in humans could be explained by genetic (i.e. inherited) factors. Further, the distribution of muscle fiber type can be altered in both un-trained and trained individuals through exercise intervention such as resistance training (Adams et al., 1993; Zaras et al., 2013) and sprint cycling training (Linossier et al., 1993).

2.3.2 Biomechanical factors

2.3.2.1 Kinetics

The shoe-pedal interface integrates the foot and lower limb with the crank arm and is the primary site of energy transfer from the cyclist to the cycle ergometer. Traditionally the pedal is positioned near or directly under the first metatarsal bone of the forefoot via flat or cleated shoes allowing the foot to act as a rigid platform for force transfer from lower limb joints to the pedal (Raasch et al., 1997). Effective or tangential force acts perpendicular to the crank driving the crank forwards, while the ineffective or radial component acts parallel to the crank, contributing little useful external work (Cavanagh & Sanderson, 1986). Using sophisticated measurement systems the force applied to the left and right cranks can be measured independently via strain gauges. Assessment of these kinetic profiles shows that effective force or crank torque/tangential force for a single pedal varies throughout the pedal cycle. Typically, a large positive propulsive force occurs in the downstroke phase at around 90° (Figure 2.7) while minimal or negative forces occur in the upstroke phase during both submaximal and maximal cycling (Dorel et al., 2010; Dorel et al., 2012; Gregor et al., 1985) (Figure 2.7). The negative values observed indicate that tangential pedal force is in the opposite direction to that observed for the crank, which results in a force that

is resistive for the contra-lateral limb (Coyle et al., 1991). At the top (i.e. TDC) and bottom (i.e. BDC) of the torque is low as the forces applied to the pedal are not directed toward rotating the crank. As the two pedals on a bicycle are connected, rotating 180° out of phase, the combined effect of the forces acting on both pedals represents total crank torque and which is commonly measured. Total crank torque can be quantified using commercially available systems such as SRM power meters which have been used in research, providing valid information regarding total torque and power (i.e. the sum of the force produced by the left and right legs) derived from the chain ring (Abbiss et al., 2009; Duc et al., 2007; Gardner et al., 2004). Like tangential or effective forces, total crank torque varies across a pedal cycle, with two distinct peaks corresponding to left and right downstroke portions of the pedal cycle, as illustrated in Figure 2.7. Although, unlike torque measured from a single pedal, there is no negative component observed. This is because each of the peaks observed represents the downstroke pedal force for one side (i.e. the right) as well as the upstroke pedal force for the contralateral side (i.e. the left). Two lows occurring within the torque profile indicate the transitions of the two cranks through the TDC/BDC of the pedal cycle. Although the total crank torque approach of assessing forces applied to the pedal/crank is well used in research (Abbiss et al., 2009; Barratt, 2008) and offers a cost effective solution, it is unable to offer the same level of detail as the assessment of single pedal forces like outlined above.

A greater crank power output can be achieved by increasing the magnitude of the effective force applied during the downstroke (Dorel et al., 2010) and/or through an improvement in pedal force effectiveness (i.e. ratio of effective force and resultant force) via a change in pedalling technique (Bini et al., 2013; Korff et al., 2007). Although, the general pattern of force applied to the crank (total or tangential) has been illustrated over the pedal cycle, the pattern can be perturbed by increasing workload (Dorel et al., 2012), cadence (Samozino et al., 2007; Sarre & Lepers, 2007) and changing the kinematics of the lower limb joints (Caldwell 1998). Dorel et al. (2012) documented that increasing exercise intensity from submaximal (150 W) to maximal cycling generated more positive torque during the upstroke phase, while Sarre and Lepers (2007) and Samozino et al. (2007) showed that peak crank torque occurred later in the pedal cycle as cadence increased (e.g. a forward shift of ~20° occurred between 123 rpm to 170 rpm).

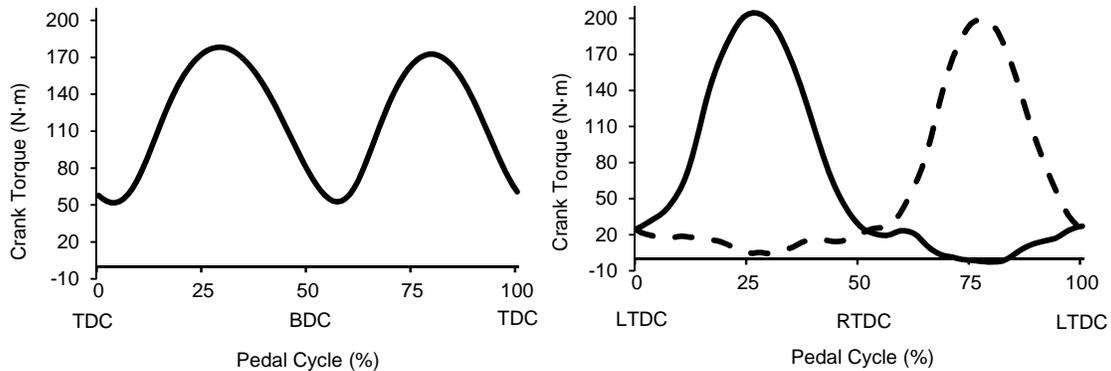


Figure 2.7. Crank torque profiles. **A:** torque profile from SRM cranks measuring total crank torque (i.e. sum of left and right cranks) and **B:** torque profiles from Axis cranks measuring the torque applied to the left and right crank separately. Solid line shows torque applied to the left crank, dashed line shows torque applied to the right crank. TDC indicates top-dead-centre, BDC indicates bottom-dead-centre, LTDC indicates left TDC, RTDC indicates right TDC.

Force measured at the pedal is composed of both muscular and non-muscular (e.g. gravity, segmental mass and inertia) components and therefore is not solely dictated by the contribution of force from the cyclist's lower limb muscles (Kautz & Hull, 1993). The effects of gravity remain fairly constant across different cadences for the same body position, though the effects of inertia appear to influence kinetic changes observed at higher cadences. More specifically, Neptune and Herzog (1999) found that non-muscular pedal forces linearly increased from low (60 rpm) to moderate (120 rpm) cadences during submaximal cycling, while the muscular component of pedal forces decreased. In a study which investigated the effect of manipulating cadence and inertia of the thigh (via the addition of masses ranging from 0 to 2 kg), altered coordination of the lower limb muscles was observed (Baum & Li, 2003). Investigating the individual and combined effects of cadence and inertia in this study, allowed these researchers to show that the inertial properties of the lower limbs in concert with cadence influence muscular activity during the pedalling movement. As such, these results can be used to understand the relative contribution of muscular and non-muscular forces on the torque vs cadence and power vs cadence relationships.

2.3.2.2 Kinematics of the lower limbs

Given that maximal muscle force is produced at an optimal muscle length (i.e. L-T relationship), optimal joint angles would lead to the maximisation of force production during single-joint and multi-joint movements. The optimisation of joint angles in movements that are multi-joint such as cycling becomes harder for the CNS to control due to movement requiring the coordinated

activation and movement of many muscles and joints moving 180° out-of-phase. As such, the kinematics of the lower limbs can be altered via a myriad of factors such as a change in saddle height, body position, crank length and distance of the axis of pedal rotation in relation to the ankle joint (Bobbert et al., 2016; Christiansen et al., 2008; Danny & Landwer, 2000; Inbar et al., 1983; Martin & Spirduso, 2001). Accordingly, to enable thorough assessment of the effect of lower limb kinematics on NMF these variables must be considered.

During maximal cycling exercise the range of motion and angular velocities reached by the ankle have been shown to be quite narrow in comparison to that exhibited by the proximal hip and knee joints (Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014). Recently, McDaniel and colleagues (2014) showed that a higher and greater range of velocities was reached by the knee joint (~ 150 to $425^\circ \cdot s^{-1}$) compared to the hip (~ 80 - $250^\circ \cdot s^{-1}$) and ankle (~ 80 - $110^\circ \cdot s^{-1}$) joints during maximal cycling exercise over a cadence range between 60 and 180 rpm. The results from this study suggest that not all muscles involved in the pedalling movement are shortening at the same velocity at a given cadence and these muscles may be operating at different parts of the F-V relationship. Similarly, at a moderate cadence of 120 rpm the ankle has an approximate range of motion of 30°, while values for the hip and knee are much larger at approximately 50° and 75° respectively (Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014). These results indicate that the muscles surrounding the hip and knee joints may be operating at a greater range of muscle lengths compared to the ankle (i.e. different sections of the L-T relationships).

Majority of studies investigating the lower limb kinematics during cycling exercise assess the movement of the joints in the sagittal plane (i.e. antero-posterior dividing the body into left and right) allowing hip and knee flexion and extension and ankle plantar-flexion and dorsi-flexion to be assessed. Typically, two dimensional (2D) video-based motion analysis measurements are used in these studies to quantify joint angles and derived range of motion, as well as joint angular velocity. However, as cycling involves out-of-plane limb motions, more sophisticated three dimensional (3D) motion capture systems (e.g. Vicon motion capture and Optotrak Certus motion tracking) in concert with the use of 3D position data, 3D joint angle computation methods can be used provide a more sensitive quantification of joint angles and angular velocities (Chiari et al., 2005). Getting accurate 3D locations of body markers contributes only one small part in the process of accurately defining joint motion. More specifically, errors in joint motion can occur from mis-location of calibration markers and from poor positioning of tracking markers (e.g. soft tissue artefact and wobbling body mass) so should be minimised where possible (Leardini et al., 2005).

2.3.2.3 Joint powers

Using kinematic data (i.e. joint angles, angular velocities), kinetic data (i.e. pedal forces) and the inertial properties of the body, estimations of the amount of force generated by the muscles and the amount of power produced at the joints can be calculated via the method of inverse dynamics (Broker & Gregor, 1994; Hasson et al., 2008; Martin & Brown, 2009). The application of this biomechanical analysis in maximal cycling has shown that the lower limb joints exhibit joint-specific parabolic relationships between power and cadence, with the apex of curve (i.e. maximal joint power) occurring at around 120 rpm for hip and knee joints. This cadence is in line with that mentioned previously in this review for the C_{opt} at which P_{max} occurs (Dorel et al., 2005; Gardner et al., 2007; Martin et al., 2000b). The relative contribution of the ankle to overall external power decreases as cadence increases (i.e. contributes approximately 18% at 60 rpm but only 10% at 180 rpm), while the contributions of the hip and knee increase from near 38% to 45% (McDaniel et al., 2014). More specifically, when assessing the contribution of the joints based upon their joint action (i.e. extension or flexion) with increasing cadence, relative hip extension and knee flexion power increased, whereas relative hip and ankle plantar flexion powers were reduced. Also, the amount of power produced by the joints varies over a pedal cycle. The ankle joint produces the greatest amount of power in synchrony with the hip and knee during the downstroke phase (i.e. 0-50% of the pedal cycle) but contributes very little during the upstroke phase. Due to the bi-articular nature of several lower limb muscles crossing the knee joint (e.g. HAM, GA, RF) power produced at this joint exhibit a double burst at the beginning of the downstroke and upstroke portions of the pedal cycle, irrespective of cadence. Regardless of cycling intensity (i.e. maximal or submaximal) hip extension is the predominant power producing action, while power produced during knee flexion is much higher than that observed at submaximal intensities (Elmer et al., 2011; McDaniel et al., 2014). Similarly, the contribution of the upper body segments appears to be greater at maximal cycling intensities indicated by a larger transfer power from the pelvis to the leg, particularly during the extension phase of the pedal cycle (Elmer et al., 2011; Turpin et al., 2016).

2.3.3 Motor control and motor learning factors

Motor control is the underlying process for how humans initiate, control and regulate the muscles and limbs upon performance of a voluntary movement or motor task, which requires the cooperative interaction between the CNS (consisting of the brain and spinal cord) and the musculoskeletal system. The first step in initiating a movement is the receipt of information by the prefrontal motor cortex, regarding the goal of the intended movement or task. The primary motor cortex generates a neural signal descending down its axons through the pyramidal tract of

the spinal cord. Neurons in the pyramidal tract (more specifically the corticospinal tract) relay the signal down the spinal cord exciting the alpha motor neurons that initiate the sequence of muscle contraction (see section 2.3.1) in those skeletal muscles/muscle groups required to perform the movement. To ensure the stability or control of a task executed the CNS receives constant sensory (afferent) feedback from proprioceptors (e.g. Golgi tendon organs and muscle spindle receptors) about limb position and exerted force (Gandevia, 1996). This feedback is used to adjust and correct the subsequent descending neural drive and thus the planning and execution of the task. At the level of the spinal cord, central pattern generators have been shown to help regulate motoneuron firing through the receipt of sensory feedback (Pearson, 1995). Central pattern generators are located between the brain and the motor neurons and have been shown to produce automatic movements such as locomotion through coordinated motor patterns (Brown, 1911; Pearson & Gordon, 2000). In ballistic movements, due to their rapidity, sensory feedback cannot be relied upon to the same extent and instead the movement is regulated using feedforward control (i.e. responding to a control signal in a pre-defined way) (Kawato, 1999). Although it is suggested that the optimal control of movement is suggested to result from a combination of both feedback and feedforward processes (Desmurget & Grafton, 2000). Practice of a particular skill or task improves the automaticity of the movement, requiring less conscious control. This can be described by the concept of a motor program, which is defined as the establishment of precise timing of muscle activations to achieve a given movement or task. Using EMG analyses, the existence of motor programs have been suggested to control locomotion (e.g. walking and running) (d'Avella & Bizzi, 2005; Ivanenko et al., 2004, 2006).

Due to the multiple degrees of freedom available to the motor system within the body's subsystems, there exist multiple ways in which a movement can be executed to achieve the same task goal. This 'problem' arises from the redundancy of the motor system, first illustrated by Nikolai Bernstein (1967) through the observation of the hammering technique of expert blacksmiths. Bernstein found that while the end point of the hammer strokes were consistent with repeated execution of the task (i.e. low between-trial/within-subject variability of hammer trajectory), the kinematic patterns executed at the shoulder, elbow and wrist varied with each repetition (i.e. greater between-trial/within-subject variability). Redundancy has long been considered a problem for the motor system. However, this classical formulation has been questioned by researchers who suggest that the CNS does not suffer from a problem of motor redundancy, but instead may be fortunate to have the "bliss of motor abundance" (Gelfand & Latash, 1998; Latash, 2000; Latash, 2012). The multiple degrees of freedom of the motor system provide greater flexibility for performing a movement but also make understanding the control of movement very complex, particularly for tasks that are multi-joint, such as maximal cycling exercise.

Several studies have highlighted that the CNS reduces the number of coordination strategies required to accomplish a task goal (e.g. the maximisation of power), in an attempt to reduce the complexity of the pedalling movement (Raasch et al., 1997; van Soest & Casius, 2000; Yoshihuku & Herzog, 1996). One particular strategy which has been evidenced by EMG and modelling analyses is that the CNS divides the neural drive between groups of muscles (i.e. muscle synergies), instead of each individual muscle, as a means to simplify the number of motor outputs required for a given task. The notion of muscle synergies have been shown for walking (Cappellini et al., 2006), upper limb reaching movements (d'Avella et al., 2008), rowing (Turpin et al., 2011) and cycling (Hug et al., 2010; Raasch & Zajac, 1999). Specific to the pedalling movement, the CNS appears to simplify the control of pedalling movement by sending a common neural drive to only three or four groups of muscles (or synergies). More specifically, Raasch and Zajac (1999) identified an extensor group (over the downstroke phase), a flexor group (during the upstroke phase) and two groups acting across TDC (RF and TA) and BDC (HAM, GAS and SOL) transition zones respectively; while several years later Hug et al. (2010) using EMG identified three synergies: 1) knee (VAS and RF) and hip (GMAX) extensors; 2) knee flexors (HAM) and ankle plantar-flexors (GAS); and 3) ankle dorsi-flexors (TA) and RF (Figure 2.8). Although, the theory of muscle synergies as a motor control strategy has recently been confronted with alternative assumptions put forward such as the minimal intervention principal (Kutch & Valero-Cuevas, 2012; Valero-Cuevas et al., 2009).

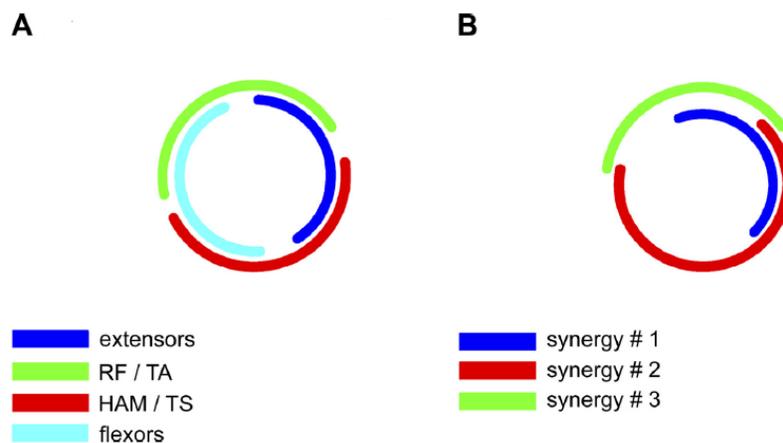


Figure 2.8. Schematic representations of muscle synergies identified for maximal cycling. **A:** illustrates synergies identified by Raasch and Zajac (1999), while **B:** illustrates synergies identified by Hug et al. (2010). Synergy # 1 includes VAS, RF and GMAX, synergy #2 includes HAM and GAS and synergy #3 includes TA and RF. Taken from Hug et al. (2010).

2.3.3.1 *Changes in inter-muscular coordination*

As outlined in section 2.3.1.1 above, individually the lower limb muscles have different functional roles and patterns of activation throughout a pedal cycle; however the effective application of force to the crank requires coordination of all these muscles (i.e. inter-muscular coordination). Inter-muscular coordination provides an insight into how the CNS and musculoskeletal systems interact to perform a movement or task (Pandy & Zajac, 1991). Indeed, previous studies have illustrated that optimal patterns of muscle activation and co-activation of the lower limb muscles determines how muscle power is transferred to the crank, and the resulting level of maximal external power produced (Dorel et al., 2012; Hug et al., 2011; Raasch et al., 1997; Rouffet & Hautier, 2008; van Ingen Schenau, 1989). Using normalised EMG profiles the co-activation (or co-contraction) of two muscles during a given time frame can be quantified using an equation to calculate an index of co-activation. This index has been used previously to assess muscle co-activation with regards to joint laxity (Lewek et al., 2004), knee osteoarthritis (Hubley-Kozey et al., 2009), walking (Arias et al., 2012) and more recently fatigue in sprint cycling (O'Bryan et al., 2014).

The co-activation of agonist-antagonist muscle pairs (e.g. GMAX-RF and VAS-HAM) is necessary in activities such as running, jumping and cycling to transfer forces across the lower limb joints and control the movement being executed (i.e. the direction of external force) (van Ingen Schenau, 1989; Van Ingen Schenau et al., 1992). Although, the co-activation of these opposing muscle pairs has been suggested as uneconomical due to their contributing forces cancelling out (Gregor et al., 1985). Further, the co-activation of agonist-antagonist muscle pairs has been suggested to provide joint stability (Hirokawa, 1991). EMG analyses have also indicated that the coordination of the lower limb muscles are sensitive to factors such as training history (e.g. novice vs trained cyclist (Chapman et al., 2008a)), power output (e.g. submaximal vs maximal) (Dorel et al., 2012; Ericson, 1986), pedalling rate (Baum & Li, 2003; Marsh & Martin, 1995; Neptune et al., 1997; Samozino et al., 2007), cycling posture and surface incline (Li & Caldwell, 1998), bicycle setup (Ericson, 1986), shoe/pedal interface (Cruz & Bankoff, 2001) and fatigue (Dorel et al., 2009; O'Bryan et al., 2014).

2.3.3.2 *Changes in movement variability*

The redundancy within the motor system allows the CNS to produce slightly different EMG and joint motion patterns even when the movement or task goal performed is the same (Bernstein, 1967; Srinivasan & Mathiassen, 2012). Indeed, a level of variability (albeit low for some) exists with the execution of a movement task, regardless of the population, stemming from inherent variability within the subsystems of the neuro-musculo-skeletal systems. Even during repetitive

tasks that are highly constrained (e.g. the kinematics of the cycling movement), variability is ever present (Enders et al., 2013). The views of Hamill (1999) suggests that variability is an essential part of normal human functioning supporting the dynamical systems approach to motor control. Several different theories and methods (e.g. variance ratios, coefficient of variation) exist for the assessment of variability which can make interpretation across studies difficult. In cycling research the calculation of a variance ratio (VR) has been used for the analysis of intra-individual and inter-individual variability of EMG and mechanical patterns (Burden et al., 2003; Hug et al., 2008; Rouffet & Hautier, 2008). Using this method a lower VR indicates less variability in the pattern assessed. Using VR and other analyses such as coefficient of variation, the variability of lower limb EMG and joint kinematics during the pedalling movement have been shown to depend on the individual muscle (typically dependent on the architecture and function of the muscle, i.e. mono-articular vs bi-articular) (Hug et al., 2008; Ryan & Gregor, 1992); exercise intensity and the skill of the population (Chapman et al., 2006; Hug et al., 2004).

In contrast to simple movements (e.g. those involving a single-joint), multi-segment movements (such as cycling) for which muscles contribute to forces acting at joints they do not cross can further complicate the understanding of motor variability (Zajac & Gordon, 1989), despite the cycling movement being constrained by the circular trajectory of the pedal, unlike open kinetic chain movements such as running and jumping. Ryan and Gregor (1992) were some of the first authors to investigate the between-cycle variability in EMG patterns during cycling at a cadence of 90 rpm and workload of 250 W. Their analyses showed that single-joint GMAX, VM and VL exhibited the least amount of variability while bi-articular HAM had the highest. Some years later, Hug et al. (2008) found that EMG patterns of bi-articular muscles (but in particular RF and HAM) demonstrated higher levels of inter-individual variability at submaximal intensities (150 W and 250 W), however this level of variability was not reflected in the level of force applied to the pedal (i.e. kinetics). When interpreted using concepts presented by Muller and Sternad (2009), the result variable (e.g. force amplitude) could still be maintained through different execution variables (e.g. activation of muscles exercises). Although these findings provide detail for constant cycling, it has been shown that muscular demand (e.g. power output) appears to affect EMG variability, with less variability occurring at 300 W than at 150 W (Enders et al., 2013). These authors and others have noted that the solution space (defined as the combination of solutions that are actually used by humans rather than the theoretical number of possible solutions available to the task) reduces as the intensity of the exercise increases or when the task being performed needs to be executed in a specific manner (Hasson et al., 2012).

As noted in a plethora of motor control experiments, variability can be reduced following a period of practice interpreted as an indicator of learning and thus improved performance of the given task (Muller & Sternad, 2009). Sides and Wilson (2012) reported that lower levels of

variability in patterns of joint motion and force application for expert performers of a task. With regards to training status and history, those unskilled in the pedalling movement appear to exhibit muscle recruitment strategies that are less refined than those who are trained, evidenced by greater intra-individual variation in EMG patterning (Chapman et al., 2008a). Bi-articular GAS and BF have been shown to exhibit less variable patterns of activation in well-trained cyclists compared to novices, potentially reflective of a strategy to control force across BDC through to the upstroke (De Marchis et al., 2013; Hug et al., 2010). In a later study by Chapman (2009) cycling novices exhibited greater variability in the lower limb joints which was considered to be reflective of meagre movement skill. In contrast to the reduction observed in most of the lower limb muscles, TA presented more inter-individual variability for competitive riders (Neptune et al., 1997), while the kinematics of the ankle joint varied within a cohort of similar ability cyclists (Kautz et al., 1991). Although variability appears to be lower in most muscles for those highly trained in cycling, a low-level of EMG variability is still present (Hug et al., 2004; Hug et al., 2008). Based on these findings and others that suggest that the CNS is already employing the most effective coordination strategy to produce maximal power output (Raasch et al., 1997; Yoshihuku & Herzog, 1996), EMG variability would be further reduced when cycling is performed at maximal intensities (e.g. sprinting), however there is currently no research to support this. Further the effect of variability in EMG or kinematic profiles on the limits of NMF have not been previously explored. Knowing that the CNS cannot eliminate variability and that the role of variability is equivocal, perhaps understanding what the optimal level of variability for a given movement is may be most important.

2.4 Methodological considerations for assessing NMF on a stationary cycle ergometer

There are several exercises which can assess the NMF of the lower limbs including vertical jump (Samozino et al., 2010), leg extension (isometric and dynamic), leg press (Bobbert, 2012), inclined push offs (Samozino et al., 2014) and maximal cycling exercise (Arsac et al., 1996; Buttelli et al., 1996; Dorel et al., 2010; Dorel et al., 2005; Driss et al., 2002; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007; Sargeant et al., 1981; Seck et al., 1995). With regards to maximal cycling exercise, the limits of NMF (i.e. maximal torque, maximal cadence, maximal power and corresponding optimal cadence) are assessed using a force-velocity (F-V) test on a stationary cycle ergometer.

2.4.1 Familiarity with stationary cycle ergometers

Prior experience cycling on a stationary ergometer appears to influence the validity and reliability of measured maximal power. Martin and colleagues (2000a) found that maximal power did not differ within and between 3.5-s sprint bouts over four days in cycle-trained men. On the contrary, active men who had limited cycling experience took approximately 10 exercise bouts before stable power values could be produced. Increases in power of 5.1% was observed from exercise bout one and two on the first day, 4.3% between the first and second days and 2.5% between the second and third day. These results reiterated those of Capriotti et al. (1999) who also showed that values of mean power were stable following two days of practice. Doré et al. (2003) and Mendez-Villanueva et al. (2007) on the other hand showed that one familiarisation session was adequate to obtain reproducible measurements of maximal power in young non-cyclist adults performing 5 to 8-s sprint bouts, although Mendez-Villanueva et al. (2007) did suggest that a second familiarisation session could be included to ensure greater stability of power outputs.

2.4.2 Test protocols

The approach used to investigate the relationship between force and velocity of muscles in-vitro (Fenn & Marsh, 1935; Hill, 1938) and in-vivo (Bobbert, 2012; Thorstensson et al., 1976a; Wilkie, 1950) requires the implementation of testing protocols that allow maximal levels of force and power at different contraction velocities to be recorded. Accordingly, the assessment of torque vs cadence and power vs cadence relationships during maximal leg cycling exercises should follow a similar method, enabling the measurement and selection of maximal levels of torque and power over a wide range of cadences. Various testing protocols, usually referred to as a Force-Velocity (F-V) (or Torque-Velocity) test, are used to collect the experimental data required to characterise the relationships between torque (i.e. the product of the tangential component of external force applied to the pedal/crank and the crank length) and cadence, but also power (i.e. the product of crank torque and cadence) and cadence during leg cycling exercises (Arsac et al., 1996; Buttelli et al., 1996; Dorel et al., 2010; Dorel et al., 2005; Driss et al., 2002; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007; Sargeant et al., 1981; Seck et al., 1995).

During maximal intensity exercise the onset of fatigue is rapid (<5-s), with the exercised muscles ability to generate and sustain force limited by the supply of adenosine triphosphate and the diminishing replenishment from phosphocreatine (McComas, 1996). Fatigue development does not just limit the supply of adenosine triphosphate it also influences other physiological processes that impairs contractility of the muscle. Additionally, fatigue can affect biomechanical (e.g. alterations in limb movement) and motor control processes (e.g. movement patterns and

coordination) (Dingwell et al., 2008). As such, there is no single cause of fatigue, with the interaction of physiological, biomechanical and motor control processes changing the demands of the task performed (Enoka & Duchateau, 2008).

Typically, the duration of a sprint bout performed as part of a F-V test is one that would allow fatigue-free data to be collected (i.e. a duration of 3 to 4-s). However, a review of the literature revealed that several studies have employed sprints lasting longer than a duration considered to be relatively fatigue-free (i.e. more than 5-s) (Arsac et al., 1996; Davies et al., 1984; Samozino et al., 2007; Sargeant et al., 1981). As such, the possibility of fatigue affecting the level of torque and power produced towards the end of these sprints (typically occurring at higher cadences) is increased. Therefore, the duration of a sprint bout should be taken into account when employing a F-V test, especially as the P-C relationship can be used to investigate fatigue related changes in power (Gardner et al., 2009). The length and type of recovery between exercise bouts is also an important consideration for F-V tests requiring multiple sprints. Enough time must be given between sprints to ensure phosphocreatine stores are replenished (i.e. conversion of adenosine diphosphate to adenosine triphosphate), while still maintaining the performance enhancing effects of muscle potentiation from previous muscle contractions (Robbins, 2005). Previous studies have shown that the effect of fatigue can be avoided by separating repeated sprint efforts (<7-s in duration) by 5-min of passive recovery (Linossier et al., 1996; Seck et al., 1995).

Torque and power outputs obtained from a F-V test can also depend on a variety of other factors including modifiable parts of the cycle ergometer (e.g. crank length, pedals, seat height) (Inbar et al., 1983; Martin & Spirduso, 2001) and positioning of the body (e.g. seated or standing) (Bertucci et al., 2005; Reiser Ii et al., 2002). For example, it has been shown that body position on a cycle ergometer can affect power production (Welbergen & Clijsen, 1990), alter muscle coordination (Chapman et al., 2008b; Ericson et al., 1985) and kinematic patterns of the lower limbs (Bini et al., 2010). Typically in maximal cycling research participants are asked to remain seated on the saddle and maintain hand position on the dropped portion of the handlebars to allow for a more robust standardisation of body position (Dorel et al., 2010; Dorel et al., 2005; Rouffet & Hautier, 2008). In fact, studies have shown that the musculature of the upper body contributes to power production when a standing position is adopted (Stone & Hull, 1993). Subsequently, standardisation of the cycle ergometer set up and testing protocol appears imperative to allow valid comparison of results obtained between subjects and also between testing sessions. In addition to the duration of the sprint bouts used and an individual's set-up on a cycle ergometer, the type of ergometer (e.g. isokinetic or isoinertial) the F-V test is performed on has important implications for P-C and T-C relationships.

2.4.2.1 Isokinetic ergometers

Isokinetic ergometers provide a variable resistance to a constant velocity whereby power is calculated as a product of the constant velocity (or cadence) and the force applied to the pedal. In this mode, maximal sprints are performed at various set cadences allowing a series of experimental data points to be obtained for the given cadence of interest (Beelen & Sargeant, 1991; McCartney et al., 1983a; McCartney et al., 1983c; McCartney et al., 1985; Sargeant et al., 1981). Subsequently, the highest value of torque and power can be chosen from the data points collected for a given cadence taking into consideration the inherent variability observed in the motor system (Bernstein, 1967; Latash, 2012) that could affect the force and power output achieved from one pedal cycle to the next (as outlined in section 2.3.3). Further, isokinetic F-V protocols can use the advantageous effects of post-activation potentiation, improving power at a set cadence from muscular contractions occurring in the preceding pedal cycles (Robbins, 2005; Sale, 2002). Most studies employing F-V tests on an isokinetic dynamometer include approximately 10 sprints to enable torque and power to be assessed at 10 different (normally evenly spaced) cadences. Although, the high number of sprints typically required is a limitation of this approach as they may cause fatigue, affecting sprints performed during the latter part of the F-V test. Also, isokinetic cycle ergometers tend to be more expensive than a standard isoinertial ergometer making them less obtainable in both research and clinical settings.

2.4.2.2 Isoinertial ergometers

The second and apparently more popular method of F-V test protocol used in the literature is with the use of an isoinertial cycle ergometer. With isoinertial cycling, a single (Martin et al., 2000a; Martin et al., 1997; Seck et al., 1995) or series (Arsac et al., 1996) of maximal sprint efforts are performed against constant external resistances applied to the flywheel of the ergometer allowing a series of experimental points to be obtained at various cadences from a single sprint bout. In 1997, Martin and colleagues published an article outlining how maximal cycling power could be determined with the use of a single-effort using the inertial-load method which was employed in later studies by this group (Gardner et al., 2007; Martin et al., 2000a). With the inertial-load method the resistance at which subjects cycled against was provided solely by the moment of inertia of the flywheel. Subjects start their single effort with the flywheel stationary and accelerate maximally for 3 to 4-s, which provides data from 6.5 pedal revolutions, typically between cadences of approximately 80 rpm to 175 rpm. Although, these authors showed that this method was both valid and reliable (an intraclass correlation coefficient of 0.99 was obtained), it does not allow for the inherent variability of the motor system (Bernstein, 1967; Latash, 2012) and the inability of humans to consistently produce a maximal effort over

continuous pedal cycles, particularly for those unskilled in the pedalling movement. Additionally, using the single-sprint inertial-load method average power was only calculated from 80 rpm upwards, leaving lower cadences unexplored which may result in errors if torque and power are extrapolated beyond the experimental data collected. Further, the cadence at which a sprint is initiated may also have an effect on the level of torque and power that can be produced. Studies implementing a single all-out effort such as those employed by Martin et al. (1997) start the sprint with the flywheel stationary (i.e. zero cadence). However with this type of sprint start there are no preceding muscular contractions before the commencement of the maximal effort to enhance the neuromuscular state of the muscle and the amount of force that can be produced (Robbins, 2005; Sale, 2002).

In contrast, Arsac et al. (1996) employed six all-out 8-s sprints against constant external resistances ranging between 0.25 to 0.75 N.kg⁻¹ which permitted a total of 126 to 204 data points from all six resistances over a cadence range of 17-214 rpm. The acquisition of such a large number of data points would allow data points corresponding to the highest level of power/torque for a given cadence to be chosen (removing the effect of movement/activation variability) allowing for a better assessment of an individual's maximal power/torque. However, it appears that Arsac et al. (1996) included all data points they collected during the F-V test to characterise torque vs cadence and power vs cadence relationships. As such, the multiple experimental data points obtained for a given cadence may not all reach the same level of torque or power which has the potential to reduce values of torque and power predicted from these relationships.

2.4.3 The inability to consistently produce maximal levels of torque and power

Combining the findings of Bernstein (1967) regarding motor system variability with those from Allen (1995) regarding muscle active state it appears that humans may not be able to activate their muscles maximally and optimally every time a movement is executed (e.g. over recurring pedal cycles), which may affect the level of maximal torque and power output achieved. This is of particular importance if the movement or task (e.g. maximal cycling exercise) performed requires the coordination of multiple muscles, as not all muscles will be active to the same level throughout the movement (e.g. throughout a pedal cycle). Also, due to more variability observed in non-expert performers, if the pedalling movement is novel, then consistent levels of torque and power may not be produced between recurring pedal cycles for these individuals. The findings of Martin et al. (2000a), Capriotti et al. (1999), Doré et al. (2003) and Mendez-Villanueva et al. (2007) are reiterated here, as each of these studies showed that familiarity with performing maximal cycling on a stationary cycle ergometer affects power outputs. When designing or choosing a test to assess NMF it seems important to keep these factors in mind to ensure that the actual limits of NMF are

being reported correctly and not underestimated. Due to the discrepancy in the ergometers and protocols used by researchers, the number of data points and range of cadences over which they are collected during a F-V test appears to vary between studies (Table 2.1). Visualisation of T-C and P-C relationships illustrated in the studies listed in Table 2.1 indicates that data points obtained using a single or series of all-out efforts using isoinertial cycle ergometers may not be distributed evenly across the range of cadences covered (like observed using isokinetic cycle ergometers) and all may not be truly maximal (i.e. different values of torque and power observed for the same cadence). For example, Doré et al. (2003) and Hintzy et al. (1999) collected approximately 20 to 50 points from their F-V tests, while the Martin group used less than 10 (Martin et al., 2000a; Martin & Spirduso, 2001; Martin et al., 1997). It is unknown if authors implementing F-V tests filter the experimental data prior to creation of T-C and P-C relationships as little to no mention regarding the process is given in the methods section. As a F-V test provides information regarding maximal torque and power output and thus allows the calculation of the limits of NMF, it seems imperative that only experimental data reflecting true maximal values for a given cadence are included.

In order to accurately assess the limits of lower limb NMF, the performance of indeed a maximal effort by the individual is required. Prior to performing a sprint, participants are instructed to execute an all-out effort or reach the highest level of acceleration possible within the bout if performed on an isoinertial ergometer. However, intrinsic motivation can influence human behaviour and thus the level of effort given (Ryan & Deci, 2000). Consequently, studies employing exercise protocols such as F-V tests which require a maximal effort note that participants were provided with a high degree of external encouragement throughout the exercise bout to help facilitate a maximal effort (Arsac et al., 1996; Dorel et al., 2005).

2.4.4 Prediction of power-cadence and torque-cadence relationships

As previously outlined in this review single muscle fibers and single muscles (i.e. single joint movements), exhibit a F-V relationship that is hyperbolic (Perrine & Edgerton, 1978; Thorstensson et al., 1976a; Wilkie, 1950). Although, the nature of this relationship differs when the movement being performed involves multiple joints and muscles such as half-squats (Rahmani et al., 2001), leg press (Bobbert, 2012; Samozino et al., 2012), arm cranking (Driss et al., 1998; Jaafar et al., 2015), jumping (Bobbert & Van Ingen Schenau, 1990) and cycling (Bobbert et al., 2016). The shape of the relationship between external force and velocity obtained during these multi-joint exercises is usually described as linear or “almost linear” (Bobbert, 2012). It has been proposed that neural factors may account for the shift from a hyperbolic to linear appearance in voluntary movements of greater complexity (Yamauchi et al., 2007). A recent

review by Jaric (2015) outlined that the F-V relationships of maximal multi-joint movements (including cycling) appear to be reliable, valid and accurate enough to detect differences in maximal force, maximal power and maximal velocity within and between sporting and clinical populations. However, in the case of maximal leg cycling, there is a lack of consensus regarding the most appropriate way to model and therefore characterise the shape of T-C and P-C relationships (see Table 2.1).

The earliest study by Dickinson in 1928 described the relationship between cadence and braking force as linear. Studies conducted some years later using an isokinetic cycle ergometer showed that the relationship between torque and cadence was linear (Sargeant et al., 1981). Thereafter, in the majority of studies, researchers have predicted T-C relationships using linear regressions, while P-C relationships were consequently characterised using symmetrical parabolas (i.e. second order polynomial/quadratic regressions) (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007) as depicted in Figure 2.5B. Maximal T-C and P-C relationships have also been described during the acceleration phase (i.e. downstroke) of a single all-out exercise (Seck et al., 1995). Further, one study applied the same method of prediction, to describe the P-C and T-C relationships for different phases of the pedalling movement, allowing cadence-specific power and torque to be calculated during the downstroke (i.e. 0-180° of a pedal cycle) and upstroke (180-360° of a pedal cycle) (Dorel et al., 2010).

However, as presented in a paper by Vandewalle et al. (1987) the force-velocity relationship during cycle ergometry was suggested to exhibit an approximately linear relationship, as the force values obtained at heavy and light braking forces were downwardly inflected. In later studies, some researchers opted to employ higher order polynomials to predict T-C and P-C relationships using second and third order polynomials/cubic regressions respectively, refuting simple linear and symmetrical parabola shapes (Arsac et al., 1996; Hautier et al., 1996; Yeo et al., 2015). The earliest of these studies demonstrated that a higher order regression provided systematically higher r values compared with a linear function (0.96 ± 0.02 vs 0.91 ± 0.04) for T-C relationships (Arsac et al., 1996). Although different modelling procedures were not investigated for P-C relationships, as a consequence of second order polynomials employed for T-C relationships, these researchers used a third order polynomial to characterise P-C relationships, which resulted in individual r values between 0.95 and 0.99. In a more recent study by Yeo et al. (2015) T-C relationships were well fit with second order polynomials (r^2 of 0.99 for both conditions), while third order polynomials provided a good fit for P-C relationships (r^2 of 0.97 for both conditions). Other than the loose investigation of regression order comparison by Arsac and colleagues (1996), it appears that no direct comparison between the different modelling procedures typically employed in the literature has been conducted previously.

Considering the results reported from F-V tests of other movements, the implementation of higher order polynomials may offer greater flexibility for modelling T-C and P-C relationships, providing a more comprehensive approach for characterising the shapes of these relationships (and the calculation of the limits of NMF). Recently, Bobbert (2012) showed that the force vs velocity relationship was not perfectly linear for maximal leg press, a multi-joint movement requiring a low level of external force control. As depicted in Figure 2.5B the dotted line on the F-V relationship indicates the reduction in total external force observed by Bobbert at high forces and high velocities. Considering that maximal leg cycling is a multi-joint movement requiring a high level of external force control, the possibility that T-C relationships may be better predicted using non-linear regressions while the P-C relationships may be better predicted using parabolas of asymmetrical shapes cannot be ruled out. Similarly, due to the complexity of dynamic, poly-articular movements such as maximal cycling, it seems very plausible that not every individual would exhibit exactly the same shaped T-C and P-C relationships.

According to previous studies, a variety of phenomena (discussed in section 2.3.1 above) could potentially cause the shape of T-C relationships to deviate from linearity and the shape of P-C relationships to deviate from a symmetrical parabola in maximal cycling. Previous findings suggest that torque and power production may be limited by neural inhibitions (Babault et al., 2002; Perrine & Edgerton, 1978; Westing et al., 1991; Yamauchi et al., 2007) and/or non-maximal skeletal muscle potentiation (Robbins, 2005) when producing maximal pedalling movements at low cadences. While pedalling at high cadences, torque and power production could be limited by activation-deactivation dynamics (van Soest & Casius, 2000), alterations in the motor control strategy (McDaniel et al., 2014).

2.4.5 Key variables used to describe the limits of NMF

Due to the discrepancy in the protocols and ergometers used by researchers as outlined in Table 2.1, the range of cadences captured during a F-V test and modelling procedure employed (i.e. order of polynomial) may have an impact on the variables commonly extracted from the T-C and P-C relationships to describe the limits of NMF. The equations describing the relationships between torque and cadence is used to predict maximal torque (T_0) which corresponds to the intercept of the T-C relationship with the torque (y) axis; maximal cadence (C_0) which corresponds to the intercept with the cadence (x) axis and the slope of the relationship (when the T-C relationship is assumed to be linear) (Dorel et al., 2010; Dorel et al., 2005; Martin et al., 1997; Yeo et al., 2015). Similar to studies investigating the F-V relationship in isolated muscle or single joint movements, F_0/T_0 indicates the theoretical maximal force/torque the limbs can produce at zero velocity, while V_0/C_0 represents the theoretical maximal velocity/cadence at

which the limbs can move (Dorel et al., 2005; Gardner et al., 2007; Martin et al., 1997; Samozino et al., 2012). In a general sense, F_0/T_0 is suggested to provide an estimate of strength of the lower limbs (Driss et al., 2002), while C_0 provides an estimate of speed characteristics (Samozino et al., 2012). Extrapolated values of C_0 and T_0/F_0 have been reported as high as 260 rpm and 236 N·m respectively in elite sprint cyclists (Dorel et al., 2005; Martin et al., 1997), but somewhat less in active volunteers with ~236 rpm and ~180 N·m respectively (Dorel et al., 2010). It is common practice in studies investigating the mechanical capabilities of single muscle fibers or single muscles to include both calculated and experimental measures of V_0 via the slack test method (Claflin & Faulkner, 1985; Edman, 1979). However, in maximal cycling research C_0 is typically calculated by extrapolation and not measured experimentally. It appears that the collection of an experimental measure of maximal cadence (C_{\max}) has only been reported once in the literature by McCartney and colleagues (1985) who documented values between 181-192 rpm for their cohort of non-cyclist females. Given that the removal of the chain from the cycle ergometer is quite easy and likely to result in pedalling against an external resistance that is close to zero, the lack of experimental cadence values reported in the literature is surprising.

As power is a product of force and velocity, estimated F_0 and V_0 have been used previously as a method of calculating maximal power (P_{\max}), using $0.5F_0$ and $0.5V_0$, or $0.25V_0F_0$, based on the assumption that the T-C relationship is linear (Driss & Vandewalle, 2013; Seck et al., 1995), while $2C_{\text{opt}}$ values have been used to predict C_0 (Driss & Vandewalle, 2013), based upon the notion that the P-C relationship reflects a symmetrical parabola. However, given that the T-C and P-C relationships may not represent linear or symmetrical shapes, the implementation of these types of equations may misrepresent important information regarding the limits of the neuromuscular system.

More often in the literature P-C relationships are created alongside those for T-C (as outlined in Table 2.1), allowing the shape of the respective relationships to be visualised. Just as maximal power in isolated muscle or muscle fibers occurs at optimal shortening velocities (Figure 2.5) (Hill, 1938), it is well known that maximal mechanical power is achieved at optimal velocities/cadences (C_{opt}) during maximal cycling (Arsac et al., 1996; Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Martin et al., 1997; Samozino et al., 2007; Vandewalle et al., 1987; Yeo et al., 2015). Though, it has been shown with forward dynamics modelling (i.e. using joint torques to predict resultant motions) that C_{opt} (i.e. the cadence at which maximal power is optimised) is not only determined by the relationship between power and velocity, but also by activation-deactivation dynamics (Bobbert et al., 2016; van Soest & Casius, 2000). The regressions fit to P-C relationships can be used to predict P_{\max} at the apex of the curve and the C_{opt} to which it corresponds (Dorel et al., 2005; Driss et al., 2002; Gardner et al., 2007; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2012). Absolute P_{\max} values achieved in maximal

cycling have been reported around 1100 W in non-cyclists (Dorel et al., 2010) and up to 2500 W in highly trained elite cyclists (Dorel et al., 2005; Martin et al., 2006). When expressed in relation to body mass, power outputs of $12.4 \text{ W}\cdot\text{kg}^{-1}$ have been reported for active, non-cyclists (Davies & Sandstrom, 1989), $17.1 \text{ W}\cdot\text{kg}^{-1}$ for power athletes (Vandewalle et al., 1987) and $19.3 \text{ W}\cdot\text{kg}^{-1}$ for elite track cyclists. Values of C_{opt} typically reported range between 110 and 140 rpm for both trained and un-trained cyclists (Dorel et al., 2005; Gardner et al., 2007; Martin et al., 2000b).

Although, T_0 , C_0 , P_{max} , C_{opt} provide important information regarding the limits of NMF with some of these parameters directly linked with performance (Hautier et al., 1996; Vandewalle et al., 1987) and strength of the knee extensor muscles (Driss et al., 2002) they characterize only a few points on the T-C and P-C curves. Recent studies have gone beyond interpretation of F_0/T_0 and V_0/C_0 values separately and have assessed the F-V mechanical profile during sprint running (Morin et al., 2002), squat jumping (Giroux et al., 2016; Samozino et al., 2014) and ballistic inclined push offs (Samozino et al., 2012) using the slope of the F-V relationship calculated from a linear regression (Giroux et al., 2016; Morin et al., 2002; Samozino et al., 2014; Samozino et al., 2012). Although these studies highlight the individualism of force and velocity producing capabilities, consideration must be given to their methods for F-V line fitting procedures. The slope appears to provide a nice method for assessment of NMF, but like the calculation of T_0 and C_0 , if the relationship between force/torque and velocity is not actually linear then this approach may not be accurate.

This literature review section has outlined the current practices for the evaluation of NMF using stationary cycle ergometers. Although the F-V test has been commonly implemented to assess the limits of NMF of the lower limbs such as their torque and power producing capabilities, there exist several different methods for data collection and analysis. It appears that perhaps due to the complexity of the pedalling movement and different factors affecting the level of power that can be produced over a range of cadences (i.e. specifically either side of P_{max} at low and high cadences) that previous methods may have overestimated the limits of NMF.

Table 2.1. Summary of studies that have used force-velocity test protocols on stationary cycle ergometers.

Author(s) (Date)	Test	n	Participants	No. Sprints	Sprint Length	Rest Length	# Data Points	T-C/P-C Regressions
Arsac et al. (1996)	Isoinertial	15	Trained-Marathon, volleyball	6	8-s	5-min	~126- 200	2 nd order/ 3 rd order
Buttelli et al. (1999)	Isoinertial	11	Well-trained males	1	6-s	-	-	Linear
Capmal & Vandewalle (1997)	Isoinertial	6	3 active & 3 cyclist males	2	To maximal velocity	5-min	~13	Linear
Capmal & Vandewalle (2010)	Isoinertial	4	Competitive cyclists	1	7-s	-	~15	Linear/ 2 nd order
Davies et al.(1984)	Isokinetic	5	Healthy males	8-10	10-s	5-min	-	Linear & Hyperbolic
Doré et al. (2003)	Isoinertial	27	14 females & 13 males	4	5 to 8-s	4-min	~30	2 nd order/ 3 rd order
Dorel et al. (2005)	Isoinertial	12	Elite cyclists	3	5-s	5-min	~50	Linear/ 2 nd order
Dorel et al. (2010)	Isoinertial	14	Active males	3	5-s	5-min	-	Linear/ 2 nd order
Driss et al. (2002)	Isoinertial	12	Male volley ball players	6-8	6-s	5-min	6	Linear/ 2 nd order
Gardner et al. (2007)	Isoinertial	7	Elite cyclists	2	3 to 5-s	3-min	~12	Linear/ 2 nd order
Hautier et al. (1996)	Isoinertial	10	Trained cyclists	3	5-s	5-min	15	2 nd order/ 3 rd order
Hintzy et al. (1999)	Isoinertial	22	Trained non-cyclists	4	6-s	5-min	55	Linear/ 2 nd order
Linossier et al. (1996)	Isoinertial	10	8 men & 2 women, active	2-3	4 to 8-s	5-min	6	Linear/ 2 nd order
Martin et al. (1997)	Isoinertial	13	Active males	1	3 to 4-s	-	6.5	Linear/ 2 nd order
Martin et al. (2000)	Isoinertial	48	13 cycle trained & 35 active men	1	3 to 4-s	-	6.5	-/2 nd order
Martin & Spirduso (2001)	Isoinertial	16	Trained cyclists	1	3 to 4-s	-	6.5	-/2 nd order
McCartney et al. (1983b)	Isokinetic	12	Healthy males	6	<10-s	2-min	6	Linear
McCartney et al. (1985)	Isokinetic	7	Healthy females	10	<10-s	2-min	10	Exponential / 2 nd order
Nakamura et al. (1985)	Isoinertial	26	Active males	8	10-s	>2-min	-	Linear
Pearson et al. (2006)	Isoinertial	14	7 young & 7 older men	15	1 to 5-s	30-s	~30	-/3 rd order
Rouffet & Hautier (2008)	Isoinertial	9	Recreationally trained males	2	-	5-min	-	-
Samozino et al. (2007)	Isoinertial	11	Trained cyclists	4	8-s	5-min	12-31	Linear/ 2 nd order
Sargeant et al. (1981)	Isokinetic	5	Untrained cyclists	8	20-s	-	8	Linear/ 2 nd order
Sargeant et al.(1984)	Isokinetic	55	31 adults & 24 children	4 or more	20-s	-	-	Linear/ 2 nd order
Seck et al. (1995)	Isoinertial	7	Healthy males	4	7-s	5-min	-	Linear/ 2 nd order
Yeo et al. (2015)	Isoinertial	24	Competitive cyclists	3	5-s	6-min	15	2 nd order/ 3 rd order

n represents the number of participants in the study

2.5 Improving NMF using ballistic exercises

2.5.1 Training interventions

As highlighted earlier in this review, the ability to produce a high level of power is fundamental for a good performance across many sports, particularly in exercises such as maximal cycling and as such, the improvement in lower limb neuromuscular power is a major focus in many training programs (Cormie et al., 2011; Cronin & Sleivert, 2005). The load/resistance, the velocity at which this resistance is moved and the pattern of the movement performed all influence the enhancement of maximal power and need to be taken into consideration when designing a training program. Common exercises used to improve power production of the lower limbs include traditional resistance training exercises such as squats, lunges and leg press; plyometrics such as bounding and hopping; and ballistic exercises such as jump squat (Cormie et al., 2007; McBride et al., 2002).

