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Assessment of joint kinetics in elite sprint cyclists

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Assessment of Joint Kinetics in Elite Sprint Cyclists

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In Loving Memory

Fuji Munro

2004-2018

“Blessed”

DEDICATION

To Kenneth George Sinclair Munro.

My dad. Your name on this thesis skips time for me.

In remembrance and gratitude for the life you gave me.

I am so much of you. I am who you made me, and proud that I am like you in so many ways.

I know you are never far and are standing with me through my life.

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I am your daughter also.

You gave me unconditional love, belief and guidance throughout so much of my life.

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When the sun sets and we think about what tomorrow will bring, we understand that better is not we what do. It is who we are.

In every PhD thesis there is a story. For every PhD scholar ultimately realises that the thesis and academic body of work are the smallest, almost least significant part of the tale. We grow in these years through the challenges of life. We are shaped everyday by the hurdles we face, the burdens we bear and the crags of the cliffs we climb every single day til we reach, in this moment, the top of this particular mountain. We glimpse the view briefly before continuing our journey with our deepest learning - that we are resilient beyond all measure and know ourselves to depths greater than ever before. We chose to face this challenge. And in doing so we said with a loud voice to the Universe “I am ready to grow”. In facing ourselves and truly learning about who we are, our PhD years have taught us to be people that can better serve this world.

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ABSTRACT

Sprint cycling requires the production of explosive muscle power outputs up to very high pedalling rates. The ability to assess muscular function through the course of the sprint would aid training practices for high-level performers. Inverse dynamics provides a non-invasive means of estimating the net muscle actions acting across any joint contributing to movement. However, analysis of joint kinetics requires motion-capture techniques that present some unique challenges for cycling. This thesis presents three studies investigating the application of a custom-designed force pedal system to examine the joint kinetics of elite trained track sprint cyclists. To provide the basis for selecting appropriate testing procedures, study one evaluated differences between two- and three- dimensional techniques while assessing joint kinetics of seated and standing sprint cycling at optimal cadence (the cadence where peak power is delivered). Study two examined the impact of cadence and seating position on joint kinetics, while determining testing reliability using the three-dimensional process. Coefficients of variation were established for between- and within- days repetitions of sprint performance at optimal cadence, and cadences 30% lower and 30% higher, in both seated and standing positions. Study three compared joint kinetics of sprint cycling performance with commonly-applied resistance-training exercises in an elite cycling cohort, in order to better understand training specificity. Joint-specific torque-angular velocity relationships were established from seated and standing sprinting at three cadences and the clean exercise at three loads, with other strength-based exercises examined at maximal load only.

Study one determined that flattened projections of the 3D motion into 2D resulted in significant differences in joint powers calculated in the sagittal-plane. When using 2D methods, knee joint power was significantly lower and hip transfer power significantly greater, while hip range of motion was lower and the angle where hip peak power occurred later in the crank cycle. These results indicate that 3D processes should be used where evaluation of absolute values are important, although 2D processes

may still be acceptable where relative differences are being assessed. It was observed in Study two that, while crank and total muscle power upheld a quadratic power-cadence relationship, joint-specific powers were uniquely related to cadence and riding position. Crank and joint-specific optimal cadences for power production were distinctly different. The hip displayed a linear maximum power-cadence relationship in seated but quadratic in standing position, with the reverse observed at the knee. Ankle and hip transfer powers both linearly declined with cadence irrespective riding position. In such a case, joint-specific power contribution, hence distribution of muscular effort, cannot be directly inferred from power assessed at the crank. Reliability was highest for crank and total muscle power, particularly at the riders' optimal cadence. Reliability of joint powers were somewhat lower and uniquely dependent on joint, joint action and trial condition. Results indicate that external power output at the crank is relatively stable across sprints, despite variation in the underlying muscular contributions. Results of study three showed equivalence in the torque-angular velocity relationships at the hip in sprint cycling and different phases of the clean. No such relationship was evident at the knee or ankle. In contrast to the negative linear relationships observed in all other conditions, ankle mechanics in sprinting showed a positive linear relationship highlighting a distinct functional role of this joint. Highest maximal torques at the hip and knee were observed during unilateral single rack pull and step-up exercises, respectively, supporting their efficacy for improving the maximum strength characteristics at these joints.

The results of this thesis indicate that joint kinetics are an effective means of assessing muscular performance in highly-trained track sprint cyclists and provide information on the underlying strategies that could not be assessed through conventional testing of power at the crank. The use of 3D processes is recommended where accuracy of assessment and absolute values are important. Flexibility of 2D processes may be advantageous in field-based settings and may be acceptable where only relative change is of interest. High reliability of 3D testing supports its use in monitoring of athletes, with the reliability data presented in this thesis providing an indication of the smallest meaningful changes in various trial conditions. Low coefficients of variation observed in crank and muscle power terms,

despite greater variation in joint powers, suggest motor control strategies dynamically respond to task conditions while maintaining a consistent external power. Resistance exercises are seen to display joint-specific profiles that characterise relative hip- or knee- dominance. The comparison of these profiles with those of sprint cycling can help inform exercise selection for strength development of elite riders. The ability to monitor changes and target training intervention at joint level provides a unique approach to athlete development. Outcomes of this thesis support the practical application of joint kinetic assessment in aiding training practices to the highest levels of competition in track sprint cycling. Indeed, the equipment, methods and knowledge obtained from this research is currently applied in the preparation of Australia's best sprint cyclists.

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1 Introduction

1.1 Overview

This doctoral thesis presents the results of three applied research studies examining joint kinetics in highly-trained track sprint cyclists during cycling and resistance-training exercises. The overall aim of the thesis is to investigate the application of a custom-designed force pedal system and dynamic modelling process to improve understanding of the biomechanics of expert performers in track sprint cycling, with the wider intention to inform elite training practices for the sport. Study one compared sagittal-plane joint kinetics determined by simple two-dimensional (2D) techniques with those using a three-dimensional (3D) system. Given that cycling movement is predominantly in the sagittal plane, the goal of this study was to determine the extent to which tri-planar assessment processes impact results. Study two examined the effects of cadence and riding position on joint kinetics, while determining the reliability of the 3D testing process. This study aimed to quantify the effects of testing conditions on intra-subject variability. Study three compared the joint kinetics of elite standard track sprint cyclists in sprint testing and in performance of six resistance exercises commonly prescribed in training. The goals of this, final, study was to determine which resistance exercises are most biomechanically comparable to the sprint cycling movement in order to further aid training prescription. Data for the studies were obtained from testing two squads of athletes over a single testing block in each case. Some common methodologies are, therefore, shared between the studies. As an inherent part of this project, a novel system facilitating 3D assessment of cycling biomechanics was engineered and constructed. The constituent parts and use of the system are described within the methodologies as relevant to the execution of the studies. However, full details of the electrical and mechanical componentry are out-with the scope of this thesis and hence are not included.

1.2 Background

In elite sport, performance differences of the smallest margins can distinguish competitive outcomes [2]. Since the magnitude of training adaptations diminish with advancing training status, the ability to

effect small but meaningful gains in performance characteristics is critical to success [3]. Expert performance requires task specific application of muscular force in creating appropriate movement patterns. Successful outcomes are dependent on the capacity of the physiological structures involved and the adoption of effective motor control strategies [4]. Performance improvement, then, requires prescription of appropriate training interventions to enhance these qualities. The principles of dynamic correspondence establish criteria supporting the effective transfer of training activities to performance in a given sport, aiding in the design of training programmes [5]. A well-defined understanding of the biomechanics of the goal movement is central to this process. In the sport of cycling, performance is largely delineated by the ability to deliver mechanical power to the crank [6]. In submaximal conditions, the optimal motor control strategy is a function of reducing metabolic and neuromuscular demand and maximising mechanical efficiency [7]. In contrast, the unique task demands of sprint cycling require production of maximal muscle power output [8]. While a substantial body of literature has described the biomechanics of steady-state cycling, there is currently a paucity of research identifying the unique characteristics of all-out sprinting [9].

Locomotion in cycling is achieved through the application of pedal force to produce torque at the crank. Three points of contact (feet, hands and pelvis) provide a loci of force transfer to the bike-rider system but equally constrain movement within fixed parameters established by bike set-up and rider position [10]. Bike geometry, rider kinematics, gear ratio and crank length will dictate performance along the force-length-velocity relation of contributing muscles, while the cyclical pattern of activity further limits the time available to develop force [11]. The interaction of pedalling rate, i.e. cadence, and crank length determines linear velocity of the pedal, ultimately establishing shortening velocity of the uni-articulate muscles contributing to pedal stroke [12]. Selection of gear ratio will affect the cadence range over which the sprint will be conducted and identification of the impact of cadence on muscle mechanical performance is, therefore, of substantial interest. Muscle power is the product of muscle force and contraction velocity, and is governed by the intrinsic properties of the muscle. The development of maximal external mechanical power, therefore, represents a compromise of muscle

function within the imposed limits of operation [13]. In this context, motor control strategy is critical. Indeed, Wakeling et al. [14] have suggested that coordinative pattern, rather than capabilities of the individual muscle, is the most important limiting factor to power production.

At any point in the crank cycle, only the proportion of pedal force perpendicular to the crank is effective in producing crank torque [15]. Driss and Vandevalle [16] suggested, therefore, that muscles have two distinct task demands in cycling, firstly to produce locomotive force, and secondly to maintain limb position round the crank cycle. Pedal force effectiveness, the ratio of perpendicular force produced to total resultant force delivered, is considered representative of the rider's pedalling technique [17]. However, skilled improvements in pedal stroke apparently have limited influence on enhancing pedal power [18]. An examination restricted to external force production is, consequently, inadequate for the purposes of assessing neuromuscular function in highly-trained athletes [19]. In contrast, Hug et al. [20] observed that, where inter-individual variance in pedal stroke is low in experienced riders, the activity patterns of contributing muscles were far more divergent. Electromyographical (EMG) studies suggest that there is, at least, a common sequencing of muscles around the crank cycle [21, 22]. Attempts have been made to associate changes in the magnitude and timing of activity of contributing muscle groups, with modifications in the environmental conditions such as workload [23, 24], cadence [25-28], fatigue [29], bike-set up [30, 31] and riding position [32-34]. However, while there is a clear relationship between muscle activity and power output, the relationship is not systematically perfect in maximal conditions. Indeed, Dorel et al. [23], showed that only the triceps surae and quadriceps muscle groups were activated maximally in sprinting trials. Given the influence of muscle redundancy, the central nervous system (CNS) has the ability to vary the recruitment strategy in response to environmental changes, and it is clear that sprinting requires a unique solution to motor control [16, 35].

Assessing the distribution of muscular effort at each joint offers a more intuitive means of assessing sprint performance. The use of inverse dynamics processes provides a non-invasive means of predicting

forces and moments within a linked-segment system, allowing evaluation of the net muscle activity affecting movement at each joint [36]. Kautz and Hull [37] further demonstrated that the process could decompose the contribution of muscular and non-muscular (i.e. gravitational and inertial) contributions to pedal force, benefiting a clear assessment of neuromuscular function in generating external power. Hip, knee and ankle joints dominate power production at the crank but are augmented by power generated by the upper body and transferred across the hip [38]. Extensor moments, from approximately 0 to 180° of the crank cycle, are the primary contributors to forward motion; while between roughly 180 and 360°, flexor moments dominate as the pedal returns to top dead centre and contralateral limb extends [18]. Comparisons of submaximal and maximal pedalling conditions reveal that the task demands of maximising power output affect an increased contribution of hip extension and knee flexion, and decreased contribution of knee extension to total muscular power [39]. Martin and Brown [40] further determined that muscle redundancy was certainly exploited in sprinting, helping prolong the crank phase of joint extension. Despite the clear differences in sprint performance, only limited research is available examining joint-specific power in maximal conditions. In the only research presenting joint kinetic data using sprint-trained participants, results of a case study of a competitive sprinter suggest that skilled sprinters may have unique functional strategies to aid maximal power delivery [41]. Wheat and Barratt [42], additionally acknowledge that the morphologies of sprint-, as compared to endurance- trained riders, would impact the inverse dynamics solutions.

The physical characteristics of sprint cyclists are uniquely specified by the demands of the sport. Sprint cyclists are more commonly mesomorphic, being heavier, stronger and with larger segmental girths than their endurance counterparts [43]. Newton’s first principle dictates that the riders must be able to produce high levels of force in order to affect the explosive accelerations required to attain high end velocities [44]. Both cross-sectional area of muscle and lower-leg lean volume are related to performance [16], indicative of the strength capacities required by the athletes. Resistance training occupies a substantial part of the sprint cyclists’ programme, not only to assist hypertrophic development but also for the purposes of increasing maximal strength, the ability to exert absolute

maximal levels of force, as well as the overall functional capacities of the rider [45, 46]. A significant body of research has also examined the development of muscle power using resistance training techniques [47-49], and, indeed, strong relationships exist between measures of maximal power output assessed in the gym and ballistic sports performance [50-53]. However, 'strength' represents the ability of the athlete to apply force under specific movement conditions [5], and, in such a case, effective use of resistance training techniques requires a clear understanding of how the movements utilised relate to movement in the sport. Specificity of a prescribed resistance exercise involves ensuring the muscles are stressed in similar functional conditions to those of the goal movement, such as the ranges of length and velocity of operation. With triple extension of the lower-limb providing the primary locomotor force in cycling, squat-based patterns and derivatives are commonly applied in resistance training [54]. Joint-specific kinetics have been examined in a number of these exercises with results showing that the biomechanical demands of the lift, including placement of the load with respect to centre of mass [55, 56], magnitude of the load itself [57, 58], stance width [59, 60], depth [61] and supporting leg position in unilateral lifts [62], all affect the distribution of muscular effort. Given the number of confounding factors, as well as the impact of skill level on performance [63], assessment of the lifts in the cyclists themselves would provide the only accurate means of relating the exercise conditions to the sprint action.

Joint-specific kinetics, therefore, provide a highly beneficial approach to improving biomechanical understanding of track sprint cycling athletes. However, the use of inverse dynamics is not without problem [64]. With movement kinematics, kinetics and rider anthropometry as model inputs, the process applies Newton-Euler equations to solve for unknown moments and forces within each segment of the linked system [4]. Segment inertial parameters, positional data tracking, positioning of markers and/or sensors, location of joint centres, estimation of the centre of pressure location, soft-tissue artifacts and errors in force plate measurements, all contribute sources of error [42, 65-69]. Testing methodology is, therefore, critical in minimising inaccuracies. Commercially available 3D motion capture systems have become the gold-standard for data acquisition and processing, providing high data resolution,

accuracy in position tracking and integrated modelling algorithms [70]. However, in these systems, kinetic information is assumed to be associated with ground reaction forces and therefore kinetic model inputs are synchronised directly with in-floor force platforms. Bespoke solutions are therefore required for cycling assessment where forces are produced at the shoe-pedal interface. Given that the cycling movement is predominantly in the sagittal-plane, 3D assessment of cycling biomechanics has been assumed as unnecessary [71]. Yet comparison of 2D and 3D analysis methods in other sports has suggested that the influence of biomechanical coupling, where one direction of movement influences movement in another direction, can critically affect outcomes [72]. Although *lower-limb* movement is predominantly sagittal plane in cycling, sprint cycling is known to have an increased contribution of power transferred across the hip [73]. Hence the impact of coupling action at the lumbar-pelvic-hip complex may be significant. Currently few studies exist utilising 3D processes in cycling, yet critical discrepancies have been quantified in kinematic analysis of cycling between 2D and 3D systems [74]. Two-dimensional analysis imposes some additional errors including oversimplification of movement patterns, particularly at the hip [75], camera parallax, which impacts segmental lengths during motion, and the coordinate reference system not being coincident with the true axis of rotation of the joint [76]. These in-accuracies would, therefore, be compounded in the inverse dynamics analysis.

To date, no assessment has been made of the benefits of 3D systems in modelling cycling. The reliability of data from these biomechanical processes has similarly not been established for cycling research. Flexibility of the motor domain is critical in sporting action, with the CNS utilising available degrees of freedom to ensure stability in performance outcomes [5]. Although Martin and Brown have demonstrated changes in effort distribution with ensuing fatigue during all-out sprint [40], the consistency of net joint moments in more stable performance conditions is yet to be established. Indeed, no study has examined intra-individual variability of joint-specific power distribution in repeat performances of a cycling test. Systematic testing must also be able to differentiate the contribution of typical process error in order to determine meaningful change in the athletes' performance [2]. Performance variability in average cycling power has been reported as being as low as 3% [77, 78],

leading to recommendations that testing be able to detect smallest worthwhile changes of 1.5% [79]. To be utilised as a regular component of an elite testing battery, results of joint kinetic analysis must, therefore, be understood in the context of inherent system and athlete variability. In doing so the data can then serve, not only to provide an understanding of performance and informing training prescription but can further provide a means of evaluating changes following training intervention.

1.3 Significance of the Research

The system designed to facilitate this research provides a novel solution to 3D biomechanical assessment of cyclists. This system solution may be used to examine biomechanics of performance on any cycling ergometer and may be utilised for a wide range of testing and research purposes. Results of the research presented in this thesis will further our understanding of the biomechanics of highly-trained track sprint cyclists both on the bike and in performance of key resistance exercises commonly used in training. This information will provide an assessment of the distinguishing biomechanical characteristics of highly skilled sprint performance, thereby assisting determination of the training needs of the athletes. Critically this will aid coaching and support staff to improve prescriptive practices developing athletes to the highest levels of performance in the sport. Key aspects of the results will inform the choice of testing practices used for biomechanical assessment. Specifically, athlete support personnel will have a qualitative means of determining whether a simple 2D system or more sophisticated 3D testing process is most appropriate for a specific analysis or purpose. Findings additionally support the accurate interpretation of results from repeat testing of the population, allowing evaluation of functional change consequent to the training interventions employed. In this context, variability data will provide practitioners with the means of determining the significance of any performance changes observed during testing. Hence, athlete progress can be more accurately monitored, while the effectiveness of training methodologies can be assessed.

1.4 Research Aims

This thesis aimed to develop a custom calibrated system in order to investigate the joint kinetics of sprint cycling performance in highly-trained track sprint cyclists. Key outcomes were, then, to determine the impact of changing cadence and riding position on joint-power distribution. Further aims were to determine the accuracy and reliability of the testing methods and outcomes and to assess the compatibility of key resistance exercises for rider development through comparison of joint kinetics in each mode of activity.

Specific aims for each study presented were:

1.4.1 Study One

Identify discrepancies between two- and three- dimensional methodologies for joint kinetic assessment in order to provide recommendations for practical implementation of testing practices.

