The effects of strength training on intermuscular coordination during maximal cycling

BURNIE, Louise Annabelle

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The effects of strength training on intermuscular coordination during maximal cycling

Louise Burnie

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Candidate Declaration

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2. None of the material contained in the thesis has been used in any other submission for an academic award.

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Name	Louise Burnie
Award	Doctor of Philosophy
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Director(s) of Studies	Professor Jon Wheat

Abstract

In natural movement tasks individual muscles are seldom required to generate force in isolation and instead most functional movements arise from the cooperation of several muscles acting together – intermuscular coordination. Contemporary studies of movement coordination are often undertaken using the ecological dynamics theoretical framework and Newell's model of constraints. Ecological dynamics examines human performance from a person-environment scale of analysis considering how people interact with a specific task and the performance environment, and the role these constraints play in the emergent coordination patterns. Pedalling is an ideal task to study intermuscular coordination since it is a natural movement task that can be accurately manipulated. Sprint cyclists often undertake gym-based strength training to increase muscle strength and size. Therefore, the aim of this programme of research was to understand how cyclists adapt their intermuscular coordination patterns during maximal cycling owing to changing organismic constraints (muscle size, strength and fatigue) caused by the gym-based strength training using the theoretical framework of ecological dynamics.

In accordance with the theoretical framework of ecological dynamics and Newell's model of constraints this programme of research highlighted the influence of the constraints acting on the cyclists' coordination patterns that emerge. Different movement and coordination patterns were observed for maximal cycling when the task constraints were changed from sprinting on a fixed ergometer in the laboratory to a track bicycle in the velodrome. This finding implies it is important to undertake biomechanical analyses of movement organisation in elite sports practice in a representative environment. Also, following a gym-based strength training intervention the cyclists' crank power increased, but there were no changes in joint moments, power or muscle activation which suggested that the cyclists might adopt individual coordination strategies following the change in their organismic constraints after the strength training intervention.

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Table of contents

List of figures

List of tables

1 Introduction

This programme of doctoral study was in collaboration with the English Institute of Sport working with the Great Britain Cycling Team.

1.1 Motivation for research

Intermuscular coordination has been defined as the interaction between muscles to control a movement (Young, 2006). In natural movement tasks individual muscles are seldom required to generate force in isolation and instead most functional movements arise from the cooperation of several muscles acting together. Therefore, the amount of force that can be generated in a particular movement context is determined not only by intramuscular factors such as muscle fibre size and type, pennation angle and neural drive, but also by the effectiveness of intermuscular coordination (Carroll, Riek, & Carson, 2001). The neuromusculoskeletal system has many more degrees of freedom than needed to perform many motor tasks (Latash, 2013). The Russian physiologist Bernstein (1967), therefore, defined coordination as the process of mastering the many redundant degrees of freedom involved in a particular movement to reduce the number of independent variables that need to be controlled (Bernstein, 1967; Turvey, 1990).

Contemporary studies of movement coordination are often undertaken using the ecological dynamics theoretical framework. Ecological dynamics examines human performance from a person-environment scale of analysis considering how people interact with a specific task and the performance environment, and the role these constraints play in the emergent coordination patterns (Brymer & Davids, 2014). In accordance with ecological dynamics, Newell (1986) proposed in his model of constraints that patterns of coordination emerge from the confluence of constraints acting on the human movement system. Constraints are boundaries or features that shape the organisation of these emergent coordination patterns. Newell proposed three categories of constraints: organism, task and environment, that interact to influence the emergence of functional patterns of coordination and control for any activity (Newell, 1986; Newell & Jordan, 2007).

Pedalling is an ideal task to study intermuscular coordination since it is a natural movement task that can be accurately manipulated (Neptune, Kautz, & Hull, 1997; Neptune & Kautz, 2001). It is also a less complex multi-joint movement compared to running owing to the mechanical coupling of the cranks and the fixed trajectory of pedal which constrain the kinematics of the lower limbs (Dorel, 2018a). However, it differs from running in that the athlete needs to coordinate the pedalling action with respect to a bicycle, thereby forming a more complex adaptive system. The type of the bicycle used for training can potentially affect the coordination pattern that will emerge, as the task constraints differ depending on whether a person is cycling on a fixed ergometer or on a bicycle that is free to move, such as on a track in the velodrome.

During pedalling, the lower limb segments need to be moved in such a way that the foot moves on a circular trajectory of the pedal (Dorel, 2018b), to achieve this the timing and magnitude of muscle activation has to be coordinated appropriately to allow an efficient energy transfer from the muscles though the body segments to the pedal (Neptune & Kautz, 2001; Raasch, Zajac, Ma, & Levine, 1997). Short-term maximal cycling is an important paradigm for studying physiological capacity (Coso & Mora-Rodríguez, 2006), evaluating force and power characteristics of lower limbs (Dorel, 2018a), muscle coordination and motor control strategies, as well as having direct relevance to a range of competitive cycling environments (Martin, Davidson, & Pardyjak, 2007). Therefore, short-term maximal cycling was chosen to study intermuscular coordination and the effect of changing task and organismic constraints on the coordination pattern. Short-term maximal cycling is a generic term used to refer to an all-out unpaced effort on a bicycle or an ergometer of typically less than 6 seconds to avoid metabolic fatigue (Gardner, Martin, Martin, Barras, & Jenkins, 2007) and will subsequently be referred to as maximal cycling in this thesis. The goal of maximal cycling is to maximise mechanical power output (van Soest & Casius, 2000; Yoshihuku & Herzog, 1996; Yoshihuku & Herzog, 1990). One mechanism that will influence the maximum crank power produced is a cyclist's intermuscular coordination pattern, which needs to maximise energy transfer from the limbs to the crank to deliver maximum effective crank force.

The intermuscular coordination pattern of a sporting movement can be influenced by types of training undertaken by an athlete. Sprint athletes often undertake gym-based strength training, where they perform exercises which are not specific to their sporting movement with the aim to increase muscle size and strength, in addition to their sport-specific training (Delecluse, 1997; García-Pallarés & Izquierdo, 2011; Parsons, 2010). This may affect the intermuscular coordination pattern of the sporting movement. Therefore, this raises questions about how best to prepare athletes for sports performance in sports that vary in the technical demands of coordinating parts of the body together to achieve a task goal and developing the amount of force and power required to achieve that goal (Young, 2006). The training programmes of track sprint cyclists commonly consist of gym-based strength training (where they perform traditional resistance training exercises that are not specific to their sporting movement), and sport-specific training (Parsons, 2010). The proportions of these different types of training vary depending on the goal of the training phase and the proximity to target competitions (Parsons, 2010). The role of the strength training for track sprint cyclists is to increase muscle mass and size with the aim to improve maximal strength, and hence increase maximal power (Parsons, 2010). However, empirical evidence shows that transfer of strength training to sports performance varies (Carroll et al., 2001; Young, 2006). Generally, there is positive transfer to sports performance. However, sometimes there is no effect or even a negative transfer (i.e. strength training is detrimental to performance) (Blazevich & Jenkins, 2002; Carroll et al., 2001; Moir, Sanders, Button, & Glaister, 2007; Young, 2006). Further research is therefore required to investigate the transfer of strength training to sports performance in track sprint cyclists.

Intermuscular coordination is a mechanism which might explain the varied transfer of strength training to sports performance in two ways. Firstly, muscle recruitment patterns associated with a strength training task could retard sports performance when expressed during the sport movement (Carroll et al., 2001). For example, the strength training programme of a sprint cyclist commonly consists of non-specific strength training exercises, such as squats, deadlifts and leg presses (Parsons, 2010). These exercises, however, have very different intermuscular coordination patterns compared to the act of pedalling (Koninckx, Van Leemputte, & Hespel, 2010). If the intermuscular coordination patterns from the non-specific strength training exercises start to be expressed during pedalling, cycling performance could be reduced. Secondly, increases in muscle strength from strength training may need to be accompanied with a change in intermuscular coordination to improve sport performance. This was demonstrated by Bobbert and Van Soest (1994) who used a musculoskeletal simulation to show that an increase in leg strength must be accompanied by a change in intermuscular coordination in order for vertical jump height to increase. This notion that the coordination patterns need to adapt in response to changing constraints (e.g. muscle size, strength and fatigue) is captured by key ideas in ecological dynamics and Newell's model of constraints (Newell, 1986). This raises the questions about the interactions between gym-based strength training and coordination (sport-specific) training and how best to manage the competing demands in track sprint cycling to improve sports performance.

Therefore, this programme of research focusses on understanding how cyclists adapt their intermuscular coordination patterns during maximal cycling due to changing organismic constraints (muscle size, strength and fatigue) caused by the gym-based strength training using the theoretical framework of ecological dynamics. A cyclist's intermuscular coordination can be investigated experimentally by measuring biomechanics variables such as joint kinematics and kinetics and EMG activity. The findings of this programme of research can be used by coaches and sport science practitioners to inform the design of elite athletes' training programmes to achieve a more successful transfer of strength training to sports performance. It will also contribute to the empirical evidence for Newell's model of constraints (Newell, 1986), identifying how athletes' coordination patterns adapt to changing constraints.

1.2 Aim and objectives

The aim of this programme of research was to investigate intermuscular coordination in maximal cycling and whether it is influenced by strength training. The theoretical framework of ecological dynamics was used to enhance understanding of the relationship between strength training and intermuscular coordination in maximal cycling.

The objectives were:

1. To understand coaches' philosophies on the transfer of strength training to elite sports performance.

2. To identify variables that describe intermuscular coordination in maximal cycling.

3. To compare the biomechanical data of a sprint cyclist in the velodrome and in the laboratory.

4. To quantify the test-retest reliability of biomechanical variables (crank powers and forces, joint angles, angular velocities, moments and powers and EMG activity) measured during maximal cycling on an ergometer.

5. To investigate the effect of gym-based strength training on intermuscular coordination in maximal cycling.

1.3 Thesis structure

This programme of research is presented as a thesis comprising seven chapters:

- Chapter 2 provides a critical review of the literature related to this programme of research. The literature review discusses intermuscular coordination, ecological dynamics theoretical framework, intermuscular coordination of cycling and in particular maximal cycling, and the effect of strength training on intermuscular coordination in maximal cycling. Also, the methods that can be used to measure the biomechanical variables that describe intermuscular coordination in maximal cycling are discussed and evaluated.
- Chapter 3 presents coaches' philosophies on the transfer of strength training to elite sports performance. This was a qualitative study designed to capture coaches' experiential knowledge and insights regarding strength training, and the range of factors and ideas believed to affect transfer of strength training to sport performance. The coaches were from a selection of sports demanding maximal effort over a short period of time (track sprint cycling, bicycle motocross (BMX), athletics sprinting, sprint kayaking and rowing) as there are clear parallels between the sports, so coaches' experiences could be synthesised.
- Chapter 4 compares the biomechanical data (crank powers and forces, joint angles, angular velocities, moments and powers and EMG activity) of a sprint cyclist in the velodrome and on an ergometer in the laboratory. The study

investigated how changing the task constraints from a fixed ergometer in the laboratory to riding a moving track bicycle in the velodrome affected intermuscular coordination. This was important to understand as the theoretical framework of ecological dynamics and Newell's model of constraints suggest that changing the task and environmental constraints will influence the emergent coordination patterns.

- Chapter 5 quantifies the test-retest reliability of biomechanical variables (crank powers and forces, joint angles, angular velocities, moments and powers and EMG activity) measured during maximal cycling on an ergometer. Although the study presented in Chapter 4 revealed differences in sprint cycling biomechanics between sprinting on the ergometer and on the track, a decision was made to use the ergometer in the laboratory for the testing protocol for the following studies due to data collection challenges associated with the measuring on track biomechanical data of a cyclist. Therefore, an understanding of test-retest reliability of the maximal cycling ergometer testing protocol was required to allow the assessment and interpretation of the strength training intervention in Chapter 6.
- Chapter 6 investigated the effect of gym-based strength training intervention on intermuscular coordination in maximal cycling. The effect of the strength training intervention on maximal cycling intermuscular coordination was assessed using the key mechanical features associated with maximal cycling, and by comparing the pre and post intervention magnitude and timing of joint moments and powers and EMG activation patterns during maximal cycling.
- Chapter 7 discusses the main findings of this programme of research, the practical applications of the research, followed by the limitations, areas for further research and the contribution to knowledge.

2 Literature review

2.1 Introduction

This chapter reviews the literature on intermuscular coordination and how it can be studied within the theoretical framework of ecological dynamics and Newell's model of constraints (sections 2.2 and 2.3). Section 2.4 and 2.5 reviews the literature on intermuscular coordination in cycling and specifically sprint cycling. Section 2.6 discusses the relationship between strength training and intermuscular coordination in maximal cycling. Section 2.7 discusses and evaluates the measurement techniques and analysis methods that can be used to measure the variables that describe intermuscular coordination in maximal cycling.

2.2 Intermuscular coordination

Intermuscular coordination has been defined as the interaction between muscles to control a movement (Young, 2006). The amount of force that an isolated muscle can exert is influenced by factors such as: muscle fibre size, pennation angle and muscle fibre type (Abernethy, Jürimäe, Logan, Taylor, & Thayer, 1994; Cormie, McGuigan, & Newton, 2011b). However, in natural movement tasks individual muscles are seldom required to generate force in isolation and instead most functional movements arise from the cooperation of several muscles acting together. Therefore, the amount of force that can be generated in a particular movement context is determined not only by intramuscular factors but also by the effectiveness of intermuscular coordination (Carroll et al., 2001).

2.3 Ecological dynamics

Contemporary studies of movement coordination are undertaken under the ecological dynamics theoretical framework, where it has been proposed that the neuromusculoskeletal system has many more degrees of freedom than needed to perform many motor tasks. The coordination of a movement is the process of mastering these redundant degrees of freedom to form a controllable system. This problem of motor redundancy is commonly known as the Bernstein problem, after the Russian physiologist who conceptualised this issue (Bernstein, 1967; Glazier & Davids, 2009; Latash, 2013).

Synergies can provide a solution to the degrees of freedom problem by linking neuromusculoskeletal components so they act coherently together (Bernstein, 1967; Riley, Shockley, & Van Orden, 2012; Turvey, 1990). Synergy formation between muscles the body's degrees of freedom (e.g. muscles, limbs, joints and bones) can be described using an ecological dynamics approach. Synergies or coordinative structures are temporary assemblies of system components so that they behave as a single functional unit (Glazier & Davids, 2009; Riley, et al., 2012). Coordinative structures can be defined as a group of muscles spanning several joints constrained to act as a single functional unit (Tuller, Turvey, & Fitch, 1982). For example, research by Kelso and colleagues demonstrated the presence of a synergy formed between the upper and lower lip to make specific sounds during speech. When a perturbation was introduced during speech (an unpredictable downward tug to the jaw), to maintain the relationship with the upper lip, the lower lip begins to stretch, demonstrating the flexibly assembled coordinative structures in speech (Kelso, Tuller, Vatikiotis-Bateson, & Fowler, 1984). The formation of synergies enable ultrafast action and compensations to sudden and unexpected environmental perturbations (Riley, et al., 2012). This form of rapid skill adaptation process is useful for athletes seeking to reorganise coordination patterns as performance conditions suddenly change (Stone, Maynard, North, Panchuk, & Davids, 2017). In cycling this could be during a match sprint, when a cyclist has to suddenly respond to a rapid acceleration or tactical manoeuvre by their opponent. These synergies or coordinative structures are temporarily assembled under constraints of the task and environment to help achieve the goal of the task (Glazier & Davids, 2009; Riley, et al., 2012; Seifert et al., 2014).

Newell (1986) proposed that patterns of coordination emerge from the constraints imposed on the action. Constraints are boundaries or features that shape the organisation of these emergent coordination patterns. Newell proposed three categories of constraints that interact to influence the emergence of functional patterns of coordination and control for any activity. These are organism, task and environment constraints (Newell, 1986; Newell & Jordan, 2007) (Figure 2.1).

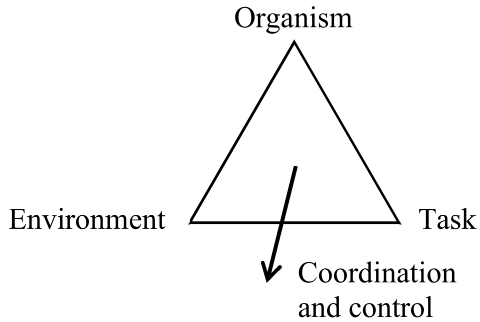

Figure 2.1: Newell's model of interacting constraints
A schematic diagram of the categories of constraints that specify the optimal pattern of coordination and control (Newell, 1986).

Organismic constraints reside at the level of the organism and are subdivided into structural and functional constraints. Structural constraints are relatively time independent (they change very slowly) such as, body height, mass, anthropometrics, muscle properties and architecture (Newell, 1986). Whereas, functional constraints have a relatively fast rate of change and can be physical or psychological such as, intentions, emotions, perception, decision-making and memory (Glazier & Davids, 2009; Newell, 1986). Environmental constraints are external to the organism and are generally not manipulated by the researcher and are relatively time independent such as, gravity, natural ambient temperature and natural light (Newell, 1986). In 2007, Newell and Jordan revisited the definition of environmental constraints and extended the definition to include any physical constraint beyond the boundary of the organism. Therefore, implements, tools or apparatus originally classified as task constraints were reclassified as environmental constraints under the revised model (Glazier & Davids, 2009; Newell & Jordan, 2007). Task constraints are related to the specific task and include the goal of the task and any specific rules or instructions that specify or constrain the response dynamics (Glazier & Davids, 2009; Newell, 1986). Many sports have rules that constrain the movement of the task such as, shot put or breaststroke swimming, and therefore, the performer's task is to optimise their performance within the imposed task constraints (Newell, 1985). One of the most profound conceptual implications of Newell's model of constraints is that functional patterns of coordination and control emerge from the interaction of the constraints acting on the neuromusculoskeletal system through a process referred to by Glazier and Davids as 'self-organising optimality' (Glazier & Davids, 2009). Therefore, the optimal pattern of coordination for a given task will be individual, as it emerges from the performer's unique set of constraints, i.e. for a given set of task and environmental conditions, different optimal patterns of coordination will emerge depending on the individual differences in organismic constraints (Newell, 1986).

Ecological dynamics as a theoretical framework is founded on two theories: dynamical systems and ecological psychology. The self-organisation of coordination patterns is often described using dynamical systems theory (Hristovski, Balague Serre, & Schollhorn, 2014). Dynamical systems are those that change over time. The system will converge into a stable state under a given set of constraints. This stable state is known as an attractor because it attracts all the nearby initial states of the system. The opposite is an unstable state called the 'repeller' which repels all initial states further away from it. A perturbation (or change in constraints) will cause the system to find a new stable state. Changing from one attractor state to another owing to a change of constraints is known as a phase-transition and this can happen quite suddenly (Hristovski et al., 2014). Gibson (1979) developed the ecological approach to psychology, with his theory of direct perception (Gibson, 1979; Turvey, 1990). The basis of this approach is perception is specific to information, and information is specific to the environment and one's movements (Gibson, 1979; Turvey, 1990). Ecological psychology assumes performer-environment mutuality and reciprocity so they combine to form a system (Araujo, Davids, & Hristovski, 2006; Turvey, 1990).

The theoretical framework of ecological dynamics combines the two theories to integrate biology and physics with psychology (Araujo et al., 2006). Ecological dynamics considers athletes as complex adaptive systems, and how such systems coordinate their actions with events, objects and surfaces in a performance environment (Araujo et al., 2006; Rothwell, Stone, Davids, & Wright, 2017). This makes it an appropriate framework for studying intermuscular coordination in maximal cycling.

2.4 Intermuscular coordination in cycling

How the motor system degrees of freedom, such as muscles are coordinated during functional movements is an issue that warrants further research. Pedalling is an ideal task for this purpose since it is a natural movement task that can be accurately manipulated (Neptune et al., 1997; Neptune & Kautz, 2001). During cycling the timing and magnitude of muscle activation has to be coordinated appropriately to allow an efficient energy transfer from the muscles though the segments to the pedal (Neptune & Kautz, 2001; Raasch et al., 1997). Although pedalling is a constrained lower limb task, it seems to require complex muscle coordination as evidenced by recorded EMG (Ryan & Gregor, 1992; So, Ng, & Ng, 2005). Furthermore, intermuscular coordination patterns, even in experienced cyclists, differ between individuals (Hug, Bendahan, Le Fur, Cozzone, & Grélot, 2004), although pedal force profiles remain similar (Hug, Drouet, Champoux, Couturier, & Dorel, 2008). Intermuscular coordination in cycling has been investigated both experimentally (Blake & Wakeling, 2012; Blake, Champoux, & Wakeling, 2012; Dorel, Guilhem, Couturier, & Hug, 2012) and computationally (Neptune & Hull, 1998; Raasch et al., 1997). These methods will be discussed in sections 2.7.1 and 2.7.2.

Many studies have investigated the effects of manipulating various constraints in cycling on the coordination pattern, which are detailed in Table 2.1.

Table 2.1: Summary of research studies investigating the effect of manipulating different constraints in cycling on coordination patterns

Constraint	Study	Variables measured
Saddle height	(Ericson, Bratt, Nisell, Németh, & Ekholm, 1986)	Joint kinetics (moments)
Saddle setback	(Menard, Domalain, Decatoire, & Lacouture, 2016)	Crank kinetics
Pedal-foot position	(Ericson et al., 1986)	Joint kinetics (moments)
Chainring shape	(Carpes, Dagnese, Mota, & Stefanyshyn, 2009; Cordova, Latasa, Seco, Villa, & Rodriguez-Falces, 2014; Hintzy, Grappe, & Belli, 2016; Rankin & Neptune, 2008)	3D joint kinematics Physiological variables and crank power Physiological variables and crank power Crank power, muscle mechanical work and activation
Crank length	(Barratt, Korff, Elmer, & Martin, 2011)	2D joint kinematics (angular velocities) and joint kinetics (powers)
Cycling position (upright or aerodynamic)	(Chapman et al., 2008; Dorel, Couturier, & Hug, 2009)	3D joint kinematics and EMG EMG, crank kinetics and physiological variables
Seated or standing	(Turpin, Costes, Moretto, & Watier, 2016; Turpin, Costes, Moretto, & Watier, 2017; Wilkinson, Lichtwark, & Cresswell, 2019)	EMG and crank kinetics EMG Crank kinetics, joint kinetics (powers) and EMG
Terrain (level or uphill)	(Sarabon, Fonda, & Markovic, 2012)	EMG
Pedalling rate	(Bieuzen, Lepers, Vercruyssen, Hausswirth, & Brisswalter, 2007; Blake & Wakeling, 2015; Ericson et al., 1986; McDaniel, Behjani, Brown, & Martin, 2014; Neptune et al., 1997)	EMG Crank kinetics and EMG Joint kinetics (moments) 2D joint kinematics (angular velocities, excursions) and joint kinetics (powers) EMG

Table 2.1: Summary of research studies investigating the effect of manipulating different constraints in cycling on coordination patterns (continued)

Constraint	Study	Variables measured
Intensity	(Blake & Wakeling, 2015; Dorel et al., 2012; Elmer, Barratt,	Crank kinetics and EMG
	Korff, & Martin, 2011; Ericson et al., 1986)	Crank kinetics and EMG
		2D joint kinematics (duty cycles) and joint kinetics (powers)
		Joint kinetics (moments)
Fatigue	(Billaut, Basset, & Falgairette, 2005; Bini, Diefenthaeler, & Mota,	Crank power and EMG
	2010; Brochner Nielsen et al., 2018; Martin & Brown, 2009;	2D joint kinematics and kinetics (moments)
	O'Bryan, Brown, Billaut, & Rouffet, 2014)	Crank and joint kinetics (powers), and EMG
		2D joint kinematics (joint excursions, duty cycles) and joint kinetics (powers)
		Crank power and EMG
Skill level	(Bini et al., 2016; Chapman, Vicenzino, Blanch, & Hodges, 2009)	3D joint kinematics
		3D joint kinematics and EMG

2.5 Sprint cycling and intermuscular coordination

To clarify the terminology used in this literature review of cycling: maximal cycling is a generic term referring to an all-out unpaced effort on a bicycle or an ergometer of typically less than 6 seconds to avoid metabolic fatigue (Gardner et al., 2007), whereas sprint cycling refers to the actual sport: short track cycling events or the sprint at the end of a cycling race.

Of the 28 cycling world-championship races governed by the Union Cycliste Internationale (UCI), 8 are all-out sprint events (men’s and women’s sprint, 500/1000 m time trial, Keirin, and BMX), 4 are often decided in the finishing sprint (men’s and women’s road race and scratch race) and 2 require repeated sprints (men’s and women’s points race) (Martin et al., 2007). Thus, sprint performance is a major determinant of most cycling world-championship racing events (Martin et al., 2007). In the Olympic Games the track cycling events that are classified as sprint are the match sprint, keirin and the team sprint.

Sprinting performances rely on a fast acceleration at the start of the sprint and on the capability to maintain the high velocity in the phase following the start (van Ingen Schenau, de Koning, & de Groot, 1994). One potential area to investigate that is key to a successful performance for track sprint cyclists is the acceleration phases that occur out of a starting gate or before the entry to the start of a flying 200 m. It is during these phases that the highest power outputs are produced (Dorel et al., 2005; Gardner, Martin, Barras, Jenkins, & Hahn, 2005; Martin et al., 2007). Another area is ‘getting a jump’ on an opponent in match sprinting, which is a rapid acceleration at an opportune time in the race, which typically occurs at speed, meaning peak power is a very important factor for cyclists to win match sprint races (Parsons, 2010). The maximum velocity phase of the flying 200 m is also another possible race phase to investigate (Dorel et al., 2005). An athlete’s maximal cycling power mainly depends on pedalling rate (cadence), muscle size, muscle-fibre type distribution, cycling position and fatigue (Martin et al., 2007). The highest recorded power output averaged for one pedal revolution is 2517 W (Martin, Gardner, Barras, & Martin, 2006). An important aspect of achieving high peak power output in cycling is the intermuscular coordination pattern (Dorel et al., 2012).

The typical intermuscular coordination pattern used in maximal cycling has been reported in the literature (Dorel et al., 2012; Raasch et al., 1997). Such muscle activation patterns are shown in Figure 2.2. When interpreting EMG activation patterns, researchers need to consider the electromechanical delay (EMD) - the time lag between the muscle activation and the muscle force production (Cavanagh & Komi, 1979; Hug & Dorel, 2009; Hug, 2011). There are various values reported in the literature for electromechanical delay from 30 ms to 100 ms (Cavanagh & Komi, 1979). The maximal sprint in Figure 2.2, was performed at 80% of optimal pedalling rate for maximum crank power production (f_{opt}), f_{opt} is typically around 130 rpm for elite track sprint cyclists (Dorel et al., 2005). Therefore, 80% of f_{opt} is approximately 104 rpm and assuming an EMD of 50 ms this equates to 31° of the crank cycle between EMG activation and muscle force production (Hug et al., 2008).

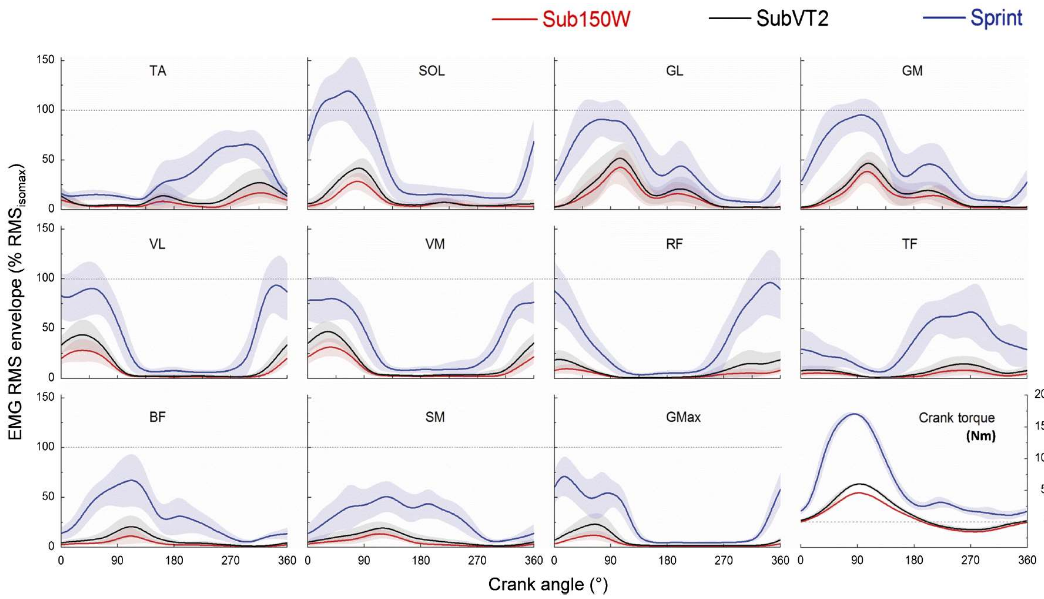

Figure 2.2: Muscle activation patterns of the lower limb muscles in maximal cycling. Ensemble-averaged EMG patterns of the 11 recorded muscles and crank torque profile. For the all-out sprint condition performed at 80% of f_{opt} (i.e. maximal cycling), refer to the blue line. The EMG and torque patterns were averaged across 6-7 consecutive pedal cycles and expressed as a function of the crank position (highest position: Top dead centre (TDC) = 0°). TA = tibialis anterior, SOL = soleus, GL = gastrocnemius lateralis, GM = gastrocnemius medialis, VL = vastus lateralis, VM = vastus medialis, RF = rectus femoris, TF = tensor fascia latae, BF = biceps femoris, SM = semimembranosus, GMax = gluteus maximus (Dorel et al., 2012, p2159).

Describing the role and activations of the muscles during maximal cycling, starting with the pedal at top dead centre (TDC = 0°), a force needs to be applied to the pedal in a forward horizontal direction and then vertically downwards. Therefore, the gluteus maximus, vastus lateralis and the vastus medialis start to activate before top dead centre to extend the hip and knee joints respectively in the downstroke. These uni-articular muscles are the primary power producers in cycling (Dorel et al., 2012; Raasch et al., 1997; Rankin & Neptune, 2008; van Ingen Schenau, Boots, De Groot, Snackers, & Van Woensel, 1992). To allow this force to be transferred to the pedal the ankle needs to provide a rigid link, so the activation of the uni-articular hip and knee muscles (gluteus maximus and vastii) need to be coordinated with the ankle plantar flexors (gastrocnemius and soleus) (Dorel et al., 2012; Raasch et al., 1997). The highest torque applied to the crank is at approximately 90° - although this will shift to slightly later in the downstroke with increased pedalling rate (McDaniel et al., 2014) (Figure 2.3). To stop the knee hyper–extending as the pedal approaches the bottom dead centre (BDC = 180°), the gastrocnemius and the hamstrings activate and start flexing the knee during the upstroke. The hip flexors (psoas and iliacus) then activate to start flexing the hip joint in the upstroke. Both these actions work to pull the pedal upwards. As the pedal reaches the second part of the upstroke (270°) the tibialis anterior activates to dorsi-flex the ankle. The hamstrings (biceps femoris long head, semimembranosus and semitendinosus) and the rectus femoris muscles activate to smooth the pedal stroke at the transitions at top dead centre and bottom dead centre (Raasch et al., 1997). Van Ingen Schenau and colleagues also identified that the bi-articular muscles have an important role to play in controlling direction of the force applied to the pedal (van Ingen Schenau et al., 1992). They identified that the paradoxical coactivation of the mono-articular agonists (vastii) with the bi-articular antagonists (hamstrings) is so the bi-articular muscles can help control the desired direction of the force applied to the pedal by adjusting the relative distribution of net moments over the joints (van Ingen Schenau et al., 1992). Also, over TDC there is coactivation of the mono-articular hip extensors (gluteus maximus) which deliver positive work with bi-articular antagonists (rectus femoris) which act to help control the direction of the force applied to the pedal (van Ingen Schenau et al., 1992). In maximal cycling, cyclists also use the actions of the upper body and torso to transfer power across the hip, as demonstrated by the hip transfer power (Elmer et al., 2011).

The resulting maximal cycling joint powers produced by these muscle activation patterns are shown in Figure 2.3 (McDaniel et al., 2014). Martin and Nichols demonstrated using simulated work loops that the joints have different roles and that during maximal cycling humans maximise muscle power at the hip and knee joints but the ankle acts to transfer (instead of maximise) power (Martin & Nichols, 2018). More specifically, the ankle works in synergy with the hip joint to transfer power produced by the muscles surrounding the hip joint to the crank (Fregly & Zajac, 1996). When investigating the influence of skill and performance level of cyclists on joint powers in maximal cycling, Barratt found no difference in relative joint powers during maximal cycling between world class track sprint cyclists and sub-elite cyclists (Barratt, 2014). However, the world class track sprint cyclists had greater cycling specific strength (peak joint moments at 60 rpm) in knee extension and flexion, and ankle extension and flexion compared to sub-elite cyclists (Barratt, 2014).

There are several key differences between maximal cycling compared to submaximal cycling. In maximal cycling knee flexion power is relatively more important and duty cycle values increase – i.e. the joints are in extension for a greater portion of the crank cycle which is an important strategy to increase maximum power (Elmer et al., 2011). Another difference between maximal and submaximal cycling is that, at optimal pedalling rates and below for maximum power production cyclists actively pull up during the upstroke, generating positive power, whereas in submaximal cycling the upstroke may be more passive (Dorel, Drouet, Couturier, Champoux, & Hug, 2009; Dorel, 2018b; Dorel et al., 2010). Dorel and colleagues found positive relationships between upstroke power and average power over a complete revolution, and between the index of mechanical effectiveness (IE - ratio of effective crank force to the total crank force) and power output during the upstroke in maximal cycling (Dorel, 2018b; Dorel et al., 2010).

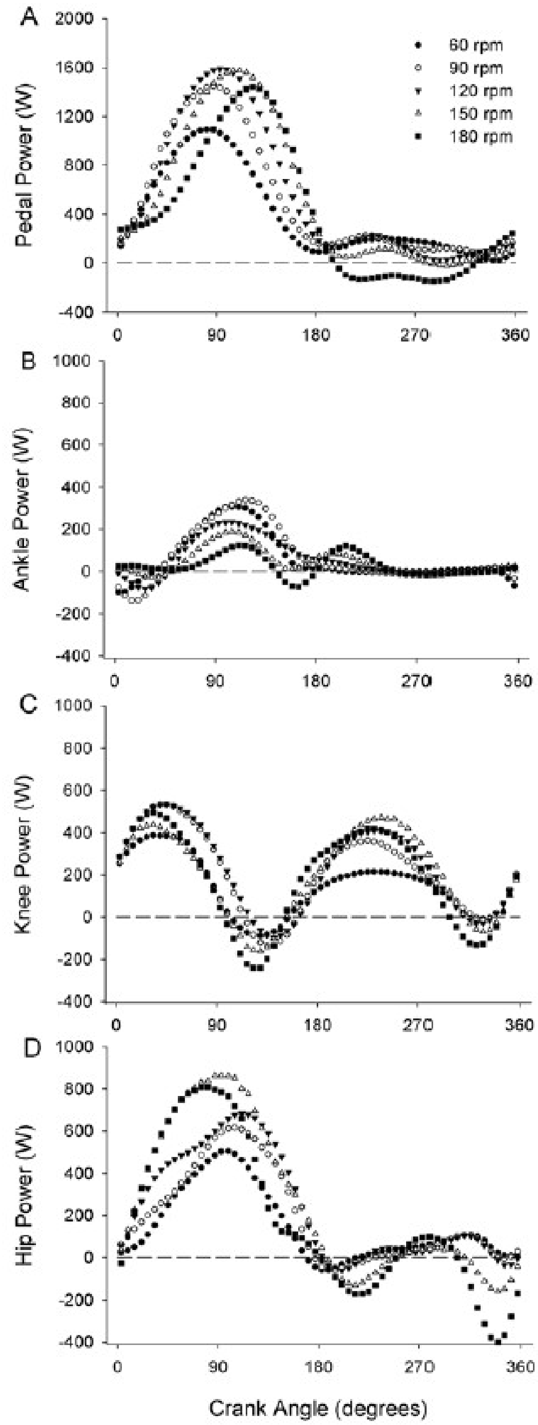

Figure 2.3: Pedal power and ankle, knee and hip joints powers during maximal cycling at different pedalling rates

From (McDaniel et al., 2014, p425)

Pedalling rate is a constraint that affects cyclists' intermuscular coordination patterns and joint powers (Dorel, Couturier, & Guilhem, 2011; McDaniel et al., 2014; Samozino, Horvais, & Hintzy, 2007). Primarily, pedalling rate affects muscle activation timings but not amplitude (Dorel et al., 2011; Samozino et al., 2007). McDaniel and co-workers demonstrated that, with increasing pedalling rate, relative ankle plantarflexion power decreased, whereas relative hip extension and knee flexion power increased (McDaniel et al., 2014). Relative knee extension power was not affected by changing pedalling rate (McDaniel et al., 2014). In addition, pedalling rate influences the maximal crank power cyclists can produce, which is described by a polynomial power-velocity relationship (Dorel et al., 2005; Dorel et al., 2011). The optimal pedalling rate (f_{opt}) is typically between 120 and 130 rpm (Dorel et al., 2005; Martin, Wagner, & Coyle, 1997), which matches the pedalling rate recorded at peak power by elite track cyclists during match sprint races (Gardner et al., 2005). In track sprint cycling pedalling rate is influenced by the choice of gear, which is a task constraint. As the cyclists get more powerful they can increase the gear size to maintain their optimum pedalling rate (Dorel et al., 2005). The power-velocity relationship suggests that the optimum pedalling rate is one that allows the muscles to contract close to their optimal shortening velocity (van Soest & Casius, 2000).

Another factor that influences optimal pedalling rate is activation-deactivation dynamics (the process of calcium release and reuptake from the sarcoplasmic reticulum) (van Soest & Casius, 2000), which mean a muscle cannot instantaneously produce maximal force at the beginning of a muscle contraction nor instantaneously relax at the end of a contraction (McDaniel et al., 2014). Van Soest and Casius used a simulation model to show the how activation-deactivation dynamics affect the optimal pedalling rate and maximum power output in maximal cycling (van Soest & Casius, 2000). When activation-deactivation dynamics were excluded from the simulation model the optimum pedalling rate increased from 120 to 200 rpm and the maximum power from 1076 W to 1754 W (van Soest & Casius, 2000). Their model demonstrates the large role activation-deactivation dynamics play in a cyclist's maximum power output. Neptune and Kautz then demonstrated that as pedalling rate increases the influence of activation-deactivation dynamics increases and they suggested it may be the governing muscle property that limits performance (Neptune & Kautz, 2001). This is because as pedalling rates increase, the time taken for a complete the crank cycle reduces, and therefore, due to activation-deactivation dynamics, it may not be possible to fully activate a muscle, or limit the proportion of the crank cycle a muscle can be fully activated (McDaniel et al., 2014). Therefore, intermuscular coordination strategies that limit the impact of activation-deactivation dynamics and maximise muscle force production in the downstroke are beneficial.

Another constraint that will influence the cyclist's intermuscular coordination pattern is fatigue, as the shortest track sprint cycling event is just under 10 seconds, and the longest around 60 seconds (Martin et al., 2007). Typically, in the literature only the first three seconds of an all-out sprint are assumed to be fatigue free (Martin et al., 2007; Martin & Brown, 2009). Power output in maximal cycling decreases with fatigue, and the reduction in power output with fatigue is greater at higher pedalling rates (Beelen & Sargeant, 1991; Martin et al., 2007). Martin and Brown demonstrated that during a 30 second maximal cycling effort fatigue occurred at different rates for the hip, knee and ankle joints (Martin & Brown, 2009). The power and range of motion of the ankle joint decreased more than at the knee and hip joints, which they suggested might be caused by two possible mechanisms: the cyclists trying to simplify the task by reducing the degrees of freedom, or that the ankle plantar flexor muscles fatigued faster than the other lower limb muscles (Martin & Brown, 2009). O'Bryan and co-workers also investigated the effect of fatigue on a 30 second maximal cycling effort and they found for the lower limb muscles the EMG amplitude reduced and the activation timings changed with fatigue (O'Bryan et al., 2014). In support of the findings of Martin and Brown, they found a significant reduction in medial and lateral gastrocnemius muscles' EMG activation levels with fatigue and reduced coactivation between gastrocnemius muscles and main power producing muscles (GMAX/VL/VM) (Martin & Brown, 2009; O'Bryan et al., 2014). However, this programme of research investigated short-term maximal cycling (4 second sprints), therefore, fatigue within a sprint was not a constraint that will influence the intermuscular coordination patterns.

Dorel and colleagues stated the role of intermuscular coordination in maximal cycling as a key factor contributing to limiting global power output (Dorel et al., 2012), which was highlighted when they found that the intrinsic muscle properties (muscle strength) only explained about 50% of the variance in force during cycling (Dorel, 2018b; Dorel et al., 2012). They also highlighted that intermuscular coordination plays an increasingly important role to achieve maximum power production at high pedalling rates, particularly those above f_{opt} (Dorel, 2018a; Dorel, Couturier, & Hug, 2014; Samozino et al., 2007). They recommended that further studies are required to investigate whether the intermuscular coordination pattern can be optimised with training (Dorel et al., 2012). Although the effect of changing various constraints on intermuscular coordination in cycling has been investigated, these studies all examined the effects of acute within-session interventions such as changes to bicycle set-up or riding position, intensity or fatigue levels. Consequently, there is a lack of research into how intermuscular coordination in cycling changes with training, and at different time points throughout the season. How intermuscular coordination adapts and changes with training is important for researchers and coaches to help them understand how training type influences intermuscular coordination, the mechanisms that underpin this and how coordination influences cycling performance. Training would be expected to change the cyclists' organismic constraints such as muscle size, strength, fatigue. Therefore, in accordance with the principles of ecological dynamics theoretical framework and Newell's model of constraints by changing the constraints acting on the athlete, new coordination patterns will emerge (Newell, 1986).

2.6 The relationship between strength training and intermuscular coordination in maximal cycling

Muscular strength and muscle size are constraints that will likely influence the functional pattern of coordination in cycling. There are many studies documenting that resistance training can lead to increases in muscular strength and size (Carroll et al., 2001; Zatsiorsky & Kraemer, 2006). The increase in muscle size is known as hypertrophy and is caused by the enlargement of the cross sectional areas of the individual muscle fibres (MacDougall, 2003; Zatsiorsky & Kraemer, 2006). Although resistance training increases the cross sectional area of all fibre types, most studies have indicated that a greater relative hypertrophy occurs in the Type II fibres which are designed for generating higher muscle power outputs (MacDougall, 2003).

Strength training also causes neural adaptations which can increase muscle strength. These include: recruitment or more consistent recruitment of the highest threshold motor units, increased motor unit firing rates and the synchronisation of motor unit firing that activate the muscle fibres within a muscle, collectively these are known as intramuscular coordination (Aagaard, Simonsen, Andersen, Magnusson, & Dyhre-Poulsen, 2002; Carroll et al., 2001; Sale, 2003; Zatsiorsky & Kraemer, 2006). Strength training can also induce changes in muscle architecture such as change in pennation angle and fascicle length which increase muscle strength (Aagaard et al., 2001; Cormie, McGuigan, & Newton, 2011a). Resistance training can also modify the connective tissue which is within and around the muscles and makes up the tendons by increasing the maximum tensile strength and amount of energy the tendons can absorb before failure (Stone, & Karatzaferi, 2003). Following a period of gym-based strength training, 'gym strength' (assessed by the amount of mass that can be lifted in non-specific strength exercises, such as the squat, deadlift and leg press) may increase not just because of changes in muscular and tendon properties but also by improvements in participants' intermuscular coordination during the gym exercise which allows them to lift greater load (Cormie et al., 2011a).

Elite track sprint and BMX cyclists routinely undertake strength training, to increase muscular strength with the aim of improving cycling performance (Parsons, 2010). They commonly use traditional resistance training exercises such as the squat, deadlift and leg press to increase muscular strength of the lower limbs (Parsons, 2010). The physiological adaptations typically responsible for the increases in maximal strength following strength training programmes consisting of traditional resistance training exercises are increases in whole muscle cross-sectional area (CSA) and neural drive (Cormie et al., 2011b). The traditional squat with high load is a popular exercise for sprint cyclists as it is a lower limb multi-joint movement that targets the muscles that cross the hip, knee and ankle joints (Farris, Lichtwark, Brown, & Cresswell, 2016). In particular, the squat exercise targets the hip extensor (gluteus maximus) and the knee extensor (vastii) muscles (Farris et al., 2016; Swinton, Lloyd, Keogh, Agouris, & Stewart, 2012). These are the main power producing muscles in maximal cycling (Dorel et al., 2012; Raasch et al., 1997).

There is little research into the influence of resistance training on muscle strength and peak power in sprint cyclists. However, elite track sprint cyclists have been found to have greater thigh girth and lower limb strength than endurance or sub-elite cyclists (Barratt, 2014; McLean & Parker, 1989), which have been found to correlate well with sprint cycling performance. For example, Dorel and colleagues found that the lean thigh volume of elite track sprint cyclists was positively correlated to maximum power (the apex of power-velocity relationship) (Dorel et al., 2005). These findings were supported by a study by Pearson and colleagues that showed peak cycling power was significantly correlated to lower limb lean volume in young and elderly men (Pearson, Cobbold, Orrell, & Harridge, 2006). 'Gym strength' (measured in this example by the isometric mid-thigh pull) has been shown to be strongly positively correlated to sprint cycling peak power and track sprint cycling times (Stone, et al., 2004). Individual joint torque has also been positively correlated to peak power output, with the magnitude of peak knee joint extensor torque being the best predictor of peak power output in sprint cycling (Kordi et al., 2017).

In support, therefore, of the empirical evidence positively correlating lower limb muscle size and strength to sprint cycling performance, coaches will typically place a large emphasis on gym-based strength training in specific training phases to improve these characteristics (Burnie et al., 2018; Parsons, 2010). However, although there is research demonstrating that gym-based strength training can increase muscle size and strength (Cormie et al., 2011b), there is a relative paucity of research to inform the optimum design and scheduling of gym-based strength training for track sprint cycling performance. For endurance cyclists there is conflicting evidence on the transfer of gym-based strength training to endurance cycling performance. Koninckx and colleagues showed that a programme of resistance training (parallel half squat and leg press [inclined at 45°]) increased endurance cyclists' peak power (11-15%) across a range of pedalling rates (40-120 rpm) (Koninckx et al., 2010). They did not measure whether this change was due to an increase in muscle cross-sectional area. A similar study was carried out by Ronnestad and colleagues, but they also measured thigh cross sectional area. They found a 12 week period of concurrent strength and endurance training increased thigh cross sectional area (4.6 ± 0.5%), maximal isometric force (21.2 ± 4.9%) and peak power in the Wingate test (9.4 ± 2.9%) (Rønnestad, Hansen, &

Raastad, 2010). However, in contrast, Bishop and colleagues found a gym-based strength training intervention in female endurance cyclists increased ‘gym strength’ (one repetition maximum (1RM) for concentric squat) but did not elicit any change in cycling performance variables (Bishop, Jenkins, Mackinnon, McEniery, & Carey, 1999). In strength-trained athletes, traditional weight training improved peak power in a 6 second maximal cycling test (Wilson, Newton, Murphy, & Humphries, 1993). These studies all had notable limitations for researchers interested in track sprint cycling because participants either had little strength training experience (endurance cyclists), or the participants were not cyclists. Consequently, the results from these studies might not be applicable to track sprint cyclists.

It is, therefore, clear that the transfer of strength training to performance in sport can vary dramatically (Carroll et al., 2001; Young, 2006). Generally there is positive transfer to sports performance although sometimes there is negative transfer, i.e. the strength training is detrimental (Carroll et al., 2001; Young, 2006). The negative transfer was demonstrated by Moir and colleagues who found a gym-based strength training intervention worsened 20 m acceleration time in physically active men (Moir et al., 2007). Consequently, Carroll and colleagues state that research is required to understand how the physiological adaptations associated with resistance training transfer to sporting performance (Carroll et al., 2001).

The training principle of specificity states the closer the resistance training resembles the sporting movement, the greater the transfer of strength (Bosch, 2015; Carroll et al., 2001; Young, 2006; Zatsiorsky & Kraemer, 2006). It implies that the intermuscular coordination is an important component in achieving transfer to the sport (Young, 2006). It has been stated for elite athletes the specificity of the training needs to increase for adaptations and improvements in sports performance to continue (Zatsiorsky & Kraemer, 2006). It has also been suggested that in order to achieve training specificity load should be added to a sporting movement, for example resisted running by pulling a sledge (Young, 2006).

The following two studies support the training principle of specificity. Leirdal and colleagues compared two different training regimens on the effect on vertical jumping performance (Leirdal, Roeleveld, & Ettema, 2007). One group performed squats and plantar flexions separately and the other group squats ending in plantar flexion, for both groups the squats were performed with no load and were a “squat jump type” movement. When both groups were retested they had increased their peak power but there was no change in vertical jump height. However, the two groups used different coordination strategies, identified by measuring the muscle activity of the vastus medialis and gastrocnemius medialis. The group who performed squats ending in plantar flexion showed training movement specific coordination effects that may be a forerunner to improvements in vertical jumping (Leirdal et al., 2007). This shows how training can influence intermuscular coordination and potentially aid transfer. Rumpf and colleagues reviewed studies on the effect of different types of training on sprinting performance and found that specific sprint training (free sprinting, resisted running or downhill running) was more effective than non-specific training (gym-based strength training, plyometrics, power training) (Rumpf, Lockie, Cronin, & Jalilvand, 2016). In contrast to this Koninckx found non-specific resistance training improved endurance cyclists’ peak power over a range of pedalling rates (Koninckx et al., 2010). However, this may be because the endurance athletes would not normally use resistance training and the novel stimulus encouraged adaptation. Therefore, this may not apply in highly trained track sprint cyclists who regularly use resistance training.

Bosch developed upon the movement specificity principle further by describing it in terms of ecological dynamics theoretical framework (Bosch, 2015). He described human movement patterns as made up of stable and variable components so they can be adapted to suit the dynamics of changing performance environments. The fixed components (attractors) are stable and economical. The changeable components (fluctuators) are variable and have high energy costs. It is extremely difficult to tell which components of movement are fixed and which are changeable.