Ballistic exercises are explosive movements whereby the limbs are rapidly accelerated against resistance. This type of training requires the CNS to coordinate the limbs to produce a large amount of force over the shortest time possible. Unlike traditional resistance training exercises, during ballistic movements like sprint cycling, the limbs accelerate throughout their range of motion providing a longer time to produce more force and power and for maximal muscle activation (Cormie et al., 2007; Cormie et al., 2011). Exercises which are ballistic in nature are commonly recommended in favour of more traditional resistance training exercises when improvements in power are sought, due to their specificity to many sports, allowing better transfer of adaptations to performance (Cady et al., 1989; Cronin et al., 2001; Kraemer & Newton, 2000; Kyröläinen et al., 2005; Newton et al., 1996). For example, volleyball players showed greater improvements (~6%) in vertical jump performance (e.g. jump height) following 8 weeks of ballistic jump squat training compared to traditional resistance training exercises of leg press and squat (Newton et al., 1999). Although not viewed as a traditional form of ballistic exercise or training, sprint cycling training has the potential to induce neural adaptations that could lead to improvements in NMF. Surprisingly, there are few studies which have implemented training programs to improve power in sprint cycling. Creer et al. (2004) found that four weeks of bi-weekly sprint cycle training, totalling only 28 minutes over the entire training period, lead to improvements in peak power and mean power output of approximately 6% each. The participants in this study were well trained cyclists, habituated to the cycling exercise for at least two years. Similarly, Linossier et al. (1993) found an increase of $2.8 \text{ W}\cdot\text{kg}^{-1}$ following sprint training, however these efforts were much shorter in duration (5-s) compared to those employed by Creer and colleagues which were 30-s in duration, while the training program ran for eight weeks

instead of four. Neither of these sprint cycling interventions accounted for cadence in their assessment of the efficacy of training on power production.

It has been shown that the transfer of training effects between exercises, performed at different speeds or against different resistances may be limited (Baker et al., 1994). The mode of exercise selected (task-specificity), the load or resistance (load-specificity) and velocity (velocity-specificity) at which the exercise is performed during training all appear to influence improvements in maximal power production observed for a given task or movement (Cormie et al., 2011). Just as specificity of the task performed in training influences the gains in power output observed for the given task, so does the level of resistance the exercise is performed against. Therefore, training at a given resistance would influence how F-V (i.e. T-C in cycling) and P-V (i.e. P-C in cycling) relationships are affected. In fact, it has been previously shown by Kaneko and colleagues (1983) that elbow flexor training against different resistances (0%, 30%, 60% and 100% of maximal isometric force) elicited specific changes in F-V and P-V relationships in previously un-trained males. Those who trained at 100% of maximal isometric force showed greatest improvements in force/power at high-force, low-velocity regions of the relationships, while those training at 0% of maximal isometric force improved their ability to produce force and power at the low force, high-velocity regions. Consideration should be given to the fact that only a single-joint was trained in this study, and due to the greater complexity of multi-joint movements it is unknown if the full training effect would be seen in exercises such as maximal cycling.

Velocity-specific responses to isokinetic training have been previously observed with low-velocity training typically leading to improvements in force and power predominantly at lower velocities while high-velocity training leading to improvements at high velocities (Caiozzo et al., 1981; Coyle et al., 1981; Lesmes et al., 1978). Following isoinertial training of single joint movements improvements in power and force were greatest at the velocities used in training (Kaneko et al., 1983). These observed responses of velocity-specific training have been shown to extend to dynamic multi-joint movements. Subjects who trained in jump squatting at high resistances (80% 1RM) improved their performances at low and moderate velocities with no change seen at higher velocities, while those participants who trained against low resistances (30% 1RM) had vast improvements in power at high, moderate and low velocities (McBride et al., 2002). While cadence-specific cycle training improved peak power for those training at low cadences (60-70 rpm) compared to those training at high cadences (110-120 rpm) as evidenced by a 4% mean high-low difference in peak power, with the low cadence group improving more than the high (Paton et al., 2009). However, it should be noted that the training performed was at submaximal intensities. In contrast to these findings, one study showed that regardless of the velocity at which participants trained, increases in maximal force output occurred at both low and

high velocities (Doherty & Campagna, 1993), a second that showed training at low velocities improved performance over a range of velocities (Caiozzo et al., 1981) and a third study contradicting the second which saw high velocity training improve performance at both high and low velocities (Coyle et al., 1981). Mohamad et al. (2012) indicated that 12 weeks of high-velocity (low-resistance) squat training may be equal, if not better than low-velocity (high-resistance) training when equated for training volume (i.e. average power, total work, time that muscle is under tension). Also, it has been suggested that the intended rather than the actual speed of the movement performed could be attributable to velocity-specific adaptations, with those studies showing high and low velocity improvements giving their participants specific instructions to perform the movement as fast as possible (Behm & Sale, 1993; Petersen et al., 1989).

The magnitude of potential power adaptations following training is highly influenced by each individual's specific neuromuscular characteristics. Therefore improvements in maximal power following a bout of training will differ depending on an individual's ability to produce force and power at low and high velocities, rate of force development, muscle coordination and skill in the task/movement/exercise being performed (Cormie et al., 2011). Those individuals who are already well trained in some of these characteristics have less potential to improve, whereas those who are untrained have greater potential for maximal power development (Adams et al., 1992; Wilson et al., 1997; Wilson et al., 1993). For example, Wilson et al. (1997) found a negative correlation between the load lifted during a pre-training one repetition maximum squat exercise (i.e. strength) and the improvement in jump height and 200-m sprint following 8 weeks of heavy strength training. An indicator that stronger individuals (i.e. those who could squat a load >1.8 times their body mass) at baseline did not improve performance outcomes to the same extent as those individuals considered to be weaker (i.e. those who could squat <1.80 times their body mass).

2.5.2 Neural and morphological adaptations

It is well recognised that neural mechanisms contribute substantially to increases in NMF (particularly strength and power) in the absence of hypertrophy, at the beginning of a training program, with the time course for neural adaptations shown to occur as little as three weeks into a high-intensity strength-training program as illustrated in Figure 2.9 (Hakkinen et al., 1985; Kyröläinen et al., 2005; Moritani & DeVries, 1979). Although, the complexity of the movement being performed affects the time course for neural adaptations, with more complex tasks requiring additional time for neuromuscular adaptations to occur (Chilibeck et al., 1998).

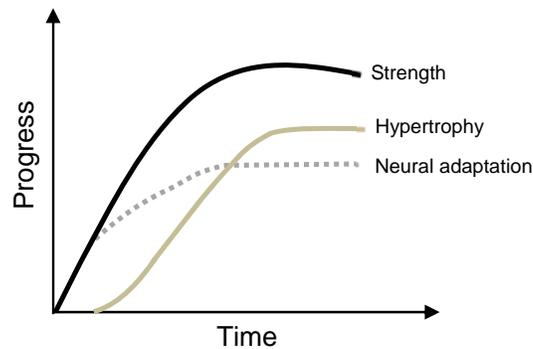


Figure 2.9. Time course for neural and hypertrophy adaptations leading to strength improvements following resistance training. Strength gains early in training are attributable to neural adaptations, while muscle hypertrophy contributes later. Adapted from Moritani and DeVries (1979).

Substantial evidence supports the role of neural factors in neuromuscular adaptations to exercise training, however the specific mechanisms responsible for these adaptations are less conclusive (Carroll et al., 2001b; Sale, 1988). Improved capacity to recruit motor-units (i.e. motor-unit recruitment) and simultaneously contract motor-units or with minimal delay (i.e. motor-unit synchronisation), motor-neuron excitability and the specificity and pattern of neural drive have all been cited as potential neural adaptations accompanying changes in strength and power (Enoka, 1997). In a general sense, increases in strength occurring within only a few weeks of training have been attributable to an improved ability to activate and coordinate muscles (Rutherford & Jones, 1986). Indeed, Rutherford (1988) suggests that improved coordination of the muscle groups used in training rather than alterations in the intrinsic strength of the individual muscles improves the performance of a movement task. Almasbakk and Hoff (1996) attributed early velocity-specific strength improvements following bench press training to more efficient coordination and activation patterns, although muscle activation (i.e. EMG) was not directly assessed. A more recent study showed that 12 weeks of high-resistance power training improved voluntary muscle activation in the knee extensor muscles (~6%) of older adults with mobility impairments that was linked to an improvement in muscle strength and gait speed (Hvid et al., 2016). Another facet of inter-muscular coordination, the simultaneous activation of agonists with their antagonist pairs (i.e. co-activation) is said to be reduced following a period of training to enable agonists to reach a higher level of activation and thus produce more net joint power (Basmajian & De Luca, 1985). Though as observed in trained sprint runners a greater level of co-activation between the knee extensor and flexor muscles has been indicated as beneficial for the performance of rapid movements (Osternig et al., 1986). Further, Carroll and colleagues (2001a) found that training the index finger extensor muscles at increasing frequencies resulted in reduced

variability in patterns of muscle activation. These authors stated that this finding was suggestive of a change within the CNS controlling the activation and coordination of the movement.

The inclusion of ballistic-type exercises in training programs offer the opportunity to maximally activate muscles over a larger part of the movement, facilitating greater neural adaptations (Cormie et al., 2011). The neural adaptations associated with improved power output following ballistic training against high resistances are suggested to include an increased rate and level of neural activation and improved inter-muscular coordination (Hakkinen et al., 1985; McBride et al., 2002). In particular, the improvement of maximal neural drive has been shown to be heightened in individuals who have not been previously exposed to strength training (Aagaard et al., 2002; Cormie et al., 2010). The improvements in maximal power output noted above in the study by Creer et al. (2004) four weeks of high-intensity sprint training were attributable to neural adaptations, in particular an increase in vastus lateralis muscle fiber recruitment as evidence by elevated RMS values. However, these neural adaptations were not thoroughly investigated in this study with only the quadriceps muscles assessed. Further, the EMG signals were not normalised to a reference value (as per the recommendations outlined in section 2.3.1.1), which clouds the comparisons that can be made between EMG results obtained from the same subject on different days.

Muscle hypertrophy (e.g. increase in the number and size of muscle fibers) tends to occur several weeks into a strength training program, following on from neural adaptations. Surface EMG makes it possible to assess the neural contribution following a training program, especially as adaptations responsible for training induced improvements in NMF are generally believed to occur within the nervous system and/or trained muscle (Coyle et al., 1981). In addition to EMG, anthropometry provides a straight forward assessment of volume, adipose and fat-free components of the lower limbs making it an ideal measure for assessing hypertrophic changes following training. Using limited equipment, girth and skinfold measurements obtained from the lower limbs have been used to estimate total and lean leg volume using derived and validated by previous researchers (Jones & Pearson, 1969; Knapik et al., 1996). The advancement of more sophisticated technology, has led to the assessment of body composition using dual-energy x-ray absorptiometry, whereby x-ray beams with different energy levels pass through the tissues distinguishing lean mass from fat mass (Ellis, 2000). Although considered to be a 'gold standard' method of body composition measurement, dual-energy x-ray absorptiometry scanners are expensive and require trained and certified personnel to conduct the tests.

Upon review of the current literature, it appears that knowledge regarding the efficacy of training programs focused on improving power production using maximal cycling is scarce. As such, the findings are inconclusive regarding the potential offered by maximal exercise on a stationary cycle ergometer to improve NMF (e.g. modification of T-C and P-C relationships).

Further, the studies that have been conducted, have not illustrated how sprint cycling interventions can be used to improve the level of torque and power that can be produced against high resistances (i.e. low cadences) and at high velocities (i.e. high cadences). Nor have studies thoroughly investigated the effect of maximal cycling interventions on the physiological, biomechanical and motor control factors outlined in section 2.3 known to affect the limits of NMF on a stationary cycle ergometer.

2.6 Role of ankle joint on lower limb NMF

2.6.1 Functional role of the ankle muscles during ballistic exercise

Simulation studies have alluded to the specific role of the ankle in ballistic exercises such as jumping, running and cycling, though due to the difficulties with the assessment of individual muscles *in vivo*, few studies have explored this in humans. Mechanical models of the vertical jump have illustrated that the inclusion of GAS as a bi-articular muscle maximised jump height in comparison to a model for which GAS was modelled using a mono-articular muscle (Pandy & Zajac, 1991; van Soest et al., 1993). Further, power produced at the ankle during a maximal effort vertical jump was considerably higher than the level of power generated during isolated ankle plantar-flexion (van Ingen Schenau et al., 1985). Although, with regards to the interpretation of these findings, the moment arms of the knee and ankle need to be considered. During slow- and medium-paced running (i.e. up to 7 ms⁻¹) the power output of the ankle plantar-flexor muscles have been shown to play a considerable role in increasing stride length (and thus running speed) via higher support forces generated during contact with the ground (Dorn et al., 2012). Combined these results enhance our understanding that bi-articular muscles (e.g. GAS, HAM and RF) play a role in transferring mechanical energy during jumping, running and cycling (Bobbert & Van Ingen Schenau, 1988; Gregoire et al., 1984; Prilutsky & Zatsiorsky, 1994; van Ingen Schenau, 1989).

Following on from the work of Raasch and colleagues (1997) assessing the contribution of the lower limb muscles in maximum speed pedalling, using a simulation of submaximal cycling at a cadence of 60 rpm, Zajac (2002) found that GMAX and VAS were able to produce the most energy of all the lower limb muscles, but these muscles were unable to directly deliver their full energy contribution to the crank (i.e. they deliver less energy to the crank than they produce). Conversely, the muscles surrounding the ankle joint (e.g. GAS, SOL and TA) were able to deliver more energy to the crank than they produced, transferring ~56% of the energy produced by proximal GMAX and VAS to the crank at the end of extension and during the transition from extension to flexion, as shown in Figure 2.10. Like noted in other ballistic movements (e.g. jumping and running), it has been suggested that the ankle plantar-flexor muscles work co-

actively with the proximal hip and knee extensor muscles to enable effective force transfer to the pedal (Kautz & Neptune, 2002; Van Ingen Schenau et al., 1995). However, there may be a limit to the amount of co-activation within a given muscle pair, with Dorel and colleagues (2012) suggesting that the amount of power generated by the hip extensors may be limited by the ankle plantar flexors ability to effectively transfer the mechanical energy from powerful GMAX to the pedal.

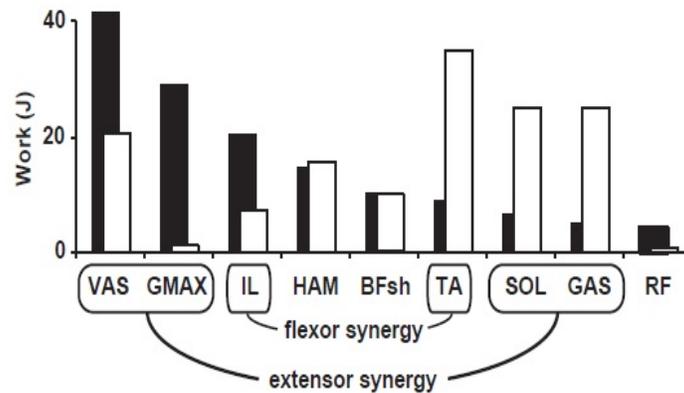


Figure 2.10. Work output of muscles during simulated submaximal cycling at 60 rpm. Filled bars represent the amount of work produced by each muscle, while unfilled bars represent the energy delivered directly to the crank. VAS (*vastii*), GMAX (*gluteus maximus*), IL (*ilipsoas*), HAM (*semimembranosus*), BFsh (*biceps femoris* short head), TA (*tibialis anterior*), SOL (*soleus*), GAS (*gastrocnemii*), RF (*rectus femoris*). Taken from Zajac (2002).

Unlike the hip and knee, ankle joint kinematics appear to be much more amenable to change, with a reduction of ~58% in ankle range of motion observed with a 120 rpm increase in cadence (McDaniel et al., 2014) and a 10° reduction following a 30-s fatiguing exercise bout (Martin & Brown, 2009). Similarly, stiffening of the ankle joint via a 13° reduction in range of motion - stemming from less plantar-flexion - and a concomitant 132% increase in TA activity has been observed after learning to single leg cycle (Hasson et al., 2008). The authors of these studies suggested that the change in range of motion and muscle activation observed at the ankle joint may represent a motor control strategy employed by the CNS to a) stiffen the ankle joint to improve force transfer from proximal muscles and/or b) to simplify the pedalling movement, perhaps as a means to restrict the degrees of freedom afforded by the task, reducing the complexity of the cycling exercise. Although, these findings from single-leg cycling should be approached with caution as this task is different to two-legged cycling requiring a larger contribution of the muscles during the upstroke portion to counteract for no contribution from contra-lateral leg. Further, it has been suggested that a stiffer musculotendinous unit may enhance the work

performed during ballistic hopping movements (Belli & Bosco, 1992). As such, the finding of McDaniel et al. (2014) - the contribution of the ankle to external power diminishes as cadence increases - may highlight the importance of a stiffer ankle during maximal cycling exercise.

2.6.2 Effect of ankle taping on the ankle joint and power production

Prophylactic interventions such as taping and bracing have been implemented in many sports to prevent the high incidence rate of ankle injuries (Garrick & Requa, 1988; Pedowitz et al., 2008). Indeed, injury to the ankle joint is the most common injury reported in sports (Ekstrand & Tropp, 1990; Garrick & Requa, 1988), typically for those ballistic in nature such as basketball (Smith & Reischl, 1986), netball (Hopper et al., 1995) and volleyball (Beneka et al., 2009). It is thought that ankle taping reduces the risk of injury primarily by providing greater structural support and/or mechanical stiffness (Alt et al., 1999; Zinder et al., 2009) but also by enhancing proprioceptive and neuromuscular control (Cordova et al., 2002; Glick et al., 1976; Heit et al., 1996; Wilkerson, 2002). Although, the exact mechanisms regarding enhanced proprioceptive and neuromuscular control are still relatively equivocal.

Taping techniques commonly used by clinicians and sport scientists to improve structural support and/or mechanical stiffness (e.g. open and closed basket weave, combinations of stirrups and heel locks) all restrict ankle joint range of motion (to a certain extent) (Fumich et al., 1981; Purcell et al., 2009). A meta-analysis of 19 studies investigating the effect of different forms of ankle support on range of motion found that the application of rigid adhesive tape on average restricted plantar-flexion by 10.5° (a large standardised effect, based upon Cohen (1988)) and restricted dorsi-flexion by 6.6° (a medium standardised effect) prior to performing exercise (Cordova et al., 2000). Following an exercise bout, plantar-flexion remained reduced by 7.6° (a medium standardised effect) and dorsi-flexion by 6.0° (a small standardised effect), indicating the integrity of the tape was still well preserved.

Based upon the findings in the section above, altering the kinematics of a movement is likely to affect the amount of external force and power that can be produced. Although ankle taping may be beneficial in reducing the risk of injury, the restriction imposed on the joint may impact performance. The effect of ankle taping on performance capabilities have been well investigated, but among these studies the findings have been inconsistent. Ankle taping has been shown to decrease sprint running and vertical jump performance in college level athletes on average by 4% and 3.5% respectively, although as the standard deviations associated with these decreases were not reported, the variation in response to ankle taping cannot be interpreted (Burks et al., 1991). Other studies have shown trivial effects of ankle taping on vertical jump and 40-yard sprint performance (Greene & Hillman, 1990; Verbrugge, 1996). More recently, ankle taping

improved peak isometric plantar-flexion strength by approximately 20% compared to a non-taped control ankle in previously uninjured females (Hopper et al., 2014). As per the relationship between force and velocity, high levels of force are produced at low velocities during a concentric muscle contraction (or movement) with the greatest amount of force produced at zero velocity (i.e. isometric). With this fundamental concept in mind, the improvement in plantar-flexion strength with ankle taping, observed by Hopper and colleagues (2014) may have implications for improving the amount of torque and power that could be produced on a stationary cycle ergometer at low velocities. Further, based upon the findings of Belli and Bosco (1992) ankle taping may help to stiffen the musculotendinous unit enhancing the work performed during ballistic movements. However, as shown by McDaniel et al. (McDaniel et al., 2014) ankle joint power and the range of cadences over which the ankle operates is greatest at low cadences, so perhaps ankle taping may not have such a favourable result improvement in power.

2.7 Summary

An adequate level of NMF is necessary in humans not only for the execution of a good sporting performance but also for the successful execution of everyday tasks. Stationary cycle ergometry offers a safe and effective means by which to explore the limits of lower limb NMF, in particular the capability to produce power and torque. As such, there exist are a plethora of studies using force-velocity tests on various types of stationary cycle ergometer to investigate the power producing capabilities of the lower limbs, although the current procedures for predicting T-C and P-C relationships may overestimate torque and power for a given cadence and thereby inadequately assess NMF of the lower limbs. The level of power produced during maximal cycling exercise can be affected by numerous physiological, biomechanical and motor control factors, making it complex to understand. Using techniques to evaluate muscle activation, joint kinematics and kinetic, the contribution of each of these can be assessed during dynamic, multi-joint movements. Each of the lower limb muscles provide a unique functional role during pedalling, but it is their well-timed co-activation within a pedal cycle which enables force to be transferred to the pedal at appropriate sections. Although, due to the abundant degrees of freedom afforded by the complexity of the pedalling movement the inherent variability observed in the neuro-musculo-skeletal systems, cannot be overlooked. As such the CNS appears to employ motor control strategies to simplify the movement. When faced with changing task constraints such as increasing exercise intensity and/or cadence, it appears that the kinematics and activation of muscles surrounding the distal ankle joint are altered more than proximal hip and knee in an attempt to maintain power output. Simulation studies have alluded to the substantial role that the ankle plays in maximal cycling, though few studies have explored this in humans. Taping

procedures provide greater structural support for the ankle joint, but the effect of taping during the performance of ballistic movements is equivocal. Therefore the effect of ankle taping on the limits of lower limb NMF on a stationary cycle ergometer is unknown and may provide further information regarding the role of the ankle joint during maximal cycling exercise. Also, power production can be improved from as little as two days to three weeks using sprint cycling interventions with the improvements attributed to motor learning and neural adaptations in the absence of morphological adaptations. However the adaptations suggested for the power improvements were not thoroughly investigated. Further, high-resistance and high-velocity training have been shown to improve power production, hence could be useful for improving power across a wide range of cadences during stationary cycle ergometry.

2.8 Study Aims

2.8.1 Study One (Chapter 3)

Aims:

- 1) To measure variations in torque and EMG profiles between maximal and non-maximal pedal cycles during a force-velocity test performed on a stationary cycle ergometer.
- 2) To compare the ability of two modelling procedures previously used in the literature to predict the shapes of T-C (i.e. linear regressions vs second order polynomial) and P-C (i.e. second order polynomials vs third order polynomials) relationships and quantify the limits of NMF.

Assumptions:

- 1) The selection of pedal cycles based on a higher level of torque, for a given cadence would be associated with higher EMG amplitude and less variable EMG profiles of the lower limb muscles.
- 2) The shapes of individual T-C and P-C relationships and associated limits of NMF (i.e. maximal power, optimal cadence, maximal torque and maximal cadence) would be better predicted using second and third order polynomials respectively.

2.8.2 Study Two (Chapter 4)

Aims:

- 1) To investigate if the adaptations of the limits of NMF would be specific to the two different training interventions (i.e. against high resistances and at high cadences).

- 2) To investigate if different motor control adaptations would accompany the changes in the limits of NMF for the two different training interventions.

Assumptions:

- 1) Training against high resistances would alter the limits of NMF on the left side of the P-C relationship (i.e. increase T_0 and the power generating capacity at low to moderate cadences), while training at high cadence would alter the limits of NMF on the right side of the P-C relationship (i.e. increase the power generating capacity at moderate to high cadences and C_0).
- 2) Changes in the limits of the NMF would be due to neural adaptations, while the variability of crank torque, kinematic and EMG profiles would be reduced after training for the same cycling condition. Further, modifications to torque applied to the crank, inter-joint and inter-muscular coordination after training may also explain the potential change in the limits of NMF.

2.8.3 Study Three (Chapter 5)

Aims:

- 1) To investigate the effect of ankle taping on the limits of NMF on a stationary cycle ergometer.
- 2) To assess how ankle taping affects crank torque application, lower limb kinematics, inter-muscular coordination and movement variability.

Assumptions:

- 1) Due to the role of the ankle in maximal cycling the limits of lower limb NMF on a stationary cycle ergometer would be affected, in particular those on the left side of the P-C relationship.
- 2) It was assumed that taping would affect the kinematics of the ankle joint, leading to compensatory changes in the kinematics of the proximal joints (hip and knee). Also, it was assumed that the neural drive to the ankle and proximal muscles could be affected, potentially affecting inter-muscular coordination through changes in the co-activation between muscle pairs. Additionally, an increase in inter-cycle and inter-participant movement variability was assumed due to the novelty of the task performed.

Chapter 3 Assessing the Limits of Neuromuscular Function on a Stationary Cycle Ergometer

3.1 Introduction

The effective execution of physical tasks requires adequate NMF. Tasks can range from those performed as part of daily life to those required in the sporting arena. Stationary cycle ergometry is an ecologically valid, safe, multi-joint movement that can be performed by a wide range of populations, making it a useful tool for the assessment of the lower limb muscles. Force-velocity (F-V) tests performed on a stationary cycle ergometer are commonly used to measure the maximal levels of torque and power that an individual can produce at a given cadence (i.e. torque vs cadence (T-C) and power vs cadence (P-C) relationships), enabling assessment of their functional capacity (Arsac et al., 1996; Dorel et al., 2005; Samozino et al., 2007; Yeo et al., 2015). Variables commonly estimated from the F-V test including maximal power (P_{max}), optimal cadence (C_{opt}), maximal torque (T_0) and maximal cadence (C_0) allow the limits of NMF to be estimated (Dorel et al., 2005; Driss et al., 2002; Gardner et al., 2007; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2012). However, the type of ergometer used (i.e. isoinertial vs isokinetic), F-V test protocol employed (i.e. single vs multiple sprints) and approach for modelling the experimental values of torque and power varies in the literature, with no previous studies investigating the best method for the assessment of the limits NMF (Arsac et al., 1996; Dorel et al., 2005; Driss et al., 2002; Gardner et al., 2007; Hautier et al., 1996; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007; Sargeant et al., 1981; Yeo et al., 2015).

Unlike the assessment of force and velocity in single muscle fibers, the level of force and power produced on a stationary cycle ergometer can be affected by many factors other than cadence. The pedalling movement is complex, requiring the coordination of a large number of muscles for the effective transmission of force/power to the crank (Raasch et al., 1997; Zajac, 2002). However, as shown previously, regardless of an individual's experience with cycling, they may be unable to produce maximal levels of force/power during every pedal cycle within a test protocol (Arsac et al., 1996; Samozino et al., 2007). Indeed, humans may not maximally and optimally activate and coordinate their lower limb muscles every time a movement is executed (i.e. every revolution of a pedal cycle) (Allen et al., 1995). Based upon previous findings, if the main the muscles used in the pedaling movement are not maximally and optimally activated, power production may suffer (van Ingen Schenau, 1989; Zajac et al., 2002). Further, due to the complexity of the pedaling movement, there is an abundance of solutions offered by the human body (Bernstein, 1967; Latash, 2012) which may cause variability in the recruitment and coordination of the lower limb muscles. If these patterns deviate too far from optimal power

production could be negatively impacted. These deviations may be even greater for those who are novice cyclists, with those unskilled in the pedalling movement exhibiting muscle recruitment strategies that are less refined (Chapman et al., 2006; Hug et al., 2008; Muller & Sternad, 2009). In the current literature, it is unclear if researchers take the points listed above into consideration when designing their methodologies, whereby a method chosen allows the true maximal ability of an individual to produce torque and power over a range of cadences is exhibited. Also, if some form of data filtering or selection takes place it does not get stated. Test protocols that include multiple sprints may allow for the collection of several data points for a given cadence (Arsac et al., 1996; Dorel et al., 2010; Dorel et al., 2005), increasing an individual's chance of producing maximal levels of torque and power, unlike the use of a single maximal effort (Martin et al., 1997).

A review of literature reveals that there is a lack of consensus regarding the shapes of T-C and P-C relationships. As two methods have been used previously to predict the shapes of these relationships, it is unclear which method provides the best fit for experimental data points collected. In the majority of studies, the T-C relationship is fit with a linear regression, relying on the assumption that the relationship is linear, while the P-C relationship appears as a symmetrical parabola and as such is typically fit with a second order polynomial (i.e. a quadratic) (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007). Although, in a few studies (Arsac et al., 1996; Hautier et al., 1996; Yeo et al., 2015), the methods used to fit the experimental data points (i.e. second order polynomials for T-C and third order polynomials for P-C) relied on the notion that the T-C relationship might not be perfectly linear while the P-C relationship might be an asymmetrical parabola. Bobbert (2012) reported that the shape of the force vs velocity relationships obtained from a F-V test during maximal leg press exercise was not perfectly linear. The author observed slight curvatures of the external force vs velocity relationship that could have been attributed to the reduced ability of individuals to control the external force at high speeds. Considering that maximal leg cycling exercise is a multi-joint movement requiring a higher level of external force control (compared to maximal leg press exercise), it seemed necessary to consider the possibility that the shapes of T-C and P-C relationships obtained using F-V tests performed on a stationary cycle ergometer might be more complex than previously assumed in a large number of studies (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007). A variety of physiological mechanisms identified during other forms of maximal intensity exercise could potentially limit the production of torque and power when cycling at low (Babault et al., 2002; Perrine & Edgerton, 1978; Robbins, 2005; Westing et al., 1991; Yamauchi et al., 2007) and high cadences (McDaniel et al., 2014; van Soest & Casius, 2000). Upon consideration of the potential roles played by these various factors on the

ability to produce torque and power, it is of importance to investigate and define the best way to describe T-C and P-C relationships obtained from F-V tests performed on a stationary cycle ergometer.

The first aim of this study was to measure variations in torque and EMG profiles between maximal and non-maximal pedal cycles during a F-V test performed on a stationary cycle ergometer. It was assumed that higher levels of torque and power would be calculated from maximal pedal cycles associated with higher levels of peak EMG and co-activation and less variable EMG profiles of the lower limb muscles. The second aim of this study was to compare the ability of two modelling procedures previously used in the literature to describe the shapes of T-C (i.e. linear regressions vs second order polynomial) and P-C (i.e. second order polynomials vs third order polynomials) relationships and quantify the limits of NMF. It was assumed that higher order polynomials would provide a better fit for the experimental data points, leading to more accurate description of T-C and P-C relationships as well as key variables (i.e. P_{\max} , C_{opt} , C_0 , T_0) calculated to define the limits of NMF.

3.2 Methods

3.2.1 Participants

Seventeen low-to-moderately active, young healthy males volunteered to participate in this study (mean \pm SD: age = 26 ± 4 y, body mass = 82.1 ± 11.2 kg, height = 180.3 ± 7.6 cm). Participants were involved in recreational physical activities such as resistance training and team sports, but did not have any prior training in cycling. The experimental procedures used in this study were approved by Victoria University's Human Research Ethics Committee and carried out in accordance with the Declaration of Helsinki. Subjects gave written informed consent to participate in the study if they accepted the testing procedures explained to them.

3.2.2 Study protocol

Each participant performed two separate tests within one week of each other which were carried out at the same time of day. In the first session participants were familiarised with the testing equipment, warm-up protocol and the testing procedures they would perform during the F-V test in the main testing session. Previously it has been shown that one familiarization session is adequate to obtain reproducible measurements of maximal power in young non-cyclist adults (Doré et al., 2003). Participants were also asked to refrain from consuming caffeinated beverages and food 12 hours prior to each test.

3.2.2.1 Force-velocity test

An electro-magnetically braked cycle ergometer (DynaFit Pro Velotron, RacerMate Inc., Seattle, WA, USA), equipped with 170 mm scientific SRM[®] PowerMeter cranks (Schoberer Rad Messtechnik International, Jülich, Germany) connected to Torxstar[™] data logging system was used to run the F-V test, similar to methods used previously (Barratt, 2008; Yeo et al., 2015). The analog torque signal was recorded by Torxstar[™] via strain gauges positioned within the spider of the SRM powermeter at a frequency of 250 Hz. A static calibration of the SRM cranks while connected to Torxstar[™] was performed prior and after data collection, following procedures previously described (Wooles et al., 2005). Additionally, Torxstar[™] was used to detect left-top-dead centre (LTDC) and right-top-dead centre (RTDC) crank positions and identify the start/end of each pedal cycle (i.e. LTDC to LTDC and RTDC to RTDC) completed during each sprint of the F-V test.

The external resistances used during the F-V test (including the warm-up) were adjusted and controlled using the Velotron Wingate software (v1.0, RacerMate Inc., Seattle, WA, USA).

The cycle ergometer was fit with clipless pedals (Shimano, PD-R540 SPD-SL, Osaka, Japan) and participants were provided cleated cycling shoes (Shimano SH-R064, Osaka, Japan). Saddle height was set at 109% of inseam length (Hamley & Thomas, 1967), while the handlebars were adjusted vertically and horizontally to the requirements of each subject.

At the beginning of both sessions, participants performed a standardised warm-up which included 8-min of cycling at 80 to 90 rpm, and two 7-s sprints at a workload of $1.2 \text{ W}\cdot\text{kg}^{-1}$, controlled by Velotron Coaching software (RacerMate Inc., Seattle, WA, USA). Following 5-min of passive rest, participants performed a F-V test that consisted of six all-out, 6-s sprints interspersed with 5-min rest periods, in accordance with methods previously described (Arsac et al., 1996; Dorel et al., 2005). More specifically, the different sprints completed by each participant were as follows: 1) a sprint from a stationary start against an external resistance of $4 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ using an 85 tooth front sprocket and 14 tooth rear sprocket; 2) a sprint from a stationary start against an external resistance of $1 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ using a 62 tooth front sprocket and 14 tooth rear sprocket; 3) a sprint from a stationary start against an external resistance of $2 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ using an 85 tooth front sprocket and 14 tooth rear sprocket; 4) a sprint from a rolling start with an initial cadence ~ 80 rpm against an external resistance of $0.5 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ using a 62 tooth front sprocket and 14 tooth rear sprocket; 5) a sprint from a rolling start with an initial cadence ~ 100 rpm against an external resistance of $0.3 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ using a 62 tooth front sprocket and 14 tooth rear sprocket; 6) a sprint from a stationary start against no external resistance (the chain was removed) in order to obtain an experimental measure of the participants maximal cadence (C_{max}). All sprints were performed on the same cycle ergometer, with the front sprocket changed from the 85 tooth to the 62 tooth and vice versa, as required during the five minute rest period given between sprints. The external resistances listed for the different sprints above correspond to the torques exerted on the flywheel of the cycle ergometer. The order of the sprints was randomized for each subject. Rolling starts were implemented for sprints performed against low external resistance in order to enable participants to reach high cadences within the 6-s sprint duration. To achieve the rolling starts, the flywheel was accelerated by the experimenter immediately prior to the sprint so that participants could initiate their sprints at the target cadence without prior effort. Participants were instructed to remain seated on the saddle, keep hands on the dropped portion of the handlebars and to produce the highest acceleration possible throughout the sprint. Participants were vigorously encouraged throughout the duration of each sprint.

Surface electromyography (EMG) signals were bilaterally recorded from seven muscles of the lower limbs: *gluteus maximus* (GMAX); *rectus femoris* (RF); *vastus lateralis* (VAS); *semitendinosus and biceps femoris* (HAM); *gastrocnemius medialis* (GAS); *tibialis anterior* (TA). These muscles were selected as they are considered to be the main lower limb muscles used in the pedalling movement (Raasch et al., 1997; Zajac et al., 2002). Disposable pre-gelled Ag-

AgCl surface electrodes (Blue sensor N, Ambu, Ballerup, Denmark) were used to record the EMG signals. Electrodes were positioned at an inter-electrode distance of 20 mm apart (centre to centre), aligned parallel to the muscle fibres in accordance with the recommendations of SENIAM (Hermens et al., 2000). Prior to placement of the electrodes, the skin was prepared by shaving, light abrasion and cleaned with alcohol swabs. Electrodes and wireless sensors were secured with adhesive tape to ensure good contact with the skin and to reduce movement artefact. EMG signals were recorded continuously and sent in real-time to a wireless receiver (Telemyo DTS wireless Noraxon Inc., AZ, USA) connected to a PC running MyoResearch software (Noraxon Inc., AZ, USA) at a sampling rate of 1500 Hz. Closure of a reed switch generated a 3-volt pulse in an auxiliary analogue channel of the EMG system which synchronised crank position (i.e. LTDC) with the raw EMG signals.

3.2.2.2 Data processing

All mechanical and EMG signals were later analysed using Visual3D software (version 5, C-Motion, Germantown, MD, USA). First crank torque signals were low-pass filtered (10 Hz, 4th order Butterworth filter). Then, using the time synchronised events of LTDC and RTDC, average cadence was derived from time duration of the pedal cycle (i.e. LTDC-LTDC for left leg and RTDC-RTDC for right leg). Average crank torque values were calculated over the same time interval, while average power was computed using Eq. 1 below (Martin et al., 1997):

$$Power = Torque \times Cadence \times \frac{\pi}{30}$$

Eq. 1

Raw EMG signals were processed using the following steps: i) removal of low-frequency artefact by using a 20 Hz high-pass Butterworth filter, ii) rectified using a root mean squared (RMS) with a 25-ms moving rectangular window and iii) smoothed using a low-pass Butterworth filter with a 10 Hz cut-off. The amplitude of the RMS of each muscle was normalised according to the methods previously defined by Rouffet and Hautier (2008).

3.2.3 Maximal vs non-maximal pedal cycles

3.2.3.1 Identification of maximal and non-maximal pedal cycles recorded during the force-velocity test

In order to assess the effect of data point selection on the shape of the T-C relationship, average cadence, and average torque values from all pedal cycles from the five sprints (against external resistance) of the F-V test were used to create individual T-C relationships. From all the data points/pedal cycles collected: 1) the highest values of torque per every 5 rpm cadence interval were selected and used to characterize a set of maximal cycle T-C relationships for each participant and; 2) the lowest values of torque per every 5 rpm cadence interval were selected and used to characterize a second set of non-maximal cycle T-C relationships for each participant. A linear regression was then fit to each individual's maximal pedal cycle and non-maximal pedal cycle T-C relationships and the equation of the lines used to predict average torque values at cadences of 60 rpm, 115 rpm and 170 rpm.

Total crank torque profiles (i.e. the sum of the force applied to the left and right cranks) were created for each participant between LTDC-LTDC and RTDC-RTDC and time normalized to 100 points (i.e. 100%) for each pedal cycle. Peak crank torque was then identified for cycles corresponding to maximal pedal cycles and non-maximal pedal cycles, as defined above for average torque. Maximal cycle peak crank torque vs cadence and non-maximal pedal cycle peak crank torque vs cadence relationships were created for each participant and fit with linear regressions. The equations of the regression lines were then used to predict peak crank torque at cadences of 60 rpm, 115 rpm and 170 rpm.

3.2.3.2 EMG activity of the lower limb muscles during maximal and non-maximal pedal cycles

Peak EMG was identified for cycles corresponding to maximal pedal cycles and non-maximal pedal cycles and used to create two peak EMG vs cadence relationships for each participant and each muscle. Individual relationships were fit with linear regressions and the equations used to predict peak EMG at the same cadences for which average torque and peak crank torque were predicted- 60 rpm, 115 rpm and 170 rpm.

Similar to crank torque profiles, EMG profiles were created for each muscle between LTDC-LTDC for left leg and RTDC-RTDC for right leg and time normalized to 100 points (100%) for each pedal cycle. Differences in the average EMG profiles observed between maximal and non-maximal cycles were investigated for each muscle.

3.2.3.3 Co-activation of the lower limb muscles during maximal and non-maximal pedal cycles

Based upon the biomechanical models of cycling (van Ingen Schenau, 1989; Zajac et al., 2002), co-activation values were calculated from the normalised EMG profiles for VAS-GAS, GMAX-VAS, VAS-HAM and GMAX-RF muscle pairs using the Co-Activation Index (CAI) shown in Eq. 2 below (Lewek et al., 2004). Average CAI profiles were created for non-maximal and maximal cycles for each muscle pair. Average CAI values were then calculated for each muscle pair and each condition.

$$CAI = \frac{1}{100} \sum_{i=1}^{100} \left[\frac{\text{lower EMGi}}{\text{higher EMGi}} \times (\text{lower EMGi} + \text{higher EMGi}) \right]$$

Eq. 2

3.2.3.4 Variability of crank torque, EMG and co-activation profiles during maximal and non-maximal pedal cycles

An index of inter-cycle (intra-individual) variability was calculated for crank torque, EMG and CAI profiles obtained for maximal and non-maximal pedal cycles using variance ratios (VR). VR values were calculated for each participant and each variable separately to quantify the variability of the profiles between-cycles using Eq. 3 below.

$$VR = \frac{\sum_{i=1}^k \sum_{j=1}^n (X_{ij} - \bar{X}_i)^2 / k(n-1)}{\sum_{i=1}^k \sum_{j=1}^n (X_{ij} - \bar{X})^2 / (kn-1)}$$

$$\bar{X} = \frac{1}{k} \sum_{i=1}^k \bar{X}_i$$

Eq. 3

where k is the number of intervals over the pedal cycle (i.e. 101), n is the number of pedal cycles (i.e. 11), X_{ij} is the mean EMG value or crank torque value at the i th interval for the j th pedal cycle and \bar{X}_i is the mean of the EMG values or crank torque values at the i th interval calculated over the 11 pedal cycles (Burden et al., 2003; Rouffet & Hautier, 2008).

3.2.4 Prediction of lower limb NMF during maximal cycling exercise

3.2.4.1 Prediction of individual T-C relationships and derived variables (T_0)

Individual maximal cycle T-C relationships were fit with 2nd order polynomial regressions in reference to methods previously described (Arsac et al., 1996; Hautier et al., 1996; Yeo et al., 2015), and also with linear regressions as per the methods traditionally used in most studies (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999). Using the equations of the 2nd order polynomials and linear regressions torque was predicted at 10 rpm intervals ranging from 40 to 200 rpm. Values of the intercept of the T-C relationship with the y-axis (theoretical maximal torque: T_0) using the equations of the 2nd order polynomials and linear regressions were calculated and compared.

3.2.4.2 Prediction of individual P-C relationships and derived variables (P_{max} , C_{opt} and C_0)

As per the filtering methods performed with the torque data, the highest values of power (one for every 5 rpm cadence interval) were selected from all pedal cycles collected during the F-V test and used to characterize a set of maximal cycle P-C relationships for each participant. Individual maximal cycle P-C relationships were then fit with 3rd order polynomial regressions with a fixed y-intercept set at zero in reference to methods previously described (Arsac et al., 1996; Hautier et al., 1996; Yeo et al., 2015), and with 2nd order polynomial regressions with a fixed y-intercept set at zero as per the methods most frequently used in studies (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999). Microsoft Excel Solver (version 2010) was used to predict the values of power (maximal power: P_{max}) and cadence (optimal cadence: C_{opt}) at the apex of the P-C relationships using both the equations of 3rd order polynomials and 2nd order polynomials. Values of the intercept of the P-C relationship with the x-axis on the right side of the relationship (theoretical maximal cadence: C_0) using the equations of the 3rd and 2nd order polynomials were calculated and compared. C_0 values obtained using 3rd and 2nd order polynomials were compared with experimentally measured maximal cadence (C_{max}). Then, using the equations of the 3rd and 2nd order polynomials power was predicted at 10 rpm intervals ranging from 40 to 200 rpm. The ratio of C_{opt}/C_0 was also calculated.

The shapes of P-C curves were further assessed by calculating and comparing the levels of power reduction associated to positive (cadence shifting towards higher values) and negative (cadence shifting towards lower values) deviations of cadence in reference to C_{opt} , using 3rd and 2nd order polynomials. These comparisons were made for a series of 5 rpm cadence intervals from -80 rpm to +80 rpm in reference to C_{opt} . To eliminate the effect of variations in C_{opt} predicted

using 3rd and 2nd order polynomials, C_{opt} values calculated from the respective equations were used.

3.2.4.3 Goodness of fit

The goodness of fit provided by low and high order polynomials was compared by calculating and comparing standard error of the estimate (SEE) and r^2 values of the different regressions fit to T-C and P-C relationships (i.e. 2nd order polynomials vs linear regressions for T-C and 3rd order polynomials vs 2nd order polynomials for P-C). Torque and power residuals were also calculated for the different regressions at a low cadence interval of 40-50 rpm, a high cadence interval of 170-180 rpm and a cadence interval of 100-110 rpm, covering the middle portion of the relationship.

3.2.5 Statistical analyses

Comparison of mean outcome variables were performed with a customized spreadsheet using magnitude-based inferences and standardization to interpret the meaningfulness of the effects (Hopkins, 2006b). First, differences in means between the pedal cycles identified as maximal and non-maximal at three different portions of the torque vs cadence relationships (60, 115 and 170 rpm) were analysed for the following variables: average crank torque, peak crank torque, peak EMG, average co-activation index, and variance ratio. Second, differences in means between high and low order polynomial regressions were analysed for the following variables: values of average torque and power predicted every 10 rpm between 40 and 200 rpm as well as the key variables traditionally extracted (T_0 , C_0 , P_{max} and C_{opt}). Third, differences in means between C_0 values predicted from high order polynomials and maximal cadence measured during the sprint performed against no resistance (C_{max}) were analysed. The standardised effect was calculated as the difference in means divided by the standard deviation (SD) of the reference condition and interpreted using thresholds set at <0.2 (trivial), ≥ 0.2 (small), ≥ 0.6 (moderate), ≥ 1.2 (large), ≥ 2.0 (very large), ≥ 4.0 (extremely large) (Cohen, 1988; Hopkins et al., 2009). As illustrated in Figure 3.1, coloured bands were used in the results section to highlight the magnitude of the standardised effect in tables and figures, with small standardised effects highlighted in yellow, moderate in pink, large in green, very large in blue, extremely large in purple. Trivial effects are indicated by no coloured band. Estimates were presented with 90% confidence intervals (\pm CI) or confidence limits (lower CL to upper CL). The likelihood that the standardized effect was substantial was assessed with non-clinical magnitude-based inference, using the following scale for interpreting the likelihoods: $\geq 25\%$, possible; $\geq 75\%$, likely; $\geq 95\%$, very likely and $\geq 99.5\%$, most likely (Hopkins et al., 2009). Symbols used to denote the likelihood of a non-trivial/true standardised

effect are * possibly, ** likely, *** very likely, **** most likely. The likelihood of trivial effects are denoted by ⁰ possibly, ⁰⁰ likely, ⁰⁰⁰ very likely, ⁰⁰⁰⁰ most likely. Unclear effects (trivial or non-trivial) have no symbol. Data are presented as mean \pm standard deviation (SD) unless otherwise stated.

Finally, to assess the goodness of fit for the different models standard error of the estimates (SEE) and r^2 values were used. Each participant's value of SEE was log-transformed, because the sampling distribution of a SD is approximately log-normal. SEE values were compared using the same statistical approach as for difference in means above, but magnitude thresholds for assessing the SDs and for comparisons of SDs were halved for comparing means (Smith & Hopkins, 2011). Thresholds for r^2 and for changes in r^2 were derived by a novel approach also based on standardization. Since $r^2 = \text{variance explained} = \text{SD}^2 / (\text{SD}^2 + \text{SEE}^2)$, substituting threshold values of 0.1, 0.3, 0.6, 1.0 and 2.0 for SEE gives thresholds for interpreting a given r^2 of 0.99, 0.92, 0.74, 0.50 and 0.20 for extremely high, very high, high, moderate and low values respectively (Hopkins, 2015). To evaluate whether a clear improvement or trivial change in r^2 was seen between comparisons, it was assumed that a substantial improvement would be one that increased the r^2 value from one magnitude threshold to the next higher threshold (e.g., a change from 0.74 to 0.92, a change of 0.18). Threshold changes for r^2 values falling between the magnitude thresholds for r^2 were determined by interpolation.

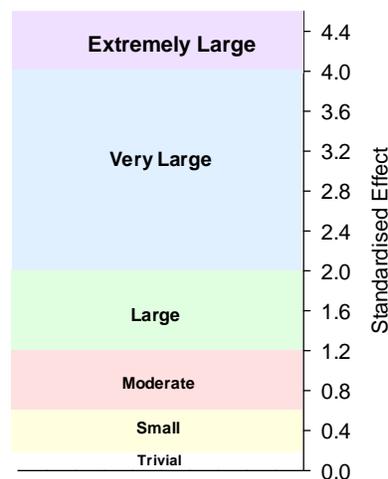


Figure 3.1. Thresholds and associated colour bands used for interpreting the magnitude of the standardised effect throughout the thesis for all variables except SEE and r^2 . Adapted from Cohen (1988) and Hopkins et al. (2009).

3.3 Results

3.3.1 Maximal vs non-maximal pedal cycles

From all the sprints of the F-V test an average of 62 ± 16 data points were collected for each subject, between cadences of 41 ± 7 rpm to 180 ± 10 rpm for sprints against resistance and between 97 ± 23 rpm to 214 ± 20 rpm for the sprint against no resistance. Maximal cycle T-C and P-C relationships were created using 24 ± 3 pedal cycles, while non-maximal cycle T-C and P-C relationships were created using 19 ± 5 pedal cycles, as per Figure 3.2.