1.4.2 Study Two

Assess the effects of changing cadence and riding position on joint-specific kinetics of sprint cycling performance, while, additionally, determining the reliability of the three-dimensional test process and evaluating intra-athlete variability in test performance.

1.4.3 Study Three

Assess joint-specific kinetics of key resistance exercises and determine their association to those of sprint cycling in the same athlete population.

1.5 Research Questions and Hypothesis

Research questions (Q) and hypothesis (H) pertaining to each study are as itemised below:

1.5.1 Study One

Q1: Does utilisation of three-, as compared to two- dimensional processes, significantly impact the assessment of sagittal-plane joint kinetics in high performance sprint cyclists?

H1: Significant differences will be observed between outcomes in each case, suggesting that utilising a three-dimensional process will increase the accuracy of results.

Q2: Is the magnitude of any apparent difference in three-, as compared to two- dimensional processes, equivalent at each contributing joint?

H2: The magnitude of difference between three- and two- dimensional analysis will be greatest at the hip joint, and, to a lesser degree the knee joint, with ankle joint showing little apparent difference.

1.5.2 Study Two

Q3: Is joint-specific power of the lower limbs during sprint cycling affected by cadence?

H3: Joint-powers will demonstrate a parabolic relationship with cadence, with the cadence at which peak power is observed being joint-specific. Distribution of joint-power will vary with cadence, with changes across each joint showing a clear trend with increasing or decreasing cadence.

Q4: Is joint-specific power affected by changing riding position from seated to standing?

H4: The distinct performance conditions of seated and standing sprinting will affect joint-power distribution, particularly observing an increased contribution of cross-hip power in the standing position.

Q5: Does a three-dimensional joint kinetic assessment process provide a reliable means of assessing highly-trained sprint cyclists?

H5: Evaluation of results from repeated testing will demonstrate coefficients of variation (CV) within acceptable margins of reliability (i.e. CV <10%).

Q6: Are joint-specific powers of expert performers during sprint cycling consistent across multiple repeats of the test?

H6: A small degree of intra-individual variability will be observed in joint-specific power across test repetitions.

1.5.3 Study Three

Q7: Is there an association between the joint moment-angular velocity profiles of the hip, knee and ankle obtained in executing the clean exercise at multiple loads with those of sprint cycling performance over multiple cadences?

H7: The joint moment-angular velocity profiles of hip, knee and ankle in executing the clean exercise will have only partial association with those of sprint cycling.

Q8: Does the kinematic profile of a resistance-training exercise impact the distribution of muscular effort across the contributing joints?

H8: Resistance exercises will vary in the distribution of effort across contributing joints, with deadlift, Romanian deadlift, hip thrust and single leg rack pull showing greater dominance of hip joint torque and step up showing greater knee joint torque.

Q9: Does joint-specific torque in various resistance exercises correspond to that in sprint cycling?

H9: Resistance exercises showing a hip-dominant effort distribution will have greater peak and mean hip torque than observed in sprint cycling, while greater peak and mean knee torque will be observed in knee-dominant exercises as compared to sprint cycling.

1.6 Limitations

The testing periods of this research were constrained by the training and competitive schedules of the athletes. Given that athlete training is generally structured with different emphasis through the year, it is possible that outcomes observed may change during different training phases. Due to the high level of the athletes within this research, the ability to control individual training schedules of participants was also limited during the trial periods. Differences in training load and, hence, physical condition may be present. The necessity of having cabling attached to the pedals, and the requirement to be in a clear motion-capture space, enforced the use of a stationary cycle ergometer for testing. This limits the natural lateral sway of the bicycle that would be inherent in performance on the velodrome. However, research intentions were to characterise the fundamental pedalling patterns of riders rather than determine the extent to which movement varies in a dynamic environment. Ergometer training is a component part of the athlete's training week and athletes are, therefore, highly familiar with performance on such equipment. The specific ergometer used for the test was as used by the elite squad in regular testing and training practices. The resistance exercise component of the study was conducted during a maximal strength training block for the athletes, with exercises representing the key lifts being utilised in this phase. It is acknowledged that other exercises are commonly utilised in developing sprint cyclists and would provide an opportunity for future research.

1.7 Delimitations

This research examined only performance in track sprint cyclists at state and national level. While criteria for participant selection for these studies, therefore, limited the selection pool, the examination was only intended to be descriptive of this distinct homogenous group. In requiring at least a 2-year training history exclusively in sprint disciplines, results of the studies are unable to account for performance of riders from other cycling disciplines or of a mixed training background. It is also likely that outcomes would be different in athletes of lower performance levels. Sprint cycling performance was examined under controlled conditions, where, in competitive situations, a number of environmental

factors will influence performance. In racing, athletes are also able to choose gear ratio's to best suit their physical characteristics. However, the range of test cadences was individualised, providing a means of matching assessment to these characteristics. The impact of bike-setup would similarly allow a means of altering an individual's biomechanical profile. However, high level riders utilise expert bike fitting services to optimise their bike-set up for competitive performance and the ergometer set up used in the trials was matched to this data. Hence, in testing maximum performance, tests were completed as close to their competitive position as possible.

1.8 Definitions of Terms

2D: Two-dimension(al)

3D: Three-dimension(al)

AIC: Akaike information criterion

AJC: Ankle joint centre

AJP: Ankle joint power

AJM: Ankle joint moment

ANOVA: Analysis of variance

ASIS: Anterior superior iliac spine

BIC: Baysian information criterion

BDC: Bottom dead centre (of crank cycle)

CI: Confidence intervals

CNS: Central nervous system

COP: Centre of pressure

CSA: Cross sectional area

CT: Computed tomography

CV: Coefficient of Variation

EMD: Electromechanical delay

EMG: Electromyography

F_0: Peak isometric force

FV: Force-velocity (relationship)

FL: Force-length (relationship)

GT: Greater trochanter

ID: Inverse dynamics

ISLS: Integrated spatial linkage system

IK: Isokinetic

HJC: Hip joint centre

HJP: Hip joint power

HJM: Hip joint moment

KJC: Knee joint centre

KJP: Knee joint power

KJM: Knee joint moment

LMM: Linear mixed model

P: Power

P_{max}: Maximum power

PV: Power-velocity (relationship)

PO: Power output

PSIS: Posterior superior iliac spine

R: Correlation coefficient

R^2: Coefficient of determination

RDL: Romanian deadlift

RFD: Rate of force development

RM: Repetition maximum

RME: Relative muscular effort

RPM: Revolutions per minute

SD: Standard deviation

SRM: Schoberer Rad Meßtechnik (a portable power monitoring system for bicycles)

T_{Pk}: Peak dynamic torque

T_0: Peak isometric torque

TDC: Top dead centre (of crank cycle)

V_{opt}: Optimal cadence (cadence at peak power)

V_0: Maximum unloaded shortening velocity

2 Literature Review

2.1 Introduction

The performance capacity of highly-trained athletes has been refined over years of focused practice and continued performance improvement requires a more advanced training prescription based upon key principles of specificity [3, 44]. To this end, the principle of dynamic correspondence outlines specific criteria guiding programming decisions including understanding the timing and direction of force application, type of muscle contraction, movement strategies employed and direction and sequencing of joint movements; thus underscoring the need for a thorough biomechanical analysis [5, 80]. While the biomechanics of steady-state cycling have been frequently reported [10, 30, 81, 82], the unique characteristics of sprint performance are less well defined [11]. Further, expert performers commonly exhibit physical and technical proficiencies distinguishing their performance [83, 84], yet, of the published studies in sprint cycling biomechanics, only a single case study has examined performance in the track sprint population [41]. Evidence of some unique characteristics highlight that wider assessment in an elite track sprint cycling population may be more revealing.

Mechanical power output is known to discriminate performance in cycling and the assessment of how external power is produced and transferred to the bicycle crank is of substantial interest [6, 85]. However, specificity of task demands critically affect the optimal movement strategy. Where economy and efficiency of movement are important to steady-state cycling [86], sprint cycling requires the athlete to produce and sustain maximal muscle power [11]. Muscle redundancy allows the central nervous system (CNS) to determine a recruitment and coordinative solution specific to performance conditions [35], and, indeed, muscle activity patterns during maximal cycling are different to those observed in submaximal conditions [16, 23]. At muscle level, the motor control strategy determines both the magnitude and timing of forces, systemically affecting net force present at each joint as well as the orientation of the external force delivered [87]. Effective pedal stroke requires orientating the greater proportion of force perpendicular to the crank [31]. However, as a representative measure, changes in

so-called ‘pedal force effectiveness', are not consistently linked to pedal power production [17]. In contrast, power output does appear to affect the distribution of muscular effort, i.e. how force production is distributed across the contributing joints - even where pedal force effectiveness remains unaltered [71]. Examination of joint kinetics, the moment and power produced at joint level, can, therefore, be more informative. In fact, recent studies have shown some key changes in joint power distribution in comparing maximal versus submaximal cycling performance [39].

As an explosive sport, sprint cycling demands rapid acceleration, hence the ability to exert high levels of force production [88]. The use of weightlifting, therefore, features highly in the training week of sprint cyclists as a means of improving the strength capacities of riders [45, 46]. The predominance of squat-based patterns in exercise selection assumes an association with lower-limb triple-extension observed during the primary power delivery phase of the crank cycle [54]. Since off-bike training is intended to develop functional capacities, there are differences of opinion as to the degree of kinematic specificity that needs to exist between weightlifting and the sport movement [89, 90]. However, it is likely that the equivalence in the distribution of muscular effort is more relevant. Joint kinetics have been described in a number of weightlifting exercises, with results showing some dependence on aspects of technical application and execution [57, 61, 63, 91]. This emphasises the need for individual joint kinetic assessment, particularly within populations with refined movement patterns, such as athletes.

The tools and technologies available to the biomechanics practitioner are constantly evolving and improved methodologies may provide opportunity to refine analysis through the quality of data obtained. The seminal research in cycling biomechanics has utilised two-dimensional (2D) motion-capture systems with a single camera manually synchronised to pedal force data [36, 92-94]. Cutting-edge mechanical analysis practices now use three-dimensional (3D), multi-camera motion capture systems, internally synchronised with high resolution force acquisition technologies and packaged with

integrated analysis and modelling solutions [70]. While weightlifting studies have made effective use of this technology [95-97], their use has been of limited interest in cycling, in part due the movement being predominantly in the sagittal-plane [71]. However, cycling presents unique challenges for biomechanical evaluation, particularly that external force is applied to a moving pedal. Competitive cyclists use a cleated pedal system to secure the foot to the pedal, meaning that the shoe-pedal interface is not directly accessible for force measurement. Furthermore, commercially available motion-capture systems directly link modelling inputs with ground reaction forces, assumed, therefore, to be derived from in-floor force platforms [70]. Novel strategies are needed throughout the data acquisition process to allow a full biomechanical assessment of cycling. To date there has been little research in this regard and further research exploring innovative solutions is required.

Supporting examination of the sprint-specific disciplines of cycling, this review will focus on the competitive discipline of track sprinting. The sport will, first, be introduced through presentation of its athletic demands, focussing on the key factors influencing neuromuscular function. A brief review of seminal findings in steady-state cycling biomechanics will be presented, providing a reference against which to, thereafter, assess differences in sprint performance. A discussion of methodologies and issues in analysing cycling biomechanics will be included with key review content then focussing on examining the application of joint kinetic analysis in both cycling and weightlifting, with particular relevance to the assessment of track sprint cyclists.

2.2 Athletic Demands of Track Sprint Cycling

World-standard track sprint cycling encompasses a number of events of different length and format. Distinct from the track endurance events, sprint competition involves short, fixed-distance races where riders accelerate to maximal velocity from standing, rolling or flying start. Sprinters compete in rides against the clock, such as the flying 200m qualifier ride (elite times being <10s men, and <11s women) and individual 500m (women, ~34s) or 1000m (men, ~60s) time trials, as well as tactical rides against

other opponents including the match sprint, where two riders challenge each other over three laps, and kierin, where 6-8 riders are paced by motorbike up to the final three sprint laps [54]. In these tactical contests the length and duration of the final sprint is dependent on tactical approach employed. Typical competitions also require rapid recovery in order to sustain performance over multiple heats and rounds within a single day. Finally, the team sprint event uniquely highlights the varied conditioning requirements for sprint cycling. Held over two laps for women and three for men, the team members ride in-line, each taking the front for a single lap before pulling off. The event therefore requires distinct rider profiles, from the explosive, high force characteristics of a starter to the sustained speed-endurance qualities of the finisher [98]. Given the unique demands of the various sprint disciplines, athletes may compete as an all-rounder across the events or may become specialists in particular races. It is noteworthy that the wind-up and tactical prelude laps of sprint events additionally infers a substantial sub-maximal component to rider conditioning [99].

Performance in sprint cycling is ultimately determined by the balance of power supply and demand [11]. Power supplied is determined by the contractile properties of the contributing muscles mediated by fatigue, pedalling rate and riding position [8, 13]. The distribution of energy sources for muscle contraction varies across the sprint events, with shorter distance events critically dependent on the both the PCr and Glycolytic systems, while longer sprint distances involve a more substantial aerobic contribution [98]. Power demand includes overcoming air, rolling and bearing/drive-train resistances as well as accounting for changes in kinetic and potential energy related to mass, gravity, inertia and velocity. The impact of aerodynamics on cycling is emphatic, with the air resistance term accounting for a substantial 96% of available power when travelling at steady-state velocity on a flat surface [11]. At the outset of the sprint, where velocity is low, riders adopt a standing position which increases power delivered to the bike, power surplus then affecting an increase in kinetic energy [73]. At maximal velocity, where air resistance terms dominate, riders maintain a seated and aerodynamic position. The tucked position, though compromising function of the musculature of the hip, reduces frontal area, ultimately benefiting performance [100]. Critical trade-offs are, therefore, apparent in the ideal

functional and anthropometric characteristics of riders. The mesomorphic profile of sprint cyclists are representative of their capacity for force production [43]. However, increased cross-sectional area (CSA) may alternatively detrimental through increasing aerodynamic drag [101].

Sprint cycling requires the production of maximal forces to affect explosive accelerations [88]. However, riders must also be able to sustain the high velocities attained over extended distances. The fixed gearing of track bikes provide a unique challenge in that the gear selection imposes a compromise between the force required to overcome inertial load and accelerate from low or zero velocity, and the leg speed required to maintain peak velocity later in the sprint. A small gear will facilitate fast start times but establishes a higher, and potentially less effective, race cadence thereafter [101]. Notwithstanding a gear selection attuned to the athlete's physical attributes, performance over the course of a sprint demands function across a wide range of the power-force-velocity relationship. Elite riders will generate peak crank torques of over 250Nm in initiating a maximal acceleration and peak pedal rates of over 160rpm by the final stages of the sprint [98, 99, 102]. During acceleration, power will initially rise to a peak (around 2000-2500W in elite males and 1400-1600 in elite females), before declining towards maximum velocity and being sustained against ensuing fatigue during the velocity maintenance phase [98]. Literature commonly ascribes optimal cadence, the pedal rate at which peak power is developed, as being between 120 and 130rpm [11]. However, maximum external power output represents a compromise in the summated power-velocity relationships of contributing muscles and is therefore dependent on muscle contractile characteristics and coordination pattern [8]. Since it is uniquely specified, identifying the riders' optimal cadence provides valuable data to inform gear selection and is a key metric that is responsive to training adaptation [101]. Given that higher cadences are on the descending limb of the power-velocity curve, competition trends have supported shifts to bigger gearing and lower race cadences [101]. The use of larger gears is additionally supported by observations that fatigue may be influenced by the number of contractions required to complete the race distance [103]. Although a higher gear allows the rider to travel further with each pedal stroke, the increased force requirement further accentuates the demand for high strength capacity in the athlete.

In order to maximise the development of sprint cyclists' strength, specific structural and neural adaptations are necessary to improve contractile function. The importance of body composition has already been highlighted and changes in CSA, muscle mass and muscle architecture are primary training goals [45, 46]. The relationship of CSA to force production is well established, higher physiological CSA representing greater number of muscle fibres in parallel [104]. Strong relationships have also been observed between lower limb lean muscle volume and maximal cycling power [11]. However, a critical trade-off must be assessed in avoiding excessive size which may be detrimental to aerodynamics. Maximal strength training, seeking to improve the athlete's ability to exert absolute maximal levels of force, will emphasise neural adaptations such as increased motor unit activation, higher motor unit discharge frequency, greater motor unit synchronisation and recruitment coordination [105]. Explosive performance is also dependent on the ability to produce force during critical time periods [106]. This is particularly important in cycling given the crank cycle limits time available to develop force. For example, at 120rpm time available for contraction, i.e. half the cycle, would be 250ms, yet knee extensors have been shown to take >300ms to reach peak force [107]. With increasing cadence, the dynamics of force development and relaxation become critical limiting factors [16, 108]. Specific strength training prescription can affect a number of adaptations known to benefit the rate of force development (RFD) [109]. However, of particular relevance is the relationship of RFD to connective tissue stiffness. Improvements in tissue stiffness are commonly observed following resistance training and have been shown to positively impact RFD [110, 111]. In cycling, Watsford et al. [112] noted that riders with higher musculoarticular stiffness have a superior ability to develop effective crank torque during sprinting. Ditroilio et al. [113] also noted that stiffness characteristics impact that ability to maintain sprint performance during fatigue.

Given the importance of rider strength characteristics, sprint cyclists dedicate a substantial amount of the training week to resistance training sessions. Riders' must have advanced movement competency in a range of resistance exercises, particularly those that match the kinematics of the downward joint-extension phase of the crank cycle [54]. Squat and deadlift variations, as well as seated leg-press, are

common for pure strength development, while jumps and weightlifting movements are incorporated to aid development of muscular power. The exact content of the resistance training component, is, however, highly dependent on coaching ethos. For example, the use of either unilateral or bilateral squat patterns is hotly attested. Regardless, movement pattern specificity, maximum absolute exercise load and ability to affect maximal recruitment are key considerations when constructing a resistance training intervention for the sprint cyclist [114]. One area of consideration that is not clearly understood is whether matching critical joint angles and movement velocities during resistance training aid the transference of training effects to the sport [5]. Although some relationship of performance in weightlifting exercises to sprint cycling performance have been reported in the scientific literature [88], it is rather the overall functional capacity of the rider's in key lifts that is important to supporting the demands of the sport. Finally, it should be noted, that while the lower limb provides primary force production in cycling, the musculature of the upper body plays a significant role in force transference to the bike. In fact, Costes et al. [115] determined that increases in crank power are associated with higher magnitudes of upper limb kinetics, while McDaniel et al. [116] reported that at least 9% of the total contribution of pedal force in seated sprinting is known to derive from force transmission across the hip. In such a case, the strength capacities of the upper limb segments and torso musculature are also critical in enhancing rapid force production.