Bosch stated that movement specificity can be divided into 5 components:

- Similarity of movement owing to similarities in the internal structure of the movement
- Similarity of movement owing to similarities in the external structure of the movement
- Similarity of movement owing to similarities in energy production

- Similarity of movement owing to similarities in sensory response
- Similarity of movement owing to similarities in the intention of the movement

To achieve transfer of strength training to sporting movement Bosch speculated that the majority of these components need to be met (Bosch, 2015). The traditional approach to strength training comes from the body building approach where the training is focused on an individual muscle group (body part method). This focuses on the physiological adaptations and not coordination of movements and so transfer occurs less successfully to sport-specific actions. Bosch (2015) highlighted the benefits of contextual strength training which is specific to coordination patterns required for successful performance in a sport like cycling. These exercises consist of attractor and varying components, with the attractors seen as the building blocks of movement and unchangeable and the varying movement components facilitating movement adaptations. Strength training can improve the attractor components, whilst also maintaining specificity to the sporting movement. One point to note is that the organisation of movements from stable to unstable patterns may suddenly change via a phase transition. A phase transition is where the system suddenly jumps from one arrangement to another. An example is the transition between walking and running, where there is a sudden change in coordination and movement organisation patterns (Bosch, 2015). Therefore, specificity between low and high intensity movements is not guaranteed. He proposed using a constraints-led approach where the constraints in strength training are varied to create overload and stimulate the emergence of new movement and coordination patterns (Bosch, 2015).

Strength training also changes several other constraints such as fatigue, because heavy periods of resistance training induces fatigue that accumulates over time (Zatsiorsky & Kraemer, 2006). Therefore, athletes often require a period of rest or reduced training load to reduce fatigue to see the benefits of the strength training (Mujika & Padilla, 2003; Zatsiorsky & Kraemer, 2006). This period can be between 2 and 6 weeks (Mujika & Padilla, 2003; Zatsiorsky & Kraemer, 2006). Athletes also reduce training load before a competition which is defined as the taper, with the primary aim to reduce accumulated fatigue to optimise sports performance (Mujika & Padilla, 2003). The scientific literature on training focuses on physiological and psychological changes during the taper and not changes in intermuscular coordination of the sporting movement (Mujika & Padilla, 2003). However, by changing the athlete's organismic constraints (i.e.

fatigue) during the taper period, based on the ecological dynamics theoretical framework and Newell's model of constraints, it would be expected that new coordination patterns would emerge (Newell, 1986). Therefore, an understanding of this process may help coaches decide on the appropriate taper length for their athletes, as currently many coaches rely on their experience and use a trial and error approach to determine the optimum taper for an athlete (Mujika & Padilla, 2003). The taper length is very individual which fits within the ecological dynamic's theoretical framework as each athlete's coordinative patterns emerge from his/her unique set of constraints (Kelso, 2014; Zatsiorsky & Kraemer, 2006).

Intermuscular coordination may explain the varied transfer of strength training to sport performance in two ways. First, increases in muscle strength from strength training may need to be accompanied with a change in intermuscular coordination to improve sport performance. Congruent with the tenets of ecological dynamics theoretical framework and Newell's model of constraints, Bobbert and Van Soest used a musculoskeletal simulation to show that an increase in leg strength (an organismic constraint) must be accompanied by a change in intermuscular coordination in order for vertical jump height to increase (Bobbert & Van Soest, 1994; Newell, 1986). Second, muscle recruitment patterns associated with a strength training task could retard sports performance when expressed during the sport movement (Carroll et al., 2001). Traditional resistance exercises have very different intermuscular coordination patterns to cycling which may impair transfer of the training effects to maximal cycling power (Koninckx et al., 2010). For instance when executing a squat a stable knee joint is very important to order to decelerate the load at the end of the range of motion (Cormie et al., 2011b), to achieve this there is significant co-contraction of the hamstrings and quadriceps (Gullett, Tillman, Gutierrez, & Chow, 2009; Slater, & Hart, 2017). This is different to coordination patterns required for cycling where only a smaller amount of co-contraction is required at the knee joint to help control the direction of force applied to the pedal (Dorel et al., 2012; van Ingen Schenau et al., 1992). In this way, extensive non-specific strength training could impair pedalling coordination such that cycling performance is reduced.

2.7 Measurement techniques and analysis methods

2.7.1 Computational approach to study coordination in maximal cycling

A computational modelling approach can be used to investigate intermuscular coordination in cycling. In the last 30 years there has been great interest in neuromusculoskeletal modelling in the biomechanics research community (Hicks, Uchida, Seth, Rajagopal, & Delp, 2015). Models have been used to understand the biomechanical principles of a movement and to identify particular areas of the movement technique that could be changed to improve performance (Bobbert & Van Soest, 1994; Neptune & Hull, 1995; Yeadon & King, 2007). Musculoskeletal modelling has been used to investigate biomechanics and coordination in cycling. Fregly and Zajac used this approach to study mechanical energy generation, absorption and transfer during pedalling (Fregly & Zajac, 1996). They were able to identify that the net hip and ankle extensor torques act in synergy to deliver energy to the crank during the downstroke (Fregly & Zajac, 1996). The net hip extensor torque generates energy to the limb while the net ankle extensor torque transfers this energy from the limb to the crank (Fregly & Zajac, 1996). This work was extended by Raasch and colleagues who used a forward dynamical musculoskeletal model to investigate the individual muscle contributions to energy generation, transfer and absorption in maximum speed pedalling (Raasch et al., 1997). In support of the findings by Fregly and Zajac, they found the uni-articular hip and knee extensors (GMAX/VL/VM) provide 55% of the propulsive energy, only 44% of which is delivered directly to the crank in the downstroke, while the other 56% is delivered to the limb and transferred to the crank by the ankle plantar flexors (GAS/SOL) (Raasch et al., 1997). A similar forward dynamic simulation was used by Korff and Jenson to study age-related difference in muscular power production in cycling between adults and children (Korff & Jensen, 2007). Computer simulation studies have investigated the relationship of the bicycle set-up (crank length, seat height, seat tube angle, chainring shape) with intermuscular coordination and maximum crank power produced (Rankin & Neptune, 2008; Rankin & Neptune, 2010; Yoshihuku & Herzog, 1990). Simulation modelling has also been used to demonstrate the role of muscle activation-deactivation dynamics on intermuscular coordination patterns in cycling (Neptune & Kautz, 2001; van Soest & Casius, 2000). These studies demonstrate the potential of musculoskeletal modelling to advance our understanding of coordination and the factors that influence it.

Computer modelling approaches such as forward dynamic analyses or simulations have several advantages for studying biomechanics and coordination of human movement. Forward dynamics simulations can optimise for a function - for example producing maximum crank power in cycling (Raasch et al., 1997) - and therefore can find the theoretical optimum sports technique and coordination pattern to achieve the task (Yeadon & King, 2007). The chosen optimisation function is often based on achieving the goal of the task, e.g. in sprint cycling maximising crank power as this maximises speed of the bicycle for a given bicycle set-up. Computer modelling also allows ideal experiments to be carried out, i.e. it is possible to change just one variable (Yeadon & King, 2007). Another advantage of computer modelling is it not influenced by external factors such as the environment (weather conditions), athlete motivation or fatigue, as an experimental data collection can be. Also it is easy to change the parameters within the model, for example Bobbert and Van Soest investigated effect of muscle strength on vertical jump height (Bobbert & Van Soest, 1994). They were able to increase the muscle strength of the knee extensor muscles by 5, 10 and 20%. If an experimental approach had been chosen to study this research question, the participants would have to undergo a strength training intervention to increase knee extensor strength which would have been time consuming (Bobbert & Van Soest, 1994). Computer simulation also allows theories to be tested before asking athletes to attempt the skill or change their technique. This was approach was used by Hiley and Yeadon to demonstrate that a triple straight somersault dismount from the high bar was theoretically possible before it had been attempted by a gymnast (Hiley & Yeadon, 2005).

However, although there are many advantages of using a computer modelling and forward dynamics simulations approach, there are several important limitations. They often have limited real world impact, owing to the challenge of validating the findings and ensuring their accuracy and reliability (Hicks et al., 2015). Researchers need to be confident that they have found a global optimum rather than the local optimum solution to the task i.e. the summit of the highest mountain rather than the top of a foothill (Glazier & Davids, 2009; Yeadon & King, 2007). It is also important to evaluate the model and to ensure the model behaves in a realistic manner. This is typically done by comparing the output of the forward dynamic simulation to experimental data and assessing the similarity of the results (Yeadon & King, 2007). It can, therefore, be argued that the model is only valid for very similar movements and constraints of the experimental data used to evaluate the model. Consequently, using forward dynamic simulations to predict the outcome of changing input parameters or optimising for a solution might not be valid.

When researchers are building a computer model of the neuromusculoskeletal system to study human movement many idealisations, assumptions and simplifications needs to be made (Hicks et al., 2015; Yeadon & King, 2007). These include the behaviour and degrees of freedoms of the joints, and the control, behaviour and properties of the muscles within the model (Hicks et al., 2015; Yeadon & King, 2007). The decision also needs to be made whether to use a torque driven model or a muscle-actuated dynamic model. It is possible to make a torque driven model participant specific by measuring joint torques on a dynamometer (Yeadon & King, 2007). However, to obtain joint torques for the full joint range of motion at different angular velocities is a time consuming process and physically exhausting for the athlete (Yeadon, King, & Wilson, 2006). A limitation of a torque driven model is that these do not describe what is happening at a muscle level, i.e. muscle activation and force production. Musculoskeletal models such as OpenSim (Delp et al., 2007) allow for the calculation of muscle forces and activations. The OpenSim musculoskeletal model has been adapted for cycling motion (Lai, Arnold, & Wakeling, 2017). However, it has been adapted for recreational endurance cycling position and not for the aerodynamic position that track sprint cyclists' adopt, which is riding with a very shallow torso angle (evidenced in Appendix 9.2), and closed hip angle around the TDC position (Heil, 2002). Therefore, owing to the limits on the joint ranges of motion, this model is not suitable for studying track sprint cycling.

It is also, difficult to make the models participant specific, because the muscle properties (force-length, force-velocity and passive properties) in the Hill-type muscle model are typically derived from experiments on rat, cat and rabbit muscles (Hicks et al., 2015). Therefore, these muscle properties may not be suitable for modelling human muscles and in particular elite athletes whose muscle properties differ from the normal population. Yoshihuku and Herzog demonstrated that the calculated maximum cycling power from a lower limb model was sensitive to the muscle model and muscle properties used – 1000 to 1300 W (Yoshihuku & Herzog, 1996). Also, to model a cyclists' movement pattern within the framework of ecological dynamics, the model would need to incorporate the organismic, environment and task constraints as these shape an athlete's coordination pattern, which currently is beyond the capability of models of human movement (Glazier & Davids, 2009).

Therefore, owing to the limitations highlighted above computational methods such as forward dynamic analyses or simulations where not chosen to study intermuscular coordination in cycling. This is, in particular owing to their limitations when trying to model a complex dynamic system such as an elite athlete (Glazier & Davids, 2009). Therefore, an experimental approach was chosen to study the effect of strength training on intermuscular coordination in maximal cycling because the changes that occur when elite track sprint cyclists undertake a strength training programme can be measured and observed.

2.7.2 Experimental approach to study coordination in maximal cycling

A cyclist's intermuscular coordination can also be investigated experimentally by measuring muscle activity using EMG to determine muscle activation onset and offset times and level of activation (Dorel et al., 2012; Hug & Dorel, 2009), or by carrying out a mechanical analysis to calculate the joint kinetics at the hip, knee and ankle throughout the pedal revolution (Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014). Combining information on muscle activation from EMG and joint kinetics from inverse dynamics analysis allows for a deeper understanding of the joint and muscle actions that produce the movement (Brochner Nielsen et al., 2018; Dorel, 2018b). The relative contribution of the individual joint powers to the net pedal power can be used to investigate different strategies between cyclists – Gregor and colleagues documented the high inter-participant variability of joint moment patterns in cycling (Broker & Gregor, 1994; Gregor, Broker, & Ryan, 1991; Gregor, Cavanagh, & LaFortune, 1985), when cycling at different net powers (Elmer et al., 2011; Skovereng, Ettema, & van Beekvelt, 2016) and pedalling rates (McDaniel et al., 2014). An advantage of an experimental approach is that the actual forces applied to the pedal, the joint and segment movements and muscle activity can be measured. Therefore, the effect of the interaction of the changing constraints on the athlete coordination patterns is inherently included. The methods used to measure the kinematics, kinetics and muscle activity during maximal cycling are discussed in the following sections.

2.7.3 Kinematics

The kinematics - the movement patterns of the lower limbs and pelvis are required to study intermuscular coordination in cycling. Chapman and colleagues used kinematics and muscle activity to investigate whether the body position influenced muscle recruitment (Chapman et al., 2008). There are a variety of methods available to capture kinematic data in biomechanics. Examples of these systems are: high speed video cameras, passive or active marker motion capture camera systems, electromagnetic tracking systems, inertial measurement units (IMUs) and bespoke systems such as instrumented spatial linkage (Martin, Elmer, Horscroft, Brown, & Shultz, 2007). Each of these systems has advantages and disadvantages when being used to capture the kinematics of a cyclist riding on the track in a velodrome. These are detailed in Appendix 9.1 - Table 9.1. One problem which affects all the motion capture systems is that they measure what is happening at the surface of the body and not at the actual joints. Therefore, one of the biggest sources of error of all marker based systems is anatomical marker misplacement and soft tissue artefact (STA) (Della Croce, Cappozzo, & Kerrigan, 1999; Della Croce, Leardini, Chiari, & Cappozzo, 2005; Leardini, Chiari, Della Croce, & Cappozzo, 2005; Neptune & Hull, 1995).

High speed video cameras

High speed video cameras systems are often used to measure 2D kinematics of a sporting movement. They are suitable for measuring kinematics of movements that typically occur in one plane such as cycling and in a small capture volume, such as on an ergometer or treadmill. Barratt used a high speed video camera to measure 2D kinematics of the lower limb during maximal cycling on an ergometer (Barratt, 2014). They also have the advantages that they are easy to use, relatively low cost, cause minimal interference for the performer and can provide visual feedback (Payton, 2007). To enable camera image pixels to be converted into metres to calculate coordinates, a recording of the scaling objects in vertical and horizontal dimensions are required (Payton, 2007). High speed video cameras can be used in conjunction with infra-red ring lights to enable the tracking of passive reflective markers. There are several automated coordinate digitiser programmes such as Quintic Biomechanics v31 (Quintic

Consultancy Ltd, Birmingham, UK) or CrankCam (Centre for Sports Engineering Research (CSER), SHU) which can track and calculate marker coordinates to speed up data processing which is an important consideration when collecting many trials at a high sampling frequency (Payton, 2007).

Passive marker motion capture camera systems

Passive marker motion capture systems such as Qualisys AB (Goteborg, Sweden) and Vicon (Oxford Metrics, Oxford, UK) have been used by many researchers to measure cycling kinematics (Brochner Nielsen et al., 2018; Chapman et al., 2008; Wilkinson et al., 2019). The advantage of passive marker based systems is they only require small reflective markers to be placed on the participant and they are wireless so there is minimal interference for the participant when performing a task. A disadvantage is that they have a limited capture volume when used outside of the laboratory (Adesida, Papi, & McGregor, 2019; Pueo & Jimenez-Olmedo, 2017). Therefore, to capture cycling in the velodrome where the bicycle moves through the capture volume a large number of cameras would be required. Passive marker-based systems can also have a problem with marker occlusion, during cycling the knee marker can often be obscured around TDC (in the sagittal plane view the elbow can obscure the knee joint). This is particularly an issue when the riders are in position to minimise an aerodynamic drag - torso fully bent over (parallel to the ground) with hands on the drops portion of the handlebars and the elbows flexed (Heil, 2002). Elite and club level track cyclists will all adopt this position. However, this can easily be solved as motion capture systems have algorithms that can fill the gap in the marker trajectories where the markers are obscured for a small part of the motion – typically less than 10 frames (Qualisys AB, Goteborg, Sweden).

Electromagnetic tracking systems

Electromagnetic tracking systems are composed of sensors containing 3 small electric coils which move within an electromagnetic field created by a source box (Pueo & Jimenez-Olmedo, 2017). The location and orientation of the sensors relative to the source box is calculated due to the coils generating a small voltage or current when moving inside a constant magnetic flux (Pueo & Jimenez-Olmedo, 2017). Electromagnetic tracking systems such as Polhemus G4 (Polhemus, Vermont, USA) have been reported to interfere with the EMG data collected at same time (Pidcoe, 2001). The electromagnetic signals contaminate the EMG data so the onset of muscle activation cannot be determined (Pidcoe, 2001). The electromagnetic signal can be removed from the EMG data by applying notch filters. However, depending on the electromagnetic tracking system used several notch filters may need to be applied which will remove the EMG data at these frequencies as well (Pidcoe, 2001). In addition, any metal objects in the vicinity of the sensors may distort the magnetic field created by the electromagnetic tracking source box, thereby ruining the measurement accuracy (Pueo & Jimenez-Olmedo, 2017). As a bicycle's drivetrain is made of metal this could distort the magnetic field affecting the accuracy of the measured pedal spindle and ankle joint coordinates. This system is therefore unsuitable for measuring kinematic data during cycling.

Inertial measurement units

Inertial sensor measurement unit systems such as Xsens MVN (Xsens Technologies B.V., Enschede, the Netherlands) use sensor fusion algorithms to estimate the displacements, and rotations of the body segments from accelerometer, gyroscope and magnetometer data measured by the inertial measurement units (IMUs) (Roetenberg, Luinge, & Slycke, 2013; van der Kruk & Reijne, 2018). The magnetometer in the IMUs provide stability in the horizontal plane by using the direction of the earth's magnetic field and are used to correct the integration drift associated with the accelerometers (Roetenberg et al., 2013). Therefore, the data obtained from the Xsens system are only an estimate of the kinematic variables. The magnetic sensors can be disrupted by ferrous metal. This was a problem for Cockcroft who used IMUs to measure a cyclist's kinematics, the IMU sensors near the pedals and handlebars suffered severe magnetic interference which affected the ankle and arm kinematics (Cockcroft, 2011). The calculated kinematics are also susceptible to errors introduced by integration of acceleration data to obtain positional data (Adesida et al., 2019; van der Kruk & Reijne, 2018). IMU systems generate whole body kinematics but the person is not located in space (van der Kruk & Reijne, 2018). As such, to link the position of the cyclist relative to the bicycle moving on track in the velodrome, the data from the IMUs would have to be combined with local positional information from a positional system within the velodrome (Zuiker, 2014). The location of the cyclist relative to the bicycle is required for inverse dynamics calculations which need the location of the applied force to the pedal.

Instrumented spatial linkage system

The bespoke instrumented spatial linkage system developed by Martin and colleagues measures the position of the anterior superior iliac spine (ASIS) which can be used to infer the position of the hip joint centre (Martin et al., 2007; Neptune & Hull, 1995). To determine the lower limb kinematics the position of the pedal spindle, ankle joint and knee joint are required. The location of the ankle joint can be determined by the angular orientation of the crank and pedal, and length from the pedal spindle to the lateral malleolus, and by assuming the vector between these two points is fixed throughout the pedal cycle (Hull & Jorge, 1985). Once the location of the hip and ankle joints are known as well as the thigh and shank segment lengths, then the position of the knee joint centre can be determined by the law of cosines (Martin et al., 2007; Martin & Brown, 2009). This system was designed for measuring kinematics on a cycle ergometer in a laboratory and the linkage needs to be fixed to the floor. The linkage could potentially be modified to be fixed to the bicycle seat post. However, the linkage would also be fixed to the cyclist, so if they were to crash while riding on the track in the velodrome, they could potentially be injured by the linkage, making it unsuitable for field testing.

3D vs 2D kinematics

Most of the studies of cycling kinematics only consider the movement in the sagittal plane, assuming the movements in the other directions are negligible (van Ingen Schenau, Van Woensel, Boots, Snackers, & De Groot, 1990). Researchers have typically used 3D joint movements and moments to investigate the potential causes of knee injuries in cyclists, as the knee can move up to 2 cm medially during the downstroke (Ericson, Nisell, & Ekholm, 1984; Gregersen & Hull, 2003; Gregor et al., 1991; Ruby, Hull, & Hawkins, 1992). Umberger and colleagues tested the planar assumption during seated submaximal ergometer cycling (Umberger & Martin, 2001). They concluded that the 2D sagittal plane kinematics were similar to the respective angles measured in 3D, as long as care was taken in defining the hip angle. The range of motion of all the joints was greatest in the sagittal plane (Table 2.2) (Umberger & Martin, 2001). An advantage of collecting 2D kinematics is a simple marker set can be used with markers on the pedal spindle, ankle (lateral malleolus), knee (lateral femoral condyle), and hip (greater trochanter). This means participant preparation time in data collection sessions is much shorter than when using a full 3D kinematic marker set where 28 to 45 reflective markers have been used (Bini et al., 2016; Brochner Nielsen et al., 2018; Wilkinson et al., 2019). This is a particularly important consideration when elite athletes are participants, as they have limited time available for testing sessions. Also, sagittal plane markers can be recorded on one high speed video camera, therefore reducing the time required for laboratory set-up and camera calibration.

Table 2.2: Approximate range of motion of the hip, knee and ankle joints in the sagittal, frontal and transverse planes during submaximal cycling on a bicycle ergometer.

	Approximate range of motion (°)		
	Hip	**Knee**	**Ankle**
Sagittal	45 Flexion / extension	75 Flexion / extension	15 Dorsiflexion/ plantarflexion
Frontal	6 Abduction / adduction	11 Abduction / adduction	6 Inversion / eversion
Transverse	7 Medial / lateral rotation	15 Medial / lateral rotation	10 Abduction / adduction

Adapted from (Umberger & Martin, 2001)

Measuring hip joint centre during cycling

Neptune and Hull investigated the different methods for determining hip movement in seated submaximal cycling (Neptune & Hull, 1995). They developed a new method owing to the inherent inaccuracy of a marker placed on the superior aspect of the greater trochanter (GT) because of soft tissue artefact (STA) and pelvis rotation. Their method consisted of putting a marker on the anterior-superior iliac spine (ASIS) and determining the hip joint centre by calculating a vector of fixed magnitude and orientation in the sagittal plane between the ASIS and the GT. This assumes no rotation of the pelvis in the sagittal plane. This method was more accurate than the marker on the GT when compared to an intracortical pin fixed to the lateral iliac crest (IC). This highlights the potential for error when measuring the hip joint location. There have been no studies investigating the location of the hip joint centre in maximal cycling. It would be expected that there would be greater movement of the pelvis (pelvic tilt, obliquity (rocking) and rotation) in seated maximal cycling than in seated submaximal cycling on an ergometer (experimental evidence of the greater pelvic tilt during maximal cycling is presented in Appendix 9.4). When a cyclist is sprinting on a moving bicycle on the track their pelvis movement could be larger due to the mediolateral movement of the bicycle frame. Therefore, the marker set used to measure kinematics of the lower limbs during track sprint cycling needs careful consideration. Refer to section 4.2 and appendices 9.4 and 9.5 for details of chosen marker set used in this programme of research.

2.7.4 Kinetics

The kinetics - the magnitude and direction of the force the cyclist applies to the pedal, and the joint reaction forces and moments of the lower limbs are also required to study intermuscular coordination.

There are three methods used in biomechanics to measure the kinetic data of cycling: force cranks which measure the force applied to the cranks, force pedals which measure the forces applied to the pedals, and shoe pressure insoles (refer to Appendix 9.1 - Table 9.2 for details of these systems). Examples of these systems which are commercially available are: Factor Power Measurement Track Cranks (bf1 systems, Norfolk, UK) which measure the 2D force applied to the crank, torque applied to the crank and crank position; force pedals – model ICS4 (Sensix, Poitiers, France) which measure the three force components (Fx, Fy, Fz) and three moment components (Mx, My, Mz) on the pedal with additional devices (encoders) required to measure crank and pedal angle; pressure insoles – Pedar (Novel, Munich, Germany) which measure the pressure distribution on the shoe insole. The commercially available pedals are very similar in the design to the track force pedals designed by Drouet and colleagues (Drouet, Champoux, & Dorel, 2009). The pedals designed by Drouet and colleagues were used to investigate the relationship between crank forces, power and index of force effectiveness for elite track sprint cycling when performing all out efforts on track in the velodrome, demonstrating that force pedals can measure kinetics on track (Dorel, Drouet, Hug, Lepretre, & Champoux, 2008).

Kinetic data can also be measured at various points on the bicycle: such as the chain, back wheel hub, and bottom bracket (Driss & Vandewalle, 2013). However, in biomechanics the interest is how the cyclists apply force to the pedals as this information is required for the inverse dynamics calculations. The force pedals are the most suitable measurement system for the kinetics in maximal cycling owing to the high forces being applied to the pedal and the convenience of being able to switch them between participants track bicycles when collecting on track data. The pedal forces measured using the force pedals can then be input into the inverse dynamics calculations to calculate the joint reaction forces and moments.

2.7.5 Inverse Dynamics

To calculate the net joint forces and net joint moments in the lower limbs, inverse dynamics techniques can be used, which were first developed by Elftman for studying human locomotion (Elftman, 1939). This method has been applied to cycling by a variety of researchers using slightly different assumptions (Ericson, 1986; Gregersen & Hull, 2003; Hull & Jorge, 1985; van Ingen Schenau et al., 1990) and has been used to investigate the relative contribution of hip, knee and ankle joints to the pedal power and force (Elmer et al., 2011; Kautz & Hull, 1993; Martin & Brown, 2009; McDaniel et al., 2014; van Ingen Schenau et al., 1990).

Newtonian mechanics are applied to each individual segment starting with the foot (as the pedal reaction forces are known), refer to Figure 2.4. The sum of all the external forces that acted on a segment are taken as equal to the product of the mass of the segment and the translation acceleration of the segment centre of gravity (obtained from kinematic data) for the vertical and horizontal directions. The sum of all the moments that acted about the segment centre of mass must equal the rate of change of the angular momentum. These three equations can then be solved for the intersegmental forces and moments at the ankle. This process is repeated for the shank and the thigh. The joint powers are calculated as the dot product of the net joint moments and joint angular velocities.

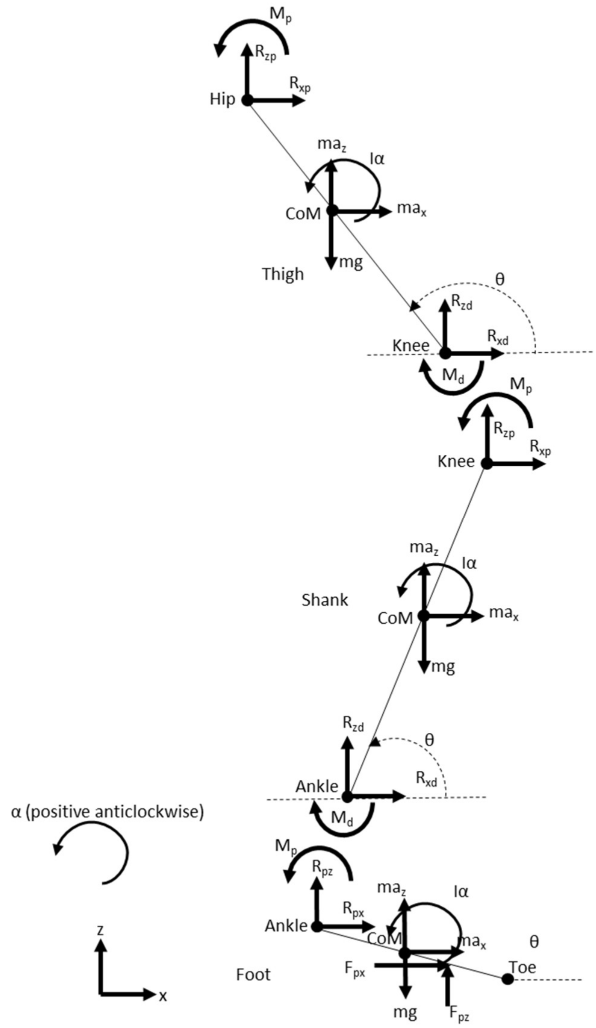

Figure 2.4: Free-body diagram of the body segments of a cyclist's leg

The leg is subdivided into three rigid links. For an explanation of the symbols: CoM = centre of mass of segment, m = mass of segment, g = gravitational acceleration, a_z = linear acceleration of CoM of segment in z direction, a_x = linear acceleration of CoM of segment in x direction, F_{pz} = vertical pedal reaction force, F_{px} = horizontal pedal reaction force, I = principal moment of inertia, θ = segment angle (angle convention - anticlockwise from horizontal), α = segment angular acceleration, R_{pz} = proximal joint reaction force in z direction, R_{px} = proximal joint reaction force in x direction, R_{dz} = distal joint reaction force in z direction, R_{dx} = distal joint reaction force in x direction, M_p = proximal joint moment, M_d = distal joint moment. Adapted from (van Ingen Schenau et al., 1990, p17).

2.7.6 Anthropometrics

The inverse dynamics calculations require body segment parameters such as segmental masses, moments of inertia and location of the mass centres. There are various methods for estimating these parameters which include: tables derived from cadaver studies (Dempster, 1955), geometric models of the human body (Yeadon, 1990), and from regression equations derived from mass scanning of young adults (de Leva, 1996; Zatsiorsky & Seluyanov, 1983). These methods require some anthropometric measurements of the participant, typically total body mass and segment lengths, however, models such as Yeadon (1990) require 95 measurements. Typically, these anthropometric parameters are derived from studies with cadavers or young adults. However, sprint cyclists have been shown to have larger thigh and calf girths then the average adult population (Foley, Bird, & White, 1989; McLean & Parker, 1989). Wheat and Barratt demonstrated using a Monte Carlo simulation that the influence of uncertainties in body segment parameters were largest on the calculated hip joint power, particularly at higher pedalling rates (Wheat & Barratt, 2015). This means the joint moments and powers calculated for sprint cyclists will differ from the true value if the anthropometric variables are calculated from the standard tables such as those in de Leva (1996) or Dempster (1955).

To obtain person specific body segment parameters there are a variety of measures that can be used, which include: manual techniques such as tape measurement and water displacement, or digital techniques such as body scanning and surface imaging (Bullas, Choppin, Heller, & Wheat, 2016). There are many different systems available to create 3D digital images from which anthropometrics can be calculated (Bullas et al., 2016). One system which is quick, low cost, commercially available and portable is a 3D surface imaging system using depth cameras (e.g. Microsoft Kinect) (Bullas et al., 2016; Clarkson, Wheat, Heller, & Choppin, 2014; Kordi et al., 2018). Kordi and co-workers demonstrated the good between-sessions reliability of the 3D depth camera system when they measured thigh volume (absolute typical error 112 cm^3 and CV 1.7%) (Kordi et al., 2018). They found the 3D depth cameras systematically measured the gross thigh volume 32.6 cm^3 (0.57%) lower than measured by an MRI scanner (Kordi et al., 2018), whereas a study by Bullas and colleagues found the 3D depth camera system systematically overestimated thigh volume (~6%) compared to a high precision 3D surface imaging system (Bullas et al., 2016). Therefore, when interpreting between-sessions changes in anthropometrics the accuracy of the 3D depth camera systems needs to be considered. The 3D depth camera system could be used to measure a cyclist's change in the body segment parameters over a period of training as this is one constraint that will influence the cyclist's intermuscular coordination pattern. It has been used to measure thigh volume, which is known to change with strength training due to muscle hypertrophy (Bullas et al., 2016; Kordi et al., 2018; Rønnestad et al., 2010).

2.7.7 Muscle Activity

Electromyography (EMG) - the measurement of muscle electrical activity is required to obtain the muscle activity sequence and level of activation of the lower leg muscles. This can be recorded using surface EMG sensors which is the most common method for studying intermuscular coordination in cycling even if the deep muscles such as the hip flexors, psoas and iliacus, cannot be measured. This method was used by Dorel and colleagues to investigate the changes in intermuscular coordination between submaximal and maximal cycling (Dorel et al., 2012).

The Delsys Trigno Lab system (Delsys Inc, Boston, MA) uses wireless surface EMG sensors. Each sensor uses a single differential electrode configuration to detect the electrical signals from the surface of a muscle. The advantage of surface EMG sensors is they are easy and quick to apply, and are not invasive, which are important considerations when working with elite athletes. However, they have several limitations, one of which is that they can't measure deep muscles because these require intramuscular fine wire electrodes which are invasive and therefore, generally not used (Hug & Dorel, 2009). In cycling this means the psoas, iliacus, adductor magnus, and biceps femoris short head muscles can't be studied. Other limitations are crosstalk, where the sensor detects electrical activity from the adjacent muscles to the one being studied (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2004), and amplitude cancellation which refers to the cancellation of the positive and negative phases of the motor unit action potentials (Farina, Merletti, & Enoka, 2004). Other factors that influence the EMG signal are: the type of electrodes used, the skin surface, the amount of subcutaneous fat, blood flow, muscle temperature, muscle length, depth of muscle below the surface and the location of electrodes (Robertson et al., 2004). Researchers can control several of these factors to improve the quality of the EMG signal: by reducing skin impedance through preparing the skin surface (shave and clean the site of the electrode) and carefully locating the sensors – De Luca recommends the sensor should be placed on the midline of the muscle belly, between the myotendinous junction and the nearest innervation zone, with the detection surface orientated perpendicularly to the length of the muscle fibres (De Luca, 1997).

Once the EMG data have been collected the EMG signals are processed. There are a variety of methods used in the literature to smooth and filter the signal to produce a smooth linear envelope. These are: the Butterworth filter (the signal is rectified first) (Challis, 1999), root mean squared (RMS) (Robertson et al., 2004), and integrated EMG (Singh & Latash, 2011). There is no agreement in the literature of what is the best method to process the EMG signal. In cycling the EMG signal is often averaged over a number of pedal cycles to obtain an averaged linear envelope, which is normalised to crank angle (Brochner Nielsen et al., 2018; Dorel et al., 2012; Hug, 2011). The linear envelope is representative of the EMG activity and increases the signal to noise ratio (Dorel et al., 2012; Hug, 2011). However, by averaging the signal, the between pedal revolution variability is lost which can be important in understanding the intermuscular coordination strategies.

To allow EMG data to be compared between participants, different muscles, different test conditions and different testing sessions EMG data are normalised, i.e. expressed in relation to a reference value obtained during standardised and reproducible conditions (Burden, 2010; Mathiassen, Winkel, & Hägg, 1995). Researchers have used a variety of methods. These include normalising the EMG data to the maximum voluntary contraction (MVC) for each muscle - there is set of guidelines on how to acquire the MVC for each muscle (Konrad, 2005). Performing an MVC for each muscle (which is usually taken as the maximum value from a set of three trials) is time consuming (Hug, 2011). One of the arguments against the use of the standard MVCs is that they are performed at different joint angles, muscle lengths and contraction type to those required for the task being studied, particularly when applied to dynamic tasks (Mirka, 1991). This is disputed by Burden (2010), who states that the task specific isometric MVCs produce similar output to standard MVCs and do not appear to be affected by contraction mode or joint kinematics (Burden, 2010). Ericson developed a specific set of MVCs with the joint positions relevant to cycling, but these are still isometric contractions and therefore, potentially do not represent the maximum activation in dynamic tasks (Ericson, 1986). Hunter and colleagues compared a static on bicycle method where the MVC for the RF muscle was measured when the pedal has been fixed to achieve a certain knee joint angle with a traditional isometric MVC for the knee extensors on a dynamometer (Hunter, St Clair Gibson, Lambert, & Noakes, 2002). The on bicycle static method did not elicit as large a muscle activation as a traditional isometric MVC (Hunter et al., 2002). Kordi compared single joint unilateral isometric MVCs performed on a dynamometer with multi-joint isometric cycling task MVCs on a cycling ergometer (Kordi, Folland, Goodall, Barratt, & Howatson, 2019). They concluded that isometric reference tasks may not be suitable to ascertain changes in peak muscle action over time in sprint cycling tests (Kordi et al., 2019).

Researchers have developed normalisation methods specifically for cycling and to reduce the time needed to carry out the normalisation. These include measuring the maximum muscle activity during a maximal sprint on a bicycle (Albertus-Kajee, Tucker, Derman, & Lambert, 2010; Rouffet & Hautier, 2008). However, as shown by Dorel and colleagues maximal cycling does not maximally activate all the muscles (Dorel et al., 2012). Dorel and colleagues used two methods and selected the highest EMG activity to overcome some of the shortcoming of the various methods (Dorel et al., 2012). These methods were isometric MVCs at a variety of joint angles similar to those used pedalling, and isokinetic MVCs using joint ranges of motion similar to cycling both of these were carried out on a dynamometer (Dorel et al., 2012). However, even using this procedure during an all-out sprint the soleus muscle activity exceeded the MVC value showing that the MVC procedure does not always elicit the maximum activity (Dorel et al., 2012). However, combining these two methods has been shown to be the most reliable at getting a maximal response from each muscle (Burden, 2010; Dorel et al., 2012; Hunter et al., 2002). This method requires the use of a dynamometer so it can only be applied in laboratory-based testing sessions and not in the field. Sinclair and colleagues compared different EMG normalisation methods for cycling: isometric MVCs on a dynamometer, 5 minute submaximal cycling at 180 W to obtain mean and peak activation for each muscle, and 10 second cycling sprint to obtain peak activation for each muscle (Sinclair et al., 2015). They found the most reliable normalisation method was to normalise using the peak muscle activation from the submaximal cycling trial (Sinclair et al., 2015).

Two simple methods to avoid having to carry out an MVC are to normalise the data to the peak value in the signal – the peak dynamic method (Brochner Nielsen et al., 2018; Ryan & Gregor, 1992) - and to the mean of the signal – the mean dynamic method (Burden & Bartlett, 1999). However, Burden and Bartlett recommended that the peak or mean dynamic method should not be used if wanting to compare between different trials, muscles, individuals, or to retain the natural variation between individuals (Burden & Bartlett, 1999).

Currently there is no agreement between researchers on what is the best normalisation procedure to use (Burden & Bartlett, 1999; Hug, 2011). In a review of EMG normalisation procedures, Burden (2010) recommended the use of the arbitrary isometric MVC as there is no strong evidence at present to suggest that the isometric specific MVC or the isokinetic specific MVC need to be used instead (Burden, 2010). Therefore, in Appendix 9.7, the test-retest reliability of different EMG normalisation protocols are assessed for between-sessions comparisons of EMG activity in maximal cycling.

To compare intermuscular coordination strategies, often the onset and offset timing of muscle activity is determined from the EMG signal (Baum & Li, 2003; Bieuzen et al., 2007; Dorel et al., 2012; Duc, Bertucci, Pernin, & Grappe, 2008). Researchers have used a number of different methods to determine the threshold which defines the onset of muscle activity: a number of standard deviations above the baseline values (1, 2 and 3 SD) (Hodges & Bui, 1996; Uliam Kuriki, Mícolis de Azevedo, de Faria Negrao Filho, Ruben, & Alves, 2011), and a percentage of the peak value (Dorel et al., 2012; Jobson, Hopker, Arkesteijn, & Passfield, 2013; Konrad, 2005). The EMG signal needs to exceed the threshold for a minimum period of time for the muscle to be defined as on. Another method is by visual inspection to determine muscle onset, if this is done by an experienced researcher in EMG it can be highly repeatable between days (Hodges & Bui, 1996). Again, there is no agreement on what threshold should be used to determine onset and offset of the muscle and therefore, the researcher needs to choose an appropriate threshold. In Appendix 9.8, the suitability and reliability of onset/offset timings to define bursts of EMG activity during maximal cycling is investigated.

This programme of research requires intermuscular coordination patterns to be compared over time and therefore, the reliability of the method to measure EMG activity is important. Dorel and colleagues found good intra-session repeatability (or test-retest reliability) of lower limb muscle activation patterns in submaximal cycling - the two testing sessions were separated by a 53 minute training session (Dorel, Couturier, & Hug, 2008). Laplaud and colleagues found the muscle activity levels for eight lower limb muscles to be highly repeatable in cyclists pedalling to exhaustion in two trials separated by separated three days (Laplaud, Hug, & Grélot, 2006). In contrast, Jobson and colleagues found low between-sessions reliability for amplitude of EMG activity measured during submaximal cycling (Jobson et al., 2013). They also found lower reliability for onset and offset of muscle activity for several muscles (tibialis anterior, soleus, gastrocnemius and rectus femoris) (Jobson et al., 2013)

A solution to the degrees of freedom problem proposed by Bernstein is the muscle synergy hypothesis which states that the brain and spinal cord simplify the control of the numerous muscles by grouping them into functional units called muscle synergies which represent pre-structured motor programmes; however, this hypothesis is yet to be proven (Bernstein, 1967; Kutch & Valero-Cuevas, 2012; Tresch & Jarc, 2009). Further analysis of the EMG signal can be carried out to extract the muscle synergies either by principal component analysis (Singh & Latash, 2011), or non-negative matrix factorisation (De Marchis et al., 2013). Muscle synergies have been extracted in cycling in previous studies to try and explain the locomotor strategy used for pedalling (Blake et al., 2012; Hug, Turpin, Couturier, & Dorel, 2011). Principal component analysis or non-negative matrix factorisation can also be used as a data reduction method for large multivariate data sets, such as EMG data for many muscles and testing conditions, and not just to extract muscle synergies (Blake & Wakeling, 2015). The muscle synergy hypothesis which assumes the extracted muscle synergies from the EMG data are a static representation of pre-structured motor programmes, does not fit within the ecological dynamics framework as it assumes the brain is the central controller and that the coordination patterns are not self-organising (Riley, et al., 2012).

2.7.8 Representative experimental design

Brunswik (1956) proposed the concept of *representative experimental design* which refers to the composition of the experimental task constraints so that they represent the behavioural setting to which the results of an investigation are intended to be generalised (Araújo, Davids, & Passos, 2007; Brunswik, 1956; Pinder, Davids, Renshaw, & Araujo, 2011). In the context of investigations of sports performance this suggests intermuscular coordination cannot be assessed in environments where the constraint that differ from those required for the sports performance. This notion is highlighted in a study by Barris and colleagues which compared springboard diving in dry-land and aquatic training facilities (Barris, Davids, & Farrow, 2013). They demonstrated that the task constraints are not similar and therefore, the dry-land training facility is not representative of diving. Similar studies have been undertaken comparing overground and treadmill running, which found the kinematic and kinetic trajectories of the treadmill gait were similar to overground gait but there were some significant differences between knee kinematics, peak ground reaction forces and joint moments (Riley, et al., 2008). Lamb (1989) also found differences in arm kinematics when comparing ergometer and on-water rowing (Lamb, 1989).

In cycling most of the studies have been undertaken on an ergometer in the laboratory, which is not representative of the task being studied. When studying coordination under an ecological dynamics theoretical framework it is important that the environment and the conditions during the experiment are as similar as possible to the scenario you want to study, so the constraints acting on the athlete are the same. Therefore, it is unknown how applicable the studies undertaken on a cycling ergometer are to riding on the track. Some of the specific differences between cycling on an ergometer and on a bicycle on the track are: air resistance when moving around the track, significant out of plane movement of the bicycle and the rider system, track cycling takes place on a banked oval track where the bends of the track can be at a 45° angle to the horizontal and when sprinting on the track the riders also have to control the bicycle direction and stability whilst trying to produce maximal power (Gardner et al., 2007). In previous studies cyclists were generally asked to remain seated, whereas in sprint cycling at the start of their effort the cyclist will adopt a standing position which increases power output (Davidson, Wagner, & Martin, 2004).

The differences between field and laboratory cycling tests have been demonstrated by Bertucci and colleagues who found that maximal aerobic crank torque profiles were significantly different with a higher rate on perceived exertion on an ergometer compared to a field road cycling tests (Bertucci, Grappe, & Groslambert, 2007). The differences were also highlighted for BMX cyclists who produced higher peak power and reduced time to peak power in field tests compared to on an ergometer in laboratory (Rylands, Roberts, & Hurst, 2015). However, in contrast, Gardner and colleagues found similar maximal torque- and power-pedalling rate relationships between sprints on an inertia ergometer and when cyclists performed a standing start 65 m on a velodrome track, concluding that ergometer data can be used to model sprint cycling performance (Gardner et al., 2007). However, they did not record detailed biomechanics variables such as crank forces, joint angles, angular velocities, moments and powers and EMG activity that characterise intermuscular coordination in sprint cycling.

There are only two field studies of intermuscular coordination in cycling: the study of muscle activity during an outdoor 18.8 km cycling time trial (Blake & Wakeling, 2012), and thigh muscle activity during track cycling (Watanabe et al., 2016). These studies highlighted several differences in intermuscular coordination between laboratory and field conditions. Blake and Wakeling found intermuscular coordination fluctuated depending on terrain and pacing strategy (Blake & Wakeling, 2012). Therefore, they concluded that care should be taken when applying the findings from laboratory studies to outdoor cycling and highlighted the importance of measuring coordination in the field or careful reproducing the outdoor environment in the laboratory (Blake & Wakeling, 2012). Watanabe and colleagues found significantly higher EMG activity for the BF muscle for the right leg compared to the left leg (Watanabe et al., 2016). Typically, muscle activity would be assumed to be similar for both legs, again highlighting the effect of task constraints on coordination patterns.

2.7.9 Statistical parametric mapping

Statistical parametric mapping (SPM) (Pataky, 2010) is being used in many contemporary biomechanics studies to compare biomechanical time series data between conditions (Colyer, Nagahara, & Salo, 2018; Judson et al., 2019; Pataky et al., 2008; Warmenhoven et al., 2018). SPM allows the data to be compared along the whole time series and not just for key discrete data points, i.e. maximum and minimum values of a variable (Warmenhoven et al., 2018). Pataky and colleagues demonstrated subsampling the centre of pressure data during walking may obscure or reverse statistical trends compared to using SPM to compare the whole time series centre of pressure data (Pataky et al., 2014). Judson and co-workers also highlighted the benefit of using SPM when they compared left and right foot ground reaction forces in bend sprinting – SPM identified asymmetries between left and right foot mediolateral forces for parts of the stance phase (Judson et al., 2019).

When interpreting SPM results Colyer and colleagues suggested that the SPM cluster size had to be larger than 5 nodes (5% of the time series) to be considered meaningful when comparing ground reaction forces in sprinting (Colyer et al., 2018). However, the researcher needs to use their judgement when deciding on the size of the SPM cluster that can be considered meaningful for their time series data. The method used to smooth the one dimensional (1D) time series biomechanical data can influence the outcome of the SPM analysis, and care needs to be taken not to over-smooth the data as this can lead to systematically biased 1D data yielding high false positive rates (Pataky, Robinson, Vanrenterghem, & Challis, 2018). SPM requires temporal normalisation of the data. This processing of the data can distort the location of the peaks (Sadeghi, Mathieu, Sadeghi, & Labelle, 2003; Warmenhoven et al., 2019). This can be a particular problem for gait biomechanical data where the data are typically normalised to % of the gait cycle, as participants have different stride lengths and frequencies. However, in cycling, the biomechanical data are normalised to crank angle which is measured and in this study the pedalling rate was also controlled, so each crank cycle takes the same time. Therefore, cycling biomechanical time series data does not have the same problems as gait when time normalising the data.

2.8 Summary

This programme of research used an ecological dynamics theoretical framework to investigate how strength training influenced intermuscular coordination in maximal sprint cycling. In accordance with concepts in ecological dynamics, there is clearly a need for more research on intermuscular coordination in cycling performance outside of the laboratory, where the task and environmental constraints are more representative of training and competition conditions. It is proposed that strength training will change the individual constraints of each athlete by changing muscle strength, body segment parameters, muscle fatigue, and intramuscular coordination. The key focus of this programme of research was therefore understanding how each athlete adapts their intermuscular coordination patterns due to changing personal constraints caused by strength training.

3 Coaches' philosophies on the transfer of strength training to elite sports performance

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3.1 Introduction

Coaches of sports requiring maximal effort over a short period of time (<60 s), such as track sprint cycling, sprint kayaking (200 m), and sprinting (athletics) often consider strength training (repetitive muscle actions against high loads) to be a fundamental aspect of an athlete's training programme (García-Pallarés & Izquierdo, 2011). Accordingly, sprint athletes from a range of sports routinely undertake strength training in addition to sport-specific training (García-Pallarés & Izquierdo, 2011; Parsons, 2010).

Despite the common prescription of strength training in elite sport, empirical evidence shows that transfer to sports performance varies (Carroll et al., 2001; Young, 2006). Generally, there is positive transfer to sports performance; for example, Blazevich and Jenkins found strength training improved 20 m start time in elite junior sprinters (Blazevich & Jenkins, 2002). However, sometimes there is no effect or even a negative transfer (i.e. strength training is detrimental to performance) (Carroll et al., 2001; Young, 2006). Moir and colleagues found a similar strength training intervention worsened 20 m acceleration time in an equivalent cohort of athletes (Moir et al., 2007).

Strength training increases muscle strength and size (Carroll et al., 2001; Zatsiorsky & Kraemer, 2006), so the focus of non-specific strength training ("traditional" gym-based strength exercises that are not specific to the sport movement e.g. squat, deadlift, and leg press) is often on these muscular adaptations (Knuttgen & Komi, 2003). It also causes neural adaptations such as recruitment, or more consistent recruitment, of the highest threshold motor units, increased motor unit firing rates, and an increase in tendency of motor units to fire synchronously, collectively referred to as intramuscular coordination (Carroll et al., 2001; Sale, 2003; Zatsiorsky & Kraemer, 2006). These neuromuscular adaptations are well correlated with performance in high-intensity locomotive sports (e.g. sprint cycling, running, kayaking and rowing). The maximum power of elite sprint cyclists, for example, is strongly correlated with maximal torque, which in turn, is correlated with lean leg volume (Dorel et al., 2005). 'Gym strength' (assessed by the amount of mass that can be lifted in a non-specific strength exercise - measured in this example by the isometric mid-thigh pull) has also been correlated with sprint cycling power and sprint cycling times (Stone, et al., 2004). Similar relationships between determinants of strength and sports performance have been found for sprinting and rowing (Kumagai et al., 2000; Slater, et al., 2005).