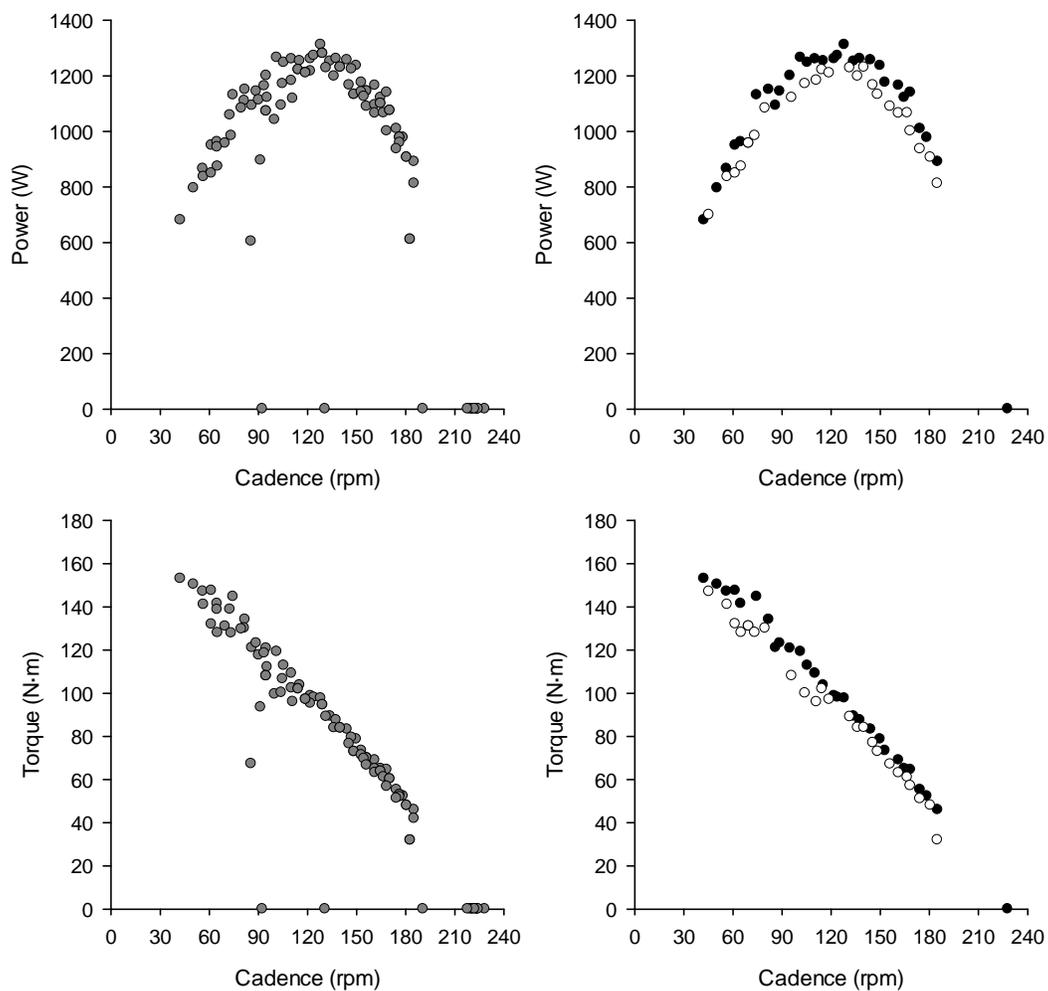


Figure 3.2. Methods used to select maximal and non-maximal cycles for each participant. Grey circles represent torque and power values for every cycle collected from all sprints of the F-V test, while black circles represent the points corresponding to maximal cycles and unfilled circles represent points corresponding to non-maximal cycles.

1.1.1.1 Differences in average torque

At 60 rpm and 115 rpm average torque was likely higher for maximal cycles compared to non-maximal cycles, with values of $132 \pm 25 \text{ N}\cdot\text{m}$ vs $126 \pm 24 \text{ N}\cdot\text{m}$ and $94 \pm 17 \text{ N}\cdot\text{m}$ vs $89 \pm 17 \text{ N}\cdot\text{m}$ respectively. Smaller differences were observed between maximal and non-maximal cycles at the higher cadence of 170 rpm ($56 \pm 12 \text{ N}\cdot\text{m}$ vs $53 \pm 13 \text{ N}\cdot\text{m}$; Figure 3.3).

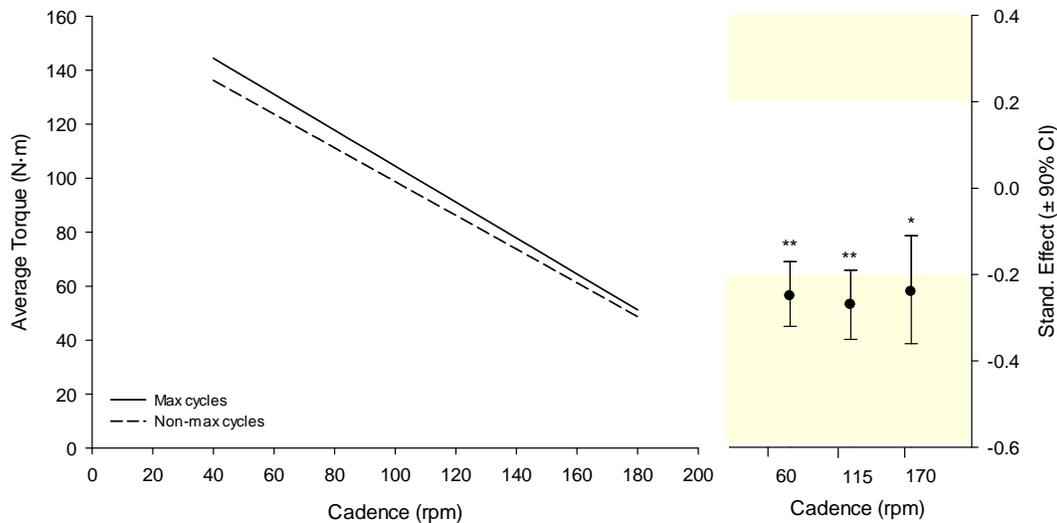


Figure 3.3. Average torque predicted from maximal and non-maximal cycles. Lines represent means with SD lines omitted for clarity. Graph to the right illustrates standardised effect \pm 90% CI of the difference between maximal and non-maximal cycles at 60 rpm, 115 rpm and 170 rpm. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely.

1.1.1.2 Differences in peak crank torque

Higher peak crank torque values were observed for maximal cycles compared to non-maximal cycles at 60 rpm ($205 \pm 44 \text{ N}\cdot\text{m}$ vs $192 \pm 32 \text{ N}\cdot\text{m}$), 115 rpm ($144 \pm 28 \text{ N}\cdot\text{m}$ vs $135 \pm 23 \text{ N}\cdot\text{m}$) and 170 rpm ($82 \pm 18 \text{ N}\cdot\text{m}$ vs $77 \pm 22 \text{ N}\cdot\text{m}$), with the largest differences observed at the lower cadences (Figure 3.4).

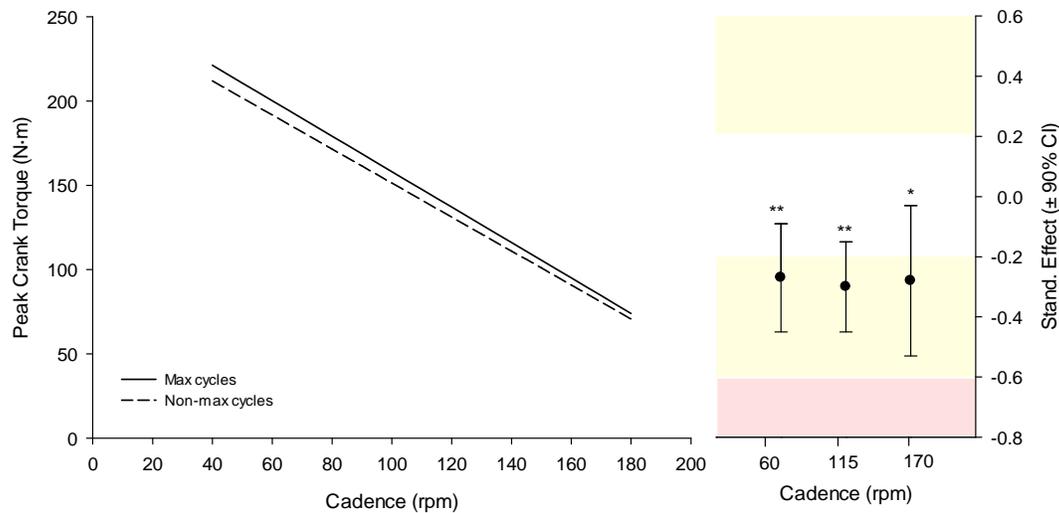


Figure 3.4. Peak crank torque predicted from maximal and non-maximal cycles. Lines represent means with SD lines omitted for clarity. Graph to the right illustrates the standardised effect \pm 90% CI of the difference between maximal and non-maximal cycles at 60 rpm, 115 rpm and 170 rpm. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely.

1.1.1.3 Differences in EMG of the lower limb muscles

Quantification of the difference in peak EMG associated with maximal and non-maximal pedal cycles revealed that the difference in peak EMG between the two conditions was not the same for each muscle or uniform across the range of cadences assessed. A fairly uniform difference in peak EMG between maximal and non-maximal pedal cycles was seen for GAS ($4 \pm 8\%$; $4 \pm 6\%$; $4 \pm 13\%$), TA ($4 \pm 6\%$; $4 \pm 4\%$; $3 \pm 9\%$) and VAS ($2 \pm 6\%$; $2 \pm 4\%$; $2 \pm 8\%$) across the range of cadences assessed (60 to 115 to 170 rpm, respectively), although greater variability was evident at the highest cadence (Figure 3.6). A trivial difference was observed between maximal and non-maximal pedal cycles at 60 rpm ($-1 \pm 8\%$) for RF, while larger differences were seen at 115 rpm ($2 \pm 4\%$) and 170 rpm ($4 \pm 7\%$). The opposite trend was observed for HAM with substantial differences observed at 60 rpm ($4 \pm 7\%$) and 115 rpm ($2 \pm 6\%$) and trivial differences at 170 rpm ($1 \pm 9\%$). GMAX peak EMG of maximal pedal cycles was possibly $3 \pm 11\%$ lower than those pedal cycles corresponding to non-maximal cycles at 60 rpm, while trivial differences were observed at 115 rpm and 170 rpm (Figure 3.6).

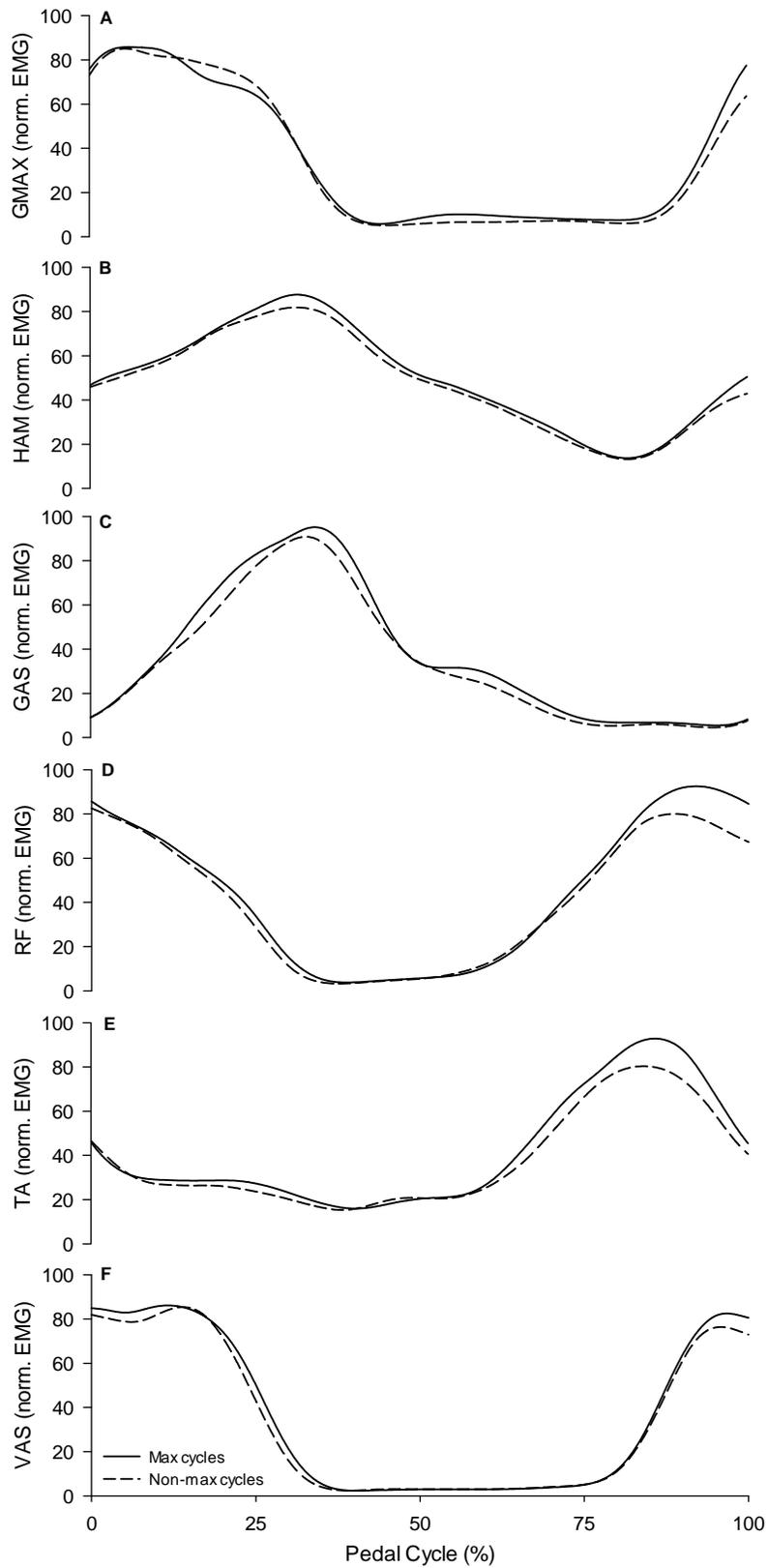


Figure 3.5. EMG profiles from maximal and non-maximal pedal cycles. **A:** GMAX, **B:** HAM, **C:** GAS, **D:** RF, **E:** TA, **F:** VAS. Lines represent means with SD lines omitted for clarity.

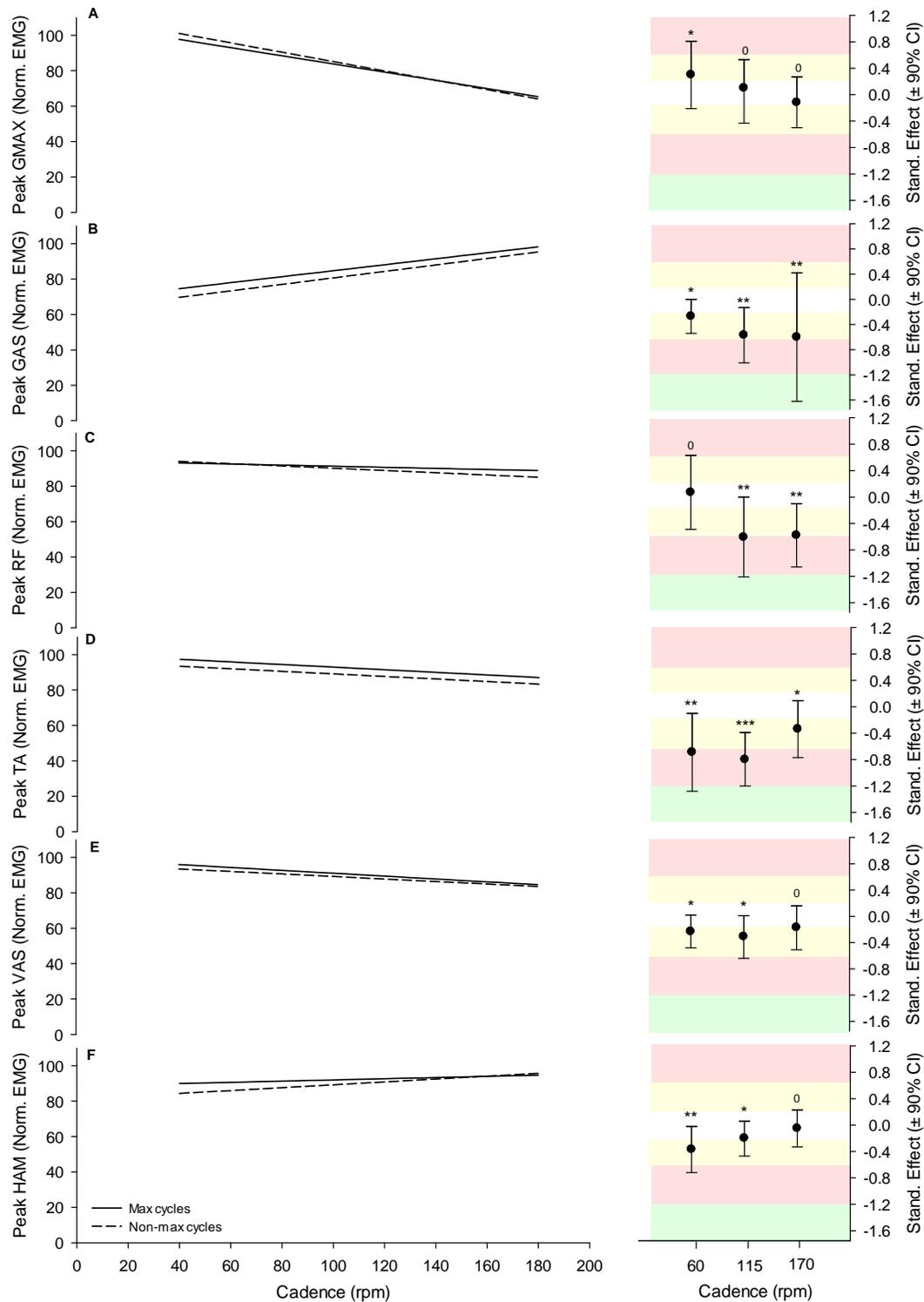


Figure 3.6. Peak EMG predicted from maximal and non-maximal cycles. **A:** GMAX, **B:** GAS, **C:** RF, **D:** TA, **E:** VAS, **F:** HAM. Lines represent means with SD lines omitted for clarity. Graphs to the right illustrate the standardised effect \pm 90% CI of the difference between maximal and non-maximal cycles at 60 rpm, 115 rpm and 170 rpm. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely or *** very likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly.

1.1.1.4 Differences in co-activation of the lower limb muscles

CAI values were higher for all muscle pairs by small to moderate magnitudes when calculated from EMG profiles obtained from maximal cycles compared to those obtained from non-maximal cycles (Figure 3.7).

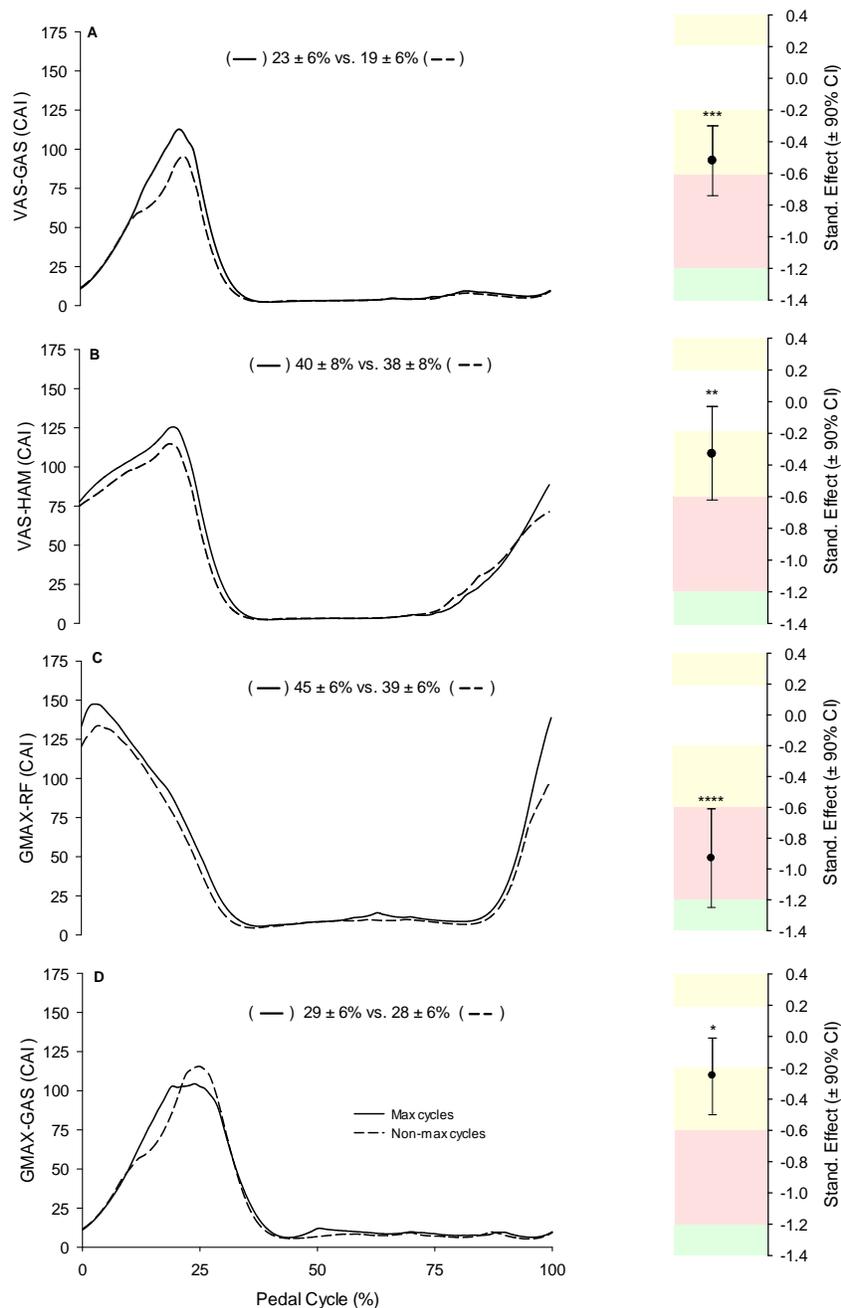


Figure 3.7. Average co-activation profiles and average CAI values for maximal and non-maximal cycles. **A:** VAS-GAS, **B:** VAS-HAM, **C:** GMAX-RF, **D:** GMAX-GAS. Lines represent means with SD lines omitted for clarity. Percentages stated on the graphs are average CAI values for maximal and non-maximal cycles. Graphs to the right illustrate the standardised effect \pm 90% CI of the difference between average CAI for maximal cycles vs non-maximal cycles. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely or **** most likely.

1.1.1.5 Differences in variability of crank torque and EMG profiles

Inter-cycle crank torque profile VR was likely lower for maximal cycle profiles compared to non-maximal cycle profiles (Figure 3.8 and Table 3.1). Similarly, inter-cycle VR for EMG profiles were lower for maximal cycles compared to non-maximal cycles for all muscles except for GMAX (Table 3.1).

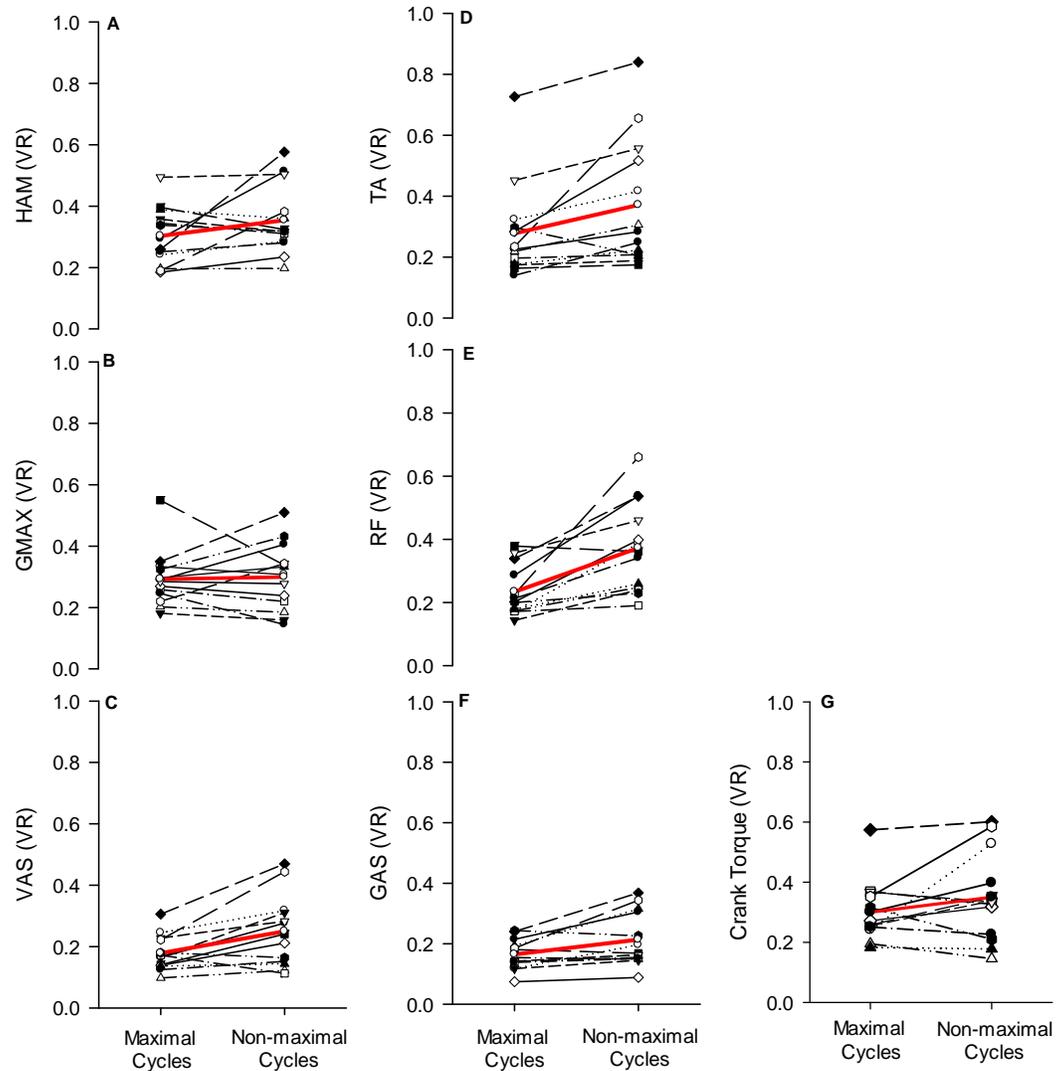


Figure 3.8. Between-cycle VR of EMG profiles and crank torque from maximal and non-maximal cycles. **A:** HAM, **B:** GMAX, **C:** VAS, **D:** TA, **E:** RF, **F:** GAS, **G:** crank torque. Each line represents one participant. Bold red line indicates mean response.

Table 3.1. Inter-cycle VR for crank torque, EMG and co-activation of muscle pairs from maximal and non-maximal cycles.

	Maximal Cycles	Non-maximal Cycles	Stand. Effect	Likelihood
Crank Torque	0.30 ± 0.10	0.35 ± 0.15	-0.43; ±0.52	**
GMAX	0.29 ± 0.09	0.30 ± 0.11	-0.07; ±0.51	0
HAM	0.30 ± 0.09	0.35 ± 0.11	-0.51; ±0.58	**
GAS	0.16 ± 0.05	0.21 ± 0.09	-0.90; ±0.54	***
RF	0.23 ± 0.08	0.37 ± 0.14	-1.65; ±0.70	***
TA	0.28 ± 0.16	0.37 ± 0.21	-0.55; ±0.36	**
VAS	0.18 ± 0.06	0.25 ± 0.12	-1.16; ±0.62	***
All Muscles	0.23 ± 0.09	0.29 ± 0.13	-0.71; ±0.21	****
VAS-GAS	0.24 ± 0.07	0.26 ± 0.11	0.26; ±0.38	*
GMAX-RF	0.25 ± 0.09	0.27 ± 0.09	0.21; ±0.42	*
VAS-HAM	0.23 ± 0.07	0.23 ± 0.09	0.06; ±0.76	0
GMAX-GAS	0.26 ± 0.07	0.26 ± 0.12	0.07; ±0.45	0
All Pairs	0.25 ± 0.08	0.26 ± 0.10	0.16; ±0.25	0

Data presented are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely or **** most likely. Likelihood of a trivial standardised effect is denoted as 0 possibly.

3.3.2 Prediction of individual T-C and P-C relationships

The number of data points selected for maximal cycles was 24 ± 3 . This subset of data was used in the analyses below to compare methods for predicting individual T-C and P-C relationships.

3.3.2.1 T-C relationships

Goodness of fit

Individual T-C relationships fit with high order polynomials had lower SEE values (3 ± 1 N·m vs 5 ± 2 N·m; factor of 0.7, 90% confidence limits 0.6 to 0.8), marginally higher r^2 values (0.98 ± 0.02 vs 0.96 ± 0.04 ; Figure 3.9A) and lower residuals between 40-50 rpm (5 ± 4 N·m vs 7 ± 6 N·m), 100-110 rpm (2 ± 3 N·m vs 4 ± 3 N·m) and 170-180 rpm (2 ± 1 N·m vs 5 ± 4 N·m; (Figure 3.9B) compared to low order polynomials. Additionally, less heteroscedasticity was seen for SEE, r^2 and residuals values when T-C relationships were described using high order polynomials (Figure 3.9B).

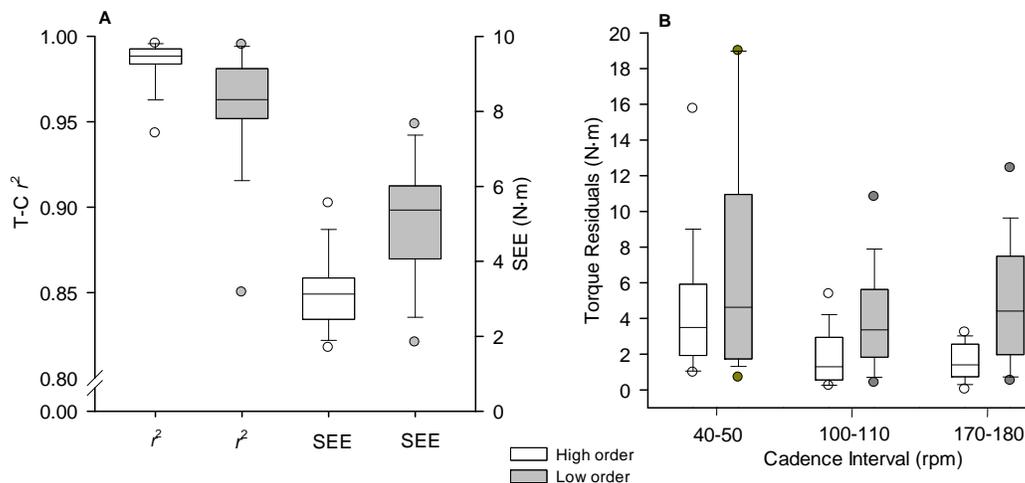


Figure 3.9. Goodness of fit variables and residuals estimated from T-C relationships fit with high and low order polynomials. **A:** calculated r^2 and SEE values, **B:** torque residuals. Box plot horizontal lines indicate median values, outliers (circles) indicate 5th/95th percentiles.

Prediction of average torque and T_0

At low cadences, torque values predicted using high order polynomials were very likely lower compared to those predicted using low order polynomials, as illustrated by differences observed for T_0 (144 ± 43 N·m vs 170 ± 33 N·m; Figure 3.12) and at 40 rpm (133 ± 26 N·m vs 144 ± 24 N·m) and 50 rpm (130 ± 23 N·m vs 137 ± 23 N·m; Figure 3.11). At high cadences, torque values predicted from high order polynomials were most likely and very likely lower than those calculated from low order polynomials, as illustrated by the differences observed at 170 rpm (50 ± 12 N·m vs 54 ± 11 N·m), 180 rpm (40 ± 13 N·m vs 47 ± 11 N·m), 190 rpm (29 ± 13 N·m vs 40 ± 12 N·m) and 200 rpm (18 ± 14 N·m vs 33 ± 12 N·m; Figure 3.11).

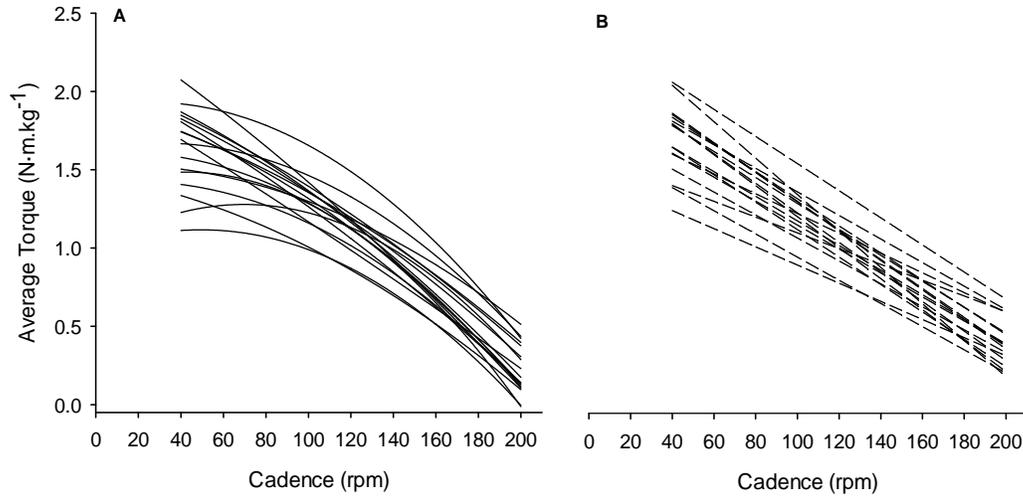


Figure 3.10. T-C relationships fit with high and low order polynomials. Individual relationships predicted from **A:** high order polynomials and **B:** low order polynomials. Average torque values are normalized to participant's body mass and each line represents one participant.

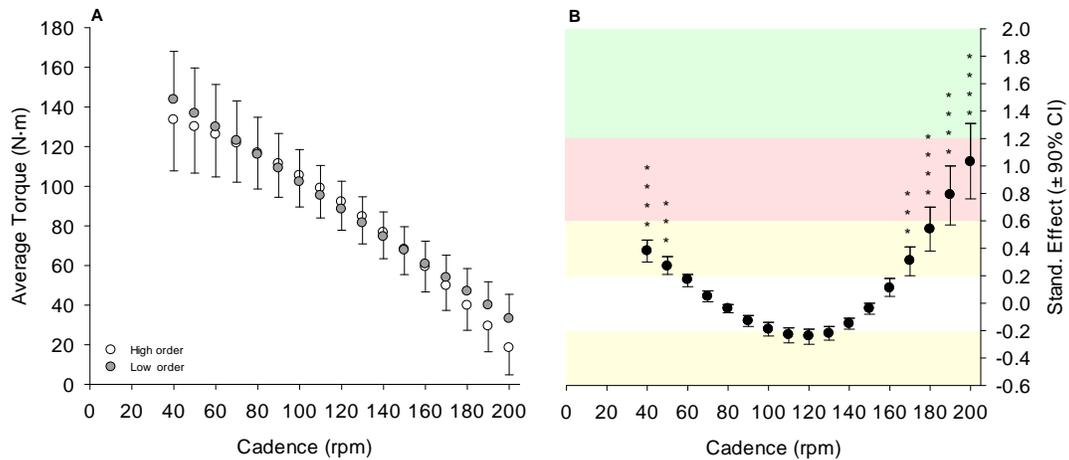


Figure 3.11. Torque predicted from T-C relationships fit with high and low order polynomials **A:** mean \pm SD torque, **B:** Standardised effect \pm 90% CI of the difference between torque predicted from high and low order polynomials. Likelihood of a non-trivial standardised effect is denoted as *** very likely or **** most likely (illustrated in the vertical direction).

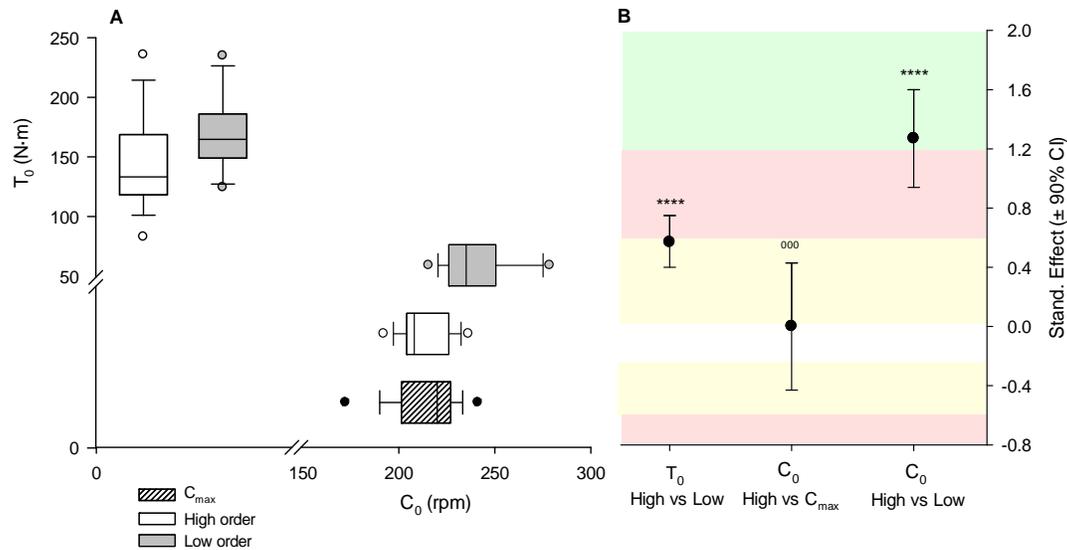


Figure 3.12. Limits of NMF- T_0 and C_0 fit with high and low order polynomials. **A:** Maximal torque (T_0) and maximal cadence (C_0) and experimentally measured maximal cadence (C_{max}). Box plot horizontal lines indicate median values, outliers (circles) indicate 5th/95th percentiles; **B:** standardised effect \pm 90% CI of the difference between variables predicted from high and low order polynomials. Likelihood of a non-trivial standardised effect is denoted as *** very likely or **** most likely.

3.3.2.2 P-C relationships

Goodness of fit

Individual P-C relationships were well described using high order polynomials, providing lower SEE values (29 ± 7 W vs 53 ± 20 W; 0.6, 0.5 to 0.7; Figure 3.13A), substantially higher r^2 values (0.97 ± 0.02 vs 0.89 ± 0.6 ; Figure 3.13A) and lower residuals at 40-50 rpm (37 ± 44 W vs 57 ± 35 W), 100-110 rpm (20 ± 17 W vs 26 ± 19 W) and 170-180 rpm (21 ± 14 W vs 53 ± 43 W; Figure 3.13B) compared to low order polynomials. Additionally, lower inter-individual dispersion was observed for SEE, r^2 and residual variables for high order polynomials.

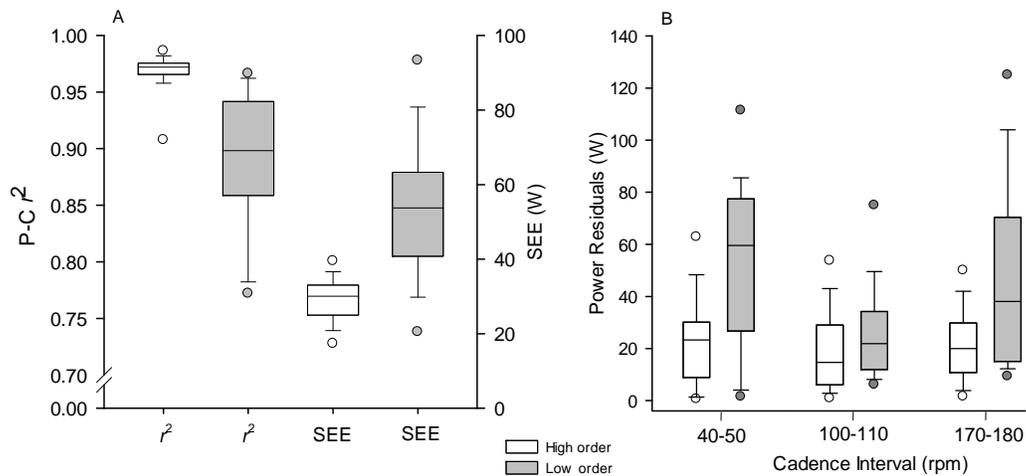


Figure 3.13. Goodness of fit variables and residuals estimated from P-C relationships fit with high and low order polynomials. **A:** calculated r^2 and SEE values, **B:** power residuals. Box plot horizontal lines indicate median values, outliers (circles) indicate 5th/95th percentiles.

Prediction of power, P_{\max} , C_{opt} and C_0

At low cadences, the power values predicted using high order polynomials were most likely lower than those predicted using low order polynomials, as illustrated by differences observed at 40 rpm (550 ± 114 W vs 629 ± 101 W), 50 rpm (673 ± 128 W vs 747 ± 119 W), 60 rpm (787 ± 139 W vs 849 ± 135 W) and 70 rpm (889 ± 148 W vs 934 ± 148 W; Figure 3.15). At high cadences, the power values predicted using high order polynomials were likely lower than those predicted using low order polynomials, as illustrated by the differences observed at 180 rpm (726 ± 266 W vs 829 ± 213 W), 190 rpm (545 ± 295 W vs 725 ± 227 W) and 200 rpm (328 ± 331 W vs 604 ± 245 W; Figure 3.15). Further, C_0 estimated from high order polynomials was reduced by a large magnitude compared to C_0 estimated from low order polynomials (214 ± 14 rpm vs 240 ± 20 rpm; Figure 3.12). C_0 values estimated using high order polynomials were not substantially different to the maximal cadences experimentally measured during the sprint performed against no external resistance (C_{\max} ; 214 ± 20 rpm), whereas C_0 values estimated using low order polynomials were most likely larger than C_{\max} . The apex of the P-C relationships (P_{\max}) calculated using high order polynomials was possibly higher compared to the apex calculated using low order polynomials (1174 ± 184 W vs 1132 ± 185 W; Figure 3.16), and likely higher when expressed in percentage of body mass (14.4 W.kg⁻¹ vs 13.9 W.kg⁻¹). Concomitantly, the cadence corresponding to the apex of the P-C relationships (C_{opt}) was likely higher when extracted from high order polynomials compared to low order polynomials (123 ± 9 rpm vs 120 ± 10 rpm; Figure 3.16). The C_{opt}/C_0 ratio

was most likely higher when calculated using high order polynomials compared to low order polynomials (0.57 ± 0.03 vs 0.50 ± 0.00).

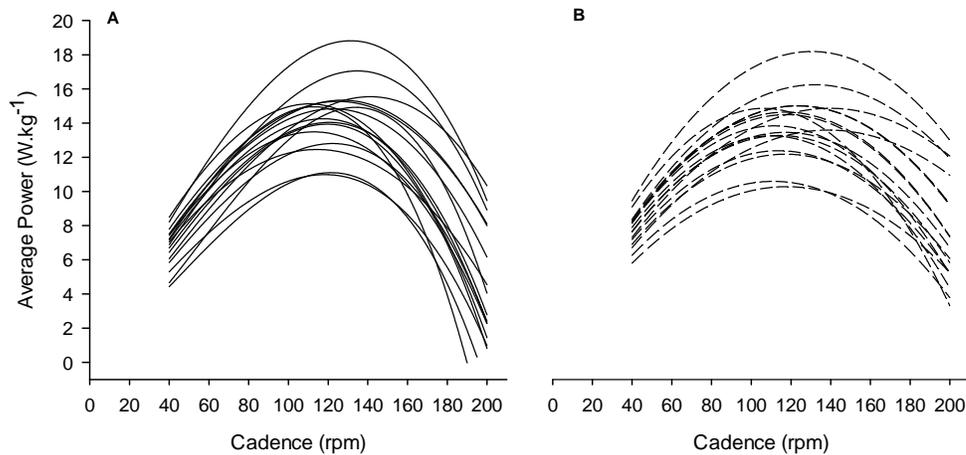


Figure 3.14. P-C relationships fit with high and low order polynomials. Individual relationships predicted from **A**: high order polynomials and **B**: low order polynomials. Average power values are normalized to participant's body mass and each line represents one participant.

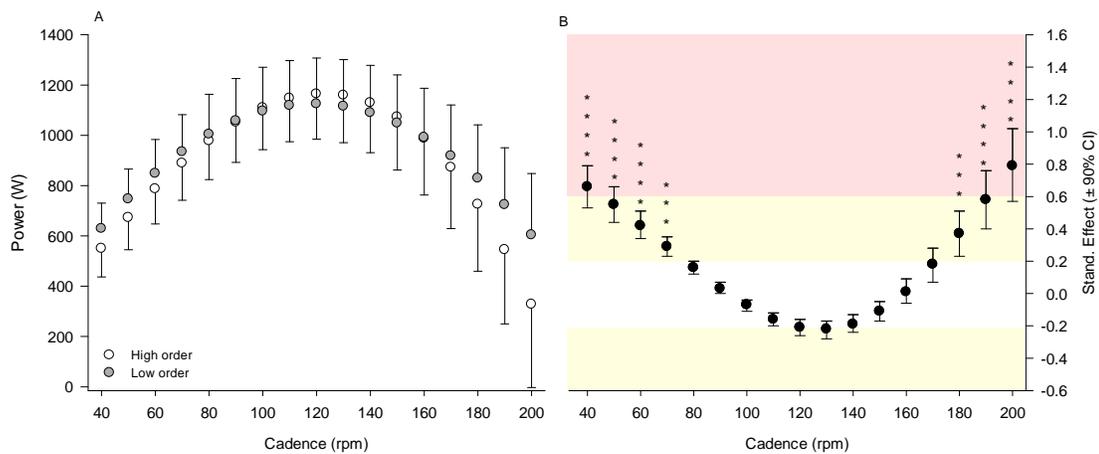


Figure 3.15. Power predicted from P-C relationships fit with high and low order polynomials **A**: mean \pm SD power, **B**: standardised effect \pm 90% CI of the difference between power predicted from high and low order polynomials. Likelihood of a non-trivial standardised effect is denoted as *** very likely or **** most likely (illustrated in the vertical direction).

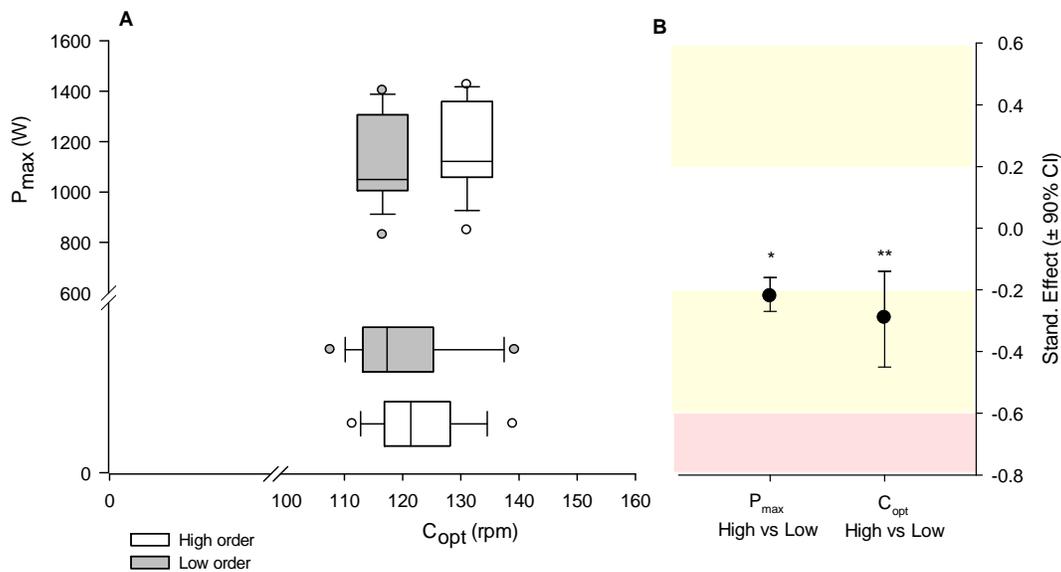


Figure 3.16. Limits of NMF- P_{max} and C_{opt} fit with high and low order polynomials. **A:** Maximal power (P_{max}) and optimal cadence (C_{opt}). Box plot horizontal lines indicate median values, outliers (circles) indicate 5th/95th percentiles; **B:** standardised effect \pm 90% CI of the difference between variables predicted from high and low order polynomials. Likelihood of a non-trivial standardised effect is denoted as * possibly and ** likely.

When the shape of individual P-C curves were predicted using high order polynomials, predicted power values on the right side of the P-C curve were not different to predicted power values on the left side of the P-C curve when cadence deviates from C_{opt} less than 35 rpm. Beyond 35 rpm predicted power values on the right side of the P-C curve were likely lower compared to predicted power values on the left side of the P-C curve with the difference ranging from most likely small when cadence deviated by 40 rpm from C_{opt} (966 ± 181 W vs 1006 ± 175 W; -0.22 ; ± 0.05 ; Figure 3.17) to most likely, very large differences when cadence deviated by 80 rpm from C_{opt} (263 ± 244 W vs 585 ± 144 W; -2.1 ; ± 0.4 ; Figure 3.17).

Trivial differences were observed between the power values predicted from high and low order polynomials on the left side of the P-C curves whereas power values predicted on the right side of the P-C curves were very likely lower at 45 rpm (908 ± 182 W vs 971 ± 166 W; 0.33 ; ± 0.08) and most likely lower at 50 (841 ± 184 W vs 933 ± 163 W; 0.48 ; ± 0.12), 55, 60, 65, 70, 75 and 80 rpm (263 ± 244 vs 623 ± 145 W; 1.4 ; ± 0.33) when using high order polynomials compared to low order polynomials (Figure 3.17).