2.2.1 Functional Assessment of Athletes

The functional profile of athletes can be described by their power-force-velocity relationship [117]. Peak mechanical power is a key discriminator of sprinting ability and can be assessed directly on a bike or cycle ergometer in short (e.g.~ 6-second) all-out efforts [16]. Using appropriate protocols and instrumentation, the individual's unique mechanical profile can be derived from the contributing factors of force and velocity during pedalling. In controlled laboratory conditions, a number of protocols have been utilised including repeated sprints on friction-braked ergometers against set braking forces [118], and isokinetic ergometers [119] at multiple cadence, as well as single all-out tests against an appropriate inertial load [120]. Tests alternatively capture data after reaching maximal velocity in the sprint [16] or

during the acceleratory phase [121]. High reliability is generally observed during such tests irrespective of the ergometer or protocol [16]. Within days and between days reliability has been established in both isokinetic (interday and intraday test-retest correlation coefficient, r > 0.9 for peak power) [119, 122] and inertial load tests (e.g. peak power interday r = 0.99, intraday r = 0.97) [120, 123]. Force-velocity profiles have also been presented from field-based performances of all-out sprinting over 65m [124] and 80m [125] on gear-ratios providing a fixed inertial load. Comparisons have been made of functional profiles derived from laboratory- and field- based assessments [124, 126]. Although Bertucci et al. [126] found some distinction between profiles derived using a stationary-mounted standard racing bike with that in field performance, Gardiner et al. [124] compared maximal torque- and power- pedalling rate relationships using a laboratory-based ergometer with those on moving bicycles and found lack of any significant difference in either regression coefficients or calculated variables. The Gardner et al. results are noteworthy for the current studies, having critically examined a specifically elite track sprint cycling cohort and, indeed, utilising an ergometer commonly used by this population in training.

A number of key functional descriptors can be derived in addition to maximal power (P_{max}) from force- or torque- velocity testing. Optimal cadence (V_{opt}) is notably related to fibre type distribution [127, 128] and has been proposed as a surrogate means of fibre-type testing. Extrapolation of the force-velocity (FV) regression line allows for the prediction of maximal isometric force (F_0) and maximal unloaded velocity (V_0), with the slope of the regression analysis then uniquely specifying the individuals' functional characteristic [129]. Driss et al. [118] showed that F_0 is related to a rider's strength characteristics, showing a significant relationship between the indice and knee extensor force production. Additionally, Jaafer et al. [130] showed that the value of F_0 was highly reliable in repeat testing on a friction-loaded ergometer, while V_0 demonstrated more divergence. Representing the shortening velocity of the fastest muscle fibres, Sasaki and Ishii [131] have demonstrated that, *in vivo*, the value of V_0 is affected by recruitment pattern. In such cases this may reflect the challenge of affecting a consistent and successful control strategy in high leg speed pedalling [41]. Although Bertucci et al. [132] were unable able to delineate performance level using these 'classical' profile

parameters, the authors established further parameters from the functional relationships, including maximal power duration criteria, that uniquely distinguished the pedalling characteristics of elite performers.

The bike, of course, provides the ideal monitoring tool. However, since a large component of sprint cyclists' training is conducted in the weight room, monitoring of performance progression in this environment is also important. In many sports, weight-room based testing utilises loaded or unloaded jumps to assess mechanical power, with such tests providing an indicative measure of the contribution of stretch-shortening cycle function to force production [133, 134]. Their relevance in a predominantly concentric sport is less clear, in fact, maximal power in a single jump test is not consistently related to that assessed on the bike [127, 135-139]. While isometric force production in both single- and multi-joint leg extension has been shown to be related to sprint cycling performance [9, 88], disparity is known to exist between maximal strength expressed by athletes in dynamic lifts and their on-bike performance. Functional FV profiles can be derived during resistance exercise performance from force and positional transducers during execution of multi-joint lifts at incremental loads [140, 141]. In-keeping with bike tests, results in the leg press [142], multi-joint leg extension [143], jump squats [144], squats [145] and ballistic pushoffs [117], have demonstrated a quasi-linear relationship, contrasting the classic parabolic Hill relationship observed in isolated muscles [146]. In weightlifting this has been explained by the influence of segmental dynamics, increasingly cancelling muscle force production at higher limb velocities [142]. Cycling characteristics are, instead, critically affected by activation dynamics, with the relatively slow rise and decay time of the muscle active state increasingly impeding force production as pedal rate increases, limiting time available [8, 147, 148]. Such differences make comparison of function expressed in the two modalities challenging at system level. Indeed the disparate means of deriving external 'force' (i.e. braking force or acceleratory torque at a flywheel, crank torque, pedal force at the drive train, or ground reaction force, which may or may not include body mass with load), is similarly obstructive. In this regard it is worth noting, that, although the FV profiles in cycling represent combined action of both flexor and extensor torques, Dorel [149] confirmed that the profiles

produced by distinct extension and flexion phases of the crank cycle are similarly linear. It is apparent that gaps exist in understanding the association of on- and off- bike training. Bridging that gap would help improve the application of gym training for sprint cyclists, and provide the basis for a more meaningful interpretation of functional changes affected through gym intervention.

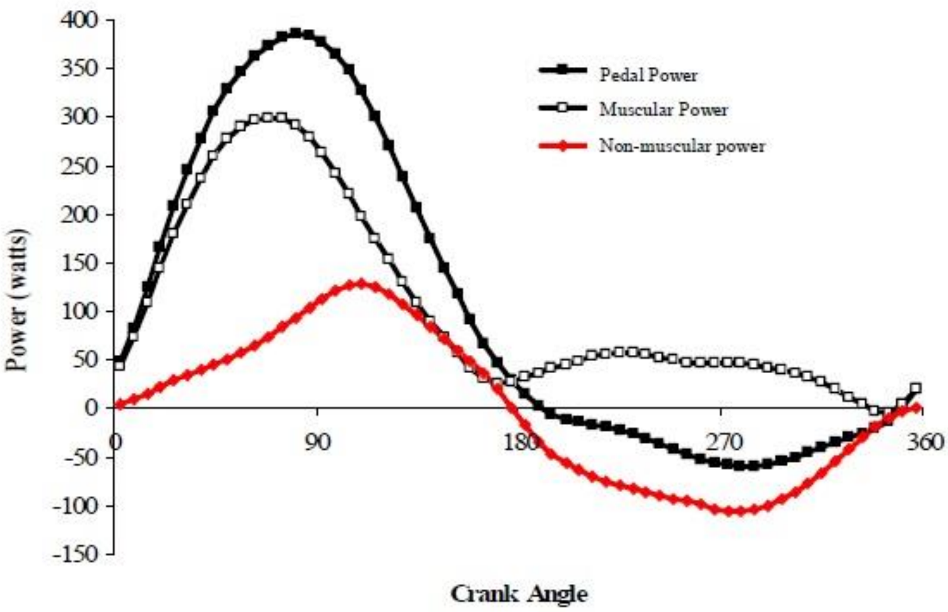
2.3 Biomechanics of Cycling

2.3.1 Bike Set-Up and Pedal Stroke

Cycling biomechanics are fundamentally characterised by the interaction of bike and rider. The bike-rider interface constrains a number of aspects of biomechanics through, for example, bike geometry and set-up, gear selection and crank length [13]. Establishing the correct bike fit is critical given that it affects the ranges of motion through which the limbs can travel and hence operation along the force-length (FL) relationships of muscles [150, 151]. A number of aspects can be altered, of which seat height and set back (distance behind the crank bottom bracket) have been most frequently examined, given their impact on positioning and hence kinematics and kinetics of performance [152-154]. Gearing and crank length also effect muscle operation, both critically impacting pedal speed, which ultimately determines shortening velocity of the uniarticulate muscles, and pedal rate, which affects muscle excitation state [155]. Crank length has generally been considered as influencing power production since it represents the moment arm about which pedal force acts. However, Martin et al. [12] determined that, in cyclical activities, force production does not vary in a purely velocity-dependent manner, but is interactively affected by the time-dependent effects on activation state. With longer cranks, maximal power is achieved at lower pedal rates but at higher pedal velocities, power production being distinctly affected by two different physiological constraints. Consequently, while crank length has been confirmed as affecting metabolic cost, Martin et al. [156] determined that crank length actually only substantially affected maximal power at extremes of length. The authors determined that standard crank lengths used in practice (e.g. 170mm) would only compromise taller and shorter riders' performance by at most 0.5%.

To achieve maximum crank torque, application of force requires coordinating the limbs to deliver the greater percentage tangential to the crank [157]. Pedalling technique is therefore considered a skilled aspect of cycling performance, with 'pedal stroke effectiveness', the ratio of effective force (normal to the pedal) to total force, referenced as a representative measure [17]. Motor control strategy affects timing and activation of contributing muscles not only to deliver maximum force, but to ensure movement kinematics of lower limb segments match the pedal trajectory [158]. While, Korff et al. [159] demonstrated that a more effective pedal stroke can benefit mechanical efficiency (power output to physiological cost), Bini and Diefenthaeler [160] determined that improving pedalling technique alone cannot benefit power generation. Using pedal force as the basis for understanding cycling kinetics is further confounded by observations that total force consists of both muscular and non-muscular (gravitational and inertial) components (Figure 2.1). Although it is recognised that most of the effective force, and hence positive power, is generated during the downstroke, the assumption that poor pedalling technique can affect negative power during the upstroke is misguided [1]. Deconstructing the force contributions shows that most of the negative power observed is due to gravity. Examination of intentional alterations in pedalling technique, further shows that improving pedal force effectiveness does not necessarily mean a more beneficial movement strategy [159]. For example increased pulling action results in a higher contribution of leg flexors, which ultimately prove to be less efficient musculature. Evaluation of kinetics at the pedal alone is insufficient for the purposes of a full biomechanical analysis.

Figure 2.1: Deconstruction of Pedal Power into Muscular and Non-Muscular Components



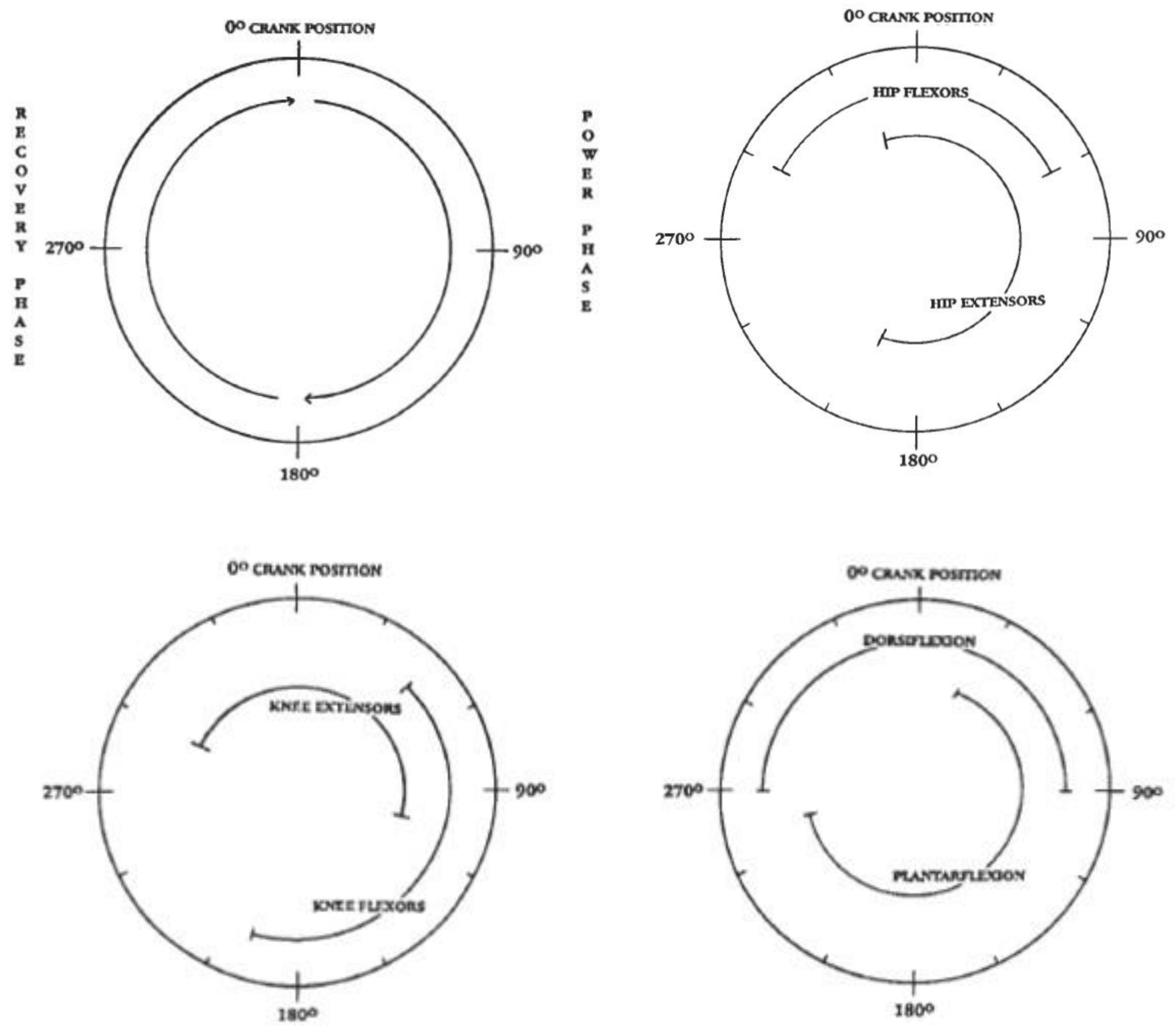
Reproduced from [1]

2.3.2 Coordination and Recruitment During Pedal Stroke

Use of electromygraphical (EMG) techniques provide a means of determining muscular contributions during the crank cycle. A full revolution can be divided into four key phases: the primary propulsive/downstroke phase from approximately 0 to 180° of the crank cycle, the pulling/upstroke phase from around 180 to 360°, and two transition phases, a forward motion bridging up to down at top dead centre (TDC, 0°) and rearward motion bridging down to up at bottom dead centre (BDC, 180°) [161] (Figure 2.2). During the downstroke, triple extension of the hip, knee and ankle provides the primary power production for forward motion, while lower limb flexion dominates during the recovery phase. Of particular note is that cycling involves all major muscle groups of the body [16]. The cleated shoe-pedal interfaces allow engagement of all the major leg and hip muscles during the full crank cycle, while muscles of the upper body are active affecting a contralateral sling action above the pelvis and ipsilateral pull against the handlebars, thus counterbalancing and counteracting force production in the lower limb [22]. Although a number of factors have been shown to impact muscle activity, a common sequencing of muscles is observed [21, 23, 25, 33, 34, 162, 163] (Figure 2.2). The muscles crossing the hip and knee are predominant in producing propulsive force. The quadriceps, acting in knee extension are active early in the downstroke, while hip extensors reach peak activity closer to the primary force

delivering position at 90°. The plantar flexors assume a more critical role in transferring limb segment energy to the crank, while also optimising limb position to follow pedal trajectory. Through the bottom of the pedal stroke, plantar flexors help control end range leg extension and act in combination with the knee flexors to affect rearward transition through BDC. Following transition to upstroke, the knee and hip flexors dominate the pulling action, with dorsi flexion of the ankle also contributing, before co-contracting with the rectus femoris to control segment position through transition forwards at TDC [164-166].

Figure 2.2: Muscle Activity around the Crank Cycle



Reproduced from [161]

Analysing muscle activity around the crank cycle, it is apparent that the uniarticulate muscles are the primary power producers, while the biarticulate muscles more ostensibly contribute to limb positioning, effective distribution of force between segments and fine control of net joint moments [22, 166]. Two-joint muscle activity is seen to be more highly variable than uniarticulate muscle activity, suggestive of their disparate roles in motor control strategies [16]. Co-contractions involving biarticulate muscles are observed at a number of key phases during the crank cycle to enhance control, steer transition at TDC and BDC and support joint stabilisation [165, 166]. The impact of such co-contractions can be seen to alternatively affect net extensor or flexor torques, and, hence, also directly impact pedal force production [167, 168]. The ankle proves to be a unique joint with respect to motor control solutions [40]. Modelling of muscle excitation has determined that the extent of joint motion at the ankle may depend on its primary task demands, i.e. whether it is acting to increase stiffness and stability for force transference, store and release energy, or fine tune overall limb kinematics to maximise power production and delivery to the pedal [22]. Martin and Brown [40] observed a reduction in ankle range of motion under conditions of fatigue, and determined that, since the ankle ultimately transfers all muscular power to the pedal, a stiffer ankle joint would minimise power loss, and, by reducing degrees of freedom, simplify task demands. Modifications of ankle position also apparently allow the extension phase to be extended benefiting increased power through the downstroke.

Electromyographical studies have demonstrated alterations in relative timing and coordination of muscle activity in response to increasing limb speed [27]. Electromechanical delay represents a relatively greater part of the pedal stroke at higher cadences, provoking suggestion that control strategies may advance onset timing in an attempt to maintain peak force delivery at optimal crank angles [27]. This so-called “activation dynamics” hypothesis has been proposed as the primary muscle property limiting performance [169] and forward dynamics modelling has further inferred its importance in determining optimal cadence [8]. Although the effects of excitation and relaxation on power output is irrefutable, the association with coordination strategies has been challenged; studies have observed

retardation of peak torque or peak power angle with cadence, while results with respect to muscle activation timing are equivocal [148]. Ettema et al. [170] and Baum and Li [171] both present evidence that excitations of lower limb muscles are disparately affected by cadence, meaning coordinative strategy is certainly altered. Proximal to distal differences in activity changes consequently led to questions around the influence of limb inertia [25]. Baum [172] decoupled the effects of movement speed and inertia by contrasting performance at multiple cadences with performance where weight was added to the distal thigh. Results suggest movement speed affects activity *time*, while limb inertia affects activity *magnitude*. There is no doubt that high movement speeds present a unique motor control challenge. EMG studies have demonstrated increasing antagonist activity and greater variability of patterns with higher cadences [22, 173]. Increases in negative work, highlighting the need for greater control of movement, have also been observed at fast limb speeds [94].