Despite beneficial neuromuscular adaptations, some whole-body mechanisms such as intermuscular coordination may explain the reduction in performance sometimes associated with strength training. Intermuscular coordination could influence the transfer of strength training to sport performance in two ways. First, increases in muscle strength from strength training may need to be accompanied with a change in intermuscular coordination to improve sport performance. This idea was supported by Bobbert and Van Soest who showed that an increase in leg strength must be accompanied by a change in intermuscular coordination in order for vertical jump height to increase (Bobbert & Van Soest, 1994). This idea that the coordination patterns need to change in response to changing constraints (e.g. muscle strength) is captured by key ideas in ecological dynamics and Newell's model of constraints (Newell, 1986). Newell proposed that patterns of coordination emerge from the constraints imposed on an individual during action. Constraints are boundaries or features that shape the organisation of emergent coordination patterns (Newell, 1986). Accordingly, strength training may be expected to change the individual constraints of each athlete by changing muscle strength, body segment parameters, muscle fatigue, and intramuscular coordination. Second, muscle recruitment patterns associated with a strength training task could retard sports performance when expressed during the sport movement (Carroll et al., 2001). For example, the strength training programme of a sprint cyclist commonly consists of non-specific strength training exercises, such as squats, deadlifts and leg presses (Parsons, 2010). These exercises, however, have very different intermuscular coordination patterns compared to the act of pedalling (Koninckx et al., 2010). In this way, extensive non-specific strength training could impair pedalling coordination such that cycling performance is reduced. This notion is further supported by the training principle of specificity, which states that the closer the strength training resembles a sport movement, the greater the transfer of strength, particularly in elite athletes (Young, 2006; Zatsiorsky & Kraemer, 2006).

The motivation for conducting this study was simple. Before a more positivistic programme of research to examine how to integrate strength training with that of coordination training was conducted, the aim was to understand the current beliefs, existing knowledge and ideas that underpinned elite coaches' approaches to the idea of integration. In essence, this is considered to be the 'philosophy' behind the day-to-day practice of elite coaches' design of strength and conditioning and coordination training in sports such as cycling (Gearity, 2010).

Elite coaches and athletes are highly motivated and have years of experience to evolve and improve their training protocols to achieve successful transfer of strength training to sports performance. This group's philosophies may, therefore, be regarded as current 'best practice'. Here, a qualitative approach was chosen to enable exploration of coaches' experiential knowledge and insights, an approach used previously in sport science research to provide insights to enhance understanding for empirical and applied research (Greenwood, Davids, & Renshaw, 2014; Jones, Bezodis, & Thompson, 2009; Phillips, Davids, Renshaw, & Portus, 2014). Also, there is very little information in the literature about elite coaches' approaches to strength training and sports performance. A selection of sports demanding maximal effort over a short period of time were chosen for analysis as there are clear parallels between sports, and so coaches' experiences can be synthesized. The aim of this study, therefore, was to explore elite coaches' philosophies regarding strength training and the range of factors and ideas believed to affect transfer of strength training to sport performance.

3.2 Methods

Thirteen participants (12 male and 1 female) were recruited by purposive (criterion-based) sampling (Patton, 2002). The criteria was that the participants were elite coaches or athletes in the sports of track sprint cycling, BMX, sprint kayaking, rowing and athletics sprinting. The participants were composed of 11 elite coaches and 2 athletes. The coaches all worked at international level and coached either the development (3) or senior squads (5), with some specialising in strength and conditioning (3). Coaching experience ranged from 2.5 to 31 years. Six of the coaches had prepared athletes for the Olympic Games, with five having coached Olympic medallists. Both athletes were Olympic medallists who had competed at international level for over 12 years. Participants were recruited through a high-performance sport network and a regional elite sports club and were provided with the details of the study and signed the consent form. The study was approved by the Sheffield Hallam University Faculty of Health and Wellbeing Research Ethics Sub-Committee.

To address the research aim a combination of epistemological constructionism and ontological relativism to inform a interpretivism research paradigm was adopted (Sparkes & Smith, 2013). The interviews were semi-structured with open-ended questions to allow participants to express thoughts and expand on topics (Sparkes & Smith, 2013). The list of questions that formed the interview framework started with general warm-up questions on sport background and experience, moving to more specific questions asking about coaching philosophy, athlete attributes, design of training programmes, strength training, and the transfer of strength training to sports performance (refer to Appendix 9.3 for interview guide). Probing questions were used to obtain more detail (Sparkes & Smith, 2013). A pilot interview was conducted to assess question suitability. All interviews were conducted by the primary researcher and took place at the participant's place of work. Interviews were between 19 and 55 minutes in length (mean 34 minutes) and were recorded on a digital voice recorder.

The interviews were transcribed verbatim and small grammatical changes were made to improve the flow of the text. To enhance data trustworthiness a process of member checking was carried out (Lincoln & Guba, 1985). For this process, transcripts were sent to participants to check for accuracy, correctness of researcher interpretations and for clarification on any transcript passages where the meaning was unclear.

The primary researcher undertook an initial analysis and coding of the transcripts using inductive reasoning in the software programme NVivo (QSR NVivo 10). This approach allowed the primary researcher to identify emerging data saturation. Following the 11th interview, a decline in new information was observed. After the 13th interview theoretical saturation was identified as all new data fitted into the existing organisation system without the emergence of new themes (Cote, Salmela, Baria, & Russell, 1993).

A thematic analysis was conducted (Braun & Clarke, 2006; Patton, 2002; Sparkes & Smith, 2013). Data were initially coded into raw themes, which were grouped into lower and higher order themes. Themes were reworked and refined by repeatedly reviewing generated themes, and the original data. Another method used to enhance output trustworthiness was analyst triangulation (Cote et al., 1993).A second researcher analysed a sample of the interview transcripts independently and discussed themes generated with the research team before final themes were agreed.

3.3 Results

Key themes emerging from interview data were grouped into 'strength training', and 'transfer of strength training to sports performance' (Figure 3.1, Figure 3.2). Transcripts revealed 30 initial data nodes, further grouped into higher and lower order themes. Coaches' philosophies were similar, although key areas where viewpoints differed included quantity and scheduling of training sessions.

3.3.1 Strength Training

Coaches believed that non-specific strength training was important for increasing athletes' muscle size and strength (Figure 3.1). The key role of non-specific strength training was muscle-level adaptations, typically in isolation from sports performance, as highlighted by the following coaches:

> *"Bigger muscles are generally stronger muscles, so the first part of our preseason is about muscle mass not typical hypertrophy ..... we want size but we also want strength as well. So it's obviously it is heavy weights ... it's our philosophy that size is one of the biggest contributors towards how strong the muscle will be."* P5 – Coach

"We are in the gym for hypertrophy and muscle mass, basically building a bigger stronger muscle." P7 – Coach

Participants also stated that an athlete's expression of strength needs to be specific to their sport:

"Strength for an [athlete] is different to strength in the gym, because it is task specific." P4 – S&C coach

"[We need] explosive power, but we need the base of strength first, and then that needs to be synchronised with the art of pedalling." P5 - Coach

During preseason training the coaches' focus was increasing the athletes' muscle size and strength. Because of this, non-specific strength training was prioritised in training programmes, as this coach expresses:

"Generally at the start of [the pre-season] …. the athletes would be in the gym three times a week and … [have sport-specific training] probably twice a week which is maintenance really." P5 - Coach

However, during the competitive season, the aim was to maintain the athletes' strength by reducing the number and volume of gym sessions, but maintaining intensity, as one of the strength and conditioning coaches highlighted:

"So for [gym sessions close to competition] dropping volume, maintaining intensity ….. and including some …. slightly more dynamic efforts" P1 – S&C coach

Coaches typically talked of prescribing non-specific strength training exercises in gym sessions, for example squat, leg press, deadlift, bench press:

> *"Being an upper body sport: bench press, bench pull, chin up [which] would be ..... in any kayak programme from club to international level."* P1 – S&C coach

> *"From a gym strength side of things, we have standards for the big core lifts, so things like, squat, deadlift, trap bar deadlift."* P4 – S&C coach

> *"Preferably most of [the strength training] is done in the gym using primary movements: squat, deadlift, cleans, leg press."* P10 - Coach

When increased specificity was desired, rather than trying to make the gym exercises more sport-specific - i.e. by mirroring the sport movement patterns, coaches instead preferred to add resistance to sporting movements. Examples of resisted sporting movements would be resisted rowing, resisted running, or over-geared (increased resistance) pedalling:

> *"Parachute or a bucket off the boat so they would still be doing resisted work but it would be in the context of rowing."* P2 - Coach

> *"Resistance running – hills .... [and] sledge work at the right time though I don't do any more than 20 m in a rep with sleds and I don't put too much weight on either."* P13 - Coach

Coaches believed that this resisted sport movement training transferred quicker than non-specific strength training exercises, as they included similar movement patterns to the sport. Accordingly, these sessions were used as a bridge between the non-specific strength training and sports performance as this coach describes:

> *"So for a couple of those athletes the gym structure would be anywhere between two or three sessions a week in a heavy gym, heavy strength*

block with some [resisted movement training] as well to encourage as much crossover as we can in that period." P9 – Coach

Only two participants included gym exercises that were sport-specific as they thought the coordination aspects transferred:

> *"There are some athletes that are doing slightly more similar exercises like RDL's [Romanian deadlifts] or the first pull of a clean off the floor which are very similar [to the sports movement], and I do think there is some transfer and there is coordination aspects of that which are really useful."* P8 – S&C coach

In contrast, as highlighted by the following quote, for most of the participants the role of gym sessions is to increase muscle size and strength, and sport-specific training for improving explosivity and coordination:

> *"I think my views and my philosophies …… have changed over time…. and years ago I would have had a stronger … view on training in a different way where it was more strength based, and then the gym exercises ….changed after the strength period to be light and explosive and lifted rapidly….. Whereas now …. we aim for strength and size from the gym, but the importance of the bike in the equation is so much higher which really makes sense when you think about it, and so that coordination and the explosivity that you want …we just get on the bike and so we manipulate the volume of that work and what it looks like in the training week, .. rather than try and go and get it in the gym."* P5 - Coach

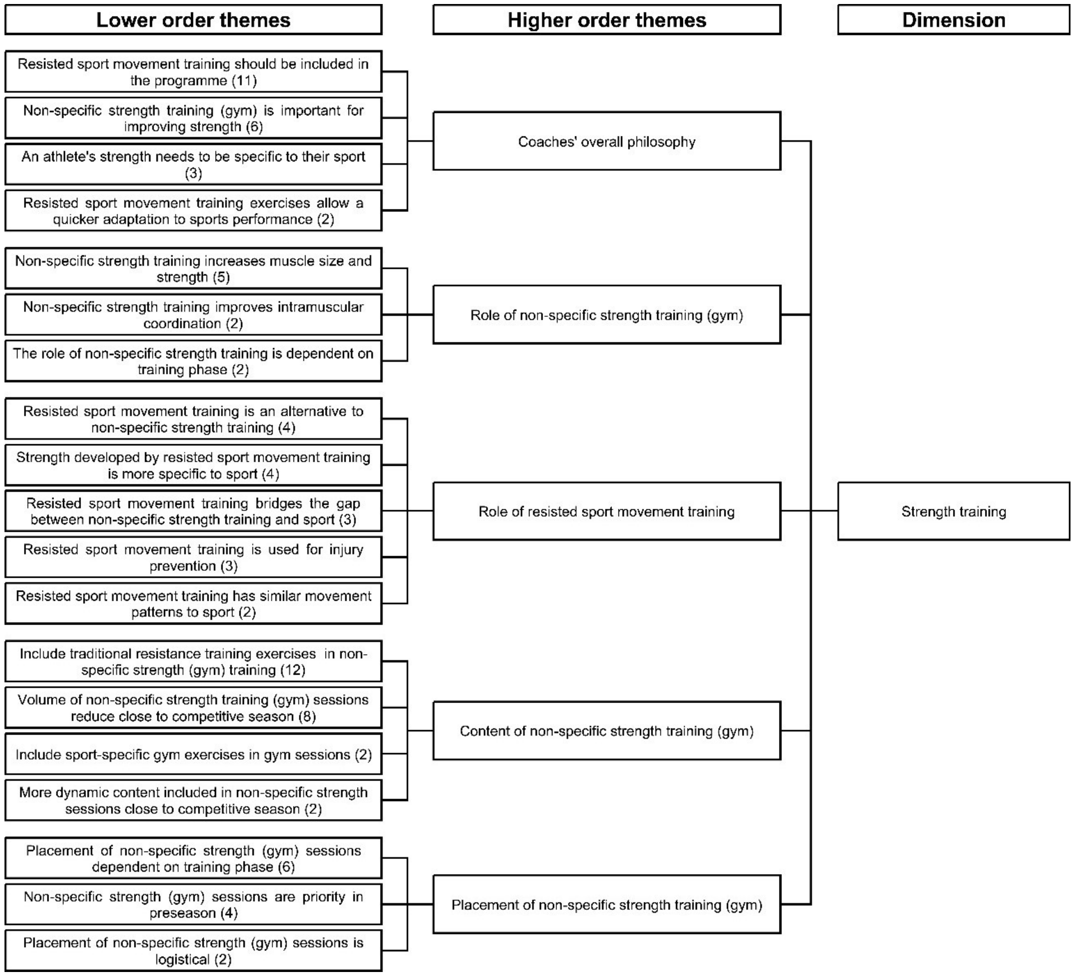

Figure 3.1: Strength training: lower and higher order themes (Number of expert sources in brackets)

3.3.2 Transfer of strength training to sports performance

Participants believed that the transfer of strength training to sports performance was not as simple as athletes getting stronger in the gym then immediately getting quicker at their sport (Figure 3.2). Therefore, they did not believe there was a direct correlation between 'gym strength' and sports performance, and that sometimes increased 'gym strength' did not transfer to performance speed at all, as one athlete discussed:

> *"I have known athletes that can lift a lot more in the gym but are slower on the [track] so it leads me to think for me there is not necessarily a direct correlation, although for some people there is. One of the guys*

that I am coaching it seems to be that he has had a quite linear progression in the gym in terms of his 1RM, squat 1RM and it seems to translate directly to the [track] without any period of adaption at all."
P12 – Athlete

Coaches identified several training protocols, athlete attributes and factors that they thought affected the transfer effectiveness and the length of the transfer period (Figure 3.2). Including speed and technique sessions during a non-specific strength training phase was one of the training protocols coaches thought improved transfer:

"So in the spring period we would add into the weights part of the programme some maximal bouts of sprint work on the rowing machine in order to make sure gym work is relevant to a more rowing specific movement." P3 - Coach

"For a couple of those athletes …… in a heavy gym, heavy strength block include maybe it's a style of warm-up or what we would call a recovery session on the bike, we should have a little speed element or a little bit of acceleration in there with a general fitness underlying thing. So we are still keeping relatively fit, there is still a little bit of pedalling and speed work in there but the aim is getting stronger." P9 - Coach

One factor affecting transfer of strength training was fatigue generated from a period of heavy strength training, which meant that an ensuing recovery period was required to observe performance benefits as a coach proposed:

"I don't think you see any [immediate] transfer at all because of the amount of fatigue that the strength places the athlete under. However,

where you do see the benefits is when you freshen them up, that's when you see the reward." P3 - Coach

Participants also highlighted sports technique as being an important factor for transfer:

"We have already talked about that the influence of technique. I have seen a lot of people who have got more strength and have never got anything on the water, so it is not just a long time scale; it is almost if there is something extra that has to come with having the increase in strength. Even in the most specific exercises we have in gym, the transfer is far from the given, so it can be time poorly invested if you cannot put it down at all." P2 - Athlete

"So, your technique needs to adapt, to go with your strength, is that it?" Interviewer

"Yeah, exactly, both in terms of movement speed and coordination." P2 - Athlete

Participants also believed that speed and technique training sessions were needed in an athlete's training programme to maintain technical performance during a period where non-specific strength training was prioritised, and these sessions facilitated a quicker transfer of 'gym strength' to sports performance:

"So if we took an athlete and said "Right, we need you to get stronger we are going to spend the whole year trying to get you stronger" …. and that's possible to do with almost any athlete. But … if we weren't teaching them…. [movement] dynamics …. or speed…. potentially they just get slower from being stronger." P10 - Coach

Coaches also believed that the transfer period from increased 'gym strength' to improved sports performance was individual and could be lengthy as these two coaches expressed:

> *"I do think there is a lot of individual response, particularly in elite athletes who are fundamentally different to most populations you would be able to test on."* P4 – S&C coach

> *"It's a fairly long transition [from non-specific strength training to improved sport performance] and there has definitely been periods where we have got athletes stronger but not quicker and there have definitely been periods where we have got an athlete quicker but not any stronger."* P9 - Coach

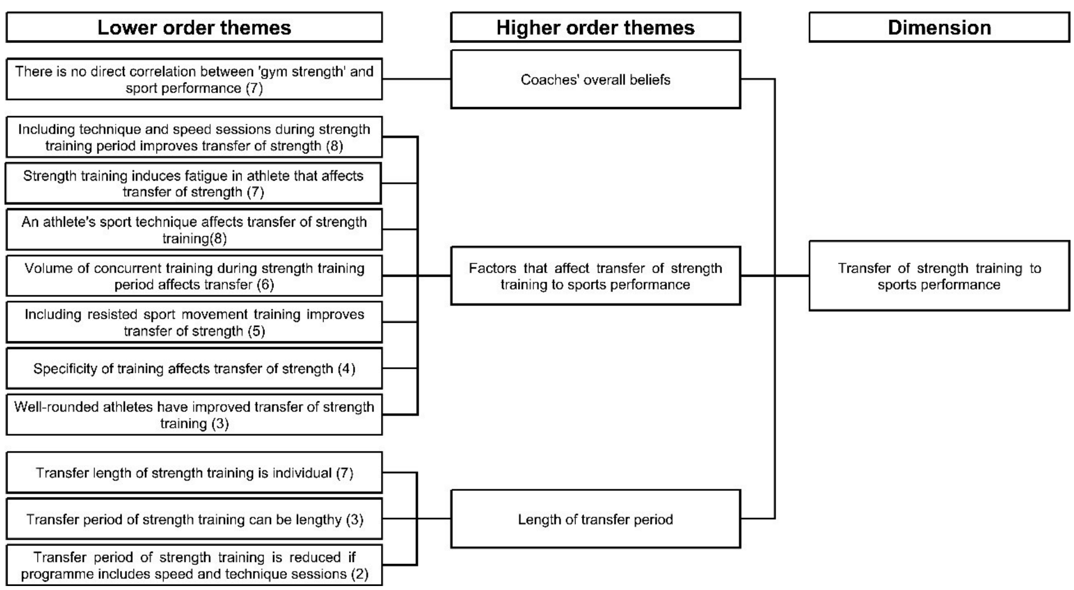

Figure 3.2: Transfer of strength training to sports performance: lower and higher order themes
(Number of expert sources in brackets)

3.4 Discussion

The philosophy and ideology behind strength training in elite performance programmes in sports demanding maximal effort over a short period of time was examined. The main findings suggest that coaches viewed task-specific strength as important for sports performance, and that this is typically achieved with a combination of non-specific strength training and resisted sport movement training.

The coaches' rationale for including non-specific strength training in the athletes' programmes was predicated on muscle-level adaptations, increasing muscle size and strength, a notion clearly supported in the scientific literature as key adaptations to strength training (Carroll et al., 2001; Rønnestad et al., 2010; Zatsiorsky & Kraemer, 2006). Muscle size and strength have been correlated with sports performance in maximal sports, further supporting the coaches' philosophy (Dorel et al., 2005; Pearson et al., 2006). A few participants specifically mentioned using strength training to achieve neural adaptations which typically improve rate of force development, important in explosive sports (that require high acceleration from the start) (Aagaard et al., 2002; Cormie et al., 2011b). Despite strength training having been shown to lead to other adaptations which contribute to increased muscle strength, such as changes to muscle-tendon stiffness and compliance, tendon properties (Zernicke & Loitz-Ramage, 2003) and muscle architectural changes (Aagaard et al., 2001), the coaches did not specially refer to these adaptations.

Only a few coaches applied the training principle of specificity when selecting and designing gym exercises, contrasting with some previous literature stating that specificity of the strength training needs to increase for elite athletes to keep improving sports performance (Bosch, 2015; Young, 2006; Zatsiorsky & Kraemer, 2006). Coaches chose gym exercises to increase strength of muscles required for the sport movements. However, coaches supplemented these exercises with resisted movement training, by using the sport movement with added resistance, to achieve specificity of load. This approach has been suggested in the literature as a method for achieving specificity (Rumpf et al., 2016; Young, 2006).

Coaches perceived that there was not a direct correlation between increased ‘gym strength’ and improved sporting performance. This view slightly contradicts their view that non-specific strength training is important for improving an athletes’ strength. However, they acknowledged that the transfer of strength training to sports performance is not inevitable and that the correct training protocols (for example by including speed and technique sessions during a strength training block) are required to achieve a successful transfer of strength. The belief that there is no direct correlation between increased ‘gym strength’ and improved sporting performance concurs with previous findings showing that transfer of strength training to sports performance can vary (Carroll et al., 2001; Young, 2006). Coaches identified the key factors that they considered to influence transfer. Specifically, they highlighted that a period of rest or reduced training load is required to reduce fatigue and thus enhance the benefit from strength training, a notion which is supported in the literature (Mujika & Padilla, 2003; Zatsiorsky & Kraemer, 2006).

Coaches also considered the role of coordination in the transfer process as they believed that it was important to maintain an athlete’s sport technique (sport-specific coordination and movement patterns) and speed during a strength training period. In agreement with this idea, some researchers consider that coordination has an important role in achieving successful transfer of strength training to sports performance (Bosch, 2015; Carroll et al., 2001). Carroll et al, for example proposed that intermuscular coordination has a role to play in training transfer, suggesting that negative transfer may occur if the intermuscular coordination patterns of the training task retard sport specific performance (Carroll et al., 2001). Beyond aspects of training specificity, however, some researchers have suggested that intermuscular coordination may also be the mechanism to explain the timeframe – as identified by the coaches in the present investigation – between increased strength and enhanced sports performance. Bobbert and Van Soest, for example, used a musculoskeletal simulation to demonstrate that an increase in leg strength must be accompanied by a change in intermuscular coordination in order for vertical jump height to increase (Bobbert & Van Soest, 1994). The idea that each athlete needs to adapt intermuscular coordination in response to a change in his/her unique set of “organismic constraints” (e.g. muscle strength) in an individualised way is very well described by the ecological dynamics theoretical framework and Newell’s model of constraints (Newell, 1986). The associated period of intermuscular coordination adaptation may, therefore, explain the timeframe associated with a successful transfer of strength training to sports performance, as highlighted by the coaches in the present study.

This study added to the literature examining experiential knowledge, beliefs and understanding of a sample of elite coaches in high performance sport. Further empirical research is needed to determine the relative importance of each factor identified by the coaches that affect transfer of strength training to sports performance to inform coaching practice. This would allow the development of a theoretical framework on how best to combine the benefits of non-specific strength training, which causes muscle-level adaptations, with sport-specific training that improves coordination and technical ability to perform a sport movement. The participants for this study were all recruited from sports that require maximal effort over a short period of time, which involve a cyclical action (for example stroke in rowing and kayaking, stride in running and crank revolution in pedalling) and are relatively closed skills sports. Therefore, it is not clear whether the findings may be applicable to understanding training for other sports, such as team games, which contain more open skills, despite the requirement for maximal bursts of effort. These maximal bursts of effort in team sports are intermittently repeated throughout a whole competitive match, which differs from the sports in this study which require one all-out effort by an athlete.

3.5 Conclusion

The main findings are that coaches view task-specific strength as important for sports performance, and that this is best achieved with a combination of non-specific strength training and resisted sport movement training. The transfer of strength training to sports performance was believed to be a complex process, with factors associated with fatigue and coordination having particular importance. The importance the coaches place on coordination is supported by a theoretical model that demonstrates increases in muscle strength from strength training may need to be accompanied with a change in intermuscular coordination to improve sport performance (Bobbert & Van Soest, 1994). The idea that each athlete needs to adapt intermuscular coordination in response to a change in his/her unique set of “organism constraints” (e.g. muscle strength) is well described by the ecological dynamics theoretical framework and Newell's model of constraints (Newell, 1986).

The coaches' experiential knowledge and the factors they identified as being important in the transfer of strength training to sports performance were considered in the interpretation of the strength training intervention in Chapter 6 and Objective 5.

4 Comparison of biomechanical data of a sprint cyclist in the velodrome and in the laboratory

4.1 Introduction

In biomechanics research the measurement of key variables in sport performance are typically undertaken in laboratory settings, although some previous studies have revealed differences with measures recorded in a performance environment: in diving, running and rowing (Barris et al., 2013; Button, Moyle, & Davids, 2010; Dingwell, Cusumano, Cavanagh, & Sternad, 2001; Lamb, 1989; Riley, et al., 2008). Brunswik (1956) proposed the concept of *representative experimental design*, referring to the design of experimental task constraints so that they represent the behavioural setting to which the results of an investigation are intended to be generalised (Araújo et al., 2007; Brunswik, 1956; Pinder et al., 2011). This idea is in accordance with the ecological dynamics theoretical framework and Newell's model of constraints that consider it is important to study athlete behaviours under specific environmental and task constraints that faithfully simulate competitive performance (Newell, 1986). These differences raise questions of specificity of movement coordination measures recorded under certain laboratory task constraints (e.g., when using ergometers or treadmills), compared to the performance environment.

To exemplify, in biomechanical analyses of cycling performance, most studies have investigated movement behaviours on an ergometer fixed in a laboratory, which is not representative of cycling on a track owing to the differences in task and environmental constraints between sprinting on a laboratory ergometer and a track bicycle in the velodrome. In previous work, Gardiner and colleagues compared maximal torque- and power-pedalling rate relationships between sprints on an inertia ergometer and when cyclists performed a 65 m effort from a standing start on a velodrome track. They found similar relationships between laboratory and field data, concluding that ergometer data can be used to model sprint cycling performance (Gardner et al., 2007). However, they did not record detailed biomechanical variables such as crank forces, joint angles, angular velocities, moments and powers, and muscle activity that characterise intermuscular coordination in sprint cycling.

Muscle activity during cycling has been shown to differ between laboratory and field conditions (Blake & Wakeling, 2012; Watanabe et al., 2016). Blake and Wakeling investigated muscle activity during an outdoor 18.8 km cycling time trial and found muscle coordination fluctuated depending on terrain and pacing strategy (Blake & Wakeling, 2012). Therefore, they concluded that care should be taken when applying the findings from muscle coordination studies undertaken on a cycling ergometer in the laboratory to outdoor cycling (Blake & Wakeling, 2012). Watanabe and colleagues found significantly higher EMG activity for the BF muscle for the right leg compared to the left leg during track cycling, again highlighting different coordination strategies may be adopted when riding on track in the velodrome compared to an ergometer (Watanabe et al., 2016).

Although similar maximal torque- and power-pedalling rate relationships were found during the acceleration phase of sprint cycling between the ergometer and the track, other research has demonstrated differences in muscle activity during cycling between laboratory and field conditions. Consequently, research is required to investigate the effect of the different task and environmental constraints on intermuscular coordination in sprint cycling between an ergometer and the track. Therefore, the aim of this study was to measure biomechanical variables that describe elite sprint cycling in a velodrome and compare the results to a performance on an ergometer in a laboratory.

4.2 Methods

Participants

Participants were seven elite track sprint cyclists: 2 males and 5 females, age: 17.8 ± 0.4 yr, body mass: 71.3 ± 10.2 kg; height: 1.69 ± 0.13 m, flying 200 m personal best (PB): 11.3 ± 0.7 s (male flying 200m PB: 10.6 ± 0.3 s and female: 11.6 ± 0.4 s). Participants were members of the national team and competed at under 23 international level. Participants were provided with study details and gave written informed consent. The study was approved by the Sheffield Hallam University Faculty of Health and Wellbeing Research Ethics Sub-Committee.

Experimental protocol

An isokinetic ergometer was set up to replicate each participant's track bicycle position - all participants' crank length was set to 165 mm, which they all rode on their track bicycle. Riders undertook their typical warm-up on the ergometer at self-selected pedalling rate and resistance for at least 10 minutes, before performing 3 x 4 s seated sprints at a pedalling rate of 135 rpm on the isokinetic ergometer with 4 minutes recovery between efforts. All participants had previously undertaken sprints on the isokinetic ergometer, so were familiar with the protocol. A pedalling rate of 135 rpm was chosen as this is a typical pedalling rate during the flying 200 m event in track cycling and within the optimal pedalling rate range for track sprint cyclists (Dorel et al., 2005).

On the track, riders undertook their typical warm-up on their track bicycle on rollers for 10 minutes, before performing 3 seated half lap sprints, motor paced up to a speed of 62.5 km/h before starting the half lap effort (the participants followed a motor bike on the track up to the required speed when the motor bike exited the track). This track effort was chosen as it was closest to a sprint on an isokinetic ergometer, with similar pedalling rate, and the riders were motor paced up to speed, so they did not have to overcome the inertia of the fixed wheel before starting their effort. Riders were held stationary at the start of each effort in the camera capture volume where they performed three heel raises with the left leg, which was used to synchronise the kinetic and kinematic data in post processing, before rolling away to be motor paced up to speed. Participants typically had 4 minutes recovery between efforts, and laboratory and track sessions were conducted either on the same day or a day apart.

Isokinetic ergometer

A SRM Ergometer (Julich, Germany) cycle ergometer frame and flywheel were used to construct an isokinetic ergometer (Figure 4.1). The modified ergometer flywheel was driven by a 2.2-kW AC induction motor (ABB Ltd, Warrington, UK). The motor was controlled by a frequency inverter equipped with a braking resistor (Model: Altivar ATV312 HU22, Schneider Electric Ltd, London, UK). Using the motor enabled the participants to start their bouts at the target pedalling rate, rather than expending energy in accelerating the flywheel.

Figure 4.1: Participant set-up for laboratory testing on the isokinetic ergometer

The ergometer was fitted with Sensix force pedals (Model ICS4, Sensix, Poitiers, France) and a crank encoder (Model LM13, RLS, Komenda, Slovenia), sampling data at 200 Hz. Normal and tangential pedal forces were resolved using the crank and pedal angles into the effective (propulsive) and ineffective (applied along the crank) crank forces (Figure 4.2).

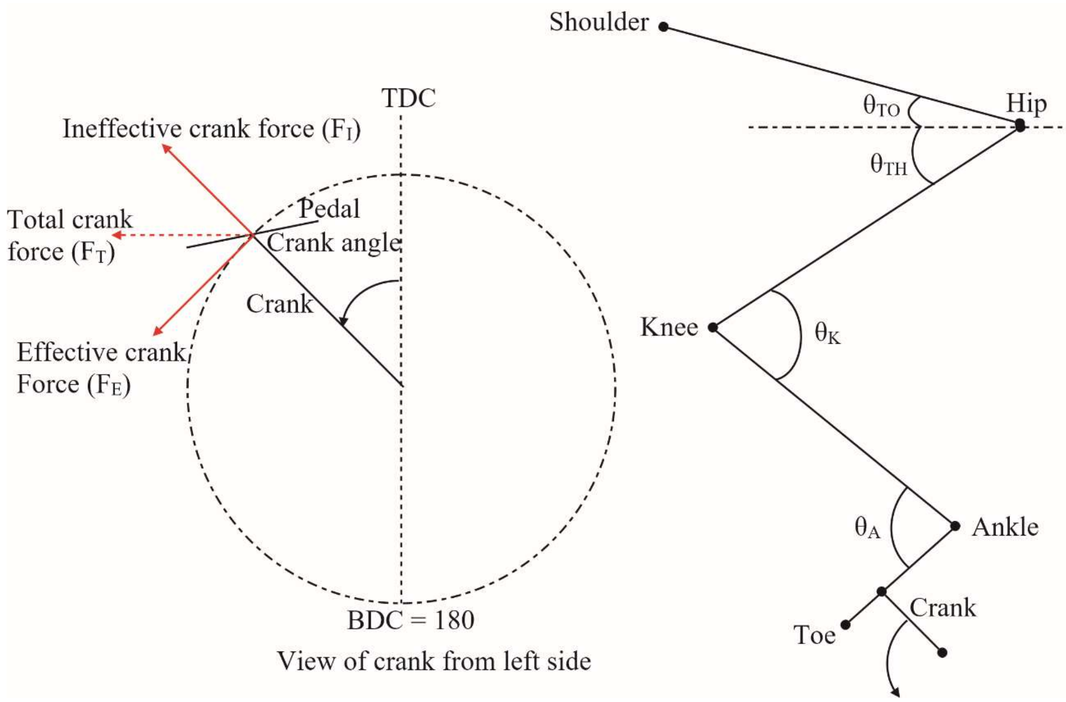

Figure 4.2: Left side joint angles and crank forces convention
TDC = top dead centre, BDC = bottom dead centre, θ_{TH} = thigh angle, θ_K = knee angle, θ_A = ankle angle, θ_{TO} = torso angle

Laboratory kinematic and kinetic data acquisition

In contemporary biomechanics studies of cycling, typically full 3D kinematics are measured which requires a large number of markers to be placed on the participant (Bini et al., 2016; Brochner Nielsen et al., 2018; Wilkinson et al., 2019). However, during cycling, the movement is predominantly in the sagittal plane (Umberger & Martin, 2001; van Ingen Schenau et al., 1990). Previous studies that have investigated maximal cycling have just considered the sagittal plane actions, as this is the plane where muscles produce power to generate effective crank force (Barratt et al., 2011; Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014). Therefore, for this programme of research, 2D kinematics in the sagittal plane was measured using a simple marker set (pedal spindle to define the point of force application, and lateral malleolus, lateral femoral condyle and greater trochanter to define the ankle, knee and hip joint centres respectively). This simple marker set has the added benefit of reducing time required for data collection sessions which is an important ethical consideration when working with elite athletes. Appendix 9.4 and 9.5 discusses and justifies the choice of using the greater trochanter marker to define the hip joint centre during maximal cycling.

Two-dimensional kinematic data of each participant's left side were recorded at 100 Hz using one high speed video camera with infra-red ring lights (Model: UI-522xRE-M, IDS, Obersulm, Germany). Reflective markers (15.9 mm diameter) were placed on the pedal spindle, lateral malleolus, lateral femoral condyle, greater trochanter and acromion (Figure 4.2). The same researcher attached the markers for all sessions. Kinematics and kinetics on the ergometer were recorded by CrankCam software (CSER, SHU, Sheffield, UK), which synchronised the camera and pedal force data, (down sampled to 100 Hz to match the camera data) and was used for data processing, including auto-tracking of the marker positions.

Velodrome kinematic and kinetic data acquisition

On the track the two-dimensional kinematics of each participant's left side were captured using 8 Qualisys Opus 7+ cameras, recorded at 200 Hz by Qualisys track manager software (QTM 2.14, Qualisys, Gothenburg, Sweden). Cameras were located in the track centre and covered a capture volume of 14 m x 1.5 m x 1.5m, along the black line from pursuit line to start of the bend (Figure 4.3).

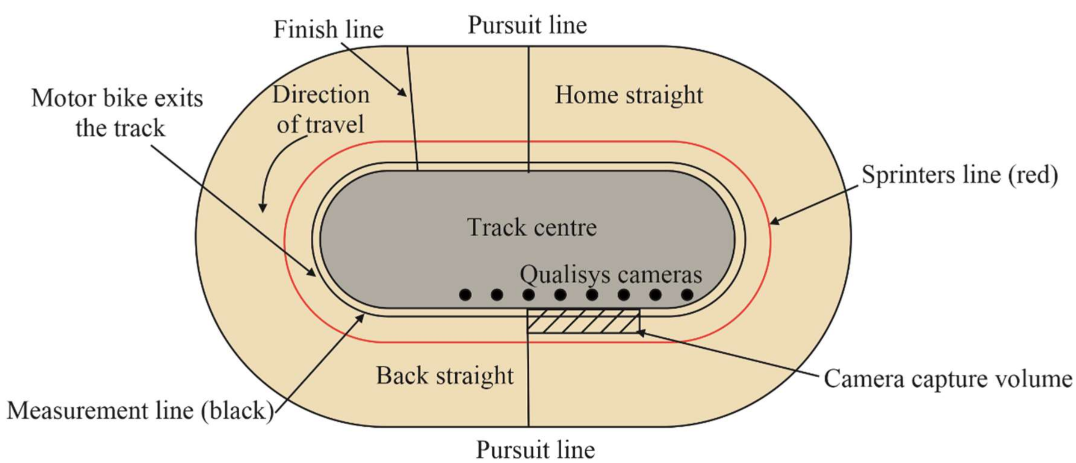

Figure 4.3: Schematic of 250 m track in velodrome with camera capture volume

The same marker set as used in the laboratory was supplemented with five markers to the left side of the bicycle frame to define the bicycle reference frame (rear wheel axle, seat stay, seat tube, downtube, front wheel axle – refer to Figure 4.4). Larger 19 mm diameter markers were used for the track data collection (Figure 4.4). A left force pedal (Model ICS4, Sensix, Poitiers, France) with a pedal strap was fitted to the rider's track bicycle. A 2 m cable ran from the pedal to a backpack containing a junction box (Sensix, Poitiers, France), Wi-Fi DAQ (Model cDAQ-9191+NI9205, National Instruments Corporation (U.K.) Ltd, Newbury, UK) and power source (Model Sony CP-V9B smartphone charger black portable charger 8700 mAh, Sony Europe B.V., Weybridge, UK) to transmit data from the force pedal to a laptop in track centre. The cable was attached to the riders' leg using Velcro straps.

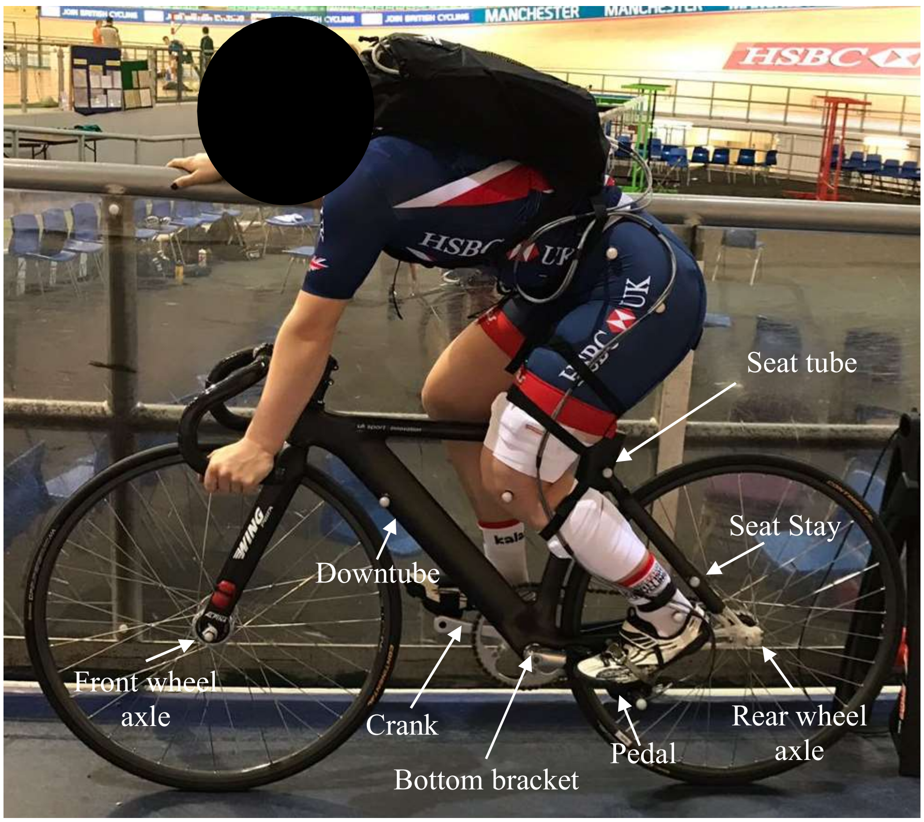

Figure 4.4: Participant set-up for track testing

EMG data acquisition

EMG signals were recorded continuously from nine muscles of the left leg: vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA), long head of biceps femoris (BF), semitendinosus (ST), lateralis gastrocnemius (GL), soleus (SO), and gluteus maximus (GMAX) with Delsys Trigno wireless surface EMG sensors (Delsys Inc, Boston, MA). The skin at the electrode placement sites was prepared by shaving the area then cleaning it with an alcohol wipe. The EMG sensors were then placed in the centre of the muscle belly - with the bar electrodes perpendicular to the muscle fibre orientation and secured using wraps to reduce motion artefacts during pedalling. The same researcher attached the EMG sensors for all sessions. A Delsys wireless sensor containing an accelerometer (148 Hz sampling rate) was attached to the left crank arm to obtain a measure of crank angle synchronised with the EMG signals. In the laboratory the EMG system was operated and recorded in EMGworks Acquisition software (Delsys Inc, Boston, MA), sampling data at 1926 Hz.

The track set-up was very similar, but the EMG system was operated and recorded via the Qualisys track manager software (QTM 2.14, Qualisys, Gothenburg, Sweden) which up sampled the EMG and accelerometer data to 2000 Hz. The EMG data were synchronised with the kinematic data using a Delsys trigger module (Model: SP-U02, Delsys Inc, Boston, MA) and an external trigger.

Body segment parameters

The body segment parameters (segmental mass, centres of mass and principal moments of inertia) were estimated via the tables of de Leva (1996) which were used in the inverse dynamics calculations for this programme of research (Chapters 4, 5 and 6). Sprint cyclists, however, have been shown to have larger thigh and calf girths then the average adult population (Foley et al., 1989; McLean & Parker, 1989). Wheat and Barratt demonstrated using a Monte Carlo simulation that the influence of uncertainties in body segment parameters were largest on the calculated hip joint power, particularly at higher pedalling rates (Wheat & Barratt, 2015). This means the joint moments and powers calculated for sprint cyclists will differ from the true value if use the body segment parameters are calculated from the tables in de Leva (1996). To obtain person specific foot, shank and thigh body segment parameters for each participant at each session would have been difficult and time consuming even using the 3D depth camera system described in (Bullas et al., 2016; Kordi et al., 2018) which is quick and easy to use. The research questions in this programme of research were interested in within- rather than between-participant differences, meaning the influence of error in the input body segment parameters will have a small impact on the conclusions of the studies, therefore, the tables produced by de Leva (1996) were used for this programme of research.

Data processing

All kinetic and kinematic data were filtered using a Butterworth fourth order (zero-lag) low pass filter with a cut of frequency of 14 Hz selected using residual analysis (Winter, 2009) (refer to Appendix 9.6 for details). The same cut off frequency was chosen for the kinematic and kinetic data as recommended by Bezodis and colleagues to avoid data processing artefacts in the calculated joint moments (Bezodis, Salo, & Trewartha, 2013). Instantaneous crank power was calculated from the product of the left crank torque and the crank angular velocity. The average left crank power was calculated by averaging the instantaneous left crank power over a complete pedal revolution. Joint angles were calculated using the convention shown in Figure 4.2. Torso angle was calculated as the angle between horizontal and a line connecting the acromion and greater trochanter (Wilkinson et al., 2019). Joint moments were calculated via inverse dynamics (Elftman, 1939), using pedal forces, limb kinematics, and body segment parameters (de Leva, 1996). Joint extension moments were defined as positive and joint flexion moments as negative. Joint powers at the ankle, knee and hip were determined by taking the product of the net joint moment and joint angular velocity.

Data were analysed using a custom Matlab (R2017a, MathWorks, Cambridge, UK) script. Each sprint on the ergometer lasted for 4 s providing six complete crank revolutions which were resampled to 100 data points around the crank cycle. During maximal cycling, fatigue occurs on a revolution by revolution basis (Tomas, Ross, & Martin, 2010). Therefore, to compare the same revolutions from the ergometer sprints (six revolutions per trial) to the track bicycle sprints (one revolution per trial), the fourth revolution from each ergometer sprint was selected to represent that trial. This assumed on track that 3 revolutions (which corresponds to distance of 24 m for the chosen track bicycle gear size) were completed after the cyclist was released by the motor-bicycle before they entered the capture volume, and therefore, the same revolution with the same level of fatigue was compared between the ergometer and track bicycle sprints. The laboratory session mean values for the times series variables for each participant were calculated from 3 revolutions (4^{th} revolution of each of the 3 sprints), with the exception of one participant who owing to technical problems only had data from 2 sprints, so the session average for this participant was created from 2 revolutions. Duty cycles for each joint were calculated as the ratio of time for extension to the time for flexion (Martin & Brown, 2009).

Track kinematics and kinetics were processed using a similar method used for the analysis of laboratory sprints. However, as the bicycle moved through the capture volume, marker trajectories were converted to a local coordinate system relative to the bicycle (the x direction was defined as the direction of travel and the z direction was defined as normal to the vertical plane of the pedal spindle) to match the laboratory coordinate system. The track force pedal data were synchronised with kinematic data using a Pearson's correlation to find the strongest correlation between the pedal angle measured by the force pedal encoder during the 3 heel raises and the pedal to ankle angle measured by the motion capture system, to identify the number of frames between the two sets of data so they could be synchronised. It was not possible to fit a crank encoder to the track bicycles due to the type of bottom bracket; therefore, the crank angle was calculated from the pedal marker trajectory. Due to the small capture volume of the cameras, only 1 revolution (7.93 m distance) for each trial was captured. There were also technical problems with force pedals Wi-Fi DAQ losing connection with the laptop during trials. Therefore, 3 efforts were obtained for 4 participants, 2 efforts for 1 participant, and 1 effort for 2 participants. The mean values of the time series variables for the track session were calculated from 1 to 3 revolutions, depending on the number of efforts recorded for each participant.

The accelerometer data for the crank arm were filtered using a Butterworth fourth order low pass filter with a cut off frequency of 10 Hz. The minimum value of the acceleration of the sensor in the direction of the crank arm corresponded to top dead centre (TDC) crank position. To synchronise the EMG data with the kinematic and kinetic data, the TDC locations from the accelerometer on the crank arm were matched to the corresponding TDC measured by the crank encoder. An additional analysis step had to be carried out for the accelerometer data from the track bicycle crank arm to resample the data to its native sampling frequency (148 Hz) as the QTM software up-sampled the data to 2000 Hz.

The raw EMG signals for the sprint efforts were high pass filtered (Butterworth second order, cut off frequency 30 Hz) to diminish motion artefacts (De Luca, Gilmore, Kuznetsov, & Roy, 2010), root mean squared (RMS, 25 ms window) and then low pass filtered (Butterworth second order, cut off frequency 24 Hz) (Brochner Nielsen et al., 2018). The laboratory data were then interpolated to 100 data points around the crank cycle (using spline interpolation method) and then averaged over 6 crank revolutions to create a linear envelope for each muscle. The EMG signals were normalised to the mean value in the linear envelope across the crank cycle for each muscle. The laboratory session mean EMG linear envelope was created from 18 revolutions. The track EMG signals were processed using the same method as used for the laboratory sprints. However, fewer revolutions (9.5 ± 4.7 revolutions) for each rider for the track session were obtained owing to problems of Wi-Fi signal drop out between the EMG sensors on the rider and the base station in track centre when the rider was on the opposite side of the track. During the track data collection several of the EMG sensors broke – typically they started reading a constant voltage. Therefore, the data for the VL (1 participant), SO (1 participant) and GL (2 participants) were removed from the laboratory session mean linear envelope to allow the same participants to be compared between the laboratory and track sprints.

Statistical analysis

Differences between the ergometer and track sprints for the discrete variables of pedalling rate, average left crank power over a complete revolution and duty cycles were assessed using paired *t*-tests. Differences in time series data (instantaneous crank powers, crank forces, joint angles, angular velocities, moments, powers and normalised EMG linear envelopes) between the ergometer and track sprints were assessed using statistical non-parametric mapping (SNPM); paired *t*-tests were used for all variables except crank forces where Hotelling's paired T^2 test was used (Pataky, 2010). Crank force consists of two vector components (effective and ineffective crank force), and therefore a multivariate statistical test was required (Pataky, 2010). The level of statistical significance was set to $P < 0.05$ for all tests.

4.4 Results

Discrete variables

There was a significant difference between average left crank power over a complete revolution for the sprints on the ergometer (516.1 ± 78.7 W) and the track (435.4 ± 59.3W) (P = 0.001) (which gives an indicative total crank power over a complete revolution for both cranks of 1032 and 871 W for the ergometer and track sprints respectively). There was a significant difference between the knee joint duty cycle for the sprints on the ergometer and the track (Table 4.1). The mean pedalling rate for the sprints on the ergometer was 135.4 ± 1.2 rpm and for the track 137.5 ± 1.8 rpm (P = 0.061).

Table 4.1: Duty cycles for the ergometer and the track sprints.

	Mean (SD)		
	Laboratory	**Track**	***P***
Ankle	0.79 ± 0.21	0.86 ± 0.38	0.424
Knee	1.07 ± 0.06	1.01 ± 0.04	0.041*
Hip	0.92 ± 0.07	0.93 ± 0.15	0.835

* Indicates significant difference between conditions ($P < 0.05$)

Time series variables

The crank powers were significantly greater ($P < 0.05$) for the ergometer compared to track sprints for parts of the crank cycle (39° to 88° and 338° to 350°) (Figure 4.5). The crank forces were significantly greater ($P < 0.05$) for the ergometer compared to track sprints between crank angles 328° and 340° (Figure 4.6).