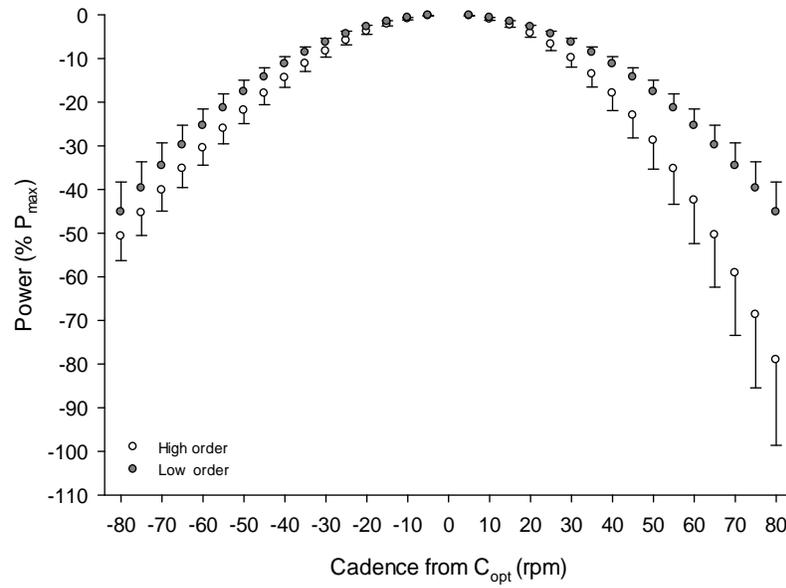


Figure 3.17. Power predicted from P-C relationships fit with high and low order polynomials at 5 rpm intervals moving away from C_{opt} on the ascending (i.e. negative values) and descending (i.e. positive values) limbs of the relationship. Data presented are mean \pm SD.

3.4 Discussion

The first purpose of this study was to measure variations in torque and EMG profiles between maximal and non-maximal pedal cycles obtained during a F-V test on a stationary cycle ergometer and secondly to compare the ability of two modelling procedures to predict T-C and P-C relationships and to quantify the limits of NMF. Analyses first show that selecting maximal pedal cycles at regular cadence intervals (i.e. every 5 rpm) over a wide range of cadences (from 40 to 180 rpm) resulted in an average value of torque that was higher than that predicted from non-maximal pedal cycles recorded during the F-V test. In association with this finding, peak crank torque, peak EMG, and co-activation of the lower limb muscles were higher for maximal cycles. Further, crank torque and EMG profiles exhibited less inter-cycle variability for maximal cycles. Secondly, higher order polynomials provided a better goodness of fit (improved r^2 and SEE and lower torque and power residuals) for both T-C and P-C relationships. The use of low order polynomials resulted in an overestimation of torque and power values predicted at low (<70 rpm) and high (>170 rpm) cadences and the estimation of T_0 and C_0 variables.

3.4.1 The effect of maximal data point selection

The method of F-V test employed in this study, made up of multiple sprints from a combination of rolling and stationary starts against varying external resistances enabled the collection of a large number of data points (57 ± 22) over a wide cadence range (41 ± 7 rpm to 180 ± 10 rpm), similar to that of Arsac et al. (1996). The large pool of data points collected allowed the highest measured value of torque to be selected within a given cadence interval (i.e. one per 5 rpm) which is not possible using F-V tests consisting of a single sprint effort (Martin et al., 1997). Further, to capture a similar range of cadences using a F-V test on an isokinetic cycle ergometer would require approximately 20 sprints, which is not feasible when assessing fatigue-free maximal torque and power production.

Comparison of maximal and non-maximal cycle revealed that torque values varied between pedal cycles and sprints at similar cadences by up to 6%. Although participants were instructed to produce a maximal effort for every sprint, the value of torque attained was not always maximal in the data recorded as illustrated in Figure 3.2. The within session increase we observed (following a single familiarization session on a separate day) was similar to the 4.3% increase in maximal power previously observed following two sequential days of practice in non-cyclists (Martin et al., 2000a). As such, the present findings suggest that filtering experimental data to include only the most maximal pedal cycles can have a similar effect as task familiarization on torque (and power) values. As power is a product of torque and cadence it is reasonable to conclude that selection of maximal power values would have mimicked those seen for T-C

relationships, resulting in P-C relationships that reflected a substantially higher level of power over the range of cadences measured. The collection of maximal data is important in circumstances where changes in power need to be precisely quantified such as the assessment of fatigue related changes in power, the efficacy of a training program (Cormie et al., 2010; Creer et al., 2004), and/or when kinematics of the pedalling movement are modified (Bini et al., 2010).

When delving into the results further, mechanical, EMG and co-activation profiles provided some insight into mechanisms behind the differences in torque observed between maximal and non-maximal pedal cycles. The magnitude of the force applied to the crank was substantially higher for maximal pedal cycles, with larger peak crank torque values observed (Figure 3.4). Similarly, in conjunction with the higher peak torque for maximal cycles, peak EMG was up to 11% higher for five of the lower limb muscles (HAM, GAS, RF, TA, VAS), of which four have been previously identified as the main contributors to the production and transfer of forces to the pedals during the extension (VAS and GAS) and flexion (RF and TA) phases of the pedal cycle (Zajac, 2002). Accordingly, it appears that participants could not maximally recruit their lower limb muscles for every pedal cycle and each sprint that they performed. As cycling is a complex, poly-articular movement, it is unlikely that every muscle being used will reach a maximal level of active state during each consecutive pedal cycle of a sprint bout. In fact, it has been shown that due to this high variability many repetitions of a movement is necessary to reach a voluntary maximal level of muscle activation (Allen et al., 1995). Further, more co-activation was observed for GMAX-RF, GMAX-GAS, VAS-GAS and VAS-HAM muscle pairs (Figure 3.7), which suggests that better inter-muscular coordination was observed during maximal cycles. In accordance with the biomechanical models of cycling, the greater co-activation observed for VAS-GAS, GMAX-RF and GMAX-GAS muscle pairs may have increased the amount of power transferred across the hip, knee and ankle joints and delivered to the crank during extension (Raasch et al., 1997; van Ingen Schenau, 1989; Zajac, 2002).

Finally, the analyses of inter-cycle variance ratios of crank torque, EMG and co-activation profiles revealed less variability in these profiles for maximal cycles (Figure 3.8), indicating that inter-muscular coordination was more optimal during maximal pedal cycles, in reference to motor learning theories (Muller & Sternad, 2009). Although variability is thought to be small for maximal intensity/high mechanical demand movements, a low level of variability in the neuro-musculo-skeletal subsystems of the body is ever present (Enders et al., 2013) and as shown in this study, should be accounted for by implementing adequate selection procedures for data recorded during a F-V test. Additionally, patterns of lower limb muscle recruitment appear to be more variable in novice cyclists (Chapman et al., 2008a), therefore the issue of EMG variability (and the need to filter data) becomes even more relevant for those who are unskilled in performing the pedalling movement, like the participants in this study. The use of F-V test

protocols like that employed in this study seems essential for the assessment of the limits of NMF in not just cycling but also in other voluntary exercise (e.g. jumping, running), as it increases the likelihood of recording and selecting data points that truly reflect the maximal force and power producing capabilities of an individual.

3.4.2 Prediction of T-C and P-C relationships

The results from the second half of the analyses clearly demonstrated that the shapes of the T-C and P-C relationships were better predicted using high order polynomials, in line with the approach adopted by a few previous studies (Arsac et al., 1996; Hautier et al., 1996; Yeo et al., 2015). The improved prediction of T-C and P-C relationships, using second and third order polynomials respectively was evidenced by higher r^2 values (Figure 3.9 and Figure 3.13) similar to values previously reported by Arsac et al. (1996), also in a non-cyclist population. The increased r^2 values were accompanied by a reduction of SEE values and average torque and power residuals, showing that T-C and P-C relationships described using higher order polynomials allowed for more accurate and valid predictions of torque and power values. Another important finding of this study is the observed reduction of the heteroscedasticity of r^2 , SEE and torque/power residual values associated with the use of higher order polynomials, indicating that higher order polynomials resulted in good prediction of T-C and P-C relationship shape for most participants. On one hand, it appeared that T-C relationships exhibited by two participants were almost perfectly linear while the shape of their P-C relationships was almost a symmetrical parabola (see Figure 3.10 and Figure 3.14). For these participants, the shape of T-C and P-C relationships could be successfully predicted using low order polynomials with the use of higher order polynomials only having a minor impact on the quality of the prediction, as reflected by small changes in r^2 and SEE values (e.g. one participant presented with the same r^2 (0.97) and SEE (16 W) values for both low and high order polynomials). However, on the other hand, the use of higher order polynomials had a much larger impact on predicted T-C and P-C relationship shapes of other participants, as reflected by large changes in r^2 and SEE values (e.g. one participant showed a substantial improvement of P-C relationship r^2 (0.86 to 0.97) and SEE (58 W to 25 W) values using high order polynomials). For the participants showing substantial improvement, visual inspection showed the importance of using higher order polynomials considering the curvilinear shapes of T-C relationships and asymmetrical parabolic shapes of P-C relationships. Altogether, these results show that higher order polynomials are more suited to predict the shapes of T-C and P-C relationships of non-cyclists, as the shapes of their relationships can deviate from the linear and symmetrical parabolas commonly assumed by researchers (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007).

3.4.3 Prediction of the limits of lower limb NMF

Analysis of the results obtained on the left side of the T-C and P-C relationships revealed that predicted values of torque and power were lower below 50 rpm and 70 rpm respectively, while a 22% reduction in T_0 was observed using higher order polynomials. As illustrated in Figure 3.11 and Figure 3.15 these results quantify the downward curvature that was observed at low cadences in the T-C and P-C relationships of some participants. Further, the reduction in torque/power observed at low cadences corroborates with previous studies which have indicated that neural inhibitions (Babault et al., 2002; Perrine & Edgerton, 1978; Westing et al., 1991; Yamauchi et al., 2007) and/or muscle potentiation (Robbins, 2005) may reduce the level of torque/power that can be produced during movements performed at low velocities. As depicted in Figure 3.10 the amount of downward curvature observed in T-C relationships at low cadences was variable between participants when higher order polynomials were used. This variability in downward curvature at low cadences did not appear to be associated with the maximal power participants could produce which is in contrast to Vandewalle et al. (1987) who observed greater downward inflections in powerful males ($>17 \text{ W}\cdot\text{kg}^{-1}$) when torque was high. For example, the most powerful participant in this study ($18.8 \text{ W}\cdot\text{kg}^{-1}$) did not exhibit the same degree of downward inflection at cadences below 70 rpm as participants with lower maximal power abilities (i.e. $11.1 \text{ W}\cdot\text{kg}^{-1}$ and $12.8 \text{ W}\cdot\text{kg}^{-1}$). Further, the difference observed in extrapolated T_0 indicate that linear regressions used in previous studies may not provide a valid estimation for all participants, and hence could misreport knee extensor muscle strength, as the two variables have been previously linked (Driss et al., 2002).

Analysis of the results obtained on the right side of the T-C and P-C relationships revealed that at higher cadences, values of torque and power were lower predicted from high order polynomials. Although, values of maximal cadence (C_0) extrapolated from low order polynomial P-C relationships were similar to those reported previously in non-cyclist populations (Dorel et al., 2010; Driss et al., 2002; Martin et al., 1997), when C_0 was predicted from high order polynomials the values were ~ 26 rpm lower. Like noted for T_0 , it appears that values of C_0 previously reported may have been overestimated in studies using linear regressions. Fortunately, due to the nature of the cycling exercise an experimental measure of maximal cadence (C_{max}) was easily attainable via chain removal from the cycle ergometer, even though inclusion of a sprint at zero external resistance is not usually included in a F-V test (McCartney et al., 1985). When C_0 values predicted from T-C relationships fit with higher order polynomials were compared to C_{max} , there was no difference in the two variables (i.e. a trivial difference), providing further support for the use of high order polynomials. The reduced ability of the non-cyclist participants to produce power/torque on the right side of the curve (including C_0 and C_{max}) may have been attributable to the increasing effect of activation-deactivation dynamics as cadence moved beyond

their optimal (>120 rpm), in line with findings of Van Soest and Casius (2000) and/or changes in their motor control strategy (McDaniel et al., 2014).

Providing further support for the notion that P-C relationship is not always a symmetrical parabola are the results showing that power predicted from higher order polynomials were substantially different between the ascending and descending limbs at comparative cadences of either side of C_{opt} (i.e. below C_{opt} and above C_{opt} respectively) (Figure 3.17). The magnitude of the difference became larger as cadence assessed moved further from C_{opt} , indicating that the P-C relationship remains symmetrical over the apex, but becomes more asymmetric moving towards the limits of NMF, as the presence of aforementioned mechanisms affecting power production at low and high cadences start to become more relevant. The participant's ability to produce power was reduced more at higher cadences indicating that the mechanisms impacted by high movement frequencies such as activation-deactivation dynamics may have a greater effect than those suggested to affect power production at low cadences (e.g. neural inhibitions) (Babault et al., 2002; van Soest & Casius, 2000; Yamauchi et al., 2007). Just as the shape of the F-V relationship has been shown to change from hyperbolic in muscle (Hill, 1938; Thorstensson et al., 1976a; Wilkie, 1950) to near linear in other multi-joint movements (Bobbert, 2012), the downward inflections in T-C and P-C curve shape observed at low and high cadence intervals Figure 3.11 and Figure 3.15 may in part occur due to the complexity of leg cycling exercise requiring a higher level of external force control. Due to these inflections the collection of data points below 70 rpm and above 180 rpm is encouraged as the cadence range to which regression lines are fit are likely to affect extrapolated T_0 and C_0 . Indeed, an advantage of the F-V test protocol employed in the current study was the obtainment of a large number of data points over a wide range of cadences which enabled a more accurate estimate of T_0 and C_0 values.

Recent studies have gone beyond interpretation of F_0/T_0 and V_0/C_0 values separately and have assessed the F-V mechanical profile using the slope of the F-V relationship calculated from a linear regression (Giroux et al., 2016; Morin et al., 2002; Samozino et al., 2014; Samozino et al., 2012). However, as the results show T_0 and C_0 values extrapolated from T-C relationships fit with linear regressions were overestimated by 22% and 13% respectively, using these values to calculate the slope of the relationship in maximal cycling is likely to lead to an inaccurate calculation. If the T-C relationship is not linear and as a consequence the slope cannot be accurately assessed it may be better to assess and compare the shape of individual P-C curves using predicted torque and power at regular cadence intervals as an alternative. Moving towards the apex of the P-C curve the results showed that predicting the shapes of P-C relationships using third order polynomials resulted in a possible, small increase of P_{max} ($4 \pm 2\%$) associated with a likely, small reduction of C_{opt} (-3 ± 1 rpm). These findings show that higher order polynomials appear to have only a possible impact on estimated P_{max} and C_{opt} suggesting that these values

previously estimated in research employing low order polynomials are still likely to be valid (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007).

3.5 Conclusion

In summary, due to the inability of individuals to maximally and optimally activate their lower limb muscles, F-V test protocols consisting of multiple sprints should be employed to enable the collection of a large number of data points for a given cadence. Further, the identification of pedal cycles representing a true maximal value of torque and power should be chosen prior to modeling T-C and P-C relationships. Maximal pedal cycles modeled with higher order polynomials provided an improved goodness of fit of the T-C and P-C relationships, leading to lower predicted torque and power values at low (<70 rpm) and high (>170 rpm) cadences compared to more commonly used low order polynomials. As such, the T-C relationship does not appear to be linear and the P-C relationship a symmetrical parabola as previously thought in maximal cycling, which can affect variables commonly estimated to assess the limits of lower limb NMF.

Chapter 4 The Effect of High Resistance and High Velocity Training on a Stationary Cycle Ergometer

4.1 Introduction

Maintaining and improving NMF is necessary for sustaining healthy movement across the lifespan (Martin et al., 2000c). Therefore, the improvement of the limits of lower limb NMF (i.e. maximal power, maximal force, maximal velocity and optimal cadence) is often a major focus in training programs for a wide range of populations from athletes and healthy individuals (Cormie et al., 2011; Cronin & Sleivert, 2005) to the elderly, the injured and those with movement disorders (Fielding et al., 2002; Marsh et al., 2009). Traditional resistance training programmes (e.g. squat, leg press) are often used to improve the amount of force and power that can be produced (Cormie et al., 2007; McBride et al., 2002). However, ballistic training (e.g. squat jump) is commonly recommended in favour of more traditional resistance training exercises when improvements in power are sought, due to their specificity to many sports, allowing better transfer of adaptations to performance (Cady et al., 1989; Cronin et al., 2001; Kraemer & Newton, 2000; Kyröläinen et al., 2005; Newton et al., 1996). Although not viewed as a traditional form of ballistic exercise training, sprints performed on a stationary cycle ergometer also requires individuals to maximally activate muscles over a larger part of the movement, facilitating greater adaptations and thus may be beneficial for improving the limits of NMF. Further, the external resistance at which the exercise is performed can be easily and safely manipulated on a stationary cycle ergometer, making it an ideal exercise for interventions aimed at improving the power producing capacities of the lower limb muscles.

It is well known that improvements in power can occur as little as three weeks into an exercise program. The gains in power are attributable to neural adaptations such as increased neural drive and more optimal inter-muscular coordination of the trained muscles (Enoka, 1997; Hakkinen et al., 1985; Hvid et al., 2016; Kyröläinen et al., 2005; Moritani & DeVries, 1979). Indeed, neural adaptations have been suggested to be behind the improvements in power observed after just two days of maximal cycling practice in untrained cyclists (Martin et al., 2000a), and after longer interventions of between 4 to 8 weeks (Creer et al., 2004; Linossier et al., 1993). Although these studies are useful for quantifying the overall efficacy of training, these authors did not analyse the changes in the limits of the NMF, only changes in P_{\max} or power produced over a sprint.

It is well known that cadence affects the amount of torque and power that can be produced during maximal cycling, as illustrated by the torque-cadence and power-cadence relationships. The production of a high level of power at a given cadence requires optimal coordination of the lower limb muscles and joints to produce high levels of power (Raasch et al., 1997). In particular, co-

activation of proximal-distal muscle pairs has been suggested as essential for effective force/power transfer to the crank (Kautz & Neptune, 2002; Van Ingen Schenau et al., 1995). However, our ability to produce power on the left side of the T-C and P-C relationships (i.e. low cadences and high resistances) may be affected by different physiological mechanisms such as neural inhibitions and muscle potentiation (Babault et al., 2002; Perrine & Edgerton, 1978; Robbins, 2005; Westing et al., 1991; Yamauchi et al., 2007) compared to those playing a role on the right side of these relationships (i.e. at high cadences) which include activation-deactivation dynamics and altered motor control strategies (McDaniel et al., 2014; van Soest & Casius, 2000). Further, there is an abundance of motor solutions offered within the human body to produce power using different movement strategies (Bernstein, 1967; Latash, 2012). Training appears to reduce the variability in execution variables (Muller & Sternad, 2009) (i.e. joint kinematic, EMG, co-activation and crank torque profiles) optimizing joint motion and inter-muscular coordination (Chapman et al., 2008b; Hug et al., 2008; Wilson et al., 2008). Indeed, less variability is well accepted as an indicator of motor learning and movement control, exhibited by those well-trained in a task (Hug et al., 2008; Muller & Sternad, 2009). Despite these findings, the key adaptations occurring with intervention-specific training on a stationary cycle ergometer have not been previously examined, nor have the adaptations been linked to changes in the limits of lower limb NMF.

In light of the previous literature, the first aim of this study was to investigate if the adaptations of the limits of NMF would be specific to the training intervention selected. To investigate this power produced between 60-90 rpm and 160-190 rpm and key variables calculated from T-C and P-C relationships (i.e. P_{max} , C_{opt} , T_0 and C_0) were assessed before and after the training. Extending upon principles of training specificity, it was assumed that training against high resistances would alter the limits of NMF on the left side of the P-C relationship (i.e. increase T_0 and the power generating capacity at low to moderate cadences), while training at high cadence would alter the limits of NMF on the right side of the P-C relationship (i.e. increase the power generating capacity at moderate to high cadences and C_0). The second aim of this study was to investigate if different motor control adaptations would accompany the changes in the limits of NMF. Also, due to the short duration of the intervention (only four weeks), it was assumed that any changes in the limits of the NMF would be due to neural adaptations. To investigate this aim, changes in the amplitudes of crank torque and joint angle profiles and average co-activation of muscle pairs were assessed before and after training, as well as inter-cycle and inter-participant variance ratios were calculated for crank torque, hip, knee and ankle joint, EMG and co-activation profiles. It was assumed that the variability of crank torque, kinematic and EMG profiles would be reduced after training for the same cycling condition (i.e. at low to moderate cadences for those training against high resistances and at moderate to high cadences for those training at high

velocities). Further, modifications to torque applied to the crank, inter-joint and inter-muscular coordination after training could also explain the potential change in the limits of NMF.

4.2 Methods

4.2.1 Participants

Seventeen low-to-moderately active males volunteered to participate in this study. These participants were the same group that had previously volunteered for study one (Chapter 3) and were involved in recreational physical activities such as resistance training and team sports, but did not have any prior training experience in cycling. The experimental procedures used in this study were approved by Victoria University's Human Research Ethics Committee and carried out in accordance with the Declaration of Helsinki. Subjects gave written informed consent to participate in the study if they accepted the testing procedures explained to them.

4.2.2 Experimental design

Participants attended familiarisation sessions to allow participants to become accustomed with the cycle ergometer setup and maximal cycling exercise. It has been shown that one familiarization session is enough to obtain reproducible measurements of maximal power in young non-cyclist adults (Doré et al., 2003). Within one week of the familiarisation session, participants performed a baseline assessment that consisted of a Force-Velocity (F-V) test on an isoinertial cycle ergometer; and anthropometric measurements of the lower limbs. Within four days of baseline testing completion, participants started a four-week training intervention in Victoria University's Exercise Physiology laboratory. All sessions were supervised and participants were verbally encouraged to produce maximal efforts for each of the sprints they performed. Within one week of completion of the training interventions, participants returned to the laboratory for post-training testing. In an attempt to minimise the influence of additional exercise performed outside of the prescribed training, subjects were asked not to deviate from their normal exercise routine between testing sessions. Participants in this study had previous training history in team sports (soccer and Australian Rules Football) and in resistance based exercises performed 2-3 times per week, with minimal discrepancy of participant training history between the groups. Participants were also asked to refrain from consuming caffeinated beverages and food 12 hours prior to each test.

4.2.3 Training interventions

The study was a two-group (high resistance; RES and high cadence; VEL) parallel design, controlled trial. Participants were pair-matched for maximal power (P_{max}) (RES: $14.5 \pm 1.7 \text{ W}\cdot\text{kg}^{-1}$ vs VEL: $14.4 \pm 2.3 \text{ W}\cdot\text{kg}^{-1}$; standardised effect; 0.04) and optimal cadence (C_{opt}) (RES: $122 \pm 10 \text{ rpm}$ vs VEL: $122 \pm 7 \text{ rpm}$; standardised effect; 0.02) using results from baseline F-V tests. Participants were then randomly allocated to either a high resistance (mean \pm SD; RES; $n = 9$; age,

27 ± 4 yr.; body mass, 81 ± 12 kg; height, 182 ± 8 cm) or high cadence (VEL; n = 8; age, 25 ± 4 yr.; body mass, 83 ± 11 kg; height, 179 ± 8 cm) training group.

Participants allocated to RES group performed all-out efforts against high external resistances, while participants allocated to the VEL group performed all-out sprints at high cadences via the use of low external resistances (Figure 4.1). Both groups performed two training sessions per week, separated by at least 48 hours over four weeks. An electromagnetically braked cycle ergometer (Velotron, Racermate Inc., Seattle, USA) was used for all training sessions. Training sessions for each group were matched for number of revolutions completed, number of revolutions per sprint and sprints per session (Tomas et al., 2010). A 4-min rest period was given between sprints to allow adequate recovery. The first training session included four sprints consisting of 16 pedal revolutions per sprint, with a total of 64 revolutions completed. An additional sprint was added to each session thereafter with each sprint within the sessions consisting of one less pedal revolution (i.e. session two included five, 15 revolution sprints; session three included six, 14 revolution sprints; session four included seven, 13 revolution sprints and so forth, until session eight, the last session, which included 11, 9 revolution sprints).

Prior to the sprints, subjects completed a warm up of 5-min of cycling at 80-100 rpm at a workload of 1.2 W.kg⁻¹. For the RES training sessions, high resistances were set (4-8 N·m.kg⁻¹) and the cycle ergometer was fit with an 85 tooth front sprocket and a 14 tooth rear sprocket. For the VEL training sessions, low resistances were set (0.1-0.5 N·m.kg⁻¹) and cycle ergometers fit with a 62 tooth front sprocket and a 14 tooth rear sprocket were used and each sprint bout was started with the flywheel rolling at approximately 130 rpm. For these rolling starts, the experimenter accelerated the flywheel immediately prior to the sprint so participants could initiate their sprints at the target cadence without prior effort. The external resistance was individually adjusted throughout the programme to enable participants to perform within the confines of their training condition. Each subject's riding position during training was kept consistent with the F-V protocol by using the same ergometer set-up features (i.e. saddle height, saddle fore-aft position, handlebar height and handlebar fore-aft position).

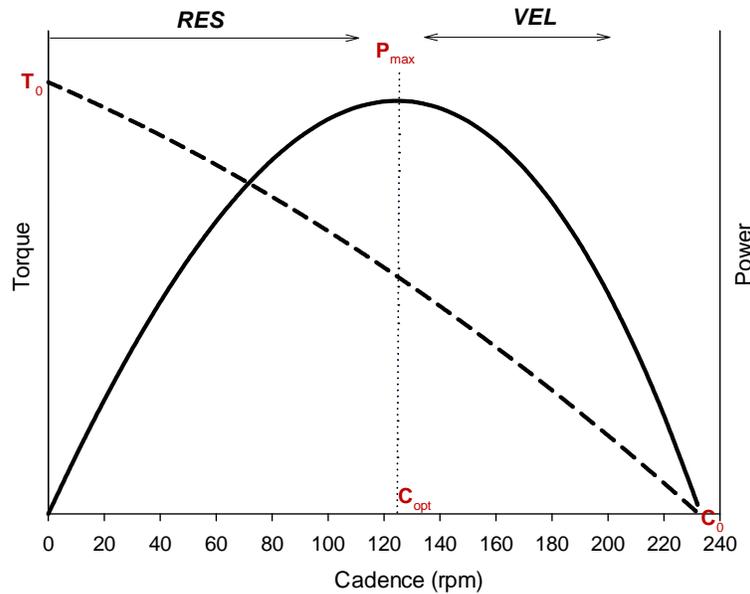


Figure 4.1. Sections of the T-C and P-C relationships for which RES and VEL trained during the four week intervention.

Training intervention adherence was good with all participants attending every training session and all completing both pre- and post-training testing sessions. The RES group cycled against external resistance of $4\text{-}8\text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ during training which resulted in a cadence range of 0 rpm to 122 ± 15 rpm covered in training. The VEL group cycled against external resistances of $0.1\text{-}0.5\text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ which enabled participants to cycle at high cadences ranging from 131 ± 5 rpm to 211 ± 10 rpm during training.

4.2.4 Evaluation of RES and VEL training interventions on NMF

4.2.4.1 Limits of NMF during maximal cycling exercise

Force-velocity test protocol

A F-V test of five sprints was implemented before and after the training interventions. This test was the same as that described in Study one (section 3.2.2.1) with the exclusion of the sprint against zero external resistance. The order of the sprints were randomised for each subject.

Analysis of T-C and P-C relationships

Torque-cadence (T-C) and power-cadence (P-C) relationships were analysed using the same methods that described the identification of maximal pedal cycles and implementation of 2nd and 3rd order polynomials respectively in section 3.2.2.1. In brief, for each pedal cycle average torque and cadence were derived from the SRM powermeter and TorxtarTM, and power was calculated using Eqn. 1. Pedal cycles corresponding to the highest values of torque and power were then selected, one value for every 5 rpm cadence interval and used to create individual T-C and P-C relationships. Further, as per the methods outlined in section 3.2.4.1 individual T-C relationships were fit with 2nd order polynomials with the equation of the regression line used to calculate T_0 . Individual P-C relationships were fit with 3rd order polynomials as per section 3.2.4.2 with a fixed y-intercept set at zero. P_{max} was identified as the apex of the P-C relationship, and C_{opt} the value corresponding to P_{max} . C_0 was estimated as the intercept of the P-C relationship with the x-axis. Using estimated C_{opt} and C_0 values a C_{opt}/C_0 ratio was also calculated. Power was predicted from P-C regression equations at 5 rpm intervals. Average power was then calculated for two cadence intervals using data points between 60-90 rpm and between 160-190 rpm.

4.2.4.2 Control of the pedalling movement

Crank torque profiles

Total crank torque signals were time-normalised to 100 points using the time synchronised events of left and right top-dead-centre (i.e. LTDC-LTDC for left leg and RTDC-RTDC for right leg) to create crank torque profiles for each pedal cycle as per methods outlined in section 3.2.3.1. Average crank torque profiles were then calculated for the two cadence intervals- 60-90 rpm and 160-190 rpm. Average values of peak and minimum crank torque were then identified from these profiles for the two cadence intervals.

Kinematics of the lower limb joints

Three-dimensional kinematic data were collected using a VICON motion capture system (Oxford Metrics Group Plc, Oxford, UK) that consisted of ten T-40s cameras, sampling at 250 Hz. Retro-reflective markers (14 mm diameter) placed on the lower limbs were tracked and recorded using VICON NEXUS 1.7 software (Oxford Metrics Group Plc, Oxford, UK). A biomechanical calibration model of each participant's lower limb was created using a combination of retro-reflective markers and virtual calibration markers at relevant anatomical landmarks using Visual3D Real-Time software (version 5, C-Motion Pty). Figure 4.2 illustrates the marker placement set-up that was adopted for obtaining a six-degrees-of-freedom biomechanical model, where clusters of

tracking markers were attached to the pelvis, thigh, shank and foot. This type of marker set-up is designed for reconstructing 6-DOF segment kinematics as recommended by Cappozzo et al. (1995). To avoid soft tissue artefact caused by the thigh and shank muscles, the marker clusters were fixed to plastic shells and secured to the lateral and distal regions of the segment using adhesive tape (Stagni et al., 2005). Four tracking markers were placed in a non-collinear array on the lateral aspect of semi-rigid cycling shoes (Figure 4.4). Calibration markers were digitised with respect to relevant segment cluster of tracking markers using a digitising pointer (C-Motion, Pty). Calibration markers included manually palpated anatomical landmarks to identify the pelvis (anterior superior iliac spine; ASIS and posterior superior iliac spine; PSIS), hip joint (lateral greater trochanter), knee joint (lateral and medial epicondyles), ankle joint (medial and lateral malleoli), and metatarsal-phalangeal joints (2nd and 5th metatarsal heads) (Figure 4.2). Calibration markers were used to reconstruct a three-dimensional model of the pelvis, hip, knee and ankle using Visual3D (version 5, C-Motion Pty). Kinematic data were recorded for all sprint trials. Target markers of each test trial were labelled in VICON NEXUS, exported as *.c3d files and post-processed in Visual 3D.

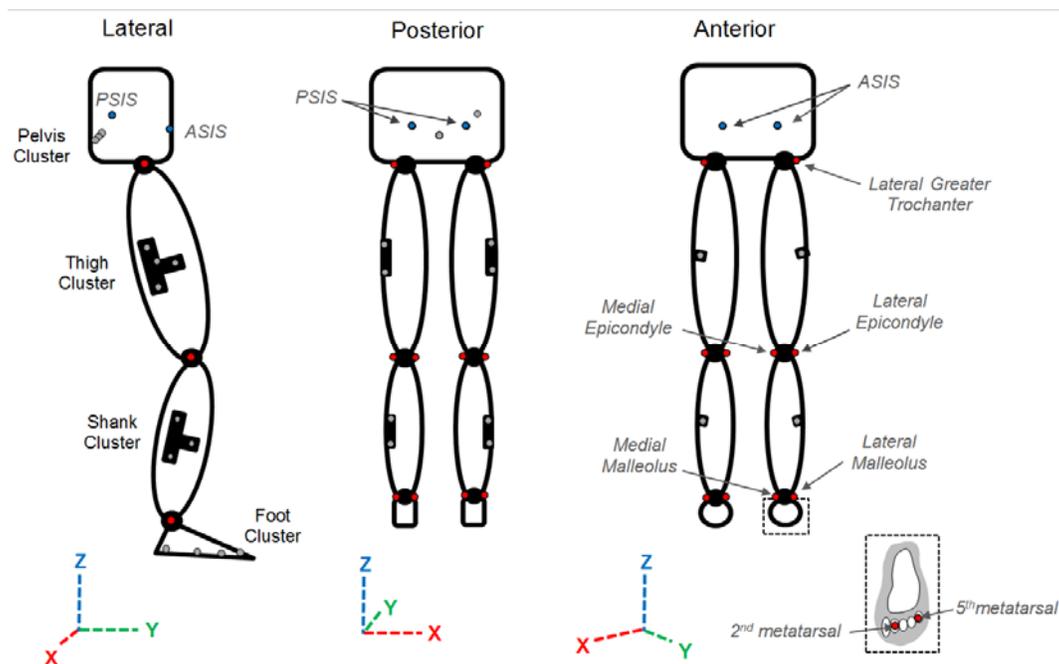


Figure 4.2. Motion capture marker set up. Grey circles indicate the location of the tracking markers on the pelvis, thigh, shank and foot (cycling shoe). Red circles indicate the calibration markers used for building a three-dimensional model of the lower limbs. Blue circles indicate the markers used for both tracking and calibration. XYZ indicate the coordinates of the laboratory.

Data analysis of the sprint trials was performed using Visual3D (C-Motion). Raw kinematic data was interpolated and low-pass filtered using a 4th order Butterworth digital filter using a cut-off frequency of 10 Hz. The three-dimensional static model was fitted to the processed data of the test trials using a least-squares procedure in Visual-3D. A six degrees of freedom method (least-squares segment optimization) was applied to determine optimal segment position and orientation (Challis, 1995). Three-dimensional kinematic details of sprint trials was obtained from local segment coordinate systems defined in Visual3D, by adopting the method of Grood and Suntay (1983). The X-axis of the pelvis coordinate system was defined from the origin (mid-point between the ASIS markers) towards the right ASIS, the Z-axis perpendicular to the XY plane and the Y-axis as the cross product of the X-axis and Z-axis. The XYZ coordinate system of the thigh had its origin at the hip joint centre with positive Z-axis directed superior and in-line with knee joint center. The positive Y-axis was directed orthogonal and anterior to the frontal plane, and the positive X-axis directed orthogonal and lateral to the sagittal YZ plane. The XYZ coordinate system of the shank had its origin at the knee joint center (mid-point of the inter-epicondylar axis), with positive Z-axis directed superior and in-line with ankle joint center. The positive Y-axis was directed orthogonal and anterior to the frontal plane, and the positive X-axis directed orthogonal and lateral to the sagittal YZ plane. The XYZ coordinate system of the foot had its origin at the ankle joint center (mid-point of the inter-malleolar axis), the Z-axis directed proximally and in-line with the second metatarsal head, the Y-axis orthogonal and anterior to the frontal plane, and the medio-lateral axis directed lateral and orthogonal to the sagittal YZ plane.

Angular displacement signals of the hip, knee and ankle joints were computed in Visual3D using an XYZ Cardan sequence convention (e.g. Cole et al. (1993)), where X defines the medio-lateral direction, Y defines the anterior-posterior direction, and Z defines the vertical direction. Hip, knee and ankle joint displacement signals were time-normalised to pedal cycle using time events of LTDC and RTDC with extension (plantar-flexion) and flexion (dorsi-flexion) identified by local minimum and maximum metric values of the hip, knee and ankle joint angle signals within each pedal cycle. Joint range of motion (ROM) was derived for each cycle by taking the difference between the maximum and minimum angles (Figure 4.3). Average joint angle profiles (hip, knee and ankle) were created for two cadence intervals: 60-90 rpm and 160-190 rpm from the same pedal cycles used for the analysis of torque profiles. Average minimum and maximum joint angles and ROM were also calculated from these pedal cycles.

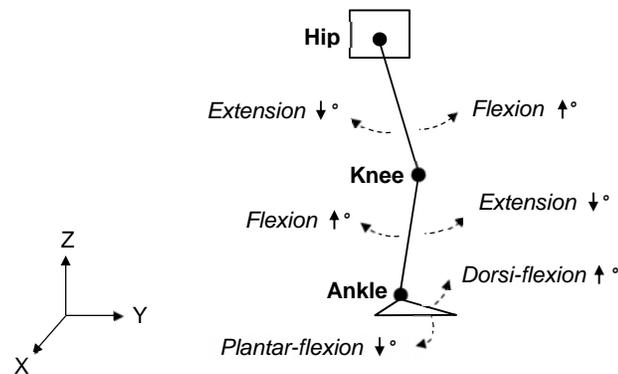


Figure 4.3. Interpretation of hip, knee and ankle joint movement. Dashed arrows indicates the direction the limb segment for a given phase of movement (e.g. extension). Solid arrows indicate that as joint angle decreases the joint is moving into extension/plantar-flexion, while as joint angle increases the joint is moving into flexion/dorsi-flexion. XYZ indicate the coordinates of the laboratory.

EMG activity of the lower limb muscles

Surface EMG signals were recorded from GMAX, RF, VAS, HAM, GAS and TA muscles. Attachment of the electrodes and filtering process of the raw EMG signal were consistent with the methods outlined in study one (section 3.2.3.2). Positions of the electrodes were marked on the participant's skin at baseline testing and throughout the training intervention to ensure better reproducibility of electrode placement in the post-training testing session. The processed EMG signals were time-normalised to 100 points between LTDC-LTDC and RTDC-RTDC for each muscle. The amplitude of the RMS of each muscle was normalised to the maximum (peak) amplitude which was recorded during the respective F-V test (i.e. pre-training EMG normalised to peak amplitude recorded during pre-training F-V test, post-training EMG normalised to peak amplitude recorded during post-training F-V test). This amplitude normalisation technique follows the methods recommended by Rouffet and Hautier (2008) to limit the impact of non-physiological factors on EMG signals (Farina et al., 2004). Co-activation profiles were calculated for each pedal cycle for VAS-GAS, GMAX-VAS, VAS-HAM, GAS-TA and GMAX-RF muscle pairs using normalised EMG profiles as per the methods and Eqn. 2 described in section 3.2.3.3. An average co-activation index value (CAI) was then calculated for each pedal cycle and each muscle pair. Average EMG profiles (GMAX, RF, GAS, TA, VAS, HAM) and CAI profiles (VAS-GAS, GMAX-VAS, VAS-HAM, GAS-TA, GMAX-RF) were created for two cadence intervals: 60-90 rpm and 160-190 rpm from the same pedal cycles used for the analysis of crank torque and kinematic profiles.

Although EMG profiles were normalised using peak amplitudes obtained pre- and post-training to enable the construction of EMG profiles, due to the potential for maximal sprint training to alter the level of activation that could be reached (i.e. peak RMS) for each of the muscles it was not appropriate to perform statistical analyses on measures of peak EMG.

Variability of crank torque, kinematic, EMG and co-activation profiles

Variance ratios (VR) were used to measure each participant's inter-cycle variability and also inter-participant variability (pre- and post-training) of the following signals: crank torque, kinematics of the hip, knee and ankle joints and EMG of the lower limb muscles. For inter-cycle variability, a VR metric was obtained for the set of seven pedal cycles within the two cadence intervals: 60-90 rpm and 160-190 rpm for each group using Eqn. 3 stated in section 3.2.3.4.

Using the same equation (Eqn. 3), inter-participant variability was calculated for each group where k is the number of intervals over the pedal cycle (i.e. 101), n is the number of participants (i.e. 9 for RES and 8 for VEL), X_{ij} is the mean EMG, crank torque or joint angle value at the i th interval for the j th participant and \bar{X}_i is the mean of the EMG, crank torque or joint angle values at the i th interval calculated over the nine or eight participants for each group.



Figure 4.4. Experimental set up for data collection, including the equipment used for mechanical, kinematic and EMG data acquisition.

4.2.4.3 Estimation of lower limb volume

Anthropometric measures were obtained from both left and right lower limbs pre and post-training to calculate total leg volume (TLV) and lean leg volume (LLV) using the previously validated method of Jones and Pearson (Jones & Pearson, 1969) This method partitions the leg into six segments (Figure 4.5). Circumferences and heights of the segments were measured using a flexible metal tape. Skinfold thickness was measured using calipers (Harpندن, Baly Int. West Sussex, UK) at the anterior and posterior thigh at one-third of subischial height and at the lateral and medial calf at maximum calf circumference. Volumes of each segment were calculated using Eqn.4.

$$V = \frac{\pi h}{3} (R^2 + r^2 + Rr)$$

Eq. 4

where V represents volume, R represents the superior radii of the segment, r represents the inferior radii of the segment and h represents the segment length. LLV was calculated using the formula above but corrected for subcutaneous fat estimated from the skinfold measurements.

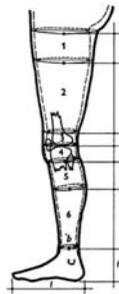


Figure 4.5. Illustration of the sites for anthropometric measurements and the six segments used to calculate lower limb volume. Taken from Jones and Pearson (1969).

4.2.5 Statistical analyses

Comparison of mean outcome variables were performed with customized spreadsheets using magnitude-based inferences and standardization to interpret the meaningfulness of the effects (Hopkins, 2006a). The within-groups differences in means (post-pre) at two sections of the power vs cadence relationship (60-90 rpm and 160-190 rpm) were analysed for the following variables: average power, peak and minimum crank torque, estimated key variables (T_0 , C_0 , P_{\max} and C_{opt}), hip, knee and ankle joint angles and range of motions, average co-activation index, variance ratio and lower limb volumes. Between-groups differences in means were assessed for average power,

crank torque and lower limb volumes. Data are presented as mean \pm standard deviation (SD) unless otherwise stated. The standardised effect was calculated as the difference in means divided by the standard deviation (SD) of the reference condition and interpreted using thresholds set at <0.2 (trivial), ≥ 0.2 (small), ≥ 0.6 (moderate), ≥ 1.2 (large), ≥ 2.0 (very large), ≥ 4.0 (extremely large) (Cohen, 1988; Hopkins et al., 2009) changes. As illustrated in Figure 3.1 (section 3.2.5) small standardised effects are highlighted in yellow, moderate in pink, large in green, very large in blue, extremely large in purple and trivial effects are indicated by no coloured band. Estimates were presented with 90% confidence intervals (\pm CI). The Likelihood that the standardised effect was substantial was assessed with non-clinical magnitude-based inference, using the following scale for interpreting the likelihoods: $\geq 25\%$, possible; $\geq 75\%$, likely; $\geq 95\%$, very likely and $\geq 99.5\%$, most likely (Hopkins et al., 2009). Symbols used to denote the likelihood of a non-trivial/true standardised effect are * possibly, ** likely, *** very likely, **** most likely. The likelihood of trivial effects are denoted by ⁰ possibly, ⁰⁰ likely, ⁰⁰⁰ very likely, ⁰⁰⁰⁰ most likely. Unclear effects (trivial or non-trivial) have no symbol. If differences were observed between groups at baseline, data sets were adjusted to the mean baseline value of the two groups combined. Comparisons of mean group data at baseline were analysed on a magnitude basis but not inferentially as per the recommendations of Hopkins (2006a).

4.3 Results

4.3.1 Effect of training on lower limb volume

RES training had a very likely trivial effect on TLV (9.3 ± 1.6 L to 9.4 ± 1.6 L; 0.04; ± 0.13) and a most likely trivial effect on LLV (8.1 ± 1.7 L to 8.2 ± 1.8 L; 0.02; ± 0.09). VEL training also had a very likely trivial effect on TLV (9.3 ± 1.7 L to 9.4 ± 1.5 L; 0.01; ± 0.12) and LLV (7.8 ± 1.7 L to 7.8 ± 1.5 L; 0.00; ± 0.11).

4.3.2 Effect of training on the limits of NMF

4.3.2.1 Effect of RES training

Following RES training, a very likely increase in power was observed at 60-90 rpm (11.5 ± 1.2 W.kg⁻¹ to 12.4 ± 1.4 W.kg⁻¹), whereas a trivial difference in power was seen at 160-190 rpm (9.4 ± 3 W.kg⁻¹ to 9.6 ± 2.9 W.kg⁻¹) (Figure 4.8). Figure 4.6 illustrates the change in T-C and P-C relationships pre- to post-training for a typical subject. The average T-C curve illustrates small to large increases in torque below 130 rpm, after training, indicating the relationship became more linear (Figure 4.6). T_0 values were most likely 0.40 ± 0.27 N·m.kg⁻¹ higher following RES training, while P_{\max} was likely 0.61 ± 0.86 W.kg⁻¹ higher. Decreases in C_{opt} and C_0 of 3 ± 5 rpm and 8 ± 21 rpm, respectively, occurred following RES training (Table 4.1).

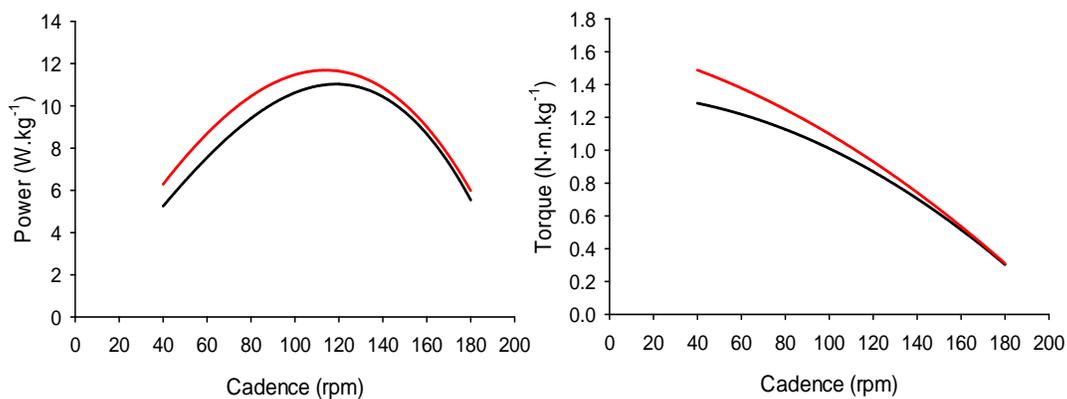


Figure 4.6. P-C and T-C relationships of a single participant before and after RES training. Black line shows pre-training relationships, red lines show post-training relationships.

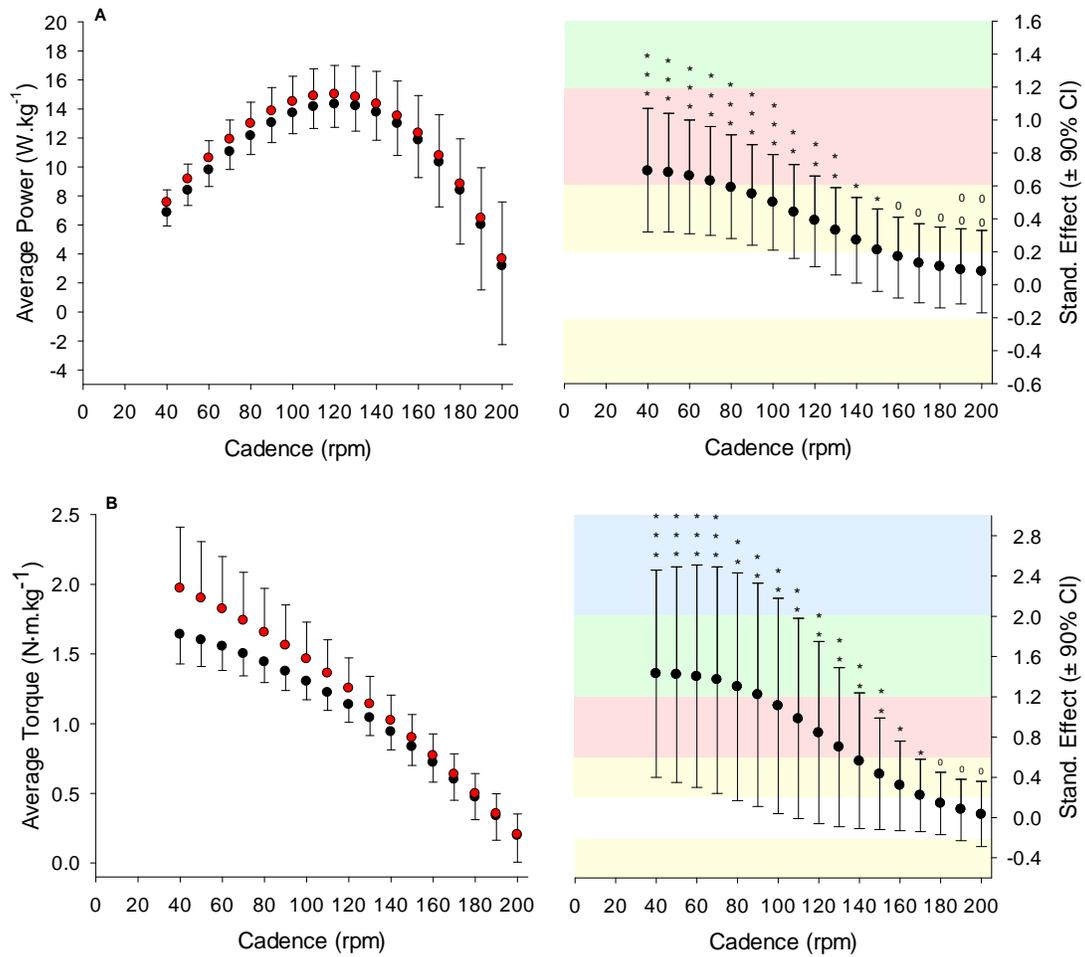


Figure 4.7. Power predicted from P-C relationships and torque predicted from T-C relationships before and after RES training. **A:** Mean \pm SD power, **B:** Mean \pm SD torque. Black points shows pre-training relationships, red points show post-training relationships. Graphs to the right illustrate the standardised effect \pm 90% CI for the Post-Pre change in power and torque produced. Likelihood of the non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely. Likelihood of the trivial standardised effect is denoted as ⁰ possibly, ⁰⁰ likely.

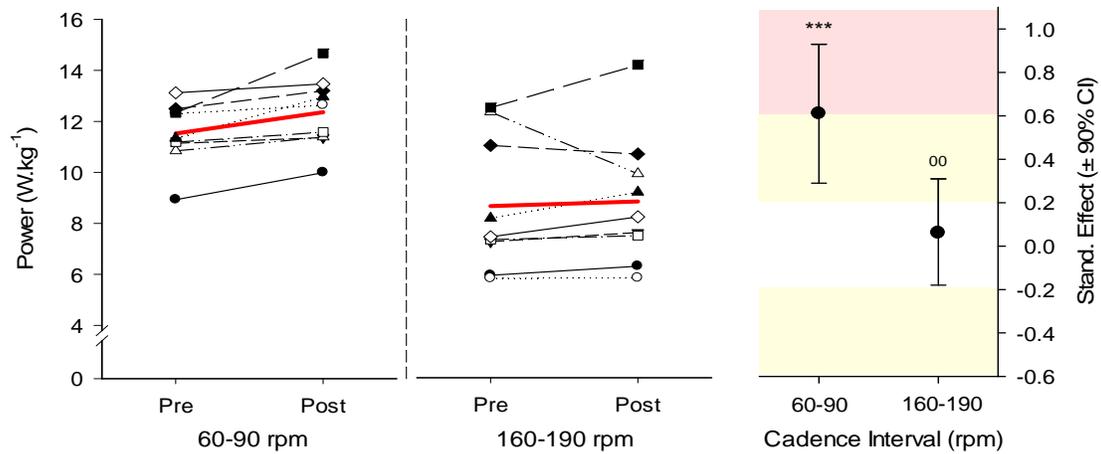


Figure 4.8. Power production at 60-90 rpm and 160-190 rpm before and after RES training. Black lines indicate individual responses to training; red line indicates mean response to training. Graph to the right illustrates the standardised effect \pm 90% CI for the Post-Pre change in power produced between 60-90 rpm and 160-190 rpm following RES training. Likelihood of the non-trivial standardised effect is denoted as *** very likely. Likelihood of the trivial standardised effect is denoted as ⁰⁰ likely.