Refined skill development might suggest coordinative pattern be more consistent in expert performers. However, Hug et al. [20] observed a high degree of variability in the activation patterns of highly trained cyclists that was particularly prominent in the recovery phase of the crank cycle. Muscle redundancy further confounds assessment of consistent patterns of muscle force production during motion. The study of motor control suggests that recruitment strategies may utilise synergetic groups to alleviate the complexity of motor control associated with skilled performance [35]. The phases of the pedal stroke have led to a number of authors proposing that motor control strategies consist of activating distinct functional groups around the crank cycle [174, 175]. Using forward dynamics processes, Raasch et al. [166] determined that cycling performance could be modelled by two contralaterally alternating agonist-antagonist groups (hip-knee flexors, hip-knee extensors). However, Hug et al. [174] assessed EMG signals during both submaximal and maximal pedalling conditions, applying a decomposition algorithm to determine the weighting of muscles within synergetic groups (“Muscle synergy vector”) and relative contribution of an observed synergy to the activity pattern (“Synergy activation coefficient”). Results determined that variability across a range of torque-velocity and posture conditions could be explained by *three* synergetic groups. However, it is yet to be established whether

motor control solutions do indeed reflect the action of muscle synergies, as opposed to simply being an optimisation function that acts to minimise motor effort [35]. Understanding the characteristics of elite function through coordination strategy is, consequently, problematic.

2.3.3 Kinematics of the Pedal Stroke

The kinematics of cycling can be assessed by tracking limb segments and pedal motion using either single- (2D) or multi- (3D) camera motion-capture systems. As a predominantly sagittal-plane motion, most studies have concentrated on two-dimensional patterns [71], with displacements, velocities and accelerations significantly influenced by riding position [34, 176], bike set-up [32, 152, 177], workload [81], cadence [178] and fatigue [160]. Critical differences are observed in sagittal plane joint angles when using 3D as compared to 2D video analysis, suggesting increased accuracy of 3D methodologies [74]. Internal and external tibial rotation [179], hip adduction [81], frontal plane motion of the knee [180], inversion and eversion of the foot [181], and other motions out with the sagittal plane have also been observed and assessed [182], albeit more particularly referenced to injury predisposition than cycling performance. Limb trajectories, mediated by seat position, fundamentally influence the patterns of force production through FL and FV relationships [183, 184]. With seat height additionally impacting the flexion/extension ratios of the lower limb joints, methods for achieving optimal bike-set up are critical when attempting to maximise power production [31, 177]. Describing motion around the crank cycle, standard crank trajectory observes a circular motion through 360°, with pedal orientation assuming a sinusoidal pattern varying from ~65° to 115° angle over a full revolution [185]. Pedalling cadence and crank length also dictate velocity at the pedal. In seated pedalling the excursion of the hip during the crank cycle is around 38-45°, while knee excursion ranges from 66-75°. Ultimately, however, the excursion of the hip and knee is impacted by the riding position [161]. The thigh never reaches vertical orientation and, in fact, neither the hip nor knee achieve full extension. Seated pedalling sees the hip joint centre constrained to remaining behind the crank spindle, while in standing the rider moves the hip towards the handlebars [176]. Effects of removing this constraint are substantial, with far greater joint ranges of motion observed across the lower limb joints when in standing position [186].

Descriptions of ankle range of motion are more divergent than those of the hip and knee. Krause et al. [176] presented increases from 17-20° in seated pedalling to 25° in standing, while Shemmel and Neal [187] observed a 25° to 40° increase moving from seated to standing. In the seated riding position the ankle transitions between dorsi- and plantar- flexion, while the standing position requires the ankle to remain in plantarflexion contributing little to the pull pattern during upstroke. Based upon these data Shemmel and Neal [187] highlight this would increase contribution of the plantarflexors to pedal force during standing.

2.3.4 Upper Body Contribution

While upper body motion is, to an extent, constrained by gripping the handlebars, the high contribution of hip joint reaction forces to pedal power suggests a substantial impact of upper body dynamics [116]. Transition from seated to standing is further known to affect an increase in upper body contribution [115]. Upper body position also uniquely influences both internal and external factors associated with power production [188]; trunk angle and forward lean are known to affect both force development in the lower limb, as well as aerodynamics through frontal area, in other words critically affecting the trade-off between power supply and demand. Savelberg et al. [34] determined that upper body position had a significant impact not only on lower body kinematics but on patterns of lower limb muscle activity. Observing that ankle musculature (though notably not knee) was particularly affected, the authors suggest that changes in FL operation of the muscles crossing the hip may alter joint torque distributions and, consequently, operation of more distal muscles. This appears to be confirmed by Emanuelle et al. [188], who observed changes in joint-specific powers when shifting from an upright to dropped/racing position. Savelberg et al. [34] further observed that upper body position influenced the extent to which eccentric activity is observed through the pedal stroke. Although the bike itself is a constraint that limits full limb extension and hence contribution of elastic tissue properties to force production [189, 190], there is apparently some ability to modify conditions for energy storage and release.

Upper body motion is also discriminatory of skill level. Participants of lower training status show a greater propensity for changes in neuromuscular control due to body position [32], while greater lateral and rotational spinal movement has also been observed in less-skilled riders [81]. Specific motion of the pelvis is unique in cycling, having large angular excursions in non-sagittal planes [191] that facilitates energy transfer from the upper to lower body [31, 192] and augments force production at the hip [193]. Although the contribution of the pelvis has been examined with respect to power production and injury predisposition, it has been acknowledged that inaccuracies in the marker definition for pelvic tracking has impeded evaluation of its true contribution [194]. Despite the apparent simplicity afforded by a prevalent plane of movement in cycling, such observations of non-sagittal plane movement of the pelvis and upper body suggest that differences in outcomes through bi- as compared to tri- planar biomechanical assessments might be observed.

2.3.5 Effects of Workload and Task Demands on the Biomechanics of Pedalling

A number of authors have provided evidence that the biomechanics of steady-state cycling are affected by absolute [157] and relative [81] intensity of power demands. Kinematics analysis shows small but significant changes, most prominently a reduction in hip range of motion and increase in ankle range of motion, as workload increases [183, 195]. Examining changes at the crank and pedal in elite performers, Kautz et al. [196] observed that higher workloads resulted in an increased vertical pedal force and shift in pedal angle suggestive of a more ‘toe-up’ position during downstroke, while a positive (propulsive), rather than negative (resistive), torque was observed during the upstroke. Electromyography studies have observed greater magnitudes of activity as power demands increase [31]. However, increased EMG activity is not always concurrent with higher power demands and, in fact, power output is not clearly associated with the levels of activity of individual muscles [14]. Evidently a number of interactive changes in movement mechanics are affected in order to improve whole limb power generation.

Beyond a simple increase in workload, demands of all-out sprinting present a substantially different motor control problem to steady-state cycling conditions. Studies in running confirm that the strategies providing optimal solutions in sprinting are quite distinct from those of paced running [197]. Although there is a paucity of studies examining sprint cycling, existing results confirm critical differences in recruitment strategies. Dorel et al. [23] observed a change in the relative contributions of muscles during all-out sprint cycling, and, somewhat unexpectedly, found that activity levels were not systematically maximal. Activity of the hip extensors, hip flexors and knee flexors were also seen to be greater than in submaximal conditions. Driss et al. [16], therefore, proposed that sprinting is dependent on four, rather than three synergetic groups: the uniarticulate hip and knee extensors, plantar flexors and biarticulate hip extensors, uniarticulate hip and knee flexors, and dorsi flexors and biarticulate hip flexors. The unique function of the ankle in sprinting also places exceptional demands on force production by the plantar flexors. Indeed, the ankle has been suggested as representing a limiting factor in extended sprints [40] and evidence suggests that energy transference during sprint cycling would require plantar flexor force production in excess of the forces achieved in a maximal isometric contraction [16]. Eccentric contractile conditions have been reported in the triceps surae complex during all-out pedalling, which may reflect a strategy that optimises force production as well as effectively supporting energy storage and release [198]. Finally, differences in EMG patterns between maximal (no-load) and submaximal cadence conditions have been observed, suggesting that differences exist in pedalling strategies for steady-state and sprint cycling performance [41]. Importantly, sprint cadences distinguish sprint trained performers executing sprint trials; no-load maximal conditions can be performed at upwards of 250rpm in skilled sprinters. This distinction is important in evaluating research since many studies report ‘high cadence’ conditions in the range of 120rpm, which would represent mid-range cadences of highly trained sprinters.

The ability to discriminate the effects of task-specific conditions on functional strategy is critical to a concise biomechanical assessment. Understanding changes in the overarching muscle coordinative pattern is difficult to infer from EMG analysis, while kinematic analysis clearly only provides

information about how the body is moving not how movement is being created or indeed, power produced. Limitations are further apparent in assessing performance kinetics solely at the pedal or crank. Muscle coordination is defined as being "*a distribution of activation of force among individual muscles to produce a given combination of joint moment*" [23]. Hence, inverse dynamics provides a useful solution to assessing cycling kinetics, allowing prediction of the summated muscle forces and moments acting to affect movement at any joint, while additionally allowing deconstruction of muscular and non-muscular components of force [37, 199]. Existing studies on joint kinetics have been shown to augment the findings of EMG and kinematic analysis. For example, greater knee flexion and cross-hip contributions to power production are observed in maximal as compared to submaximal conditions [39]. Changes in effort distribution through cadence [200], workload [173, 201], riding position , bike set-up [202] and fatigue [203] have been evaluated using inverse solutions, while a small number of studies in sprint cycling have already illustrated the need for further assessment of this distinct competitive category [38-40, 42, 204, 205]. Following discussion of methodological processes for cycling-specific analysis, a detailed examination of joint kinetics studies will, hereafter, be presented.

2.4 Measuring Cycling Kinetics and Kinematics

Advancing technology has made aspects of performance assessment accessible to the average rider. Most competitive riders will have an on-board personal computer providing them with the ability to monitor metrics such as speed, time, heart rate and cadence [206]. Power monitoring devices are becoming similarly accessible, although accuracy and data resolution of consumer-level products are not sufficient for high quality analysis [207]. The ability to take direct measurement from the bike makes evaluation of basic performance mechanics comparatively easy in cycling. On the contrary, detailed analysis is more challenging, with some aspects of current practices impeded by the involvement of the bike; spatial limitations of utilising motion-capture systems, for example, restrict testing protocols to using stationary bikes, while camera fields of vision to view kinematic markers are further occluded by the bike frame [208]. The importance of testing processes and equipment has

already been presented and an understanding of the issues and options provides the basis for better decision-making in this regard.

2.4.1 Kinetics

A number of devices exist for measuring external power in cycling. Most commonly used is the *Schoberer Rad Meßtechnik* (SRM) system which uses strain gauge technology integrated into the drive-side crank. Although both crank torque and angular velocity data can be extracted, the system cannot provide independent assessment of forces produced by each limb. Further, Bini et al. [207] demonstrated that crank based torque analysis overestimated power, underestimated torque and increased peak torque angle as compared to instrumentation directly at the pedal. Assessing force at the pedals facilitates independent bilateral force assessment and using triaxial sensors, provides information pertaining to the three contributing orthogonal forces, given by the vertical, anterioposterior and mediolateral components. Force transducers are further able to provide comparatively high resolution data (e.g. 1000Hz compared to 250Hz of professional level crank technology) [209].

Although instrumented pedals are more accurate, integrated pedal systems are not readily available [210]. Custom design solutions are therefore required and systems have been presented using either strain gauges or piezoelectric transducers placed into the pedal body [211], housed in mounting systems, or placed under the base of the pedal [212]. Most of the early literature in cycling used pedal systems that restricted the use of standard pedal cleats commonly used in cycling [36]. The use of cleated pedals by competitive riders preclude the use of an internal housing, and, to facilitate riders using their own shoes/cleats, systems mounted directly under the cleat system are required. Voltage output from the sensors is proportional to the deformation of the material through force applied. Determining the exact relationship of force to voltage is critical to data accuracy. At low loads the material response is non-linear hence initial processes require preloading the sensors so that applied forces occur during the linear region of the deformation-voltage relationship. Thereafter static calibration of the pedal system requires

application of known loads in each direction, with voltage output then entered into a linear regression model whose coefficients describe the sensor response along the current axis [213]. Calibration processes are almost universally described as being conducted with pedals removed from the bike and either suspension of calibrated weights [209] or force-displacement units [214] used to apply a known load. *In situ* calibration processes have not been identified in any known studies. Accurate orientation of the pedal is critical to ensuring force applied is in the intended direction and inclinometers are commonly used in this regard. In defining the relationship of voltage to force, this process fundamentally establishes the accuracy of kinetic data.

Temperature will also affect voltage readings during operation, with sensors known to drift over time. Bini and Carpes [215] recommend that a 20-30 minutes warming up period, where the sensors are powered up with no load applied, is adequate practice to ensure temperature stability. Critical sources of error are introduced due to material response. Application of force in any direction compresses the material causing expansion in the other two directions. Due to this 'Poisson effect', any voltage along the primary axis is therefore accompanied by small cross-talk voltages along the other two axis [215]. Recording the cross-talk voltages allows determination of an inverse correction matrix that can be applied as part of the calibration process [212, 216].

To provide associated angular velocity data, pedal systems utilise integrated or frame-mounted encoders or potentiometers [212, 217]. Although providing a simple solution, encoders or potentiometers have the potential to add considerable system complexity and/or alter the inertial load. Motion capture systems provide an alternative, with velocity data able to be determined by differentiation of pedal coordinate data. Pedal coordinate data can similarly be used to determine pedal inclination, providing a means of resolving the applied forces into the global coordinate system for input to subsequent modelling [209]. As will be discussed in greater detail later in this chapter, motion capture does introduce process errors through inaccuracies of tracking, digitising and post-processing of positional

data [68]. However, the use of motion capture is, in any case, a necessity for dynamic modelling and errors can be minimised by taking due care in methodological process.

With appropriate attention given to minimising sources of error, force pedal systems provide a sensitive means of directly evaluating force production of each limb [207]. Methodologies for their use are well established in cycling. Bini et al. [195] have reported that instrumented force pedals are highly reliable in testing competitive riders. While most sensors offer the opportunity to assess triaxial forces, the crank is constrained to move in two dimensions. Mediolateral forces do not directly contribute to crank rotation and are generally excluded from performance related assessments [71]. However, one critical aspect of 3D versus 2D modelling processes is locating the centre of pressure (COP) for kinetic input to the model. Carmago-Junior et al. [66] determined the errors in COP data could significantly affect calculation of joint moments in gait assessment, with errors being more prominent in distal segments, and at higher movement velocities. While COP information is directly available from force platform data, the intermediate cleat system is preventative to measuring COP directly in cycling. However, the soles of cycling shoes are designed to be as stiff as possible in order to transmit force directly to the shoe-cleat interface. Given that forces are fundamentally directed to the crank through the pedal spindle, and that the shoe-cleat interface is optimally aligned immediately above the spindle, the pedal spindle provides an appropriate axis for COP location [218].

2.4.2 Kinematics

Kinematic assessment requires positional tracking of joints and segments of the body in either 2D or 3D space. A number of systems are available either recording body position by tracking visual landmarks or identifying the body in space through magnetic sensors [70, 208]. Visual systems use either single- (2D) or multiple- (3D) camera setups to record the trajectories of anatomical and segment tracking markers allowing evaluation of translation and rotational motion in the global coordinate system [4]. The accuracy of the system can be defined by its ability to reconstruct position time-

histories, distinguish markers in close proximity, measure and represent distance and be sensitive to fine movement [70]. Capture rate and the resolution of cameras fundamentally affects accuracy and lower precision induces substantial errors, particularly where limb speeds are high and/or markers are closely positioned. Considering the potential for divergent accuracy, Fonda et al. [177] found significant differences between methods of assessing cycling kinematics and concluded that 3D systems represented the gold standard. Bouillod et al. [219] further confirmed the high reliability of two different 3D systems and recommended that the Vicon system represented the best option for research purposes. Only limited assessment has been made of existing technologies, of which no study has specifically examined the accuracy of different methodologies in examining the high velocity motion inherent in sprint cycling.

Seminal papers in cycling biomechanics have utilised stand-alone high-speed cameras, requiring subsequent manual digitisation of trajectories and manual synchronisation of the kinetic and kinematic data, adding further potential for error [36, 92-94]. Although cycling movement is predominantly in the sagittal plane, Umberger and Martin [74] compared sagittal plane kinematics captured from 2D and 3D processes and demonstrated some small but noticeable differences that may significantly impact assessment of joint kinetics. While the emergence of 3D technology presents the opportunity to improve accuracy of results, some barriers to use are evident with respect to cycling. Software for 3D systems not only provide data acquisition solutions, but provide integrated post-processing and modelling functions [70]. In such cases, kinetics are internally time-synchronised to positional data, but with force data inherently linked to acquisition by in-floor force platforms. Pedal forces must, instead, be measured on dynamically moving platforms that will be auxiliary to the primary system. The 3D systems also present some limitations in being inherently less portable. Recognising that laboratory-constrained as compared to field-based testing could potentially trade-off accuracy and validity, Elliot et al. [67] compared the Vicon laboratory-based system with a video-based set up in cricket performance and demonstrated a lower root mean square error for reconstructing joint angles in the Vicon system.

The marker set used to define the body is critical to ensuring accurate representation of segment position [64]. A number of technical models and approaches have been used and supported by reliability testing [220]. The pelvis model is particularly crucial in 3D and commonly tracks position of the left and right anterior superior iliac spine (ASIS) and posterior superior iliac spine (PSIS) [221]. However, occlusion of the ASIS marker is common in sporting movements affecting pelvic tracking [222], and would be likely in the spinally-flexed cycling position. Alternative models have been proposed [222, 223], with Borhani et al. [223] recommending addition of a cluster of markers at the sacrum to the standard ASIS and PSIS markers. Markers represent one of the most critical errors in motion-analysis, through inaccuracies of positioning on the body, tracking and coordinate reconstruction [68]. The largest error contribution is frequently cited as "soft-tissue artifacts" due to movement of the markers on the soft tissue with respect to the underlying skeleton [65]. While this is an unavoidable aspect of marker use, careful marker application processes can minimise impact. The use of bony prominences is ideal for accurate positioning being easily and repeatedly identifiable through palpation [224]. Such locations are frequently used for identifying key anatomical landmarks defining the local coordinate system of the segments.