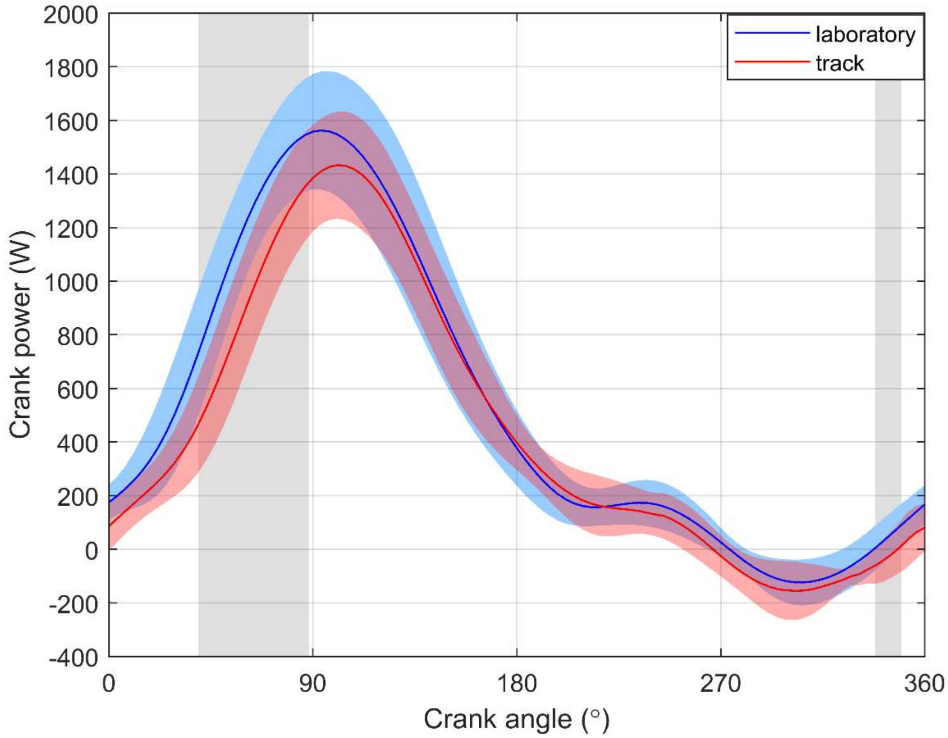

Figure 4.5: Comparison of mean crank powers for the ergometer and the track sprints.

Areas of the graph shaded grey where the SNPM is significant.

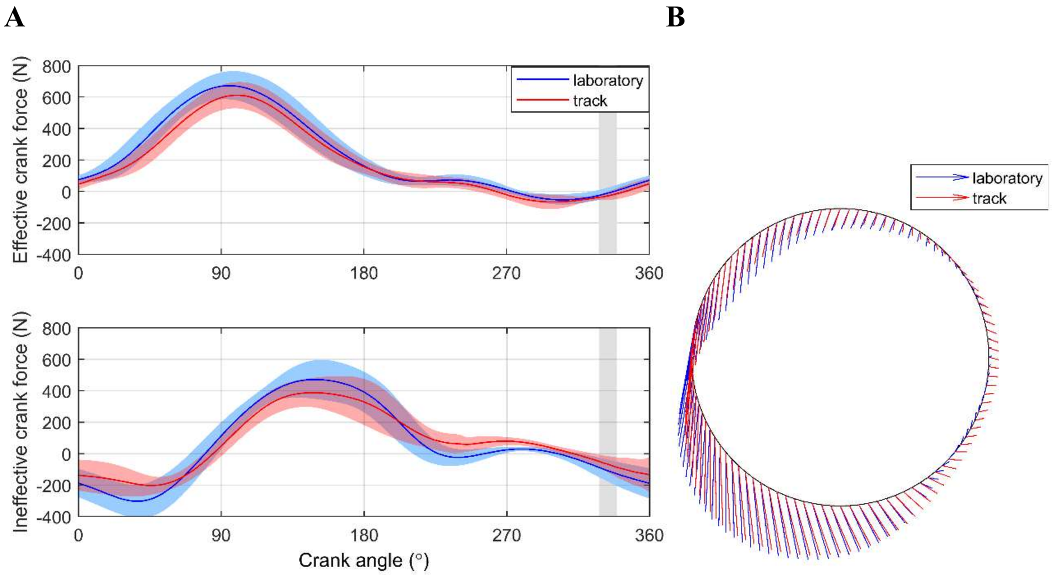

Figure 4.6: Comparison of mean crank forces for the ergometer and the track sprints.

A: Crank force separated into effective and ineffective components

B: Visualisation of crank forces

Areas of the graph shaded grey where the SNPM is significant.

The joint angles were significantly greater ($P < 0.05$) for the ergometer compared to the track sprints between crank angles: knee: 23° to 220°, hip: 152° to 192°, and torso: 0 to 360° (Figure 4.7 and Figure 4.8). The joint angular velocities were significantly different ($P < 0.05$) between the ergometer and track sprints for proportions of the crank cycle (knee: 202° to 226° (greater flexion angular velocity on the ergometer) and 355° to 46° (greater extension angular velocity on the ergometer), and hip: 133° to 159° (greater extension angular velocity on the ergometer), 202° to 232° (greater flexion angular velocity on the track) and 355° to 42° (greater extension angular velocity on the ergometer)) (Figure 4.7). The joint moments were significantly different ($P < 0.05$) between the ergometer and track sprints for proportions of the crank cycle (ankle: 248° to 300° and 336° to 346°(greater on the track), knee: 234° to 267° (greater flexion moment on the track), and 335° to 12° (greater extension moment on the ergometer), and hip: 281° to 311° (greater flexion moment on the ergometer)) (Figure 4.7). The joint powers were significantly different ($P < 0.05$) between the ergometer and track sprints for proportions of the crank cycle (knee: 184° to 193° (greater on the ergometer), and

324° to 330° (greater on the track), and hip: 10° to 21° and 228° to 316° (greater on the ergometer)) (Figure 4.7).

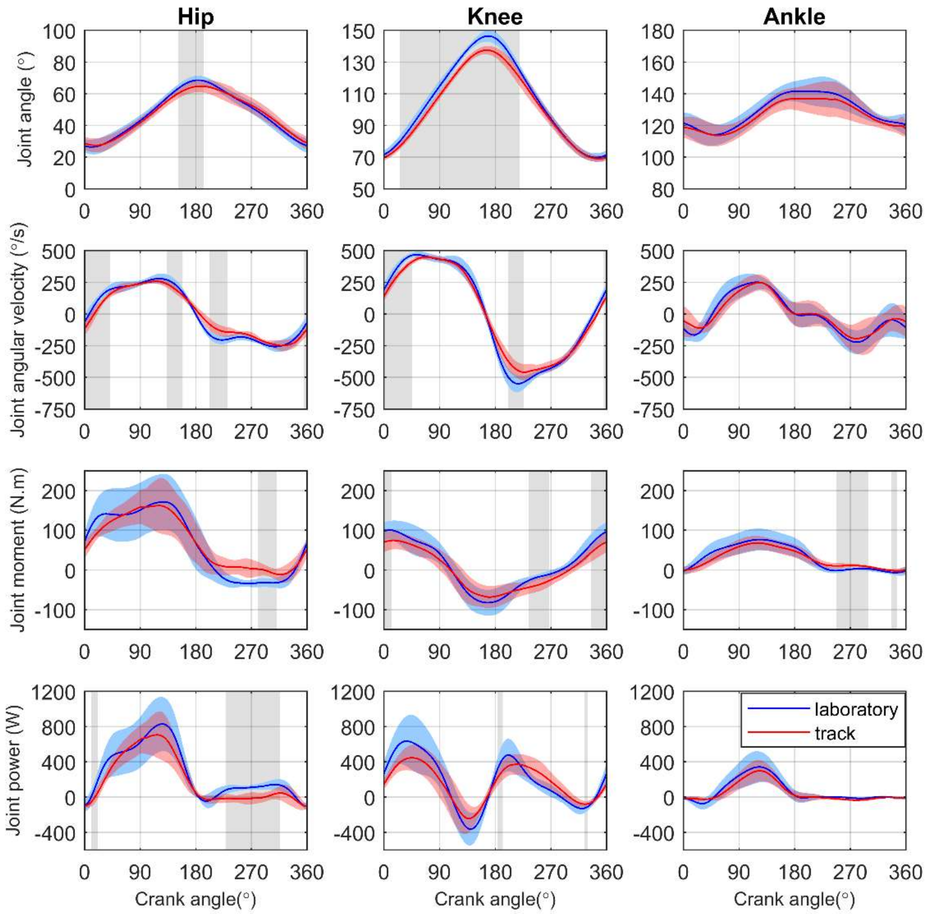

Figure 4.7: Comparison of mean joint angles, angular velocities, moments and powers for the ergometer and the track sprints.

Areas of the graph shaded grey where the SNPM is significant.

For ease of presenting the data the thigh angle and angular velocity are presented as hip angle and angular velocity

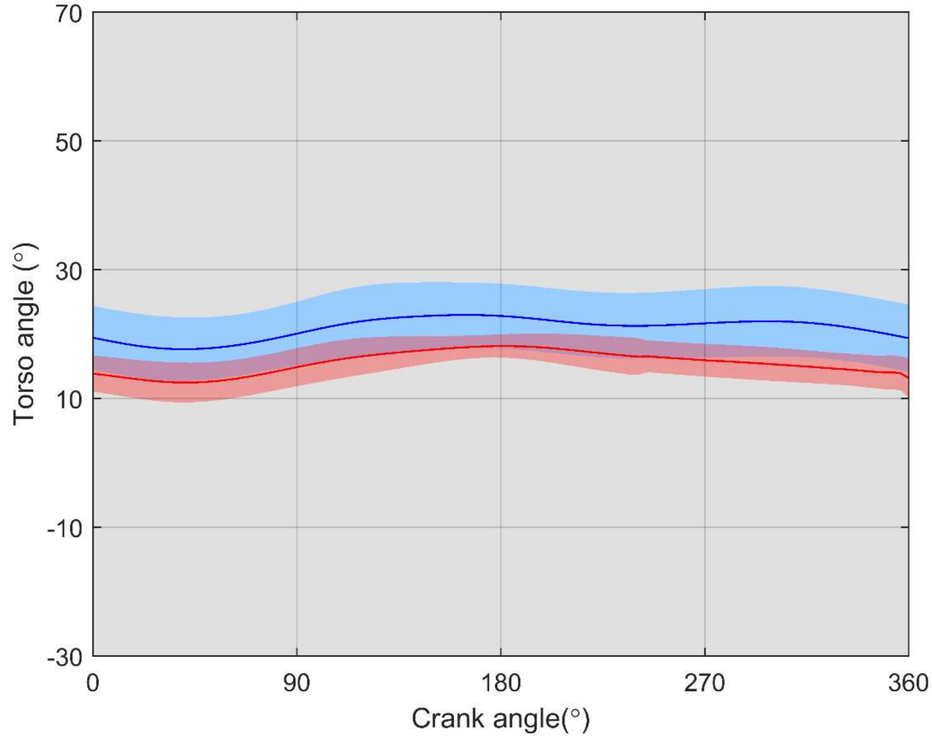

Figure 4.8: Comparison of torso angle for the ergometer and the track sprints. Areas of the graph shaded grey where the SNPM is significant.

EMG activity was the significantly different ($P < 0.05$) between ergometer and track sprints between crank angles for the VL: 84 to 95°, RF: 226° to 244°, VM: 74° to 90°, TA: 210° to 233°, ST: 205° to 243°, SO: 85° to 98° and GMAX: 85° to 109° (Figure 4.9).

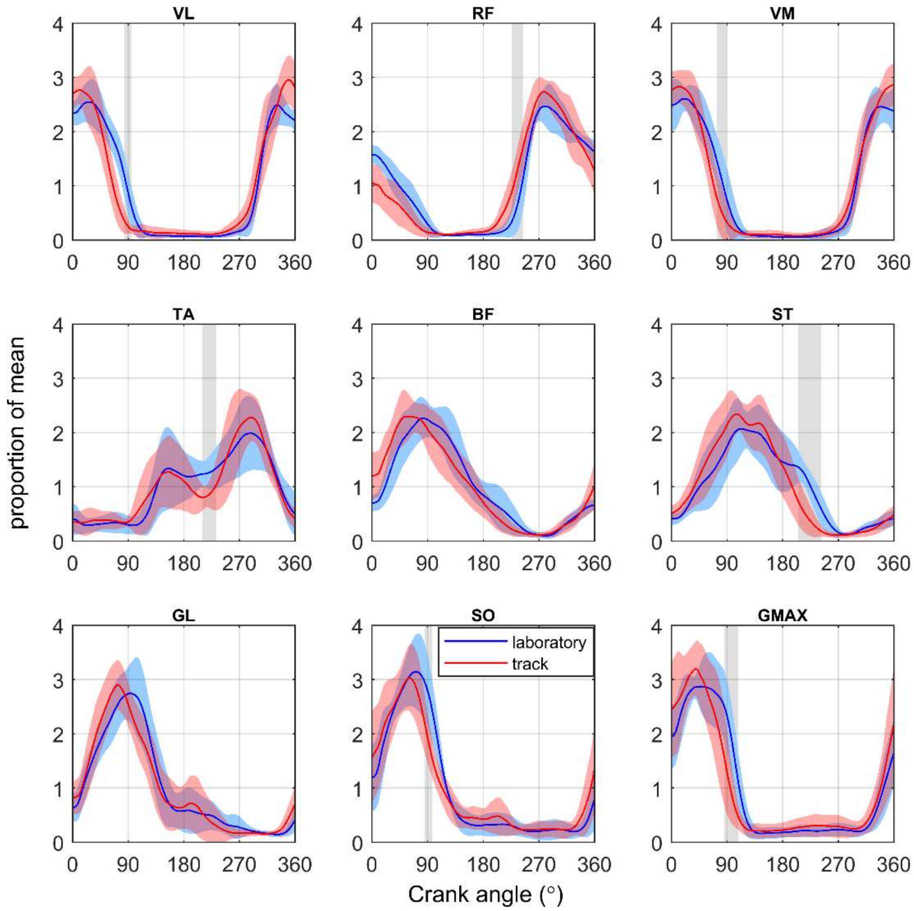

Figure 4.9: Comparison of EMG linear envelopes (normalised to mean value in signal) for each muscle for the ergometer and the track sprints.
VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF = biceps femoris, ST = semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus. Areas of the graph shaded grey where the SNPM is significant.

4.5 Discussion

Cyclists produced higher crank power on an isokinetic ergometer than in the flying half lap efforts on the track. There was a higher peak crank power, differences in rider position, greater knee moment over TDC and later offset of main power producing muscles for the sprints on the ergometer compared to the track. When sprinting on the ergometer, the riders only have to focus on producing maximum power, whereas on the track they also have to control the bicycle direction and stability whilst trying to produce maximal power (Gardner et al., 2007), which may be a limiting factor in producing maximum power on the track.

The maximum crank power measured for the ergometer sprints (left crank: 516.1 W) is similar to the maximum pedal power measured by in a study by Elmer and co-workers - 563 W for a 3 second sprint at 120 rpm on an isokinetic ergometer (Elmer et al., 2011). This is slightly higher than measured in the current study. However, this would be expected as their participants were all male and the effort was at slightly lower pedalling rate. Both of these factors would be expected to increase power output. The shape and magnitude of the joint powers throughout the crank cycle for the ergometer sprints are very similar to previous studies of maximal cycling (Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014).

Thigh and knee angles were larger during the downstroke on the ergometer than the track bicycle, signifying the cyclists were pedalling with a straighter leg on the ergometer (Figure 4.7). The duty cycle for the knee joint was significantly greater for the ergometer compared to the track sprints (Table 4.1), indicating the knee joint spent longer extending throughout the crank cycle during the sprints on the ergometer. Increased duty cycles have previously been shown to be an important strategy to maximise power production in cycling (Elmer et al., 2011; Martin & Brown, 2009). An increased duty cycle increases the portion of the cycle during which a muscle can produce work, hence increasing power production (Askew & Marsh, 1997; Martin & Brown, 2009). The increased duty cycle on the ergometer was associated with a significantly larger knee moment around TDC for the sprints on the ergometer compared to the track (Figure 4.7). Both of these could be contributing mechanisms for the higher crank power observed for the sprints on the ergometer. The EMG activity patterns were similar between the ergometer and track sprints (Figure 4.9). However, the main power producing muscles (VL/VM/GMAX) had a significantly later offset of EMG activity for the ergometer compared to the track sprints. This meant that these muscles were active for a larger portion on the crank cycle, in particular during the major power producing phase of the crank cycle, which might also be a contributing factor to the increased duty cycles and crank power produced during the downstroke for the ergometer sprints.

On the ergometer participants displayed a tendency to hover over the saddle (despite the instruction to remain seated) possibly because they did not have to control stability and direction of a moving bicycle. This altered riding position potentially allowed them to produce more crank power on the ergometer by increasing knee joint duty cycle and knee moment over TDC with delayed offset of the main power producing muscles (VL/VM/GMAX). Changes in cycling position have been shown to alter EMG activation patterns for lower limb muscles (Dorel et al., 2009). Therefore, the slightly altered riding position on the ergometer compared to the track bicycle might have influenced the EMG activation patterns. The cyclists also had a shallower torso angle on the track bicycle compared to the ergometer (Figure 4.8), which might be due to the different environmental constraints between the velodrome and the laboratory, as the ergometer was set-up to match the track bicycle position. The altered environmental and task constraints for the track sprints (for example: the control of a moving bicycle, the banking of the track, air resistance when moving around the track and the environment in the velodrome) might have influenced the joint angles and rider position, although further research is required to investigate which of these constraints caused the changes in rider position.

There were significant differences during part of the upstroke for the bi-articular RF and ST muscles between the ergometer and track sprints (Figure 4.9). During this region of the crank cycle, these muscles activate to flex the hip and knee joints to actively pull up during the upstroke. Therefore, slight alterations in these muscles activation patterns in the upstroke might explain the differences in knee and hip joint moments during the upstroke between the ergometer and track sprints.

The negative ankle power between TDC and 45° was smaller for track sprints compared to the ergometer, although not statistically significant. During the downstroke the gastrocnemius muscle is thought to experience a stretch-shortening cycle, where it actively lengthens (energy is absorbed and stored in the muscle-tendon unit) and then shortens, potentially reusing the elastic strain energy that was stored in the muscle-tendon unit (Gregor et al., 1991). McDaniel and co-workers demonstrated that the negative power absorption at the ankle joint during maximal cycling was greatest at a pedalling rate of 90 rpm and then gradually reduced with increasing pedalling rate, with minimal negative power at 150 rpm and at 180 rpm positive power is produced in this region of the crank cycle (McDaniel et al., 2014). This reduction in negative ankle power with increased pedalling rate was associated with a reduction in peak positive ankle power produced in the downstroke, potentially due to less elastic strain energy being stored in the muscle-tendon unit between TDC and 45° (McDaniel et al., 2014). Therefore, the greater negative ankle power in the ergometer sprints and the role of the stretch-shortening cycle might partially explain the increased peak positive ankle power during the downstroke for the ergometer sprints compared to the track.

The finding that cyclists produced higher crank power on an ergometer than on the track is incongruent with data reported by Gardiner and colleagues who found similar peak power values between ergometer and track sprints, with considerable individual variability (Gardner et al., 2007). However, their experimental protocol was different to the current study. They used an inertial load ergometer in the laboratory, and both the ergometer and track efforts were from a standing start, so the power was measured during the acceleration phase. This meant maximum speed of the track bicycle in the Gardner et al. (2007) study was 41 km/h with a maximum pedalling rate of 100 rpm, compared to 62.5 km/h and 135 rpm in this study. They also recorded power data using an SRM power meter which only samples at 5 Hz (Gardner et al., 2007), whereas the Sensix force pedals used in the current study sampled at 200 Hz. The differences in the experimental protocol between the two studies may account for the differences in the findings. Also, in contrast to the findings of Gardner et al. 2007, differences in crank-level variables between field and laboratory testing have been found in other cycling disciplines such as road and BMX (Bertucci et al., 2007; Gardner et al., 2007; Rylands et al., 2015). A study comparing sprints on an ergometer in the laboratory to riding on a BMX indoor track found significantly higher peak power (34%) and reduced time to peak power production in field tests compared to laboratory (Rylands et al., 2015). Their findings are opposite to this study which found track sprint cyclists produced higher crank power on the ergometer compared to the track. However, there are specific differences between seated flying efforts on the track and riding a BMX bicycle down a starting ramp. BMX riders use their upper body to contribute to power production by oscillating the bicycle from side to side. On a fixed ergometer they are unable to do this, and hence produce reduced crank power (Rylands et al., 2015).

There were several differences in the kinetic measuring equipment between the ergometer and track sprints. The riders' track bicycles were fitted with the left force pedal with a normal pedal on the right-hand side. Force pedals are wider and deeper than a standard pedal and the right pedal would have hit the track around the banked bends. There was a difference in mass between the two pedals on the track. However, as a flying effort was used, the effect of this mass difference was expected to be small. The riders also, did not report any perceived differences in bicycle handling. The participants had to wear a backpack when riding on the track and this may have influenced their pedaling technique and position on the bicycle. However, the backpack was light, and therefore the effect was expected to be small. For future development of the on-track testing method, possible solutions to measure kinetics without the need for the backpack should be investigated to minimise interference to the performer.

There were several technical problems when collecting data in the field related to maintaining Wi-Fi connection between the EMG sensors and the Wi-Fi DAQ for the force pedal on the rider and the laptop recording the data in track centre. The Delsys EMG system suffered from dropped data when the Wi-Fi signal between the base station and sensors was lost when the rider moved away from the base station. A possible solution to this problem would be to have a data logger on the bicycle to record the EMG data. However, this would make synchronising the EMG data with the kinematic data difficult as currently a hardwired trigger switch is used. It would also mean it would not possible to assess the quality of the EMG data during the testing session which was a problem identified by Blake and Wakeling in their study of outdoor cycling (Blake & Wakeling, 2012). A data logger on the bicycle would also be a solution for the problems with the force pedals Wi-Fi connection. Another limitation of the testing method was that it was not possible to measure crank angle directly owing to the type of bottom bracket on the track bicycles which meant a crank encoder could not be fitted. Therefore, the crank angle had to be calculated from the pedal spindle marker trajectory which might have introduced small errors in the crank angle.

This study had a small sample size (7 participants) and relatively few crank revolutions were measured for the track sprints, which reduced the statistical power and there was relatively large inter-participant variability. Therefore, further research is needed with a larger sample size and observing more crank revolutions for the track efforts to investigate if the same behaviour is observed.

4.6 Conclusion

There are relatively small differences in movement organisation between sprinting on a velodrome track and on an ergometer. However, the static task constraints of ergometer cycling led the cyclists to adopt a different position, increasing knee joint duty cycle, knee moment over TDC and later offset of muscle activity in the main power producing muscles. All these factors might contribute to an increased overall crank power output on the ergometer compared to the track where the cyclists also needed to control the stability and direction of the bicycle. Future research is needed to assess whether the differences in joint angles, EMG activity and crank powers were due to the different environmental and task constraints between the ergometer and the track bicycle sprints. The on-track data collection method has the potential to be a useful tool to help coaches assess pedalling on a track. The findings imply it is important to undertake biomechanical analyses of movement organisation in elite sports practice in a representative environment. Further research is needed to investigate differences between cycling on an ergometer and a track bicycle using a larger sample of participants and observing a greater number of crank revolutions of track data.

Although this study revealed differences in sprint cycling biomechanics between sprinting on the ergometer and on the track, a decision was made to use the ergometer in the laboratory for the testing protocol for the following studies (Chapter 5 and 6). The reasons for this decision were: the current on-track data collection method can only measure one revolution per effort on track due to the limitations of the equipment available, which is insufficient to study coordination due to inherent between-revolution variability in maximal cycling. In addition, there were technical problems during the on-track data collection sessions where Wi-Fi connection was lost between the EMG sensors, force pedals and the laptop recording the data which meant data from trials were lost. It was also very difficult to obtain the track at the velodrome for testing sessions. This meant there was a risk that it would be not be possible to collect data from many participants or at the time intervals required by the research question.

Therefore, the test-retest reliability of the biomechanical variables measured during maximal cycling on the ergometer is quantified in the next chapter.

5 Biomechanical measures of maximal cycling on an ergometer: a test-retest study

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5.1 Introduction

The reliability of a clinical or sports science test is defined as the consistency or reproducibility of a performance when a test is performed repeatedly (Hopkins, Schabort, & Hawley, 2001). This is an important consideration for researchers, clinicians and applied sports scientists as the better the reliability of the measurement, the easier it is to detect a real change in outcome (Hopkins, 2000). If the reliability of a test is low, then the outcome of a test may conceal the true effect of an intervention. Conversely, if the reliability of a test is not known then small random deviations may be misinterpreted as a meaningful change in performance (Yavuzer, Öken, Elhan, & Stam, 2008).

Applied biomechanics researchers are often interested in assessing the short- or long-term effects of interventions that aim to improve clinical or sports performance outcomes. In clinical gait analysis, for example, the results of biomechanical assessments are used to inform clinical decision making, by evaluating the effectiveness of interventions such as surgery, physical therapy, medication or orthotics on gait biomechanics (Kadaba et al., 1989; McGinley, Baker, Wolfe, & Morris, 2009; Yavuzer et al., 2008). Test-retest reliability studies of clinical gait have found that the sagittal plane kinematics and kinetics have high values of reliability in comparison to the data collected in the transverse and coronal planes (McGinley et al., 2009). Furthermore, knee abduction/adduction and hip, knee and foot rotation joint angles demonstrate the lowest reliability (McGinley et al., 2009), with the size of the measurement error the same order of magnitude as the real joint motion in these planes. In the context of clinical gait reliability studies have therefore proved valuable by identifying those variables that need to be interpreted with particular caution in order to effectively inform clinical decision making (McGinley et al., 2009).

An understanding of test-retest reliability has similar relevance when assessing sporting movements, as biomechanical measures are often used to evaluate the effectiveness of longitudinal interventions such as changes to training programmes or equipment modification (Costa, Bragada, Marinho, Silva, & Barbosa, 2012; Milner, Westlake, & Tate, 2011). Cycling is a commonly used sporting movement for this purpose, as it is a relatively constrained movement that can be accurately manipulated (Neptune et al., 1997; Neptune & Kautz, 2001). Whilst the reliability of submaximal or “endurance” cycling is well reported (Bini & Hume, 2013; Hopkins et al., 2001; Jobson et al., 2013; Laplaud et al., 2006), only a small amount by comparison is known about the reliability of short-term maximal cycling. This comparative deficit exists despite maximal cycling being an important paradigm for studying physiological capacity (Coso & Mora-Rodríguez, 2006), muscle coordination and motor control strategies, as well as having direct relevance to a range of competitive cycling performance environments (Martin et al., 2007). Therefore, quantifying test-retest reliability in maximal cycling biomechanics is important. Test-retest reliability has been quantified for overall net crank power output on an inertial load cycling ergometer within- and between-session (Coso & Mora-Rodríguez, 2006; Hopkins et al., 2001; Mendez-Villanueva, Bishop, & Hamer, 2007), with trained cyclists producing reliable power within the first testing session (Martin, Diedrich, & Coyle, 2000). There have been no studies quantifying the within- and between-session reliability of biomechanical variables (crank power and forces, joint angles, angular velocities, moments and powers and EMG activity) for short-term maximal cycling despite these measures being important descriptors of the outcome, technique and intermuscular coordination of a movement (Brochner Nielsen et al., 2018; Jacobs & van Ingen Schenau, 1992; Wakeling, Blake, & Chan, 2010).

EMG activity can be used to determine muscle activation onset and offset times and level of activation (Dorel et al., 2012; Hug & Dorel, 2009). This is important when investigating intermuscular coordination in cycling, as the timing and magnitude of muscle activation has to be coordinated appropriately to allow an efficient energy transfer from the muscles though the body segments to the pedal (Neptune & Kautz, 2001; Raasch et al., 1997). Joint kinetic measures (moments and powers) at the hip, knee and ankle throughout the pedal revolution describe the action and contribution of the joints to pedal power and can be used to identify different coordination strategies between cyclists (Elmer et al., 2011; Martin & Brown, 2009; McDaniel et al., 2014). Combining information on muscle activation from EMG and joint kinetics from inverse dynamics analysis provides a deeper understanding of the joint and muscle actions that produce the movement. Hence, both are required to describe intermuscular coordination in maximal cycling and were chosen for measurement and analysis in this study (Brochner Nielsen et al., 2018; Dorel, 2018b).

The aim of this study was to quantify the test-retest reliability of kinematic, kinetic, and muscle activation variables during maximal sprint cycling. It was hypothesised that within-session reliability would be better than between-sessions reliability.

5.2 Methods

Participants

Fourteen track sprint cyclists participated in the study. Participants regularly competed at track cycling competitions at either Master's international and national level (10), or Junior national level (4). Although the participants were varied in their anthropometrics (7 males and 7 females, age: 40.5 ± 17.7 yr, body mass: 72.5 ± 8.5 kg, height: 1.71 ± 0.06 m), they were similar with respect to cycling performance level (flying 200 m PB: 11.98 ± 0.90 s). Participants were provided with study details and gave written informed consent. The study was approved by the Sheffield Hallam University Faculty of Health and Wellbeing Research Ethics Sub-Committee.

Experimental protocol

An isokinetic ergometer was set up to replicate each participant's track bicycle position. All participants' crank lengths were set to 165 mm, which was what they rode on their track bicycles. Riders undertook their typical warm-up on the ergometer at self-selected pedalling rate and resistance for at least 10 minutes, followed by one familiarisation sprint (4 seconds at 135 rpm). Martin and colleagues demonstrated that trained cyclists can produce valid and reliable results for maximal cycling power from the first testing session (Martin et al., 2000), and so one familiarisation sprint was deemed appropriate.

Riders then conducted 3 x 4 s seated sprints at a pedalling rate of 135 rpm on the isokinetic ergometer with 4 minutes' recovery between efforts. Participants undertook an identical session 7.6 ± 2.5 days apart, at approximately the same time of day (0.11 ± 2.18 h). A pedalling rate of 135 rpm was chosen as this is a typical pedalling rate during the flying 200 m event in track cycling and within the optimal pedalling rate range for track sprint cyclists (Dorel et al., 2005). The competitive level and typical training volume of the participants meant that it was not feasible to ask them to stop exercising 24 hours prior to the testing sessions, so instead they were instructed to undertake the same training in the preceding 24 hours before both sessions.

Isokinetic ergometer

A SRM Ergometer (Julich, Germany) cycle ergometer frame and flywheel were used to construct an isokinetic ergometer. The modified ergometer flywheel was driven by a 2.2-kW AC induction motor (ABB Ltd, Warrington, UK). The motor was controlled by a frequency inverter equipped with a braking resistor (Model: Altivar ATV312 HU22, Schneider Electric Ltd, London, UK). This set-up enabled the participants to start their bouts at the target pedalling rate, rather than expending energy in accelerating the flywheel. The ergometer was fitted with Sensix force pedals (Model ICS4, Sensix, Poitiers, France) and a crank encoder (Model LM13, RLS, Komenda, Slovenia), sampling data at 200 Hz. Normal and tangential pedal forces were resolved using the crank and pedal angles into the effective (propulsive) and ineffective (applied along the crank) crank forces (Figure 4.2).

Kinematic and kinetic data acquisition

Two-dimensional kinematic data of each participant's left side were recorded at 100 Hz using one high speed video camera with infra-red ring lights (Model: UI-522xRE-M, IDS, Obersulm, Germany). Reflective markers were placed on the pedal spindle, lateral malleolus, lateral femoral condyle, greater trochanter and iliac crest. The same researcher attached the markers for all sessions. The choice and justification for the marker set used, and the definition of hip joint centre are discussed in section 4.2 and appendices 9.4 and 9.5. Kinematics and kinetics on the ergometer were recorded by CrankCam software (CSER, SHU, Sheffield, UK), which synchronised the camera and pedal force data, (down sampled to 100 Hz to match the camera data) and was used for data processing, including auto-tracking of the marker positions.

EMG data acquisition

EMG signals were recorded continuously from nine muscles of the left leg: vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA), long head of biceps femoris (BF), semitendinosus (ST), lateralis gastrocnemius (GL), soleus (SO), and gluteus maximus (GMAX) with Delsys Trigno wireless surface EMG sensors (Delsys Inc, Boston, MA). The skin at the electrode placement sites was prepared by shaving the area then cleaning it with an alcohol wipe. The EMG sensors were then placed in the centre of the muscle belly - with the bar electrodes perpendicular to the muscle fibre orientation and secured using wraps to reduce motion artefacts during pedalling. The same researcher attached the EMG sensors for all sessions. A Delsys wireless sensor containing an accelerometer (148 Hz sampling rate) was attached to the left crank arm to obtain a measure of crank angle synchronised with the EMG signals. The EMG system was operated and recorded in EMGworks Acquisition software (Delsys Inc, Boston, MA), sampling data at 1926 Hz.

Data processing

All kinetic and kinematic data were filtered using a Butterworth fourth order (zero-lag) low pass filter with a cut off frequency of 14 Hz selected using residual analysis (Winter, 2009) (refer to Appendix 9.6 for details). The same cut off frequency was chosen for the kinematic and kinetic data as recommended by Bezodis and colleagues to avoid data processing artefacts in the calculated joint moments (Bezodis et al., 2013). Instantaneous crank power was calculated from the product of the left crank torque and the crank angular velocity. The average left crank power was calculated by averaging the instantaneous left crank power over a complete pedal revolution. Owing to a technical fault with the force measurement in the right pedal, it was not possible to calculate total average crank power per revolution (sum of left and right crank powers). Joint angles were calculated using the convention shown in Figure 4.2. Joint moments were calculated via inverse dynamics (Elftman, 1939), using pedal forces, limb kinematics, and body segment parameters (de Leva, 1996). Joint extension moments were defined as positive and joint flexion moments as negative. The use of de Leva (1996) body segment parameters for this programme of research is discussed in section 4.2. Joint powers at the ankle, knee and hip were determined by taking the product of the net joint moment and joint angular velocity.

Data were analysed using a custom Matlab (R2017a, MathWorks, Cambridge, UK) script. Each sprint lasted for 4 s providing six complete crank revolutions which were resampled to 100 data points around the crank cycle. Crank forces and powers, joint angles, angular velocities, moments and powers were averaged over these revolutions to obtain a single ensemble-averaged time series for each trial.

The accelerometer data for the crank arm was filtered using a Butterworth fourth order low pass filter with a cut off frequency of 10 Hz. The minimum value of the acceleration of the sensor in the direction of the crank arm corresponded to top dead centre (TDC) crank position. To synchronise the EMG data with the kinematic and kinetic data, the TDC locations from the accelerometer on the crank arm were matched to the corresponding TDC measured by the crank encoder.

The raw EMG signals for the sprint efforts were high pass filtered (Butterworth second order, cut off frequency 30 Hz) to diminish motion artefacts (De Luca et al., 2010), root mean squared (RMS, 25 ms window) and then low pass filtered (Butterworth second order, cut off frequency 24 Hz) (Brochner Nielsen et al., 2018). The data were then interpolated to 100 data points around the crank cycle (using spline interpolation method) and then averaged over 6 crank revolutions to create a linear envelope for each muscle. The EMG signals were normalised to the mean value in the linear envelope across the crank cycle for each muscle. The reliability of different EMG signal normalisation methods is discussed in Appendix 9.7 and evidence provided for the choice of normalising the EMG signals to the mean value for this programme of research. Appendix 9.8 provides the justification of why muscle activity bursts 'on/off' were not used in the EMG analysis for this programme of research.

Statistical analysis

Differences in discrete values between sessions were assessed using paired *t*-tests. Differences in time series data (instantaneous crank powers, crank forces, joint angles, angular velocities, moments, powers and normalised EMG linear envelopes) between sessions were assessed using statistical parametric mapping (SPM); paired *t*-tests were used for all variables except crank forces where Hotelling's paired T^2 test was used (Pataky, 2010). Crank force consists of two vector components (effective and ineffective crank force), and therefore a multivariate statistical test was required (Pataky, 2010). The level of statistical significance was set to $P < 0.05$ for all tests.

The reliability of the discrete variables between sessions was assessed using intra-class correlation coefficient (ICC) tests. ICCs were calculated using IBM SPSS Statistics Version 24 (IBM UK Ltd, Portsmouth, UK), based on average measures, absolute agreement, two-way mixed effects model (ICC (3,*k*) - where *k* is equal to the number of trials in a session, which in this study is three). The ICCs were interpreted using Koo and Li's guidelines: values less than 0.50 are indicative of poor reliability, between 0.50 and 0.75 indicates moderate reliability, 0.75 to 0.90 indicates good reliability and > 0.90 indicates excellent reliability (Koo & Li, 2016). For a variable to be considered as having excellent reliability, both upper and lower bounds of the 95% confidence intervals must fall within the excellent range (i.e. > 0.9) (Koo & Li, 2016).

Standard error of measurement (SEM) for between sessions was calculated using the formula (Weir, 2005), where SD is standard deviation of the mean difference:

$$SEM = SD\sqrt{1 - ICC} \tag{1}$$

Minimal detectable difference (MDD) was calculated for between sessions using the formula (Weir, 2005):

$$MDD = SEM \times 1.96 \times \sqrt{2} \tag{2}$$

The coefficient of variation (CV) was calculated for the average left crank power over a complete revolution (Hopkins, 2000).

The adjusted coefficient of multiple determination (R_a^2) was calculated for the kinematic, kinetic and EMG time series data to evaluate the reliability of these waveforms within- and between-session (Kadaba et al., 1989). When the waveforms are similar R_a^2 tends to 1, and if they are dissimilar R_a^2 tends to 0. The adjusted coefficient of multiple correlation (CMC) was calculated by taking the square root of the adjusted coefficient of multiple determination (Kadaba et al., 1989). CMC for within- and between-session for each time series variable were calculated for each participant and then averaged across all participants to obtain the mean and SD value of CMC. Growney and colleagues suggested values of CMC greater than 0.8 represent a fairly high degree of reproducibility (Growney, Meglan, Johnson, Cahalan, & An, 1997).

Apart from this reference there are no published criteria for interpretation of CMC: therefore, the guidelines for interpreting the Pearson's correlation coefficient were used, where values < 0.5 indicate low repeatability, 0.5 to 0.7 indicates moderate repeatability, 0.7 to 0.9 indicates good repeatability and > 0.9 indicates excellent repeatability (Hinkle, Wiersma, & Jurs, 2003). The EMG data were visually inspected for signal quality and the frequency spectrum of the raw and filtered EMG signal calculated. EMG signals with a high frequency content below 20 Hz, indicates low frequency noise due to movement artefact (De Luca et al., 2010) and therefore, these trials were discarded. Therefore, the CMC for within- and between-session for the EMG linear envelopes of the VL, VM, ST, and GMAX muscles were calculated using 13 participants. At least 2 trials for each muscle per session per participant were required to calculate CMC. The calculated reliability of the EMG data, therefore, is the upper bound as very noisy trials were discarded.

The cross-correlation coefficient (*R*) was calculated to compare the temporal effects of within- and between-session EMG linear envelopes (Wren, Do, Rethlefsen, & Healy, 2006). The between-sessions cross-correlation coefficient was calculated comparing the session mean EMG linear envelope, and within-session the cross-correlation coefficient was calculated comparing the EMG linear envelope for two trials.

5.3 Results

Discrete variables

All discrete variables demonstrated good to excellent within-session reliability ICC(3,1) > 0.833 (Table 5.1). Discrete crank level variables demonstrated good to excellent between-sessions reliability ICC(3,k) > 0.756 (Between-sessions reliability for kinematic and kinetic variables. Average crank power over a complete revolution for the left side only was 445.3 ± 95.7 and 438.8 ± 111.5 W for session 1 and 2 respectively (Table 5.2), which gives an indicative total crank power over a complete revolution for both cranks of 891 and 878 W. MDD between-sessions for peak crank power and forces was 21 W and between 9 to 72 N respectively (Table 5.2). Peak joint angle values typically demonstrated moderate to excellent reliability, with MDD between-sessions from 1.1 to 4.4° (Table 5.2). Peak joint angular velocity between-sessions reliability was typically moderate to excellent, except for peak knee flexion and thigh extension angular velocity which had poor to good reliability (Table 5.2). MDD between-sessions for peak joint angular velocities ranged from 14 to 59°/s (Table 5.2). Peak joint moments demonstrated moderate to excellent between-sessions reliability, except for peak knee flexion moment which demonstrated poor to moderate reliability (Table 5.2). Maximum ankle and knee joint powers demonstrated good to excellent reliability between-sessions, whereas maximum hip power showed poor to good reliability (Table 5.2). MDD between-sessions for peak joint moments ranged from 2 to 26 N.m and for maximum joint powers 30 to 144 W. CV for average left crank power over a complete revolution was 3.0 ± 1.5% and 4.6 ± 1.9% for within- and between-session respectively.

Table 5.1: Within-session reliability for kinematic and kinetic variables

Variable	Units	Mean (SD) Sprint 1	Sprint 2	Sprint 3	Mean differences	ICC (3,1)	95% LB	95% UB	*P*	SEM	MDD
Power (average over a complete revolution - left only)	W	437.3 ± 109.0	434.2 ± 110.9	437.9 ± 113.4	-2.5	0.996	0.991	0.999	0.17	0.9	3
Pedalling rate	rpm	134.7 ± 1.3	134.6 ± 1.4	134.7 ± 1.4	-0.1	0.993	0.984	0.998	0.37	0.0	0.1
Max effective crank force	N	581.0 ± 131.0	578.1 ± 134.2	578.4 ± 128.6	-0.2	0.997	0.993	0.999	0.79	0.9	2
Max ineffective crank force	N	605.7 ± 164.9	598.3 ± 173.5	616.7 ± 166.1	-12.2	0.986	0.966	0.995	0.37	5.1	14
Min ineffective crank force	N	-208.9 ± 84.5	-206.8 ± 77.9	-214.7 ± 89.0	5.3	0.983	0.959	0.994	0.53	3.9	11
Max instantaneous crank power	W	1351.6 ± 315.7	1347.0 ± 325.1	1348.6 ± 311.8	-1.1	0.997	0.993	0.999	0.92	2.2	6
Peak ankle plantarflexion angle	°	142.6 ± 11.4	142.5 ± 11.6	142.1 ± 11.7	0.2	0.992	0.980	0.997	0.79	0.3	0.7
Peak ankle dorsiflexion angle	°	113.8 ± 5.7	113.4 ± 5.6	114.0 ± 6.3	-0.4	0.981	0.953	0.993	0.48	0.3	0.9
Peak knee extension angle	°	143.8 ± 6.5	143.1 ± 5.5	143.6 ± 5.6	-0.3	0.973	0.934	0.990	0.50	0.3	0.9
Peak knee flexion angle	°	70.2 ± 3.5	70.1 ± 3.5	70.2 ± 3.4	-0.1	0.981	0.955	0.994	0.74	0.1	0.3
Peak hip extension angle	°	68.3 ± 4.5	68.5 ± 4.7	68.4 ± 4.9	0.1	0.981	0.954	0.993	0.82	0.2	0.6
Peak hip flexion angle	°	25.3 ± 4.1	25.9± 4.4	25.5 ± 4.4	0.3	0.978	0.947	0.992	0.32	0.2	0.7
Peak ankle plantarflexion angular velocity	°/s	248.3 ± 62.0	257.0 ± 78.3	240.5 ± 66.5	11.0	0.930	0.833	0.976	0.35	9.6	27
Peak ankle dorsiflexion angular velocity	°/s	-270.0 ± 105.4	-278.5 ± 116.0	-266.6 ± 104.6	-7.9	0.982	0.957	0.994	0.45	4.7	13
Peak knee extension angular velocity	°/s	481.8 ± 36.9	476.4 ± 33.4	480.3 ± 34.0	-2.6	0.969	0.925	0.989	0.38	2.8	8
Peak knee flexion angular velocity	°/s	-518.8 ± 49.1	-511.6 ± 40.8	-514.0± 49.1	1.6	0.931	0.833	0.976	0.64	5.9	16
Peak hip extension angular velocity	°/s	274.4 ± 19.4	274.8 ± 24.4	275.5 ± 20.2	-0.4	0.967	0.920	0.989	0.91	1.8	5
Peak hip flexion angular velocity	°/s	-275.6 ± 37.1	-271.9 ± 32.0	-275.4 ± 37.1	2.3	0.982	0.956	0.994	0.42	1.6	4
Peak ankle plantarflexion moment	N.m	81.1 ± 20.8	81.3 ± 21.4	82.0 ± 19.3	-0.5	0.986	0.965	0.995	0.86	0.6	2
Peak ankle dorsiflexion moment	N.m	-12.3 ± 5.8	-12.3 ± 6.5	-12.5 ± 6.2	0.2	0.973	0.933	0.990	0.90	0.3	1

Table 5.1 (continued)

Variable	Units	Mean (SD) Sprint 1	Sprint 2	Sprint 3	Mean differences	ICC (3,1)	95% LB	95% UB	*P*	SEM	MDD
Peak knee extension moment	N.m	82.9 ± 32.9	82.4 ± 33.6	84.0 ± 35.1	-1.1	0.995	0.989	0.998	0.55	0.4	1
Peak knee flexion moment	N.m	-59.1 ± 16.5	-58.4 ± 15.2	-56.8 ± 15.7	-1.1	0.942	0.859	0.980	0.61	1.6	4
Peak hip extension moment	N.m	142.8 ± 32.9	141.7 ± 33.7	140.5 ± 32.9	0.8	0.972	0.931	0.990	0.52	1.2	3
Peak hip flexion moment	N.m	-42.8 ± 16.4	-41.3 ± 18.2	-42.0 ± 17.6	0.4	0.990	0.976	0.997	0.42	0.5	1
Maximum ankle power	W	261.1 ± 111.5	264.7 ± 111.9	253.9 ± 105.4	7.2	0.987	0.968	0.995	0.42	2.9	8
Maximum knee power	W	615.3 ± 250.1	620.9 ± 248.4	636.5 ± 267.0	-10.4	0.984	0.961	0.994	0.76	9.1	25
Maximum hip power	W	600.7 ± 175.1	580.5 ± 170.3	560.8 ± 125.3	13.1	0.957	0.896	0.985	0.15	13.8	38

*** indicates significant difference between sessions ($P < 0.05$), ICC(3,1) = Within-session intraclass correlation with lower (LB) and upper (UB) bound confidence intervals, SEM = standard error of measurement, MDD = minimal detectable difference**

Table 5.2: Between-sessions reliability for kinematic and kinetic variables

Variable	Units	Mean(SD) Session 1	Mean(SD) Session 2	Mean difference	***P***	ICC (3,*k*)	95% LB	95% UB	SEM	MDD
Power (average over a complete revolution - left only)	W	445.3 ± 95.7	438.8 ± 111.5	-6.5	0.429	0.979	0.938	0.993	4.3	12
Pedalling rate	rpm	134.8 ± 1.3	134.7 ± 1.4	-0.2	0.021*	0.986	0.935	0.996	0.0	0.1
Max effective crank force	N	593.3 ±126.2	579.0 ± 130.9	-14.4	0.072	0.986	0.952	0.996	3.2	9
Max ineffective crank force	N	603.5 ± 172.1	605.3 ± 165.4	1.8	0.944	0.923	0.756	0.975	25.9	72
Min ineffective crank force	N	-192.7 ± 65.2	-207.3 ± 82.3	-14.7	0.136	0.937	0.805	0.980	8.7	24
Max instantaneous crank power	W	1387.2 ± 309.2	1348.4 ± 316.5	-38.7	0.043*	0.986	0.946	0.996	7.7	21
Peak ankle plantarflexion angle	°	141.7 ± 11.3	142.3 ± 11.5	0.6	0.446	0.983	0.948	0.994	0.4	1.1
Peak ankle dorsiflexion angle	°	113.1 ± 5.0	113.8 ± 5.8	0.7	0.281	0.955	0.863	0.985	0.5	1.3
Peak knee extension angle	°	142.7 ± 6.4	143.5 ± 5.7	0.8	0.489	0.864	0.580	0.956	1.6	4.4
Peak knee flexion angle	°	70.0 ± 3.6	70.2 ± 3.4	0.2	0.715	0.857	0.550	0.954	1.0	2.6
Peak thigh extension angle	°	68.1 ± 5.0	68.4 ± 4.6	0.3	0.720	0.893	0.665	0.966	1.0	2.8
Peak thigh flexion angle	°	26.1 ± 4.3	25.6 ± 4.2	-0.5	0.447	0.916	0.746	0.973	0.7	1.9
Peak ankle plantarflexion angular velocity	°/s	236.6 ± 65.7	247.1 ± 65.0	10.4	0.441	0.839	0.509	0.948	19.7	55
Peak ankle dorsiflexion angular velocity	°/s	-262.0 ± 91.2	-268.5 ± 107.2	-6.6	0.561	0.957	0.868	0.986	8.6	24
Peak knee extension angular velocity	°/s	472.8 ± 43.2	479.1 ± 33.8	6.3	0.434	0.838	0.504	0.948	11.8	33
Peak knee flexion angular velocity	°/s	-507.5 ± 57.6	-513.3 ± 43.6	-5.8	0.635	0.772	0.279	0.927	21.4	59
Peak thigh extension angular velocity	°/s	265.6 ± 29.1	273.8 ± 21.9	8.2	0.141	0.814	0.447	0.939	8.5	24
Peak thigh flexion angular velocity	°/s	-277.6 ± 30.7	-273.4 ± 35.1	4.2	0.390	0.924	0.769	0.975	4.9	14
Peak ankle plantarflexion moment	N.m	78.6 ± 18.6	81.4 ± 20.2	2.8	0.372	0.910	0.729	0.971	3.4	9
Peak ankle dorsiflexion moment	N.m	-14.0 ± 7.0	-12.3 ± 6.0	1.8	0.049*	0.928	0.743	0.978	0.8	2
Peak knee extension moment	N.m	90.0 ± 34.5	82.9 ± 33.5	-7.1	0.028*	0.965	0.852	0.990	2.0	6
Peak knee flexion moment	N.m	-50.7 ± 20.9	-57.7 ± 15.0	-7.0	0.151	0.697	0.127	0.900	9.4	26
Peak hip extension moment	N.m	132.3 ± 30.7	140.4 ± 32.8	8.1	0.086	0.919	0.737	0.974	4.6	13

Table 5.2 (continued)

Variable	Units	Mean(SD)		Mean difference	*P*	ICC	95%	95%	SEM	MDD
		Session 1	Session 2			(3,*k*)	LB	UB		
Maximum ankle power	W	259.6 ± 111.7	258.5 ± 107.8	-1.1	0.937	0.951	0.846	0.984	10.9	30
Maximum knee power	W	659.6 ± 321.7	620.4 ± 253.6	-39.2	0.160	0.968	0.901	0.990	17.6	49
Maximum hip power	W	519.8 ± 186.3	578.1 ± 153.0	58.3	0.104	0.826	0.474	0.944	52.1	144

*** indicates significant difference between sessions (*P* < 0.05), ICC(3,k) = Between-sessions intraclass correlation with lower (LB) and upper (UB) bound confidence intervals, SEM = standard error of measurement, MDD = minimal detectable difference**

Time series variables

Crank power demonstrated excellent within- and between-session reliability, CMC ≥ 0.995 (Figure 5.1). Crank power was significantly different ($P < 0.05$) between sessions one and two, between crank angles 340 to 6° (7.2% of crank cycle) (Figure 5.1). The ineffective crank force was slightly less repeatable (CMC ≥ 0.988) than effective crank force (CMC ≥ 0.995) within- and between-session, although it still demonstrated excellent reliability (Figure 5.2). The crank forces were significantly different ($P <$ 0.05) between sessions one and two, between crank angles 191 to 199° (2.2% of crank cycle), and 347 and 1° (3.9% of crank cycle) (Figure 5.2).