Table 4.1. Effect of RES training on the limits of NMF estimated from P-C and T-C relationships.

	Pre	Post	Stand. Effect	Likelihood
P_{\max} (W.kg ⁻¹)	14.5 \pm 1.7	15.1 \pm 2.0	0.33; \pm 0.28	**
C_{opt} (rpm)	122 \pm 10	119 \pm 7	-0.26; \pm 0.27	*
T_0 (N·m.kg ⁻¹)	1.8 \pm 0.4	2.1 \pm 0.3	1.01; \pm 0.43	****
C_0 (rpm)	218 \pm 14	210 \pm 18	-0.50; \pm 0.84	*

Variables estimated from P-C relationship are P_{\max} (maximal power) and C_{opt} (optimal cadence). Values estimated from T-C relationships are T_0 (maximal torque) and C_0 (maximal cadence). Data presented are mean \pm SD; standardised effects are presented with \pm 90% CI. Likelihood of the non-trivial standardised effect is denoted as * possibly, ** likely or **** most likely.

4.3.2.2 Effect of VEL training

A possible increase in power production was observed at 160-190 rpm ($9.7 \pm 2.9 \text{ W.kg}^{-1}$ to $10.5 \pm 2.8 \text{ W.kg}^{-1}$; Figure 4.11). As illustrated in Figure 4.9 participant responses to the VEL training were varied at 160-190 rpm. A likely trivial difference was observed from pre-training ($11.4 \pm 1.7 \text{ W.kg}^{-1}$) to post-training ($11.3 \pm 1.4 \text{ W.kg}^{-1}$) at 60-90 rpm. Figure 4.9 illustrates the change in P-C and T-C relationships pre- to post-training for a typical subject. Evaluation of the average T-C curve for VEL revealed small increases in torque above cadences of 180 rpm post-training, indicating a reduction in the downward inflection observed prior to the training intervention (Figure 4.10). Following VEL training, likely trivial differences were observed in P_{\max} and T_0 , while a possible decrease of $4 \pm 24 \text{ rpm}$ was seen for C_0 . The most substantial change in one of these variables indicating the limits of NMF was C_{opt} with a likely increase of $3 \pm 6 \text{ rpm}$ observed post-training (Table 4.2).

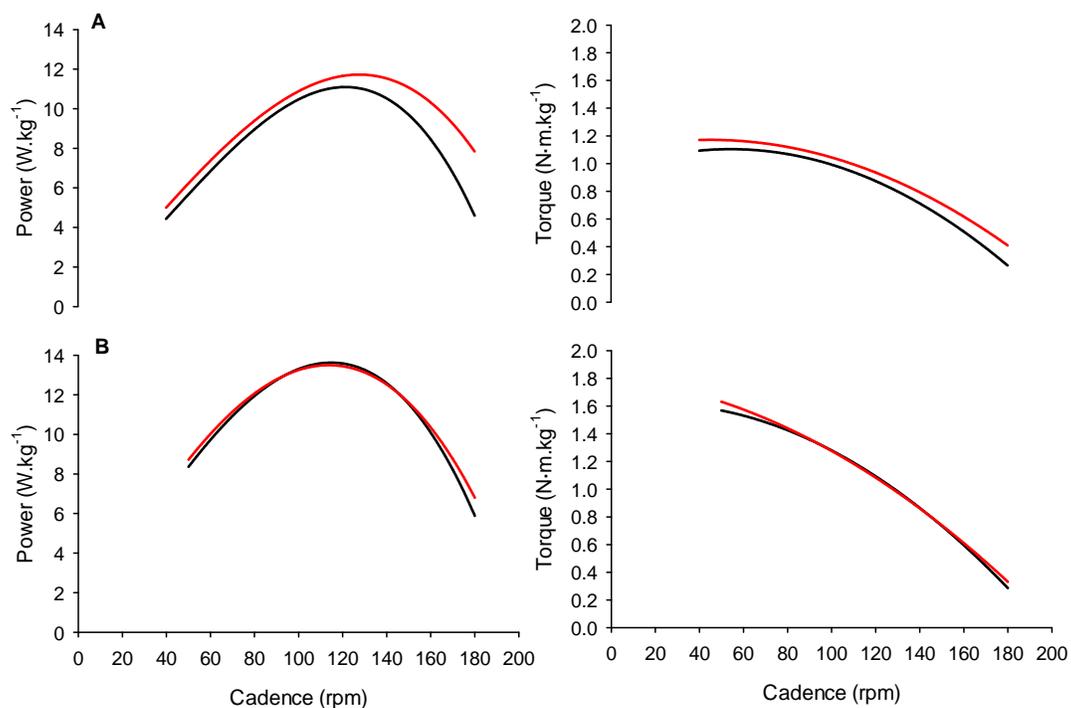


Figure 4.9. P-C and T-C relationships of two participants before and after VEL training. **A:** a participant who responded positively to VEL training, **B:** a participant that showed little response to training. Black lines show pre-training relationships, red lines show post-training relationships.

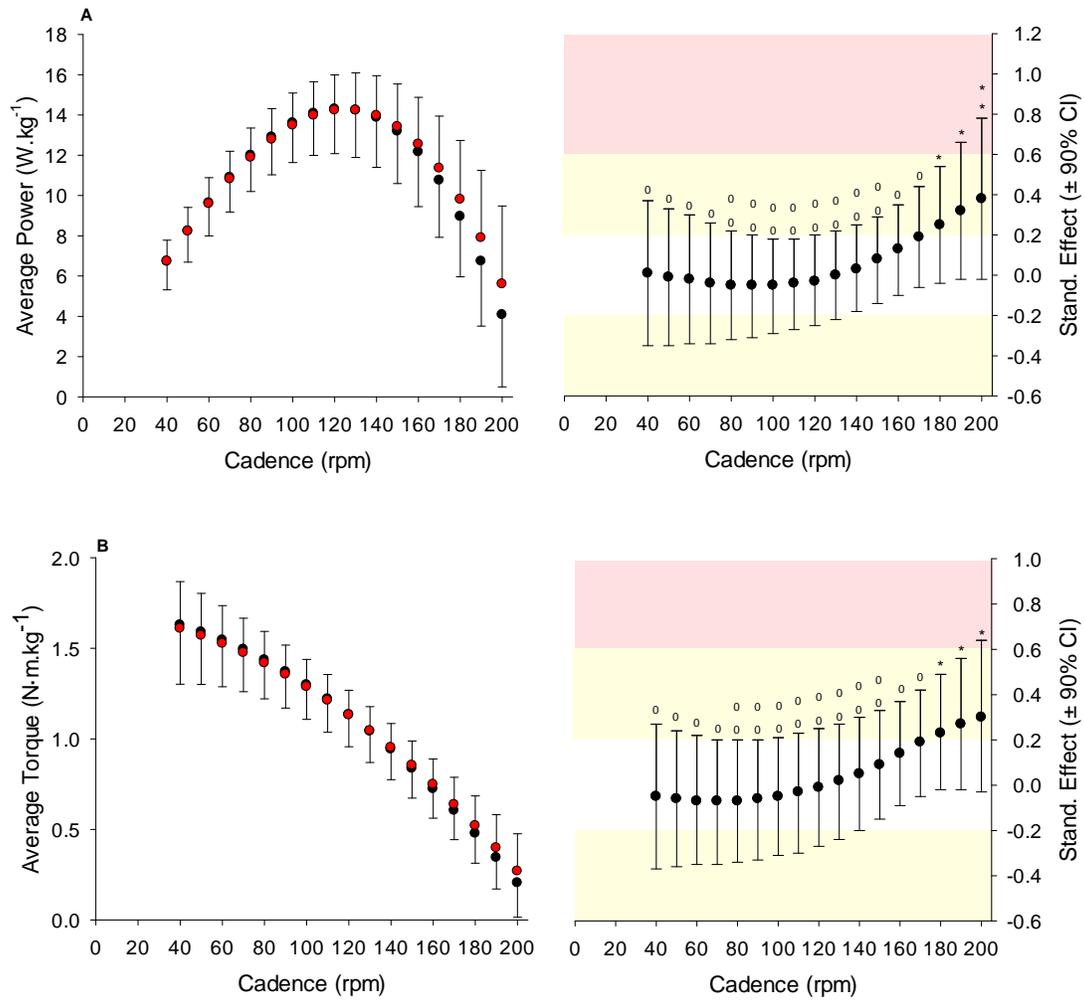


Figure 4.10. Power predicted from P-C relationships and torque predicted from T-C relationships before and after VEL training. **A:** Mean \pm SD power, **B:** Mean \pm SD torque. Black points shows pre-training relationships, red points show post-training relationships. Graphs to the right illustrate the standardised effect \pm 90% CI for the Post-Pre change in power and torque produced. Likelihood of the non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely. Likelihood of the trivial standardised effect is denoted as ⁰ possibly, ⁰⁰ likely.

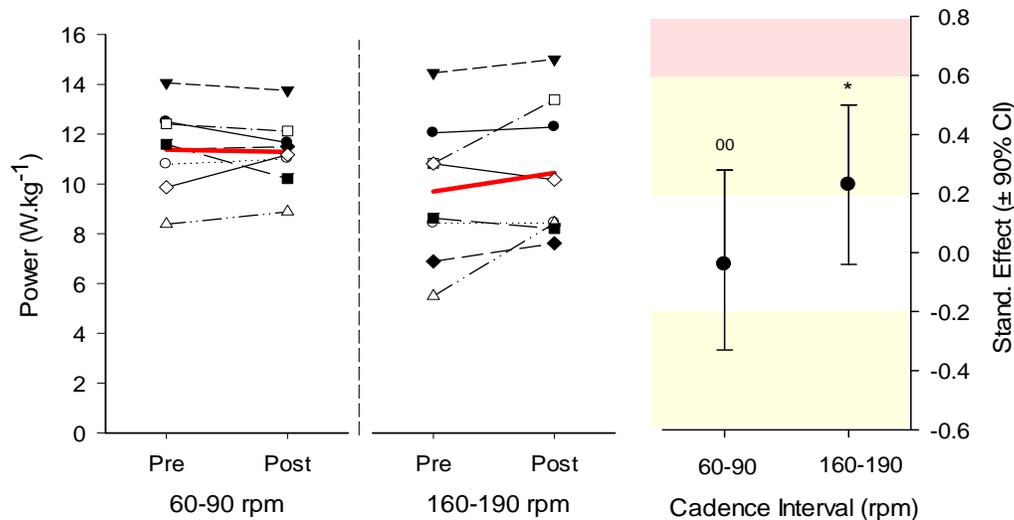


Figure 4.11. Power production at 60-90 rpm and 160-190 rpm before and after VEL training. Black lines indicate individual responses to training; red line indicates mean response to training. Graph to the right illustrates the standardised effect \pm 90% CI for the Post-Pre change in power produced between 60-90 rpm and 160-190 rpm following VEL training. Likelihood of a non-trivial standardised effect is denoted as * possibly. Likelihood of a trivial standardised effect is denoted as ⁰⁰ likely.

Table 4.2. Effect of VEL training on the limits of NMF estimated from P-C and T-C relationships.

	Pre	Post	Stand. Effect	Likelihood
P_{\max} (W.kg ⁻¹)	14.4 \pm 2.3	14.3 \pm 1.8	-0.02; \pm 0.22	⁰⁰
C_{opt} (rpm)	122 \pm 7	126 \pm 7	0.40; \pm 0.48	**
T_0 (N·m.kg ⁻¹)	1.7 \pm 0.5	1.8 \pm 0.3	0.18; \pm 0.55	0
C_0 (rpm)	217 \pm 17	213 \pm 35	-0.20; \pm 0.85	*

Variables estimated from P-C relationship are P_{\max} (maximal power) and C_{opt} (optimal cadence). Values estimated from T-C relationships are T_0 (maximal torque) and C_0 (maximal cadence). Data presented are mean \pm SD; standardized effect are presented with \pm 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly or ⁰⁰ likely.

4.3.3 Effect of training on crank torque, kinematic and EMG profiles

4.3.3.1 Crank torque profiles

Following RES training, a likely increase in peak crank torque (2.30 \pm 0.21 N·m.kg⁻¹ to 2.55 \pm 0.40 N·m.kg⁻¹), and a likely decrease in minimum crank torque (0.60 \pm 0.12 N·m.kg⁻¹ to 0.55 \pm 0.15 N·m.kg⁻¹) were observed after RES training (Figure 4.12).

Following VEL training, a small reduction in minimum crank torque (0.49 \pm 0.10 N·m.kg⁻¹ to 0.43 \pm 0.13 N·m.kg⁻¹) and peak crank torque (0.96 \pm 0.14 N·m.kg⁻¹ to 0.91 \pm 0.13 N·m.kg⁻¹) was observed at 160-190 rpm following VEL training (Figure 4.13). Peak crank torque occurred

later in the pedal cycle ($33 \pm 9\%$ to $39 \pm 3\%$; $1.71; \pm 2.53$) and minimum crank torque occurred earlier in the pedal cycle ($16 \pm 4\%$ to $14 \pm 6\%$; $0.54; \pm 1.07$) after VEL training.

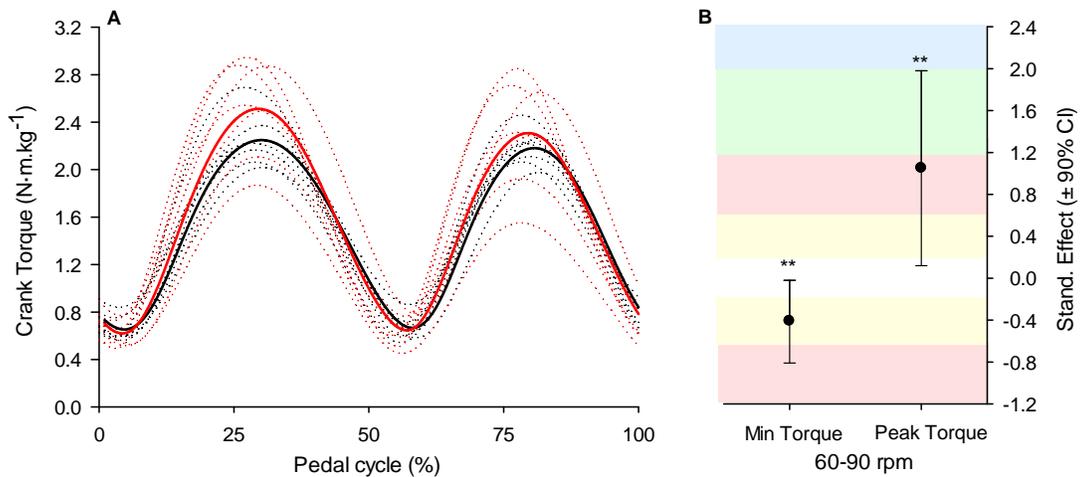


Figure 4.12. Crank torque profiles before and after RES training at 60-90 rpm. **A:** Mean crank torque pre- (solid black line) post- (solid red line) training. Dotted lines indicate individual responses, **B:** standardised effect \pm 90% CI for the change in minimum and peak crank torque produced between 60-90 rpm following RES training (B). Likelihood of the non-trivial standardised effect is denoted as ** likely.

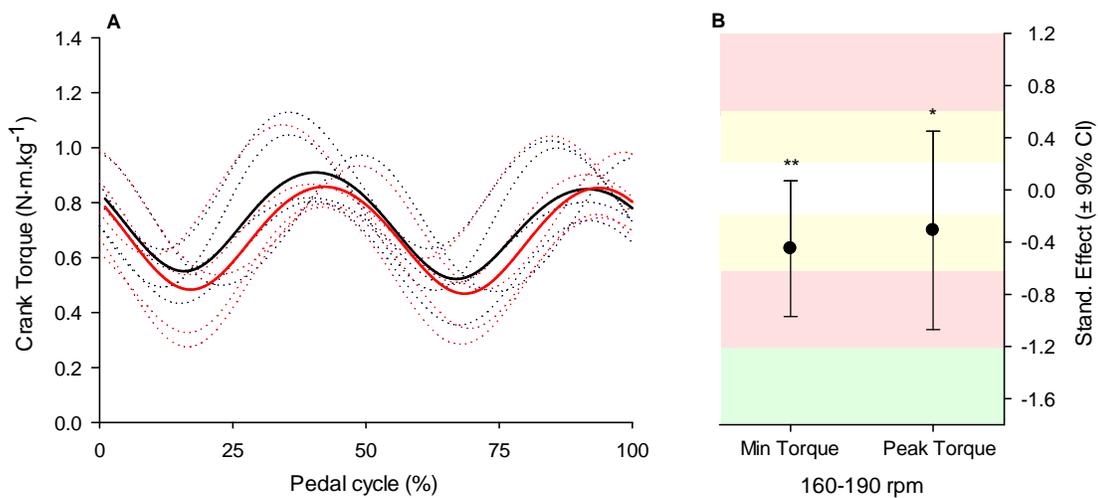


Figure 4.13. Crank torque profiles before and after VEL training at 160-190 rpm. **A:** Mean crank torque pre- (solid black line) post- (solid red line) training, **B:** standardised effect \pm 90% CI for the change in minimum and maximum crank torque produced between 160-190 rpm following VEL training (B). Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely.

4.3.3.2 Kinematic profiles

Following RES training, a likely increase in hip ROM was observed at 60-90 rpm ($43 \pm 3^\circ$ to $45 \pm 3^\circ$) and a possible increase in maximal hip flexion angle ($80 \pm 9^\circ$ to $82 \pm 11^\circ$) (Figure 4.14A). Maximal knee flexion angle increased ($101 \pm 4^\circ$ to $104 \pm 5^\circ$) (Figure 4.14B). A very likely reduction in ankle joint ROM was observed at 60-90 rpm following RES training ($52 \pm 7^\circ$ to $46 \pm 7^\circ$), which appeared to result from a higher maximal plantar-flexion angle between 50-75% of the pedal cycle ($44 \pm 7^\circ$ to $49 \pm 5^\circ$) (Figure 4.14C).

Following VEL training, it was likely that the maximal dorsi-flexion angle of the ankle was reduced ($80 \pm 6^\circ$ to $76 \pm 11^\circ$) between 160-190 rpm, but this did not result in a substantial change in ankle ROM (Figure 4.15C). At this cadence range, a possible increase in hip ($50 \pm 3^\circ$ to $51 \pm 4^\circ$) and knee ($77 \pm 4^\circ$ to $78 \pm 6^\circ$) joint ROM was observed (Figure 4.15A and B).

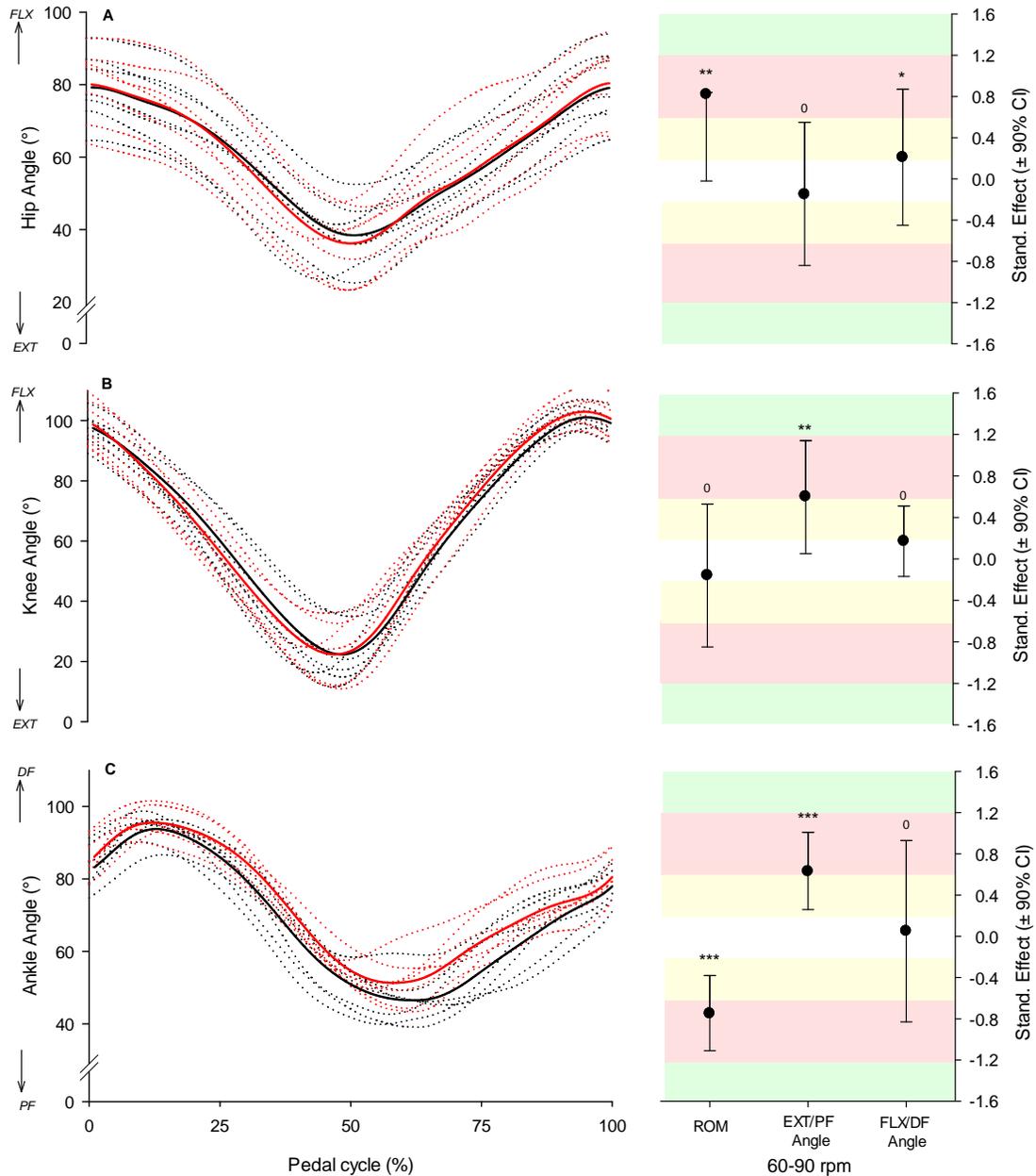


Figure 4.14. Joint angle profiles before and after RES training for 60-90 rpm. **A:** hip joint, **B:** knee joint, **C:** ankle joint. Solid lines indicate mean pre- (black) post- (red) training response. Dotted lines indicate individual responses. EXT and PF on graph axes indicate that the joint is moving into extension or plantar-flexion, while FLX and DF indicate that the joint is moving into flexion or dorsi-flexion. Graphs to the right of the joint angle profiles illustrate the standardised effect \pm 90% CI for the change in ROM and flexion (FLX)/dorsiflexion (DF), extension (EXT) /plantar-flexion (PF) angles produced between 60-90 rpm following RES training. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely or **** most likely. Likelihood of a trivial standardised effect is denoted as 0 possibly.

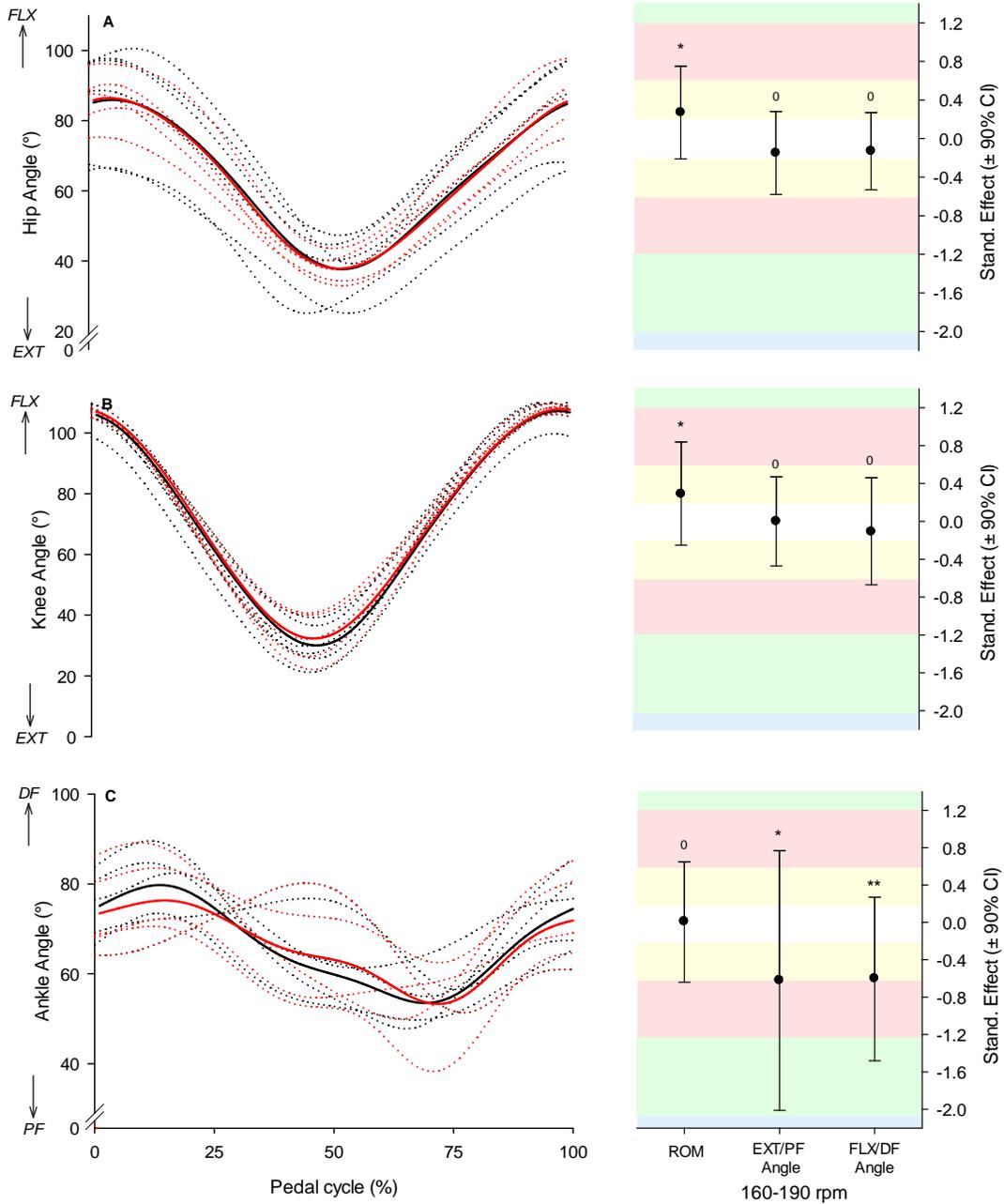


Figure 4.15. Joint angle profiles before and after VEL training for 160-190 rpm. **A:** hip joint, **B:** knee joint, **C:** ankle joint. Solid lines indicate mean pre- (black) post- (red) training response. Dotted lines indicate individual responses. EXT and PF on graph axes indicate that the joint is moving into extension or plantar-flexion, while FLX and DF indicate that the joint is moving into flexion or dorsi-flexion. Graphs to the right of the joint angle profiles illustrate the standardised effect ($\pm 90\%$ CI) for the change in ROM and flexion (FLX)/dorsi-flexion (DF), extension (EXT) /plantar-flexion (PF) angles produced between 160-190 rpm following VEL training. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly.

4.3.3.3 EMG and CAI profiles

Individual and mean EMG signals before and after RES and VEL training have been illustrated in Figure 4.16 and Figure 4.17 respectively. However, due to all-out sprint training potentially increasing the level of activation that could be reached (i.e. peak RMS) following training, it was not appropriate to report and compare EMG amplitude changes on measures of peak EMG pre- and post-training. It was possible to report changes in average co-activation index (CAI) values.

Following RES training, average CAI was likely lower for VAS-GAS muscle pair (27 ± 2 a.u. to 24 ± 5 a.u.) and possibly lower for GMAX-GAS (44 ± 7 a.u. to 42 ± 7 a.u.) at 60-90 rpm, while a very likely increase was observed for VAS-HAM (36 ± 4 a.u. to 41 ± 8 a.u.) and possible increases for GMAX-RF (32 ± 6 a.u. to 36 ± 12 a.u.) and GAS-TA (23 ± 6 a.u. to 25 ± 7 a.u.) muscle pairs, as shown in Figure 4.18.

Following VEL, training a likely lower average CAI values for GMAX-RF muscle pair (46 ± 11 a.u. to 39 ± 8 a.u.) at 160-190 rpm, while possible increases were observed for GMAX-GAS (29 ± 4 a.u. to 32 ± 6 a.u.) and GAS-TA (25 ± 5 a.u. to 27 ± 9 a.u.) (Figure 4.19).

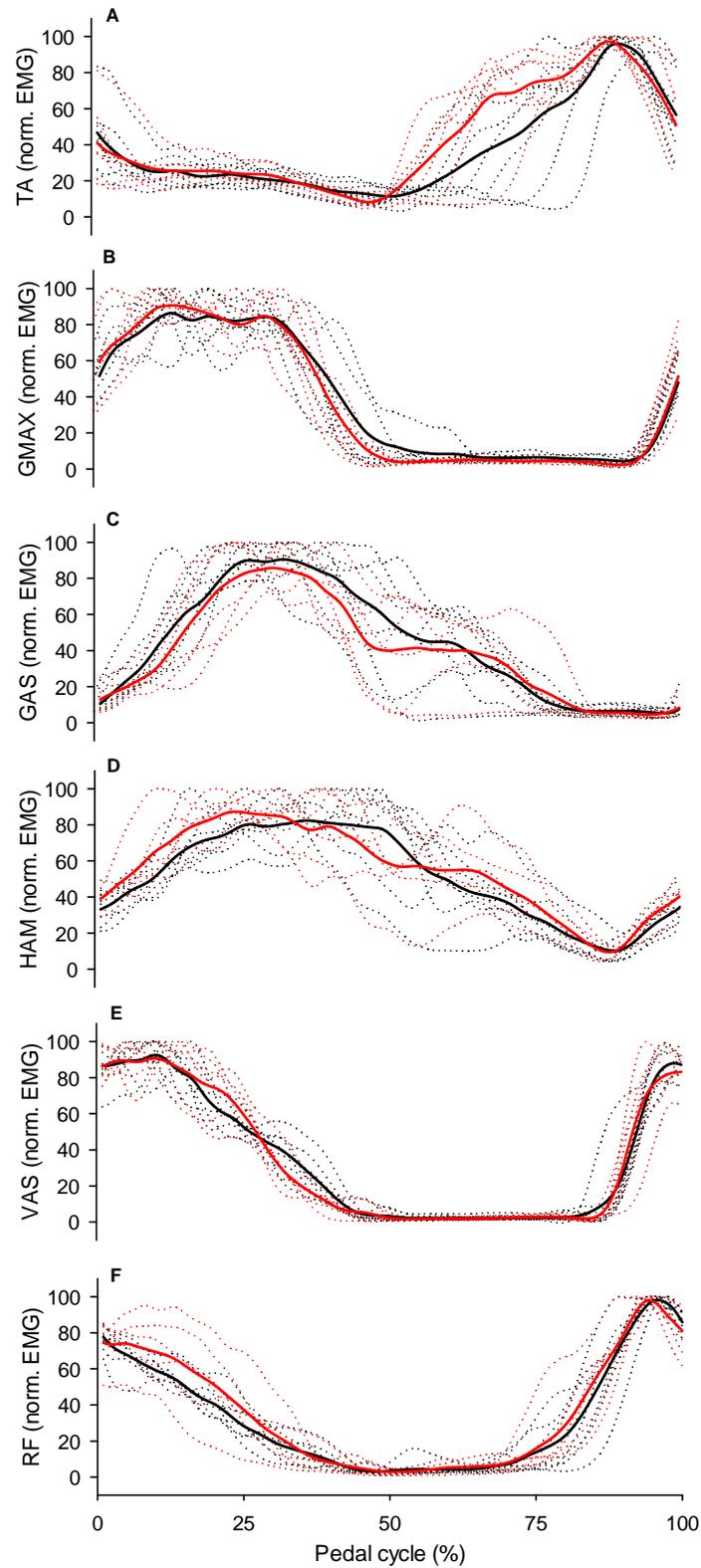


Figure 4.16. EMG profiles before and after RES training at 60-90 rpm. **A:** TA, **B:** GMAX, **C:** GAS, **D:** HAM, **E:** VAS and **F:** RF. Solid lines indicate mean pre- (black) post- (red) training response. Dotted lines indicate individual responses.

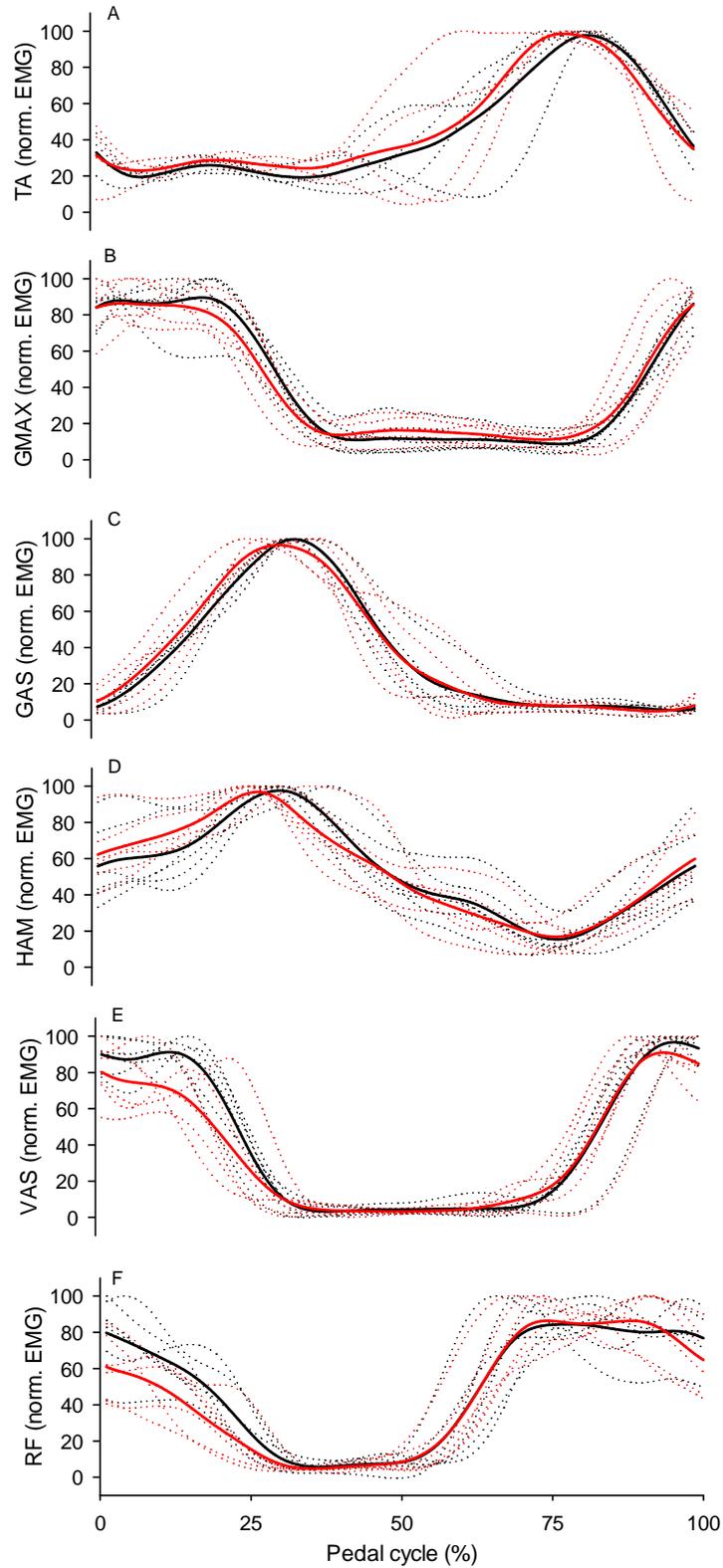


Figure 4.17. EMG profiles before and after VEL training at 160-190 rpm. **A:** TA, **B:** GMAX, **C:** GAS, **D:** HAM, **E:** VAS and **F:** RF. Solid lines indicate mean pre- (black) post- (red) training response. Dotted lines indicate individual responses.

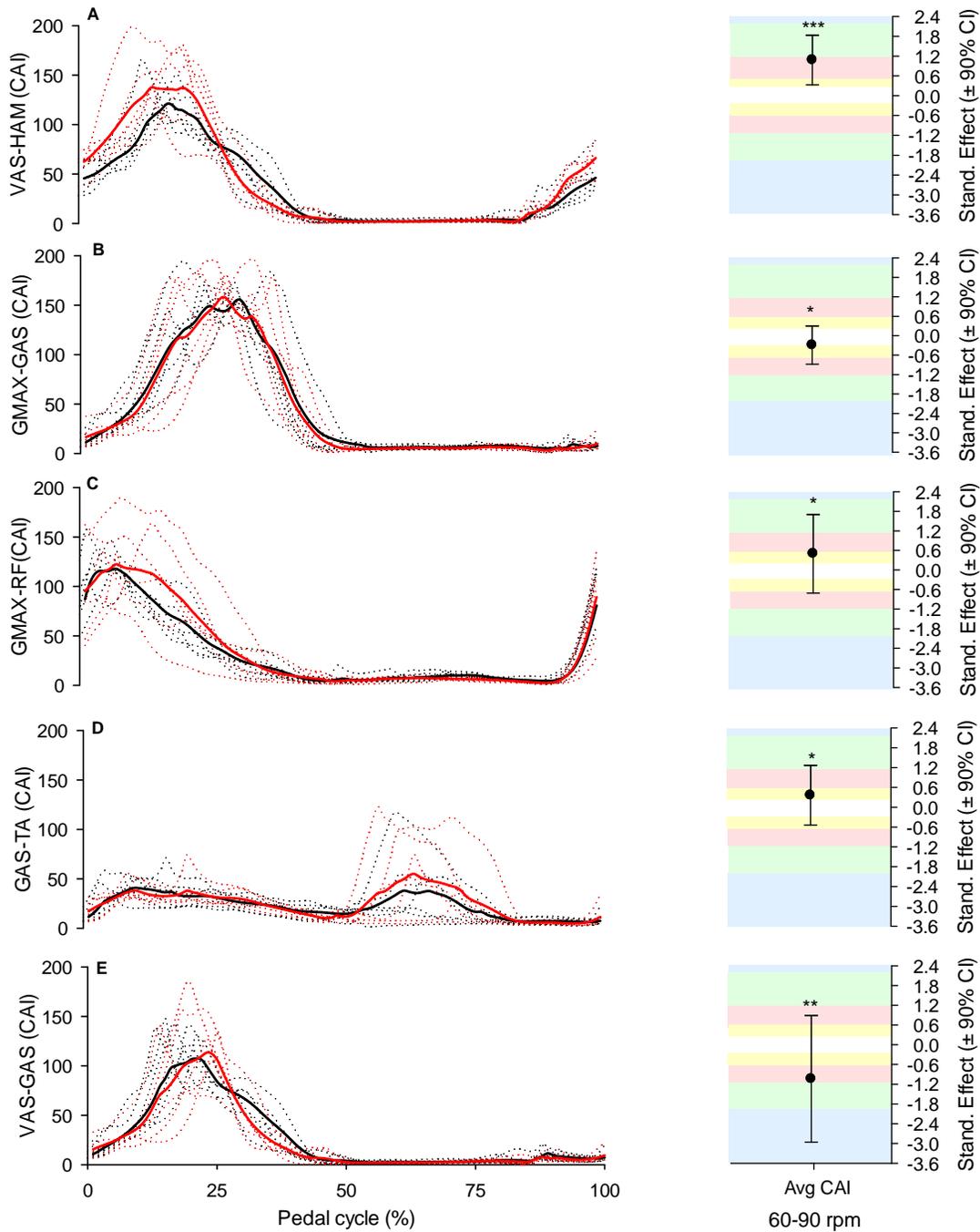


Figure 4.18. CAI profiles before and after RES training at 60-90 rpm. **A:** VAS-HAM, **B:** GMAX-GAS, **C:** GMAX-RF, **D:** GAS-TA and **E:** VAS-GAS. Solid lines indicate mean pre- (black) post- (red) training response. Dotted lines indicate individual responses (A). Graphs to the right of the CAI profiles illustrate the standardised effects \pm 90% CI for the change in average CAI for the various muscle pairs between 60-90 rpm following RES training. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely or *** very likely.

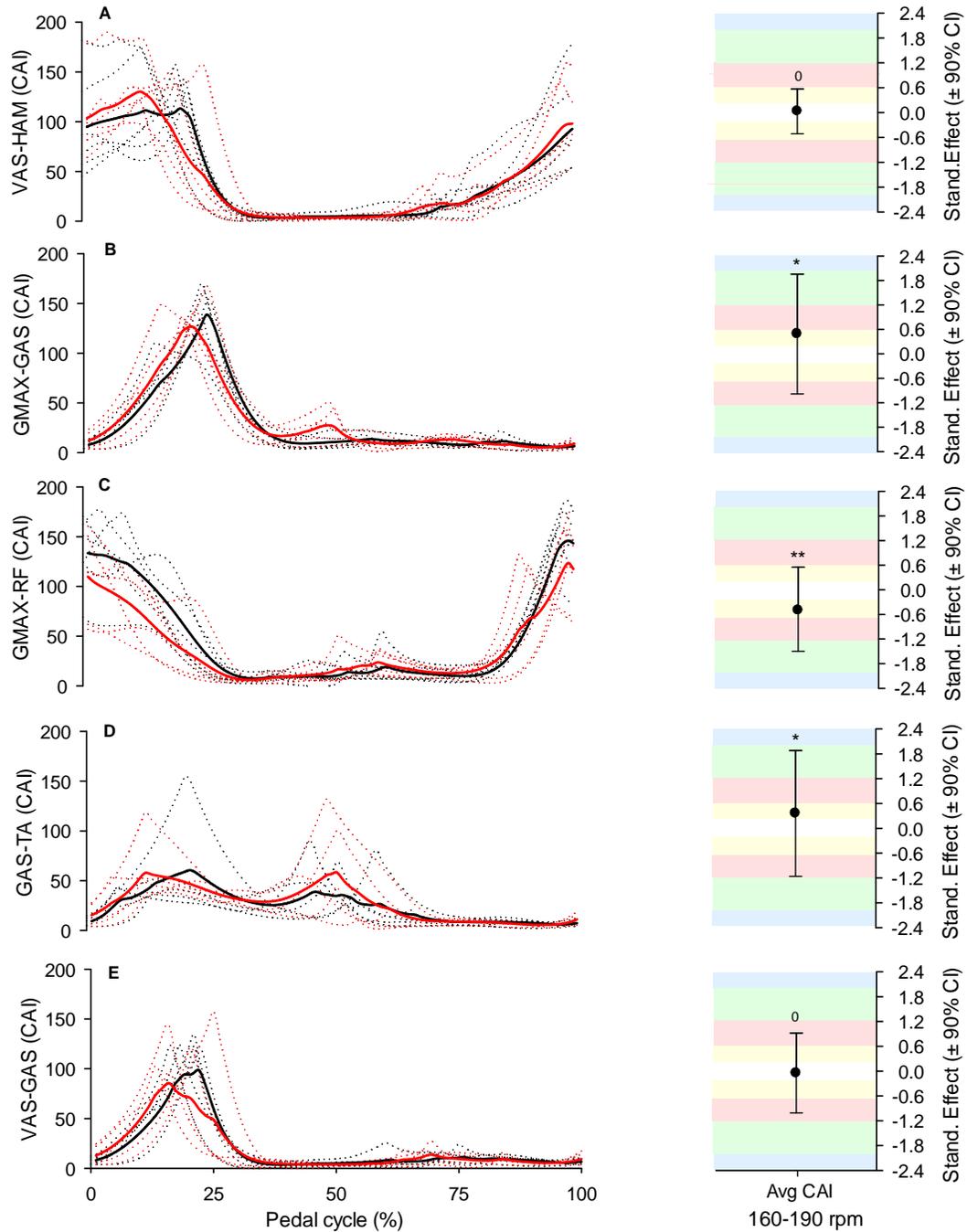


Figure 4.19. CAI profiles before and after VEL training at 160-190 rpm. A: VAS-HAM, B: GMAX-GAS, C: GMAX-RF, D: GAS-TA and E: VAS-GAS. Solid lines indicate mean pre- (black) post- (red) training response. Dotted lines indicate individual responses. Graphs to the right of the CAI profiles illustrate the standardised effects \pm 90% CI for the change in average CAI for each muscle pair at 160-190 rpm following VEL training. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely. Likelihood of a trivial standardised effect is denoted as 0 possibly.

4.3.4 Effect of training on variability of crank torque, kinematic and EMG profiles

4.3.4.1 Inter-cycle variability

Following RES training, clear differences were observed for hip, knee and ankle joint profile VR with all reduced post-RES training at 60-90 rpm. At this same cadence interval, a reduction in VR was observed for GMAX, while increases were seen for TA, RF and HAM. With regards to inter-cycle VR values for CAI profiles, reductions were observed for all muscle pairs: GMAX-GAS, GMAX-RF, VAS-HAM and VAS-GAS at 60-90 rpm, except for an unclear change seen for GAS-TA. All VR values and magnitudes of change can be found in Table 4.3.

Following VEL training, as outlined in Table 4.4, hip, knee and ankle joint profile VR increased by moderate, large and small magnitudes respectively. Assessment of VR for individual muscles revealed likely increases for GAS, TA, HAM and possible increases for GMAX, and VAS. With all muscles combined a likely small increase in VR was observed for VEL at 160-190 rpm. VEL training led to possible reductions in VR for GAS-TA, VAS-GAS, VAS-HAM and a likely reduction for GMAX-RF muscle pairs. In contrast a possible increase in VR was observed for GMAX-GAS muscle pairs.

Table 4.3. Inter-cycle VR for crank torque, joint angle, EMG and CAI, before and after RES training at 60-90 rpm.

	Pre	Post	Stand. Effect	Likelihood
Crank torque	0.10 ± 0.06	0.11 ± 0.08	0.07; ±0.54	unclear
Hip joint	0.08 ± 0.03	0.05 ± 0.04	-0.80; ±1.35	**
Knee joint	0.03 ± 0.02	0.01 ± 0.01	-0.97; ±0.72	***
Ankle joint	0.13 ± 0.07	0.07 ± 0.03	-0.67; ±0.59	**
All joints	0.08 ± 0.06	0.05 ± 0.04	-0.53; ±0.32	***
GMAX	0.31 ± 0.08	0.27 ± 0.08	-0.44; ±0.39	**
GAS	0.24 ± 0.08	0.23 ± 0.07	-0.14; ±1.05	0
RF	0.19 ± 0.03	0.25 ± 0.07	1.43; ±1.67	**
TA	0.20 ± 0.09	0.23 ± 0.08	0.24; ±1.07	*
VAS	0.20 ± 0.06	0.20 ± 0.04	-0.03; ±0.97	0
HAM	0.27 ± 0.05	0.32 ± 0.12	0.78; ±1.48	**
All muscles	0.24 ± 0.08	0.25 ± 0.08	0.17; ±0.36	0
GMAX-GAS	0.31 ± 0.05	0.25 ± 0.07	-0.94; ±1.02	**
GMAX-RF	0.28 ± 0.04	0.25 ± 0.13	-0.59; ±2.00	*
VAS-HAM	0.30 ± 0.09	0.19 ± 0.06	-1.02; ±0.95	**
VAS-GAS	0.29 ± 0.08	0.24 ± 0.08	-0.61; ±0.89	**
GAS-TA	0.35 ± 0.12	0.34 ± 0.17	-0.04; ±0.86	unclear
All pairs	0.31 ± 0.08	0.26 ± 0.11	-0.63; ±0.43	***

Data presented are mean ± SD; standardized effect are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely or *** very likely. Likelihood of a trivial standardised effect is denoted as 0 possibly

Table 4.4. Inter-cycle VR for crank torque, joint angle, EMG and CAI, before and after VEL training at 160-190 rpm.

	Pre	Post	Stand. Effect	Likelihood
Crank torque	0.43 ± 0.14	0.42 ± 0.08	-0.02; ±1.12	⁰
Hip joint	0.02 ± 0.02	0.04 ± 0.03	0.60; ±0.93	**
Knee joint	0.01 ± 0.00	0.02 ± 0.02	1.73; ±2.45	**
Ankle joint	0.24 ± 0.14	0.34 ± 0.24	0.57; ±1.29	*
All joints	0.09 ± 0.13	0.13 ± 0.20	0.29; ±0.43	*
GMAX	0.25 ± 0.06	0.31 ± 0.18	0.89; ±2.48	*
GAS	0.09 ± 0.04	0.15 ± 0.09	1.39; ±1.63	**
RF	0.24 ± 0.07	0.24 ± 0.14	-0.02; ±1.66	⁰
TA	0.25 ± 0.09	0.37 ± 0.23	1.04; ±2.04	**
VAS	0.17 ± 0.07	0.20 ± 0.09	0.35; ±1.42	*
HAM	0.28 ± 0.10	0.35 ± 0.16	0.60; ±0.52	**
All muscles	0.21 ± 0.09	0.27 ± 0.16	0.54; ±0.73	**
GMAX-GAS	0.21 ± 0.09	0.24 ± 0.17	0.23; ±1.53	*
GMAX-RF	0.30 ± 0.03	0.24 ± 0.10	-1.64; ±2.02	**
VAS-HAM	0.30 ± 0.19	0.21 ± 0.11	-0.36; ±0.96	*
VAS-GAS	0.26 ± 0.13	0.20 ± 0.14	-0.40; ±1.18	*
GAS-TA	0.34 ± 0.09	0.29 ± 0.20	-0.42; ±2.04	*
All pairs	0.28 ± 0.12	0.23 ± 0.14	-0.37; ±1.79	*

Data presented are mean ± SD; standardized effect are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly.