Approximations of joint centres of rotation are particularly prone to introducing errors. While the knee and ankle have reasonably well defined and identifiable axis of rotation, the hip joint centre is less obvious or accessible. Accurate location of the hip joint centre presents one of the most demanding challenges in movement analysis techniques [225]. In 2D methodologies the hip joint centre is commonly represented as the greater trochanter. However, this approach has been shown to introduce unacceptable errors [194, 225]. Examining the impact of hip joint centre definition in cycling, Neptune and Hull [226] compared existing methodology with actual location defined by a cortical pin surgically embedded at the joint centre. Thereafter defining an alternative prediction method to increase accuracy, they confirmed the differential effects of hip joint centre location on joint kinetics during seated cycling [194]. Influence of hip joint centre position has similarly been observed in joint kinetics of weightlifting. Sinclair et al. [69] determined that moments at both the hip and knee are affected, in keeping with the

impact of hip joint centre in defining the thigh anatomical frame [225]. In 3D analysis, predictive and functional methods of locating joint centres have been assessed. Bell et al. [221] compared the accuracy of a number of predictive methods and noted some degree of inaccuracy in all approaches. Inaccurate positioning has, further, been shown to impact hip joint moments [227]. By comparing actual position located by x-ray, recommendations for the most accurate predictive position of the hip joint centre have also been presented relative to the coordinate position of the ASIS and pubic symphosis [227]. Computed Tomography (CT) scans have been similarly utilised, providing a new methodology relative to the inter-ASIS distance and PSIS-ASIS distance [228]. Two-target [221, 229] and three-target [230] relative approaches have been presented and provide the basis for hip joint centre models used in modelling software.

2.4.3 Inverse Dynamics Modelling

Inverse dynamics provides a means of predicting the net forces and moments that have contributed to producing an end movement [231]. As a representation of total muscular effort this provides the ideal means of assessing sports performance [36]. Inputs to the process are the biomechanical descriptors of the end movement, i.e. movement kinetics and kinematics, alongside anthropometric information to describe the athlete. Working back from external force production, the inverse process uses Newton-Euler equations to calculate forces and moments acting on each segment in the linked system [4]. By way of comparison, *forward* dynamics predicts outcome movements in response to given inputs [232]. The inverse solution progresses through the linked kinetic chain solving for the unknown joint reaction forces and joint moments acting on each segment. The resulting joint kinetic information then provides a means of evaluating the combined muscle activity associated with affecting movement at each joint, hence creating performance of the movement [36]. As a predictive model, a number of assumptions and process errors are inherent and the accuracy and reliability of inverse dynamics solutions have been the subject of a substantial body of research [68]. For example, magnitudes of uncertainties in torque estimates have been examined in gait analysis [68]. Of the various sources of quantifiable error, primary contributors were apparently associated with segment angles and body segment parameters. Assessment

of outcomes must also acknowledge the contribution of co-contractions of agonist and antagonist muscle groups since these can confound a direct association with muscular effort [233]. A number of methodological approaches to the inverse dynamics solution can be taken, each with their own advantages and limitations [64]. However, Cleather et al. [231] compared two of the most commonly applied approaches in assessing weightlifting movements and found inter-segmental moments to be equivalent in each case.

At the heart of the process is the definition of a linked-segment model, representing the mechanical behaviour of the interconnected limb segments. Cycling research has commonly applied a closed chain five-bar linkage with thigh, shank, foot and crank as moving links, and frame as the fixed link [82, 199, 234]. More recent studies have extended the models to include the pelvis segment [235-237] or full 19-segment representations including torso and upper limb segments [115, 238]. In comparing modelling approaches, attention given to contribution of the hip-pelvis interaction is particularly important. While some studies have assumed a fixed hip position in seated cycling [82], other models have determined that hip movement is insignificant [239]. However, Neptune and Hull [194] identified that this introduces errors. The joint reaction force at the hip represents the summated action of all forces transmitted across the hip and in such a case, the hip joint force redistributes power from the pelvis to the thigh [31]. With a fixed hip position, power and work terms associated would be zero, yet evidence supports that cross-hip power is significant and can be quite substantial in sprint cycling [116].

To solve for unknown forces and moments, a free-body diagram is created of the contributing segments [4]. Inertial properties of each segment are critical to the solution [42]. Segment lengths are defined by anthropometric measures assessed through the motion capture processes or else manually measured, while segment inertial parameters, including mass, radius of gyration, centre of inertia position, are determined by using established data. This tabular data has itself been the topic of some discussion, with adjustments to commonly used values being presented [240, 241]. Given the high limb speeds

associated with sprint cycling, Wheat and Barratt [42] hypothesised the inaccurate inertial parameters could contribute substantial errors in the model. Utilising probabilistic analysis, the authors examined the uncertainties associated with differences in inertial data. Running repeat iterations on data collected during sprinting at 120rpm and 160rpm, uncertainties at the knee and ankle joint powers were found to be insignificant [42]. Magnitudes of uncertainty in peak hip joint power were considered meaningful, but were inconsistent through the pedal stroke, particularly in the higher cadence condition. With the advent and availability of 3D imaging technologies such as Magnetic Resonance Imaging and CT scanning, there is some potential to identify individualised parameters. However, it has been acknowledged that such techniques are currently time-consuming and expensive limiting opportunities for the purposes of most research [242].

Accurate acquisition and processing of data is critical to the use of inverse solutions. Errors in the positional data are multiplicative through the differentiation process that provides the movement velocities and accelerations. In the bottom-up approach (solutions commencing at the external force and working distal to proximal) both kinetic and kinematic errors propagate through the solutions process, compounding errors at the proximal joints [4]. It is well established that the design parameters for filtering kinetic and kinematic data is critical [4, 243] and exert an impact on joint kinetic results [244]. It is imperative, therefore, through all steps of the process to assess and identify opportunities to minimise errors.

Although the majority of cycling studies have been conducted using 2D data, examinations of non-sagittal plane movement in cycling have demonstrated the importance of including triaxial pedal force components in the analysis [245, 246]. In addition to aforementioned differences in kinematic data assessed in 3D as compared to 2D [74], Quintana-Dugue et al. [247] compared 3D motion capture with 2D video in calculating crank torque from accelerational data. Validating their computational method against torque directly measured at the crank, the authors showed that 3D was more accurate, but that

improvements could be made in the 2D process by correcting for pedal marker positional data. Segmental data is more susceptible to restricting the dimensions included in the analysis. In flattened projections of 3D into two dimensions, sagittal plane motion of the joints in the camera frame of reference is not coincidental with the joint axis [76]. Joint centre definitions are similarly affected by inaccuracies in the frame of reference and lack of precision in locating the actual axis of rotation [245]. Furthermore, the effects of camera parallax distort segment lengths during motion introducing dynamic errors in the calculations that will vary with crank angle [71]. While no studies have directly compared 2D and 3D joint kinetics in cycling, Alkjaer et al. [76] demonstrated significant differences in the magnitudes of joint moments during walking when using 2D and 3D methodologies. The time-normalised profiles were consistent and inter-individual variation was not affected. Adjusting the 2D joint centre locations based on position established in 3D reduced the disparity of hip and ankle results, but notably not the knee, suggesting a greater impact of frame of reference.

The hip joint is particularly susceptible to 2D-3D differences, not only through the erroneous definition of hip joint centre, but through the contribution of frontal plane motion of the pelvis and trunk. Eng and Winter [75] compared 3D gait analysis with results of existing 2D studies and identified the significant contribution of frontal plane work at the hip relating to controlling the trunk and pelvis against gravitational load. Although cycling would not impose similar control demands, large frontal plane pelvic excursions are apparent in trained cyclists [248], suggesting similar outcomes may be observed. During the golf swing it has been shown that the impact of trunk and pelvic motions on lower limb kinematics is augmented by analysis using 3D versus 2D processes [72]. Smith et al. [72] highlight that errors are introduced by oversimplification of motion in 2D, and additionally emphasise that biomechanical coupling in the lumbo-pelvic-hip complex is instrumental in defining hip kinetics. Coupling patterns exist where movement in one direction, especially the trunk, influences movement in another [249]. Studies of coupling have previously established the impact of errors using 2D as opposed to 3D analysis [250]. Given that 9% of pedal power is transmitted from above the hip in cycling, there is potential for a substantial contribution from biomechanical coupling [116]. Movement

mechanics suggest that lumbar spine coupled motions would most likely present through axial rotation and lateral flexion, thereby introducing movement not accounted for in 2D analysis. Furthermore McGill has demonstrated that interaction of the torso musculature with the hip can act to augment the power producing capabilities of the hip joint [193, 251]. In such a case, the demand for maximal muscle power in sprint cycling would likely see a greater contribution of cross-hip power. Based upon this contention the accuracy of modelling methodology may, therefore, be of increased importance during sprint cycling, and clearly warrants assessment.

2.5 Joint Kinetics in Cycling

As a biomechanical assessment, joint kinetic analysis provides the means to predict the muscular effort involved in performance and determine the relative distribution of work across the contributing joints [13]. Seminal papers used the analysis to describe function of the lower limb around the pedal stroke [36]. Results across a large number of studies are consistent in the shape of the temporal profiles observed, although a number of factors influence the magnitude and specific timing of key transitions [7, 31, 185, 239, 252]. Hip torque is negative (extensor) across a large period of the crank cycle, until around 220°, with peak value generally around 100°, maximum torque delivery by hip extensors concurrent with the observed position of peak force delivery at the pedal. The hip then generates a small flexor (positive) torque from 220° through the remainder of the cycle as hip flexors raise the thigh towards top pedal position. Knee torque demonstrates first a positive (extensor) then negative (flexor) torque during the downstroke. Peak flexor torque occurs just prior to BDC, before rising once more towards generating an extensor torque ahead of TDC. The transition of extensor to flexor torque during knee extension in the power phase, although influenced by both cadence [25] and training [253], is a consistent observation. This provides evidence that the biarticulate muscles can aid control strategy optimisation across the pedal stroke by ‘tuning’ the distribution of net joint torques to most effectively meet task demands [168]. Gregor et al. [93] determined that their contribution during this phase provides economy of motion; the two joint hip extensors continue delivering extensor moments without adding demand on the knee extensors to overcome the flexor torque this enforces at the knee. Ankle torque is

negative (plantar flexor) until close to 200°, remaining near zero through the remainder of the cycle. Similar to the hip, peak ankle torque is observed around 100°. However, its profile is seen to closely follow that of the normal pedal force and it appears that the ankle joint acts to optimise stiffness and maximise energy transfer from limb to crank [254].

Joint power profiles are predominantly positive around the crank cycle and, indeed, it has been shown that most of the mechanical energy associated with cycling is produced by concentric contraction of the lower limb muscles [255]. Profiles at each joint are seen to reflect positive generation of energy through the first half of the crank cycle, whilst the knee additionally produces a secondary positive power peak during the early part of recovery. Brief periods of energy absorption occur at the ankle immediately following TDC and BDC, between the two positive power phases in the knee, and through most of recovery in the hip [256]. Eccentric action of the ankle plantar flexors has been confirmed [92], while Hawkins & Hull [257], assessing the existence of stretch-shortening cycle action during cycling, further determined that both hip and knee extensors demonstrated brief periods of eccentric to concentric action. In many studies, however, a further (much smaller) positive power phase at the hip is seen during the second half of the crank cycle, the generation of which is associated with the torque accompanying hip flexion towards TDC [252, 258, 259]. Such differences in profile description are indicative of high variability between both subjects and test conditions. In fact, variability is a well-documented feature of biomechanical studies of cycling, with disparities particularly prevalent at the hip [20, 260].

Whilst discrepancies are present in the relative peak torque values of each joint, the hip and knee peak torques are larger than those of the ankle, reflecting their primary roles in propulsion [261]. Cycling has been suggested as a knee dominant movement, with evidence suggesting that the knee contributes over 50% of the mechanical energy across a full crank cycle [256]. Additionally, over 6% of the work across the crank cycle is derived from hip joint reaction force, confirming the contribution of musculature from above the hip. Inspecting the crank cycle over the two primary phases of down- and up- stroke, Ericson

et al. [92] apparently confirmed knee dominance of the cycling movement, finding the knee extensors contributed 39% of total positive work, hip extensors 27%, ankle plantar flexors 20%, with smaller contributions from the knee flexors (10%) and hip flexors (4%). However, more recent studies have disputed these findings, suggesting that hip extension provides the greatest contribution under a range of cycling conditions [39, 40]. Mechanical power analysis [261] provides a means of interpreting the task-specific role at each joint; net hip and knee torques are seen to generate energy, while ankle *transfers* energy, with gravity augmenting the limb-crank energy transfer in both phases of the crank cycle. Further optimisation modelling has confirmed these findings [262], while additionally providing evidence of distinct roles for the biarticulate and uniarticulate muscles in this process [158]. Conflicting demands of energy transfer at the hip, knee and ankle into translational movement at the pedal see the biarticulate muscles adjusting the relative distribution of net moments, while uniarticulate muscles are activated only when they are in a position to shorten and hence generate positive work. This underpins the belief that biarticulate muscles function more in control and transfer of force, while the uniarticulate muscles provide primary force production [95].

2.5.1 Effects of Workload and Cadence

Early research by Kautz and Jorge [37] presented a method whereby inverse dynamics could be used to decompose pedal force into muscular and non-muscular (gravitational and inertial) components. This methodology subsequently demonstrated that manipulations of cadence affected only the non-muscular contribution, muscular moment remained relatively constant over most of the crank cycle. Other studies [199, 234] have alternatively decomposed the joint-specific moments into kinematic moments, related to the acceleration of the limb segments, and quasi-static moments associated with generating pedal force. Static moments are dominated by the ankle, demonstrating that most of the ankle torque is generating pedal power, with little needed to angularly accelerate the foot. In contrast, the combined inertia of the proximal segments affect an increasing kinematic component at the hip and to a lesser extent the knee, demonstrating their greater dependency on angular acceleration. The findings were clearly related to changes in cadence. At low cadences pedal power must be generated by greater static

contributions, while high cadences demand an increased kinematic component. Both instances may require a high total torque but by disparate contribution, suggesting an optimal cadence exists whereby both contributions are minimised at each joint. The impact of limb inertia was later underscored by Li and Caldwell [25] while examining the impact of different riding conditions on joint kinetics. Changing pedal rate induced greater effects at the hip than knee or ankle due to the interrelation of inertial properties with cadence. The impact of inertial properties of the limb segments on coordinative pattern have been confirmed elsewhere in comparisons of cycling performance in children versus adults [263]. Children are seen to alter joint torque distribution to compensate for the lower segmental mass and moments of inertia, hence substantially smaller gravitational and inertial components of pedal force.

A number of studies have examined the impact of cadence in steady-state conditions, presenting somewhat disparate results. However, some distinction can be made in the study methodologies. Evidently joint moment will increase with increasing work demands [36], hence a change in workload affected by altering test cadence will confound clear interpretation of results. Wangerin et al.[36] examined changes in submaximal workload with two cadence conditions (60 and 90rpm) in a small cohort (2 pro and 2 recreational) of cyclists and determined that higher workloads increased joint moments while higher pedal rates decreased them. Ericson et al. [252] reached similar conclusions, although found response to cadence more robust. Broker and Gregor [256] supported this finding examining the source and transfer of power through the lower limb and finding results were sensitive to workload but insensitive to cadence over a small range of 90-110rpm. Examining the interaction of cadence and workload through five different combinations, they did, however, observe that the hip joint was substantially affected by both cadence and workload.

Although the absolute magnitudes observed at each joint are certainly affected by changes in demand, of greater interest is whether joints respond differently to changing conditions. Specifically, is the relative contribution of each joint impacted by different riding conditions? This would suggest a change

in the distribution of muscular effort, i.e. control strategy, under differing performance demands. Ericson et al. [252] explored a range of cadences (40-100rpm) and found no impact on relative distribution of work. Notably, the study did not use subjects experienced in cycling and the methodology did not control workload. In such a case, increasing cadence affected a concurrent rise in power output. However, Bini et al. [200] similarly found no significant change in relative contributions of the joints to total work, although work produced by the knee did increase at the higher cadence. Here, the study concentrated on only single small increment and decrement around freely chosen cadence. Further examination of two lower cadences in a separate study by the same group found that ankle joint contribution increased at 70rpm as compared to 40rpm. Alternatively, studies of Mornieux et al. [201] and Sanderson et al. [264], maintained a fixed power output while assessing performance at 60, 80 and 100rpm, observing a decrease in the contribution of hip moment and increase in knee moment with cadence, with the ankle contribution remaining unchanged. Increasing power output while holding cadence fixed resulted in the opposite effect; a greater contribution was observed at the hip, and less at the knee in higher power output conditions, ankle again remaining unaffected. Outcomes were the same in both normoxic and hypoxic conditions, leading authors to assert that the coordinative pattern of cycling is robust to changing environmental conditions. The same outcomes have, more recently, been confirmed in joint powers of recreational cyclists over a cadence range of 60-110rpm [265].

The wide ranges of pedal rates that have been utilised by existing studies on the impact of cadence make direct comparisons somewhat difficult. Confounding results further, Hoshikawa et al. [258] examined cadences from 40rpm up to 120rpm and found the hip increased its contribution to total work, while there was a decrease in the contribution of the knee. Limited influence of cadence observed at the ankle appears to confirm the role of ankle musculature is stable across different demands, while those at the hip and knee apparently are compensatory in maintaining force production. The Hoshikawa et al. [258] study highlights the impact of training status, since relative values at the hip and ankle in trained cyclists were significantly lower while the knee was higher than those of non-cyclists across almost all cadences. This may be indicative of differences in coordinative pattern with training status and hence skill level.

In fact, while assessing improvements in pedalling technique, combined EMG and joint kinetic analysis has suggested that more effective directing of pedal force round the crank cycle is accompanied by a reorganisation of joint torque distribution and altered uni- and bi- articulate muscle patterns [253]. Unfortunately, most research in joint kinetics has utilised cyclists with limited or no training, with few studies examining those of very high training status. It is unclear whether further changes in joint kinetics would be observed in participants of advanced skill levels.

The ability to maintain effective pedal force direction with increasing limb speed is certainly a known aspect of high performing cyclists. High cadence pedalling presents coordinative challenges. Neptune and Herzog [94] tested competitive riders over cadences up to 120rpm and used the decomposition technique to assess negative work. The positive correlation of negative muscular work with cadence may suggest a reduced ability to effectively coordinate function at higher limb speeds. Ettema et al. [170] examined the phases of joint power production in competitive cyclists tested over five cadences (60-100rpm) and found the time course of muscle power was shifted later in the crank cycle with increasing cadence. The observance of a fixed time lag and onset of muscle activation unrelated to cadence, dispels the association of activation dynamics and electromechanical delay to cadence-related coordinative changes, leading to the aforementioned assertion of the influence of limb inertia. The phase shifts further means that cadence does not just alter performance along the FV relationship of muscles, but more globally the coordinative strategy i.e. pedalling technique.