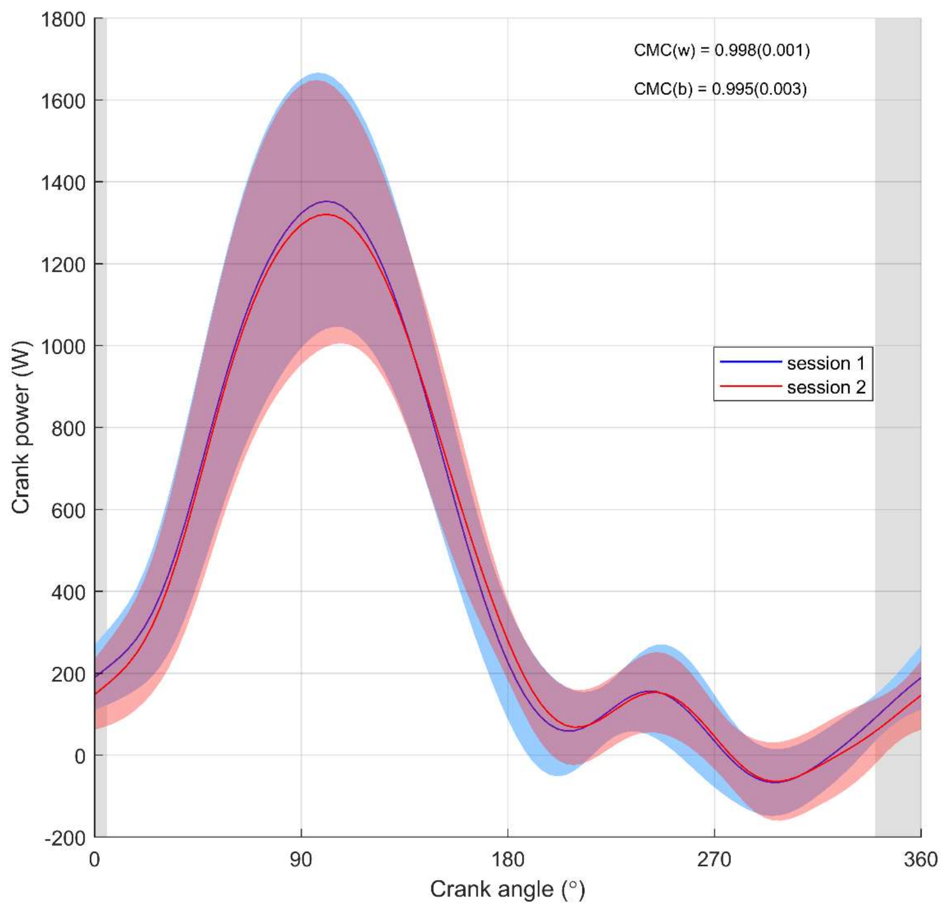

Figure 5.1: Crank power: group means for session one and two.
Areas of the graph shaded grey where the SPM is significant. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b).

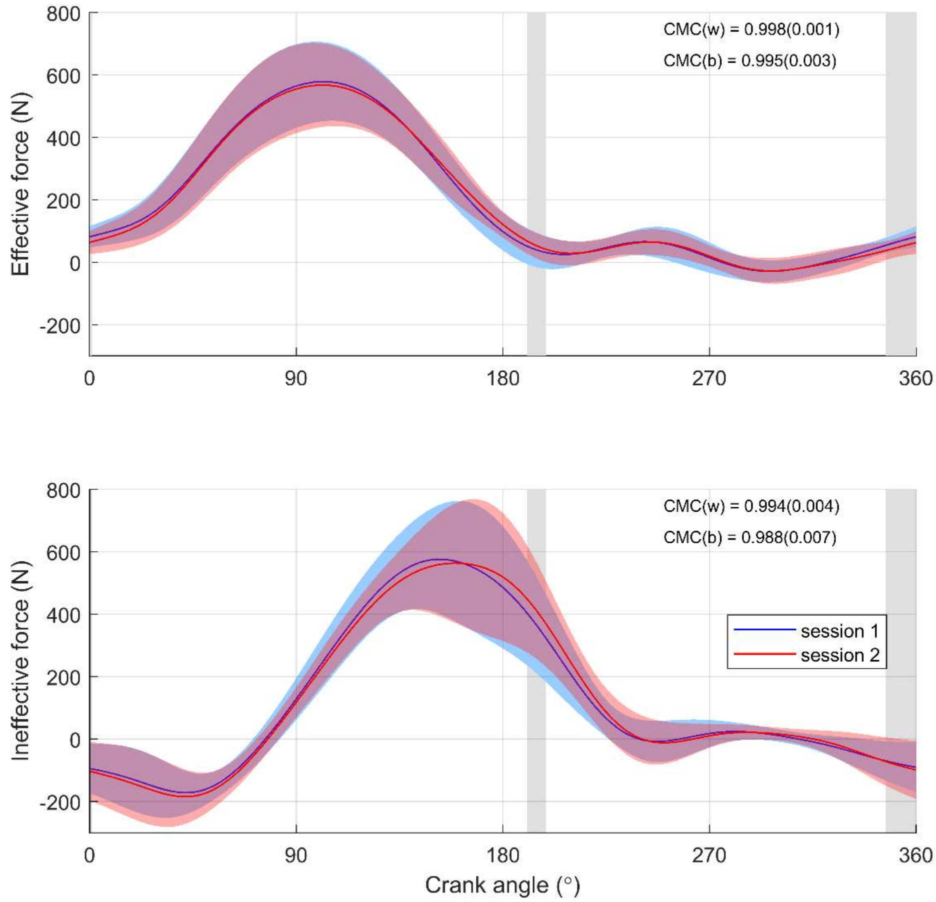

Figure 5.2: Crank forces: group means for session one and two.
Areas of the graph shaded grey where the SPM is significant. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b).

Joint angles and angular velocities demonstrated excellent within- and between-session reliability (CMC $\geq$ 0.950) (Figure 5.3). Ankle joint angles and angular velocities were less repeatable than those at the knee and hip joints. Ankle joint angular velocity was significantly different ($P < 0.05$) between sessions one and two, between crank angles 152 to 170° (5.0% of crank cycle) (Figure 5.3).

Joint moments and powers demonstrated excellent within- and between-session reliability (CMC $\geq$ 0.928) (Figure 5.3). Hip joint moments and powers were less repeatable than those at the knee and ankle joints. Ankle joint moment was significantly different ($P < 0.05$) between sessions one and two, between crank angles 340 to 6°

(7.2% of crank cycle) (Figure 5.3). Hip joint power was significantly different ($P < 0.05$) between session one and two between crank angles 340 to 2° (6.1% of crank cycle) (Figure 5.3).

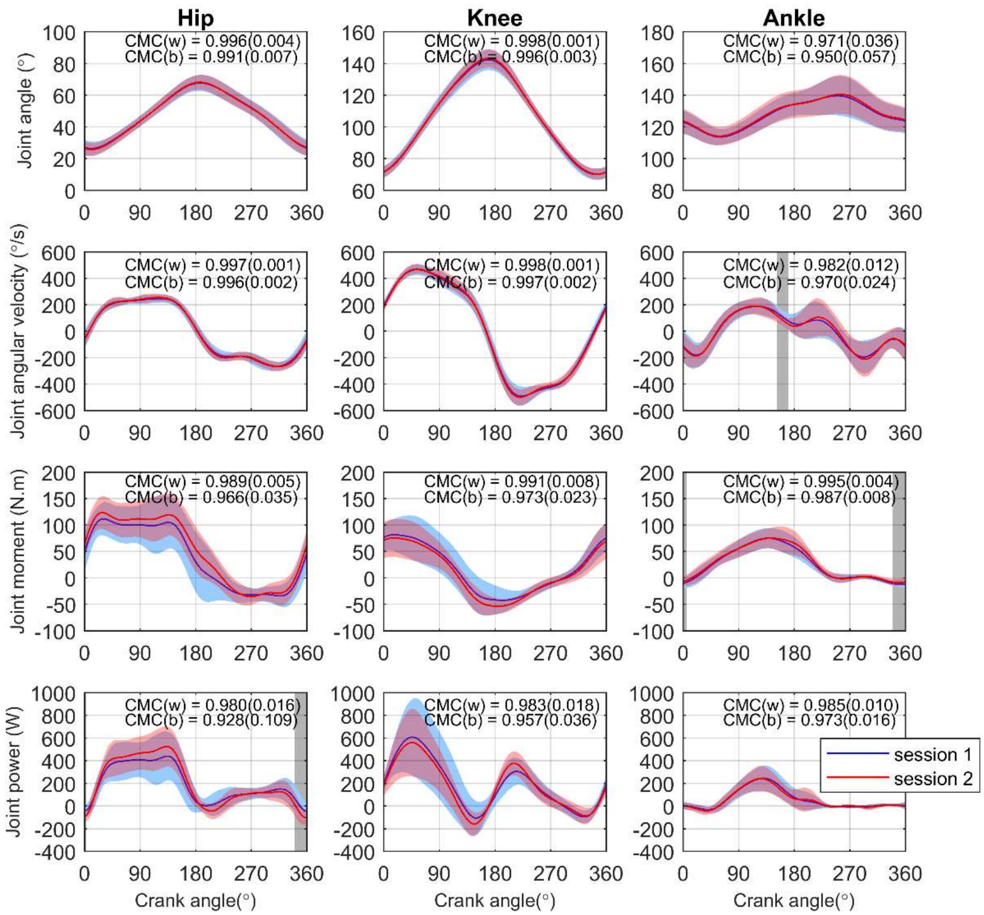

Figure 5.3: Joint angles, angular velocities, moments and powers: group means for session one and two.

Areas of the graph shaded grey where the SPM is significant. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b).

For ease of presenting the data the thigh angle and angular velocity are presented as hip angle and angular velocity

EMG linear envelope normalised to the mean value in the signal demonstrated high within- and between-session reliability (Figure 5.4). CMC values for EMG linear envelopes ranged between 0.972 to 0.985, and 0.960 to 0.981, for within- and between-session respectively. The TA, BF and ST muscles demonstrated the lowest reliability for EMG activity, and the VL and VM muscles the highest reliability (Figure 5.4). The cross-correlation coefficient (*R*) which compares timing of EMG linear envelopes between-sessions ranged from 0.976 to 0.990 (Figure 5.4).

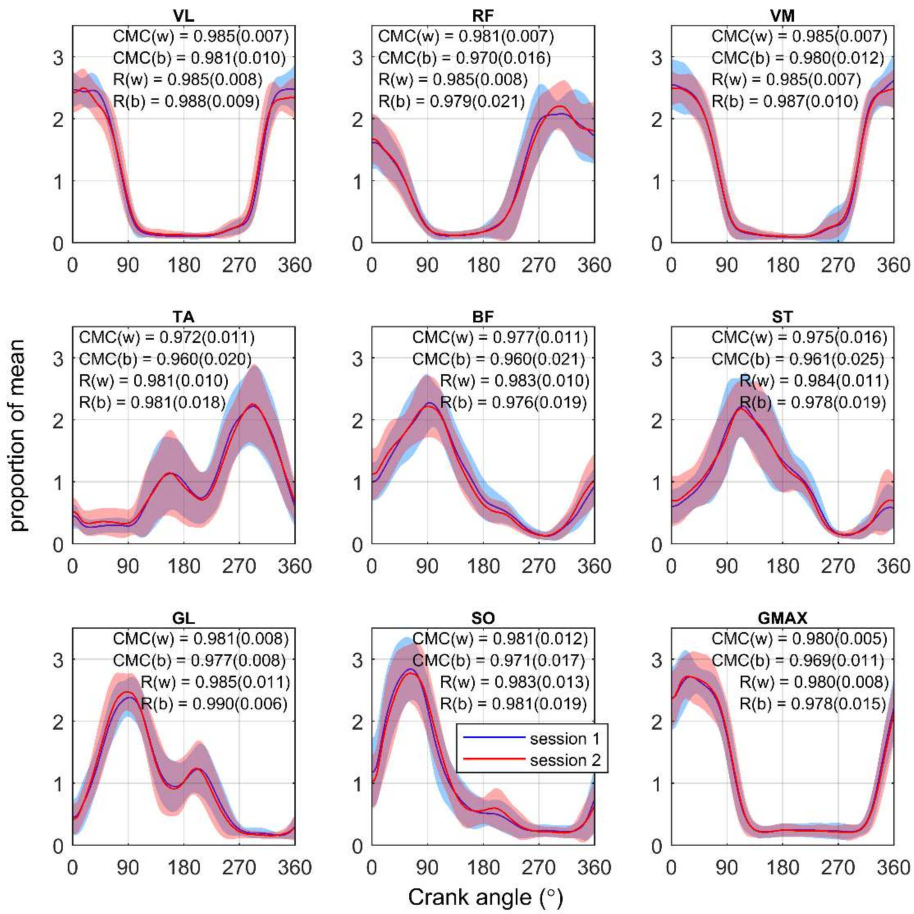

Figure 5.4: EMG linear envelopes (normalised to mean value in signal) for each muscle: group means for session one and two.
VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF = biceps femoris, ST = semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b). Mean and standard deviation of cross-correlation coefficient (R) within-session (w) and between-sessions (b).

5.4 Discussion

The purpose of this study was to quantify the test-retest reliability of kinematic, kinetic, and EMG activation variables measured during short-term maximal sprint cycling. The main findings were that between-sessions test-retest reliability level was typically moderate to excellent for the biomechanical variables that describe maximal cycling, and furthermore that within-session reliability was better than between-sessions reliability. However, some variables, such as peak knee flexion moment and maximum hip joint power demonstrated lower reliability, indicating that care needs to be taken when using these variables to evaluate changes in maximal cycling biomechanics.

Within- and between-session values of CMC for joint angles and angular velocities demonstrated high reliability (CMC > 0.950) (Figure 5.3). This agrees with the findings in the clinical gait analysis literature where high CMC's are reported for sagittal plane kinematics (CMC's typically greater than 0.920) (McGinley et al., 2009). More specifically, the ankle joint kinematics (angle and angular velocity) were found to be less repeatable than knee and thigh joint kinematics. Again, these results are consistent with gait studies where lower values of CMC are observed for ankle dorsiflexion/plantarflexion compared to knee and hip flexion/extension (Kadaba et al., 1989; McGinley et al., 2009).

The source of the lower reliability in the ankle joint kinematics data is not clear. It seems unlikely to be a measurement error, given that anatomical landmark marker placement errors for the lower limb are greatest at the hip, rather than the ankle joint (intra-examiner precision for the greater trochanter marker is 12.2 mm along the long axis of the femur, and 11.1 mm in the anterior-posterior direction, compared to lateral malleolus - 2.6 mm along the long axis fibula, 2.4 mm anterior-posterior direction) (Della Croce et al., 1999; Della Croce et al., 2005). Furthermore, the soft tissue artefact (STA) of the lower limb markers in cycling is also larger for the hip rather than the ankle joint (greater trochanter marker displacement at 30 rpm submaximal cycling, 37.3 mm anterior-posterior and 10.3 mm proximal-distal, compared to the lateral malleolus 15.8 mm anterior-posterior and 8.6 mm proximal-distal) (Li, et al., 2017). By comparison there are potential biological explanations for the lower reliability of the ankle joint kinematics. Martin and Nichols, for example, demonstrated that the ankle has a different role to the knee and hip joints in maximal cycling and acts to transfer - instead of maximise power (Martin & Nichols, 2018). More specifically, the ankle works in synergy with the hip joint to transfer power produced by the muscles surrounding the hip joint to the crank (Fregly & Zajac, 1996). The results support this notion by suggesting that cyclists may regulate their ankle angle as part of this hip-ankle synergy, in order to maintain a stable effective crank force. A specially designed experiment would be required to test this hypothesis.

In terms of joint kinetics, joint moments and powers demonstrated lower reliability at more proximal compared to distal joints – with the lowest values of CMC for the hip joint (Figure 5.3). This observation may be due to the STA and skin marker misplacement errors being largest at the hip joint, as discussed above (Della Croce et al., 1999; Li, et al., 2017). It may also be attributed to the fact that measurement errors in general (STA, marker misplacement, force pedal measurement precision) will propagate through the inverse dynamics calculations (Myers, Laz, Shelburne, & Davidson, 2015). In either scenario, this indicates that the observed differences in proximal to distal joint reliability are likely to be due to measurement error, rather than biological variability.

The peak knee flexion moment showed poor to moderate between-sessions reliability, with the largest MDD of all joint moments (26 N.m). Error due to knee marker misplacement is dependent on knee flexion angle, with previous studies demonstrating that the greater the knee flexion, the larger error in the joint angle (Della Croce et al., 1999). Marker displacement could, therefore, explain the poor reliability of the peak knee flexion angular velocity and moment data. Further work is required, using more detailed marker sets and models of STA, to reduce the influence of STA and skin marker misplacement on the calculated kinematics and kinetic variables, which may improve the reliability of the calculated knee flexion and hip joint variables.

Average left crank power output over a complete revolution was highly reliable both within- and between-session, supporting the findings of Martin and colleagues that trained cyclists are able to reproduce reliable maximal crank power within one testing session (Martin et al., 2000). Effective crank force exhibited similar reliability to crank power, whereas ineffective crank force demonstrated lower within- and between-session reliability which was associated with the large intra-participant variability in ineffective crank force between crank angles of 140° and 210° (Figure 5.2). It is unlikely that force pedals' measurement precision would provide an explanation for these observed differences in reliability between the effective and ineffective crank forces, given that the measurement precision values are the same for all components of force for the instrumented pedals used in this study (combined error - linearity and hysteresis 1% measuring range (MR) and crosstalk between the components (<1.5% MR) (Sensix, Poitiers, France)). Therefore, it seems probable that the reliability difference between effective and ineffective force may have a biological basis, a notion which can be expanded upon using the EMG results.

EMG linear envelopes generally demonstrated excellent reliability, CMC(b) > 0.960 (Figure 5.4). However, the hamstrings (BF/ST) and the TA muscles demonstrated the lowest reliability for EMG activity. Wren and colleagues suggested the lower reliability of the hamstrings may be caused by measurement error reflecting the increased sensitivity of these muscles to electrode placement owing to muscle length and overlying fat mass (Wren et al., 2006). The lower reliability of EMG activity in the hamstring muscles (BF/ST) may also have a biological basis however, given that the findings are consistent with other studies who suggest that this is related to their bi-articular function (Ryan & Gregor, 1992). Van Ingen Schenau and colleagues for example demonstrated that the bi-articular muscles are important for controlling the direction of the external force on the pedal (van Ingen Schenau et al., 1992). They identified that the paradoxical coactivation of the mono-articular agonists (vastii) with bi-articular antagonists (hamstrings) emerges so that the bi-articular muscles can help control the desired direction of the force applied to the pedal by adjusting the relative distribution of net moments over the joints (van Ingen Schenau et al., 1992).

On a mechanical basis, the goal of maximal cycling is to maximise the effective crank force as this maximises the propulsive power and thus the speed of the bicycle. Taking the crank force and EMG data together therefore, the results allows speculation that cyclists may regulate bi-articular muscles activation to control the direction of the pedal force, with the aim of maximising effective crank force and maintaining a stable outcome at the expense of the ineffective force which does not directly affect the task outcome. The bi-articular muscles (BF, ST and GL) are active in the region of the crank cycle where the ineffective crank is more variable, which could explain the biological mechanism underlying this finding. This principle has been observed in walking (Kadaba et al., 1989; Giakas & Baltzopoulos, 1997) and running (Kinoshita, Bates, & DeVita, 1985), where the propulsion and braking ground reaction forces (anterior-posterior and vertical direction) have been shown to have lower between-stride variability than the medio-lateral force. However, further, purposefully designed, experiments are required to confirm or refute these speculations.

SPM indicated a significant between-session difference for small regions of the crank cycle, for crank power, crank forces, ankle angular velocity and moment, and hip power. These differences are unlikely to be meaningful as these are less than 7.2% of the crank cycle, and typically occur in regions of low magnitude in these variables.

The experimental protocol could have introduced some variability to the kinematics, as although the participants were instructed to remain seated during the sprints on the ergometer, they tended to hover slightly over the saddle (potentially with the aim to increase crank power), which increases pelvis movement. Also, the ergometer was set-up to match each participant's track bicycle. Therefore, saddle height was not standardised to percentage of inside leg length, which is often recommended (de Vey Mestdagh, 1998). Some of the participants had a relatively low saddle height compared to their leg length, which resulted in relatively large pelvis obliquity (rocking) and transverse rotation when they sprinted. This strategy may have introduced more within- and between-trial variability, particularly at the hip joint.

5.5 Conclusion

Typically, the biomechanical variables that describe maximal cycling are reliable. However, some variables have lower reliability, indicating that care needs to be taken when using these to evaluate changes in maximal cycling biomechanics. The results allow us to speculate that biological variability is the source of the lower reliability of the ineffective crank force, ankle kinematics and hamstring muscles activation while measurement error is the source of the lower reliability in hip and knee joint kinetics. Further research using purposefully designed experiments is required to confirm or refute these speculations. These reliability data can be used to help understand the practical relevance of a longitudinal intervention on athletes' maximal cycling performance, such as the strength training intervention reported in Chapter 6.

6 The effect of strength training on intermuscular coordination in maximal cycling

6.1 Introduction

Coaches of sports requiring maximal effort over a short period of time (< 60 s), such as sprint running, track sprint cycling, sprint kayaking (200 m), and bicycle motocross (BMX) often consider strength training (repetitive muscle actions against high loads) to be a fundamental aspect of an athlete's training programme (Debraux & Bertucci, 2011; Delecluse, 1997; García-Pallarés & Izquierdo, 2011; Parsons, 2010). Accordingly, sprint athletes routinely undertake gym-based strength training in addition to sport-specific training with the aim to increase muscle size and strength (Burnie et al., 2018; Delecluse, 1997; García-Pallarés & Izquierdo, 2011; Parsons, 2010).

Although coaches from these sprint sports - interviewed in Chapter 3 - viewed strength training as a fundamental part of sprint athletes' training programmes, they do not necessarily believe there is a direct correlation between improvements in 'gym strength' (e.g. assessed by the amount of mass that can be lifted in a non-specific strength exercise with gym equipment) and sports performance (Burnie et al., 2018) (Chapter 3, Figure 3.2). This experiential observation is supported by empirical evidence which shows that the transfer of strength training to sports performance varies; generally, there is positive transfer to sports performance (i.e. strength training improves performance). However, sometimes there is no effect or even a negative transfer (i.e. strength training is detrimental to performance) (Blazevich & Jenkins, 2002; Carroll et al., 2001; Moir et al., 2007; Young, 2006).

Intermuscular coordination is a mechanism which might explain the varied transfer of strength training to sports performance in two ways. Firstly, muscle recruitment patterns associated with a strength training task could retard sports performance when expressed during the sport movement (Carroll et al., 2001). For example, the strength training programme of a sprint cyclist commonly consists of non-specific strength training exercises, such as squats, deadlifts and leg presses (Parsons, 2010). These exercises, however, have very different intermuscular coordination patterns compared to the act of pedalling (Koninckx et al., 2010). For instance when executing a squat a stable knee joint is very important in order to decelerate the load at the end of the range of motion (Cormie et al., 2011b). To achieve this aim there is significant co-contraction of the hamstrings and quadriceps (Gullett et al., 2009; Slater, & Hart, 2017). This intermuscular coordination pattern is different to coordination patterns required for cycling where co-contraction between the quadriceps and the hamstrings is required to provide fine control of the direction of force applied to the pedal, rather than to stabilise the knee joint (Dorel et al., 2012; van Ingen Schenau et al., 1992). In this way, extensive non-specific strength training could actually impair pedalling coordination, such that cycling performance is reduced.

Secondly, improvements in sports performance might only occur if increases in muscle strength are accompanied by concomitant adaptations in intermuscular coordination. This notion that the coordination patterns need to change in response to changing constraints (e.g. muscle size, strength and fatigue) is captured by key ideas in ecological dynamics and Newell's model of constraints (Newell, 1986) that propose the coordination patterns emerge from the constraints imposed on the system (Newell, 1986). Bobbert and Van Soest demonstrated using a musculoskeletal simulation that an increase in leg strength must be accompanied by a change in intermuscular coordination in order for vertical jump height to increase (Bobbert & Van Soest, 1994), providing further support for this notion. They performed a dynamic optimisation analysis to identify the intermuscular coordination pattern that maximised vertical jump height for their musculoskeletal model (Bobbert & Van Soest, 1994).

However, even without a musculoskeletal simulation it should be possible to assess how strength training affects intermuscular coordination during maximal cycling. Several key mechanical features that represent functional maximal cycling coordination patterns have been identified from previous research. First, the hip and ankle joint working in synergy during the downstroke, to enable the ankle to transfer the power produced by the hip extensor muscles to the crank (Fregly & Zajac, 1996). Second, adjustment of the bi-articular hamstring muscles activation to control the direction of the external force on the pedal. For example, if greater muscle force is produced by the vastii, the activation of the hamstring muscles will have to adapt to control the direction of the resultant force applied to the crank, so that it is directed more effectively (tangentially) (van Ingen Schenau et al., 1992). Third, at higher pedalling rates, muscle activation-deactivation dynamics are a major constraint on power production (McDaniel et al., 2014; Neptune & Kautz, 2001; van Soest & Casius, 2000), as there is insufficient time to fully activate the muscles to achieve maximal force production during a crank cycle. Therefore, coordination strategies that can maximise the muscle force production in the main power producing phase of the downstroke are beneficial. One of these strategies is to time the activation of the powerful hip and knee extensor muscles (GMAX/VL/VM), so they activated as maximally as possible at a crank angle of around 90° from top dead centre (TDC) – the location of peak crank power (Dorel et al., 2012; McDaniel et al., 2014). Fourth, at optimal pedalling rates and below for maximum crank power production, cyclists actively pull up during the upstroke generating positive crank power in maximal cycling (Dorel et al., 2009; Dorel et al., 2010). This is in comparison to submaximal cycling, where the upstroke may be more passive (Dorel et al., 2009; Dorel et al., 2010). Dorel and colleagues found positive relationships between upstroke power and average crank power over a complete revolution, and between index of mechanical effectiveness (IE - ratio of effective crank force to the total crank force) and power output during the upstroke in maximal cycling (Dorel, 2018b; Dorel et al., 2010). These key mechanical features will be used to assess the effect of strength on intermuscular coordination in maximal cycling.

Considering the evidence of possible mechanisms for how strength training might influence coordination, the aim of this study was to investigate the effects of a gym-based strength training intervention on short-term maximal cycling biomechanics and intermuscular coordination patterns. It was hypothesised that strength training will change maximal cycling coordination patterns – magnitude and timing of joint moments and powers, EMG activation patterns and the key mechanical features associated with maximal cycling.

6.2 Methods

Participants

Twelve track sprint cyclists participated in the study. Participants regularly competed at track cycling competitions at either under 23 international level (5), Master's international and national level (4), or Junior national level (3). Although the participants were varied in their anthropometrics (4 males and 8 females, age: 24.1 ±

13.8 yr, body mass: 68.2 ± 11.1 kg, height: 1.70 ± 0.07 m,), they were similar with respect to cycling performance level (flying 200 m PB: 11.61 ± 0.90 s). Participants were provided with study details and gave written informed consent. The study was approved by the Sheffield Hallam University Faculty of Health and Wellbeing Research Ethics Sub-Committee.

Experimental protocol

An isokinetic ergometer was set up to replicate each participants track bicycle position, - all participants used a crank length of 165 mm on their track bicycles. Riders undertook their typical warm-up on the ergometer at self-selected pedalling rate and resistance for at least 10 minutes, followed by one 4 s familiarisation sprint at 135 rpm. Riders then conducted 3 x 4 s seated sprints at a pedalling rate of 135 rpm, interspersed with 2 x 4 s seated sprints at a pedalling rate of 60 rpm on the isokinetic ergometer with 4 minutes recovery between efforts. The 60 rpm pedalling rate was chosen as it has been used as a measure of cycling specific strength (Barratt, 2014). At this pedalling rate the extension phase lasts for approximately 500 ms which is deemed an acceptable time to allow peak muscular force to be developed (Aagaard et al., 2002; Hannah, Minshull, Buckthorpe, & Folland, 2012; Tillin, Jimenez-Reyes, Pain, & Folland, 2010). Conversely, a pedalling rate of 135 rpm was chosen as this is representative of the pedalling rate during the flying 200 m event in track cycling and within an optimal pedalling rate range for track sprint cyclists (Dorel et al., 2005). All participants had previous experience of undertaking gym-based strength training, including traditional resistance training exercises, with many of the participants undertaking lighter strength training volume in the period immediately prior to the start of the intervention, owing to the proximity of the competition season or end of season training break. The participants then undertook a training programme for 11.6 ± 1.4 weeks of two gym-based strength training sessions per week consisting of traditional resistance training exercises: squats, leg press and deadlift. The weight lifted, number of repetitions and sets of each exercise were prescribed by each participant’s strength and conditioning coach, along with any other supplementary exercises. The overall content of the training programmes was prescribed by the participants’ cycling coaches and typically included at least two track cycling sessions and one road ride of about 60 to 90 minutes in length a week. Refer to Appendix 9.10, for details of the participants training programmes and gym exercises in the intervention period. Following the training period the participants undertook an identical testing session to the pre-test. Participants were asked to undertake similar training in the preceding 24 hours before both testing sessions.

Isokinetic ergometer

The same ergometer was used as described in section 4.2. The ergometer controlled pedalling rate to within 1 rpm for each session (mean pedalling rate: session 1, 135.1 ± 1.2 rpm, session 2, 135.2 ± 1.1 rpm). This set-up enabled participants to start their bouts at the target pedalling rate, rather than expending energy in accelerating the flywheel. The ergometer was fitted with Sensix force pedals (Model ICS4, Sensix, Poitiers, France) and a crank encoder (Model LM13, RLS, Komenda, Slovenia), sampling data at 200 Hz. Normal and tangential pedal forces were resolved using the crank and pedal angles into the effective (F_E) (propulsive) and ineffective (F_I) (applied along the crank) crank forces, and total resultant crank force (F_T) (Figure 4.2**).**

Kinematic and kinetic data acquisition

Two-dimensional kinematic data of each participant's left side were measured using the same method as described section 4.2.

EMG data acquisition

EMG signals were recorded continuously from nine muscles of the left leg: vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA), long head of biceps femoris (BF), semitendinosus (ST), lateralis gastrocnemius (GL), soleus (SO), and gluteus maximus (GMAX) with Delsys Trigno wireless surface EMG sensors (Delsys Inc, Boston, MA). The skin at the electrode placement sites was prepared by shaving the area then cleaning it with an alcohol wipe. The EMG sensors were then placed in the centre of the muscle belly - with the bar electrodes perpendicular to the muscle fibre orientation and secured using wraps to reduce motion artefacts during pedalling. The same researcher attached the EMG sensors for all sessions. A Delsys analogue sensor was connected to a reed switch which was fitted to the ergometer so it omitted a pulse when the left crank arm passed top dead centre (TDC). The EMG system was operated and recorded in EMGworks Acquisition software (Delsys Inc, Boston, MA), sampling data at 1926 Hz.

Anthropometrics

The participants' left thigh volumes were measured using a 3D depth camera scanning system described in previous research (Bullas et al., 2016; Kordi et al., 2018). In brief, the system consisted of four consumer depth cameras (Microsoft Kinect version 1, Microsoft Corporation, Redmond, USA) mounted vertically at each corner of a 1.41 m by 1.41 m aluminium frame (Bosch, Rexworth, AG). The 3D depth cameras were operated and calibrated using KinanthroScan software (KinanthroScan v1.0, CSER, SHU, UK), which was also used to record the scans and to process and analyse the data to calculate thigh volume. Thigh volume was calculated by digitising the anatomical landmarks at the superior and inferior boundaries of the thigh segment. The superior boundary of the thigh was defined as 1cm distal to the gluteal fold and the inferior boundary at the midpoint of the superior border of the patella in accordance with the standards of the International Society for Advanced Kinanthropometry (Stewart & Sutton, 2012). The thigh volume was calculated using an implementation of Green's equations (Crisco & McGovern, 1997) – full details of the procedure to calculate thigh volume is described in (Bullas et al., 2016; Kordi et al., 2018).

Gym strength

A back squat exercise was used to evaluate the effectiveness of strength training programmes as recommended by (Parsons, 2010). Participants reported details of the weight lifted, repetitions and sets for the squat they performed in their gym session closest to the laboratory testing sessions. To allow comparison of the 'gym strength' between participants and sessions, squat predicted one repetition maximum (1RM) (how much weight an individual can lift for one repetition) was calculated using the (Brzycki, 1993) formula:

$$Predicted\ 1RM = \frac{Weigh\ \ lifted}{1.0278 - 0.0278X} \tag{3}$$

Where X= the number of repetitions of the exercise performed

Subjective measures

At both testing sessions the participants were asked to rate their energy level (low (1) to high (6)), sleep quality (disrupted (1) to good (5)) and muscle soreness (no soreness (1) to high (6)). This was used to assess participants' perceived fatigue levels, as coaches believe that a heavy strength training period induces fatigue which affects sports performance in the immediate period following strength training (Burnie et al., 2018) (Chapter 3). The participants were also asked to rate how they thought they pedalled on a Likert scale (poor (0) to excellent (10)) to provide a self-reported pedalling score (Porta & Last, 2018).

Data processing

All kinetic and kinematic data were filtered using a Butterworth fourth order (zero lag) low pass filter using a cut off frequency of 10 Hz and 14 Hz for the 60 rpm and 135 rpm sprints respectively, which were selected using residual analysis (Winter, 2009). The same cut off frequency was chosen for the kinematic and kinetic data as recommended by Bezodis and colleagues to avoid data processing artefacts in the calculated joint moments (Bezodis et al., 2013). Instantaneous left crank power was calculated from the product of the left crank torque and the crank angular velocity. The average left crank power was calculated by averaging the instantaneous left crank power over a complete pedal revolution. Joint angles were calculated using the convention shown in (Figure 4.2). Joint moments were calculated via inverse dynamics (Elftman, 1939), using pedal forces, limb kinematics, and body segment parameters (de Leva, 1996). As discussed in section 4.2 the body segment parameters (segmental mass, centres of mass and principal moments of inertia) were estimated via the tables of de Leva (1996). In this study there were only small changes in participant mass (1.1 ± 1.8 kg) and thigh volume (176 ± 302 cm^3) between pre and post intervention, suggesting little change in the distribution of the leg mass between-sessions. Also, the actual mass at each testing session was always used for the calculation of the body segment parameters for input into the inverse dynamics calculations. Therefore, the tables produced by de Leva (1996) were deemed suitable to calculate the body segment parameters for this study. Joint extension moments were defined as positive and joint flexion moments as negative. Joint powers at the ankle, knee and hip were determined by taking the product of the net joint moment and joint angular velocity. The power transferred across the hip joint was calculated as the dot product of hip joint reaction force and linear velocity (Martin & Brown, 2009).

Data were analysed using a custom Matlab (R2017a, MathWorks, Cambridge, UK) script. Each sprint lasted for 4 s so provided four and six complete crank revolutions at 60 rpm and 135 rpm respectively. Crank forces and powers, joint angles, angular velocities, moments and powers were resampled to 100 data points around the crank cycle and then mean value at each time point was calculated to obtain a single ensemble-averaged time series for each trial. Owing to technical problems for two participants, their session average for the sprints at 135 rpm were calculated from two instead of three sprints. Also, the session average for the 60 rpm sprints was calculated from only one sprint for five of the participants.

Relative distribution of joint powers has been used as a measure of coordination in cycling (Barratt, 2014; Korff & Jensen, 2007; Korff, Hunter, & Martin, 2009). To calculate relative joint powers, the joint powers were averaged over the extension and flexion phases as defined by the joint angular velocities (positive velocity for extension and negative velocity for flexion) and then normalised to average left crank power over a complete revolution.

The raw EMG signals for the 135 rpm sprint efforts were high pass filtered (Butterworth second order, cut off frequency 30 Hz) to diminish motion artefacts (De Luca et al., 2010), root mean squared (RMS, 25 ms window) and then low pass filtered (Butterworth second order, cut off frequency 24 Hz) (Brochner Nielsen et al., 2018). To synchronise the EMG data with the kinetic and kinematic data the TDC locations obtained from the analogue sensor were matched to the corresponding TDCs measured by the crank encoder. The data were then interpolated to 100 data points around the crank cycle (using spline interpolation method) and then averaged over six crank revolutions to create a linear envelope for each muscle. The EMG signals were normalised to the mean value in the linear envelope across the crank cycle for each muscle. Due to noisy EMG data for specific muscles for several participants, the EMG linear envelopes for these muscles were created from averaging one or two sprints instead of three.

Assessment of key mechanical features of maximal cycling

The strength of the hip-ankle synergy was quantified by the frequency of in-phase coordination pattern between the hip and ankle moments in the downstroke, which was calculated using the vector coding method (see below for details). The activation of the bi-articular hamstring muscles to control the direction of the external force applied to the pedal was assessed by comparing the IE, and the EMG activation patterns of the BF and ST muscle pre and post strength training intervention. To assess the role of the activation-deactivation dynamics of the main power producing muscles in the downstroke, the EMG activation timings of the GMAX/VL/VM muscles were compared pre and post strength training intervention. The role of the upstroke in power generation in maximal cycling was assessed by comparing the IE and average crank power produced in the upstroke sector pre and post strength training intervention.

Quantifying hip-ankle joint synergy

It has been suggested that the hip and ankle joints need to work in synergy to transfer the power produced at the hip joint to the crank (Fregly & Zajac, 1996). Therefore, to quantify hip-ankle joint coordination and the strength of the hip-ankle joint synergy a vector coding technique was used (Chang, Van Emmerik, & Hamill, 2008; Hamill, Haddad, & McDermott, 2000; Sparrow, Donovan, Van Emmerik, & Barry, 1987). Vector coding is typically applied to kinematic data to quantify inter-segment coordination from segmental angle-angle diagrams (Chang et al., 2008). The vector coding method was applied to joint moment-moment diagrams, as these were the most appropriate variables to investigate the hip-ankle synergy, as Fregly and Zajac identified that the net hip and ankle joint torques act in synergy during the downstroke (Fregly & Zajac, 1996). The coupling angle (γ_i) was calculated from the hip-ankle moment diagrams for each point on the crank cycle (the joint moment data had been interpolated to 101 equally spaced data points around the crank cycle, using the detailed method in Appendix 9.9, equations 5, 6 and 7). The coupling angle is defined as the orientation of the vector (relative to the right horizontal) between two adjacent points on the moment-moment plot (Appendix 9.9, Figure 9.15).

The coupling angle was calculated for each instant of the crank cycle for all revolutions of the sprints at 135 rpm for each participant. Since the coupling angles are directional in nature, the mean coupling angles were computed using circular statistics (Batschelet,

1981) (Appendix 9.9, equations 8, 9 and 10). This process was repeated to calculate the group mean coupling angles pre and post strength training intervention (Appendix 9.9, Figure 9.17).

The mean coupling angle for each participant was categorised into four coordination phases: in-phase, anti-phase, hip phase and ankle phase based on the system proposed by Chang and colleagues (Chang et al., 2008) (Figure 6.1).

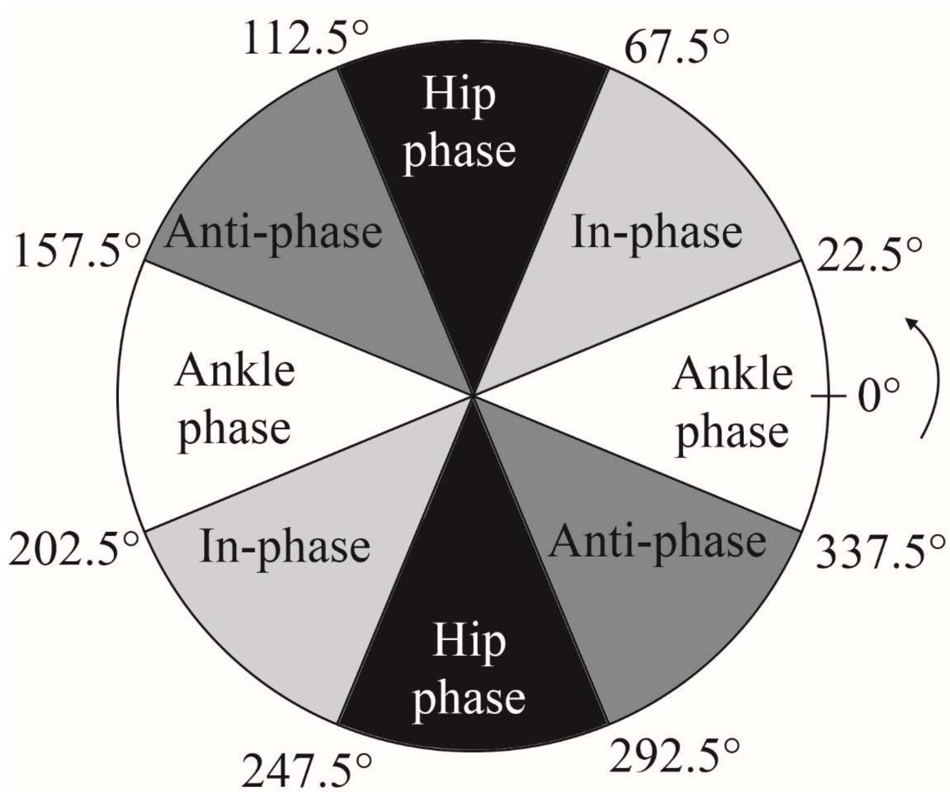

Figure 6.1: The coordination pattern classification system for the coupling angle (γ_i)
When the coupling angle values are 45° and 225° (a positive diagonal), the couple is in-phase: both the hip and ankle moments are increasing or decreasing at similar rates, i.e. the hip and ankle joints are working in synergy. Conversely, when the coupling angles are 135° and 315° (a negative diagonal), the couple is anti-phase. For example, when the hip moment is increasing whilst ankle moment is decreasing. When coupling angles are parallel to the horizontal (0° and 180°), the ankle moment is changing but not the hip moment – ankle phase. When coupling angles are parallel to the vertical (90° and 270°), the hip moment is changing but not the ankle moment – hip phase. Since the

coupling angles rarely lie precisely on these angles the unit circle was split into 45° bins as used by (Chang et al., 2008) (Figure 6.1). The frequency the mean coupling angle ($\overline{\gamma_i}$) lay within each of these coordination patterns during the downstroke (defined between crank angles of 0 to 180°) was calculated for each participant for each session.

Index of mechanical effectiveness (IE)

The overall index of mechanical effectiveness (IE) for the complete crank cycle was determined as the ratio of the linear impulse of F_E to linear integral of F_T (Dorel et al., 2010; Lafortune & Cavanagh, 1983). Mean values of the F_E, F_T, crank power, and IE were calculated for the four functional angular sectors of the crank cycle (Dorel et al., 2010; Hug et al., 2008) (Figure 6.2). The values of force and power output for the different sectors were weighted by the size of each sector relative to the entire crank cycle (i.e. 60/360 for the top, 120/360 for the downstroke).

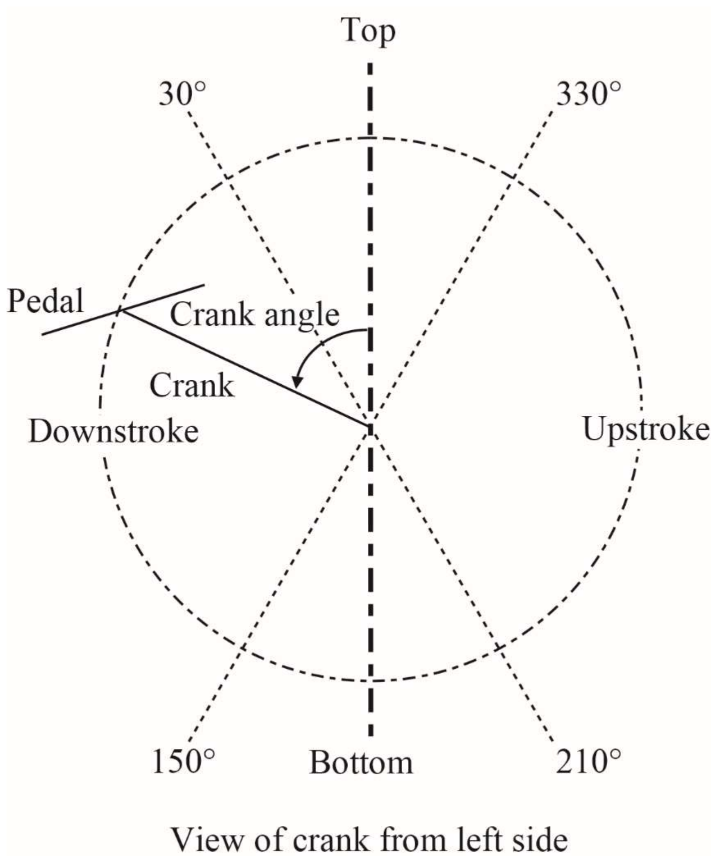

Figure 6.2: Four functional angular sectors of the crank cycle (as defined in (Dorel et al., 2010; Hug et al., 2008))

Statistical analysis

Statistical tests for discrete variables were performed using IBM SPSS Statistics Version 24 (IBM UK Ltd, Portsmouth, UK). Differences between discrete values between sessions were assessed using paired *t*-tests for the normally distributed variables and Wilcoxon matched-pairs tests for the non-parametric variables (typically the qualitative scores and coordination phase frequencies). The participants' change in squat predicted 1RM between sessions was correlated with changes in average left crank power over a complete revolution for sprints at 135 rpm between sessions using a Pearson correlation. This was to assess if the coaches' belief identified in Chapter 3, that there was not a direct correlation between changes in 'gym strength' and sports performance was evident in this study. Differences between time series data (instantaneous crank powers, crank forces, joint angles, angular velocities, moments, powers and normalised EMG linear envelopes) between sessions were assessed using statistical parametric mapping (SPM); paired *t*-tests were used for all variables except crank forces where Hotelling's paired T^2 test was used (Pataky, 2010). Crank force consists of two vector components (effective and ineffective crank force), and therefore a multivariate statistical test was required. To try and explain the increase in average left crank power over a complete revolution following the strength training intervention, the data were explored *post-hoc* by correlating the participants pre strength training relative hip and knee joint extension powers with change in average left crank power over a complete revolution between sessions for sprints at 60 rpm and 135 rpm using a Pearson correlation. The level of statistical significance was set to $P < 0.05$ for all tests. Effect size values (ES) were calculated for all discrete variables. ES for parametric variables were calculated using the following equation (Ivarsson, Andersen, Johnson, & Lindwall, 2013):

$$d = \frac{M_{post} - M_{pre}}{\sqrt{\frac{SD_{pre}^2 + SD_{post}^2}{2}}} \qquad (4)$$

Where d = ES, M_{post} and M_{pre} are the group means post and pre intervention, and SD_{pre} and SD_{post} are the groups' standard deviations.

ES were interpreted using Cohen's classification system: effect sizes between 0.2 and 0.5 were considered small, between 0.5 and 0.8 were considered moderate, and greater than 0.8 were considered large (Cohen, 1988).

6.3 Results

Discrete variables

Squat predicted 1RM increased following the strength training intervention. This increase was very close to being statistically significant (Table 6.1, $P = 0.050$, ES = 0.26). Average left crank power over a complete revolution for sprints at 135 rpm significantly increased post strength training intervention (Table 6.1, $P = 0.028$, ES = 0.29).

Table 6.1: Discrete variables pre and post strength training intervention

Variable	Units	Pre	Post	Change	*P*	Effect Size
		Mean (SD)				
Mass	kg	68.2 ± 11.0	69.2 ± 11.3	1.1 ± 1.8	0.101	0.09
Thigh volume	cm^3	6111 ± 1159	6254 ± 1293	176 ± 302	0.151	0.12
Squat predicted 1RM	kg	108.6 ± 29.5	116.2 ± 28.5	7.6 ± 11.9	0.050	0.26
Average crank power 60 rpm	W	335.2 ± 67.2	342.3 ± 53.1	7.2 ± 21.4	0.269	0.12
Average crank power 135 rpm	W	467.6 ± 88.9	494.1 ± 91.2	26.5 ± 36.2	0.028*	0.29
Energy level	1 to 6	3.7 ± 1.0	3.8 ± 1.4	0.0 ± 0.9	0.754	-0.08
Sleep quality	1 to 5	3.5 ± 1.2	3.4 ± 1.3	-0.3 ± 1.5	0.883	-0.06
Muscle soreness	1 to 6	3.1 ± 1.4	2.7 ± 1.4	-0.2 ± 1.5	0.645	-0.12
Pedalling score	0 to 10	5.9 ± 1.2	6.3 ± 1.5	0.2 ± 1.3	0.531	-0.17

- * indicates significant difference between sessions ($P < 0.05$)
- Average crank power over a complete revolution for the left crank only (Gives an indicative total power for both cranks of 935 W and 988 W for session 1 and 2 respectively, for sprints at 135 rpm, and 670 W and 685 W for session 1 and 2 respectively, for sprints at 60 rpm)

Table 6.2: Peak joint moments produced at 60 rpm pre and post strength training intervention

Peak joint moment (N.m)	Mean (SD) Pre	Post	Change	*P*	Effect size
Ankle plantarflexion	119.3 ± 22.3	123.5 ± 29.4	4.1 ± 14.3	0.331	0.16
Ankle dorsiflexion	-20.3 ± 12.7	-20.5 ± 14.4	-0.2 ± 5.4	0.869	-0.02
Knee extension	124.3 ± 78.5	107.8 ± 31.1	-16.5 ± 58.4	0.970	NA
Knee flexion	-65.5 ± 12.1	-63.4 ± 12.4	2.1 ± 11.5	0.542	0.17
Hip extension	205.2 ± 51.6	220.8 ± 33.6	15.6 ± 65.2	0.427	0.36
Hip flexion	-70.7 ± 23.8	-68.4 ± 30.9	2.3 ± 21.3	0.698	0.08

- Knee extension moment data were non-parametric

There were no significant differences in cycling specific strength (peak joint moments at 60 rpm) between pre and post strength training intervention (Table 6.2).