4.3.4.2 Inter-participant variability

Variance ratios were calculated to assess inter-participant variability. Due to its method of calculation a single value is generated for all participants, hence comment on the direction of change (i.e. an increase/decrease) could be made pre- to post-training, however statistical comparisons could not be performed on the change. After four weeks of RES training, crank torque VR increased, although little change was observed in VR for all joints and all muscles at 60-90 rpm. An increase in VR was seen for CAI of all muscle pairs combined and individually (Table 4.5).

Those training in VEL showed little change in crank torque VR at 160-190 rpm post-training as illustrated in Table 4.6. All joints combined, little change in inter-participant was observed for VEL, but individually a reduction was seen for hip joint angle VR, while an increase was seen for ankle joint angle VR. Increases in VR were observed for all muscles combined and all muscle pairs combined, though individually, reductions were observed in RF, HAM, VAS-HAM and GAS-TA (Table 4.6).

Table 4.5. Inter-participant VR for crank torque, joint angle, EMG and CAI, before and after RES training at 60-90 rpm.

	Pre	Post	Post-Pre % diff
Crank torque	0.07	0.22	214
Hip joint	0.37	0.38	3
Knee joint	0.07	0.08	14
Ankle joint	0.14	0.12	-14
GMAX	0.09	0.09	0
GAS	0.25	0.32	28
RF	0.09	0.17	89
TA	0.35	0.24	-31
VAS	0.04	0.07	75
HAM	0.35	0.34	-3
GMAX-GAS	0.14	0.19	36
GMAX-RF	0.09	0.13	44
VAS-HAM	0.11	0.14	27
VAS-GAS	0.14	0.23	64
GAS-TA	0.76	0.78	3

Data are presented as means. SD cannot be calculated for this variable. Variables highlighted in orange indicate a reduction in VR from pre- to post-training, while those highlighted in grey indicate an increase.

Table 4.6. Inter-participant VR for crank torque, joint angle, EMG and CAI, before and after VEL training at 160-190 rpm.

	Pre	Post	Post-Pre % diff
Crank torque	0.64	0.65	2
Hip joint	0.40	0.15	-63
Knee joint	0.02	0.03	50
Ankle joint	0.31	0.58	87
GMAX	0.08	0.21	163
GAS	0.07	0.12	71
RF	0.20	0.17	-15
TA	0.37	0.41	11
VAS	0.06	0.17	183
HAM	0.28	0.23	-18
GMAX-GAS	0.11	0.20	82
GMAX-RF	0.14	0.32	129
VAS-HAM	0.30	0.27	-10
VAS-GAS	0.18	0.26	44
GAS-TA	0.71	0.68	-4

Data presented are means. SD cannot be calculated for this variable. Variables highlighted in orange indicate a reduction in VR from pre- to post-training, while those highlighted in grey indicate an increase.

4.4 Discussion

The first aim of this study was to investigate if the adaptations of the limits of NMF would be specific to the training intervention selected. The results show that RES training improved the limits of NMF on the left side of the P-C relationship as revealed by the moderate increases in power production seen at 60-90 rpm ($+7 \pm 6\%$) and T_0 ($+25 \pm 19\%$). There was a small increase in P_{\max} for this group that was associated to small reductions in C_{opt} . On the right side of the curve, trivial changes in power were seen at 160-190 rpm, while C_0 was reduced by a small magnitude (-3 ± 9 rpm). VEL training led to changes on the right side of the curve as revealed by a small increase in power at 160-190 rpm ($+10 \pm 20\%$), and C_{opt} ($+3 \pm 6$ rpm). Surprisingly, C_0 was reduced following VEL training (-2 ± 11 rpm). Trivial effects on power produced at 60-90 rpm were also observed for this group.

The second aim of this study was to investigate if different motor control adaptations would accompany the change in the limits of NMF. For RES, the increase in power was linked to an increase in peak crank torque ($+11 \pm 13\%$), while adaptations at the ankle included a reduction in joint range of motion that was associated with a small increase in co-activation of GAS-TA muscle pair. Also, average VAS-HAM co-activation was greater, while moderate and small reductions were seen for VAS-GAS and GMAX-GAS respectively. Additionally, movement variability was reduced between cycles for all joints and muscle pairs. The adaptations that accompanied the increase in power following VEL training included a more plantar-flexed position of the ankle over the pedal cycle, and an associated increase in GAS-TA co-activation. In association an increase in range of motion of the proximal joints was observed, while GMAX-RF co-activation was reduced. As opposed to RES, inter-cycle movement variability increased for all joints and most muscles.

The collection of findings above confirm the first assumption that different ballistic training interventions would result in different adaptations of the limits of NMF with the greatest gains seen for exercise conditions that were used during training. This study was the first to show that the specific limits of NMF within the P-C and T-C relationships could be changed using specific sprint cycling interventions. Further, in response to the second aim, it was found that the increase in power production observed for RES was associated with motor control adaptations that were different to the ones accompanying the increase in power for VEL.

4.4.1 The effect of RES training on the limits of NMF and associated adaptations

The intervention-specific increase in power we observed at 60-90 rpm (Figure 4.8) was similar to those previously reported following a period of practice and training in both non-trained and trained

cyclists, though consideration should be given to the fact that these authors assessed changes in P_{\max} (Creer et al., 2004; Martin et al., 2000a). The trivial pre- to post-training changes in power produced at 160-190 rpm for RES further highlights that the changes in the limits of NMF were training intervention specific, in line with previous reports from single and multi-joint exercise training that power improvements are specific to sections of the F-V at which it is trained (Kaneko et al., 1983; McBride et al., 2002). As illustrated in Figure 4.10B, the inflection observed on the left side of the T-C (i.e. below 100 rpm) was reduced following training, with the relationship exhibiting a shape that was closer to linear, similar to that observed in competitive cyclists (Capmal & Vandewalle, 1997; Dorel et al., 2005). The reduction in C_{opt} , suggests a left-ward shift of the P-C curve towards lower cadences, like those at which training was performed. As the reductions in C_{opt} (-3 ± 5 rpm) and C_0 (-8 ± 21 rpm) were not even, a narrowing of the right side of the P-C relationship resulted, indicating that participants in this group were not able to produce power for the same range of cadences.

For RES, the improvement on the left side of the P-C relationship included a substantial increase in peak crank torque. This change could be due to an increase in torque produced during the downstroke and/or reduced negative torque (i.e. less negative work produced by the contralateral muscles) during the upstroke (Figure 4.12). Of the lower limb joints assessed the ankle displayed the greatest alterations in range of motion following RES training with an average reduction of $6 \pm 4^\circ$ (Figure 4.14). This change resulted from the adoption of a more dorsi-flexed position of the ankle over the full pedal cycle. These changes on the ankle joint kinematics are probably due to the increased co-activation seen for the ankle agonist-antagonist GAS-TA muscle pair. The adoption of a more dorsi-flexed position of the ankle seems to have been compensated by an increase in hip range of motion illustrated in Figure 4.14. Interestingly, this change was accompanied by a moderate increase in VAS-HAM co-activation (Figure 4.18), which may have led to an increased transfer of knee extension power to hip extension power (van Ingen Schenau, 1989; Van Ingen Schenau et al., 1995). The reduced co-activation of VAS-GAS and GMAX-GAS ($-5 \pm 14\%$ and $-8 \pm 19\%$, respectively) suggest that participants adopted an inter-muscular coordination less oriented towards the transfer of hip and knee extension powers via the ankle plantar-flexors (Figure 4.18). The EMG profiles of the different lower limb muscles (Figure 4.16 and Figure 4.17) were typical for those previously illustrated in maximal cycling (Dorel et al., 2012; Rouffet & Hautier, 2008) as were the values of average co-activation (O'Bryan et al., 2014). However, due to issues with EMG normalisation it was not possible to ascertain if neural drive to the muscles changed, even if this change is likely, based on previous research (Creer et al., 2004; Enoka, 1997; Hakkinen et al., 1985).

The changes in kinematics and inter-muscular coordination observed for RES, were associated with small to moderate reductions in inter-cycle variability, suggesting that after training

each participant adopted movement strategies that were optimal for producing power at low to moderate cadences. Indeed, less variable movement patterns are said to be an indicator of movement control occurring with learning of a new task, which is of relevance for the un-trained cyclists recruited for this study (Muller & Sternad, 2009). As inter-participant variability appeared relatively unchanged for RES, it appears that participants did not adopt similar movement strategies when receiving the same training stimulus (Table 4.5). The reduction in the inter-cycle variability for all muscle pairs, except GAS-TA, suggests that participants learnt how to co-activate their ankle joint muscles to change the ankle joint kinematics, which seems to be the major kinematic change and might be linked to the increase in power seen on the left side of the P-C curve. Additionally, it is important to note that the limits of NMF were increased in absence of a greater lean muscle mass, suggesting that the changes observed for this group were not due to modifications in muscle morphology (i.e. size or cross-sectional area).

4.4.2 The effect of VEL training on the limits of NMF and associated adaptations

Following VEL training, an increase of the limits of NMF was seen on the right side of the P-C curve, but interestingly this was not inclusive of C_0 . On average, there was a small increase in the power produced on the right side of the curve (i.e. 160-190 rpm), although the individual responses to the training intervention were highly variable, ranging from a 53% improvement to a 6% decrease in power production on the right side of the curve (Figure 4.9). The increase in C_{opt} and interestingly the concomitant reduction in C_0 resulted in a narrowing of the right side of the P-C relationship post-training, indicating that participants could not maintain power production for the same range of cadences compared to baseline. Although, it was surprising that those in VEL did not increase C_0 following training, especially as the difference between the maximal cadences of these participants at baseline and the highest cadence at which they trained was only ~ 7 rpm. Considering the very short cycle time observed at C_0 (i.e. 282 ms) activation-deactivation dynamics (i.e. delay between muscle force development and relaxation) may have limited participants ability to produce power at maximal cadences (Samozino et al., 2007), especially if it is presumed that the muscles were activated to a higher level after training. With this in mind, the effect of activation-deactivation dynamics may have also affected C_0 values for RES, especially as the participants in this group did not train at cadences near maximal. Although, anthropometric assessment indicated that lean lower limb volume did not change with training, a change in muscle fiber type distribution cannot be discounted as sprint cycling training has previously shown to change the proportions of type I and type II muscle fibers in the vastii muscles (Linossier et al., 1993). However, this change in fiber type proportions were associated with an increase in C_0 (~ 27 rpm), which was in contrast to the reduction in C_0 observed in the present study.

Further, to help explain the variable responses to training seen for this group, consideration should be given to the impact of tendon stiffness on the transfer of force from the different lower limb muscles to the pedal, especially at high cadences when muscle contraction time is short. Also, the effect of inter-individual variability in patella and Achilles tendon stiffness, on RTD could have made it harder to observe clear changes in power after VEL training (Bojsen-Moller et al., 2005; Waugh et al., 2013). Additionally, as the time course for tendon adaptations typically requires heavy load strength training for longer than eight weeks, we did not anticipate that the four weeks of ballistic training completed by the participants in this study would elicit a change in tendon stiffness (Kubo et al., 2007; Reeves et al., 2003).

The adaptations associated with the improvement in power on the right side of the P-C relationship were unique to VEL. In concert, both maximal plantar-flexion and dorsi-flexion angles were reduced, keeping the ankle in a more plantar-flexed position over most of the pedal cycle (Figure 4.15), while an associated increase in average GAS-TA co-activation occurred (Figure 4.19). The increase in the co-activity of these ankle muscles may have stiffened the ankle joint in the more plantar-flexed position observed. Given the position of the ankle, perhaps an increase in neural drive to GAS (Figure 4.17) may have been attributable, although this could not be quantified. Small changes in range of motion observed at the hip and knee joints may have been able to compensate for the larger change at the ankle joint. Perhaps this movement strategy was adopted to reduce the number of degrees of freedom, keeping the ankle in a position that was more optimal for the transfer of power from the proximal joints to the crank and would not need to be changed at a fast rate given the fast cycle time. Other inter-muscular coordination changes observed for VEL included more co-activation of GMAX-GAS which may have been a strategy to enable greater transfer of muscle force from power producing hip extensors across the ankle plantar-flexors to the crank during the downstroke. The same was not observed for GMAX-RF co-activation.

As noted in Table 4.4, some execution variables were fine-tuned after training, indicated by less variability (i.e. co-activation of most muscle pairs) while others were not (i.e. all joints, and most muscles). Perhaps these participants did not receive enough training to elicit changes in these variables or maybe less variability in the execution of the movement was not essential for power production. The increase in inter-cycle variability for all joints, indicates that these participants did not implement the same movement strategies from pedal cycle to pedal cycle. Instead, they may have exploited the abundant degrees of freedom afforded by the human body, finding their own unique kinematic or muscle activation solution for producing power at moderate to high cadences. The solutions attained for some individuals may have been beneficial, improving the level of power they could produce post-training, while for others the solutions may have been unsuccessful, resulting in little change to no change in power at 160-190 rpm.

4.4.3 Limitations

The design of the intervention matched groups for the total number of revolutions and hence muscle contractions completed per training session, based upon the findings of Tomas et al. (2010). Although, matching the interventions in this manner resulted in RES accumulating a total cycling time that was 30% greater compared to VEL (9.8 ± 0.9 min vs. 6.9 ± 0.4 min). The average cadences maintained by the groups during the sprints performed in training were 78 ± 29 rpm for RES and 177 ± 23 rpm for VEL. Taking into consideration that the majority of power is produced during the downstroke (i.e. half a pedal cycle), the time available for these muscles to reach and maintain a high active state within half a pedal cycle at these cadences was ~ 169 ms for VEL compared to ~ 385 ms for RES. Consequently, the total time for which the power producing lower limb muscles were active would have been less for VEL, particularly when the effect of activation-deactivation dynamics is considered. Neural excitation and muscle force response time delays of around 90 ms have been estimated in most of the lower limb muscles (Van Ingen Schenau et al., 1995), which would further reduce the time available for the muscles to maintain a high active state to ~ 79 ms and ~ 295 ms for RES and VEL respectively. A longer time spent active is likely to have facilitated greater neural adaptations such as an increased rate and level of neural activation, leading to large improvements in power production for those training against the high resistances. Perhaps more time spent cycling may be required for high velocity training interventions to elicit a relative increase in power that was similar to RES.

Based upon previous studies it is expected that neural drive would have increased following training, leading to higher peak EMG values recorded (Hakkinen et al., 1985; Hvid et al., 2016). However, the maximal intensity of the sprint bouts performed in training has the potential to modify maximal levels of activation for those muscles trained, which meant that normalising signals to peak EMG values like recommended in previous research (Rouffet & Hautier, 2008) was not an appropriate method for this study. As co-activation profiles were constructed using EMG signals normalised in reference to their respective time points (i.e. pre or post training) and due to the potential increase in peak EMG, the influence of training on co-activation indices and variance ratios reported in the present study may have been underestimated. Also, due to the type of crank torque system employed in this study, it was not possible to differentiate the torque produced during the downstroke and upstroke phases of the pedal cycle and relate this to the improvements in power observed. Lastly, due to the method of calculating inter-participant variance ratios, statistical comparisons could not be made between pre- and post-training values and hence some caution should be taken when interpreting these findings.

4.5 Conclusion

To conclude, four weeks of ballistic training on a stationary cycle ergometer against high resistances and at high cadences resulted in intervention-specific improvements in the limits of NMF which were associated to specific adaptations of the kinematics and inter-muscular coordination selected to produce the pedalling movement. Changes for the high resistance group included a change in the limits of NMF mainly on the left (i.e. T_0 and power produced at 60-90 rpm), while changes for the high cadence group included an increase in power produced at 160-180 rpm on the right side of the P-C relationship. C_0 was surprisingly reduced following the high cadence intervention, with the decrease observed for this limit in both interventions likely due to effect of activation-deactivation dynamics. For those training at high resistances the improvements in power were largely associated with greater application of torque to the crank during the downstroke, a more dorsi-flexed ankle position over the pedal cycle and increased co-activation of the knee flexors and knee extensors. Based on theoretical studies, this increase in co-activation could potentially lead to a greater transfer of knee extension power to the crank (van Ingen Schenau, 1989). Additionally, the movement strategy adopted (i.e. joint motion and inter-muscular coordination) by VEL was less variable from cycle to cycle. For those training at high cadences, the improvements were associated with the adoption of a more plantar-flexed ankle position, and greater reliance on the transfer of muscle force from power producing hip extensors across the ankle plantar-flexors during the downstroke. In contrast to RES, participants in VEL exhibited more variable movement strategies. It appears that the kinematic and inter-muscular coordination adaptations that took place during RES training were different to those for VEL, although the changes observed for VEL were less clear, even though the participants in both groups performed the same number of repetitions in training. As such the intervention-specific adaptations that took place for each group were not conducive for producing a higher level of power at the opposite section of the P-C relationship for which they did not train. With these findings in mind, a training program combining both high resistance and high velocity training may result in P-C and T-C relationships with inflections that are less pronounced at low and high cadences and thus exhibiting a shape that is more linear.

The increases in power we observed after just four weeks of training may be beneficial for improving the power of the lower limb muscles over the life span, potentially counteracting the previously reported 7.5% reduction in power production observed per decade of life (Martin et al., 2000c). In response to this potential increase in power, the ability to execute functional tasks requiring a large contribution from the lower limb muscles performed as part of daily living is likely to improve. Further, the specific adaptations associated with the improvement in power seen in this study could be used by sport scientists, clinicians and physiologists to provide training cues in real time feedback (i.e. ankle joint position) to individuals sprinting on a stationary cycle

ergometer, which could improve their ability to produce power at specific sections of the P-C relationship.

Chapter 5 The Effect of Ankle Taping on the Limits of Neuromuscular Function on a Stationary Cycle Ergometer

5.1 Introduction

Ankle taping procedures are commonly used in sport science, providing greater structural support, while enhancing proprioceptive and neuromuscular control for injured individuals (Alt et al., 1999; Cordova et al., 2002; Heit et al., 1996; Wilkerson, 2002). Various procedures such as open and closed basket weave with combinations of stirrups and heel locks are commonly used by clinicians and sports trainers to tape the ankle (Fumich et al., 1981; Purcell et al., 2009). These taping techniques commonly used all appear to affect the kinematics of the ankle joint to a certain extent. A meta-analysis showed that rigid adhesive tape can restrict plantar-flexion by 11° on average and dorsi-flexion by 7° during ballistic exercises (Cordova et al., 2000). Although, the effect that ankle taping can have on performance during ballistic movements is unclear. Some authors reported reductions in 40-yard sprint running performance (-4%) and standing vertical jump height (-3.5%), while others have reported non-substantial effects during these exercises (Greene & Hillman, 1990; Verbrugge, 1996). It is possible that the different taping techniques used by these authors (i.e. medial and lateral stirrups combined with heel locks vs basket weave and stirrups) could be attributable to discrepancies in performance.

In maximal cycling exercise, the ankle joint and surrounding musculature play an important role in the transfer of power to the cranks. More than 50% of the force produced by the larger hip (i.e. GMAX) and knee (i.e. VAS) extensor muscles is delivered to the crank through their co-activation with the ankle plantar-flexor muscles (i.e. GAS and SOL) (Zajac, 2002). Therefore, the ankle plantar-flexors ultimately affect the level of power measured at the crank level (Kautz & Neptune, 2002; Van Ingen Schenau et al., 1995). Previous findings show that the range of motion of the ankle and the level of power that can be directly produced by the ankle muscles are larger at low cadences and decrease as cadence increases (McDaniel et al., 2014). This group also showed that the levels of joint power produced by the plantar-flexors during the downstroke phase are much larger than the levels of joint power produced by the dorsi-flexors during the upstroke phase of the pedal cycle. Similarly, the level of crank power produced during the downstroke are largely higher than those produced during the upstroke phase of the pedal cycle (i.e. approximately 6:1) (Dorel et al., 2010). Based on the effect of ankle taping on the kinematics of the ankle joint, it is possible that ankle taping might reduce ankle joint power produced at low cadences and during the downstroke phase. The application of ankle tape while cycling is likely to cause an acute alteration that affects the movement strategy (i.e. kinematics, inter-muscular coordination) employed by the CNS to execute the pedalling task (Muller &

Sternad, 2009). The performance of a new task is characterised by a high level of variability during practice, in particular this variability can be substantial during movements that offers the human body an abundance of solutions, like cycling. Therefore, ankle taping may influence the transfer of force from the muscles through the ankle on to the crank and thus affect the limits of lower limb NMF. Although taping is common practice in other ballistic exercises, there appears to be little investigation into the effect of ankle taping on the variables considered to define the limits of NMF (i.e. power, T_0 , P_{max} , C_{opt} and C_0) of the lower limbs on a stationary cycle ergometer.

The first aim of this study was to investigate the effect of ankle taping on the limits of NMF on a stationary cycle ergometer. To address this research question, we evaluated the effect of ankle taping on the torque-cadence and power-cadence relationships over the downstroke and upstroke phases of the pedal cycle, separately. More specifically, it was assumed that due to the role of the ankle in maximal cycling the limits of lower limb NMF on a stationary cycle ergometer would be affected, in particular those on the left side of the P-C relationship. The second aim was to assess how ankle taping affected crank torque application, lower limb kinematics, inter-muscular coordination and movement variability. To address this research question, kinematic variables (i.e. minimum and maximum angles, range of motion, angular velocity), peak EMG, average co-activation of main muscle pairs and inter-cycle and inter-participant variability were compared between the two conditions at various sections of the P-C and T-C relationships - on the left (i.e. T_0 and power at 40-60 rpm), in the middle (i.e. P_{max} , C_{opt} and power at 100-120 rpm) and on the right (i.e. power produced at 160-180 rpm and C_0) from F-V tests performed on a stationary cycle ergometer with the ankles bi-laterally taped or not. It was assumed that taping would affect the kinematics of the ankle joint, leading to compensatory changes in the kinematics of the proximal joints (hip and knee). It was also assumed that the neural drive to the ankle muscles could be affected as well as the activation of proximal muscles, potentially affecting inter-muscular coordination through changes in the co-activation between various muscle pairs. Additionally, an increase in inter-cycle and inter-participant movement variability was assumed due to the novelty of the task performed.

5.2 Methods

5.2.1 Participants

Eight male (mean \pm SD: age = 26 ± 4 y; body mass = 76 ± 11 kg; height = 176 ± 10 cm) and five female (age = 26 ± 4 y; body mass = 64 ± 10 kg; height = 166 ± 4 cm) low-to-moderately active, healthy volunteers participated in this study. Participants were involved in recreational physical activities such as resistance training and team sports, but did not have any prior training experience in cycling. The experimental procedures used in this study were approved by Victoria University's Human Research Ethics Committee and carried out in accordance with the Declaration of Helsinki. Subjects gave written informed consent to participate in the study if they accepted the testing procedures explained to them.

5.2.2 Experimental design and ankle tape intervention

Participants visited the laboratory for three familiarisation sessions and one main testing session. The purpose of the familiarisation sessions was to ensure that participants were well practiced in the maximal cycling movement as it has been shown that two days of practice allows for valid and reliable measurements of maximal cycling power output in participants with limited cycling experience (Martin et al., 2000). Participants performed the familiarisation sessions without ankle taping. The same exercise protocol, a force-velocity (F-V) test was employed for familiarisation and main testing sessions. In the main testing session participants completed F-V tests in both control and ankle tape conditions. The order of condition was randomised as were the sprints within each condition. For the control condition (CTRL) the cycle ergometer was fit with clipless pedals (Shimano, PD-R540 SPD-SL, Osaka, Japan) and participants were provided with cleated cycling shoes (Shimano SH-R064, Osaka, Japan). The cleat-pedal arrangement was positioned under the forefoot as normally worn while cycling (Figure 5.1C).

In the ankle tape condition (TAPE) the same shoes and cleat-pedal arrangement was used as per CTRL, the only difference was the application of tape on both ankles to restrict the range of motion at the joint (Figure 5.1B). The range of motion of the ankle joints was reduced using rigid tape (Professional Super Rigid, 38 mm, Victor Sports Pty Ltd., Melbourne, Australia) applied in a combination of basket weave, stirrup and heel lock taping procedures previously shown to reduce plantar-flexion angle of the ankle joint (Fumich et al., 1981; Purcell et al., 2009). More specifically, anchor strips were applied to the base of the foot and midcalf, followed by two stirrup strips applied under the foot from the medial to lateral aspect of the midcalf anchor strip. Two separate heel locks were applied (one medially and one laterally) and finally a figure-of-8 (Figure 5.1A). Participants were asked to hold their feet in the most dorsi-flexed position they could while the tape was being applied to the ankle. Taping was performed by the same researcher

throughout the study for consistency. Other than performing the sprints participants' ankle movement was restricted to preserve the integrity of the tape. Participants were also asked to refrain from consuming caffeinated beverages and food 12 hours prior to each test.



Figure 5.1. Ankle taping procedure. **A:** illustration of the steps taken to tape the ankle in this study (taken from Rarick et al. (1962)); **B:** example of the taped ankle and **C:** taping + cycling shoe combination used in the TAPE condition.

5.2.3 Evaluation of the effect of ankle taping on NMF

5.2.3.1 The limits of NMF during maximal cycling exercise

Force-velocity test

A custom built isoinertial cycle ergometer equipped with 172.5 mm instrumented cranks (Axis Cranks Pty, Australia) was used to run the F-V test. Tangential force (i.e. crank torque) was recorded from the left and right cranks separately via load cells at a frequency of 100 Hz and sent in real time to Axis bike crank force vector analyser software (Swift Performance Equipment, Australia). A static calibration of the instrumented cranks while connected to Axis bike crank force vector analyser software was performed prior and after data collection, following procedures previously described (Wooles et al., 2005). The external resistances used during the F-V test (including warm up) were adjusted and controlled using an 11-speed hub gearing system (Shimano Alfine SG-S700, Osaka, Japan). The cycle ergometer saddle height was set at 109% of

inseam length (Hamley & Thomas, 1967), while the handlebars were set at a comfortable height for each subject. At the beginning of the sessions, subjects performed a standardized warm-up of 5-min of cycling at 80 to 90 rpm, at a workload of 100 W and culminated with two practice sprints. Following 5-min of passive rest, subjects performed two F-V tests in the same session, one in the CTRL condition and one in the TAPE condition. Each F-V test consisted of three 4-s sprints interspersed with a 5-min rest period. More specifically, the different sprints completed by each subject were as follows: 1) sprint from a stationary start against a high external resistance; 2) sprint from a rolling start with an initial cadence of ~ 70 rpm against a moderate external resistance and 3) sprint from a rolling start with an initial cadence of ~ 100 rpm against a light external resistance. For each sprint, subjects were instructed to produce the highest acceleration possible while remaining seated on the saddle and keeping their hands on the dropped portion of the handlebars. Subjects were vigorously encouraged throughout the duration of each sprint.

Analysis of T-C and P-C relationships

The methods for analysis of T-C and P-C relationships are the same as those described for the identification of maximal pedal cycles outlined in Study one, (section 3.2.3.1) and Study two (section 4.2.4.1). Briefly, average torque and cadence were recorded and calculated from the Axis cranks, over a full pedal cycle (i.e. LTDC-LTDC and RTDC-RTDC), downstroke (i.e. LTDC-LBDC and RTDC-RBDC) and upstroke (i.e. LBDC-LTDC and RBDC-RTDC) portions of the pedal cycle for each leg separately (Figure 5.2). Power was then calculated using Eqn. 1. The same maximal data point selection and curve fitting procedures as outlined in Study one (sections 3.2.4.1 and 3.2.4.2) were implemented for full pedal cycle, downstroke and upstroke T-C and P-C relationships. Average values of power produced in the downstroke and upstroke phases were then calculated for CTRL and TAPE for three cadence intervals: 40-60 rpm (low cadences), 100-120 rpm (moderate cadences) and 160-180 rpm (high cadences) using between 5 and 10 pedal cycles for each participant. P_{\max} , C_{opt} and C_0 were calculated from regressions fit to each of the P-C relationships (i.e. downstroke and upstroke phases), while T_0 was calculated from regressions fit to each of the T-C relationships.

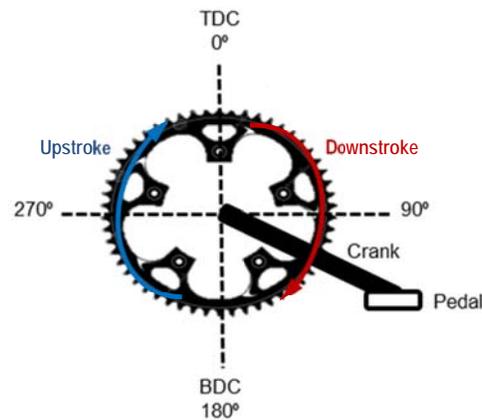


Figure 5.2. Sections of the pedal cycle. A full pedal cycle is defined between TDC and TDC, while the downstroke portion of the pedal cycle is defined between TDC and BDC and the upstroke portion of the pedal cycle is defined between BDC and TDC.

5.2.3.2 Control of the pedalling movement

Crank torque profiles

In comparison to studies one (Chapter 3) and two (Chapter 4) for which total crank torque was recorded (i.e. sum of left and right crank force), the use of Axis cranks in this study enabled the assessment of force delivered to the left and right cranks separately, allowing patterns of force application during the downstroke and upstroke phases of the pedal cycle to be illustrated and quantified. Crank torque signals were time normalised to 100 points, like study one and two using the time synchronised events of left and right top-dead-centre to create crank torque profiles for each pedal cycle. Average crank torque profiles were calculated for three cadence intervals, 40-60 rpm, 100-120 rpm and 160-180 rpm using between 5 and 10 pedal cycles for each participant. Average values of peak and minimum crank torque were then identified from these profiles for the three cadence intervals.

Kinematics of the lower limb joints

The marker setup adopted and three-dimensional kinematic data collected was as per the methods described for Study two in section 4.2.4.2 and illustrated in Figure 4.3. The neutral position of the ankle (i.e. when standing in anatomical position) was approximately 90°. Average hip, knee and ankle joint angle and angular velocity profiles were created from the same pedal cycles (encompassing both left and right pedal cycles) as those used for the analysis of mechanical data

for 40-60 rpm, 100-120 rpm and 160-180 rpm intervals. Minimum and maximum joint angles for the hip, knee and ankle were obtained for each pedal cycle within these cadence intervals, and the difference between the minimum and maximum values was used to obtain joint range of motion (ROM). Joint angular velocity profiles of the extension (plantar-flexion) and flexion (dorsi-flexion) phases of movement for each of the joints were also constructed using the same pedal cycles within the three cadence intervals. Average peak extension/plantar-flexion and flexion/dorsi-flexion joint angles, ROMs and average extension (plantar-flexion) and flexion (dorsi-flexion) angular velocities were calculated from the profiles for the three cadence intervals. Using the zero crossing of the angular velocity profiles, the section of the pedal cycle (i.e. in percent of the pedal cycle) where the joints moved from flexion/dorsi-flexion to extension/plantar-flexion and from extension/plantar-flexion to flexion/dorsi-flexion were also identified for the pedal cycles corresponding to the three cadence intervals.

EMG activity of the lower limb muscles

Surface EMG signals were recorded from four muscles surrounding the left and right ankle joints: GAS, TA, SOL and from GMAX, VAS, RF and HAM muscles on the left only. Attachment of the electrodes and filtering process of the raw EMG signal were as per the methods outlined in Study one (section 3.2.3.2) and Study two (4.2.4.2). As per these studies, synchronisation of EMG and crank torque signals was achieved via the closure of a reed switch which generated a 3-volt pulse in an auxiliary analogue channel of the EMG system which synchronised Axis crank position with the raw EMG signals.

Processed EMG signals were time normalised to 100 points and the amplitude of the RMS for each muscle normalised to the maximum (peak) amplitude recorded during the testing session according to methods previously recommended (Rouffet & Hautier, 2008). Average EMG profiles were then created from the normalised EMG signals for 40-60 rpm, 100-120 rpm and 160-180 rpm using the same pedal cycles used for the analysis of mechanical and kinematic data. Average peak EMG amplitude was then calculated for the downstroke portion of the pedal cycle for GAS, SOL GMAX, VAS, RF and HAM and both the downstroke and upstroke portions of the pedal cycle for TA at each cadence interval. As muscle force (i.e. force applied to the crank) occurs later in the pedal cycle than EMG activity (i.e. EMD) (Cavanagh & Komi, 1979; Ericson et al., 1985; Van Ingen Schenau et al., 1995; Vos et al., 1991), to enable associations to be made between muscle activation and crank torque patterns it was necessary to shift the EMG signal by a given time period or in the present study a given portion of the pedal cycle. EMD has been shown to lie between 60 ms and 100 ms dependent on the muscle, but reports suggest it is approximately 90 ms in most of the leg muscles during cycling regardless of their functional roles

(i.e. mono-articular or bi-articular) (Van Ingen Schenau et al., 1995; Vos et al., 1991). These EMD times appear to remain consistent regardless of cadence (Li & Baum, 2004) and movement complexity (Cavanagh & Komi, 1979) as such at 40-60 rpm a forward EMG shift of approximately 6% would be required (i.e. 60 ms/1200 ms), while at 100-120 rpm and 160-180 rpm the shift would be 15% and 23% respectively.

Co-activation profiles were calculated for GAS-TA, SOL-TA, GMAX-GAS, GMAX SOL, GMAX-RF, VAS-HAM, VAS-GAS and VAS-SOL muscle pairs at 40-60 rpm, 100-120 rpm and 160-180 rpm intervals for CTRL and TAPE using Eqn. 2 stated in Section 3.2.3.3. An average CAI value was then calculated for each muscle pair for the three cadence intervals for CTRL and TAPE conditions.

Variability of crank torque, kinematic, EMG and co-activation profiles

Variance ratios (VR) were used to calculate inter-cycle and inter-participant variability in crank torque, kinematic, EMG and co-activation profiles for CTRL and TAPE. Pedal cycles between 40-60 rpm, 100-120 rpm and 160-180 rpm were used in Eqn. 3 to produce a VR for each participant (inter-cycle variability) and also a VR between subjects (inter-participant variability) like described in study two, section 4.2.4.2.



Figure 5.3. Experimental set up for data collection including the equipment used for the acquisition of mechanical, kinematic and EMG data.

5.2.4 Statistical analyses

Comparison of mean outcome variables were performed with customized spreadsheets using magnitude-based inferences and standardization to interpret the meaningfulness of the effects (Hopkins, 2006a). Differences in means between CTRL and TAPE conditions were analysed for the following variables calculated for the downstroke and upstroke sections of the pedal cycle: T_0 , C_0 , P_{max} and C_{opt} . Power was also calculated and compared at 40-60 rpm, 100-120 rpm and 160-180 rpm. Comparisons between condition means were analysed for the following variables at 40-60 rpm, 100-120 rpm and 160-180 rpm: peak and minimum crank torque, hip, knee and ankle joint angles, range of motion and angular velocity, peak EMG, average co-activation and inter-cycle and inter-participant variance ratios. The standardised effect was calculated as the difference in means (TAPE-CTRL) divided by the SD of the reference condition and interpreted using thresholds set at <0.2 (trivial), ≥ 0.2 (small), ≥ 0.6 (moderate), ≥ 1.2 (large), ≥ 2.0 (very large), ≥ 4.0 (extremely large) (Cohen, 1988; Hopkins et al., 2009). As illustrated in Figure 3.1, (section 3.2.5) small standardised effects are highlighted in yellow, moderate in pink, large in green, very large in blue, extremely large in purple and trivial effects are indicated by no coloured band. Estimates are presented with 90% confidence intervals (\pm CI). The Likelihood that the standardized effect was substantial was assessed with non-clinical magnitude-based inference, using the following scale for interpreting the likelihoods: $\geq 25\%$, possible; $\geq 75\%$, likely; $\geq 95\%$, very likely and $\geq 99.5\%$, most likely (Hopkins et al., 2009). Symbols used to denote the likelihood of a non-trivial/true standardised effect are * possibly, ** likely, *** very likely, **** most likely. The likelihood of trivial effects are denoted by ⁰ possibly, ⁰⁰ likely, ⁰⁰⁰ very likely, ⁰⁰⁰⁰ most likely. Unclear effects (trivial or non-trivial) have no symbol. Data are presented as mean \pm standard deviation (SD) unless otherwise stated.

5.3 Results

5.3.1 Effect of ankle taping on the limits of NMF

5.3.1.1 T-C and P-C relationships

As illustrated in Table 5.1, T_0 estimated from for the downstroke and upstroke phases of the pedal cycle were reduced by small magnitudes in TAPE compared to CTRL. C_{opt} was increased by small magnitudes in TAPE when estimated from both downstroke and upstroke phases, while C_0 was higher in the downstroke phase (Table 5.1). Trivial differences between the two conditions were observed for P_{max} when estimated from either phase of the pedal cycle. Average power produced during the downstroke ($6.56 \pm 1.07 \text{ W}\cdot\text{kg}^{-1}$ vs $6.92 \pm 0.98 \text{ W}\cdot\text{kg}^{-1}$) and upstroke ($1.38 \pm 0.57 \text{ W}\cdot\text{kg}^{-1}$ vs $1.52 \pm 0.50 \text{ W}\cdot\text{kg}^{-1}$) phases at 40-60 rpm were reduced by small magnitudes in TAPE compared to CTRL (Figure 5.4A and B). Trivial differences in power produced during the downstroke and upstroke phases were observed between CTRL and TAPE at 100-120 rpm and 160-180 rpm. Upon comparison of power, P_{max} , T_0 , C_{opt} and C_0 estimated from the downstroke and upstroke, all variables were higher in the downstroke phase in both CTRL and TAPE conditions. More specifically, in TAPE, power calculated from the downstroke was higher than that produced during upstroke phase at 40-60 rpm ($79 \pm 7\%$), 100-120 rpm ($85 \pm 7\%$) and 160-180 rpm ($108 \pm 19\%$), while P_{max} , T_0 , C_{opt} and C_0 were $84 \pm 5\%$, $76 \pm 10\%$, $37 \pm 15 \text{ rpm}$ and $62 \pm 26 \text{ rpm}$ higher, respectively.

Table 5.1. Limits of NMF estimated from P-C and T-C relationships calculated in the downstroke and upstroke phases of the pedal cycle.

	Pedal Cycle Section	CTRL	TAPE	Stand. Effect	Likelihood
P_{max} ($\text{W}\cdot\text{kg}^{-1}$)	Downstroke	12.0 ± 2.4	11.8 ± 2.3	$-0.04; \pm 0.11$	⁰⁰⁰
C_{opt} (rpm)		125 ± 10	129 ± 10	$0.32; \pm 0.30$	*
T_0 ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$)		1.47 ± 0.20	1.39 ± 0.23	$-0.38; \pm 0.31$	**
C_0 (rpm)		224 ± 15	228 ± 21	$0.26; \pm 0.43$	*
P-C r^2		0.94 ± 0.03	0.95 ± 0.02	-	-
T-C r^2		0.97 ± 0.02	0.96 ± 0.03	-	-
P_{max} ($\text{W}\cdot\text{kg}^{-1}$)		Upstroke	2.0 ± 0.7	2.0 ± 0.8	$0.04; \pm 0.18$
C_{opt} (rpm)	88 ± 14		92 ± 16	$0.25; \pm 0.16$	*
T_0 ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$)	0.41 ± 0.17		0.35 ± 0.18	$-0.32; \pm 0.19$	**
C_0 (rpm)	164 ± 27		166 ± 27	$0.09; \pm 0.11$	⁰⁰⁰
P-C r^2	0.96 ± 0.03		0.96 ± 0.03	-	-
T-C r^2	0.97 ± 0.02		0.95 ± 0.04	-	-

Variables estimated from P-C relationship are P_{max} (maximal power) and C_{opt} (optimal cadence). Values estimated from T-C relationships are T_0 (maximal torque) and C_0 (maximal cadence). r^2 indicates the goodness of prediction. Data presented are mean \pm SD; standardised effects are presented with \pm 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly or ** likely. Likelihood of a trivial standardised effect is denoted as ⁰⁰ likely or ⁰⁰⁰ very likely.

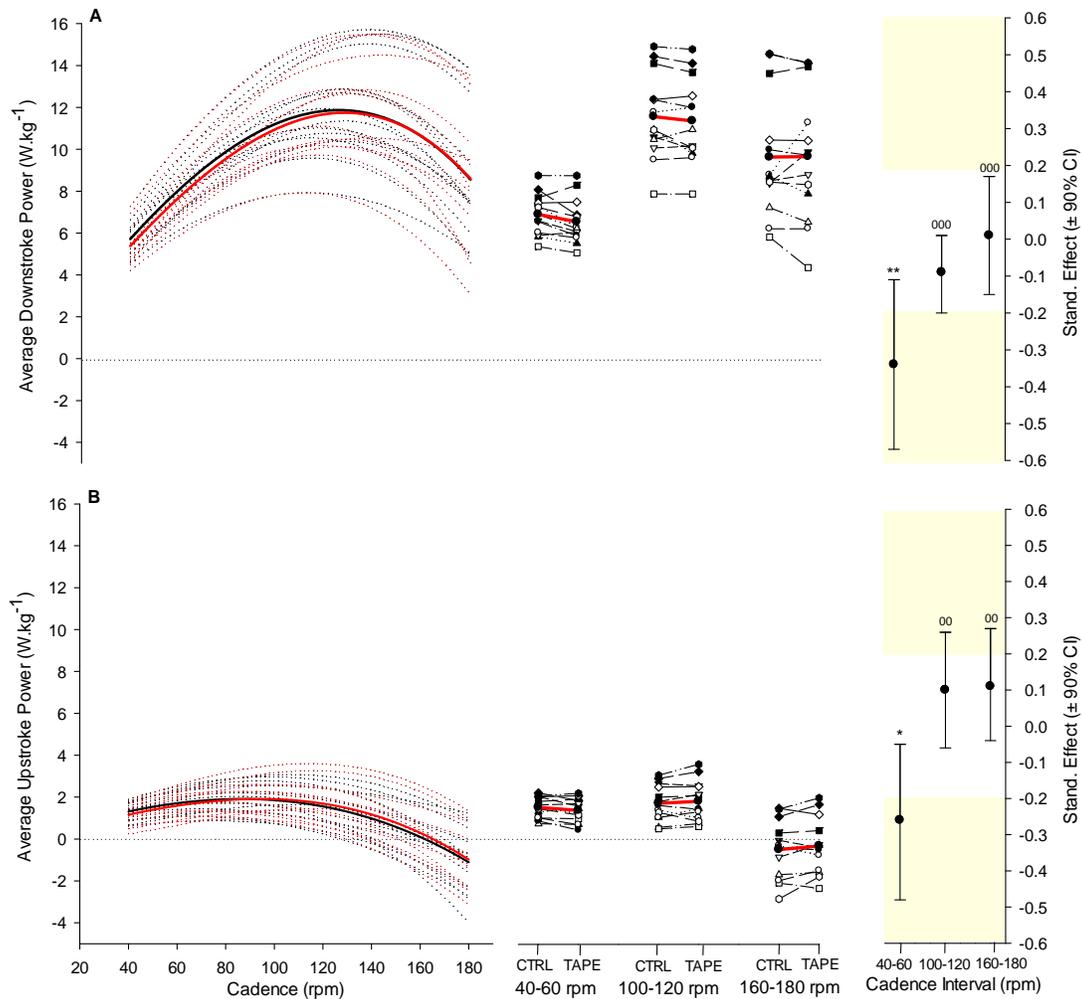


Figure 5.4. Average power produced during the downstroke and upstroke phases of the pedal cycle in CTRL and TAPE conditions. **A:** individual responses for average power produced during the downstroke phase (0-50%) and **B:** during the upstroke phase (50-100%) of the pedal cycle in CTRL (black lines) and TAPE (red lines) conditions. Solid lines indicate mean response, dotted lines indicate individual responses. Middle graphs illustrate average power predicted from the individual relationships at 40-60 rpm, 100-120 rpm and 160-180 rpm. Graphs on the right illustrate the standardised effect \pm 90% CI of the TAPE-CTRL difference at the three cadence intervals. Likelihoods for non-trivial standardised effect are denoted as * possibly or ** likely. Likelihoods for trivial standardised effect are denoted as ⁰⁰ likely and ⁰⁰⁰ very likely.

5.3.1.1 Crank torque profiles

At 40-60 rpm, during the downstroke phase there was a small reduction in peak crank torque produced during the first 25% of the pedal cycle in TAPE compared to CTRL (2.20 ± 0.31 N·m.kg⁻¹ vs 2.31 ± 0.25 N·m.kg⁻¹) (Figure 5.5). At 160-180 rpm, peak torque was lower between 25-40% of the downstroke phase in TAPE compared to CTRL (0.96 ± 0.18 N·m.kg⁻¹ vs 1.02 ± 0.23 N·m.kg⁻¹), while more negative torque (i.e. a lower value of minimum crank torque) was

generated during the latter half of the upstroke phase (i.e. 75-90% of the pedal cycle) in TAPE ($-0.22 \pm 0.09 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ vs $-0.19 \pm 0.07 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) (Figure 5.5). Trivial differences were observed between CTRL and TAPE for minimum and peak crank torque at 100-120 rpm.

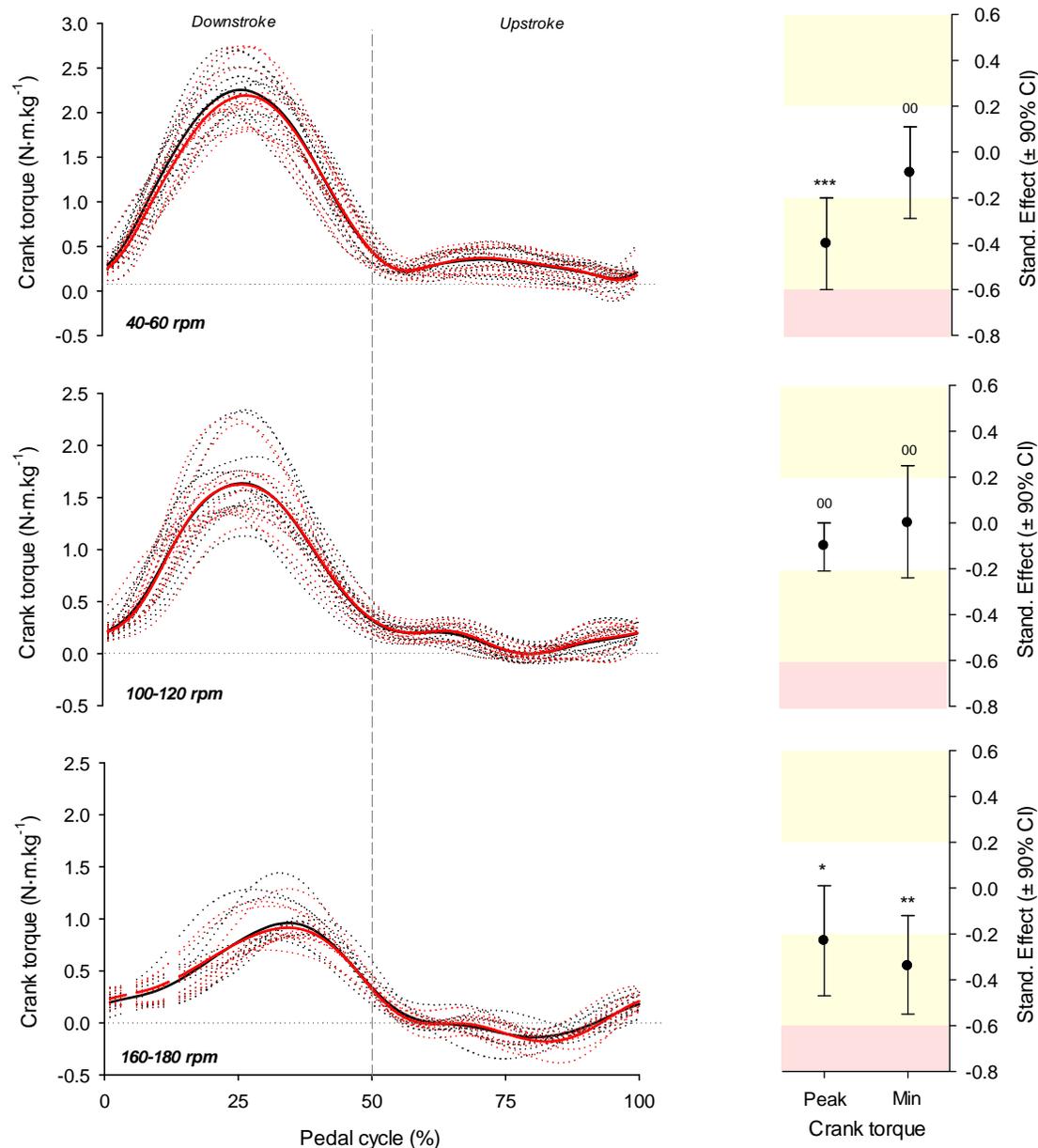


Figure 5.5. Crank torque profiles for CTRL and TAPE conditions. Lines show mean responses at 60-80 rpm, 100-120 rpm and 160-180 rpm for CTRL (black) and TAPE (red). Solid lines indicate mean response, dotted lines indicate individual responses. Graphs to the right of the profiles show standardised effect \pm 90% CI the difference between CTRL and TAPE conditions for min and peak crank torque values. Likelihoods for non-trivial standardised effect are denoted as * possibly, ** likely or *** very likely. Likelihoods for trivial standardised effect are denoted as 00 likely.

5.3.1 Effect of ankle taping on kinematic and EMG and co-activation profiles

5.3.1.1 Kinematic profiles

As illustrated in Table 5.2, few clear changes were observed in the section of the pedal cycle for which the joints moved from extension/plantar-flexion into flexion/dorsi-flexion and from flexion/dorsi-flexion to extension/plantar-flexion. Most notably, was that the ankle moved into dorsi-flexion later in the pedal cycle in TAPE at 40-60 rpm, but, the opposite was observed at 160-180 rpm, with both dorsi-flexion and plantar-flexion occurring earlier in the pedal cycle. Hip flexion started later in the pedal cycle for TAPE at 100-120 rpm.

Table 5.2. Section of the pedal cycle corresponding to the start of joint extension/plantar-flexion and flexion/dorsi-flexion.