These findings are instrumental in determining the need to distinctly assess sprint performance. Fundamentally the cadences observed in sprint competition extend beyond those assessed in the aforementioned studies, with sprinters likely being more skilled in working at these limb speeds. Further, attempts to interpret study findings in cycling are commonly approached from the perspective of efficient movement. For example, Marsh et al. [233] and Redfield and Hull [82] maintain that the observed trade-offs between hip and knee contributions reflect an optimisation strategy and that an

‘optimal pedalling cadence’ exists globally minimising work demands across the joints. However, since sprinting requires maximal muscle output, an ‘optimal cadence’ would represent operation at a global maximum of combined power-velocity relationships of muscles crossing each joint. The task demands in sprinting would likely affect a strategic alteration of muscular effort in this regard. In all-out sprint conditions, peak crank torque is higher at the end of the downstroke phase at leg speeds of 200rpm as compared to those at low cadences (around 80rpm), while distinct patterns of energy transfer as well as unique relationships in knee and ankle angles are also evident in high cadence pedalling [198]. This provides additional evidence supporting a distinct optimal control pattern for high speed pedalling.

Only two studies have made direct comparison of maximal and submaximal cycling conditions, with results providing strong evidence of the distinct neuromuscular characteristics of sprint cycling performance. During isokinetic performance at a single cadence of 90rpm, Horscroft et al. [38] determined that ankle joint power and cross-hip power (power transferred from the upper body) over a full crank revolution were substantially greater in maximal conditions. Elmer et al. [39] compared performance in a range of power outputs including maximal power. Separating the contributions of joint extension and joint flexion across the revolution and using regression analysis across power conditions, results show an increased dependency on knee flexion power at high power outputs. Notably values show high inter-individual variability highlighting individuality of coordinative strategy in this population. Previous study determined that emphasising the pulling action (i.e. flexor action) during the upstroke improved pedal stroke effectiveness to the detriment of efficiency [159]. Elmer et al. [39] proposed that the pulling strategy is, instead, beneficial under the altered task constraints of sprinting, since performance is driven by demands for maximal muscle power rather than economic movement. Subsequent 3D examination of workload and cadence in cycling supports these results, observing an increased knee flexion moment with cadence, although knee abduction moments remained unaltered [180]. Increased contribution of power transferred across the hip in maximal pedalling is a consistent finding of studies [38, 39]. While it is acknowledged that additional upper body movement is counterproductive when seeking to be efficient in submaximal performance, the upper body conversely

augments power production in maximal conditions. Elmer et al. [39] further isolated the effects of maximal versus submaximal pedalling in assessment of duty cycle, showing that maximal conditions resulted in an increased time in extension of all lower limb joints, aiding power production. Finally, repeating the maximal trial at a second higher cadence (120rpm) the authors critically acknowledge that leg speed impacts results (significant changes being particularly prominent in hip extension) and that training status may be a factor resulting in differences between studies. These suggestions underscore the need for a more highly specified evaluation of joint kinetics in sprint cyclists.

Few studies have examined cycling performance at extremely high cadences. A single study assessed joint-specific power in maximal and submaximal high cadence cycling as a part of a case study of an elite sprinter [41]. Results confirm the predominance of hip extension power, while noting a significant contribution of hip flexion and the importance of hamstrings activity in transferring mechanical energy between the knee and ankle. Martin et al. [266] compared joint-specific power across a range of cadences from 58 to 190rpm during maximal conditions and found that, while the knee followed the quadratic power-cadence relationship commonly observed at the crank, peak hip power continued to increase to the highest of pedalling rates, whereas power at the ankle decreased at higher limb speeds. In a separate study, the same laboratory also confirmed changes in relative contributions of each joint with cadences across the same range [267]. A further study, comparing relative contribution in maximal trials between 60 and 120rpm, confirmed only the decline in hip and ankle power [116]. Collectively, these results suggest changes in joint kinetics are more apparent at the extremes of high cadence pedalling. Since coordinative ability is challenged at high leg speeds and given the inherent variability in pedalling technique at high cadence, it may be that that, as cadence rises, the cyclist is increasingly unable to maintain the same recruitment strategy [22, 173]. However, to date, assessment at such cadence ranges has been restricted to non-sprint trained athletes and it is unclear whether skilled performers would show similar outcomes.

Research indicates that there may be unique interactions of joint-specific muscle function in maximal cycling conditions. Over a cadence range of 60-180rpm, McDaniel et al. [205] observed a decrease in relative ankle plantar flexion power and increase in hip extension and knee flexion powers with increasing cadence, knee extension power remaining unaltered. The study further observed that hip excursion increases while ankle excursion decreases with cadence. The interaction of the hip and ankle appears to support findings of Fregly and Zajac [261] who previously determined that the ankle and hip work synergistically in acting as a source (hip) and channel (ankle) for energy delivery to the crank, where the knee works more independently. Control of ankle kinematics has, elsewhere, been proposed as providing a means of affecting time in power-producing limb extension through the crank cycle [40]. Hence it would seem that the hip-ankle interaction provides both kinetic and kinematic benefit. Results of this study highlight that the factors differentiating conditions in maximal as compared to submaximal performance are likely unique to each joint. Distinct joint excursion and angular velocity profiles at each joint as cadence changes, affect both the effect of time available, hence muscle active state, as well as operation over the power-velocity relationship, producing unique constraints. With trade-offs observed in hip-knee power production and hip-ankle energy transfer, such studies highlight the importance of a joint-level analysis in understanding changes in performance at the crank.

Given the apparent importance of cross-hip power in maximal pedalling conditions, it is interesting to note that only a single study has directly examined upper limb kinetics. Costes et al. [115] determined that riding position and work conditions influence the magnitude of joint kinetics in the upper limb joints. Standing position and higher power outputs were associated with higher magnitude of upper limb kinetics, although pulling (not pushing) actions were only affected. Pulling on the handle bars is acknowledged as a key part of the sprint cycling action, with the authors concluding that a greater pulling action is needed to prevent body elevation when pedal reaction force becomes greater than body mass. This may suggest that upper limb action is not a constituent part of cross-hip power, but that alternatively, cross-hip power represents the reaction force of the back and contralateral leg.

2.5.2 Effects of Riding Position and Bike Set-Up

Since task specificity appears to have a critical impact on the distribution of muscular effort, it is unsurprising that changing the relationship between the bike and rider has been shown to affect joint-specific kinetics. Standing out of the saddle reduces the constraints on the hip position and indeed is seen to alter function further down the kinetic chain. In submaximal riding, Li and Caldwell [25] noted that a transition to standing affects an increase in ankle and knee moment and concomitant decrease in hip moment. The importance of the upper body in sprinting is further emphasised by the finding that a transition from seated to standing affects only the contribution of cross-hip reaction terms to crank power [73]. Limited information from this abstract, however, precludes further comparison with the submaximal study. Altering upper body position by changing handgrip position and hence trunk angle also appears to affect joint power. Specifically, hip power is increases when riding in the dropped handlebar position as compared to riding on the tops, with knee power increasing and ankle remaining unchanged [188].

Seat height represents one of the most commonly explored changes to performance position. Altered seat position affects knee kinematics, hence operation of muscles crossing the knee joint, to a greater extent than either the hip or ankle. Research confirms a significant impact of seat height on knee joint kinetics, although the extent of height changes appears to impact results. While Horscroft et al. [154] found joint power distribution was unaltered by lowering saddle height, Bini et al. [202] found that knee contribution to total work was only significantly different between a saddle height compared both above and below the riders' accustomed position. Tests of saddle heights above and below the position recommended as optimal in maximal cycling suggests that a lower saddle height is related to a reduction in power output, with knee, but not hip or ankle, moments decreased in this position [237]. Forward-aft position of the saddle similarly shows pronounced affects at the knee, affecting shear, but not compressive, forces present at the joint [268].

The interaction of seat height and foot position has been distinguished through the flexor and extensor phases of muscle action [30]. Seat height affects knee flexor, but not extensor, contribution, while an associated change in the anterior-posterior position of the foot on the pedal additionally altered dorsiflexion moment. A later study by the same authors [269] determined that seat height and foot position actually had a small effect on joint loading at both the knee and hip in some subjects suggesting some variability in relative contribution of the joints. The shoe-pedal relationship introduces a number of variables impacting ankle elevation, including pedal platform height, shoe sole thickness, cleat thickness and ankle position above the foot sole. Pedal platform height adds complexity to the kinetic analysis since the centre of pressure of the foot is not coincident with the pedal spindle axis [270]. Although pedal force application is unaffected by platform height, varying the height of the platform *does* have an interactive effect with cadence that affects joint moments by up to 13% [270]. Combined measures such as 'posture height', being saddle height, crank length, shoe cleat position, and saddle setback and 'posture length' additionally associated with handlebar reach, are also seen to affect performance at the pedal. Support for this contention can be found in the study by Hayot et al. [236] where a preferred or forward position resulted in a larger knee power than a more backward position.

Crank length also affects the kinematic relationship of the linked-segment model. With cadence held constant, a longer crank affects pedal speed, hence muscle shortening velocities of uniarticulate muscles, while crank length alternatively affects the excursion length of the same muscles [11]. Barratt et al. [7], compared two submaximal conditions where pedal rate and pedal speed were alternatively held constant allowing independent assessment of the effects of pedal speed and crank length on joint powers. Across cranks lengths from 150-190mm, the authors found increases in pedal speed and crank length both affected increases in knee and hip angular excursion. While joint moments and powers were less affected, a trend for decreases in knee extension power and increases in hip extension power was observed with longer cranks. As previously discussed, Martin [1] dispelled myths around the influence of crank length on maximum pedal power by demonstrating that the counteractive effects on pedal speed and pedal rate must be considered in the analysis. Barratt et al. [204] extended this approach in

examining the effects of crank length on joint-specific power in maximal conditions. Examining the same range of crank lengths as in the submaximal trials, at a constant pedalling rate (120rpm) crank length effected small but significant differences in hip and knee joint powers between the extreme crank length conditions. However, when cadence was optimised for maximum power production crank length had no significant effect on any lower limb joint.

2.5.3 Effects of Fatigue and Other Influencing Factors

Mechanisms of fatigue are largely distinct in steady-state and maximal conditions. However, the neuromuscular effects of fatigue are similarly explained through impaired force production and movement control. The effects of fatigue on coordinative pattern have been demonstrated [29, 271] and the relationship of joint kinetics to muscular effort provides similar insight into muscular stress [203]. While Amoroso et al. [272] observed changes in kinematics and kinetics of performance during a constant cadence trial to exhaustion in competitive cyclists, the study failed to examine consequences at joint level. However, conducting a similar protocol, Bini et al. [203] observed that ankle joint contribution decreased while total absolute moment and hip and knee moments increased at the end of the trial. It should be noted that cadence also declined by the final stages of the trial which may confound results. In another study, Bini and colleagues [160] also assessed the impact of an incremental test to exhaustion, thereby altering conditions to increase workload during the onset of fatigue. Here knee joint contribution increased, while total absolute joint moment, knee moment, plantar flexor moment and hip flexor moments were also observed alongside changes in hip and ankle kinematics.

The impact of fatigue on joint kinetics in maximum conditions has also been examined. Elmer et al. [273] compared joint kinetics in maximal sprinting pre- and post- performance of a 10-minute time trial at constant cadence. Joint-specific powers were non-significantly altered by the end of the time trial. However, the post-time trial maximal test affected a decline in joint powers, with ankle plantarflexion and knee flexion showing greater fatigue than both hip extension and knee extension. With maximal

performance affected and the time trial submaximal performance largely unaffected, the authors conclude a distinct functional consequence of fatigue on maximal performance. However, it should be noted that fixed duration time trial present substantially different performance strategies than time trials to exhaustion and it is unclear whether the time trial would have represented similarly fatiguing steady-state conditions. Assessing the effects of fatigue induced by maximal pedalling conditions themselves, Martin and Brown [40] compared joint-specific power at the start, middle and end of a 30s all-out sprint at 120rpm. Relative ankle plantar flexion power declined by the middle interval and was significantly less than knee flexion and hip extension power in the final interval. Knee extension power was also reduced below that of hip extension by the end of the trial, where knee flexion power was not significantly different from the power achieved during hip extension. Time spent in joint extension also declined.

Fatigue appears to most critically affect the ankle joint. Studies have highlighted the unique role of the ankle in being responsible for force transference. Mornieux et al. [201] suggested that the ankle musculature should function to maximise joint stiffness to aid function in this regard. A reduction in ankle moment suggests impairment of this functional capacity during fatigue. Interestingly, in submaximal conditions the ankle appears to increase range of motion during fatigue, while in maximal conditions ankle joint excursion is reduced. The increased range of motion observed at the ankle during steady-state fatigue has been suggested as a mechanism to offset the decline contractile capabilities by increasing lengthening velocity [203]. Alternatively, in all-out conditions there is less movement in the ankle that would minimise power loss while further reducing the degrees of freedom and constraining control optimisation to support maintenance of maximum power [40]. While the influence of central and peripheral components of fatigue are likely to be present in both submaximal and maximal performance conditions, the differences in compensatory mechanics support the assertion that strategic alterations in joint are at least a component part of functional changes observed at this joint.

Inverse dynamics approaches have also been used to examine a number of other factors influencing the distribution of muscular effort in pedalling. A noncircular chainring apparently produces a similar joint kinetic profile to standard circular chainring [274], although the *extent* of chainring ovality can affect both knee and hip joint kinetics; chainrings of high ovality reduces relative knee power and increases contribution of the hip, suggesting benefit to power production [275]. Elmer et al. [276] examined eccentric cycling in a recumbent position finding that most of the power absorbed was at the knee, while a subsequent study using an eccentric protocol as a fatiguing intervention prior to concentric performance, confirmed knee extension as the greatest power absorber followed by hip extensor [277]. During subsequent maximal concentric pedalling knee extensor power was significantly reduced whereas hip extensor power was unaffected. Several researchers have compared assisted single leg and double-leg cycling. For example, Bini et al. [278] report reduced hip extensor moment and increased knee flexor moment in single-leg assisted pedalling as compared to two-leg. Elmer et al. [279] compared counterweighted and non-counterweighted single leg pedalling with the bilateral condition and observed that relative ankle plantar flexion and hip extension work was greater with two-legged pedalling. Counterweighting the passive pedal in single-leg pedalling has been reported as providing a means of compensating for the inertial effect of the contralateral leg. While single-leg training apparently provides beneficial peripheral adaptations, joint kinetic analysis, therefore, confirms that the coordinative pattern of pedal stroke is altered in this condition. Such outcomes highlight that joint kinetic analysis not only provides a means of describing control differences as a result of environmental conditions, but can help inform decision-making in training prescription for cyclists.

2.6 Joint Kinetics in Weightlifting

A number of weightlifting exercises are commonly used to improve the strength capacities of cyclists. The characteristic triple extension of hip, knee and ankle during primary power production of the downstroke in cycling has led to the assumption that squat-based patterns and variations are ideal resistance training exercises for cyclists [54]. However, the contribution that each joint makes to muscular effort, hence coordinative pattern of lift execution, may be far from similar. Different exercise

variations have been shown to develop unique joint kinetic distributions reflecting the kinematic pattern and position of the body through the lift, as well as kinetic differences affected by position and magnitude of the external load [56-58, 280]. In a given exercise, altering limb segment angles through changing aspects of technique, such as movement timing or foot placement [59-61, 281], can further affect the relationship of force vectors at any joint to ground reaction force. A number of studies have observed that both loading conditions and body position impact the effective moment arms at each joint, directly impacting net moment [56, 91, 282]. In such a case, an understanding of the characteristics of each exercise and ramification of changing their execution are critical to informing their application in cycling.

External load is accepted as being the primary means of changing demands of any resistance training exercise [283]. Indeed, load lifted is seen to be the main determinant of intersegmental moment at the hip and vertical forces at each of the lower limb joints [284]. However, the three joints are seen to respond uniquely to loading. Bryanton et al. [61] showed that relative muscular effort (RME) of the hip extensors and ankle plantarflexors increased with load in the squat, where loading had little effect on knee extensor RME. Split stance exercises are similarly consistent. In the lunge exercise, load has little impact at the knee, while a linear increase in work is observed at the hip and ankle [285]. Knee response to loading also appears distinct in explosive exercises. During the pull phase of the clean both hip extensor and ankle plantar flexor moments continue to increase with load beyond where moment is maximised at the knee [57].

To help improve athletic performance, weightlifting studies have frequently examined the optimal load for developing power [283, 286]. However, Farris et al. [96] determined that peak external power, observed between 40-60% 1RM (a load range also commonly referenced for power development in the literature), actually represents a compromise in powers produced by each joint. The hip, knee and ankle were found to be optimised for power production at unique loads. While Farris et al. [96] observed an

optimal load for knee power as 40% and hip 60% in the squat, Jandacka et al. [287] observed in the squat jump that knee power was maximal at 0% of 1RM load, ankle at 70% of 1RM, while hip produced a similar power across loads from 0-70% 1-RM. In contrast, Moir et al. [58], observed that the unloaded condition maximised knee power, but found a unique maximal of 42% of 1RM at the hip. Jandacka et al. determined that differences may be due to lack of control of squatting depth in the Moir study, which would affect both joint torque and velocity developed through the concentric phase of the lift.

Understanding the effects of changing the technical conditions of different exercises can aid their application by defining conditions that will maximise muscular effort of each joint [288]. However, ultimately muscular effort must be related back to the external power being produced and some consideration has been given to whether inference in joint powers could be made from examining external power only. Kipp et al.[289] examined the correlation of joint powers to four common measures of external power output in the clean exercise performed at various load, and determined that the hip and knee showed the highest correlations to external power calculated by the work-energy method at a load of 85% of 1RM. Peak sum of all joint powers was only correlated to external power calculated by the impulse-momentum method at loads of 75 and 85% of 1RM. Results, therefore, suggest joint powers are only predictive of external power outputs at higher loads and that the contribution of the ankle may be more variable. Such findings may be equally true in other exercises. Regression modelling of load lifted on hip and knee joint moments in the deadlift, further suggests substantial variability in response [290].