There were no significant differences in IE for complete crank cycle or each of the four function sectors between pre and post strength training intervention (Table 6.3). There was a significant difference in average crank power in the bottom sector between pre and post strength training intervention (Table 6.3, $P = 0.007$, ES = -0.53).

Table 6.3: Index of mechanical effectiveness (IE) and average crank power for the four functional sectors for sprints at 135 rpm: pre and post strength training intervention (left side only)

Variable		Mean (SD)				
	Units	**Pre**	**Post**	**Change**	***P***	**Effect Size**
IE complete rev	%	67.5 ± 8.0	67.7 ± 5.9	0.3 ± 3.6	0.622	NA
IE downstroke	%	84.9 ± 3.1	85.2 ± 2.2	0.3 ± 2.4	0.653	0.12
IE bottom	%	38.0 ± 9.9	38.8 ± 8.0	0.9 ± 5.4	0.587	0.10
IE upstroke	%	36.5 ± 22.8	37.6 ± 18.6	1.1 ± 14.9	0.804	0.05
IE top	%	52.8 ± 33.3	60.3 ± 28.3	7.5 ± 17.4	0.164	0.24
Average crank power downstroke	W	1093.8 ± 212.5	1140.6 ± 216.4	46.8 ± 84.1	0.080	0.22
Average crank power bottom	W	357.1 ± 73.9	401.0 ± 102.9	43.9 ± 58.6	0.007**	NA
Average crank power upstroke	W	63.0 ± 42.4	66.5 ± 36.0	3.5 ± 17.8	0.515	0.09
Average crank power top	W	147.9 ± 75.7	162.8 ± 41.2	14.9 ± 73.2	0.497	0.24

- ** indicates significant difference between sessions ($P < 0.01$)
- IE complete rev and average crank power in downstroke data were non-parametric

There was low positive correlation (r = 0.413) between change in squat predicted 1RM and change in average left crank power over a complete revolution for sprints at 135 rpm between pre and post strength training intervention (Figure 6.3).

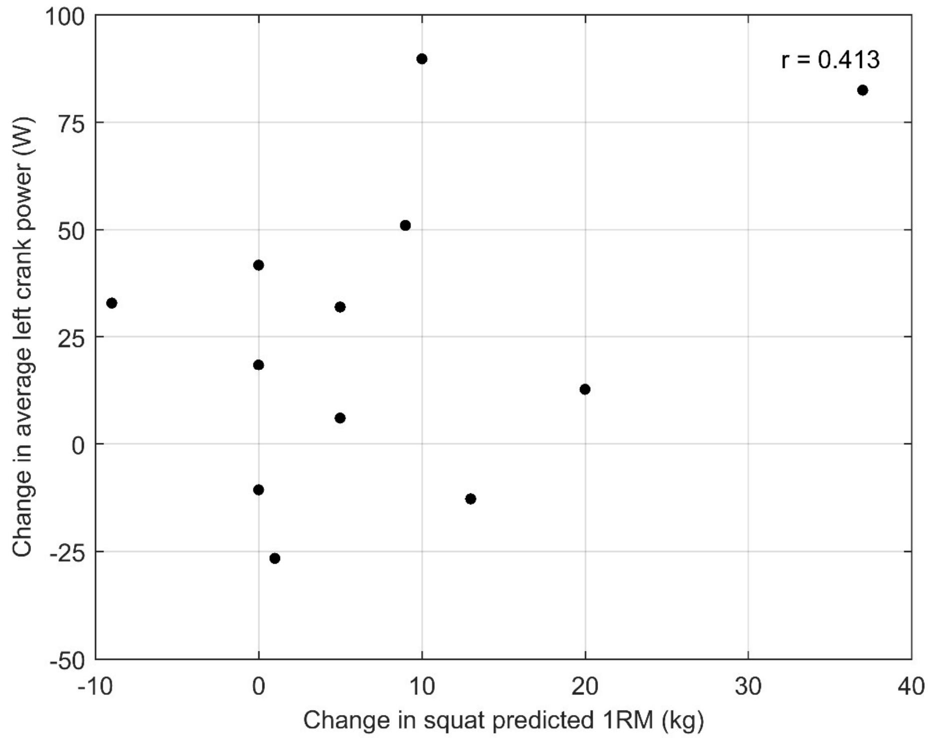

Figure 6.3: Relationship between change in squat predicted 1RM and change in average left crank power over a complete revolution for sprints at 135 rpm between pre and post strength training intervention.
Pearson's correlation coefficient (r)

There was a strong positive significant correlation (r = 0.702, *P* = 0.011) between pre strength training intervention relative hip joint extension power and change in average left crank power over a complete revolution for sprints at 60 rpm between sessions (Figure 6.4). There was also a strong negative significant correlation (r = -0.769, *P* = 0.003) between pre strength training intervention relative knee extension power and change in average left crank power over a complete revolution for sprints at 60 rpm between sessions (Figure 6.4). There was little correlation between pre strength training intervention relative hip and knee joint extension powers and change in average left crank power over a complete revolution for sprints at 135 rpm between sessions (Figure 6.5).

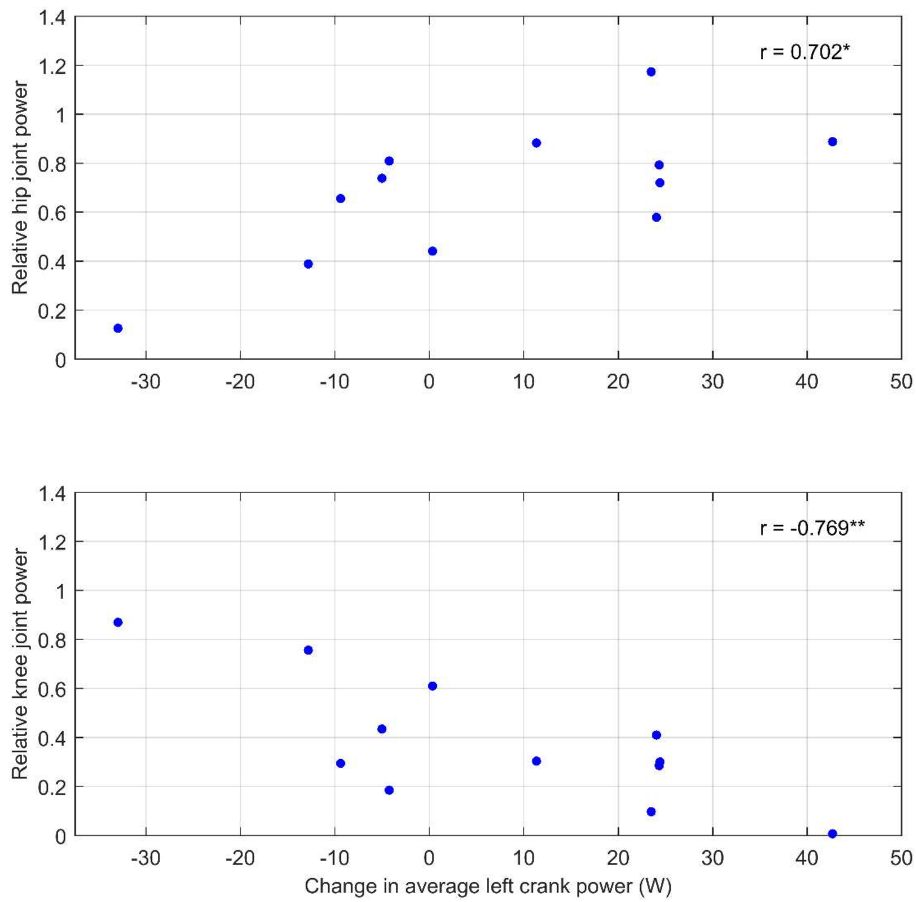

Figure 6.4: Change in average left crank power over a complete revolution for sprints at 60 rpm between pre and post strength training intervention correlated with relative hip and knee joint extension powers in pre strength training testing session.

Pearson's correlation coefficient (r), * indicates *P* < 0.05, and ** *P* < 0.01.

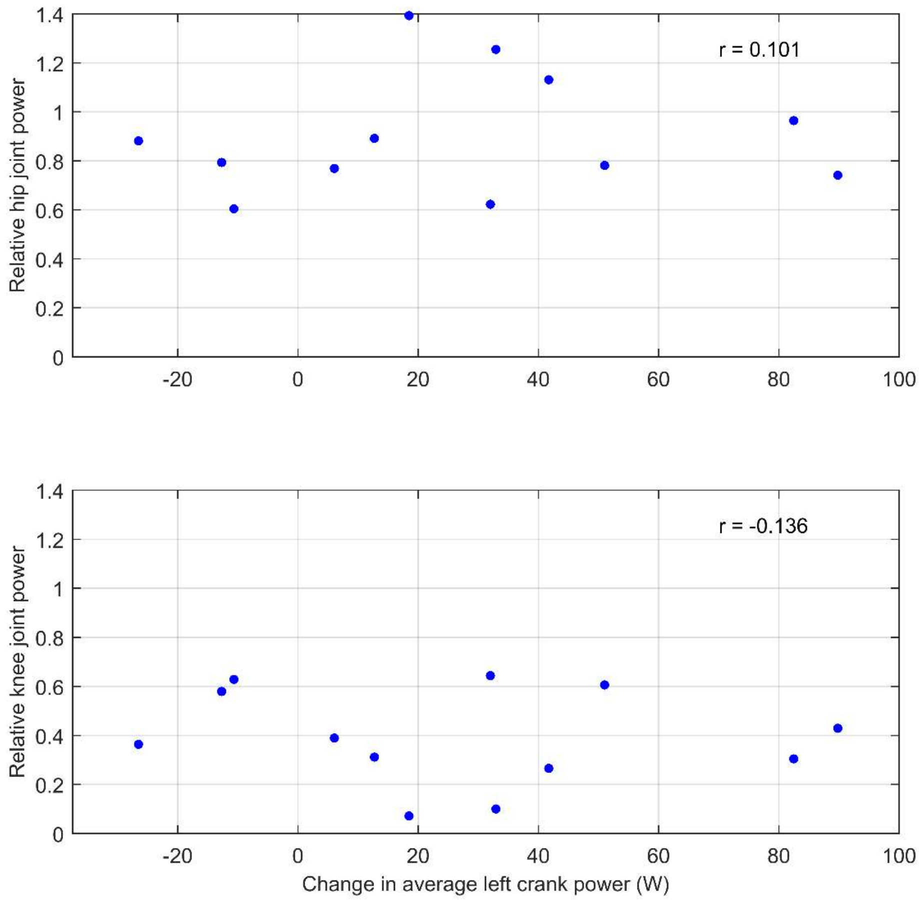

Figure 6.5: Change in average left crank power over a complete revolution for sprints at 135 rpm between pre and post strength training intervention correlated with relative hip and knee joint extension powers in pre strength training testing session.

Pearson's correlation coefficient (r)

Time series variables – Sprints at 135 rpm

Knee joint angular velocity was significantly different ($P < 0.05$) between pre and post strength training intervention, between crank angles 348 to 4° (Figure 6.8). Knee joint power was significantly different ($P < 0.05$) between pre and post strength training intervention, between crank angles 337 to 342° (Figure 6.8). There were no significant differences between instantaneous crank powers, forces and other joint angles, angular velocities, moments and powers (Figure 6.6, Figure 6.7, Figure 6.8).

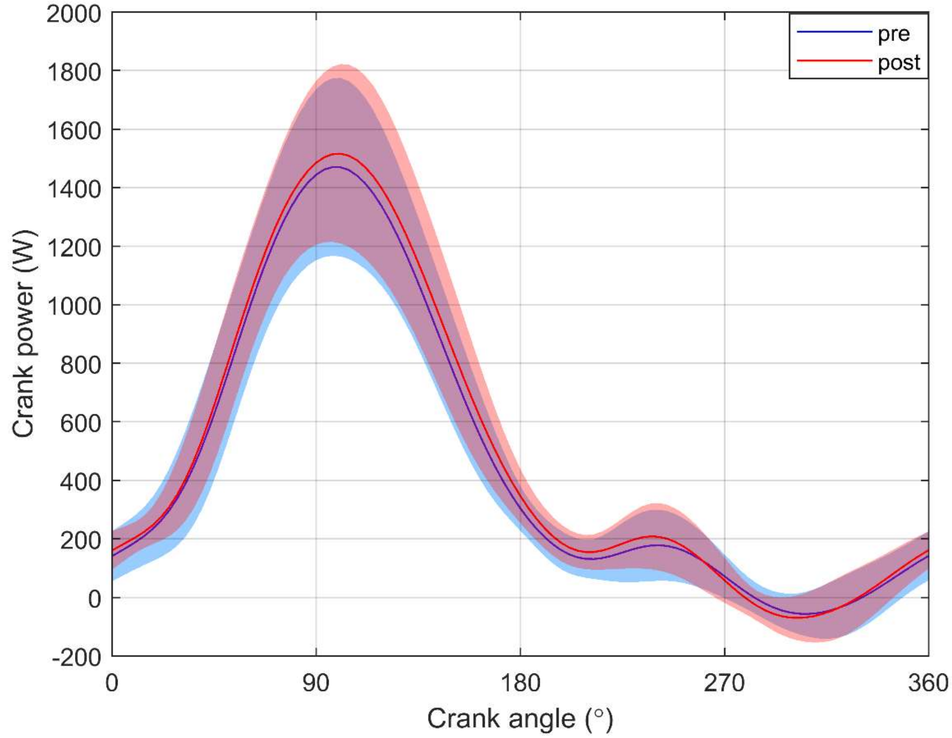

Figure 6.6: Crank power for sprints at 135 rpm: pre and post strength training intervention

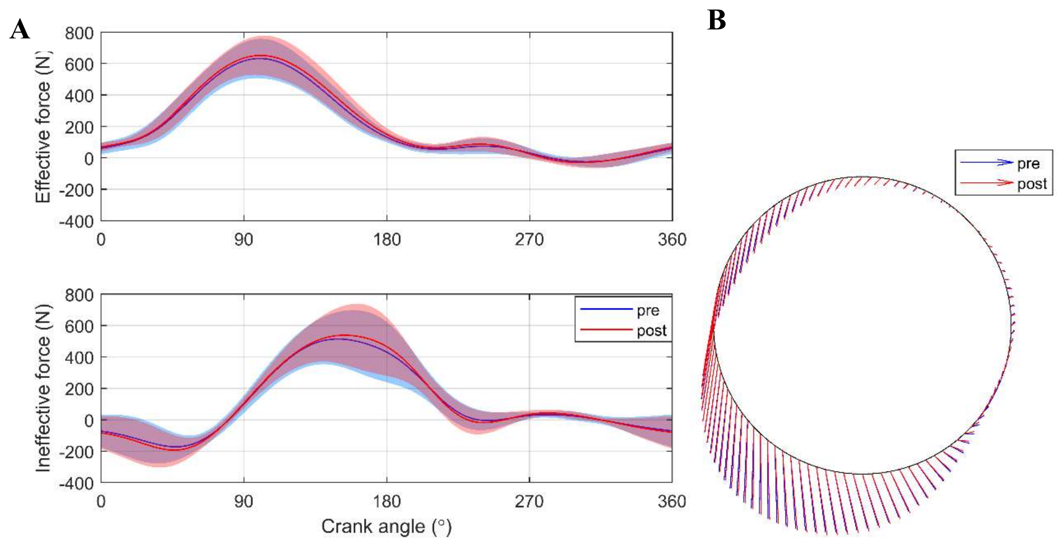

Figure 6.7: Crank forces for sprints at 135 rpm: pre and post strength training intervention

A: Crank force separated into effective and ineffective components

B: Visualisation of crank forces

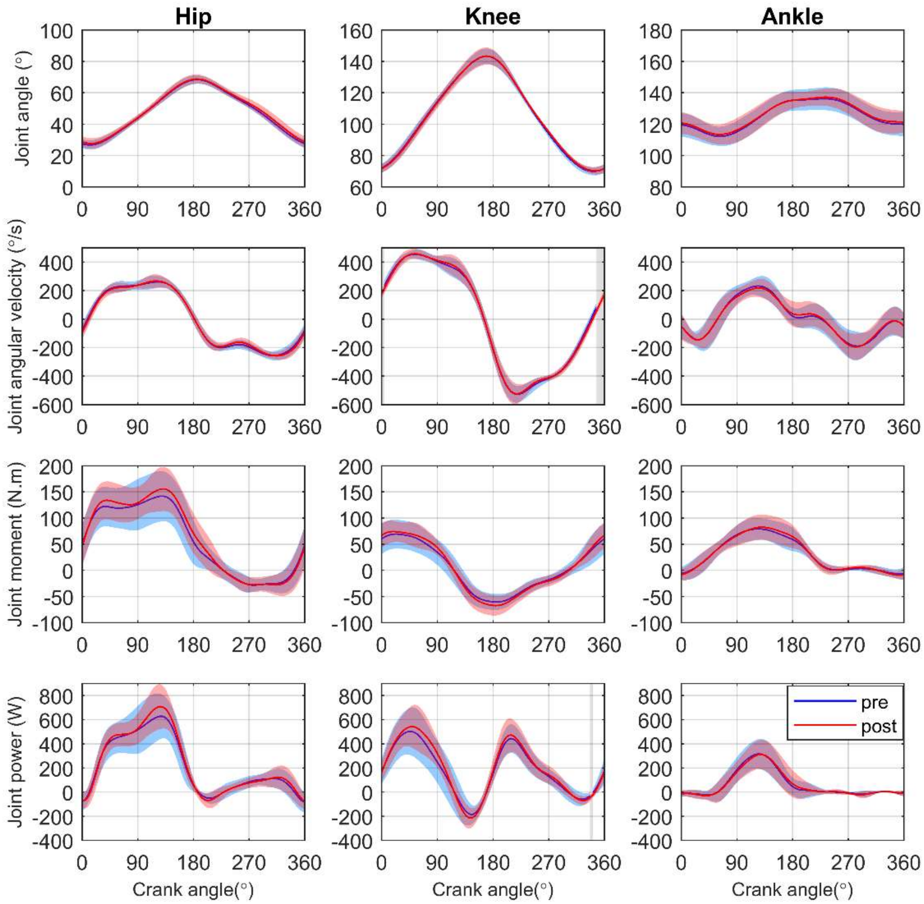

Figure 6.8: Joint angles, angular velocities, moments and powers for sprints at 135 rpm: pre and post strength training intervention
Areas of the graph shaded grey where the SPM is significant.

For ease of presenting the data the thigh angle and angular velocity are presented as hip angle and angular velocity

There was no significant differences between relative joint extension and flexion powers between pre and post strength training intervention (Figure 6.9).

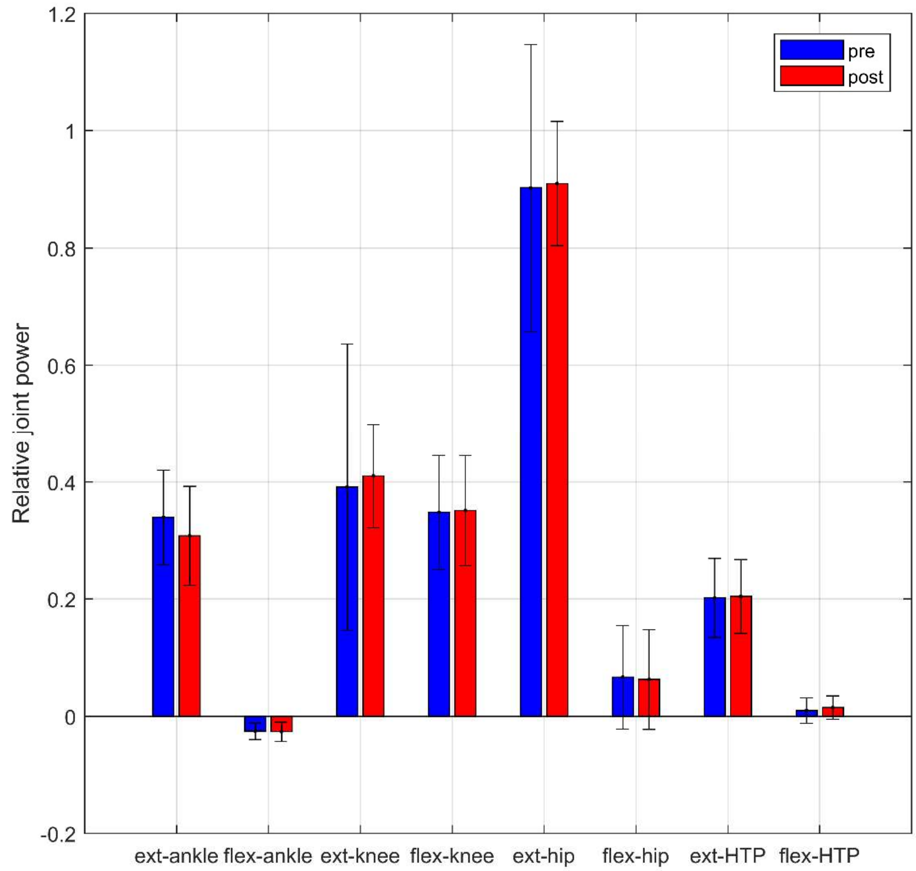

Figure 6.9: Relative joint powers in extension and flexion phases for sprints at 135 rpm: pre and post strength training intervention.
HTP = Hip transfer power

The P values and effect sizes for relative joint powers in extension and flexion between pre and post strength intervention: Ankle extension: *P* = 0.284, ES = -0.38, Ankle flexion: *P* = 0.784, ES = -0.06, Knee extension: *P* = 0.776, ES = 0.12, Knee flexion: *P* = 0.921, ES = 0.03, Hip extension: *P* = 0.924, ES = 0.04, Hip flexion: *P* = 0.838, ES = -0.04, HTP extension: *P* = 0.775, ES = 0.04, HTP flexion: *P* = 0.406, ES = 0.24

There were no significant difference between the frequency of the hip-ankle moment coordination phases during the downstroke for sprints at 135 rpm between pre and post strength training intervention (Figure 6.10).

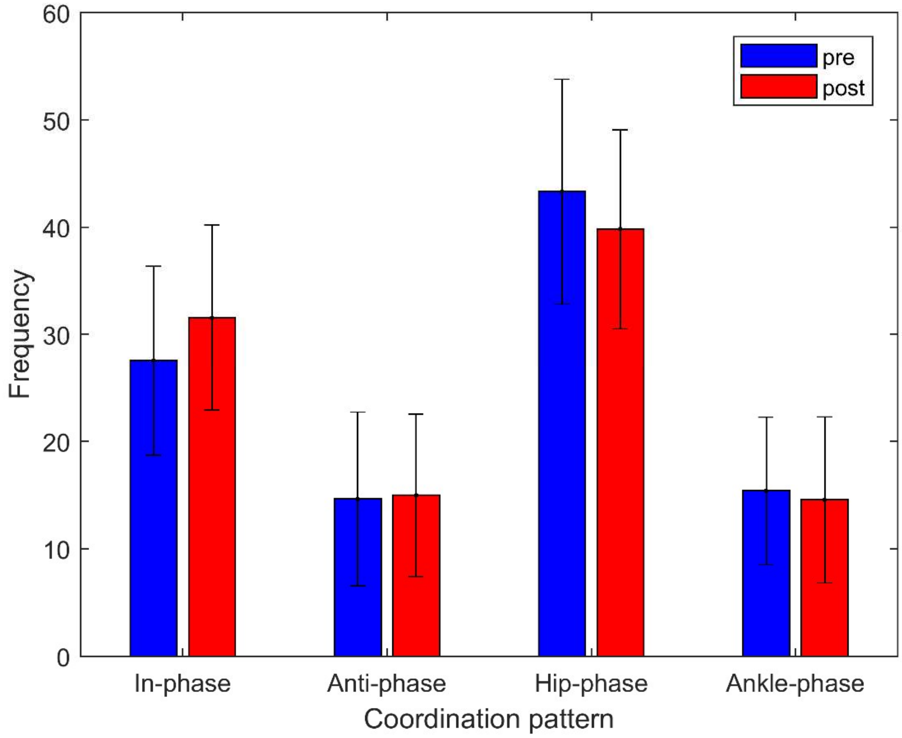

Figure 6.10: Hip-ankle moment coordination patterns during downstroke phase of the crank cycle for sprints at 135 rpm: pre and post strength training intervention The P values and effect sizes for coordination patterns between pre and post strength intervention: In-phase: *P* = 0.428, ES = -0.18, Anti-phase: *P* = 0.939, ES = -0.02, Hip phase: *P* = 0.311, ES = -0.22, Ankle phase: *P* = 0.998, ES = -0.01

EMG activity for the BF muscle was significantly different ($P < 0.05$) between pre and post strength training intervention between crank angles 107° to 119° (Figure 6.11).

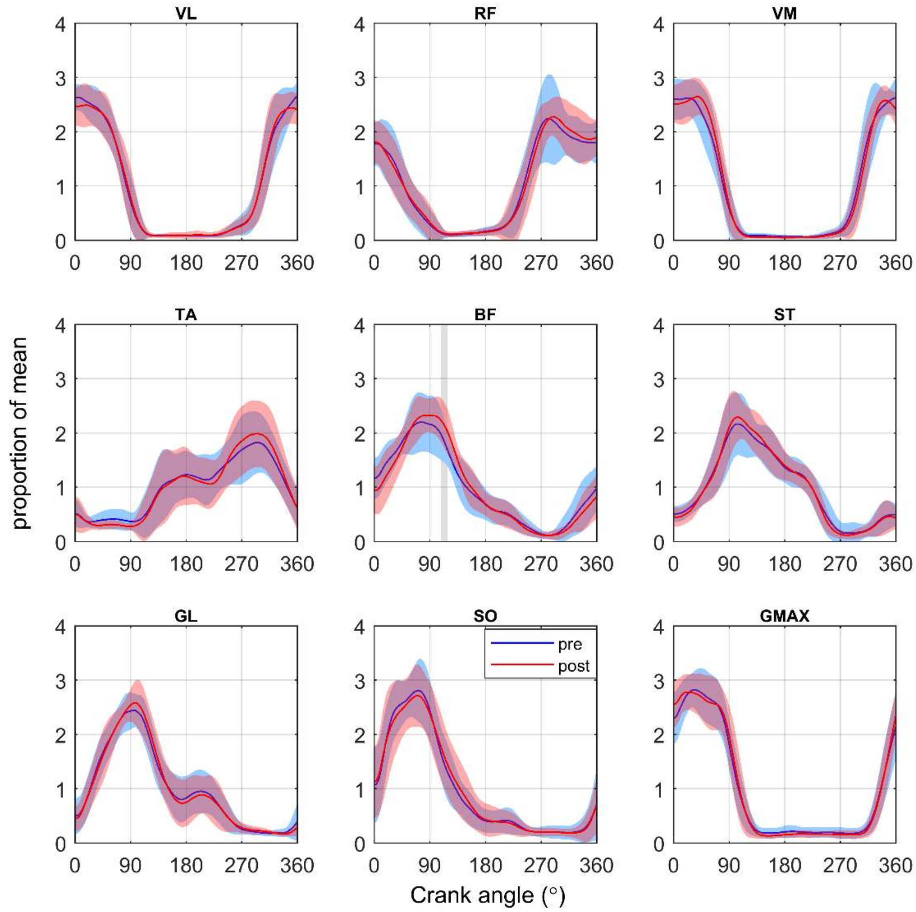

Figure 6.11: EMG linear envelopes (normalised to mean value in signal) for each muscle for sprints at 135 rpm: pre and post strength training intervention VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF = biceps femoris, ST = semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus.

Areas of the graph shaded grey where the SPM is significant.

6.4 Discussion

This study investigated the acute effects of a strength training intervention on intermuscular coordination in short-term maximal cycling. 'Gym strength', as quantified by squat predicted 1RM, increased post strength training intervention. This observation was accompanied by a significant increase in average left crank power over a complete revolution for sprints at 135 rpm with a significant increase average left crank power produced in the bottom sector of the crank cycle. This supports the findings of previous research that strength training positively correlates with cycling power (Stone, et al., 2004; Wilson et al., 1993). However, there was no change in cycling specific strength (peak joint moments at 60 rpm) and thigh volume following the strength training intervention, which would causally link the increase in 'gym strength' to the increase in cycling power at 135 rpm. Furthermore, the increase in average left crank power over a complete revolution was not associated with significant changes in the other biomechanical variables (instantaneous crank power, forces, joint angles, angular velocities, moment and powers, muscle activity, IE and hip-ankle moment synergy), suggesting that individual coordination strategies might have been adopted by the participants to increase average left crank power over a complete revolution following the strength training intervention.

This study investigated the acute effects of a strength training intervention on maximal cycling coordination. The results of this study were considered in relation to two possible mechanisms of how strength training could affect sport coordination patterns that have previously been identified in the literature. The first mechanism considered, was whether, following a period of strength training, the coordination patterns of strength training exercises could start to be expressed during sporting movement performance, thereby having a detrimental effect on sports performance (Carroll et al., 2001). This potential outcome was not supported by the results of this study, as the muscle activation patterns pre and post strength training intervention were very similar. Only the BF muscle activation pattern was significantly different between the crank angles of 107° to 119° (Figure 6.11) following the strength training intervention. When the individual participants' pre and post EMG activation patterns were explored subjectively they were similar in shape, with differences observed for only four participants for one muscle each (the shape and timing of onset and offset of muscle activity of RF, BF, ST and TA were different between-sessions). This finding suggests that the strength training exercises' coordination patterns were not expressed during maximal cycling following the strength training intervention.

The second mechanism explored how intermuscular coordination might have affected the transfer of strength training to sports performance. Specifically, whether in order to observe a performance improvement, intermuscular coordination patterns needed to be adapted to enable athletes to use their increased muscle strength resulting from the strength training. Coordination post strength training intervention was therefore assessed through analysis of four key mechanical features of maximal cycling.

First, it has been suggested that the hip and ankle joints work in synergy during the downstroke (Fregly & Zajac, 1996). It was postulated that strength training might alter the strength of the synergy between the hip and ankle. However, the findings revealed no change in the strength of the hip-ankle synergy in the downstroke following the strength training intervention i.e. the frequency of hip-ankle moment in-phase coordination pattern was unchanged (Figure 6.10).

Second, activation of the bi-articular hamstrings muscles is required to help control the direction of the applied force to the pedal (van Ingen Schenau et al., 1992). It was hypothesised that strength training might alter the hamstrings' muscle activation patterns. Following the strength training intervention there was a change in BF muscle activity for a small region of the crank cycle (Figure 6.11). This change in hamstring muscle activity might be related to the control of the applied force to the pedal. However, the IE was unchanged in all crank sectors following the strength training intervention (Table6 .3), suggesting the direction of applied force was unchanged following the strength training intervention, although the change in BF muscle activity could be to maintain the same IE. When interpreting the EMG activity in relation to muscle force, the electromechanical delay (time between EMG activity and production of mechanical force) needs to be considered. This is typically around 50 ms (Cavanagh & Komi, 1979; Hug et al., 2008), which at 135 rpm equates to 50° of the crank cycle. Taking into account the EMD when interpreting the BF muscle activity could mean the hamstring muscles were producing a force for slightly longer and with greater magnitude in the bottom sector of the crank cycle, potentially explaining the increase in the bottom sector crank power following strength training.

Third, at higher pedalling rates, muscle activation-deactivation dynamics are a major constraint on power production (McDaniel et al., 2014; Neptune & Kautz, 2001; van Soest & Casius, 2000). Therefore, coordination strategies that can maximise the muscle force production in the main power-producing phase of the downstroke are beneficial. One of these strategies is to time the activation of the powerful hip and knee extensor muscles (GMAX/VL/VM), so they activated as maximally as possible at a crank angle of around 90° from top dead centre (TDC) – the location of peak crank power (Dorel et al., 2012; McDaniel et al., 2014). It was speculated that strength training might alter the activation timing of the GMAX/VL/VM muscles. However, there were no changes in muscle activation timings and patterns of the main power producing muscles (GMAX/VL/VM) following the strength training, suggesting the participants did not alter their strategies to limit the effect of activation-deactivation dynamics on maximal crank power.

Fourth, at below and optimal pedalling rates for maximum crank power production, cyclists actively pull up during the upstroke generating positive crank power. It was postulated that strength training might alter the crank power generation in the upstroke. However, following the strength training intervention there were no changes in the IE and the crank power produced in upstroke (Table 6.3), suggesting the participants did not improve their upstroke. Therefore, with the exception of the later offset of the BF muscle activity following strength training, there was no change in any of the key mechanical features of maximal cycling to use the increased muscle strength gained from the gym-based strength training.

This study did not include a long-term follow up testing session (such as 8 to 10 weeks following the completion of the strength training intervention). It was therefore not possible to assess whether the participants adapted their coordination patterns after a period of cycling focussed training to use their increased muscle strength developed during the gym-based strength training period. This was suggested by Bobbert and Van Soest who recommended a period of sports-specific training was required following strength training to allow athletes to adapt their intermuscular coordination patterns to use their increased muscle strength obtained from strength training to improve their sports performance (Bobbert & Van Soest, 1994).

Average left crank power over a complete revolution for sprints at 135 rpm significantly increased following the strength training intervention. However, the mechanisms to explain this change are unclear as there were no significant changes in cycling specific strength (peak joint moments at 60 rpm) (Table 6.2) following the strength training intervention. Nor were there any significant changes in the instantaneous crank power and forces, joint moments and powers and muscle activation patterns, with the exception of the later offset of the BF muscle following the strength training intervention (Figure 6.6, Figure 6.7, Figure 6.8, Figure 6.11). There was a significant difference between knee joint angular velocity and power for a small region of the crank cycle between sessions. However, these are unlikely to be meaningful changes as these differences occur for very small regions of the crank cycle where the variable is of low magnitude. There was high inter-participant variability in initial coordination strategies, as evidenced in Figure 6.9, where there is a large standard deviation for the relative contribution of individual joints in the extension and flexion phases to the crank power. High inter-participant variability in coordination strategies has been identified previously by Broker and Gregor, who found high inter-participant variability in joint moment patterns, particularly at the hip joint, in submaximal cycling even amongst a homogenous group of cyclist (12 junior national team male cyclists) (Broker & Gregor, 1994). Dorel and colleagues also found high inter-participant variability in EMG patterns in elite track sprint cyclists (Dorel et al., 2012). The high inter-participant variability observed in initial coordination strategies could potentially be obscuring the changes in the biomechanical variables following the strength training intervention.

Following the strength training intervention participants might have employed individual coordination strategies to increase average left crank power over a complete revolution i.e. some participants, for example, could have improved the downstroke (increased crank power using hip and knee extensors), whereas others could have improved the upstroke (less negative or more positive contribution to crank power using hip and knee flexors). These changes could be caused by increases in muscle strength or improvements in coordination, or a combination of both. This speculation that participants developed individual strategies to increase crank power would be predicted by Newell's model of constraints (Newell, 1986), which proposes the patterns of coordination emerge from the confluence of constraints acting on the human movement system. In this study, strength training would be expected to change the organismic constraints such as muscle size, strength and fatigue. Therefore, the interaction of the participants' changing individual organismic constraints following a strength training intervention with the task constraints could have resulted in different coordination patterns emerging.

The low positive correlation between change in squat predicted 1RM and change in average left crank power over a complete revolution at 135 rpm between pre and post strength training intervention (Figure 6.3) adds empirical evidence to support some coaches' beliefs that there is no direct correlation between increases in 'gym strength' and sports performance (Chapter 3) (Burnie et al., 2018). The findings support data from previous research suggesting that the transfer of strength training to sports performance varies (positive, no change, or negative) (Carroll et al., 2001; Young, 2006). One of the factors coaches believe can influence the transfer to strength training to sports performance is fatigue induced by a period of heavy strength training (Chapter 3) (Burnie et al., 2018). However, there were no significant differences between pre and post strength training subjective measures of fatigue (energy level, sleep quality and muscle soreness) (Table 6.1), suggesting that for the participants in this study, fatigue was not a factor influencing their performance at the post strength training intervention testing session.

The increases in participants' 'gym strength' (squat predicted 1RM) following the strength training intervention were not associated with changes in thigh volume (Table 6.1). Thus, it is unlikely that the increase in 'gym strength' can be explained by an increase in thigh muscle cross sectional area (CSA) (Table 6.1). In addition, the increase in 'gym strength' did not result in any significant increases in cycling specific strength (peak joint moments at 60 rpm - Table 6.2). These two findings taken together suggest that the increase in 'gym strength' could be because of improved intermuscular coordination during the squat exercise allowing the participants to lift greater load. However, this is speculation as the participants' intermuscular coordination patterns during the squat exercise pre and post strength training intervention were not measured. As intermuscular coordination of squat exercise is independent to cycling, this means there was not a mechanism to causally link the increase in 'gym strength' to the increase in cycling power at 135 rpm, as muscle strength appears not to have increased following the strength training intervention.

When exploring the data, *post-hoc*, to explain the increase in average left crank power over a complete revolution following the strength training intervention, the participants' relative joint extension power appeared to influence the performance gain associated with the strength training intervention when sprinting at 60 rpm but not at 135 rpm.

There was a strong positive significant correlation between pre strength training intervention relative hip extension power for sprints at 60 rpm and change in average left crank power over a complete revolution - participants with greater hip extension power in the pre-test showed greater increase in average left crank power over a complete revolution following the intervention (Figure 6.4), but there was little relationship between relative joint extension powers and change in average left crank power over a complete revolution for sprints at 135 rpm (Figure 6.5). The traditional squat exercise targets the hip extensors and therefore, could elicit greatest increase in the hip extensor strength and power. It could be speculated that riders with a hip dominant strategy (greater relative joint extension power produced at hip rather than the knee joint) would achieve a greater increase in crank power following the strength training intervention. At the higher pedalling rate there were no relationships observed between the relative joint power contributions at the knee and hip and the change in crank power following the strength training intervention, which suggested at the higher pedalling rates the initial coordination strategy (whether more relative power is produced at the knee or hip) had less influence on the transfer of the strength training to maximal cycling performance. This observation is congruous to the findings by Dorel and colleagues that intermuscular coordination plays an increasingly important role to achieve maximum power production at high pedalling rates, particularly those above f_{opt} in addition to the intrinsic force- and power-velocity characteristics of the muscles (Dorel, 2018a; Dorel et al., 2014; Samozino et al., 2007).

There is a paucity of research studying elite athletes (Williams & Kendall, 2007). The requirement for research using elite athletes as participants was highlighted by Hakkinen, who suggested that for well-trained, and in particular elite strength, athletes, the magnitude and time courses of the neuromuscular adaptions to strength training may differ to untrained participants - smaller improvements over a longer time course (Hakkinen, 1989). However, typically the research into effects of strength training has been conducted using untrained or active participants, which raises the question of whether there is an adequate research base to inform training interventions and programmes for elite athletes (Williams & Kendall, 2007). Therefore, this study provides a valuable contribution to the literature.

In well-trained and elite athletes, it is often difficult to observe a statistically significant change in performance following a strength training intervention (Cormie et al., 2011b; Häkkinen, Komi, Alén, & Kauhanen, 1987; Hakkinen, Pakarinen, Alen, Kauhanen, & Komi, 1988). Therefore, changes might have occurred in joint moments and powers and muscle activity following the strength training intervention that were too subtle to be detected by this measurement protocol. In Chapter 5, the reliability of the biomechanical variables measured using this maximal cycling protocol were quantified. The peak hip joint extension moments and powers (minimal detectable differences (MDD) of 13 N.m and 144 W), and peak knee flexion moment (MDD of 26 N.m) were found to be the least reliable, with large MDDs. The magnitude of the change in these variables following the strength training intervention was smaller than the MDDs, and therefore it was not possible to determine if a 'real change' had occurred owing to the size of the measurement error. This issue is especially relevant to this study as one of the key gym exercises in the strength training intervention was the squat with heavy load, which targets the hip extensor muscles in particular. Therefore, as peak hip power had a large MDD, a large change in magnitude this variable would be required for a 'real change' to be observed. As discussed in Chapter 5, further development of the methods for measuring lower limb kinematics is required, using more detailed marker sets and models of STA, to reduce the influence of STA and skin marker misplacement on the calculated kinematics and kinetic variables, which may improve the reliability of the calculated knee flexion and hip joint variables and reduce the MDDs.

A possible limitation of this study concerns the lack of a control group (i.e. a group that did cycling training sessions only during the intervention period). However, as the aim was to recruit elite and high-level track sprint cyclists as participants for this study, it would have been unethical to ask one sample of elite athletes to act as controls for treatment groups owing to the potential for interference in their scheduled training for high-level competitions. This issue, however, makes it difficult to ascertain whether the changes / lack of change are due solely to the strength training intervention. The use of elite and high-level athletes also meant it was not possible to standardise the content of the strength training programmes (number of sessions per week, exercise sets and reps), although the programmes all included similar exercises, as it was infeasible to interfere with their performance preparation to such a large extent. Therefore, a more observational analytic approach was implemented in this study to advance our understanding further of elite athletes which are not well represented in the scientific research (Williams & Kendall, 2007).

Another limitation of this study was that it was not possible to assess whether co-contraction of the muscles around the knee joint increased following the strength training intervention, as the EMG activity was normalised to mean value in signal and not a maximal voluntary contraction (MVC). Therefore, it cannot be determined whether the amplitude of the activation of the antagonist muscles at the knee increased during the knee extension phase of the crank cycle which is required to determine if the level of co-contraction changed. The decision was made not to normalise the EMG signals to a MVC for each muscle as this method has been shown to be unreliable for between-sessions comparisons (for more details refer to Appendix 9.7) (Sinclair et al., 2015) and even when using cycling specific MVC procedures maximal muscle activity is not always elicited (Dorel et al., 2012). Therefore, future research needs to consider methods of how to assess muscle co-contraction at the joints and to develop a more reliable EMG normalisation procedure for between-sessions comparisons.

6.5 Conclusion

Track sprint cyclists' 'gym strength' increased following a strength training intervention and this was accompanied by a significant increase in average left crank power over a complete revolution for sprints at 135 rpm. However, there was no change in cycling specific strength and thigh volume following the strength training intervention which would causally link the increase in 'gym strength' to the increase in cycling power at 135 rpm. Furthermore, the increase in average left crank power over a complete revolution was not associated with significant changes in key mechanical features of maximal cycling and other biomechanical variables that describe intermuscular coordination in maximal cycling. Therefore, the participants might have adopted individual coordination strategies to increase crank power following strength training. Such observations are consistent with the tenets of ecological dynamics and would be predicted by Newell's model of constraints (Newell, 1986). Further research is required to investigate how the cyclists' intermuscular coordination patterns change after an additional period of cycling focussed training following the completion of the strength training intervention, to assess whether cyclists require a period of sport-specific training to enable them to learn how to use their increased muscle strength gained from the strength training intervention to increase crank power.

7 Overall Discussion

The aim of this programme of research was to investigate whether intermuscular coordination in maximal cycling is influenced by strength training. To achieve this, five objectives were identified. This chapter summarises the main findings of this programme of research in relation to each objective, the practical applications of the research, followed by the limitations, areas for further research and the contribution to knowledge.

7.1 Summary of findings

Objective one: To understand coaches' philosophies on the transfer of strength training to elite sports performance

Semi-structured interviews were carried out with thirteen elite coaches and athletes from the disciplines of track sprint cycling, BMX, sprint kayaking, rowing and athletics sprinting. The interviews captured the coaches' experiential knowledge regarding strength training and the range of factors believed to affect transfer of strength training to sport performance (Chapter 3). Their views indicated that the role of non-specific strength training ("traditional" gym-based strength exercises that are not specific to a sport movement) is to increase athletes' muscle size and strength. This training method is typically used in conjunction with resisted sport movement training (for example, increased resistance running, pedalling or rowing), as it is believed to achieve an effective transfer of enhanced muscle strength to sports performance. They believed the transfer of strength training to sports performance was a complex process, with factors associated with fatigue and coordination having particular importance. The importance the coaches placed on coordination is captured by the theoretical framework of ecological dynamics and Newell's model of constraints that describes how each athlete needs to adapt their intermuscular coordination patterns in response to a change in his/her unique set of "organism constraints" (e.g. muscle strength, size and fatigue) (Newell, 1986). This perspective is also supported by a musculoskeletal simulation model that demonstrated increases in muscle strength from strength training may need to be accompanied with a change in intermuscular coordination to improve sport performance (Bobbert & Van Soest, 1994). The coaches' experiential knowledge and the factors they identified as being important in the transfer of strength training to sports performance were considered in the interpretation of the strength training intervention in Chapter 6 and Objective 5.

Objective two: To identify variables that describe intermuscular coordination in maximal cycling

A review of the literature identified several key mechanical features of maximal cycling (Chapter 2 and 6). First, the hip and ankle joint are suggested to work in synergy during the downstroke, to enable the ankle to transfer the power produced by the hip extensor muscles to the crank (Fregly & Zajac, 1996). Second, adjustment of the bi-articular rectus femoris and hamstring muscles activation are suggested to control the direction of the external force on the pedal (van Ingen Schenau et al., 1992). Third, at higher pedalling rates, muscle activation-deactivation dynamics have been shown to be a major constraint on power production (McDaniel et al., 2014; Neptune & Kautz, 2001; van Soest & Casius, 2000), as there is insufficient time to fully activate the muscles to achieve maximal force production during a crank cycle. Therefore, coordination strategies that can maximise muscle force production in the main power producing phase of the downstroke are beneficial. Fourth, at below or optimal pedalling rates for maximum crank power production, cyclists actively pull up during the upstroke to generate positive power in the upstroke during maximal cycling in comparison to submaximal cycling where the upstroke may be more passive (Dorel et al., 2009; Dorel et al., 2010). Dorel and colleagues found positive relationships between upstroke power and average crank power over a revolution, and between index of mechanical effectiveness (IE - ratio of effective crank force to the total crank force) and power output during the upstroke in maximal cycling (Dorel, 2018b; Dorel et al., 2010).

There are several biomechanical variables, that can be measured experimentally, which have been used to describe intermuscular coordination during maximal cycling (section 2.7.2). These include: measuring EMG activity to determine muscle activation onset and offset times and level of activation; measuring crank kinetics and lower-limb kinematics which can be input into inverse dynamics calculations to obtain the joint kinetics at the hip, knee and ankle joints throughout the pedal revolution. Combining information on muscle activation from EMG and joint kinetics from inverse dynamics analysis provides a deeper understanding of the joint and muscle actions that produce the movement. Hence, both are required to describe intermuscular coordination in maximal cycling and were chosen for measurement and analysis during maximal cycling for experimental studies (Chapters 4, 5, and 6).

Objective three: To compare the biomechanical data of a sprint cyclist in the velodrome and in the laboratory

The study reported in Chapter 4 identified relatively small differences in movement organisation between sprinting on a velodrome track and on an ergometer. However, the static task constraints of ergometer cycling led the cyclists to adopt a different position, increase knee joint duty cycle and knee moment over TDC, and delayed the offset of the main power producing muscles. All these factors potentially contributed to an increase in overall crank power output on the ergometer compared to the track where cyclists also needed to control the stability and direction of the bicycle. Future research is needed to assess whether the differences in joint angles, EMG activity and crank powers were owing to the different environmental and task constraints between the ergometer and the track bicycle sprints. The findings imply it is important to undertake biomechanical analyses of movement organisation in elite sports practice in a representative environment.

Although this study revealed differences in sprint cycling biomechanics between sprinting on the ergometer and on the track, a decision was made to use the ergometer in the laboratory for the testing protocol for studies reported in Chapter 5 and 6 which address Objectives 4 and 5. The reasons for this decision were: the current on-track data collection method could only measure one revolution per effort on track owing to the limitations of the equipment available which is insufficient to study coordination owing to between-revolution variability in maximal cycling. There were also technical problems during the on-track data collection sessions where the Wi-Fi connection was lost between the EMG sensors, force pedals and the laptop recording the data which meant data from trials were lost. In addition, it was very difficult to obtain the track at the velodrome for testing sessions. This meant there was a risk that it would be not be possible to collect data from many participants or at the time intervals required by the research question. Therefore, the laboratory testing protocol using the ergometer was used to address Objectives 4 and 5.

Objective four: To quantify test-retest reliability of biomechanical variables measured during maximal cycling on an ergometer

The test-retest study in Chapter 5 identified biomechanical variables that describe maximal cycling on an ergometer were more reliable within-session than between-sessions. Typically, the biomechanical variables that describe maximal cycling are reliable. However, some variables, such as peak knee flexion moment and maximum hip joint power, demonstrated lower reliability, indicating that care needs to be taken when using these variables to evaluate changes in maximal cycling biomechanics. The MDDs identified in this study can be used by researchers and sports science practitioners to help understand the magnitude of the change required in the biomechanical variables for a 'real change' to have occurred. This information can be used to help them interpret the effect of longitudinal interventions such as changes to bicycle set-up and training programmes on athletes' maximal cycling performance (such as the strength training intervention reported in Chapter 6 to address Objective 5). Although measurement error (instrumentation error, anatomical marker misplacement and soft tissue artefacts) can explain some of the reliability observations reported in this study, it can be speculated that biological variability may also be a contributor to the lower repeatability observed in several variables including ineffective crank force, ankle kinematics and hamstring muscles' activation patterns.