	Cadence Int.	CTRL	TAPE	Stand. Effect	Likelihood
Hip Ext	40-60 <i>rpm</i>	2 ± 1	2 ± 1	-0.17; ±0.49	unclear
Hip Flex		51 ± 3	51 ± 2	-0.08; ±0.37	⁰
Knee Ext		96 ± 1	96 ± 2	-0.15; ±0.51	unclear
Knee Flex		48 ± 3	48 ± 2	-0.9; ±0.25	⁰
Ankle PF		13 ± 3	13 ± 4	0.08; ±0.72	unclear
Ankle DF		57 ± 6	62 ± 4	0.83; ±0.56	***
Hip Ext	100-120 <i>rpm</i>	3 ± 1	3 ± 2	0.11; ±0.71	unclear
Hip Flex		50 ± 2	51 ± 2	0.62; ±0.40	***
Knee Ext		96 ± 1	96 ± 2	0.14; ±0.40	⁰
Knee Flex		48 ± 2	48 ± 1	-0.04; ±0.27	⁰⁰
Ankle PF		16 ± 3	16 ± 3	0.18; ±0.46	unclear
Ankle DF		64 ± 6	64 ± 5	0.08; ±0.29	⁰
Hip Ext	160-180 <i>rpm</i>	5 ± 2	4 ± 2	-0.16; ±0.37	⁰
Hip Flex		52 ± 2	52 ± 2	0.00; ±0.34	⁰
Knee Ext		98 ± 1	98 ± 1	0.19; ±0.43	unclear
Knee Flex		46 ± 1	46 ± 1	0.14; ±0.26	⁰
Ankle PF		18 ± 9	15 ± 4	-0.28; ±0.53	*
Ankle DF		69 ± 5	68 ± 5	-0.20; ±0.45	*

Values indicate percent of pedal cycle and are stated as mean ± SD. Ext and PF indicate the start of extension and plantar-flexion, Flex and DF indicate the start of flexion and dorsi-flexion. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly or *** very likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly or ⁰⁰ likely.

Minimum and maximum joint angles and range of motion

At 40-60 rpm, there was a large effect of TAPE on ankle ROM, with an average reduction of $-15 \pm 6^\circ$ observed (Table 5.3). Between 0-25% of the pedal cycle the ankle displayed a moderate reduction in maximum dorsi-flexion angle (i.e. ankle was in a more plantar-flexed position) and during the upstroke phase displayed a large increase in maximum plantar-flexion angle (i.e. ankle was in a more dorsi-flexed position) in TAPE compared to CTRL (Figure 5.7). The hip joint

exhibited a greater ROM for TAPE compared to CTRL at 40-60 rpm. At 100-120 rpm, there was also a large effect of TAPE on ankle ROM, with an average reduction of $-8 \pm 6^\circ$ observed. The reduction in ankle ROM stemmed from a moderate increase in maximum plantar-flexion angle. A small increase in maximum dorsi-flexion angle was also observed (Figure 5.7). The hip joint exhibited a greater ROM for TAPE compared to CTRL at 100-120 rpm. At 160-180 rpm, a large effect of TAPE on ankle ROM, was also observed with an average reduction of $-5 \pm 7^\circ$ (less than that seen at 40-60 rpm and 100-120 rpm). Like 100-120 rpm, the reduction in ankle ROM stemmed from a moderate increase in maximum plantar-flexion angle as illustrated in (Figure 5.7) and quantified in (Table 5.3). The hip and knee joints exhibited small increases in ROM for TAPE compared to CTRL. An effect of cadence was also observed for ankle ROM, with moderate to large standardised effects observed moving from one cadence interval to the next (i.e. standardised effect \pm CI; -1.12 ; ± 0.22 for 40-60 rpm vs 100-120 rpm and -1.84 ; ± 0.27 for 100-120 rpm vs 160-180 rpm).

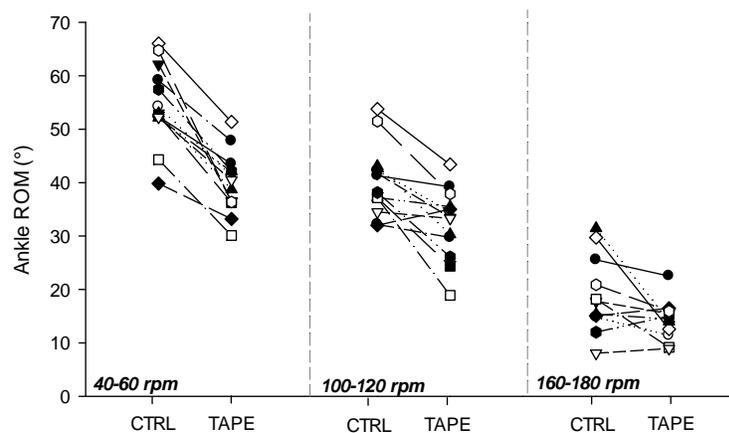


Figure 5.6. Ankle ROM for CTRL and TAPE conditions. Lines show individual responses at 60-80 rpm, 100-120 rpm and 160-180 rpm.

Table 5.3. Minimum and maximum joint angles and range of motion for the hip, knee and ankle joints in CTRL and TAPE at 40-60 rpm, 100-120 rpm and 160-180 rpm.

Degrees (°)		Cadence Int.	CTRL	TAPE	Stand. Effect	Likelihood
Hip	ROM	40-60 rpm	41 ± 9	43 ± 7	0.28; ±0.25	*
	Min		40 ± 13	40 ± 12	0.00; ±0.07	⁰⁰⁰⁰
	Max		81 ± 12	83 ± 11	0.22; ±0.20	*
Knee	ROM		79 ± 17	79 ± 11	0.01; ±0.18	⁰⁰
	Min		25 ± 15	25 ± 10	0.03; ±0.16	⁰⁰
	Max		104 ± 7	104 ± 5	0.09; ±0.14	⁰⁰
Ankle	ROM		54 ± 8	40 ± 6	-1.78; ±0.41	****
	Min		51 ± 7	61 ± 8	1.27; ±0.32	****
	Max		105 ± 6	101 ± 5	-0.64; ±0.27	***
Hip	ROM	100-120 rpm	50 ± 4	51 ± 3	0.28; ±0.35	*
	Min		38 ± 11	37 ± 10	-0.11; ±0.12	⁰⁰
	Max		88 ± 12	88 ± 11	0.00; ±0.15	⁰⁰⁰
Knee	ROM		83 ± 7	84 ± 4	0.06; ±0.32	⁰
	Min		25 ± 8	24 ± 6	-0.07; ±0.20	⁰⁰
	Max		108 ± 6	108 ± 6	-0.02; ±0.21	⁰⁰
Ankle	ROM		40 ± 8	32 ± 7	-0.95; ±0.33	****
	Min		54 ± 9	63 ± 9	0.93; ±0.36	****
	Max		94 ± 7	95 ± 5	0.20; ±0.35	*
Hip	ROM	160-180 rpm	51 ± 4	53 ± 3	0.38; ±0.39	**
	Min		37 ± 10	34 ± 11	-0.19; ±0.21	⁰
	Max		88 ± 13	88 ± 13	-0.03; ±0.13	⁰⁰⁰
Knee	ROM		78 ± 7	81 ± 3	0.43; ±0.40	**
	Min		30 ± 8	29 ± 5	-0.21; ±0.25	*
	Max		108 ± 6	110 ± 5	0.21; ±0.26	*
Ankle	ROM		21 ± 7	15 ± 4	-0.89; ±0.47	***
	Min		57 ± 11	66 ± 9	0.73; ±0.39	***
	Max		79 ± 8	81 ± 8	0.26; ±0.28	*

ROM indicates joint range of motion, Min indicates minimum angle, while Max indicates maximum angle. Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely or **** most likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly, ⁰⁰ likely, ⁰⁰⁰ very likely or ⁰⁰⁰⁰ most likely.

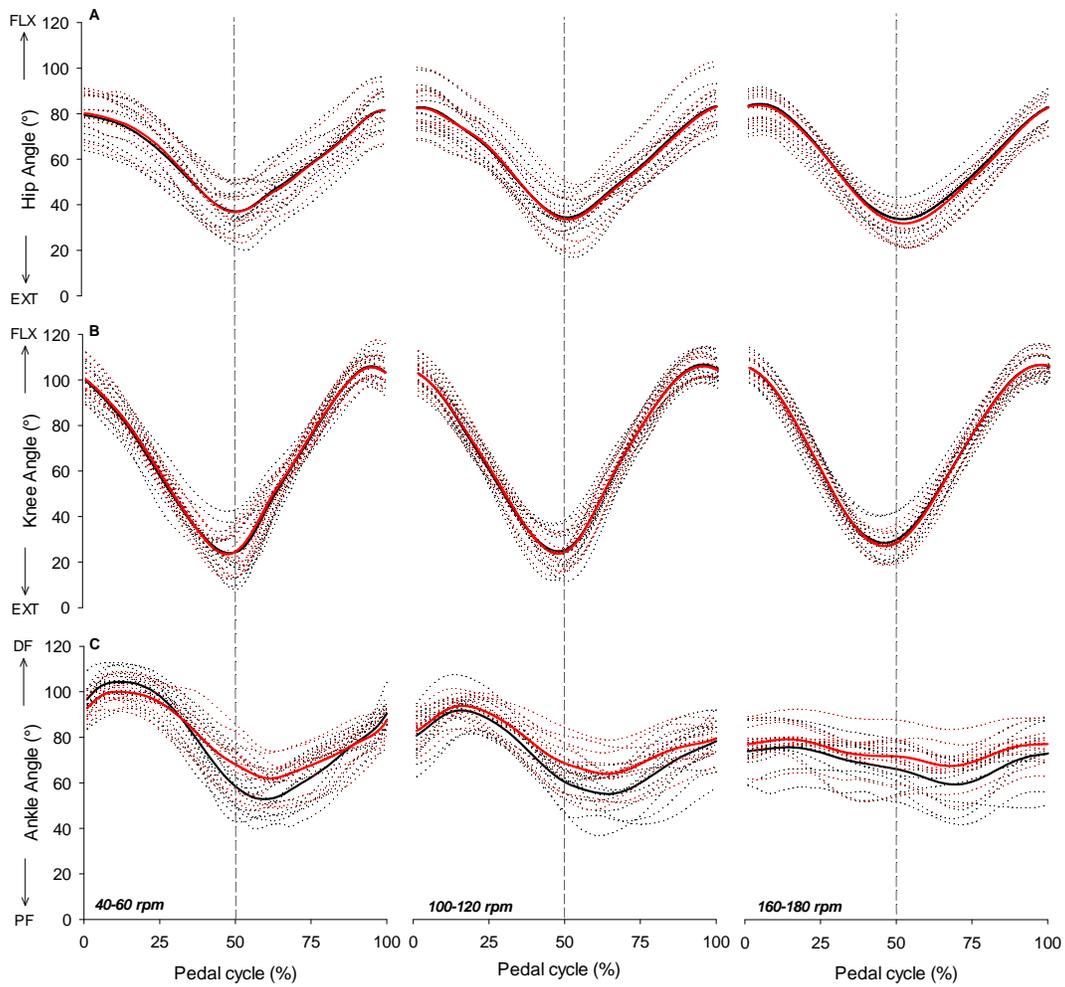


Figure 5.7. Joint angle profiles for CTRL and TAPE conditions. **A:** hip joint, **B:** knee joint and **C:** ankle joint profiles at 40-60 rpm, 100-120 rpm and 160-180 rpm. Solid lines show mean responses for CTRL (black) and TAPE (red) conditions. Dotted lines show individual responses. On the graph axes, EXT and PF indicate that the joint is moving into extension or plantar-flexion, while FLX and DF indicate that the joint is moving into flexion or dorsi-flexion.

Angular velocity of joint phases

At 40-60 rpm, average ankle plantar-flexion and dorsi-flexion, and hip and knee flexion velocities were reduced by large to small magnitudes in TAPE, but a small increase was observed in hip extension velocity (Table 5.4). Average plantar-flexion and dorsi-flexion velocity were reduced by moderate magnitudes at 100-120 rpm, while there was a small increase in average hip flexion velocity (Table 5.4). At 160-180 rpm, average ankle plantar-flexion and dorsi-flexion velocities were still reduced, and average hip flexion velocity increased, with all the changes small in magnitude (Table 5.4).

Table 5.4. Extension/plantar-flexion and flexion/dorsi-flexion velocities for the hip, knee and ankle joints in CTRL and TAPE at 40-60 rpm, 100-120 rpm and 160-180 rpm.

Degrees per second ($^{\circ}.s^{-1}$)	Cadence Int.	CTRL	TAPE	Stand. Effect	Likelihood
Hip Ext Vel	40-60 rpm	79 ± 13	82 ± 15	0.25; ±0.48	*
Knee Ext Vel		154 ± 33	149 ± 24	-0.16; ±0.24	0
Ankle PF Vel		74 ± 20	47 ± 16	-1.29; ±0.38	****
Hip Flex Vel		82 ± 14	77 ± 16	-0.28; ±0.44	*
Knee Flex Vel		154 ± 27	142 ± 27	-0.44; ±0.35	**
Ankle DF Vel		67 ± 21	43 ± 15	-1.11; ±0.36	****
Hip Ext Vel	100-120 rpm	184 ± 22	186 ± 20	0.05; ±0.44	0
Knee Ext Vel		294 ± 32	294 ± 31	0.01; ±0.36	0
Ankle PF Vel		74 ± 33	48 ± 27	-0.76; ±0.29	****
Hip Flex Vel		175 ± 16	180 ± 15	0.29; ±0.55	*
Knee Flex Vel		314 ± 26	312 ± 16	-0.05; ±0.50	0
Ankle DF Vel		71 ± 31	41 ± 26	-0.90; ±0.38	****
Hip Ext Vel	160-180 rpm	287 ± 37	292 ± 25	0.14; ±0.29	0
Knee Ext Vel		434 ± 43	448 ± 23	0.30; ±0.36	*
Ankle PF Vel		34 ± 50	27 ± 35	-0.14; ±0.42	0
Hip Flex Vel		262 ± 19	271 ± 10	0.41; ±0.50	**
Knee Flex Vel		404 ± 39	418 ± 21	0.33; ±0.31	**
Ankle DF Vel		47 ± 31	32 ± 27	-0.44; ±0.42	**

Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, or **** most likely. Likelihood of a trivial standardised effect is denoted as 0 possibly.

5.3.1.2 EMG profiles

At 40-60 rpm, a moderate reduction in peak SOL EMG and small reductions in peak GAS, TA and HAM were observed for TAPE during the downstroke phase (Table 5.5 and Figure 5.8). TAPE also moderately reduced peak TA during the upstroke phase. VAS was the only muscle to show a small increase in peak amplitude at 40-60 rpm in TAPE. At 100-120 rpm, peak EMG of GAS, SOL, TA (upstroke) and GMAX were reduced by small to moderate magnitudes, while VAS increased (Table 5.5). At 160-180 rpm, small increases were observed for peak EMG of TA, GAS and VAS activity during the downstroke phase (Figure 5.8 and Table 5.5).

Table 5.5. Peak EMG values in CTRL and TAPE conditions at 40-60 rpm, 100-120 rpm and 160-180 rpm.

	Cadence Int.	CTRL	TAPE	Stand. Effect	Likelihood
GMAX	40-60 rpm	100 ± 12	99 ± 16	-0.11; ±0.85	0
VAS		94 ± 10	98 ± 11	0.35; ±0.92	*
RF		99 ± 8	100 ± 17	0.15; ±1.20	unclear
HAM		98 ± 13	91 ± 20	-0.49; ±1.07	*
GAS		94 ± 14	89 ± 20	-0.29; ±0.63	*
SOL		117 ± 20	100 ± 23	-0.76; ±0.66	**
TA (downstroke)		56 ± 21	52 ± 24	-0.21; ±0.23	*
TA (upstroke)		129 ± 29	102 ± 15	-0.87; ±0.54	***
GMAX	100-120 rpm	79 ± 16	72 ± 16	-0.39; ±0.38	**
VAS		84 ± 12	87 ± 20	0.20; ±0.49	*
RF		90 ± 27	91 ± 33	0.04; ±0.20	00
HAM		98 ± 22	102 ± 26	0.19; ±0.47	unclear
GAS		103 ± 11	100 ± 14	-0.22; ±0.62	*
SOL		112 ± 9	107 ± 12	-0.61; ±0.99	**
TA (downstroke)		57 ± 20	58 ± 19	0.05; ±0.31	0
TA (upstroke)		106 ± 8	99 ± 11	-0.81; ±0.79	**
GMAX	160-180 rpm	71 ± 21	69 ± 30	-0.09; ±0.33	0
VAS		71 ± 19	78 ± 19	0.34; ±0.34	**
RF		85 ± 18	83 ± 10	-0.13; ±0.33	0
HAM		111 ± 28	111 ± 24	0.00; ±0.8	0
GAS		106 ± 10	110 ± 12	0.41; ±0.53	**
SOL		82 ± 16	83 ± 21	0.10; ±0.74	0
TA (downstroke)		50 ± 26	60 ± 35	0.36; ±0.54	*
TA (upstroke)		97 ± 18	97 ± 20	-0.04; ±0.64	0

Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, or *** very likely. Likelihood of a trivial standardised effect is denoted as 0 possibly.

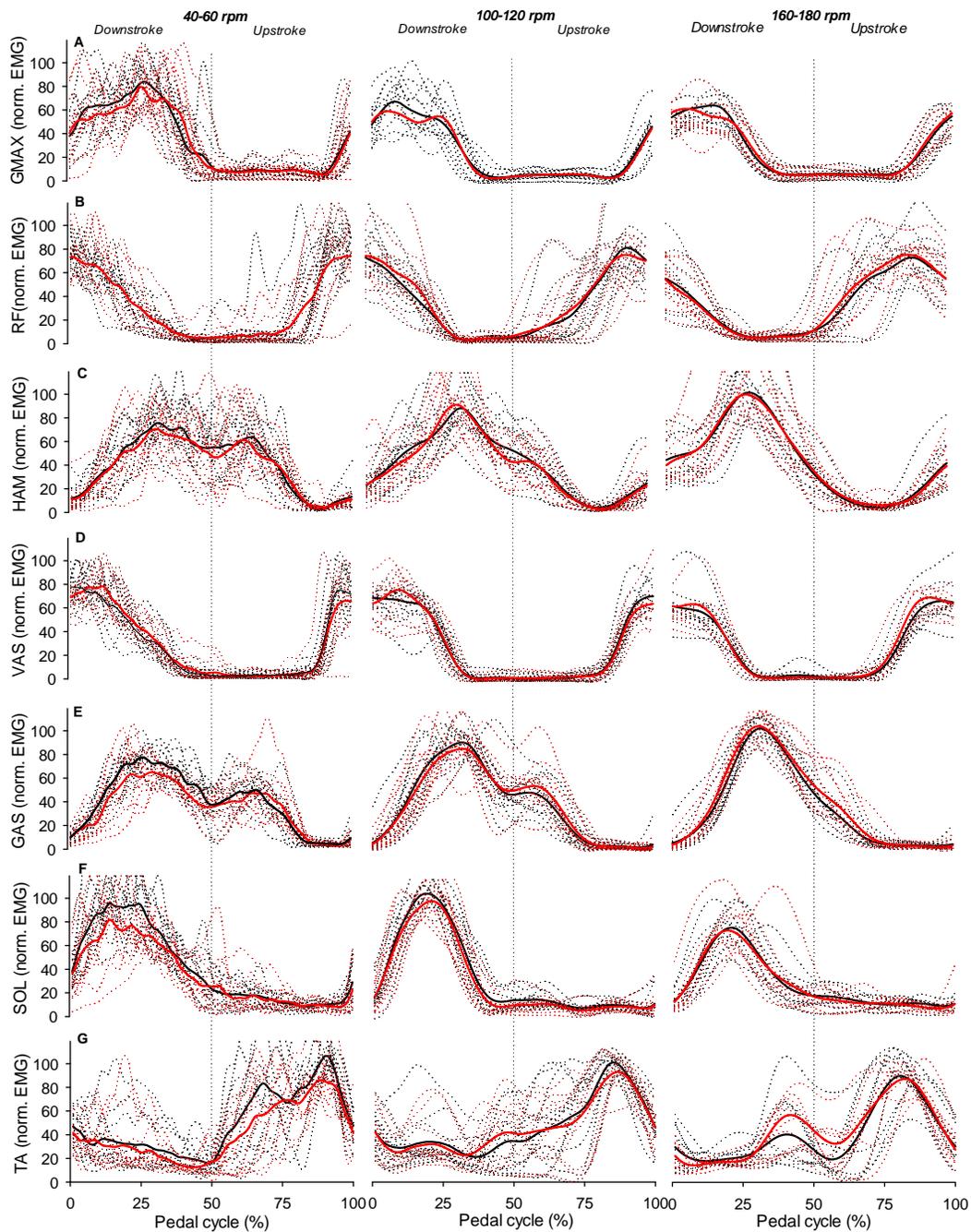


Figure 5.8. EMG profiles for CTRL and TAPE conditions. **A:** GMAX, **B:** RF, **C:** HAM, **D:** VAS, **E:** GAS, **F:** SOL and **G:** TA at 40-60 rpm, 100-120 rpm and 160-180 rpm. Solid lines show mean responses for CTRL (black) and TAPE (red) conditions. Dotted lines show individual responses.

5.3.1.1 CAI profiles

As illustrated in Figure 5.9 and quantified in Table 5.6, at 40-60 rpm, TAPE led to small reductions in the average co-activation of GAS-TA and SOL-TA during both downstroke and upstroke phases and small and moderate reductions in GMAX-GAS and GMAX-SOL muscle pairs respectively. At 100-120 rpm large reductions in average co-activation of GMAX-GAS and GMAX-SOL were seen, while there were small increases in GMAX-RF and GAS-TA co-activation in the upstroke. At 160-180 rpm, a moderate decrease in co-activation of GMAX-SOL and small increase in GAS-TA (upstroke) muscle pairs were observed for TAPE.

Table 5.6. Average CAI values in CTRL and TAPE at 40-60 rpm, 100-120 rpm and 160-180 rpm.

	Cadence Int.	CTRL	TAPE	Stand. Effect	Likelihood
GMAX-GAS	40-60 rpm	41 ± 10	36 ± 9	-0.43; ±0.41	**
GMAX-SOL		51 ± 9	45 ± 8	-0.67; ±0.52	**
GMAX-RF		29 ± 9	28 ± 7	-0.11; ±0.29	0
VAS-HAM		20 ± 7	21 ± 5	0.07; ±0.33	0
VAS-GAS		23 ± 7	22 ± 8	-0.02; ±0.40	0
VAS-SOL		36 ± 10	36 ± 14	0.00; ±0.36	0
GAS-TA (downstroke)		25 ± 17	21 ± 12	-0.26; ±0.16	*
GAS-TA (upstroke)		20 ± 7	18 ± 8	-0.20; ±0.49	*
SOL-TA (downstroke)		25 ± 16	21 ± 13	-0.21; ±0.13	*
SOL-TA (upstroke)		11 ± 5	10 ± 4	-0.31; ±0.37	*
GMAX-GAS	100-120 rpm	41 ± 10	24 ± 5	-1.55; ±0.36	****
GMAX-SOL		48 ± 9	35 ± 8	-1.29 ±0.36	****
GMAX-RF		27 ± 9	30 ± 9	0.39; ±0.23	**
VAS-HAM		24 ± 8	23 ± 7	-0.17; ±0.25	0
VAS-GAS		20 ± 6	19 ± 4	-0.16; ±0.31	0
VAS-SOL		35 ± 14	35 ± 15	0.01; ±0.15	000
GAS-TA (downstroke)		26 ± 14	26 ± 15	0.01; ±0.25	00
GAS-TA (upstroke)		15 ± 7	19 ± 7	0.56; ±0.46	**
SOL-TA (downstroke)		23 ± 12	21 ± 11	-0.19; ±0.20	0
SOL-TA (upstroke)		13 ± 7	12 ± 8	-0.01; ±0.68	0
GMAX-GAS	160-180 rpm	19 ± 4	20 ± 5	0.12; ±0.39	0
GMAX-SOL		31 ± 6	28 ± 8	-0.60; ±0.38	***
GMAX-RF		25 ± 7	25 ± 6	-0.06; ±0.33	0
VAS-HAM		24 ± 7	23 ± 5	-0.12; ±0.24	0
VAS-GAS		12 ± 5	12 ± 3	-0.11; ±0.32	0
VAS-SOL		24 ± 14	24 ± 17	-0.01; ±0.19	00
GAS-TA (downstroke)		23 ± 13	26 ± 12	0.18; ±0.36	0
GAS-TA (upstroke)		8 ± 4	11 ± 5	0.52 ± 0.55	**
SOL-TA (downstroke)		18 ± 9	17 ± 5	-0.18; ±0.35	0
SOL-TA (upstroke)		11 ± 5	12 ± 7	0.16; ±0.68	0

Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely, *** very likely or **** most likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly, ⁰⁰ likely or ⁰⁰⁰ very likely.

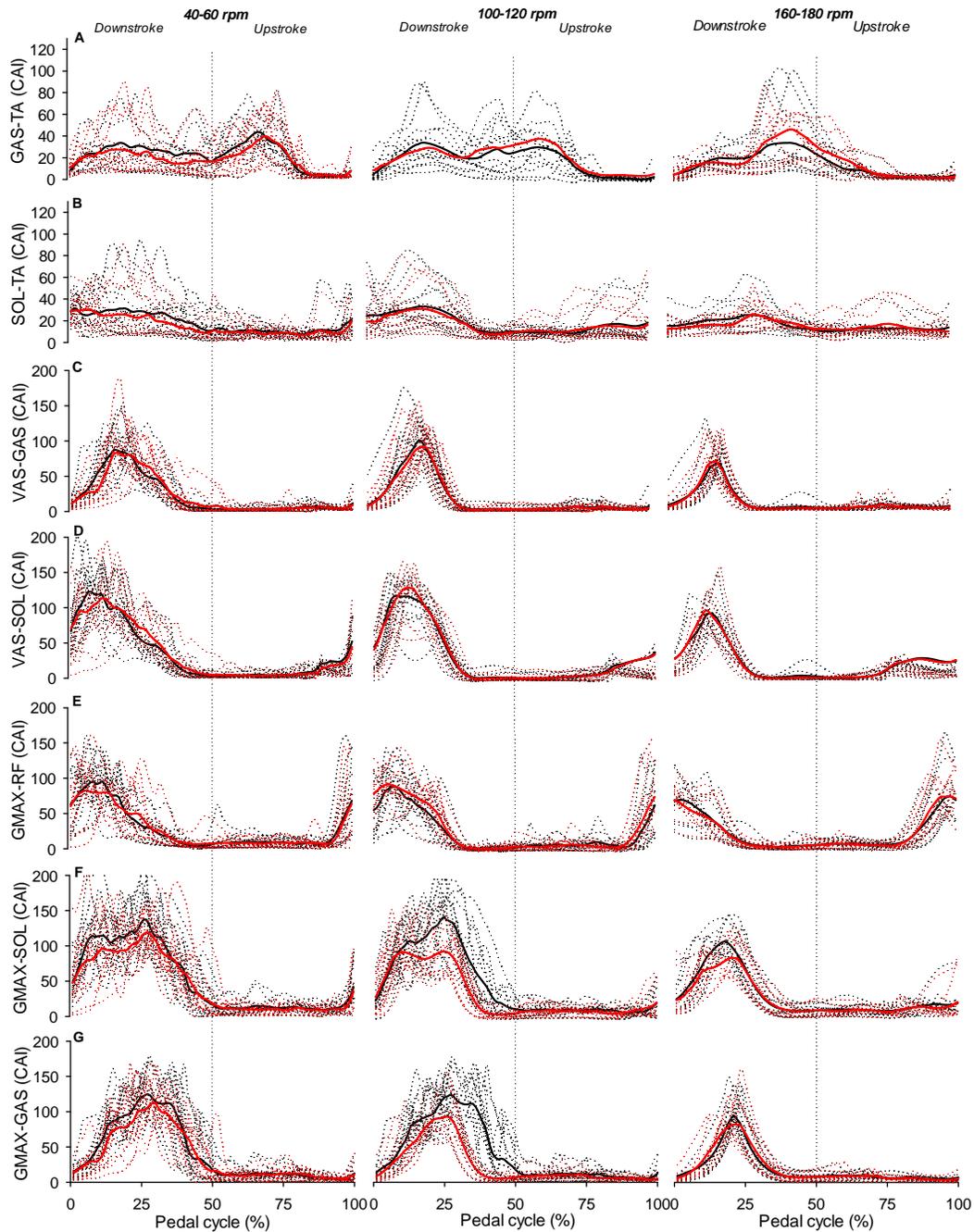


Figure 5.9. Co-activation profiles for CTRL and TAPE conditions. **A:** GAS-TA, **B:** SOL-TA, **C:** VAS-GAS, **D:** VAS-SOL, **E:** GMAX-RF, **F:** GMAX-SOL and **G:** GMAX-GAS at 40-60 rpm, 100-120 rpm and 160-180 rpm. Solid lines show mean responses for CTRL (black) and TAPE (red) conditions. Dotted lines show individual responses.

5.3.2 Variability in crank torque, kinematic, EMG and co-activation profiles

5.3.2.1 Inter-cycle variability

At 40-60 rpm, inter-cycle VR was increased for the ankle joint, VAS, HAM, GAS and TA by small magnitudes (Table 5.7). At 100-120 rpm, inter-cycle VR was increased for GMAX, VAS and RF, but decreases were observed for crank torque and the knee joint (Table 5.8). At the highest cadence interval of 160-180 rpm, inter-cycle VR was increased for crank torque and VAS by small magnitudes and HAM by a moderate magnitude, whereas small reductions were seen for GMAX and SOL (Table 5.9).

Table 5.7. Inter-cycle VR for crank torque, kinematic and EMG profiles for CTRL and TAPE conditions at 40-60 rpm.

	CTRL	TAPE	Stand. Effect	Likelihood
Crank torque	0.03 ± 0.02	0.03 ± 0.02	-0.08; ±0.29	0
Hip joint	0.08 ± 0.05	0.07 ± 0.05	-0.05; ±0.30	0
Knee joint	0.02; ±0.01	0.02 ± 0.01	0.16; ±0.65	unclear
Ankle joint	0.09 ± 0.08	0.14 ± 0.10	0.56; ±0.34	***
GMAX	0.23 ± 0.16	0.23 ± 0.08	-0.02; ±0.36	0
VAS	0.13 ± 0.06	0.16 ± 0.11	0.36; ±0.72	*
RF	0.18 ± 0.10	0.15 ± 0.14	-0.26; ±0.38	*
HAM	0.18 ± 0.06	0.21 ± 0.08	0.42; ±0.82	*
GAS	0.31 ± 0.13	0.37 ± 0.15	0.45; ±0.48	**
SOL	0.34 ± 0.20	0.33 ± 0.15	-0.06; ±0.33	0
TA	0.39 ± 0.24	0.47 ± 0.27	0.34; ±0.75	*

Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely or *** very likely. Likelihood of a trivial standardised effect is denoted as 0 possibly.

Table 5.8. Inter-cycle VR for crank torque, kinematic and EMG profiles for CTRL and TAPE conditions at 100-120 rpm.

	CTRL	TAPE	Stand. Effect	Likelihood
Crank torque	0.03 ± 0.02	0.02 ± 0.01	-0.46; ±0.52	**
Hip joint	0.04 ± 0.04	0.04 ± 0.04	-0.03; ±0.19	00
Knee joint	0.01 ± 0.01	0.01 ± 0.00	-0.42; ±0.50	**
Ankle joint	0.18 ± 0.14	0.18 ± 0.14	-0.02; ±0.13	000
GMAX	0.17 ± 0.09	0.23 ± 0.08	0.62; ±0.38	***
VAS	0.12 ± 0.05	0.15 ± 0.06	0.59; ±0.70	**
RF	0.14 ± 0.10	0.19 ± 0.07	0.48; ±0.34	**
HAM	0.20 ± 0.06	0.19 ± 0.08	-0.12; ±0.75	unclear
GAS	0.22 ± 0.11	0.22 ± 0.12	0.06; ±0.34	0
SOL	0.24 ± 0.16	0.24 ± 0.25	-0.04; ±0.60	0
TA	0.39 ± 0.22	0.40 ± 0.21	0.08; ±0.30	0

Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly, ** likely or *** very likely. Likelihood of a trivial standardised effect is denoted as 0 possibly, 00 likely or 000 very likely.

Table 5.9. Inter-cycle VR for crank torque, kinematic and EMG profiles for CTRL and TAPE conditions at 160-180 rpm.

	CTRL	TAPE	Stand. Effect	Likelihood
Crank torque	0.06 ± 0.02	0.07 ± 0.03	0.34; ±0.85	*
Hip joint	0.02 ± 0.01	0.01 ± 0.00	-0.09; ±0.53	⁰
Knee joint	0.01 ± 0.00	0.01 ± 0.01	0.29; ±0.76	unclear
Ankle joint	0.42 ± 0.14	0.42 ± 0.20	0.05; ±0.51	⁰
GMAX	0.16 ± 0.04	0.14 ± 0.07	-0.49; ±1.07	*
VAS	0.10 ± 0.04	0.12 ± 0.05	0.50; ±1.00	*
RF	0.13 ± 0.05	0.13 ± 0.06	-0.01; ±0.63	unclear
HAM	0.10 ± 0.04	0.14 ± 0.04	0.80; ±0.54	***
GAS	0.18 ± 0.09	0.16 ± 0.10	-0.13; ±0.27	⁰
SOL	0.34 ± 0.20	0.26 ± 0.21	-0.32; ±0.53	*
TA	0.28 ± 0.10	0.30 ± 0.13	0.19; ±0.73	⁰

Data are mean ± SD. Standardised effects are presented with ± 90% CI. Likelihood of a non-trivial standardised effect is denoted as * possibly or *** very likely. Likelihood of a trivial standardised effect is denoted as ⁰ possibly.

5.3.2.2 Inter-participant variability

Due to the method of calculation for inter-participant variance ratios requiring profiles of all participants together, a single value is generated. Hence statistical comparisons could not be performed on the difference between conditions, only comment provided regarding the direction of the change (i.e. increase or decrease). As shown in Table 5.10, at 40-60 rpm, variance ratios were higher in TAPE for profiles of the ankle joint, all muscles except TA, and all co-active muscle pairs. At 100-120 rpm and 160-180 rpm, there was a reduction in variability for crank torque, knee joint, HAM, GMAX-GAS, VAS-GAS, GMAX-RF and VAS-HAM, while an increase in variability was observed for the other muscles (RF, GAS, SOL, TA), VAS-SOL, GAS-TA and SOL-TA muscle pairs (Table 5.10).

Table 5.10. Inter-participant VR for crank torque, kinematic, EMG and CAI profiles for CTRL and TAPE conditions at 40-60 rpm, 100-120 rpm and 160-180 rpm.

	<i>40-60 rpm</i>		<i>100-120 rpm</i>		<i>160-180 rpm</i>	
	CTRL	TAPE	CTRL	TAPE	CTRL	TAPE
Crank torque	0.07	0.08	0.12	0.10	0.17	0.10
Hip joint	0.33	0.29	0.23	0.25	0.25	0.28
Knee joint	0.07	0.05	0.06	0.04	0.06	0.03
Ankle joint	0.18	0.29	0.35	0.41	0.83	0.92
GMAX	0.29	0.34	0.21	0.21	0.26	0.34
VAS	0.13	0.19	0.11	0.14	0.16	0.16
RF	0.26	0.33	0.32	0.40	0.40	0.31
HAM	0.36	0.47	0.32	0.29	0.30	0.29
GAS	0.25	0.34	0.18	0.22	0.09	0.11
SOL	0.20	0.36	0.10	0.13	0.25	0.37
TA	0.52	0.46	0.49	0.53	0.48	0.50
GMAX-GAS	0.24	0.27	0.24	0.21	0.29	0.27
GMAX-SOL	0.27	0.33	0.25	0.24	0.26	0.33
VAS-GAS	0.27	0.33	0.26	0.25	0.31	0.26
VAS-SOL	0.28	0.36	0.33	0.34	0.43	0.53
GMAX-RF	0.35	0.42	0.34	0.23	0.36	0.32
VAS-HAM	0.34	0.36	0.34	0.32	0.31	0.29
GAS-TA	0.76	0.77	0.68	0.73	0.69	0.81
SOL-TA	0.75	0.76	0.75	0.81	0.86	0.89

Data presented are means. SD cannot be calculated for this variable. Variables highlighted in orange indicate a decrease in VR from pre- to post-training, while those highlighted in grey indicate an increase.

5.4 Discussion

The first aim of this study was to investigate the effect of bi-lateral ankle taping on the limits of NMF on a stationary cycle ergometer on the left, at the apex and on the right side of the P-C relationship during different phases of the pedal cycle (i.e. downstroke and upstroke). Ankle taping led to reductions in crank power on the left side of the curve, as reflected by reductions in power produced at 40-60 rpm and decrease in T_0 calculated during both the downstroke and upstroke phases. Ankle taping led to increases in C_{opt} for both phases while no difference in P_{max} or power produced at 100-120 rpm were seen. Ankle taping also led to some minor changes on the extreme section of the right side of the curve which consisted of an increase of C_0 calculated for the downstroke phase, but there was no difference for downstroke or upstroke power produced at 140-160 rpm.

The second aim of this study was to assess how ankle taping affected crank torque application, lower limb kinematics, inter-muscular coordination and movement variability at 40-60 rpm, 100-120 rpm and 160-180 rpm. At 40-60 rpm, taping caused a small reduction in peak crank torque that was accompanied by a change in ankle joint kinematics and a compensatory increase in range of motion and extension velocity at the hip joint. In concomitance there was a reduction in the peak EMG, average co-activation of the ankle muscles, as well as GMAX-GAS and GMAX-SOL muscle pairs. More inter-participant variability was observed for ankle kinematics and inter-muscular coordination. At 100-120 rpm, changes in ankle joint kinematics and EMG were seen, that were compensated by changes in average co-activation (i.e. increases in GAS-TA and GMAX-RF and decreases in GMAX-GAS and GMAX-SOL). In addition, an increase in hip range of motion and reduction in peak GMAX EMG, lead to a large reduction in GMAX-GAS and GMAX-SOL co-activation. At 160-180 rpm, taping caused a reduction in peak torque during the downstroke and minimum torque in the upstroke. The more dorsiflexed position adopted by the ankle across the pedal cycle with changes at the hip and knee joints were seen in response. Linked to the change at the ankle greater average GAS-TA co-activation of was seen in the upstroke for which there was more negative torque. Also, the changes in inter-cycle and inter-participant variability at this cadence interval were not cohesive. Additionally, the reduction in range of motion imposed by the ankle tape was not as substantial at 100-120 rpm and 160-180 rpm, compared to 40-60 rpm as indicated by lower standardised effects in Table 5.3, therefore both condition and cadence had an effect.

5.4.1 Effect of ankle taping on the left side of the P-C relationship

Our results show that ankle taping produced its largest effect on the left side of the P-C relationships and more specifically during the downstroke phase of the pedal cycle, as revealed

by a $0.35 \pm 0.49 \text{ W}\cdot\text{kg}^{-1}$ reduction in crank power at 40-60 rpm and a $0.1 \pm 0.1 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ reduction in T_0 (Figure 5.4). While possible small reductions were also observed for upstroke power at 40-60 rpm and T_0 with ankle taping (Figure 5.4) the ratio of downstroke to upstroke power was high, similar to that observed by (Dorel et al., 2010), highlighting the greater importance of the downstroke phase for power production.

The reductions in power produced during the downstroke were accompanied by reductions in peak crank torque, produced during the first part of the pedal cycle (Figure 5.5). Ankle taping also had the greatest effect on the ankle joint kinematics at these low cadences, with the ankle less dorsi-flexed during the downstroke phase while its angular velocity was also reduced. As such, it appears that the restriction imposed by tape caused participants to plantar-flex their feet to a great degree, earlier in the pedal cycle (Table 5.2), which enabled plantar-flexion to be maintained $\sim 5\%$ longer in the pedal cycle, perhaps in an attempt to increase the duty cycle of the leg (Elmer et al., 2011). In compensation to the adjustment at the ankle, the range of angles covered by the hip joint was increased, associated with an increase in hip extension velocity, leading the hip extensors to operate on a different part of the power vs velocity curve. The reductions in crank torque and power during the downstroke were associated with a reduction in neural drive to the ankle musculature (GAS, SOL and TA) as illustrated in Figure 5.8. This finding suggests that these muscles were less active. The increase in peak VAS EMG suggests that this muscle was more activated, which may have resulted in an increased power production of the knee extensors during the downstroke phase. The reduction in the neural drive to plantar-flexing GAS and SOL and dorsi-flexing TA resulted in less co-activation of these agonist-antagonist muscle pairs over the downstroke (Figure 5.9). As such, ankle taping may have passively increased the stiffness of the joint, reducing the need for co-activation between agonist and antagonist muscles to actively stiffen the joint. Upon consideration of EMD, the reductions in peak EMG of the ankle muscles occurred around the same section of the pedal cycle (15-30%) for which the decrease in peak crank torque was observed. The co-activation of muscle pairs considered to work co-actively to produce and transfer force (i.e. VAS-GAS and VAS-SOL) (Zajac, 2002), were relatively unaffected by taping, perhaps due to the increase in VAS activation accounting for the decreased activation of SOL and GAS. In contrast, the average co-activation of other muscle pairs that work to produce and transfer positive force from the hip extensors to the ankle plantar-flexors during the downstroke (i.e. GMAX-SOL and GMAX-GAS) were reduced with taping, which potentially contributed to the reduction in power output observed.

In the upstroke phase, the ankle adopted a more dorsi-flexed position which may not have required the ankle joint to rotate at the same velocity for this joint action. With this new ankle position, the hip and the knee did not appear to require the same flexion velocity to return the joints back to their position at TDC. The more dorsi-flexed ankle position was concomitant with

more TA activation in the upstroke, although this did not result in more co-activation with the plantar-flexor muscles. Substantial increases in ankle joint variability that accompanied the changes in the amplitude of the profiles, indicates that participants were not able to find a consistent solution to overcome the perturbation, nor did they execute a similar strategy as a group.

Inter-cycle variability was greater for ankle joint movement and several of the distal and proximal muscles (Table 5.7) and inter-participant variability greater for the ankle joint, most muscles and all co-active muscle pairs. Participants may have used the abundance of movement solutions offered by the human body and searched for their own unique solution to the acute perturbation at the ankle. Participants were required to produce maximal power on the cycle ergometer with little prior experience of the pedalling movement itself, let alone with the unfamiliar addition of ankle tape. Indeed, greater movement variability is typically observed in those unskilled or novice to a task (Sides & Wilson, 2012). Further to this, the varied responses in crank torque patterns and ankle joint motion between individuals may in part be attributable to Achilles tendon stiffness. It is known that tendon stiffness influences the transmission of force from the muscle, and that inter-individual variability in tendon stiffness is substantial within and between populations (e.g. men vs. women) (Magnusson et al., 2007; Waugh et al., 2013). Therefore, participants with stiffer Achilles tendons may have displayed larger reductions in power production as a result of the ankle taping, assuming that taping provided the same level of ankle stiffness across all participants.

Overall, it appears that ankle taping may have restricted the contribution of the ankle joint at a section of the P-C relationship (i.e. low cadences) for which the joint has been shown to contribute most to external power (particularly in the downstroke), while operating over a wide range of joint angles (McDaniel et al., 2014).

5.4.2 Effect of ankle taping on the middle of the P-C relationship

At the apex of the P-C relationship, C_{opt} calculated during both the downstroke and upstroke phases were ~ 4 rpm higher when the ankles were taped. This finding combined with the increase in hip flexion velocity implies that the power producing muscles surrounding the hip may have been operating at a different section of their force-velocity relationship. P_{max} (Table 5.1) and power produced between 100-120 rpm (Figure 5.4) during both the downstroke and upstroke phases were similar between conditions. Like observed at low cadences, ankle joint kinematics, including, range of motion and angular velocities in both its movement phases were still moderately reduced with ankle tape. As shown in Figure 5.6, a more dorsi-flexed position across the whole pedal cycle was exhibited. The range of motion of the hip and the portion of the pedal

cycle for which it was extended increased, perhaps to account for the reduction in plantar flexion over the downstroke. Although the activation of GAS and SOL were reduced (Figure 5.8), the level of co-activation between their agonist-antagonist pairs were not affected during the downstroke (Figure 5.9) indicating that these muscles may have worked together to maintain a stable joint position, providing adequate support for force transfer to the crank. The reduction in average co-activation of GMAX with GAS and SOL over the first 50% of the pedal cycle indicates that the transfer of power from the hip extensors to the ankle plantar-flexors may have been less effective. Additionally, this decrease may not have contributed to the reduction in power due to an increase in power transfer from hip extensor muscles to the knee extensors at the same section of the pedal cycle (i.e. increased co-activation of GMAX-RF) (Figure 5.9). Less variability in crank torque profiles was seen between cycles, indicating that participants repeatedly executed a pattern which was favourable for maintaining power production in the downstroke and upstroke despite the perturbation of tape. More variability observed for proximal GMAX, VAS and RF suggests that participants explored strategies that altered the elemental variables (i.e. level of neural drive across the pedal cycle), in attempt to maintain the result variables (i.e. maintaining power).

5.4.3 Effect of ankle taping on the right side of the P-C relationship

On the right side of the relationship, there was a small increase in C_0 calculated in the downstroke (Table 5.1). This may have resulted from ankle taping reducing the complexity of the movement (i.e. reducing the degrees of freedom) and as such the pedalling movement became less variable. However, taping had a trivial effect on the level of power produced at 160-180 rpm during both the downstroke and upstroke phases. Although a reduction was observed in peak crank torque during the downstroke and more negative torque, as illustrated in Figure 5.5, more torque was applied to the crank during the first half of the downstroke which may have compensated for these reductions, and thus power production was maintained. Despite the lack of difference in power at these high cadences, ankle taping still had a moderate effect on the kinematics of the ankle with a more dorsi-flexed position adopted over the pedal cycle. As illustrated in Figure 5.7, the range of angles at which the ankle joint operates (irrespective of ankle taping) narrows as cadence increases. Combining this finding with a lesser contribution of the ankle to crank power (McDaniel et al., 2014) may help to explain why the effect of tape was not like that observed when cycling at low cadences. In compensation to the reduction in ankle range of motion, the hip and knee joints moved through a greater range of angles, for which were covered at a faster velocity during extension for the knee and flexion for the knee and hip. The portions of the pedal cycle for which the hip extended, a heightened level of neural drive was observed and like in the other two cadence intervals may have been a strategy to produce power in compensation for the

perturbation at the ankle. Interestingly, GAS and TA were more activated in the downstroke, however as noted in Table 5.6, average co-activation was not different (Figure 5.9). Only one of the two co-active pairs including GMAX were moderately reduced (i.e. GMAX-SOL), as such the power from the hip extensors to the ankle plantar-flexors was better maintained at high cadences. More variability was observed in the way participants applied force to the crank from cycle to cycle, but equivocal differences were seen in the profiles of the lower limb joints and several muscles. It appears that participants explored different execution strategies (i.e. decreased variability between cycles for GMAX and SOL, but increased variability for VAS) via the many movement solutions offered by the human body (Latash, 2012) but were still able to produce the same result variable (i.e. the maintenance of power while the ankle was taped).

5.5 Conclusion

In summary, ankle taping reduced the limits of lower limb NMF on the left side of the P-C relationship (i.e. T_0 and power produced at 40-60 rpm), particularly during the downstroke phase of the pedal cycle, but had limited impact in the middle (i.e. power produced at 100-120 rpm) and on the right side (i.e. power produced at 160-180 rpm) of the relationship. Taping induced substantial reductions in the range of angles for which the ankle could operate, the velocity at which they rotated and lower neural drive to the surrounding muscles, causing an acute perturbation to the motor system. In response, altered crank torque application, compensations at the proximal muscles and changed inter-muscular coordination was seen. Due to the novelty of the movement performed, individually, participants did not appear to implement cohesive strategies from cycle to cycle and as a group did not respond the same way to the restriction imposed by the ankle taping. The findings of this study provide further insight into the substantial role of the ankle joint for power production on a stationary cycle ergometer, in particular that a substantial ankle joint range of motion is required for maximal power production to be achieved when cycling against high resistances/low cadences, while not vital for maintaining power production at moderate and high cadences. As such, cycling coaches and sport scientists could use real time feedback of ankle joint position and application of torque to the crank to provide their athletes with cues, teaching them to make better use of their ankle muscles.

Chapter 6 General Discussion and Conclusions

The ability to produce adequate power is necessary for the successful execution of functional movements in order to perform a given task. The limits of lower limb NMF on a stationary cycle ergometer are governed by physiological, biomechanical and motor control factors. Cycling is a complex exercise, requiring the optimality of these inter-related factors to enable power and torque production to be maximised. Therefore, this thesis comprised a series of related studies first to assess the limits of lower limb NMF on a stationary cycle ergometer, secondly to improve the limits of NMF using two 4-week interventions performed on a stationary cycle ergometer and thirdly to investigate how ankle taping affects the limits of NMF. The use of EMG, kinetic and kinematic measurement techniques, enabled the physiological, biomechanical and motor control factors affecting the limits of lower limb NMF on a stationary cycle ergometer to be assessed.

6.1 Summary of findings

The findings in Chapter 3 of this thesis show that participants were unable to activate their lower limb muscles in a maximal and optimal manner for every pedal cycle and as such the levels of torque and power produced oscillated between maximal sprints performed as part of a F-V test. Further, the use of higher order polynomial regressions showed that the T-C relationship was not linear for all individuals, while the P-C relationship is not a symmetrical parabola. As such, the new methodological approach outlined in this study offered a more sensitive approach for the assessment of the T-C and P-C relationships and thus the limits of lower limb NMF.

The findings in Chapter 4 provide new evidence that four weeks of ballistic training on a stationary cycle ergometer against high resistances and at high cadences resulted in intervention-specific improvements in the limits of NMF which were associated to specific adaptations of the kinematics and inter-muscular coordination that were not conducive for producing a higher level of power at the opposite section of the P-C relationship for which they did not train. Adaptations on the left side of the P-C relationship included a higher level of crank torque during the downstroke, a more dorsi-flexed ankle position over the pedal cycle, increased reliance on the transfer of knee extension power to hip extension power and the adoption of a less variable movement strategy from cycle to cycle. For those training at high cadences, the improvement on the right side of the P-C relationship were associated with the adoption of a more plantar-flexed ankle position, and greater reliance on the transfer of muscle force from power producing hip extensors across the ankle plantar-flexors during the downstroke and more variable movement strategies.

Finally, the findings in study three showed that the reduction of power produced on the left side of the P-C relationship (i.e. at low cadences) with ankle taping was associated with a reduction in ankle joint range of motion and co-activation of the main muscle pairs likely affecting the transfer of force/power from the proximal muscles to the cranks. More between-participant variability in ankle kinematics and inter-muscular coordination shows that participants adopted different movement strategies in response to ankle taping. Taping had little effect on power produced in the middle (i.e. at moderate cadences) and right side (i.e. at high cadences) of the relationship even though changes in kinematics and inter-muscular coordination were observed. Other limits of NMF within these sections, other than power were modified which included an increase in C_{opt} and decrease in C_0 . Overall it appears that a large range of motion at the ankle joint is essential for producing high levels of power at low cadences.