The relative importance of the hip joint in squat-based movements is reported by a number of studies. With hip extensor moments observed across the concentric phase of squat-based patterns, a consistent linear relationship has been reported with load, in both standard lifts (such as squats, deadlifts and lunges) [291], as well as explosive exercises such as jumps and Olympic lifts [57, 97]. Results highlight an increased proportion of hip involvement as load increases. The hip is also observed as being the

limiting joint in squatting [292], while jump height is seen to be most critically dependent on hip power [293]. In contrast, the knee and ankle appear influenced by more than just load and both exhibit distinct patterns that are affected by load and load vector, technique and body position [291].

The ratio of hip:knee contribution is commonly used to classify exercises as so-called 'hip dominant' or 'knee dominant' [294]. The actions of the two primary power producing joints are frequently observed as interacting in a compensatory manner with altered lifting conditions. Bryanton et al. [288] observed that hip extensor activity, particularly the relative contribution of gluteus maximus versus hamstrings, influences the relative muscular effort of the quadriceps, thereby affecting knee moments. The knee has been the sole focus of a number of studies in both standard and explosive lifts, with particular interest for injury and rehabilitation considerations in performing weightlifting movements [282, 295-297]. Body position uniquely affects the moment arm and kinematics of the knee through the combined effects of thigh and shank angle, such that even small technique variations can be either beneficial or detrimental to knee loading [55]. Unlike the hip, during the concentric phase of lifts the knee is commonly seen to observe phases of both flexor and extensor moment specific to lift conditions [56]. Some explanation may be provided in understanding the role of the two-joint muscles through the course of the lift. While acting in limb control and positioning as well as energy transfer between the segments, their specific lines of action could actually, by vector summation, act to augment or diminish net extensor moments [298]. Their role can certainly account for discrepancies in results and a potential limitation of standard inverse dynamics techniques is the inability to adequately account for their actions [95].

Ankle function shows substantial variability during various lifting conditions. A comparison of the sumo and conventional deadlift found that, while the conventional position affected a net plantar flexion moment at the ankle, the sumo position alternatively affect a net dorsiflexion [280]. The study similarly showed that direction of ankle moment is affected by stance width in squatting [59]. Ankle moments

are frequently shown to be substantial in weightlifting, in many cases greater than those at the knee [57, 285]. The structure of the triceps surae complex is certainly related to the demands for high force production at this joint [299], and, in fact, vertical forces in the deadlift are seen to be highest at the ankle joint [284]. However, the relationship of ankle joint kinetics with load appears highly variable [291]. Both standard lifts and explosive exercises have either no relationship [291], non-linear [63] or linear [57] relationships of load to joint torque. In contrast, the range of motion of the ankle distinctly affects power development. Kipp et al. [57] noted that joint excursion was much smaller in the ankle than at the hip or knee, and, with joint angular velocity observed as being much lower, ankle joint power was reasoned to be more highly dependent on joint torque profile. In support, Suchomel et al. [300] showed that where the knee and hip joint angular velocities were responsive to load changes in both jump squat and hang clean, the ankle was somewhat insensitive.

Disparities are often noticeable in examining the outcomes of comparable weightlifting studies and appears only partially explained by differences in specific conditions of study design. A degree of inter-subject variability is frequently reported [298]. Expertise in execution has been shown to impact both the kinematic and kinetic profile of the lift being analysed. Assessment of deadlift performance in skilled and unskilled competitors in powerlifting competition noted that unskilled performances showed significantly higher variability in linear and angular accelerations [284]. Joint kinetics terms showed similar relative distributions but differences in the magnitudes observed at each joint. However, Enoka [63] determined that the ability to lift heavier loads was not simply due to a scaled increased in joint moments but was critically impacted by the temporal organisation of intersegemental kinetics. Examining the pull phase of the clean, a successful lift required generation of sufficient joint power and optimal organisation of the phases of joint power production and absorption. The complex technical aspect of weightlifting exercises has been argued as being a barrier to application in athlete development [301]. Certainly, successful lift completion in clean pulls demands effective timing and relative magnitudes of key interactions of kinematics and kinetics [302]. Such differences are apparent even in

highly trained competitive lifters, research demonstrating that control of knee position in the snatch discriminates performance [303].

As in cycling assessment, methodological concerns are evident in weightlifting analysis. Re-examining the deadlift technique comparison, Escamilla et al. [280] observed a number of differences between results of 2D versus 3D analysis. Utilising both 2D and 3D processes in a further study of squatting, the group also showed that 2D processes incurred errors in hip angle, ankle and knee moment arms and hence moment terms [59]. The differences were significantly related to the stance width, with wide stance observing greatest errors. In fact, knee moments were alternatively greater in wide stance in 3D processing while they were less than narrow stance in 2D. Comparisons of stance elegantly highlight the effects of 2D error through both camera parallax and misalignment of the joint coordinate axis. A narrow stance places the thigh and shank segments in closer alignment with the anterior-posterior coordinate axis, where a wide stance angles the segments outwards. Results of the two studies led authors to conclude that 3D analysis is more accurate. As with cycling, squat-based weightlifting exercises are predominantly sagittal plane. In contrast, however, bilateral, and to a lesser extent unilateral, weightlifting exercises have far less hip movement out of the sagittal plane. Consequently, errors associated with hip joint centre location are less prominent. Sinclair et al. [69] compared four hip joint centre locating techniques assessing the back squat and significant differences were only observed in the frontal and transverse planes. Consistency of inverse dynamics methodologies have also been confirmed in weightlifting, with equivalent outcomes found using different analysis of joint kinetics in push jerks, vertical jumps, and squats [231]. As crucial footnote to evaluating study outcomes, an examination of the validity of joint kinetic measures observed that, while average and peak torque values were both significantly correlated with task objectives in work-related tasks, only mean values were significantly correlated in power-related tasks [304].

2.6.1 Standard Lifts

The squat has been most commonly examined and is the foundational movement pattern from which a number of exercise variations are derived through manipulation of direction and positioning of the load, limb position, joint excursion, as well as relative contribution of one or both legs [62, 91]. In squatting with no additional load, peak knee moment is greater than that of the hip or ankle [231]. In contrast a barbell squat is predominantly loaded at the hip, to a lesser extent ankle and least knee, although some individual variation is apparent [298]. Other studies have equally found the knee outweighed contribution of the ankle but not the hip [305, 306]. As previously noted, the influence of load is certainly discriminatory. A consistent linear increase of the hip moment with load, alongside less consistent effects at the knee and ankle can affect a shift in relative joint dominance at particular loads. The depth of the squat also influences outcomes, with a squat to parallel depth demonstrating lower relative muscular effort in both the hip and knee as compared to full range squat (below parallel). The interaction of load and depth can, therefore, substantially alter the relative effort present at each joint [61].

Small differences in left-right magnitudes of net joint moments can be present at each joint in squatting [305]. Hip and ankle contribution have been observed as having a greater magnitude on the left, while knee values were greater on the right [305]. In this study, all but one participant stated right leg dominance, which may suggest an underlying relationship to coordinative strategy in a stronger leg. An examination of squatting in long jumpers alternatively noticed that the take-off leg tended to show greater extensor moments at the hip and ankle and non-take off leg at the knee [306]. Whilst this appears to be at odds with the squatting study, the training/experience status and depth of the squat are quite distinct in each study. Both of these factors would, therefore, critically affect outcomes. These results serve to illustrate the impact of population and technique of execution on joint kinetic distribution. Notably no studies have directly assessed differences with a left versus right driving leg in unilateral or split stance lifts.

Key squat variations are affected by changing the bar position hence centre of mass of the system, which consequently affects joint loading. As compared to a posterior shoulder (back squat) bar position, an anterior shoulder (front squat) positioning of the bar affects a concomitant change in hip:knee relationship. Examining knee function only, Gullet et al. [55] determined that, while overall muscle recruitment was the same, the back squat had a greater knee extensor moment than the front racked position. In contrast, Russell et al. [307] found similar knee extensor demands but determined that changes in trunk inclination affected the trunk extensor moments more prevalently. Unfortunately, the relative contribution of the knee to the other joints was not presented in these studies. Yet, while the back squat has been shown in a number of studies as being dominated by hip extension [288, 298, 306, 308], knee moment is seen to be highest of the three joints in front squat [309]. However, the joint loading pattern of the exercise can be affected by the direction of the load vector applied. By replacing a standard free weight (barbell) with a flywheel/cable pulley load, Chiu et al. [309] showed that the cable lift created greater contribution of the hip and ankle while decreasing the knee contribution. In free-weight conditions there are constraints on the body position based on the necessity of maintaining the system centre of mass between forefoot and heel [310]. The utilisation of pulleys or equipment that constrain the direction of load application, provides a means of compensation by introducing reaction forces against the system. Employing the guided bar tracks of the Smith machine, research has shown that joint load distribution can be modulated by changing the relative body and equipment inclination; a backward (forward) inclination decreased (increased) knee torque and increased (decreased) hip torque [310].

Joint kinetics of the deadlift further demonstrate the impact of altering the load vector. The exercise is characterised by the bar being lifted from a ‘dead’ position from the floor up, with the external mass therefore non-axial and below the midline. The deadlift is commonly considered hip dominant and studies have confirmed a far greater contribution of the hip as compared to either the knee or ankle, citing impact of a relatively large moment arm of the bar to this joint [284]. However, different bar and stance variations affect specific distribution of muscular effort. Cholewecki et al. [290] examined

performance of the lift in powerlifters who generally use either the wider stanced, ‘sumo’, position or conventional position. The hip contributed 671Nm in sumo and slightly greater hip moment at 713Nm in conventional position, both far in excess of the knee which contributed only 18Nm in each case. Brown and Abani [284] confirmed the predominance of the hip in adolescent powerlifters, here noting that the ankle, in fact, exceeded the contribution of the knee in both lift-off and knee-passing phases of the lift. Further studies have confirmed that the two predominant lift styles affect both knee and ankle contributions, as previously mentioned the ankle notably observes a net dorsiflexion (not plantarflexion) moment, in the sumo lift [280]. Comparison of different bar types in the deadlift shows further distinctions. Hexagonal versus straight bar position affects the loading vector of the lift as well as lift kinematics. Although greater load can be lifted with the hex bar, hip and ankle moments are lower, and knee higher, as compared to a straight bar lift [56].

In addition to the impact of the load position, changes in feet and knee positions significantly affect joint moments. A comparison of stances in an unloaded squat used principle components analysis to determine that a wider stance increased knee flexion moment [295]. Hip:knee extension ratio has also been shown to be greater in the wide stance position with a compensatory increase (decrease) in hip (knee) contribution apparent through altering stance [308]. In contrast, when comparing the back squat with leg press exercise, research has shown knee forces to be higher in squatting although neither foot height (leg press) nor foot angle (squat) affect magnitudes of forces [282]. However, the technique used in the squat critically impacts the effects of stance alteration. Swinton et al. [91] found that the foot position and shin angle of a traditional, powerlifting and box squat in combination with the concomitant effect on the centre of mass displacement, created differences in the joint moments across all exercises. The traditional squat showed the greatest peak moments at the ankle, while hip moment was greatest in the powerlifting variation and knee in the box squat. Such results highlight the importance of moment arms in accounting for joint-specific changes. Knee positioning during movement is also seen to affect movement kinetics. Antererio-posterior knee position, as controlled by restricting forward knee movement, affects trunk and shank angles, hence force distribution [311], while knee medial and

anterior displacement of the knee from neutral alignment is seen to affect both sagittal and frontal plane torques of all three lower limb joints [312]. Altering toe angle also apparently affects non-sagittal plane moments at the ankle [60], while a number of studies have shown that footwear, through affecting joint angles and torso position, can alter the contributory pattern of the joints to the movement [313, 314].

Strength training ethos observes some division of opinion in the use of unilateral as compared to bilateral lifts, given that higher absolute loads can be achieved in bilateral lifts [283, 315]. However, joint-level analysis of unilateral and split stance variations present results that support their benefit with studies confirming increased joint moments, hence muscular effort, in these positions. Stuart et al. [316] compared knee joint moments in two squat variations with the lunge exercise and found a higher maximum extensor moment in the lunge than either squat condition. Van Soest et al. [317] directly compared unilateral and bilateral jumps, albeit in only unloaded, and found the single-leg execution created higher peak torques at all joints. The effects of different variables were highlighted in this study, with only mean values significantly higher in hip and ankle. Joint powers at the knee were contrastingly lower in the unilateral variation, compared with the bilateral movements, while ankle was higher. A number of aspects of technique execution and limb position can affect moment distribution in unilateral exercises. Direction of the step taken during the lunging exercise affects joint loading, with a forward direction showing higher demand on hip as compared to a lateral direction, which increases demand on ankle and knee extensors [318]. A further study confirmed the dominance of the hip extensors in the anterior lunge, while demonstrating that loading effects the three lower limb joints in a distinct manner [285]. The position of both the dominant and supporting legs in these exercises is also seen to alter contribution from the hip and ankle. Moment distribution in split squats is dependent on step length and front knee position, while load alternatively affects the hip and ankle but not knee [319]. Varying the position of the support leg in single leg squats affects similar outcomes; back (as compared to forward or mid) position increases knee extensor and decreases hip extensor moment [62]. Finally, joint contribution can also be altered by conducting the single leg squat on varying angles of a decline board, a steeper angle increasing the knee moment and decreasing moments at the hip and ankle [320].

2.6.2 Ballistic and Explosive Lifts

Exercise selection for developing power as opposed to strength per se, generally utilise ballistic or explosive movements [48, 283]. Jumping has been shown to be a hip dependent movement pattern and is seen to have a similar contribution of hip extension as the back squat when compared at the same load [293, 321]. In keeping with the increased movement speed and hence joint angular velocities of the explosive exercise, joint powers are higher in the squat jump confirming their benefit for athletic development [321]. Further, jump shrugs produce far higher joint angular velocities than hang cleans, results being related these to the more ballistic nature of the lift [300]. While a number of standard lifts have observed a higher contribution of the ankle moment than knee, in developing power the knee is reported as higher than the ankle in both jumps and Olympic lifts [57, 58]. Kipp et al. [57] further noted the differences in the relationships of joint torque and joint power at each joint to loading could be explained by the contribution of joint angular velocity in the power term. In this study, joint angular velocity had no apparent relationship to load at any joint and, in such a case, increased contribution of the knee could be more related to the moment arm of the ground reaction force about the knee than the external load.

Olympic weightlifting movements are known to demand high levels of external power delivery and have, therefore, been recommended as the superior exercises for athlete development [283]. However, the necessity for technical proficiency in these lifts has led to their application being questioned in lieu of utilising loaded jumps to the same end [301]. The relationship of Olympic lifts to loaded jumping is therefore of interest. Canavan et al. [322] compared snatch and vertical jump performance and determined that the kinetics of performance were highly similar supporting the beneficial inclusion of either type of exercise in athlete programming. However, joint level analysis is more revealing. Cleather et al. [235] compared jumping and jerking concluding they had unique sagittal plane strategies. The push jerk was observed to be more knee dominant, where vertical jump observed a more equal contribution of hip and knee. The relationship between jumping and jerking did, however, appear to be

load dependent. Cushion et al. [323] concluded that there was a partial correspondence in push jerk, unloaded countermovement jump and loaded jump squat with correlations between joint kinetic profiles significant at certain loads. The relationship of each exercise with load was demonstrably exercise depended, with push jerk displaying a much greater increase in joint moments with load than jumping.

A number of studies have described joint kinetics in Olympic lifts or derivatives. Relative importance of the hip has been reported in snatch variations. Hip joint power is seen to be a critical determinant of whole body power in the power snatch [324]. In the full snatch, knee joint moments are small, representing a third of those at the hip, and are not correlated with load lifted [303]. In fact, higher standard of lifters limit knee joint moment by control of knee position with respect to the ground reaction force vector [303]. Examining joint torque during the pull phase of the clean exercise, Kipp et al. [57] determined that hip torque dominated effort contribution, followed by the ankle. Increasing the load lifted affected an increased torque at both joints to the highest load, where knee contribution peaked at a 75% 1RM load suggesting little advantage of training heavier for this particular joint. Separately comparing the power clean and jump shrug, the research group found joint, load and lift dependent behaviour [325]. While positive work increased with load in both lifts, unique relative loads were found to maximise work for each joint, with distribution of work across the joints further impacted distinctly by load. Although Hayashi et al. [97] demonstrated that peak joint torques increased at all joints with load during the pull phase of the clean, discriminatory effects on joint *power* was less clear due to an inconsistent interaction of load with joint angular velocity. Since joint angular velocity is seen to be higher when conducting explosive exercises in a ballistic manner, the weightlifting derivatives without the catch phase may represent the ideal combination [300]. Such lifts have been shown to produce superior external power, and indeed Kipp et al. [325] determined that the jump shrug had greater load-averaged hip and knee positive work, and peak knee and ankle joint power. Further comparisons of exercises would be beneficial in this regard.

2.7 Summary and Conclusions

Inverse dynamics provides an effective means of assessing the functional performance of athletes, allowing prediction of the distribution of muscular effort across the joints during movement. Studies reporting joint kinetics of cycling have demonstrated some distinctions in trials of all-out sprinting as compared to steady-state conditions. Results also show a clear dependency on pedalling rate and riding position, in particular whether conducting the trial in a seated or standing position. Sprint cycling requires production of maximum muscular power output in both seated and standing positions, while competitive cadences extend beyond 150rpm. At such high limb speeds the ability to effectively direct force around the pedal stroke is challenged and studies have shown performance variability increases at high cadence. Although expert performers are likely to express more skilled coordinative patterns, no study has conducted a full joint-specific analysis in highly-trained sprint cyclists. In such cases, current reports of joint kinetics of sprint cycling may be inconsistent and assessment of sprinting in an appropriate population would be beneficial.

Two dimensional processes in inverse solutions are seen to be susceptible to inaccuracies when compared against gold-standard 3D systems. Studies of movement involving significant frontal and transverse plane motion or involving at least significant non-sagittal motion of the pelvis, have concluded that the use of 3D motion capture techniques is essential. Although cycling is predominantly a sagittal plane movement, non-sagittal plane motion of the pelvis is apparent, and, indeed cross-hip joint-power contributions appear discriminatory in maximal conditions, particularly in standing. The use of triaxial force data has been show as critical to accurate joint kinetic assessment in cycling. However, it is unknown to what extent motion out of the sagittal plane at either the hip, knee or ankle would additionally contribute to sagittal plane joint kinetics. Acquisition of force data and subsequent synchronisation with kinematic data can be problematic in cycling and 3D processes require bespoke solutions. System design facilitating 3D analysis, and subsequent evaluation of 3D as compared to 2D outcomes, would inform biomechanical testing practices for high performance cycling.