Objective five: To investigate the effect of gym-based strength training on intermuscular coordination in maximal cycling

The study reported in Chapter 6 found track sprint cyclists' 'gym strength' increased following a strength training intervention and this was accompanied by a significant increase in average left crank power over a complete revolution for sprints at 135 rpm. This supports the findings of previous research that strength training positively correlates with cycling power (Stone, et al., 2004; Wilson et al., 1993). However, there was no change in cycling specific strength and thigh volume following the strength training intervention which would causally link the increase in 'gym strength' to the increase in cycling power at 135 rpm. Furthermore, the increase in average left crank power over a complete revolution was not associated with significant changes in the key mechanical features of maximal cycling and the other biomechanical variables that describe intermuscular coordination in maximal cycling. The findings suggest that the participants might have adopted individual coordination strategies following the strength training intervention to increase crank power. Such observations are consistent with the tenets of ecological dynamics and would be predicted by Newell's model of constraints (Newell, 1986).

There was no evidence of the two possible mechanisms that have previously been identified in the literature of how intermuscular coordination could affect the transfer of strength training to sports performance following the strength training intervention. First, it was possible following a period of strength training that the coordination patterns of strength training exercises undertaken during the period of training could start to be expressed during sporting movement (Carroll et al., 2001). This was not however, evident in the results of the present study, as the EMG activation patterns pre and post strength training intervention were very similar. Second, it was possible that, following a period of strength training the sport movement intermuscular coordination patterns needed to be adapted to enable the athlete to use their increased muscle strength obtained from the strength training before an improvement sports performance was observed. However, there was no change in any of the key mechanical features of maximal cycling and the other biomechanical variables that describe intermuscular coordination in maximal cycling following the strength training intervention. These findings suggest that there is no immediate adaptation of the cyclists' intermuscular coordination patterns to use the increased muscle strength gained from the gym-based strength training to improve performance.

7.2 Limitations

The limitations of each study have been identified in the individual chapters. However, the main limitations to this programme of research are discussed here.

The effect of strength training on intermuscular coordination in maximal cycling was assessed with an isokinetic ergometer-based testing protocol. The fixed ergometer has different task and environmental constraints to riding a track bicycle in the velodrome, which might have affected the results presented in Chapter 6. Therefore, the changes/lack of change in intermuscular coordination patterns identified might have differed if the coordination patterns had been measured during sprint on a track bicycle in the velodrome. However, as highlighted previously there were technical limitations and problems with the on-track data collection method which meant it was not chosen to measure intermuscular coordination in maximal cycling pre and post a strength training intervention.

This programme of research used the theoretical framework of ecological dynamics and Newell's model of constraints to interpret changes in intermuscular coordination and the effects of strength training on intermuscular coordination in maximal cycling. The benefit of using the theoretical framework of ecological dynamics to study coordination is it considers athletes as complex adaptive systems, and how such systems coordinate their actions with events, objects and surfaces in a performance environment. It also incorporates how changing organismic constraints such as changes in muscle size, strength, fatigue with training effect the coordination patterns that will emerge. However, this approach does not identify the neurophysiological mechanisms that underpin these changes in intermuscular coordination with strength training. Carroll and colleagues highlighted the lack of knowledge in this area and encouraged researchers to design specific experiments to investigate these neural adaptations (Carroll et al., 2001; Carroll, Selvanayagam, Riek, & Semmler, 2011). They suggested that some of the adaptations associated with strength training could be regarded as motor learning – resistance training could enhance the effectiveness of intermuscular coordination but the precise nature of these adaptations still needs to be determined (Carroll et al., 2001; Carroll et al., 2011). However, the aim of this programme of research was not to identify the neurophysiological mechanisms that underpin changes in intermuscular coordination.

Surface EMG sensors were chosen to measure muscle activity as they are easy and quick to apply, and are not invasive, which are important considerations when working with elite athletes. However, this meant it was not possible to measure the muscle activity of the deep muscles such as the hip flexors (psoas and iliacus) which are important in maximal cycling (Hug & Dorel, 2009; Raasch et al., 1997). Therefore, the contribution of these muscles to the intermuscular coordination pattern cannot be assessed along with whether their muscle activity was influenced by strength training.

For this programme of research the decision was made not to normalise the EMG signals to a MVC for each muscle, as this method of normalising EMG to an MVC has been shown to be unreliable for between-sessions comparisons. However, it meant it was not possible to determine if the amplitude of the muscle activations changed and therefore if the levels of co-contraction of the muscles crossing a joint altered with strength training.

This programme of research used joint moments and powers in conjunction with EMG activity to investigate intermuscular coordination in maximal cycling. By their very nature joint moments and powers only describe the net moment associated with the many individual muscle actions that cross the joint. Measuring individual muscle forces would provide a more comprehensive analysis of intermuscular coordination. However, it is not possible to measure *in-vivo* muscle forces, and therefore, joint kinetics provide an experimental approximation of muscle forces and moments. To estimate individual muscle forces a computer musculoskeletal simulation would be required. However, this approach was not chosen owing to the limitations when trying to model a complex dynamic system such as an elite athlete, and the difficulty in developing participant-specific musculoskeletal models.

7.3 Recommendations for future research

The effect of one period of strength training on intermuscular coordination in maximal cycling was investigated as part of this programme of research. The findings from this programme of research suggested there was no acute effect of strength training on intermuscular coordination in maximal cycling. However, only one strength training period was considered and no follow-up testing session was carried out. It is possible that effects could be seen if a longer period of training was observed. Therefore, to help increase our understanding of how intermuscular coordination is affected by different training phases and changing constraints, longitudinal research designs are needed to allow the investigation of how sprint cyclists' coordination patterns change throughout a whole season. Also, this programme of research only investigated intermuscular coordination at a pedalling rate of 135 rpm which was chosen as this is a typical pedalling rate during the flying 200 m event in track cycling and within the optimal pedalling rate range for track sprint cyclists (Dorel et al., 2005). However, track sprint cycling requires different types of efforts at a variety of pedalling rates, such as accelerating from a standing start, as required in the team sprint and 500 m or 1000 m time trial events. Track sprint cycling also requires cyclists to pedal across a range of pedalling rates during the high velocity phase of races, as track bicycles have a fixed gear ratio, and hence the pedalling rate changes throughout an effort. It would, therefore, be useful to investigate the effect of strength training on intermuscular coordination on the different types of track sprint cycling efforts.

Further development could be undertaken to improve the sensitivity of the techniques used to measure biomechanical variables that describe intermuscular coordination in maximal cycling when carrying out future studies. Often subtle changes occur in elite and well-trained athletes in response to training interventions. Therefore, improving the sensitivity of the measurement techniques, for example, using more detailed marker sets and models of STA, to reduce the influence of STA and skin marker misplacement on the calculated kinematics and kinetic variables, may improve the reliability of the calculated knee flexion and hip joint variables and reduce the MDDs. Also, the on-track data collection method could be developed to measure more revolutions per track effort to capture the between-revolution variability in maximal cycling. Future development could be undertaken to minimise the testing equipment required to be worn or fixed to the athletes - so removing the need for backpack and cable from force pedal which is fixed to leg. This would mean the testing protocol would be less invasive and therefore, have the potential to allow easy monitoring of coordination patterns during training. It is important to measure coordination in a representative environment when studying coordination under the theoretical framework of ecological dynamics and Newell's model of constraints.

The effects of strength training intervention on participants' muscle strength and properties was quantified using 'gym strength' (defined by 1RM for the squat exercise), which was measured pre and post strength training intervention. However, there was no measurement of individual muscle properties (isolated muscle strength, muscle CSA, muscle architecture and neural drive), joint-level properties (such as isolated joint torques and rate of force development), and intermuscular coordination during the squat exercise. Therefore, it was only possible to speculate on which mechanisms may have increased the 'gym strength' post strength training and how changes/lack of changes in intermuscular coordination during maximal cycling may have related to changes in the individual muscle and joint-level properties. For future studies, it would therefore be useful to measure the individual muscle and joint properties.

The pre and post strength training intervention data in Chapter 6 were compared using a group research design approach. However, an alternative approach would be to analyse the data using an individual participant research design as advocated by James and Bates (James & Bates, 1997; James, Bates, & Dufek, 2003). An individual participant experimental design is one where the individual serves as the unit of study (James & Bates, 1997). Their performance or behaviour is typically evaluated across time or under different conditions, with the participant serving as their own control (James & Bates, 1997). The individual participant approach might identify different responses to the strength training intervention which are obscured by the inter-participant differences in initial maximal cycling coordination patterns, training histories and training programmes during the intervention period.

7.4 Practical applications

This programme of research has several practical applications for coaches and sport scientists working in sports requiring a maximal effort over a short period of time. The study reported in Chapter 3 captured the current 'best practice' on the coaching philosophies of strength training and the transfer of strength training to sport performance for sports that require maximal effort over a short period of time. This information can be used to help inform coaching practice in these sports. The on-track data collection method developed in Chapter 4 has the potential to be a useful tool to help coaches assess pedalling on a track, and in training throughout the season. This could help them to understand how their training programmes affect pedalling technique and could help them to identify any areas of a cyclist's pedalling technique that require improvement and to assess the effectiveness of their coaching interventions to improve the rider's technique. The test-retest reliability of the biomechanical variables measured during maximal cycling including the minimal detectable differences (MDD) were reported in Chapter 5. This information can be used by sport science practitioners and researchers to help understand the practical relevance of a longitudinal interventions, such as changes to bicycle set-up and training programmes on athletes' maximal cycling performance. Chapter 6 highlighted the individual response to gym-based strength training, and how coaches and sport science practitioners need to consider the effects of gym-based strength training on athletes' intermuscular coordination patterns.

7.5 Contributions to knowledge

This programme of research investigated the effect of strength training on intermuscular coordination in maximal cycling and its main contributions to knowledge are:

- Captured the experiential knowledge of elite coaches' (from the sports of track sprint cycling, BMX, sprint kayaking, rowing and sprint running) on their philosophies regarding strength training and the range of factors and ideas believed to affect transfer of strength training to sport performance. They believed the transfer of strength training to sports performance was a complex process, with factors associated with fatigue and coordination having particular importance.
- Identified the key mechanical features associated with maximal cycling from previous research.
- Developed a method to measure biomechanical variables that describe cycling on-track in the velodrome, which identified relatively small differences in movement organisation between sprinting on a velodrome track and on a fixed ergometer.
- Quantified the test-retest reliability of the biomechanical variables that describe maximal cycling. Identifying the MDDs which can be used by researchers and sports science practitioners to help understand the magnitude of change required in the biomechanical variables for a 'real change' to have occurred following an intervention.
- A gym-based strength training intervention increased 'gym strength' and average crank power over a complete revolution during maximal cycling. However, this was not associated with significant changes in any of the key mechanical features of maximal cycling and other biomechanical variables that describe intermuscular coordination in maximal cycling. The findings suggested that the participants might have adopted individual coordination strategies to increase crank power following the strength training intervention.

7.6 Conclusion

In summary, this programme of research identified that coaches consider the role of coordination important in the transfer process of strength training to sports performance as they believed that it was important to maintain an athlete's sport technique (sport-specific coordination and movement patterns) and speed during a strength training period. Typically, the biomechanical variables that describe maximal cycling were found to be reliable. However, some variables such as peak knee flexion moment and maximum hip joint power have lower reliability, indicating that care needs to be taken when using these to evaluate changes in maximal cycling biomechanics owing to interventions such as changes in training or bicycle set-up. When the task constraints were changed from sprinting on a fixed ergometer in the laboratory to sprinting on a track bicycle in the velodrome, different movement and coordination patterns were observed which is accordance with the theoretical framework of ecological dynamics and Newell's model of constraints. This finding implies it is important to undertake biomechanical analyses of movement organisation in elite sports practice in a representative environment. This programme of research demonstrated that a period of gym-based strength training did not alter intermuscular coordination in maximal cycling and the key mechanical features associated maximal cycling. Although the cyclists might adopt individual coordination strategies following the change in their organismic constraints after the strength training intervention as crank power increased.

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9 Appendices

9.1 Appendix A: Equipment options for measuring kinematic and kinetic data of a cyclist on a bicycle in the velodrome

Table 9.1: Equipment options for measuring kinematic data of a cyclist on a bicycle in the velodrome

Equipment	Product name	Manufacturer	What does the equipment measure?	Data sampling rate	Accuracy	Sensor type and size	Transmission type	Equipment requirements	Quantifying variability	Synchronisation	Set-up	Transfer between bikes and cyclists	Data processing	Problems	Pilot testing / Comments
Passive marker motion capture camera system	Opus 7 + cameras	Qualisys	Measures reflective marker coordinates in 2D and then calculates 3D coordinates in global coordinate system (GCS).	12 MP, 300 Hz (normal mode) 3MP, 1110 Hz (high speed mode)	±1 subpixel	Passive reflective markers. The size depends on size of capture volume and number of cameras (up to 25 m with 19 mm markers and up to 9 m with 4 mm markers)	Cameras daisy-chained to computer.	The cameras need to be wired together. The cameras fitted to tripods which are positioned along the handrail in track centre. Cameras require power – there are sockets in track centre.	Capture volume with the 8 cameras available will mean only 1 to 2 pedal revolutions will be recorded.	EMG can be synchronised with kinematics through Qualisys Track Manager (QTM) and using trigger module to connect Qualisys and Delsys EMG systems. Can synchronise with force pedals by matching pedal angle measured by pedal angle encoder in Sensix force pedals with pedal angle measured by markers on pedal spindle and ankle by Qualisys system.	Camera set-up time 1 hr. Calibration L-frame which defines the origin of the global coordinate system can be placed adjacent to the track at the pursuit line Wave a calibration wand along the track to create the capture volume – require track to be clear for 2 minutes to get good calibration.	Move reflective markers on the bike frame between bikes and fix reflective markers to cyclist.	Use QTM for initial data processing – marker labelling and gap filling and generation of global coordinates. If define cycling AIM model can do automated tracking of markers.	Lighting in the velodrome – if sun overhead can create many ghost markers. Also the reflective surfaces such as handrails require marker masks.	Pilot testing determined with 8 camera's and 19 mm markers can achieve a capture volume of 14.5m along the black line on the back straight of the track. The cameras were set up along the handrail in track centre to capture left side kinematics. The track is open to the public outside GB training sessions so the cameras would need to be taken down and set-up for each session which will take 1.5 hrs in addition to the testing session. Optimised camera locations and properties to limit ghost markers and the need for marker masks.
Electromagnetic tracking system	G4	Polhemus	6DOF position and orientation of sensor	120 Hz	Static accuracy within 1 m of source box, orientation 0.5° RMS and position 2 mm RMS. Drift free. However, metal objects near sensors can interfere with magnetic field.	Sensors lightweight (9.1 g) small cube which are wired to a hub (114 g) which can be worn on a belt. 3 sensors per hub. Would need to attach cables between sensors to cyclists legs.	Wired between sensor and hub. Wireless between hub and computer. In large spaces there is a problem with dropped frames. Would need to hardwire a data logger to hub, which would need to be carried by the cyclist in a backpack.	Source box requires power source. Could use a power gorilla. The source box needs to be attached to the bike. This would require a bracket to be manufactured to connect it to the seat post.	Can record as many pedal revolutions as required.	Not easy to synchronise with EMG and force pedal systems. Source box defines origin of coordinate system of Polhemus. Therefore, would need to measure dimensions to force pedal coordinate system to link the two systems together. To synchronise the system with Delsys EMG would need to hardwire into the trigger unit which would not be possible. Could potential fit Delsys sensor to crank arm so could match crank arm acceleration with crank angle measured by	Attach source box and power source to bike.	Would require source box and power source to be moved between bikes. Fix sensors and 2 hubs to cyclist (maximum of 6 sensors, both legs hip, knee, ankle)	Software outputs position and orientation of each sensor in the source box coordinate system which will need to be converted to pedal coordinate system.	Magnetic field emitted by source box contaminates the EMG signal (Pidcoe, 2001). Can filter to remove contamination to get muscle onset but not so successful if want amplitude of EMG signal. Can apply notch filter at area of most interference. Delsys suggested capturing motion data at higher rate 600 Hz. However, this is not possible with the Polhemus system. Interference from metal on the bike with Polhemus magnetic field.	Pilot testing found unable to measure ankle joint and pedal spindle location with Polhemus sensors owing to magnetic interference from the drivetrain. This system is therefore not suitable for measuring on-bike kinematics.

Equipment	Product name	Manufacturer	What does the equipment measure?	Data sampling rate	Accuracy	Sensor type and size	Transmission type	Equipment requirements	Quantifying variability	Synchronisation	Set-up	Transfer between bikes and cyclists	Data processing	Problems	Pilot testing / Comments
										crank encoder in post processing.					
Instrumented spatial linkage (ISL)		Custom - based on device described in (Martin et al., 2007)	Measures the position of the anterior superior iliac spine (ASIS) in sagittal plane. Can then infer hip joint position by assuming a constant offset from the ASIS that is measured in a static condition.	Digital encoder (2500 per revolution)	Statically the ISL had a mean horizontal error of 0.03 ± 0.21 mm and a mean vertical error of -0.13 ± 0.59 mm when compared to video based motion capture system had a mean horizontal error of 0.30 ± 0.55 mm and a mean vertical error of -0.27 ± 0.60 mm.	The cyclists wear a belt held in place with double sided tape which has a threaded connector centred on the ASIS. The end of the ISL segment is mounted to the connector and loosely held in place with a threaded fastener. The two segment ISL is constructed using aluminium segments, bearings and digital encoders.	The two digital encoders on the mechanical linkage are processed using a digital to analogue converter. This has a cable to computer where voltage represents an angle. To use this system on the track the cable would need to be connected to a data logger attached to bike.	The mechanical linkage would be required to be fixed to the seat post with a data logger.	Can measure as many revolutions as required.	Can be synchronised with force pedals by plugging into the same junction box as used for pedals. Not sure how to synchronise with EMG.	Fix mechanical linkage and data logger to seat post. Secure belt to cyclist and fix linkage to it.	Would need to move linkage and data logger between bikes. Move belt between cyclists.	Simple equation to calculate the ASIS location (end of segment 2 of the mechanical linkage) using known segment lengths of the mechanical linkage and the angle of each segment from the encoders.	Safety: the cyclist would be fixed to metal linkage which is attached to the bike. Therefore, if they crash they will stay attached to the bike potentially leading to injury.	Used in laboratory set-up at University of Utah for inverse dynamics. They used inverse kinematics to calculate knee joint location. Would require designing for use on a track bike and safety concerns with fixing linkage to track bike and the rider.
Inertial measurement system (IMU)	MVN Biomech	Xsens	Each sensor incorporates 3D gyroscopes, 3D accelerometers and 3D magnetometers. 6DOF of body segments estimated by integrating gyroscope data and double integration of accelerometer data with time. Magnetic sensor used to limit drift.	Body suit 240 Hz. Individual sensors 60 Hz.	(Supej, 2010) the movement accuracy was dependent on duration of motion < 10 s high accuracy, > 35 s low accuracy. Drifting in longer trials. (Godwin, Agnew, & Stevenson, 2009) Motion dependent error, questions raised about suitability of sensors when changing direction and during fast movements. (Cockcroft, 2011) evaluates the use of IMU's for measuring road cycling kinematics.	Onesie suit with sensor and wires built in. Problems with getting suit to fit athletes, for example rowers, the suit too small for upper body. Athletes have also found the suit inhibiting. Requires 2 Xbus Masters which contain batteries and hub which emits wireless signal to base computer, these sit in back pockets. Can use individual sensors if only interested in lower limbs. Velcro strapping to fix individual sensors and cabling to power pack (new individual sensors are wireless). (Godwin et al., 2009) found movement occurred between segment and sensor attached by Velcro.	Wireless from hub in suit back pocket to computer. 100 m zone for wireless transmission. Has been used for 100 m sprint on athletics track with computer at 50 m. It has not been used in a circular area (velodrome). (Cockcroft, 2011) used Xsens to measure cycling kinematics on the road with a car following the cyclist (30 m) within wireless range with the MVN laptop.	Only additional item is computer to receive wireless data.	Can measure as many revolutions as required. However, for data capture over 10 s there are problems with sensor drift.	(Godwin et al., 2009) Difficulty aligning Xsens and lab coordination systems. An additional system is required to achieve this. Local Positioning System used in speed skating. Can be synchronised with EMG using Delsys trigger. Would be difficult to synchronise with force pedals system. Would have to try and match foot angle from Xsens with pedal angle from force pedals	Set-up time 15 mins, if need to change suit 30 mins to change wires and sensors. Software requires minimum of height and foot size to scale biomechanical model. Can input measurement distance from sensor to bony landmark. The accuracy of the biomechanical model depends on accurate sensor placement. Calibration poses N-pose, T-pose and the dynamic hand-touch and squat calibration. In (Cockcroft 2011) difficulty in performing squat pose in cycling shoes and as foot sensors attached to shoe could not remove shoes.	No equipment would be required to be fixed to the bike. If cyclists were same size and using suit could transfer between cyclists.	Xsens uses a biomechanical model which is scaled to participant based on anthropometric measurements. Assumes bilateral symmetry. It is unclear what is used for the hip centre. Could use foot sensor to compare to pedal angle from the force pedals to foot angle as foot sensor is attached to the cycling shoe.	The IMU's can suffer significant magnetic interference from the metal on the bike. Haven't seen any research where Xsens used to measure kinematics for input into inverse dynamics calculations. As the Xsens suit is worn by the cyclist and measures segment orientations would be difficult to measure the position of the pedal spindle which is required for inverse dynamics calculation as it is the point of force application.	Pilot testing confirmed the problems with magnetic interference to the foot sensor owing to the metal on the drive train. The system is therefore unsuitable for capturing kinematics of the cyclist on the track.
Velodrome cameras	5MP Camera Board Module	Raspberry PI	Video images of straight in velodrome, from right side only. Would require digitisation of	5 megapixel native resolution sensor-capable of 2592 x 1944	The images are stitched together from cameras that run along the length of the velodrome. Low resolution of images	Passive reflective markers on joint centres.	The images are automatically captured when the cyclists moves through the back straight and saved into	None	Back straight of track so only 2/3 pedal revolutions	The performance analysis system in the velodrome uses Coordinated Universal timing (UTC) from GPS to synchronise	None, system permanently installed in the velodrome. Passive reflective markers required to be fitted	None	May be possible to automate the digitisation of the markers on the video.	Low resolution of the images and low frame rate	Owing to low resolution and frame rate, not suitable for measuring kinematics of sprint cycling.

Equipment	Product name	Manufacturer	What does the equipment measure?	Data sampling rate	Accuracy	Sensor type and size	Transmission type	Equipment requirements	Quantifying variability	Synchronisation	Set-up	Transfer between bikes and cyclists	Data processing	Problems	Pilot testing / Comments
			images to get 2D kinematics.	pixel static images supports 1080p30, 720p60 and 640x480p60/ 90 video camera. Fixed focus. Sampling frequency approx. 10 Hz	and low sampling frequency of the videos		the performance analysis database.			devices in the velodrome. This may not be accurate enough to synchronise with force pedals.	to cyclist to aid digitisation.				

Table 9.2: Equipment options for measuring kinetic data of a cyclist on a bicycle in the velodrome

Equipment	Product name	Manufacturer	What does the equipment measure?	Data sampling rate	Accuracy	Sensor type and size	Transmission type	Equipment requirements	Quantifying variability	Synchronisation	Set-up	Transfer between bikes and cyclists	Data processing	Problems	Pilot testing / Comments
Force pedals	Model: ICS4	Sensix	3 force components (Fx, Fy, Fz) and 3 moment components (Mx, My, Mz) on the pedal. Pedal and crank encoders measure the pedal and crank angle. Or can use motion capture system to measure crank angle to avoid fitting crank encoder to bike.	Maximum 250 Hz	Combined error (linearity and hysteresis) 1% measuring range. Crosstalk between components 1.5<% measuring range.	6-component force-torque sensor in pedal. 6 Wheatstone full-bridges strain gauges which produce 6 voltage outputs which can be converted the pedal force and moments. Each pedal has a cable from it which runs to a junction box connected to a computer (lab set-up). Or the cable can run to junction box which can be connected to a data logger or wireless NIDAQ which transmits data to computer in track centre – these would need to be carried in a backpack. Could either measure crank angle with crank encoder or markers on pedal spindle to calculate crank angle from kinematic data.	Wired to junction box and then can be cabled or wireless transmission to computer.	Data logger or wireless NIDAQ connected to junction box which connects to cable from pedals.	Can measure as many revolutions as required if using crank encoder to measure crank angle. If using motion capture system to measure crank angle limited to number of revolutions that can be captured by the motion capture system (Qualisys system can measure 1.5 to 2 revolutions)	Match pedal angle measured by motion capture system with pedal angle measured by pedal encoder to synchronise the systems. Motion capture system can then be synchronised with the EMG system.	When fit pedal to track bike perform calibration	Move pedal between bikes. Fix cable from pedals to cyclist's legs back into backpack containing junction box and wireless NIDAQ.	Propriety software or can be processed in Matlab.	Not designed for track use, track sprint cyclists wear pedal straps to ensure their feet does not unclip from the pedal during high power efforts. The pedals were not designed for a strap to be fitted so had develop a solution to use a cable tie to fit strap to pedal cleat.	Pilot testing identified that owing to force pedal being wider and deeper than a standard pedal, the right pedal hits the track around the banking, and therefore, only possible to use left pedal. Junction box and wireless NIDAQ need to be carried in a backpack on the rider (protected by foam so in the event of an accident should not injure cyclist). Had problems with maintaining wireless connection between NIDAQ in backpack and base computer in track centre. Therefore, broadcast bespoke Wi-Fi network using velodrome access points for force pedals to improve data transmission.
Force cranks	Factor Power measurement track cranks	bf1 systems	Torque and ineffective force applied to the crank and crank angle - 2D only.	192 Hz	Crank position within ±3°, Force/Torque accuracy ±1% @ 25°C	Force crank	Transmits wirelessly from cranks to Factor logger fitted to the bike	Factor logger ANT. Cranks powered by Li-Ion cell (10 hrs battery life) which is charged via a connector on the front of the cranks takes 3 hrs.	Can measure as many revolutions as required.	With difficultly possibly could use GPS (UTC) or use EMG accelerometer sensor fitted to crank arm to match crank angle.	Calibration - zero load before start trial	Move cranks and factor logger between bikes. Cranks should be compatible with bottom bracket used on Cervelo track bike. This needs to be checked.	Propriety software to process and output data.	Synchronising with other systems and the cranks are 170 mm only and track sprint cyclists typically use 165 mm length cranks.	Not chosen, as only have 170 mm crank length and all the riders use 165 mm. Also would be difficult to move between bikes within a track session.
Pressure insoles	Pedar insole system	Novel	Measure pressure distribution on the insole of the shoe, and centre of pressure. Only measures vertical force.	50 – 100 Hz	Hysteresis <7%, resolution 2.5 or 5 kPa, offset temperature drift < 0.5 kPa/K	Shoe insole is made up of between 85 and 99 sensors depending on insole size. The insoles are 1.9 mm thick.	Options: data stored on an SD card which is part of the power unit or transmitted via USB cable to a computer.	Insoles require power unit and data logger which is worn on the waist. Cable from each insole to power unit which would need to be attached to back of the cyclists legs.	Can measure as many pedal revolutions as required	Synchronisation can be done through using a trigger box. Normally the Novel system runs as the dominant system so it triggers the EMG and motion capture system. This would not work if needs to be a	Need to choose correct insole size that fits in the cyclists shoes. Then attach battery pack to waist and attach cables to the back of the legs. Calibration lift foot so no load	Move insoles and battery pack between cyclists.	Propriety software to process data which outputs time series centre of pressure (x and y coordinates).	Only measures vertical forces and inverse dynamics calculations require horizontal and vertical forces.	Only measures vertical forces and inverse dynamics calculations require horizontal and vertical forces. Therefore the system is not suitable for

Equipment	Product name	Manufacturer	What does the equipment measure?	Data sampling rate	Accuracy	Sensor type and size	Transmission type	Equipment requirements	Quantifying variability	Synchronisation	Set-up	Transfer between bikes and cyclists	Data processing	Problems	Pilot testing / Comments
										hardwired connection as insoles are on cyclist, who is moving around the track and the motion capture system is beside the track.	condition before the start of each trial.				measuring kinetics of sprint cycling.

9.2 Appendix B: Torso angle of track sprint cyclists during maximal cycling

Introduction

Many studies into cycling kinematics and kinetics set-up the cycling ergometer so the participants adopt an upright cycling position - Wilkinson and colleagues used a standardised torso angle of 70° (Wilkinson et al., 2019) and Umberger and Martin (2001) used a torso angle of approximately 35-44°. The cycling position can influence the pedal forces and muscle activity (Dorel et al., 2009), and therefore when comparing the biomechanics results to other studies the position of the upper body needs to be considered. Therefore, the aim of this study was to measure the torso angle of track sprint cyclists during maximal cycling with the ergometer set-up to match their track bicycle position.

Methods

The participants were 22 track sprint cyclists, twelve of which were part of the study in Chapter 6, and the remainder from a subsequent study not included in this thesis. This data was collected alongside the pelvic tilt data in Appendix 9.4. Reflective markers were placed onto the left acromion (AC) and greater trochanter (GT). The markers were recorded, tracked and processed using the methods described in section 6.2. The torso angle was calculated as the angle between the horizontal and a line connecting the AC and GT (Wilkinson et al., 2019). Each sprint lasted for 4 s providing six complete crank revolutions which were resampled to 100 data points around the crank cycle. The torso angle was averaged over these revolutions to obtain a single ensemble-averaged time series for each sprint for each participant. The mean torso angle was calculated by averaging the time series torso angle data for a single sprint from all the participants.

Results

The mean torso angle was 15.4 ± 1.7° (Figure 9.1).

Discussion

The cycling ergometer was set-up to match each participant's track bicycle position which resulted in the participants riding with a shallow torso angle 15.4 ± 1.7°, evidencing that track sprint cyclists adopt a position to minimise aerodynamic drag.

This differs from the cycling ergometer set-up used in other studies where a more upright cycling position is adopted (Umberger & Martin, 2001; Wilkinson et al., 2019).

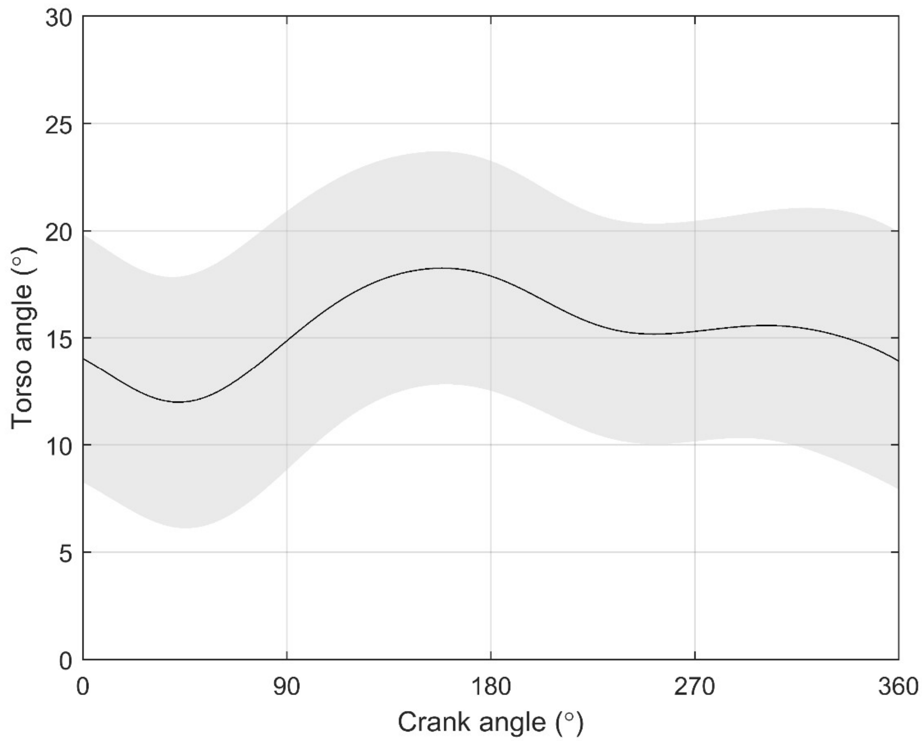

Figure 9.1: Mean torso angle during sprints at 135 rpm

9.3 Appendix C: Interview guide

Introduction

1. Name
2. What is your role?
3. How did you get into cycling?
4. How many years have you worked as a coach/ participated in the sport?

Main questions

5. What is your coaching philosophy?
6. What are the key factors and attributes for developing a world class sprint cyclist?
7. How do you design the athletes' training programme?
   a. How do the athletes training programme change throughout the season?

      (Probe: What are the reasons behind the changes in training programmes)

   b. Do the athletes have a taper before competitions?

      (Probe: Is there a difference in taper length between bike and gym-based strength training?)
      (Probe: Does the taper depend on the importance of the competition?)
      (Probe: What are the reasons behind the taper?)

8. How do you develop strength in sprint cyclists?

   (Probe: Gym training, alternatives?)
   (Probe: Details of strength training; how programmed, what types of exercises are used in strength training in the gym?)
   (Probe: How do you measure success / improvement of strength training?)

9. How do you think gym-based strength training influences an athlete's performance on the bike? Is there an adaption period?

   (Probe: Do you think there is a positive / negative impact?)
   (Probe: Any examples, anecdotal evidence?)

10. How do you develop leg speed?
11. Are there any areas you would like investigating/researching around strength training, and how it transfers to on-bike performance?
12. Is there anything else you would like to add?

9.4 Appendix D: Pelvic tilt during maximal cycling

Introduction

Neptune and Hull (1995) compared the accuracy of different methods to measure the hip joint centre in submaximal cycling. Their study determined the most accurate method for measuring the hip joint centre was where a vector between a marker on the anterior superior iliac crest (ASIS) and the greater trochanter (GT) is measured during a static trial (Neptune & Hull, 1995). This offset is assumed to be constant during the dynamic trials. However, this method assumes there is no rotation of the pelvis in the sagittal plane. This assumption is reasonable for seated submaximal cycling where small changes in pelvic tilt (2° at 200 W) have been measured (Bini et al., 2016). However, no values of pelvic tilt during seated maximal cycling have been reported in the literature to determine if this assumption is valid for maximal cycling. Therefore, the aim of this study was to measure pelvic tilt during maximal sprint trials at a pedalling rate of 135 rpm.

Methods

The participants were 22 track sprint cyclists, twelve of which were part of the study in Chapter 6, and the remainder from a subsequent study not included in this thesis. Reflective markers were placed onto the anatomical bony landmarks of the pelvis: the left anterior superior iliac spine (ASIS), the posterior superior iliac spine (PSIS) and the iliac crest (IC). The markers were recorded, tracked and processed using the methods described in section 6.2. The pelvic tilt was calculated as the angle between the horizontal and a line connecting the PSIS and the ASIS (Preece et al., 2008). During the cycling sprints the ASIS marker was often obscured by the thigh around top dead centre (TDC), and therefore to enable the position of the ASIS to be calculated throughout a trial the vector between the IC and ASIS marker was required. The vector between the marker on the IC and the ASIS was measured during a static trial where the left crank was fixed at 90° from TDC – this vector was assumed constant during the sprints. The pelvic tilt was calculated at each time point during the 4 s sprint at 135 rpm (Figure 9.2). Figure 9.2 illustrates the comparison of the pelvic tilt angle calculated either by using the tracked ASIS marker position (for this participant the ASIS was not obscured by the thigh), or by using the ASIS marker position calculated from the vector offset from the

IC measured during the static trial. The calculated pelvic tilt angles from the two methods are very similar.

Each sprint lasted for 4 s providing six complete crank revolutions which were resampled to 100 data points around the crank cycle. The pelvic tilt was averaged over these revolutions to obtain a single ensemble-averaged time series for each sprint for each participant. The range of pelvic tilt for a sprint was calculated from the difference between the maximum and minimum pelvic tilt values from the ensemble-averaged pelvic time series data for the sprint for the participant. The group mean pelvic tilt was calculated by averaging the time series pelvic tilt data for a single sprint from all the participants.

Results

The pelvic tilt for two example participants throughout a sprint at 135 rpm are shown in Figure 9.2 and Figure 9.3. The mean pelvic tilt value over the crank cycle for all participants is shown in Figure 9.4. The mean pelvic tilt range was 4.6 ± 1.3°.

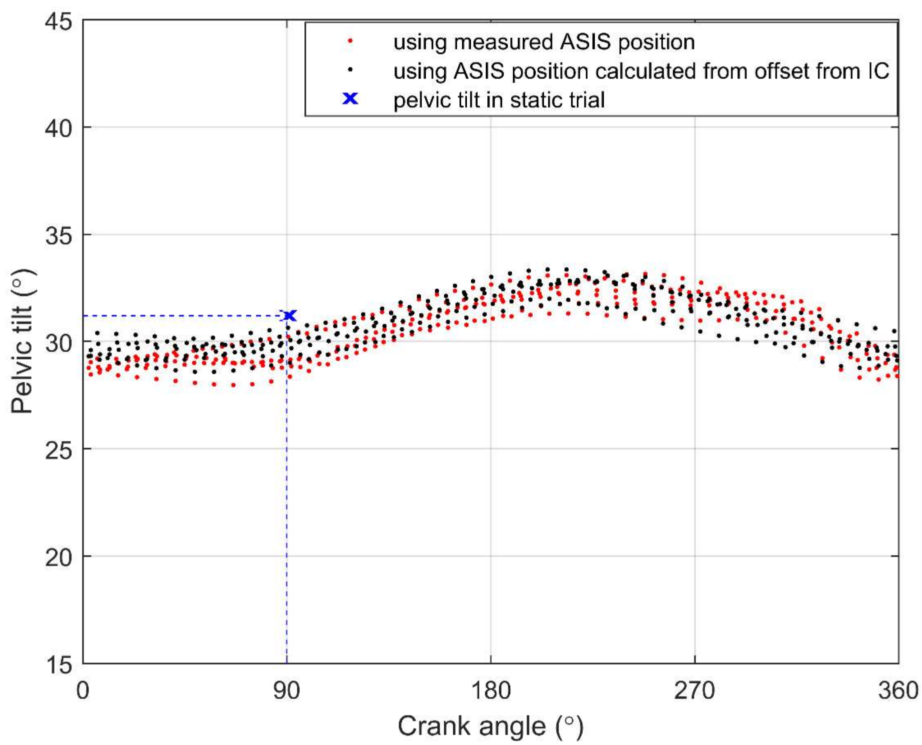

Figure 9.2: Pelvic tilt for a single participant during a single sprint at 135 rpm

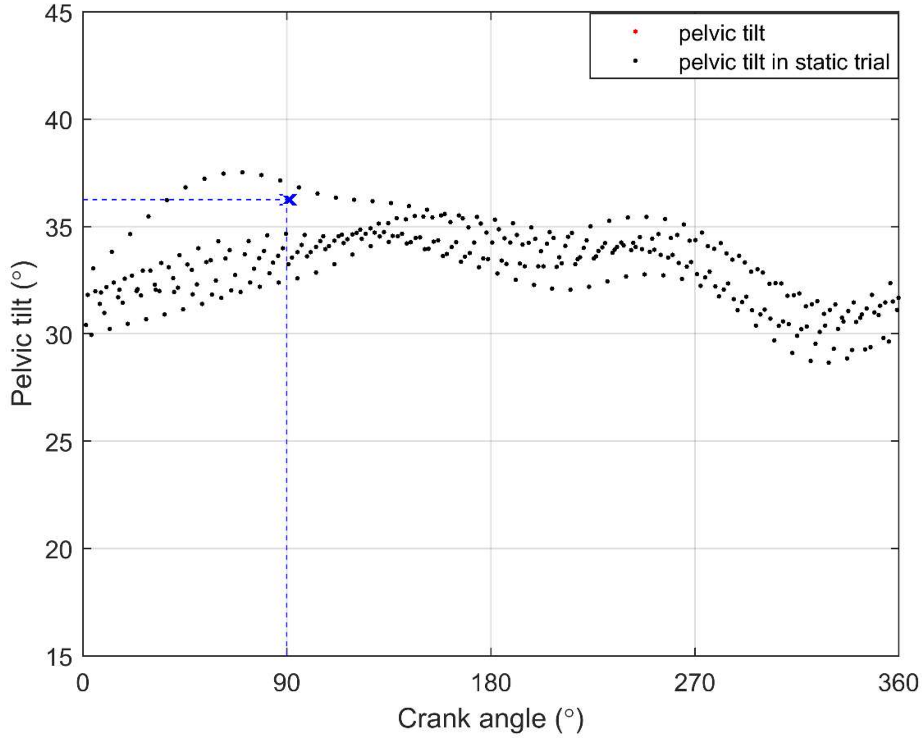

Figure 9.3: Pelvic tilt for a single participant during a single sprint at 135 rpm

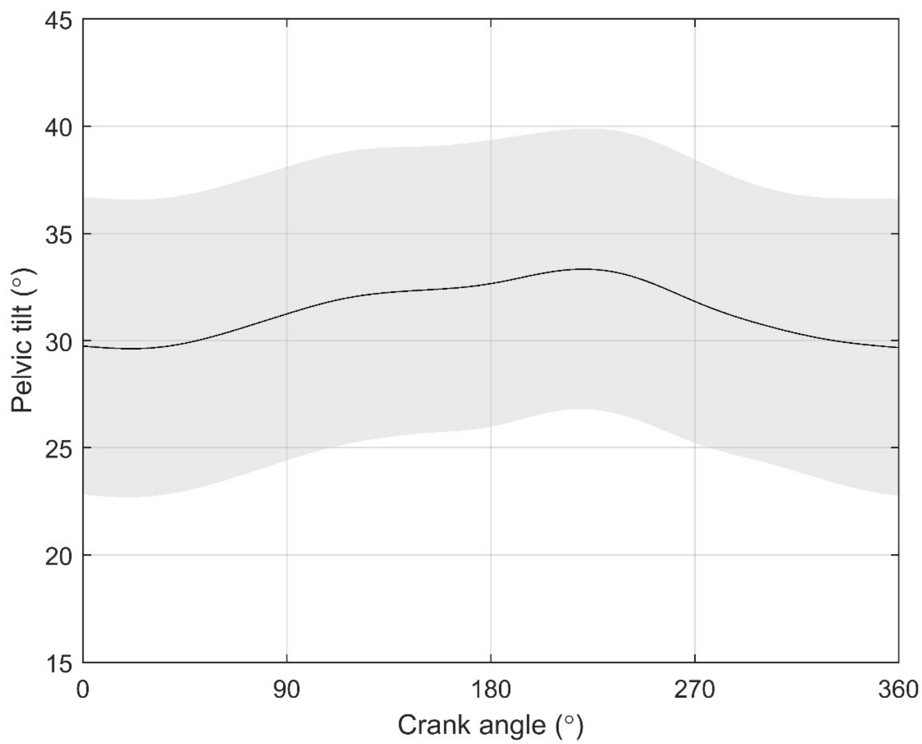

Figure 9.4: Mean pelvic tilt during sprints at 135 rpm

Discussion

The mean pelvic tilt range was measured as 4.6 ± 1.3°, which, therefore, violates the assumption of the constant offset method to define the hip joint that the pelvis does not rotate in the sagittal plane during cycling.

Limitations

The ASIS marker was attached to the front of the cyclists' shorts. Track sprint cyclists adopt an aerodynamic cycling position with a shallow torso angle (Heil, 2002), (which is evidenced in Appendix 9.2 – a mean torso angle of 15.4 ± 1.7° was measured during the same trials as used in this study). This cycling position causes the pelvis to rotate forwards which meant the ASIS marker hung from the front of the shorts creating a slight offset to the actual bony landmark. This also meant that the ASIS marker could move during the sprints creating some movement artefact. However, for most participants the ASIS marker was obscured during the dynamic trials and therefore a constant offset from the IC marker was used, which such suffers less movement artefact.

9.5 Appendix E: Test-retest reliability of hip joint variables measured during maximal cycling on an ergometer: using the greater trochanter position defined by constant offset from iliac crest marker

Introduction

In this thesis (Chapters 4, 5 and 6) the hip joint centre was defined by a marker placed on the greater trochanter (GT). However, a marker placed on the greater trochanter can experience significant soft tissue artefact (STA) during cycling (Li, et al., 2017). Neptune and Hull (1995) compared the accuracy of different methods to measure the hip joint centre in submaximal cycling. Their study determined the most accurate method for measuring the hip joint centre was where a vector between a marker on the anterior superior iliac crest (ASIS) and the GT was measured during a static trial (Neptune & Hull, 1995). This offset is assumed to be constant during the dynamic trials. They then tracked the position of ASIS marker during dynamic trials which undergoes smaller STA. This method to define the hip joint centre has been used in studies of maximal cycling (Martin et al., 2007; McDaniel et al., 2014). Therefore, the aim of this study was to assess if the test-retest reliability of the hip joint variables (joint angle, angular velocity, moment and power) is more reliable for the method where the hip joint centre (greater trochanter position) is defined by using a constant vector between the iliac crest (IC) and the GT than by a marker placed on the GT for maximal cycling.

Methods

The hip joint centre (greater trochanter position) was defined by a constant vector between the IC and the GT. This is an alternative method to the one used in Chapter 5, where the hip joint centre was defined by the tracked GT marker position during the dynamic trials. The vector between the marker on the IC and the GT was measured during a static trial where the left crank was fixed at 90° from TDC – this vector was assumed constant during the sprints (Barratt, 2014; Neptune & Hull, 1995). The GT position was calculated for the dynamic trials assuming a constant offset from the IC marker. The data in this study was collected during the same trials reported in Chapter 5. The same methods described in Chapter 5 were then used to process and analyse the kinetic and kinematic data and to calculate the hip joint variables and assess test-retest reliability.

Results

The test-retest reliability of the hip joint variables where the hip joint centre (greater trochanter position) was defined by a constant vector between the IC and the GT are presented in Figure 9.5.

Discussion

The test-rest reliability for the hip joint variables (joint angle, angular velocity, moment and power) were very similar to the method of defining the hip joint centre by tracking a marker on the GT (which was assessed in Chapter 5). The adjusted coefficient of multiple correlation between-sessions for the hip joint moment was 0.966 for both methods of defining the hip joint centre. Therefore, it was decided for this programme of research to define the hip joint centre by the location of the GT marker, as the constant offset method did not improve the between-sessions test-retest reliability. Also during maximal cycling the pelvis rotates in the sagittal plane (as evidenced in Appendix 9.4), violating the assumption on which this method is based.

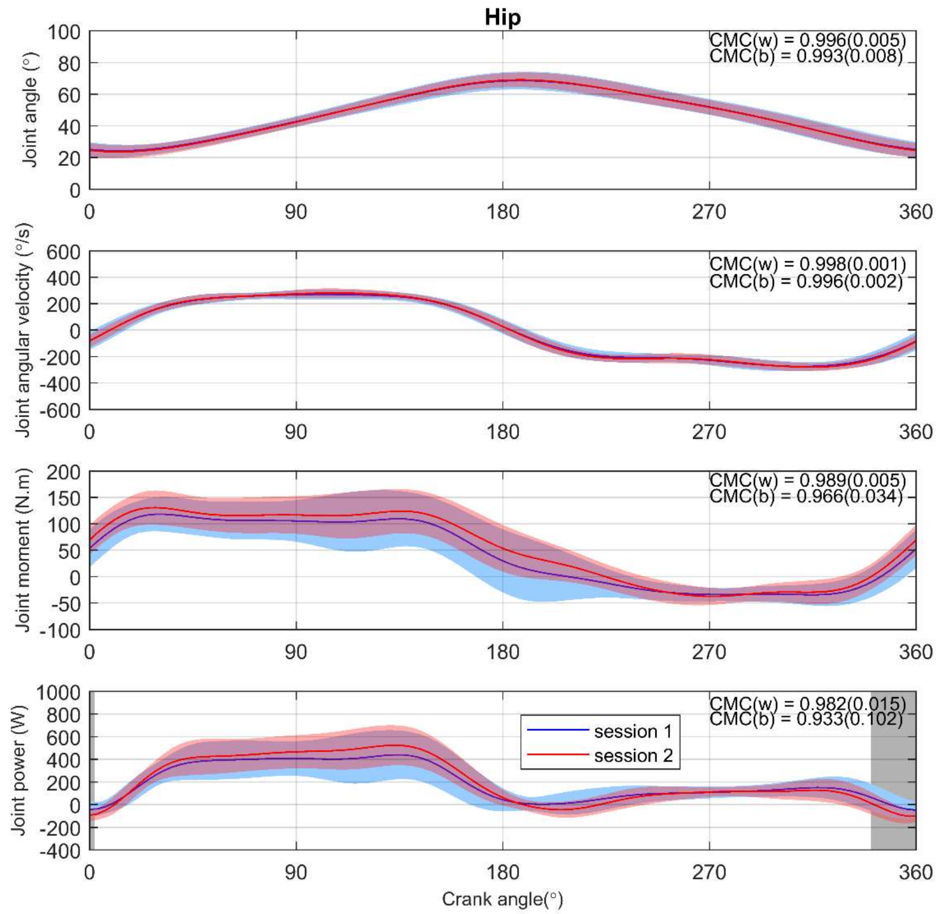

Figure 9.5: Hip joint angles, angular velocities, moments and powers: group means for session one and two.

Hip joint centre (greater trochanter position) is defined by a constant vector between the iliac crest (IC) and the greater trochanter (GT). Areas of the graph shaded grey where the SPM is significant. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b).

For ease of presenting the data the thigh angle and angular velocity are presented as hip angle and angular velocity

9.6 Appendix F: Selecting cut-off frequency for the Butterworth filter used to smooth the kinematic and kinetic data in maximal cycling

Introduction

Typically, residual analysis (Winter, 2009) is used to determine the cut-off frequency of the filter used to process biomechanical kinematic and kinetic data. Bezodis and colleagues recommend using the same cut-off frequency for the kinematic and kinetic data to avoid data processing artefacts in the calculated joint moments (Bezodis et al., 2013). Therefore, the aim of the study was to select the cut-off frequency for Butterworth fourth order (zero-lag) low pass filter for kinematic and kinetic data during maximal cycling at a pedalling rate of 135 rpm.

Methods

Residual analysis was carried out to select the cut-off frequency of the Butterworth fourth order (zero-lag) low pass filter for the crank force and marker coordinate data for sprints at 135 rpm (Winter, 2009). The crank force and marker coordinate data were from the test-retest study (Chapter 5). The maximum residual for the effective and ineffective crank forces, and all the marker coordinates at the chosen cut-off frequency of 14 Hz was calculated for each sprint for each participant (refer to Figure 9.6 and Figure 9.7 for an illustration of the maximum residual). The mean value of the maximum residual for crank forces and marker coordinates for each session for each participant were calculated. From these values the group mean for each session were calculated.