6.2 General discussion and research significance

Our first investigation in study one showed that the levels of torque and power produced by the participants fluctuated between pedal cycles for all-out sprints performed as part of a F-V test, due to an inability to always activate their lower limb muscles in a maximal and optimal manner. The novel data selection procedure used in this study enabled the selection of experimental data points that truly reflected maximal torque and power. In light of this finding, it appears that selecting maximal pedal cycles over a wide range of cadences is essential prior to modelling T-C and P-C relationships. The selection of maximal data points has particular relevance for the assessment of power and torque in those individuals who have limited prior experience with the pedalling movement, as they are not able to produce consistently high levels of power like seen in trained cyclists (Martin et al., 2000a). The second part of our investigation illustrated that the T-C relationship was not linear in most of our participants, while all participants did not exhibit a P-C relationship that was a symmetrical parabolic shape. These findings refuted the more simple modelling approaches typically used in the cycling literature (Dorel et al., 2010; Dorel et al., 2005; Gardner et al., 2007; Hintzy et al., 1999; Martin et al., 1997; McCartney et al., 1985; Samozino et al., 2007), but was in line with a previous study reporting that the F-V relationship was curvilinear during a leg press exercise (Bobbert, 2012). Due to the improved accuracy of the model, the limits of NMF (i.e. P_{max} , C_{opt} , T_0 and C_0) were more accurately calculated, suggesting that the more simple modelling methods used previously were incorrect and likely not sensitive enough to assess the true limits of NMF. Inaccurate calculations could be particularly important for the limits reported at the apex of the P-C relationship, P_{max} and C_{opt} as these variables are commonly reported in research and used as indicators of performance. This new methodological approach outlined in study one may be of great interest to coaches and sport scientists seeking a

more accurate way to quantify power and torque production on a stationary cycle ergometer and thus the evaluation of the limits of NMF. For sprint cyclists, the method we outlined may provide a more accurate assessment of an athlete's power profile to better identify their strength and weaknesses and further optimize their performances by implementing training interventions that are best suited to them. The progress we made with P-C relationship profiling may also help athletes with factors such as gear ratio selection in training and competition. However, the participants assessed in this research were not trained cyclists, therefore the profiles we observed may be different to those exhibited by an athlete. Although, regardless of expertise, due to effect of neural limitations on power production above cadences of ~ 120 rpm (van Soest & Casius, 2000), the shape of the right side of the P-C relationships may be similar in cyclists to that observed in our group of non-cyclists.

Although the present research investigated the limits of NMF on a stationary cycle ergometer, the methods described could be employed in other ballistic movements (e.g. jumping, sprint running, throwing). The new method could be used to tailor training programs, targeting specific sections of the P-C/T-C (P-V/F-V) relationships that require improvement and then used to evaluate the efficacy of the intervention. Also, the new methods developed can be used to better quantify fatigue during cycling exercises, extending on previous work (Gardner et al., 2009). Lastly, finding that methodological consideration should be given to the way in which T-C and P-C relationships should be modelled, the new approach highlighted in study one was used in the subsequent studies of this thesis to better assess the limits of NMF following training interventions (i.e. study two) and with ankle tape (i.e. study three).

The results from study two confirmed that different ballistic training interventions performed on a stationary cycle ergometer against high resistances and at high cadences leads to improvements in the limits of NMF specific to the exercise condition trained. Indeed, those participants who trained on the left side of the P-C relationship did not improve their ability to produce power on the right side of the curve and vice versa for those participants who trained on the right side of the relationship, indicating that the adaptations were specific. We learnt from the second study that once a P-C profile is obtained for an individual (using the methods from study one), targeted training could be used to change specific sections of their profile in as little as 4 weeks. For example, specific power-training interventions may be beneficial for these track cyclists competing in events such as the 200-m sprint. In this event, cadence is substantially higher (155 ± 3 rpm) for the majority of the race than the cadence corresponding to maximal power (130 ± 5 rpm) (Dorel et al., 2005) (i.e. the majority of power for the sprint duration is produced on the right side of the P-C relationship) and hence, high-velocity training could be beneficial. Further, the improvement in power and torque on the left-side of the P-C and T-C relationships with RES training and on the right-sides of these relationships for VEL suggests that an intervention

combining both high resistance and high velocity training may be beneficial in reducing the inflections observed at low and high cadences. This would likely result in relationships that were more symmetrical and closer to linear, like those previously illustrated in groups of well-trained cyclists (Capmal & Vandewalle, 1997; Dorel et al., 2005).

Specific motor control adaptations were associated with the improvement in power seen for the different interventions, as such these findings could be used in training to provide cues to athletes in real time which may facilitate a greater adaptation. For example, if an athlete's P-C profile reveals a need for the improvement of power at low cadences, feedback could be given to them by sport scientists and coaches regarding the position of their ankle joint, providing cues which allow them to adopt a similar range of motion/ankle angles over the pedal cycle that were linked with the improvement in power seen after the high-resistance training intervention. We acknowledge that it is difficult for laboratory-based tests to mimic the exact requirements of track cycling events performed in the field. However, with further technological development this gap could be closed, For example equipment could be attached to the athletes bike and provide an instantaneous auditory cue when cycling above or below a target power, pre-determined from their individual P-C and power-time profiles.

Further, it should be noted that the adaptations seen in the second study occurred in the short term, therefore those adaptations that may occur with a longer period of intervention-specific all-out sprint cycling training are unknown and warrant further investigation. From a neural point of view, the adaptations to the type of training employed in the present study appear to be specific. However, it is well accepted that morphological changes of the muscle occur past four weeks of training (Hakkinen et al., 1985; Kyröläinen et al., 2005; Moritani & DeVries, 1979), as such these adaptations taking place may not be as specific, improving power production over a wider range of cadences (i.e. the adaptation is less specific to the training conditions). Studies looking at the transfer of adaptations that occurred with stationary cycling to other movements are warranted, but due to the specificity observed within the cycling movement itself (i.e. no cross-over in cycling when moving between the left and right sides of the relationship) the gains may not be completely transferrable to a different exercise mode. Lastly, as power production has been reported to decline by 7.5% per decade of life (Martin et al., 2000c), the $7 \pm 6\%$ and $10 \pm 20\%$ increases in power we observed at specific sections of the P-C relationship, following just four weeks of high resistance and high cadence training respectively, may be useful for counteracting the decline in power over the life span.

The investigation into the effect of bilateral ankle taping on the limits of NMF in study three revealed that tape substantially restricted the kinematics of the ankle and the neural drive to the surrounding musculature over a wide range of cadences (e.g. 40-180 rpm). However, despite this perturbation, power production was only affected at low cadences (in both the

downstroke and upstroke phases) but not at moderate to high cadences. The reduction in inter-muscular co-ordination between the proximal muscles and the ankle muscles, indicates that the ankle muscles play a fundamental role in the delivery of force to the crank when the cadence is low. This finding complements that of McDaniel et al. (2014) who showed that the ankle contributes its greatest amount of power at low cadences.

Further, this study was the first to explore the effect of cadence on the functional role of the plantar-flexor muscles which was previously unexplored *in vivo* or using simulation models (Raasch et al., 1997; Zajac, 2002). The knowledge gained from this study could be applied in a sport science setting, whereby individuals are taught to make better use of their ankle muscles, in an attempt to improve their ability to transfer force from the proximal muscles to the crank. In this scenario, real time feedback of ankle position could be used to ensure that a large range of motion is covered and the variability exhibited in the motion pattern of the ankle is minimised from cycle to cycle. The maintenance of power at moderate and high cadences may have been due to a more stable ankle joint position via greater co-activation of agonist-antagonist ankle muscles, enabling an adequate transfer of force to the crank. As such, it appears that functional role of the ankle muscles changed as cadence increased beyond optimal values. Although to the merit of ankle taping, C_0 was increased. The restriction imposed by tape may have reduced the complexity of the cycling movement, reducing variability enabling participants to reach these very high cadences. With this in mind, individuals or athletes presenting with a P-C profile for which C_0 requires improvement, interventions that reduce the complexity of the pedalling task like ankle taping may be beneficial as a training tool.

Interestingly, after finding in study two, that greater power production after training against high resistances was associated with a more dorsi-flexed position adopted by the ankle, it was assumed that restricting ankle joint range of motion had potential for improving power production. However, as shown in study three, even though the ankle adopted a more dorsi-flexed position during the downstroke at low cadences, a reduction in power production was observed. On comparison of the magnitude of the reduction in ankle range of motion induced by taping ($14 \pm 7^\circ$; standardised effect; $\pm 90\%$ CI: $-1.78; \pm 0.41$) compared to training ($6 \pm 4^\circ$; $-0.75; \pm 0.36$) the reduction with taping was much greater than that seen following training, indicating that the perturbation with ankle tape was too extreme to be of benefit for producing power.

Extending on the findings of study three, a device fixing the ankle joint at a given angle, may offer an experimental manipulation that is more cohesive between participants, which may allow the full effect of the ankle on power production to be realised. The determination of joint powers using inverse dynamics may provide further information regarding the effect of ankle taping/perturbation on the amount of power produced by the joint over a range of cadences. Additionally, it would be interesting to know if a period of training with ankle tape (or with an

ankle fixing device) elicits neuromuscular and motor control adaptations similar to those found in study three following the acute manipulation. In contrast to the findings of the third study, after practice, individuals may respond more favourably to the having their ankles tape and be able to produce more power than in a control condition. As such, further investigation into the benefit of ankle taping as a training tool is warranted.

While the third study induced a kinematic perturbation directly at the ankle joint (i.e. a reduced range of motion) that affected activation of the surrounding muscles, it is also believed that the ankle muscles transfer power by taking advantage of the large moment arm between the ankle and pedal (i.e. the perpendicular distance between the line of action of the force applied to the pedal and the axis of rotation of the ankle joint) (Raasch et al., 1997). Previously shown in submaximal cycling, reducing the length of the ankle moment arm lead to changes in the control of the pedalling movement via decreased activation of the muscles surrounding the ankle (Ericson et al., 1985). However, the importance of the moment arm between the ankle and pedal in the transfer of power through the ankle to the pedal during maximal intensity cycling is unclear. Therefore, it would be of interest to investigate the effect of a mechanical constraint such as a large reduction in the length of the moment arm between the ankle and the pedal (i.e. rearward movement of the cleat towards the axis of rotation of the ankle joint) on the limits of NMF on a stationary cycle ergometer.

6.3 Limitations of this research

This thesis provides new insight into the limits of neuromuscular function on a stationary cycle ergometer. However, interpretation of the data must be considered in the context of the limitations of the research.

General limitations

- Due to the crank torque system employed in the first and second study measuring total crank torque, the contribution of the two limbs could not be dissociated. However, as the thesis progressed, measuring forces on the left and right cranks separately became possible (i.e. Axis cranks) was available and as such was implemented in study three.
- The number of pedal cycles used to calculate average values and variance ratios for a given cadence interval varied depending on the cadence interval assessed. Due to a revolution taking more time to complete at low cadences compared to high cadences, and because the sprints were performed on an isoinertial cycle ergometer, fewer pedal cycles was available for inclusion in the analysis of low cadence intervals. For example, in study

three, approximately five pedal cycles were used for analysis of the 40-60 rpm cadence interval, while approximately 10 pedal cycles were used for the analysis of the 160-180 rpm cadence interval. In addition to the effect of cadence, the number of pedal cycles included within an interval was also participant dependent (i.e. some participants could overcome the external resistance more rapidly than others leading to fewer pedal cycles performed at the beginning of a sprint).

- Although, the co-activation profiles of different muscle pairs were illustrated, values used to compare conditions were represented by an average value calculated over the full pedal cycle. As such co-activation was not calculated over different portions of the pedal cycle, except for average co-activation calculated for agonist-antagonist ankle muscles in the downstroke and upstroke phases in study three.
- Specific cadence intervals (i.e. low, moderate and high cadences) were used in the three studies to assess the effect of data selection procedures, training interventions and ankle taping on the production of power, as such, the effect of these outside of the investigated cadence intervals is unknown and only informed assumptions can be made regarding potentially changes.

Study one limitations

- With regards to the data selection procedures implemented in study one, when only one experimental data point was available for a given 5 rpm cadence interval, it was selected as a maximal cycle/data point unless the power/torque values were substantially lower than those of maximal cycles selected from the adjacent intervals. Consequently, a data point for that given cadence interval was not included in non-maximal cycle T-C and P-C relationships which lead to a small discrepancy in the number of maximal and non-maximal cycles (i.e. 24 ± 3 pedal cycles vs. 19 ± 5 pedal cycles).

Study two limitations

- Although EMG could be used to assess patterns of muscle activation and co-activation of muscle pairs, EMG amplitude could not be compared before and after training in study two due to issues with normalisation of the signal. The location of the EMG electrodes on the lower limb muscles were marked at baseline and were continued to be marked over the training period until the post-training session to ensure consistency of electrode placement. However, issues arose regarding the most appropriate reference value to normalise EMG signals to. The maximal intensity of the sprint bouts performed in training has the potential to modify maximal levels of activation for those muscles

trained, which meant that normalising signals to peak EMG values like recommended in previous research (Rouffet & Hautier, 2008) was not an appropriate method for this study.

- The use of dual-energy x-ray absorptiometry may have provided a more robust quantification of total and lean muscle volume in study two, although this equipment was not available at the time the research was conducted. However, to ensure consistency within and between subjects the same experimenter performed all anthropometric measurements pre- and post-training.
- Due to participants displaying highly variable responses to high velocity training, a larger sample size (i.e. $n > 8$) may have been required to enable the full effect of this training intervention to emerge. Further, matching groups for number of contraction cycles completed as per the work of (Tomas et al., 2010), rather than time spent training (i.e. time for which muscles were recruited), may have underestimated the training volume required to improve power production at fast movement velocities.

Study three limitations

- Ankle taping did not affect the kinematics of the ankle joint in the exact same way for all individuals (i.e. ankle joint ROM). Although, the same researcher performed the same taping procedures on all participants to ensure consistency and reduce the level of experimenter variability, variations in ankle range of motion were seen between participants at each of the different cadence intervals assessed (i.e. 40-60 rpm, 100-120 rpm and 160-180 rpm). Although the tape used was rigid, it still offered some laxity and participants may have produced forces during the sprints that the tape could not withstand causing it to deform, changing ankle joint range of motion within a pedal cycle. In the piloting phase of this study, alternative methods for stiffening the ankle joint were considered such as ankle-foot orthoses. However, these devices were not deemed sturdy enough and therefore unsafe to use during maximal cycling exercise.

6.4 Overall conclusion

The three studies presented in this thesis provide:

- a more robust methodological approach providing a more accurate assessment of T-C and P-C relationship and the limits of lower limb NMF on a stationary cycle ergometer.
- evidence for the potential offered by specific power training interventions employed on stationary cycle ergometers to improve targeted limits of lower limb NMF.
- an understanding of the effect of ankle taping on the limits of the lower limb NMF on a stationary cycle ergometer.

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Appendices

Appendix A: Study one & two participant information documentation



INFORMATION TO PARTICIPANTS INVOLVED IN RESEARCH

You are invited to participate in a research project:

Effect of training interventions at cadences above and below optimal on maximal power vs cadence relationships in non-cyclist males

This project is being conducted by a student researcher Briar Rudsits as part of a PhD study at Victoria University under the primary supervision of Dr. David Rouffet from the College of Sport and Exercise Science, Faculty of Arts, Education and Human Development.

Project explanation

High performances in sprint track cycling events rely on the maximisation of power produced at low and high cadences. During specific sprint events cyclists need to be able to produce power from a stationary start, so low cadences (0-120 rpm). During this initial acceleration phase cyclists adopt a standing position to overcome the high gear ratios and produce as much power as possible. However, once a cyclist is "wound up" they are pedalling at much higher cadences (greater than 120 rpm) and change to a seated position. Performance during these different phases of a sprint event is dependent on the relationship between power and cadence. The aim of this project is to investigate and compare the effect of different training interventions for improving the maximal power vs cadence relationship and associated changes in muscle coordination, mechanical force profiles and lower limb kinematics in non-cyclist males. Specifically, this study will investigate the benefit of changing body position to improve power production at low cadences (seated vs standing) and the benefit of using submaximal efforts to improve power production at high cadences (maximal vs submaximal). The findings from this study will provide a new insight into the effect of different training practises on the power vs cadence relationship and associated neural adaptations. It will also provide coaches with new information for the design of innovative training interventions that could lead to important performance improvements. If you wish to participate in this study you will be randomly allocated to one of four groups in which you will undertake four weeks of bi-weekly training.

What will I be asked to do?

Time Commitment

You will be asked to attend a total of 14 sessions over a maximum of six weeks. For the first four sessions we require approximately 90 minutes of your time each. During the training period we will require approximately an hour of your time for the first week, increased by an extra 20 minutes every week thereafter, as you progress through the training intervention. The two post-test sessions will each require approximately 90 minutes.

Pre-screen and Familiarisation Sessions

During these sessions you will be asked to fill out an informed consent form and health screening questionnaires. You will then begin a familiarisation session where you will become used to the procedures

you will be asked to perform (maximal cycling test, maximal torque tests) and with the equipment that will be used in the testing sessions (cycle ergometer, electromyography, kinematics). We want you to be comfortable with all of the procedures before the study begins and to perform at the peak of your ability every time. You will complete two familiarisation sessions, lasting approximately 90 minutes each. After the familiarisation sessions you will be randomly assigned to one of four groups- seated maximal sprints at cadences above optimal, seated maximal sprints at cadences below optimal, maximal sprints at cadences below optimal out of the seat or submaximal efforts at cadences below optimal.

Baseline and Post-training Testing

The exercise test you became familiarised with will be repeated on a subsequent testing day, no less than 2 days after familiarisation. Each session will take approximately 90 minutes each. Upon arrival to the laboratory reflective infra-red markers will be attached to your back and lower limbs to provide information regarding hip, knee and ankle joint angles and angular velocity. Surface electromyography electrodes will be placed on the muscle belly of both legs to provide information regarding muscle coordination. Prior to placement of electrodes the skin will be prepared by shaving and cleaning with alcohol swabs and secured using tape. You will then perform a warm up of approximately 5 minutes at a submaximal resistance (1.2 W.kg⁻¹) at a cadence of 80-90 rpm, followed by two practice sprints. Following this you will perform a torque-velocity test on a cycle ergometer. This test is comprised of a series of maximal cycle bouts of approximately 4 seconds each, with body position and resistance randomised. Each sprint will be separated by 4 minutes rest. The torque-velocity test and the instrumented cycle ergometer provide us with information regarding power output, optimal cadence, torque and forces applied to the pedals. An adequate cool down period of approximately 5 minutes at 75 W at your chosen cadence, will follow the test. During this session you we will also take anthropometric measurements of your legs. Circumference and skinfold measurements will be obtained from both left and right legs to calculate thigh muscle cross-sectional area. This will involve making several marks with pen on your thigh. Circumference and skinfold measurements will be made over these marks using a soft tape and skinfold calipers. These measures will be put in place to monitor if the changes seen in power-cadence relationship could be due to neural or hypertrophic factors.

The second baseline testing session will require you to perform tests on an isokinetic dynamometer to determine the maximal amount of torque you can produce with the hip, knee and ankle muscle groups during flexion/extension movements at a range of velocities. You will perform a warm up of 3-5 submaximal and one maximal repetition for each muscle group (i.e. knee flexion/extension) and each test velocity (i.e. 180°/s). This will also allow you to become acquainted with the movement before the test starts. Following these you will give three maximal efforts at 4 different speeds (ranging from 60-300°/s) with a rest period of four minutes between each repetition. You will be restrained during the repetitions to isolate the movement being performed. Surface electromyography will be recorded from the corresponding muscles of the hip, knee and ankle muscle groups.

Post-training testing will be conducted approximately one week after your last day of training. You will be asked to attend two testing sessions on separate days. Session one will include a torque-velocity test on a cycle ergometer and anthropometric measurements. Session two will include a torque-velocity test on an isokinetic dynamometer. All test procedures the same as described above.

Training Period

The exercise programme will last for four weeks. During this period you will train two times per week. All exercise will be performed on a cycle ergometer with each session consisting of a series of maximal (seated or standing) or submaximal efforts at high or low cadences based on a set number of revolutions. Each sprint will be separated by approximately four minutes rest. To allow progression, more sprints will be added to each session, increasing the amount of work completed each time. Sessions will begin and end with a warm up and cool down period. During the training period you will be asked not to alter your normal daily exercise routine and to keep a training diary. Training sessions will be run and monitored by the researchers.

What will I gain from participating?

We cannot guarantee that you will have direct benefits from participating in this study. However it is likely that following the training intervention will improve your fitness. During the training intervention will be trained by qualified sport scientists. We will provide feedback about your performance in the baseline and post-intervention tests conducted, allowing you to better understand your sprint ability.

How will the information I give be used?

All of the information gathered in this study is highly confidential and will be coded and stored under secure conditions. The data gathered during the study will be used in a PhD thesis, published scientific literature and conference proceedings, but no identifying personal details will be disclosed. The information you provide will be used anonymously for these purposes only.

The data gathered from this study may be used for related research studies. If you do not want your data to be used for additional studies please tick the check box on the consent form "I agree to the information collected from this study being used for related research purposes". If you agree to your data being used for related research purposes it will be done so anonymously.

During testing we might ask your permission to take photos or video footage of the experimental set up (electrode and marker placement etc) which may be used in research presentations or scientific publications. This will only be done with your prior permission, with all images made anonymous to maintain your privacy.

What are the potential risks of participating in this project?

- The maximal exercise bouts might result in some localised muscle soreness, however this will subside completely within a couple of days.
- The torque-velocity requires repeated maximal cycling bouts which may include risks of vasovagal episodes, muscle soreness and stiffness. The risk of such events is very low, especially with the appropriate warm-up and cool-down procedures that will be employed. Participants will be closely supervised and monitored at all times during testing sessions.
- Participants may become stressed or anxious whilst undertaking the study due to either exercise stress (the high intensity nature of the study) or environmental stress (the procedures being conducted upon them, laboratory surroundings). We will endeavour to minimise these risks by explaining the procedure in full beforehand. If you have any of these feelings and would like to discuss your involvement in this study, you can do so with Dr. Harriet Speed a registered psychologist at Victoria University, Ph: (03) 9919 5412, Email: harriet.speed@vu.edu.au.

How will this project be conducted?

All volunteers will be screened for cardiovascular risk factors and any health issues that prevent them from participating in this study. After explanation of the testing procedures by the researcher and you feel you fully understand the requirements of the research, you will be asked to sign an informed consent document. This study will then be conducted over a six week period, following the protocol described above.

Who is conducting the study?

College of Sport and Exercise Science, Victoria University

Chief Investigator: Dr. David Rouffet
Tel: (03) 9919 4384
Email: david.rouffet@vu.edu.au

PhD Researcher: Miss. Briar Rudsits
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Associate Investigators: Associate Professor. Andrew Stewart
Tel: (03) 9919 5200
Email: andrew.stewart@vu.edu.au

Dr. Simon Taylor
Tel: (03) 9919 9527
Email: simon.taylor@vu.edu.au

Any queries about your participation in this project may be directed to the Chief Investigator listed above.

If you have any queries or complaints about the way you have been treated, you may contact:

Research Ethics and Biosafety Manager
Victoria University Human Research Ethics Committee
Victoria University
PO Box 14428
Melbourne, VIC, 8001
Tel: (03) 9919 4148.



CONSENT FORM FOR PARTICIPANTS INVOLVED IN RESEARCH

INFORMATION TO PARTICIPANTS:

We would like to invite you to take part in the study:

Effect of training interventions at cadences above and below optimal on maximal power vs cadence relationships in non-cyclist males

CERTIFICATION BY SUBJECT

I, _____ of _____

certify that I am at least 18 years old* and that I am voluntarily giving my consent to participate in the study: **'Effect of training interventions at cadences above and below optimal on maximal power vs cadence relationships in non-cyclist males'** being conducted at Victoria University by Dr. David Rouffet, Miss Briar Rudsits, Associate Professor Andrew Stewart and Dr. Simon Taylor.

I certify that the objectives of the study, together with any risks and safeguards associated with the procedures listed hereunder to be carried out in the research, have been fully explained to me by:

Briar Rudsits (PhD Researcher)

and that I freely consent to participation involving the below mentioned procedures:

- High-intensity cycling
- Surface electromyography
- Lower limb kinematics
- Isokinetic dynamometry
- Anthropometric characteristics
- Four weeks of sprint training

I certify that I have had the opportunity to have any questions answered and that I understand that I can withdraw from this study at any time and that this withdrawal will not jeopardise me in any way.

I have been informed that the information I provide will be kept confidential and will not be published. I allow the information gathered during this research to be used after the specified study period has finished.

I agree that the information collected from this study can be used for related research purposes.

Signed: _____

Date: _____

Any queries about your participation in this project may be directed to a researcher:

Dr. David Rouffet

Tel: (03) 9919 4384

Email: david.rouffet@vu.edu.au

Miss. Briar Rudsits (PhD Student)

Tel: 0449 162 051

Email: briar.rudsits@live.vu.edu.au

Associate Professor. Andrew Stewart

Tel: (03) 9919 5200

Email: andrew.stewart@vu.edu.au

Dr. Simon Taylor

Tel: (03) 9919 9527

Email: simon.taylor@vu.edu.au

If you have any queries or complaints about the way you have been treated, you may contact the Research Ethics and Biosafety Manager, Victoria University Human Research Ethics Committee, Victoria University, PO Box 14428, Melbourne, VIC, 8001 or phone (03) 9919 4148.

Appendix B: Study three participant information documentation

INFORMATION TO PARTICIPANTS INVOLVED IN RESEARCH

You are invited to participate in a research project:

Contribution of ankle muscles to power production during maximal cycling exercises

This project is being conducted by a PhD student Briar Rudsits under the principal supervision of Dr. David Rouffet and associate supervision of Dr. Simon Taylor and Associate Professor Andrew Stewart from the College of Sport and Exercise Science at Victoria University.

Project Explanation

The muscles of the ankle (i.e. calf muscles) play an important role during maximal cycling as more than 50% of the power from the big muscles crossing the hip and knee joints can only be transferred to the pedal through the action of the muscles of the ankle. It is generally assumed that the ankle muscles transfer power to the pedal by reducing the range of motion of this joint (i.e. the magnitude of the change in the angle of the ankle joint during the pedalling cycle) and/or by taking advantage of the large moment arm between the ankle and the pedal (i.e. perpendicular distance between the line of action of the force applied to the pedal and the axis of rotation of the ankle joint). However, the importance of those two mechanisms in the transfer of power through the ankle to the pedal still remains unclear. The aims of this study are to investigate and compare: 1) the effect of a large reduction in the length of the moment arm between the ankle and the pedal on power production and movement control during maximal cycling exercise; 2) the effect of decreased range of motion of the ankle on power production and movement control during maximal cycling exercise. To investigate the effect of ankle joint moment arm length and ankle joint range of motion on power production and movement control during maximal cycling exercises, you will perform a Torque-Velocity test (a series of short, maximal sprints) in three different conditions: wearing traditional cycling shoes, wearing modified cycling shoes and wearing traditional cycling shoes with your ankles taped.

As you will have no experience with performing maximal cycling exercises, the study includes a training intervention allowing you to become accustomed to the three experimental conditions outlined above. By comparing your results obtained at baseline and after the training intervention, it will be possible to dissociate the effect of the changes in the mechanical constraints of the movement (i.e. reduction in the moment arm and reduction in the range of motion of the ankle joint) and the effect of inexperience on power production during maximal cycling exercise.

Finally, this study will include isolated testing of the ankle muscles to investigate if the mechanical constraints of the pedalling movement used in this study will have greater effect on participants with stronger ankle muscles. Investigation of this relationship will allow us to confirm the importance of the role played by the ankle muscles in terms of power production during maximal cycling exercises.

What will I be asked to do?

Time Commitment

You will be asked to attend three familiarisation sessions, four testing sessions and eight training sessions over a period of five to six weeks. Familiarisation sessions will require approximately one hour each, every testing session will take approximately two hours of your time and training sessions will take approximately one hour of your time each.

Pre-screen and Familiarisation Sessions

During this session you will be asked to fill out an informed consent form and health screening questionnaires prior to commencement of the testing session. You will then be in a familiarisation session in which you will be run through the testing procedure that will take place at baseline (prior to training) and post-training testing sessions. The testing procedure is termed a Torque-Velocity test which consists of a series of maximal and short duration (5-s each) sprints performed on a stationary cycle ergometer against different levels of external resistances (ranging from low to high). During this test you will be asked to cycle as hard and as fast as possible. During these sessions, you will be asked to wear normal cycling shoes. The objective of this familiarization period is to allow you to be comfortable with all the testing procedures before the study begins, so that we can obtain reliable measurements during the core part of the study.

Baseline and Post-training Testing

Between two and five days after your last familiarisation session, you will be asked to perform the same testing procedure (Torque-Velocity test) as you did in the familiarisation sessions. The results obtained during this session will be used as baseline measurement. Prior to the start of the test, reflective infrared markers will be attached and secured to your back and both lower limbs (using hypoallergenic tape). These markers will be used to study the movements of your hip, knee and ankle joints. Additionally, electrodes will be attached to the skin above 10 muscles on both your lower limbs. These electrodes will be used to measure the recruitment of the muscles by the central nervous system. You will then perform a warm up of approximately 10 minutes at a submaximal resistance ($1.2 \text{ W}\cdot\text{kg}^{-1}$) at a cadence of 80-90 rpm that will include three maximal sprints. Following the warm-up, you will rest for 5-min before performing a Torque-Velocity test on a stationary cycle ergometer equipped with instrumented cranks (used for measuring the force applied to the pedals, as well as the rotation of the cranks). This test is comprised of a series of maximal cycle bouts of approximately 5 seconds each at different resistances. Each sprint will be separated by 5 minutes rest. As part of this testing procedure, you will be asked to perform sprints while wearing traditional cycling shoes, others while wearing the modified cycling shoe and others with both your ankles being taped to restrict their movement. Sprints will start with the tape condition (due to the time requirements of taping but the order of the control and shoe conditions will be randomised). After the final sprint, you will be asked to exercise at a submaximal intensity for 5-min to cool down.

Within 72 hours of the Torque-Velocity test on a cycle ergometer, you will be asked to perform a test to measure the amount of force your ankle muscles can produce. Before the start of this test, you will be asked to perform a warm up protocol that will consist of a series of submaximal and maximal contractions with your ankle muscles against various resistances. For the test itself, you will be asked to perform a series of maximal contractions of the ankle muscles against a set of resistances (ranging from 1 Nm to 30 Nm) with a rest period of three minutes between each repetition. The position of your upper and lower leg will be mechanically restrained during this test to isolate the contribution of the ankle muscles to the exercise.

Post-training sessions will be conducted within one week of your last day of training. You will be asked to attend two testing sessions on separate days. Session one will include a Torque-Velocity test on a stationary cycle ergometer as per the methods described above. The final session will include the test measuring the amount of force your ankle muscles can produce.

Training Intervention

Following the baseline testing procedures, you will be randomly assigned to one of three training groups: training with traditional cycling shoes, training with modified cycling shoes or training with ankle tape. If assigned to the normal cycling shoe group you will be asked to wear normal cycling shoes with the pedal positioned under your forefoot. The modified cycling shoe group will be asked to wear a cycling shoe fitted with a custom-made adapter which allows the position of the foot in reference to the pedal to be moved rearward, so the axis of the pedal is in line with your ankle joint. Moving the pedal axis in line with the ankle joint effectively reduces the moment arm between the ankle and the pedal. The ankle tape group will wear

the normal cycling shoe but have both ankles taped with rigid sports tape to limit ankle joint range of motion and increase joint stiffness. All training sessions will be performed in the same condition, defined depending on the group you were assigned to. The training programme will last for four weeks, and will consist of two training sessions per week. The training principals of overload and progression will be applied through increased training volume (number of sprints performed) and intensity (resistance). All exercise will be performed on a stationary cycle ergometer with each session consisting of a series of short maximal sprints at a range of resistances. All sessions will begin and end with a warm up and cool down period. During the training period you will be asked not to alter your normal daily exercise routine and to keep a training diary. Training sessions will be run and monitored by the researchers.

What will I gain from participating?

We cannot guarantee that you will have direct benefits from participating in this study. We will however, provide feedback about your performance in the tests conducted, such as your ability to generate power on a cycle ergometer before and after training.

How will the information I give be used?

All of the information gathered in this study is highly confidential and will be coded and stored under secure conditions. The data gathered during the study will be used in a PhD thesis, published scientific literature and conference proceedings, but no identifying personal details will be disclosed. The information you provide will be used anonymously for these purposes only.

During testing we might ask your permission to take photos or video footage of the experimental set up (electrode placement etc) which may be used in research presentations or scientific publications. This will only be done with your prior permission, with all images made anonymous to maintain your privacy.

What are the potential risks of participating in this project?

- The maximal exercise bouts might result in some localised muscle soreness or fatigue, however this will subside completely within a couple of days.
- The maximal exercise bouts may include risks of vasovagal and very rarely heart attack, stroke or sudden death. The risk of such events is very low, especially with the appropriate warm-up and cool-down procedures that will be employed. Participants will be closely supervised and monitored at all times during testing sessions.
- Participants may become stressed or anxious whilst undertaking the study due to either exercise stress (the high intensity nature of the study) or environmental stress (the procedures being conducted upon them, laboratory surroundings). We will endeavour to minimise these risks by explaining the procedure in full beforehand. If you have any of these feelings and would like to discuss your involvement in this study, you can do so with Dr. Janet Young a registered psychologist at Victoria University, Ph: (03) 9919 4762, Email: janet.young@vu.edu.au.

How will this project be conducted?

All volunteers will be screened for cardiovascular risk factors and any health issues that prevent them from participating in this study. After explanation of the testing procedures by the researcher and you feel you fully understand the requirements of the research, you will be asked to sign an informed consent document. Following this you will be asked to undertake the activities outlined in this document.

Who is conducting the study?

College of Sport and Exercise Science, Victoria University.

Chief Investigator: Dr. David Rouffet PhD student: Miss. Briar Rudsits
Tel: (03) 9919 4384 Tel: 0449 162 051
Email: david.rouffet@vu.edu.au Email: briar.rudsits@live.vu.edu.au

Associate Investigators: Associate Professor. Andrew Stewart Dr. Simon Taylor
Tel: (03) 9919 5200 Tel: (03) 9919 9527
Email: andrew.stewart@vu.edu.au Email: simon.taylor@vu.edu.au

Any queries about your participation in this project may be directed to the Chief Investigator listed above.

If you have any queries or complaints about the way you have been treated, you may contact:

Research Ethics and Biosafety Manager
Victoria University Human Research Ethics Committee
Victoria University
PO Box 14428
Melbourne, VIC, 8001
Tel: (03) 9919 4148



CONSENT FORM FOR PARTICIPANTS INVOLVED IN RESEARCH

INFORMATION TO PARTICIPANTS:

We would like to invite you to take part in the study:

Contribution of ankle muscles to power production during maximal cycling exercises

CERTIFICATION BY SUBJECT

I, _____ of _____

certify that I am at least 18 years old* and that I am voluntarily giving my consent to participate in the study: 'Contribution of ankle muscles to power production during maximal cycling exercises' being conducted at Victoria University by Dr. David Rouffet, Miss Briar Rudsits, Associate Professor Andrew Stewart and Dr. Simon Taylor.

I certify that the objectives of the study, together with any risks and safeguards associated with the procedures listed hereunder to be carried out in the research, have been fully explained to me by:

Briar Rudsits (PhD student)

and that I freely consent to participation involving the below mentioned procedures:

- Completion of a series of maximal and short duration cycling sprints on a stationary bike ergometer while wearing standard cycling shoes
- Completion of a series of maximal and short duration cycling sprints on a stationary bike ergometer while wearing modified cycling shoes
- Completion of a series of maximal and short duration cycling sprints on a stationary bike ergometer while wearing standard cycling shoes with both ankles taped
- Completion of maximal contractions of the muscles of the ankle
- Recording of the activation of muscles of the lower limbs
- Recording of the displacement of the body segments of the lower limbs
- Recording of the forces applied to the pedals

I certify that I have had the opportunity to have any questions answered and that I understand that I can withdraw from this study at any time and that this withdrawal will not jeopardise me in any way.

I have been informed that the information I provide will be kept confidential and will not be published. I allow the information gathered during this research to be used after the specified study period has finished.

Signed: _____

Date: _____

Any queries about your participation in this project may be directed to a researcher:

Dr. David Rouffet

Tel: (03) 9919 4384

Email: david.rouffet@vu.edu.au

Miss. Briar Rudsits (PhD Student)

Tel: 0449 162 051

Email: briar.rudsits@live.vu.edu.au

If you have any queries or complaints about the way you have been treated, you may contact the Research Ethics and Biosafety Manager, Victoria University Human Research Ethics Committee, Victoria University, PO Box 14428, Melbourne, VIC, 8001 or phone (03) 9919 4148.

Appendix C: Study one (Chapter 3) participant characteristics

Participant	Age (y)	Height (cm)	Body Mass (kg)
1	23	184	84
2	29	191	94
3	32	168	55
4	20	180	87
5	26	185	79
6	19	172	72
7	25	174	74
8	23	173	75
9	22	177	74
10	32	189	93
11	26	188	91
12	32	195	101
13	29	178	84
14	29	181	96
15	22	170	74
16	24	175	78
17	25	183	78
Mean	26	180	82
SD	4	8	11

Appendix D: Study two (Chapter 4) participant characteristics

Group	Participant	Age (y)	Height (cm)	Body Mass (kg)
RES	1	30	191	95
	2	32	168	55
	3	27	185	80
	4	20	180	87
	5	25	179	75
	6	23	173	75
	7	32	189	93
	8	26	188	91
	9	25	183	78
Mean		27	182	81
SD		4	8	12
VEL	10	23	184	84
	11	19	172	72
	12	22	177	74
	13	31	195	101
	14	29	178	84
	15	29	181	96
	16	26	175	79
	17	22	170	74
Mean		25	179	83
SD		4	8	11

Appendix E: Study three (Chapter 5) participant characteristics

Participant	Gender	Age (y)	Height (cm)	Body Mass (kg)
1	Male	23	160	72
2	Male	32	165	62
3	Female	29	168	73
4	Male	24	183	89
5	Female	26	164	71
6	Male	19	177	64
7	Male	27	187	91
8	Female	23	172	70
9	Male	26	175	77
10	Male	28	187	75
11	Female	30	161	54
12	Male	25	173	74
13	Female	22	164	54
Mean		26	172	71
SD		4	9	11

Appendix F: Conference presentations

Rudsits, B. L. and Rouffet, D. M. (2015). *EMG activity of the lower limb muscles during sprint cycling at maximal cadence. European College of Sport Science Conference, Malmo, Sweden. (Oral presentation)*

Introduction: Performances produced during exercises of maximal intensity strongly influence our ability to maximally activate those muscles contributing to the movement. When the movement frequency of maximal exercises is increased, the time window available for activating and deactivating the muscles becomes narrower. According to results of a simulation study, activation-deactivation dynamics could limit sprint cycling performance when cadences increase above optimal cadence (van Soest & Casius, 2000). The aim of this study was to investigate activation and deactivation of the lower limb muscles during sprint cycling at maximal cadence.

Methods: Twelve physically active males performed a torque-velocity test and a maximal sprint against no external resistance on a stationary cycle ergometer. Surface EMG (Noraxon, US) was measured from six muscles [*gluteus maximus* (GMAX), *rectus femoris*, *vastus lateralis* (VAS), *semitendinosus* and *biceps femoris*, *medial gastrocnemius*, *tibialis anterior*]. Normalized peakEMG, minEMG and activation duration (in % of pedalling cycle duration) were calculated for all muscles at two cadences: optimal cadence (C_{opt}) and maximal cadence (C_{max}). Finally a co-activation index was also computed for two pairs of contralateral muscles (GMAX and VAS) at C_{opt} and C_{max} (O'Bryan et al., 2014). One-way ANOVAs with repeated measures were performed to analyse the effect of cadence on the various EMG variables. **Results:** A reduction in peakEMG ($88 \pm 16\%$ vs $74 \pm 21\%$, $P < 0.05$), an increase in minEMG ($3 \pm 2\%$ vs $5 \pm 4\%$, $P < 0.05$) and an increase in activation duration ($64 \pm 13\%$ vs $75 \pm 11\%$, $P < 0.05$) of the lower limb muscles was observed from C_{opt} to C_{max} . Co-activation indexes increased for both GMAX ($5 \pm 3\%$ vs $17 \pm 9\%$, $P < 0.05$) and VAS ($3 \pm 2\%$ vs $7 \pm 3\%$, $P < 0.05$) muscle pairs from C_{opt} to C_{max} . Participants' C_{max} was 218 ± 17 rpm and C_{opt} 124 ± 8 rpm. **Discussion:** The EMG results indicate a reduction in the maximal level of activation of the muscles combined with a reduction in their level of relaxation at maximal cadence. In addition, the relative duration of activation of the muscles was increased, leading to a rise in the co-activation of contralateral power producer muscles that probably caused an augmentation of the negative work produced during the pedaling cycle (Neptune & Herzog, 1999). Finally, larger standard deviation values were seen at C_{max} compared to C_{opt} , indicating greater inter-individual differences in the ability of subjects to perform at high movement frequencies.

Rudsits, B. L., Taylor, S. B. and Rouffet, D. M. (2015). How fast can we really move our legs? Sensorimotor Control Conference, Brisbane, Australia. (Poster presentation)

HOW FAST CAN WE REALLY MOVE OUR LEGS?

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INTRODUCTION

- Sprint performances are sensitive to variations in movement frequency
- Sprint cycling performances produced at high cadences seem to be limited by:
 - Activation-deactivation dynamics⁽¹⁾
 - Selection of motor control strategies to reduce complexity of the pedalling movement⁽²⁾
- One can wonder if the maximal pedalling cadence (C_{max}) reached by humans always equals twice their optimal cadence (C_{opt})⁽³⁾

AIMS

- To determine if C_{max} actually equals twice C_{opt}
- To compare the EMG activity of the lower limb muscles and joint kinematics observed at C_{opt} and C_{max}

METHODS

- Twelve males performed (i) a Torque-Velocity test and (ii) a maximal sprint without external resistance on a stationary cycle ergometer fitted with SRM cranks and Torxtar™
- The following variables were extracted at C_{opt} and C_{max} :
 - Range of motion (ROM) and angular velocity of hip, knee and ankle joints⁽²⁾
 - Normalised PeakEMG, MinEMG and EMG burst duration of GMAX, RF, VAS, HAM, GAS and TA muscles⁽⁴⁾
- Two-way ANOVAs with repeated measures were performed to analyse the effect of cadence on joint and muscle. When significant effects were detected *post hoc* pairwise comparisons were used to locate significant differences ($P < 0.05$)



Fig. 1 Test setup

RESULTS

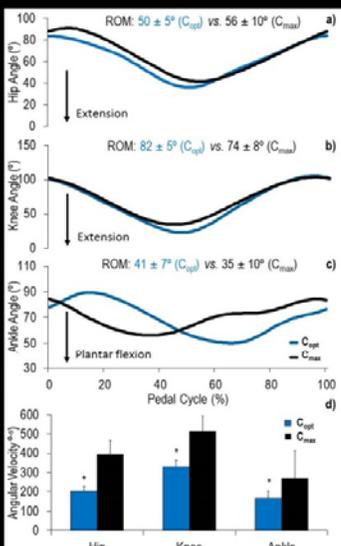


Fig. 2 a) Increase in hip ROM, b) decrease in knee ROM, c) decrease in ankle ROM (cadence x joint interaction effect, $P < 0.05$), d) increase of hip, knee and ankle joint angular velocity from C_{opt} to C_{max} (cadence main effect, $P < 0.05$)

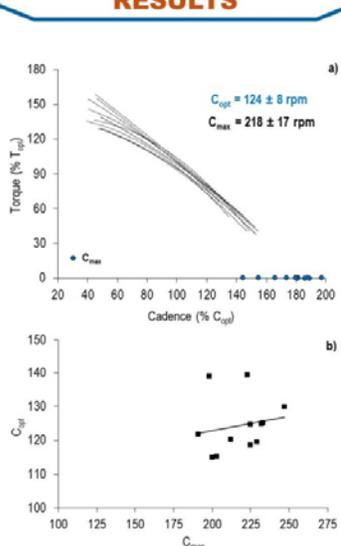


Fig. 3 a) Individual torque-cadence relationships showing measured C_{max} values. C_{max} was lower than twice C_{opt} (-31 ± 21 rpm; $P < 0.05$); b) poor correlation was seen between C_{opt} and C_{max} ($r = 0.18$, $P = 0.58$)

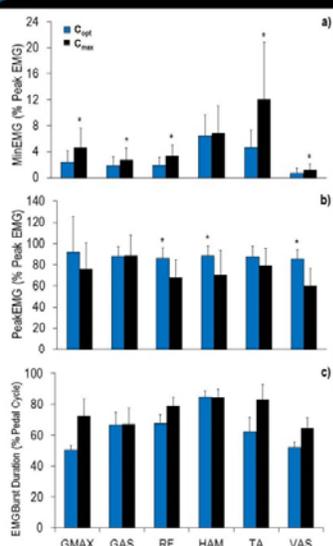


Fig. 4 a) Increase in MinEMG, b) decrease in PeakEMG from C_{opt} to C_{max} (cadence main effect; $P < 0.05$) and c) EMG burst duration increased for HAM and GAS and decreased for GMAX, RF, TA and VAS from C_{opt} to C_{max} (cadence x muscle interaction effect; $P < 0.05$)

CONCLUSION

- C_{max} cannot be predicted from calculation of C_{opt} derived from results of a Torque-Velocity test⁽³⁾
- The changes in EMG confirm that activation-deactivation time constraints prevent subjects from maximally activating and deactivating most lower limb muscles during cycling sprints performed at very high cadences⁽¹⁾
- The changes in distal joint kinematics indicate that subjects changed their motor control strategy during cycling sprints performed at very high cadences⁽²⁾

References: 1. Van Soest AJK, et al. (2000). Med Sci Sports Exerc, 32(11), 1927-1934; 2. McDaniel J, et al. (2014). JAB, 30(3), 423-30; 3. Driss T, et al. (2013), 1-40. Biomed Res Int; 4. Rouffet DM, et al. (2008). JEK, 18, 866-878

Rudsits, B. L., Taylor, S. B., Stewart, A. M. and Rouffet, D. M. (2016). Effect of cadence-specific sprint training on the maximal power-cadence relationships of non-cyclists. Exercise and Sport Science Australia Conference, Melbourne, Australia (Poster presentation)

EFFECT OF CADENCE-SPECIFIC SPRINT TRAINING ON MAXIMAL P-C RELATIONSHIPS OF NON-CYCLISTS

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INTRODUCTION

Training interventions are useful for improving and maintaining the level of power that can be produced by the lower limb muscles [1]. The inclusion of ballistic-type exercises in training programs offer the opportunity to maximally activate muscles over larger time durations to facilitate neural adaptations [2]. Maximal power output was increased by ~6% after four weeks of sprint cycling [3], however the neural adaptations remained to be identified.

AIM

To investigate changes in power production after four weeks of ballistic exercise training (in the form of sprint cycling) performed against high resistances (low cadences) and at high cadences (low resistances) and the associated neural adaptations in a group of active young adults.

METHODS

17 non-cyclist, males were allocated to either a high resistance-low cadence (HI-RES; n=9) or low resistance-high cadence (HI-CAD; n=8) training program performing 4 weeks of bi-weekly, maximal sprint cycling. A Force-Velocity test was performed pre- and post-training on a stationary cycle ergometer fit with SRM cranks and Torxtar™ [4] to enable the following variables to be extracted between cadences of 60-90 rpm and 160-190 rpm: lower limb EMG profiles [5]; EMG co-activation indices (CAI) [6] and joint kinematics [7]. To assess the anthropometric characteristics and strength of the lower limb muscles pre- and post-training, total leg (TL) and lean leg (LL) volume [8] and maximal torque produced by the knee extensor, knee flexor and ankle plantar flexor muscle groups were measured. Magnitude based inferences were used to analyse the change in variables observed after training. Standardised effects ±90% CI provided the magnitude thresholds for small (>0.2), moderate (>0.6) and large (>1.2) changes [9].

EXPERIMENTAL PROTOCOL

RESULTS

Fig 1. Following HI-RES at 60-90 rpm power production increased (**ES: 0.61 ± 0.32, very likely difference (mean ± SD) (A). TA activation was greater during the upstroke phase (50-90% pedal cycle) increasing GAS-TA CAI (**ES: 0.36, ±0.91, possible effect) (B) and ankle ROM decreased (**ES: -0.75 ± 0.36, very likely effect) (C).

		Torque (N m)	Pre	Post
HI-RES	KEXT	168 ± 49	158 ± 45	
	KFLX	105 ± 42	108 ± 29	
	APF	61 ± 16	62 ± 16	
HI-CAD	KEXT	164 ± 38	170 ± 32	
	KFLX	103 ± 19	104 ± 14	
	APF	60 ± 9	60 ± 9	

		Volume (L)	Pre	Post
HI-RES	TL	9.3 ± 1.6	9.4 ± 1.6	
	LL	8.1 ± 1.7	8.2 ± 1.8	
HI-CAD	TL	9.3 ± 1.7	9.4 ± 1.5	
	LL	7.8 ± 1.7	7.8 ± 1.5	

Research Application

The 7% increase in power observed for HI-RES suggests that sprint cycling exercises performed against high resistances could be used to improve/maintain lower limb power production across the lifespan [1].

CONCLUSION

Only the HI-RES group showed an increase in power production for similar exercise conditions (cadences ranging from 60 to 90 rpm). No increase in power production was observed for the HI-CAD group. No changes in the volume of the lower limbs and the maximal strength of the main muscle groups (KEXT, KFLX or APF) were observed after the four week training period. In the HI-RES group, we observed increased co-activation between GAS-TA agonist-antagonist muscles that was associated with a change in the ankle joint angle (i.e. more dorsiflexed). Therefore, training at HI-RES resulted in inter-muscular coordination changes responsible for a modification in the range of motion of the ankle joint.

References: 1. Martin, 2000; 2. Cormie et al. 2011; 3. Creer et al. 2004; 4. Yeo et al. 2015; 5. Rouffet et al. 2008; 6. O'Bryan et al. 2014; 7. McDaniel et al. 2014; 8. Jones & Pearson 1969; 9. Hopkins 2009