Results

Figure 9.6 and Figure 9.7 illustrate the residual analysis for the crank forces and marker trajectories for a single sprint at 135 rpm for an example participant. The mean residual for the crank forces and marker coordinates for a Butterworth cut-off frequency of 14 Hz was very similar for both sessions (Table 9.3).

Table 9.3: Mean residual for a Butterworth filter cut-off frequency of 14 Hz for crank forces and marker coordinates for sessions 1 and 2

Variable	Units	Mean (SD)	
		Session 1	Session 2
Crank forces	N	9.52 ± 2.97	9.51 ± 2.97
Marker coordinates	m	0.00074 ± 0.00023	0.00074 ± 0.00021

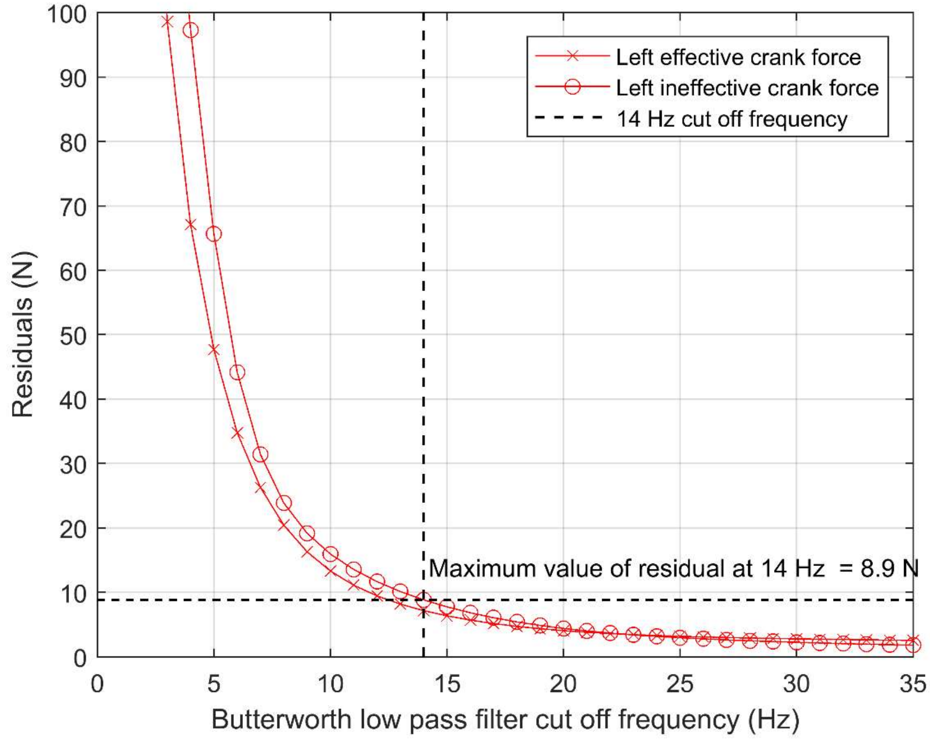

Figure 9.6: Residual analysis for crank force data for a sprint at 135 rpm for an example participant

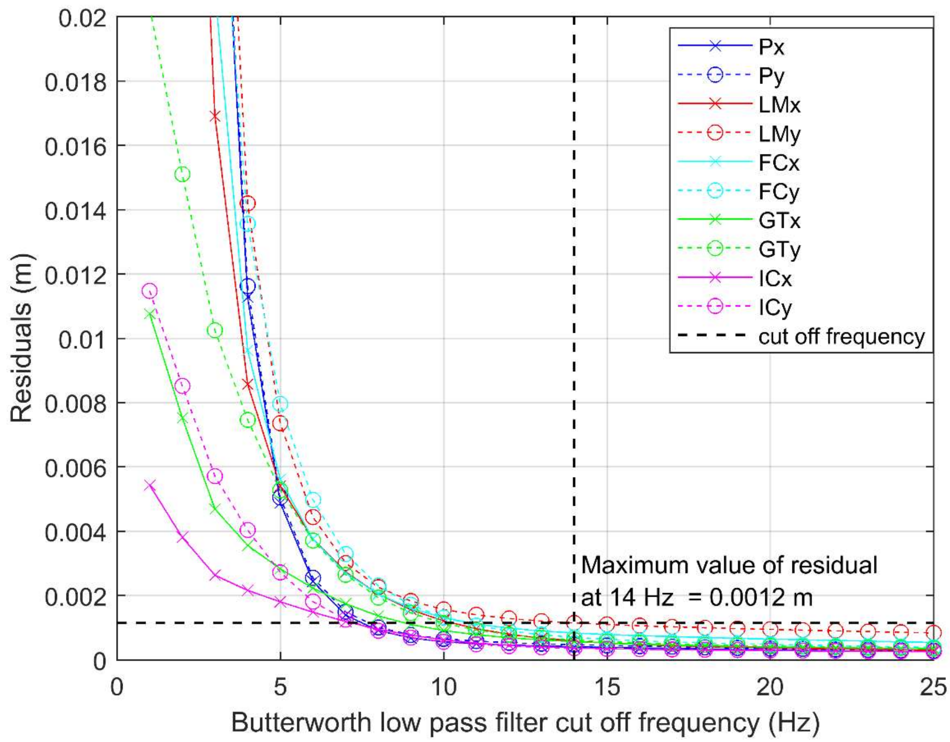

Figure 9.7: Residual analysis for marker coordinate data for a sprint at 135 rpm for an example participant
P = pedal spindle, LM = lateral malleolus, FC = femoral condyle, GT = greater trochanter, IC = iliac crest, x denotes horizontal coordinate, y denotes vertical coordinate.

Discussion

The same cut-off frequency was chosen for the kinematic and kinetic data as recommended by Bezodis and colleagues to avoid data processing artefacts in the calculated joint moments (Bezodis et al., 2013). Therefore, a cut-off frequency of 14 Hz was chosen as this best balanced over smoothing the crank force data and under smoothing the marker trajectory data (Figure 9.6, Figure 9.7). This cut-off frequency was used to process all kinematic and kinetic data for maximal cycling sprints at 135 rpm in this programme of research (Chapters 4, 5 and 6). If inverse dynamic calculations were not being carried out and just kinematic data was being used potentially a lower cut-off frequency of 12 Hz could have been used. However, this would over-smooth the crank force data.

9.7 Appendix G: Comparison of the test-retest reliability of EMG normalisation methods for maximal cycling

Introduction

To allow EMG data to be compared between participants, different muscles, different test conditions and different testing sessions, EMG data needs to be normalised to a reference value (Burden, 2010; Mathiassen et al., 1995). However, currently there is no agreement between researchers on what is the best normalisation procedure to use (Burden & Bartlett, 1999; Hug, 2011). Several cycling specific EMG normalisation methods have been developed these include: normalising EMG activity to on-bicycle static isometric maximum voluntary contractions (MVC's) (Hunter et al., 2002; Kordi et al., 2019) and normalising EMG activity to the maximum EMG activity during a maximal sprint on a bicycle (Albertus-Kajee et al., 2010; Rouffet & Hautier, 2008). However, as this programme of research investigated intermuscular coordination in maximal cycling, using this method would mean the EMG signal would be normalised by itself, and would therefore, be the peak dynamic method (Brochner Nielsen et al., 2018; Ryan & Gregor, 1992).

Dorel and colleagues used a combination of isometric and isokinetic MVCs performed on a dynamometer to obtain maximum muscle activity, which could be used to normalise EMG activity during sprint cycling (Dorel et al., 2012). However, this method has several limitations: it is a time-consuming process, meaning it is ethically not practical for use with elite athletes who have limited time available for testing sessions. In addition, performing multiple MVCs on a dynamometer induces fatigue owing to the number of maximal muscle contractions required. This could influence the participants performance and cycling biomechanics in the subsequent cycling sprints required by the experimental protocol. Therefore, for these reasons this method was not investigated as a possible solution to normalise the EMG activity.

The aim was to assess if a simple method to obtain on-bicycle isometric MVCs (performed with the left crank fixed at 90°) was reliable to normalise EMG activity during maximal cycling, and then to compare this method to the peak and mean dynamic normalisation methods.

Methods

Fourteen track sprint cyclists (the same participants as Chapter 5) performed on bicycle isometric MVCs on a custom-made cycling ergometer (BAE Systems, London, UK) that could be adjusted to make it isometric by fixing the cast iron flywheel - this was the same ergometer as used in (Kordi et al., 2019). The ergometer was set up to replicate each participant's track bicycle position and the left crank was fixed at 90° from TDC. The EMG sensors were attached to the participant as described in section 5.2. Following their typical warm-up (described in section 5.2), the participants were asked to perform 3 x 3 s isometric MVCs on the ergometer with 20 s recovery between efforts. The participants were instructed to push down as hard as possible with the left leg whilst remaining seated and holding onto the dropped handlebars. The EMG signals for all muscles were recorded continuously throughout the MVCs. The raw EMG signals for the MVC efforts were root mean squared (RMS, 200 ms window) (Kordi et al., 2019) and the peak value in rms EMG signal for each muscle was taken as the isometric MVC.

The EMG signals measured during the sprint trials (Chapter 5) were normalised to the peak isometric MVC for each muscle. Also, the EMG signals during the sprints were normalised to the peak value in the linear envelope for each muscle (peak dynamic method) as an alternative normalisation method.

Adjusted coefficient of multiple correlation (CMC) for within- and between-session and cross-correlation coefficient (R) were calculated for the normalised (peak and on-bicycle isometric MVC) EMG linear envelopes using the methods described in section 5.2.

The method and results for the EMG signals normalised to the mean value in the signal are presented in Chapter 5.

Results

When the EMG activity for each muscle was normalised to the peak on-bicycle isometric MVCs there was a large range in the maximum values of the linear envelope for the muscles - ranging from 100% of the isometric MVC for the GMAX to 600% of the isometric MVC for the TA muscle (Figure 9.8, Figure 9.9). There was also high inter-participant variability in maximum values (% of MVC) of the linear envelope for each muscle (Figure 9.8, Figure 9.9). CMC values for EMG linear envelopes ranged between 0.963 to 0.982, and 0.879 to 0.924, for within- and between-session respectively. The TA, BF and ST muscles demonstrated the lowest reliability for EMG activity, and the VL and VM muscles the highest reliability (Figure 9.8, Figure 9.9).

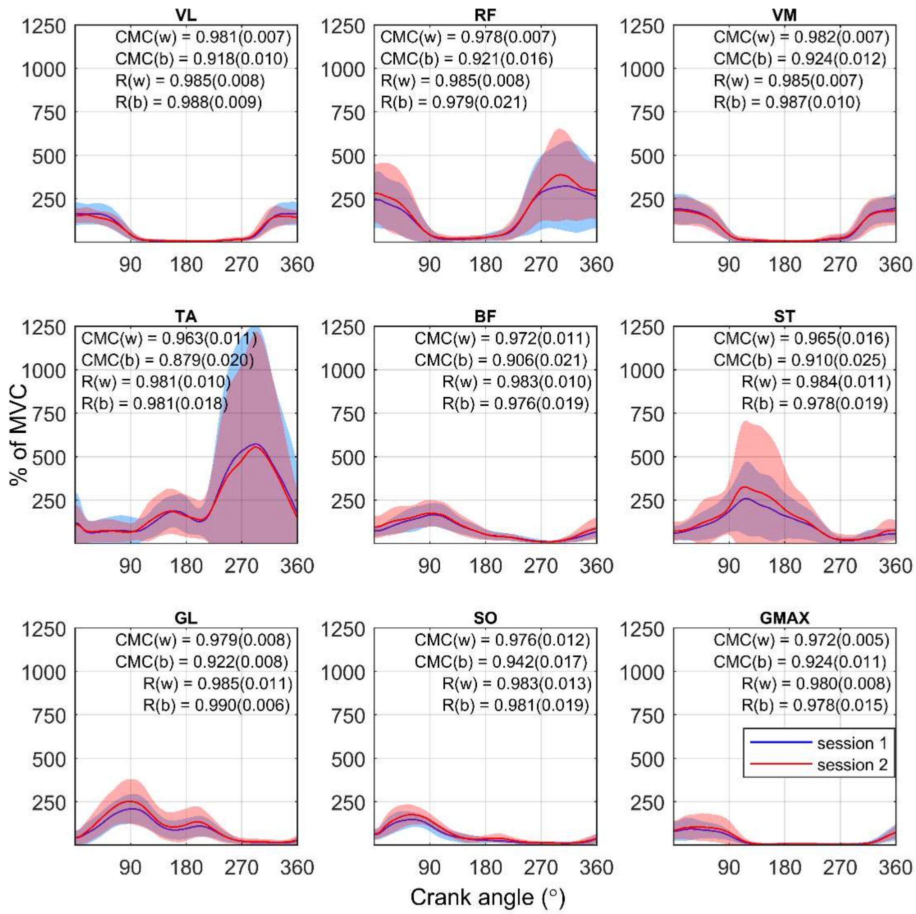

Figure 9.8: EMG linear envelopes (normalised to isometric on-bicycle MVC) for each muscle: group means for session one and two.

VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF=biceps femoris, ST= semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b). Mean and standard deviation of cross-correlation coefficient (R) within-session (w) and between-sessions (b).

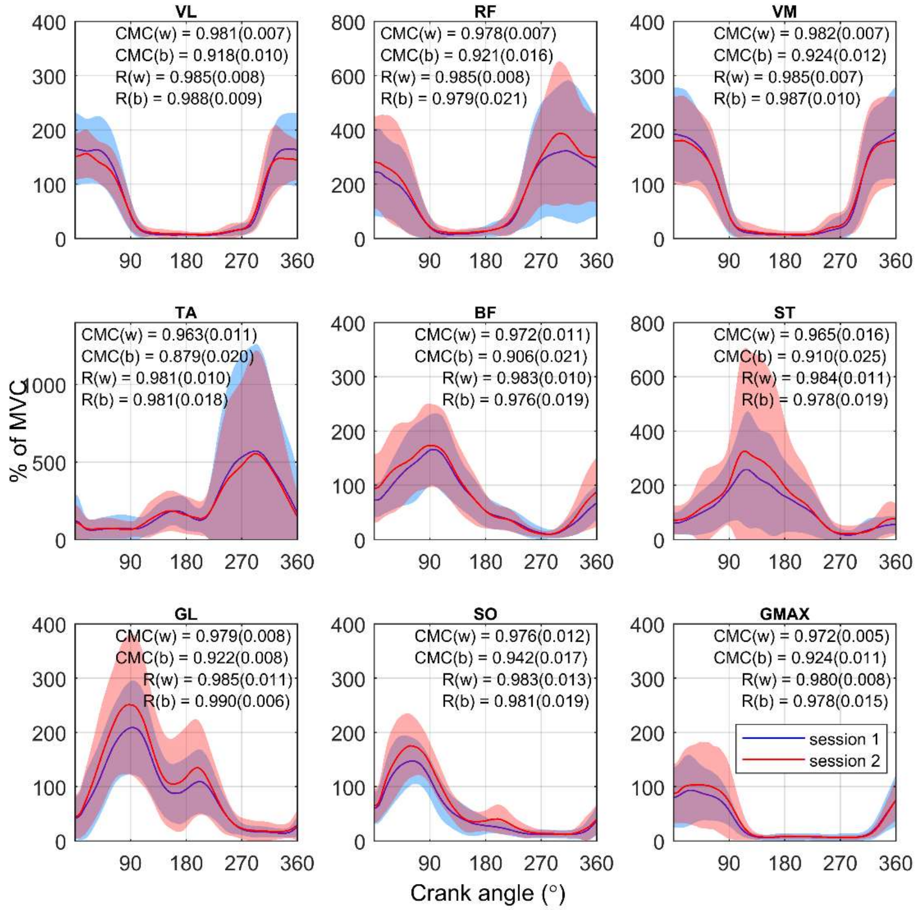

Figure 9.9: EMG linear envelopes (normalised to isometric on-bicycle MVC) for each muscle: group means for session one and two. Same as Figure 9.8 but with varied y scale on subplots for clarity.
VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF=biceps femoris, ST= semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b). Mean and standard deviation of cross-correlation coefficient (R) within-session (w) and between-sessions (b).

EMG linear envelope normalised to the peak value in the signal demonstrated high within- and between-session reliability (Figure 9.10). CMC values for EMG linear envelopes ranged between 0.966 to 0.982, and 0.950 to 0.978, for within- and between-session respectively. The TA, BF and ST muscles demonstrated the lowest reliability for EMG activity, and the VL and VM muscles the highest reliability (Figure 9.10).

SPM indicated a significant difference ($P < 0.05$) between-sessions for the GL muscle between 216° and 217° of the crank cycle for EMG activity normalised to the peak value in the signal (Figure 9.10). This is unlikely to be a meaningful difference as it is less than 0.3% of the crank cycle.

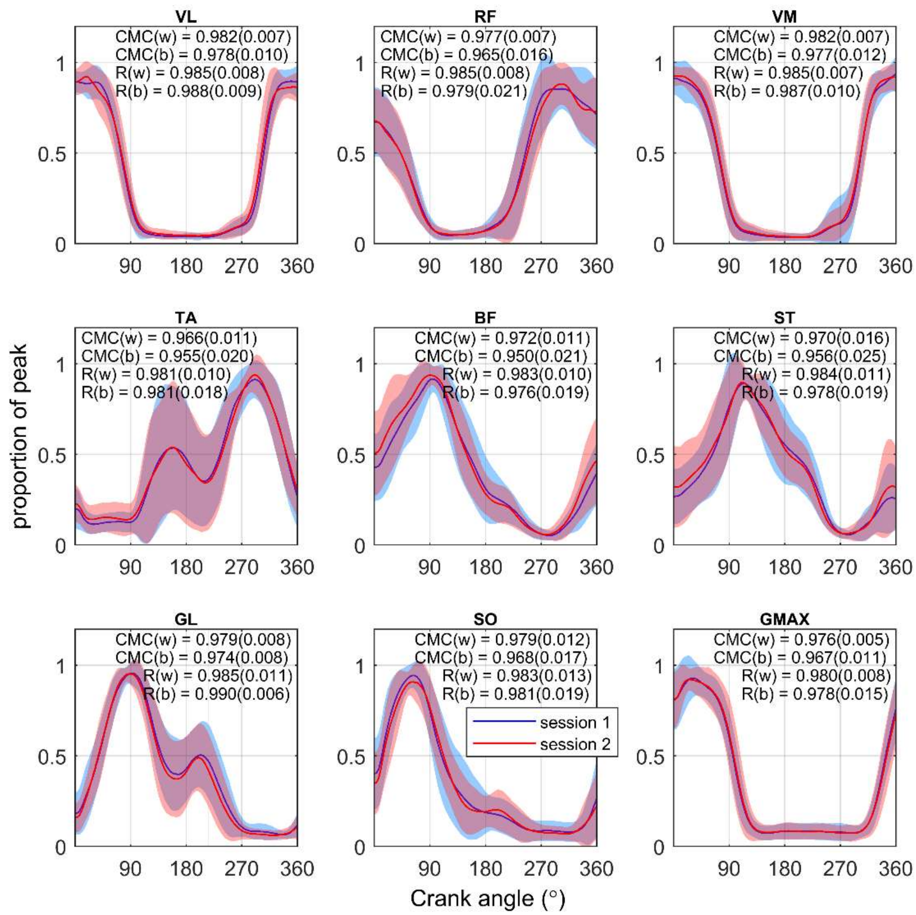

Figure 9.10: EMG linear envelopes (normalised to peak value in signal) for each muscle: group means for session one and two.

VL = vastus lateralis, RF = rectus femoris, VM = vastus medialis, TA = tibialis anterior, BF=biceps femoris, ST= semitendinosus, GL = gastrocnemius lateralis, SO = soleus, GMAX = gluteus maximus. Mean and standard deviation of adjusted coefficient of multiple correlation (CMC), within-session (w) and between-sessions (b). Mean and standard deviation of cross-correlation coefficient (R) within-session (w) and between-sessions (b). Areas of the graph shaded grey where the SPM is significant.

Discussion

The on-bicycle isometric MVC EMG normalisation method demonstrated lower reliability than the peak or mean dynamic method (Chapter 5: Figure 5.4, Figure 9.8, Figure 9.9 and Figure 9.10). When the EMG activity during maximal cycling was normalised to the on-bicycle isometric MVCs there was a large range in % of MVC activity for the muscles (ranging from 100% of the isometric MVC for the GMAX to 600% of the isometric MVC for TA muscle). This demonstrated that the on-bicycle isometric MVC was not successful at eliciting maximum muscle activation from certain muscles. There was also high inter-participant variability in the muscle activity achieved in the on-bicycle isometric MVC in comparison to the sprint. These results support the conclusions of Kordi and co-workers that on-bicycle isometric MVC's do not improve the between-sessions reliability of muscle activity measured during sprint cycling (Kordi et al., 2019). Therefore, in agreement with Kordi and co-workers the on-bicycle isometric MVC method was deemed unreliable to normalise EMG activity during maximal cycling between-sessions.

The within- and between-session reliability values were slightly higher for the EMG linear envelope normalised to the mean value in the signal (mean dynamic method) compared to the peak value in the signal (peak dynamic method). CMC values for EMG linear envelopes (normalised to mean value in signal) ranged between 0.972 to 0.985, and 0.960 to 0.981, for within- and between-session respectively (Chapter 5, Figure 5.4). Therefore, the EMG normalisation to the mean value in the signal (mean dynamic method) was chosen as the normalisation method for the experimental studies in this programme of research.

9.8 Appendix H: Suitability and reliability of onset and offset timing of muscle activity in maximal cycling

Introduction

To compare intermuscular coordination strategies between participants and conditions often the onset and offset timing of muscle activity are calculated from the EMG signals. Researchers have used a number of different methods to determine the threshold which defines the onset of muscle activity: a number of standard deviations above the baseline values (1, 2 and 3 SD) (Hodges & Bui, 1996; Uliam Kuriki et al., 2011), and a percentage of the peak value (Dorel et al., 2012; Jobson et al., 2013; Konrad, 2005). The EMG signal typically needs to exceed the threshold for a minimum period of time for the muscle to be defined as on. Another method to determine muscle onset and offset is by visual inspection which if done by an experienced researcher in EMG, can be highly repeatable between days (Hodges & Bui, 1996). There is no agreement between researchers on what threshold should be used to determine the onset and offset of the EMG activity. Also it has been demonstrated that the test-retest reliability of onset and onset timings for certain muscles during cycling can be poor (Dorel et al., 2008; Jobson et al., 2013). Therefore, the aim of this study was to investigate the suitability and reliability of onset/offset timings to define bursts of EMG activity during maximal cycling.

Methods

The EMG data from the test-retest study (Chapter 5) were analysed to calculate muscle activity bursts. The data collection included a 'quiet sit' where the participants sat still on the ergometer with the left crank at 90° from TDC for 10 s to obtain a baseline EMG signal for each muscle. Once the EMG signals for the sprints at 135 rpm had been processed to create linear envelope for each muscle for each participant for a session, the muscle activity bursts were determined. The peak EMG rms value in the linear envelope for each muscle was identified. The baseline EMG rms value was calculated as the mean value of the rms EMG (window length 25 ms) in the quiet sit trial. The threshold for determining onset and offset for a burst of EMG activity was defined as the period where the signal was above a threshold of 20% difference between the peak and the baseline rms EMG values (Dorel et al., 2012) – refer to Figure 9.11 for an example of the definition of muscle activity bursts. This definition of onset/offset threshold was chosen as it was used in a study by Dorel and colleagues into muscle activity in maximal cycling (Dorel et al., 2012). For a muscle to be defined as on, the rms EMG value needed to exceed the threshold for a minimum of 25 ms, and conversely for a muscle to be defined as off, the rms EMG value needed to be below the threshold for a minimum of 25 ms (Uliam Kuriki et al., 2011). To assess the sensitivity of the muscle activity bursts to the threshold definition an alternative definition for the onset/offset threshold was defined as the baseline rms EMG plus 20% of the difference between the peak and the baseline rms EMG values. Owing to some of the muscles, such as the TA, changing between one and two bursts of activity between-sessions (Figure 9.13 and Figure 9.14), it was difficult to assess reliability quantitatively, and therefore the reliability of the onset and offset muscle bursts was assessed by visual inspection.

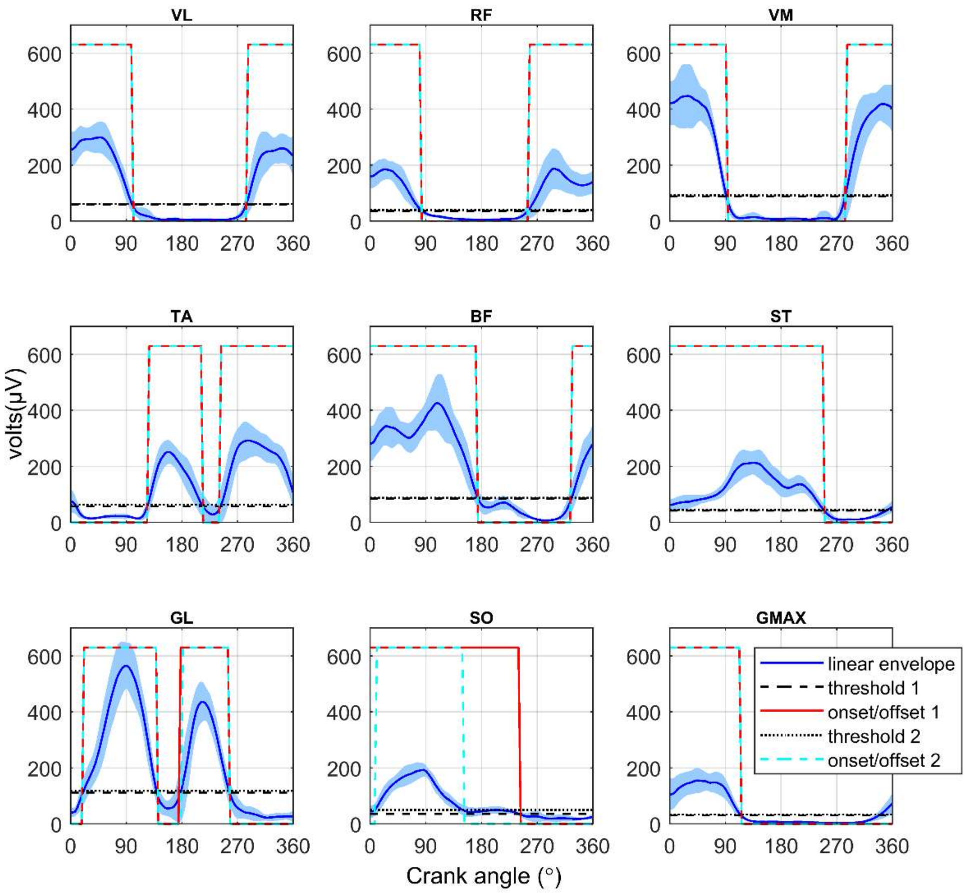

Figure 9.11: Onset/offset of muscle activity for a session for an example participant Threshold 1 defined as 20% of the difference between the peak and the baseline rms EMG. Threshold 2 defined as the baseline rms EMG value plus 20% of the difference between the peak and the baseline rms EMG.

Results and discussion

Figure 9.12 illustrates the muscle activity bursts for the VL muscle with onset and offset for all participants. All participants have one burst of muscle activity in the crank cycle for this muscle. Therefore, it would be possible to calculate the group mean and SD values of VL muscle activity bursts. However, muscles such as the TA can have one or two bursts in a crank cycle (Figure 9.13), although participant 4 had three bursts however, on visual inspection this seems an artefact owing the threshold definition and should be 2 bursts. The inter-participant variability in TA muscle activity bursts has previously been reported in the literature for maximal and submaximal cycling (Dorel et al., 2012; Hug et al., 2008). Muscles such as the GL, SO and TA, which have one or two bursts of muscle activity depending on the participant, create a problem when trying to calculate group mean muscle activity bursts. An approach used by Dorel and colleagues was to discount the second burst of muscle activity as voluntary and therefore, exclude it from the calculation of the group mean EMG activity onset/offset and duration of bursts (Dorel et al., 2012). They, also, for the SO and GL muscles, if the period between the two bursts of muscle activity was less than 15°, they considered it as one global burst of muscle activity to simplify the calculations (Dorel et al., 2012). Jobson and co-workers adopted a similar approach where the first onset of muscle activity was defined as the primary onset and the second offset the primary offset to combine the two bursts of muscle activity to create one burst (Jobson et al., 2013). However, by adopting this approach important features of the intermuscular coordination pattern will be lost.

Another issue when defining onset and offset muscle activity bursts is the definition of the threshold where the muscle activity can be defined as 'on/off'. This threshold is subjective and there is no consensus between researchers on how this should be defined and the choice of threshold can influence the results (Hodges & Bui, 1996; Li, & Caldwell, 1999). This is illustrated by Figure 9.13 and Figure 9.14 which show the TA muscle activity bursts using different threshold definitions – the number of muscle activity bursts changes for participants 1, 3, 4, and 9 depending on which threshold definition used. For the participant's muscle activity shown in Figure 9.11, the length of the muscle activity burst of SO changes - offset is 99° earlier in the crank cycle for threshold definition 2. Both of these results highlight the sensitivity of the onset and offset timings of muscle activity bursts, and number of bursts to the threshold definition. When the muscle activity burst data was visually inspected the test-retest reliability of the muscles, that can have more than one burst of muscle activity in the crank cycle, such as the TA, was low (Figure 9.13, Figure 9.14). As illustrated in Figure 9.13 and Figure 9.14, the TA muscle activity can change from one to two bursts of muscle activity between-sessions. This finding is in agreement with Dorel and colleagues, who found during submaximal cycling that several muscles, including the TA, exhibited low within-session reliability values for onset and offset timings (Dorel et al., 2008). This was supported by a study by Jobson and co-workers who found low between-sessions reliability of onset of the TA (SEM was 45.6° at 150 W), and offset of the SO, GL and RF during submaximal cycling (Jobson et al., 2013).

Based on the exploration of the EMG data it was therefore decided not to use onset and offset muscle activity bursts to study intermuscular coordination in maximal cycling, owing to the challenges and problems identified with this analysis method.

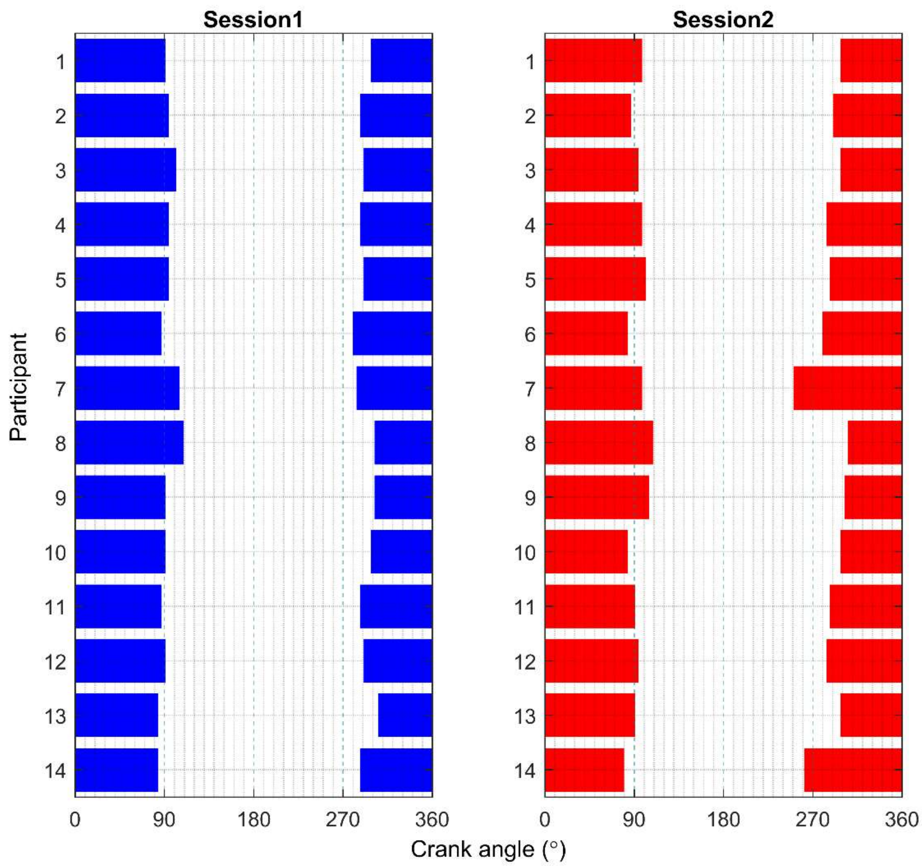

Figure 9.12: Onset and offset and duration of burst of EMG activity for the VL muscle for all participants for sprints at 135 rpm: session 1 and 2
Threshold defined as 20% of the difference between the peak and the baseline rms EMG

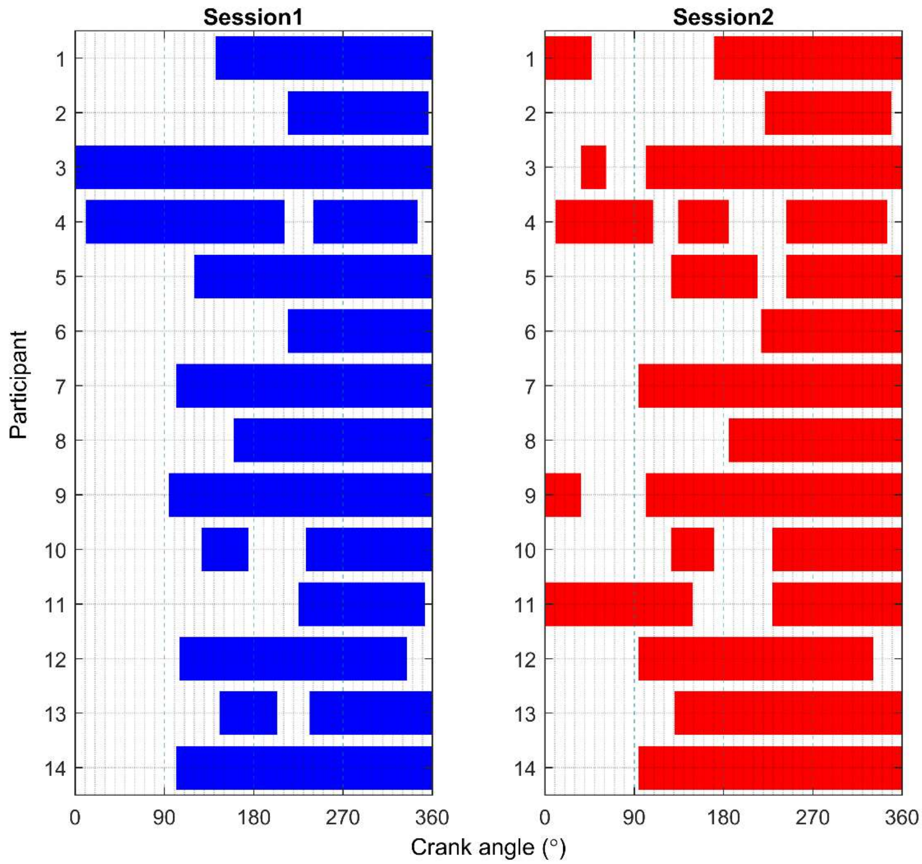

Figure 9.13: Onset and offset and duration of burst of EMG activity for the TA muscle for all participants for sprints at 135 rpm: session 1 and 2

Threshold defined as 20% of the difference between the peak and the baseline rms EMG

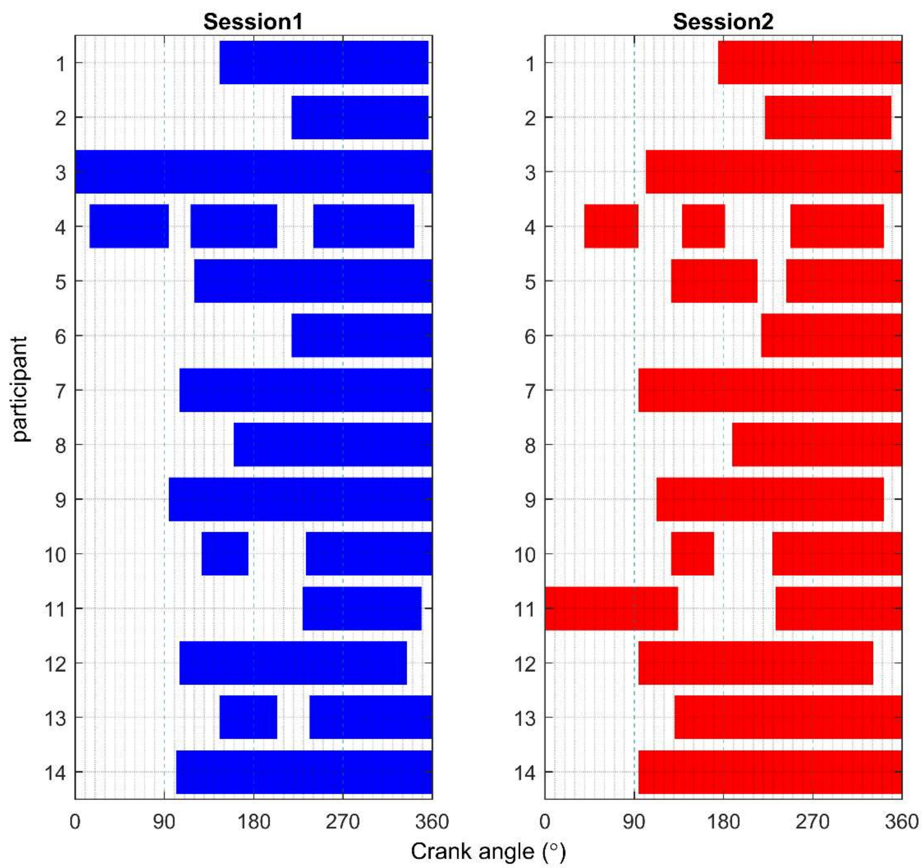

Figure 9.14: Onset and offset and duration of burst of EMG activity for the TA muscle for all participants for sprints at 135 rpm: session 1 and 2

Threshold defined as the baseline rms EMG value plus 20% of the difference between the peak and the baseline rms EMG

9.9 Appendix I: Quantifying hip-ankle moment synergy – detailed vector coding method

To quantify hip-ankle joint coordination and the strength of the hip-ankle joint synergy, a vector coding technique was used (Chang et al., 2008; Hamill et al., 2000; Sparrow et al., 1987). The coupling angle (γ_i) was calculated from the hip-ankle moment diagrams (Figure 9.16) for each point on the crank cycle (the joint moment data had been interpolated to 101 equally spaced data points around the crank cycle) using equations 5, 6 and 7. The coupling angle is defined as the orientation of the vector (relative to the right horizontal) between two adjacent points on the moment-moment plot, Figure 9.15.

$$\gamma_{i,j} = \tan^{-1}\left(\frac{y_{j,i+1}-y_{j,i}}{x_{j,i+1}-x_{j,i}}\right) \quad \text{if } \left(x_{j,i+1}-x_{j,i}\right) > 0 \ \ \& \ \left(y_{j,i+1}-y_{j,i}\right) > 0 \quad (5)$$

$$\gamma_{i,j} = 360 + \tan^{-1}\left(\frac{y_{j,i+1}-y_{j,i}}{x_{j,i+1}-x_{j,i}}\right) \quad \text{if } \left(x_{j,i+1}-x_{j,i}\right) > 0 \ \ \& \ \left(y_{j,i+1}-y_{j,i}\right) < 0 \quad (6)$$

$$\gamma_{i,j} = 180 + \tan^{-1}\left(\frac{y_{j,i+1}-y_{j,i}}{x_{j,i+1}-x_{j,i}}\right) \quad \text{if } \left(x_{j,i+1}-x_{j,i}\right) < 0 \quad (7)$$

where $0° \leq \gamma \leq 360°$, y = hip moment, x = ankle moment, i = each instant during the crank cycle of the jth crank revolution.

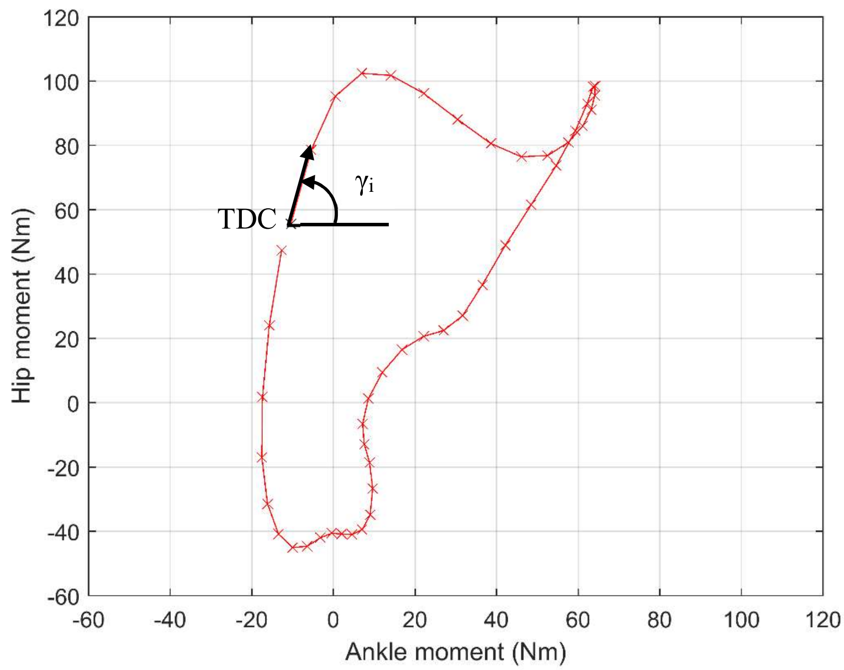

Figure 9.15: An illustration of the calculation for a coupling angle (γ_i) from hip-ankle moment plot (one single revolution during a sprint)

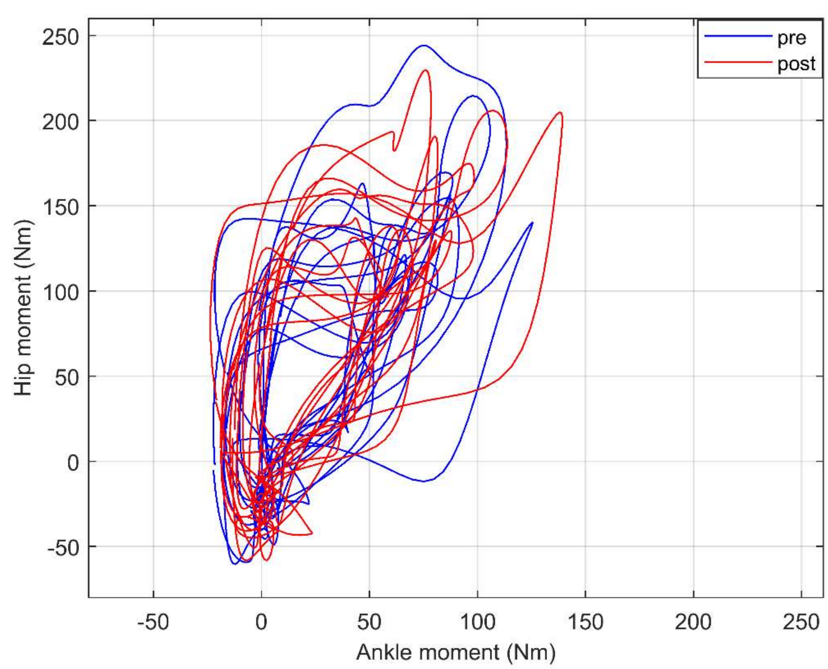

Figure 9.16: Hip-ankle moment plots for sprints at 135 rpm: pre and post strength training intervention

The coupling angle was calculated for each instant of the crank cycle for all revolutions of the sprints at 135 rpm for each participant. Since the coupling angles are directional in nature, the mean coupling angles were computed using circular statistics (Batschelet, 1981). For each participant the mean coupling angle ($\bar{\gamma}_i$) for a session was calculated from the mean horizontal ($\bar{x}_i$) and vertical ($\bar{y}_i$) components at each instant of the crank cycle, using equations 8, 9, and 10:

$$\bar{x}_i = \frac{1}{n}\sum_{j=1}^{n}\left(\cos\gamma_{j,i}\right) \tag{8}$$

$$\bar{y}_i = \frac{1}{n}\sum_{j=1}^{n}\left(\sin\gamma_{j,i}\right) \tag{9}$$

$$\bar{\gamma}_i = \begin{cases} tan^{-1}\left(\frac{\bar{y}_i}{\bar{x}_i}\right) & \text{if } \bar{x}_i > 0 \;\&\; \bar{y}_i > 0 \\ 180 + tan^{-1}\left(\frac{\bar{y}_i}{\bar{x}_i}\right) & \text{if } \bar{x}_i < 0 \\ 360 - tan^{-1}\left(\frac{\bar{y}_i}{\bar{x}_i}\right) & \text{if } \bar{x}_i > 0 \;\&\; \bar{y}_i < 0 \end{cases} \tag{10}$$

The length of the average coupling angle ($\overline{r_i}$) and the coupling angle variability ($CAVi$) was calculated using the equations from (Needham, Naemi, & Chockalingam, 2014):

$$\overline{r_i} = \sqrt{\overline{x_i}^2 + \overline{y_i}^2} \tag{11}$$

$$CAV_i = \sqrt{2.(1 - \overline{r_i})}\frac{180}{\pi} \tag{12}$$

This process was repeated to calculate the group mean coupling angles pre and post strength training intervention (Figure 9.17).

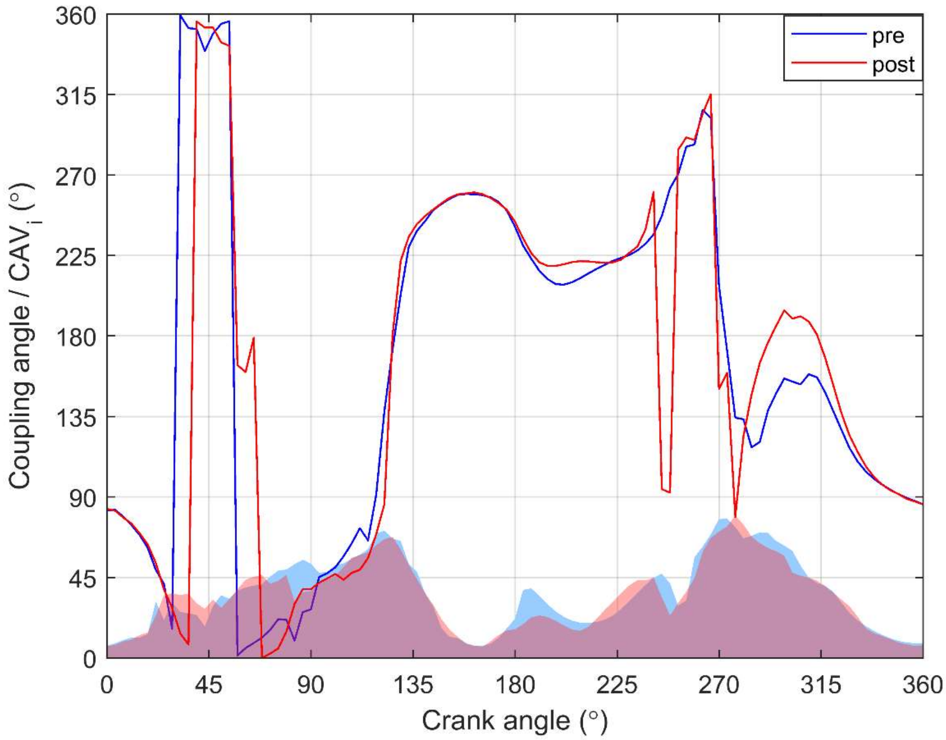

Figure 9.17: Mean coupling angle for hip-ankle moment for sprints at 135 rpm: pre and post strength training intervention

Shaded area represents coupling angle variability (*CAV_i*)

9.10 Appendix J: Details of participants training programmes in intervention period

Table 9.4: Details of participants training sessions and gym exercises in the intervention period

Participant	Training sessions per week					Main gym exercises							
	Track	**Gym**	**Road**	**Turbo or rollers**	**Other** yoga	**Squats** (back, full, half, partial, Bulgarians, front)	**Clean** (full and pull)	**Deadlifts** (normal and Romanian)	**Leg press** (single and double)	**Leg curl**	**Leg extension**	**Calf raises**	**Supplementary exercises**
1	3	3	1			X	X	X	X	X	X	X	SA DB Row, good mornings, bench press, chin ups
2	4	3	1			X	X	X	X	X	X	X	SA DB Row, bench press, chin ups, good mornings
3	4	2	1	1		X	X	X	X	X	X	X	SA DB Row, bench press, walking lunges, chin ups
4	4	3	1			X	X	X	X	X	X	X	SA DB Row, bench press, walking lunges, chin ups
5	2	1	2	2.5		X		X	X				SA DB Row, bench press, walking lunges
6	2.5	3		1	1	X		X					step ups, hip lifts, single arm rows, bench press, box jumps
7	3.5	3	1			X	X	X	X	X	X	X	SA DB Row, bench press, walking lunges, chin ups

Table 9.4: (continued)

Participant	Training sessions per week					Main gym exercises							
	Track	**Gym**	**Road**	**Turbo or rollers**	**Other** yoga	**Squats** (back, full, half, partial, Bulgarians, front)	**Clean** (full and pull)	**Deadlifts** (normal and Romanian)	**Leg press** (single and double)	**Leg curl**	**Leg extension**	**Calf raises**	**Supplementary exercises**
8	3	2	2			X		X	X			X	SA DB Row, bench press, walking lunges
9	2	2	1	3		X		X	X				SA DB Row, bench press, walking lunges
10	2	2	1	3		X		X	X				SA DB Row, bench press, walking lunges
11	1	2		2		X		X	X			X	bench press, bent over row, lat pull down
12	2	1	2	2		X		X	X				
Average number of sessions per week / Number of participants that undertake exercises	**2.8**	**2.3**	**1.3**	**2.1**	**1.0**	**12**	**5**	**12**	**11**	**5**	**5**	**7**	