Although resistance training is a constituent part of the training week of sprint cyclists, the relationship of the applied exercises to the goal movement is poorly understood. Lower limb joint kinetics in weightlifting are demonstrably affected by the skill level of the performer, position and magnitude of the load, relationship of the body segments to the load position, as well as technical aspects of body position and lift execution. For an accurate comparison with cycling, assessment of joint kinetics in weightlifting must be done with applicable exercises and loads and using the same athletes. Conducting a biomechanical analysis and comparison of both on- and off- bike training modalities in expert performers would provide an effective means of better understanding appropriate exercise selection, hence supporting improvements in training to the highest levels of performance in sprint cycling.

Chapters 3, 4 and 5 are not included in this version of the thesis

6 Summary, Conclusions and Recommendations

6.1 Thesis Summary and Implications

Biomechanical understanding of sports performance underpins training practices for athlete development. Indeed, the principles of dynamic correspondence are dependent on a thorough assessment of the movement kinematics and kinetics [5]. While being able to monitor small but meaningful changes in movement is important in athlete development and performance [2], several sources of error are common within various applied biomechanics methodologies [68]. There has been little research examining the use of motion-capture technologies to assess sprint cycling performance, and limited evidence of the reliability of the testing practices and results. Developing a means of conducting cycling kinematic and kinetic assessments that are reproducible in repeat testing would allow biomechanical data to become an integral part of performance monitoring processes for high performance sprint cyclists. Cycling biomechanics are critically influenced by cadence and riding position [364], yet detailed assessment of the effect of these factors on cycling kinematics and kinetics, particularly in sprint cycling performance, is lacking. Testing reliability may, additionally, be affected by both the cadence and riding position utilised in the testing protocol. Therefore, the aims of this thesis were to: i) determine the accuracy and reliability of assessing the joint kinetics in highly-trained sprint cyclists, ii) investigate the impact of changing cadence and riding position on joint kinetics and testing reliability and iii) determine the similarities and differences in the athletes joint kinetics during sprint cycling with those during execution of resistance training exercises used to develop athletes' strength characteristics. To achieve these aims a custom system was built and utilised in the three studies comprising this project.

Key findings of this thesis are that: i) higher accuracy of results are obtained using 3D, compared with 2D joint kinetic analysis; ii) high reliability in crank power and total muscle power were observed using 3D analysis, particularly at a rider's optimal cadence for power production; iii) reliability of joint-

specific powers were lower than observed in crank and total muscle power, with reliability values impacted by cadence, and whether the power is assessed over the extension or flexion phase of joint action; iv) mean and maximal crank and total muscle power demonstrate a quadratic relationship with cadence, whereas the relationships observed at contributing joints are quadratic or linear (both positive or negative); optimal cadence for maximising power production is, therefore, unique to each joint; v) although the standing position allows greater power to be produced at the crank, riding position distinctly effects the cadence relationships observed at the joints such that contribution of joint-specific power to crank power is position dependent; vi) joint kinetics of both seated and standing sprint cycling performance has similarities with those observed through specific phases of the clean exercise, predominantly at the hip; vii) joint kinetics of other resistance exercises examined were uniquely affected by the particular mechanics of the lift, with a relative hip- or knee- dominance observed.

Current biomechanics practices for assessing joint kinetics consider 3D methodologies as the gold standard [177]. Indeed, while process errors are well documented in both 2D and 3D methodologies, comparisons in non-cycling movements suggest 3D analysis is more accurate [72, 75]. However, 2D systems offer increased flexibility and portability, which would be of particular benefit for regular testing in the training and testing schedules of elite athletes [67]. To date, most studies in cycling have been restricted to 2D systems. This is due to the predominance of movement in the sagittal-plane, and, more critically, the challenges of integrating pedal force data into the data acquisition and modelling packages of 3D systems [71]. As such, the first important outcome within this thesis was the design, construction and refinement of a data acquisition and modelling system for high-level cycling analysis. This system utilised custom force pedals that are both robust enough to withstand the forces delivered by high level sprint cyclists but sensitive enough to detect small changes in force patterns. The data acquisition and modelling aspects of the system were then designed to have flexibility that permits use with either 2D or 3D techniques. As part of that design process a unique and custom-built solution for force sensor calibration was also achieved allowing calibration to be conducted in situ, avoiding any need to dismantle the pedal system. This system is appropriate for widespread use in cycling and has a

far wider scope than the research contained within this thesis. Indeed, the data capture system and calibration devices developed in this thesis are now being utilised in the assessment of cycling biomechanics of Australia's most talented cyclists. It is noted that non-motion capture technologies, such as the Martin laboratory's integrated spatial linkage system [329], have been successfully used in cycling research and may provide more flexible solutions in field-based settings. However, the integrated spatial linkage system still only utilises 2D coordinate system, and indeed references movement to the positioning to the hip joint centre position, which is susceptible to its own errors.

A number of limitations of biomechanical processes and current available technologies were overcome by the methodologies utilised within this thesis. Although technology to assess kinetics at the crank is widely available, Bini et al. [207] demonstrated that crank-based analysis overestimated power and underestimated torque when compared with instrumented pedals. Data resolution of crank systems are, at best, 250Hz where the pedal force system utilised within this thesis captured data at 1000Hz. Furthermore, assessing crank power, rather than pedal and joint-specific power, provides only limited information on performance. In sprint cycling, the ability to produce maximal muscular power is discriminatory and, therefore it is important to assess how such power is being developed and delivered [11]. Understanding of muscle mechanics indicates it is the coordinative pattern rather than individual muscle capabilities that is critical to power production [14]. Yet, limitations are evident in analysis of coordination patterns through EMG. Patterns are highly variable and the relationship of observed activity to performance is not exact [20]. In fact, not all contributing muscles are recruited maximally during sprint cycling performance [23]. The use of joint kinetic analysis in this thesis provided a non-invasive means of evaluating the net muscle activity affecting movement at each joint and decomposing muscular from non-muscular contributions to external power [36]. Results demonstrated clear benefit of the joint kinetics testing process, providing confirmation that performance conditions such as cadence and riding position impact the net muscular activity each joint; information unable to be ascertained at crank level.

Outcomes of the first study provide the basis for selecting either 2D or 3D processes for assessing cycling performance. Results indicate significant differences in hip, knee and hip transfer powers using 2D, compared with 3D methods. It may, therefore, be ideal to utilise 3D analysis techniques. However, given that most athlete testing is assessing the impact of acute or chronic intervention, quantifying the absolute values of data may not be critical, instead identifying relative changes may be more important. In such cases, the 2D system may still provide an acceptable testing tool so long as the practitioner appreciates the compromises in absolute profiles. The first hypothesis of this thesis was, therefore substantiated, with greater accuracy in determining joint kinetics using a 3D, as compared to 2D system. However, the second hypothesis stated that errors would be greatest at the hip. In fact, the knee and hip transfer terms were most affected. These findings were largely due to movement out of the frontal plane and, at the knee, errors in the segment lengths and moment arms through the crank cycle. The hypothesis of significant differences at the hip was only partially supported. Magnitudes of hip powers were unaffected, while phase shifts in the position of peak power around the crank cycle were apparent. These phase shifts were largely due to differences in joint angular velocities, and hence related to established range of motion differences as well as joint centre position. Alkaejer et al. [76] demonstrated that hip (but not knee) errors could be reduced by utilising 3D positional data reduced to 2D, which although confirming the relationship to hip joint centre location, does not assist improving a stand-alone 2D system. However, it is plausible that the extent of errors in 2D as compared to 3D assessment could be quantified and used to apply corrective adjustments to 2D results. This would be a fruitful area of exploration for future research. In broader terms, results also confirm that a frontal-plane component does make a significant contribution in sprint cycling. Results of Study One showed no interaction of motion-capture condition with riding position and, as such, differences between 2D and 3D were equivalent in and out the saddle. Comparisons of the two positions did, however, establish some key distinctions of joint-kinetics when sprinting in the standing position. The comparison between standing and seated sprinting was further extended in Study Two. Only a single study by Davidson et al., [73] printed as abstract only, has examined sprinting out the saddle. This thesis, therefore, provides novel data demonstrating that out of the saddle sprinting uniquely impacts the contributions at each joint.

While hip, hip transfer, muscle and crank powers across the full revolution were greater in standing, compared with seated sprinting, decomposing the movement into extension and flexion phases indicated that knee flexion power decreased, whereas hip flexion increased, during standing. Hip power therefore contributed to greater total muscle and crank power in the standing compared with the seated condition. Study Two showed that this compensatory shift of joint emphasis is further exaggerated at higher cadences. While knee range of motion increased in standing, no such change was observed at the hip. Yet the position of maxima and minima of joint excursion altered concomitant to the more forward position of the body in standing, affecting a shift in the position of peak power delivery at the crank. Joint powers also consistently reflected this phase advancement. Collectively, an extended duty cycle through extension would augment power delivery. The more detailed analysis in this thesis demonstrated that, in fact, conclusions drawn in the Davidson et al. abstract [73] only provide part explanation for positional differences. Understanding these differences can assist the training process for athletes. In terms of movement mechanics, sprinting effectively consists of two distinct movement patterns that uniquely load the contributing muscles. Not only does this impact the understanding of supply versus demand in the aerodynamic trade-offs in each position, but the strength characteristics required by standing and seated sprinting require different training stimuli or emphasis in gym-based interventions. The importance of understanding the unique relationships between sprinting and strength exercises was extensively examined within Study Three, which highlighted that the torque-angular velocity relationships in each riding position affects the relevance of particular conditioning exercises.

Study Two further determined that, while the total muscle power term reflects the quadratic power-cadence relationships observed at the crank, the joints show distinct profiles. The third hypothesis asserted that cadence conditions would affect joint power distribution. This hypothesis was, therefore, confirmed. This highlights that disparate task demands affect the optimal coordinative strategy to maximise global muscle power. Implications of this finding may be of critical importance in testing practices, since test cadence is commonly selected to maximise external power. In fact, a novel finding

of this thesis is that optimal cadence at the crank will represent a compromise in joint-specific power production. Assessment of maximal muscle function may, then require a joint-specific approach to testing and analysis. The equivalent hypothesis (hypothesis four) on the effects of riding position was confirmed with the exception of knee power. Results of Study Three additionally demonstrated the knee torque-angular velocity characteristics are equivalent in each riding position. This provided explanation for lack of main effect at the knee for riding position in Studies One and Two, though a significant interaction of cadence and seating position did distinguish that knee power continued to increase in the high cadence condition in standing. Hip power alternatively continued to increase at high cadence while seated, and again these results emphasise that unique functional characteristics are required to optimise performance in different riding conditions. In contrast, the ankle contributed greater power at low compared to higher cadences. Study Three added to the unique findings presented in this thesis with respect to the ankle. High torque conditions of low cadence pedalling require an 'ankling' motion that assists in force production, while an ankle fixated in plantar flexion improves limb control and force transference from the proximal joints at high cadences. These observations provide explanation for the ankle power-cadence relationship observed in Study Two. Collectively, these results characterise that the joints play distinct roles in different performance conditions. It is noted that joint-level analysis has its limitations, and that, indeed, the relationship between joint- and muscle- level analysis is far from trivial. However, the current research outcomes further our understanding of the biomechanics of skilled sprint cycling performance and can significantly contribute to improving the training strategies for sprint cycling. For example, previous studies have suggested that increased musculo-tendon stiffness at the ankle discriminates higher performances in sprint cycling [112], and the current study provides justification for training to that end. The distinct torque-angular velocity relationship of the ankle requires consideration when applying commonly used 'triple-extension' patterns in the gym, since the ankle does not perform in a congruent manner to the other two joints.

A number of findings of this thesis with respect to seated sprinting were consistent with previous study results [40, 205, 266] and confirm that joint kinetics of sprint performance show some disparity from

those in submaximal conditions [38, 39]. There appears to be an increased contribution of upper body and lower limb flexion in sprint cycling performance compared to submaximal riding. Although it has been suggested that these strategies are inefficient in cycling [159], unlike the movement efficiency demanded by steady-state riding, the goal task in sprinting is to achieve maximum muscular power output. Prior studies have, therefore, suggested that in this context, joint flexion and upper body movement would actually be beneficial [39]. Since existing studies have utilised non-sprint trained populations it had previously not been confirmed whether the joint kinetic profiles observed would be upheld in skilled performers. The data from this thesis indicates substantial similarities, although some further nuances in performance were observed – notably the mechanics of the ankle and high magnitudes of ankle power. However, without establishing the consistency of joint kinetic results, the relationship of small observed changes in net moments and powers cannot be irrefutably related to sprint-specific mechanics. It was hypothesised in this thesis that a high degree of reliability would be observed of results in repeat testing. This was confirmed for data through the primary power phase of performance. However, critically, flexion and upper body powers showed least within-days reliability; coefficients of variation of flexor and upper body power were substantially higher than extension and lower limb powers. Given that skilled performers are known to display greater consistency of movement patterns [83, 333], poor reliability in this cohort is an interesting outcome. The variability of joint power out with the primary power production phase may then either be related to being a less refined part of the movement pattern, or equally may be evidence that the ancillary part of the movement fulfils a dynamic optimisation role, varying in response to movement conditions in order to improve power delivery to the crank.

High within-days reliability values of crank and muscle power confirm that total muscle power output is relatively consistent within the cohort examined in this thesis. Notably, the muscle term is derived from summated joint terms, and yet shows greater reliability than observed at the individual joints. In such a case, although external power output is maintained, some inherent variability in the underlying control strategy is, therefore, likely revealed through this analysis. With ankle power showing greater

variability than the hip and knee (as evidenced by higher CV), this may provide further evidence of the joint's role in 'tuning' overall limb and muscle function, movement dynamics altered to help optimise power production around the crank cycle. The impact of cadence and riding position on the reliability of power values suggest that variance is also greater out with ideal performance conditions for maximising muscle power. Optimal cadence represents the pedal rate at which the athlete can generate most external power output, in other words where a global maximum of muscle power output can be achieved. The finding of greatest reliability at this cadence is a notable finding of this study, since it confirms that the overall motor control pattern is most consistent at this pedal rate. Flexibility of the motor domain may be more important where performance conditions are more challenging i.e. where high muscle force or contraction velocities are required. Though again this could be reflective of less refined movement control at the extremes of performance. It would be interesting to examine changes in reliability following an extended period of training at such cadences or between populations of different history.

Of course systematic error would substantially affect testing reliability. However, this would be consistent across all trial conditions. Low between-weeks variances across all dependent variables support the use of these testing practices for elite performance since accurate inference can be made about changes observed in repeat testing. A unique aspect of the statistical approach used in Study Two was the ability to partition the variance, allowing contributing factors in distinct performance conditions to be established. The impact of trial conditions on reliability results are a key finding of this thesis and provides important information for the application of joint kinetic testing. This is a consideration that likely has relevance for joint kinetic testing in other sports. Controlling for between-athletes variances in the modelling of all three studies was also a significant benefit of the methodology applied in this thesis. The between-athlete variance accounted for the greater part of variance in Study Two and, indeed, allowing the intercept of models to vary with athlete allowed identification of underlying patterns that may have otherwise been clouded by individual differences. Individual response is a challenging issue for sports science research where group means are required to confirm or refute

hypothesis. And yet individual differences are cornerstone of specificity in training prescription. Including the athlete as a random term in model fit further revealed higher between-athlete variation in resistance exercises as compared to cycling. This suggested that some individualisation in the ability to express strength during lifting, where greater consistency is observed in their skilled movement pattern.

The application of the methodologies utilised in this thesis allow coaches and sports scientists to assess how acute changes in functional conditions may impact future performance. For example, small changes in bike set-up may appear to have little effect on power measured at the crank in acute testing, where altered joint-specific power profile might provide insight into potential impact of the change. Given that Martin and Brown [40] have established that the net muscular contribution at each joint is uniquely affected by fatigue, changing the coordinative strategy could also impact performance by creating a different muscular fatigue profile over the course of the sprint. The testing process could, then, benefit determination of where an intervention to change or improve the contribution of supporting muscle groups is needed, with repeat testing informing the efficacy of the intervention itself. Results of Study Three provide particular benefit in this regard, allowing coaches to make informed decisions around exercise selection. Disparity between the cycling and lifting performance of sprint cyclists is commonly reported in training squads and, while FV profiles are generally used to examine the functional performances of athletes, those derived from on and off the bike techniques are incomparable [16, 140, 365]. Examination of the joint torque-angular velocity profiles in this thesis presented a novel means of relating the performances. This provides the means for coaches to assess relative similarities of movements and supports more informed decisions on exercise selection to target improvement in the strength capabilities of specific muscular action. Since ‘strength’ represents force capabilities under specific movement conditions, assessment of joint kinetics also provides the means to determine how well performance in any exercise is meaningful to the context of their bike performance.

The finding of some similarities between the torque-angular velocity profiles of the clean and seated and standing sprint, upholds hypothesis seven of this thesis. The association of distinct derivative phases of the clean to each riding positions highlighted the impact of movement velocity on the relationship between the exercise modalities. Ballistic exercises have previously been recommended as more beneficial to improving power characteristics due to higher movement velocities and studies have suggested that weightlifting derivatives that do not require the catch phase would be ideal [300]. Jump squats are a commonly used exercise for training and testing power performance and, indeed the lower technical demands of the jump may permit the development of higher joint torque through this exercise. Additionally, previous research suggests jump shrugs may provide a better stimulus for power development [325] and assessment of their relationship to the cycling movement would be beneficial. Future research opportunities are, therefore, apparent in this area. The variation in joint dominance of the strength exercises tested also confirmed hypothesis eight, since position of the load and kinematic profile of the lift effected a unique joint kinetic profile in each exercise. Similarly hypothesis nine was also confirmed, since hip and knee torque in each exercise was relatively greater or less than in seated or standing cycling. This suggested relevance of utilising certain exercises in a joint-targeted manner, exercise selection being focussed on a hip or knee emphasis. At the hip, maximum torque in the single leg rack pull was distinguished as being substantially higher than produced on the bike, where at the knee, the box step-up showed greater maximum torque. Greater torque demands in exercises where primary drive happens through a single leading leg, supports the efficacy of unilateral lifts for strength development.

The findings of this thesis can aid the specificity and individualisation of training prescription for sprint cyclists. Results do, however, draw into question the overall ethos of gym based exercise to improve bike related strength. Joint-specific maximum torque was higher on the bike than the bilateral (though notably not unilateral) resistance exercises. Results of this thesis suggest that on-bike performance conditions can be effectively manipulated to exaggerate loading emphasis towards joint-specific strength improvement. Given the increased variance of torque production during lifts as compared to

bike performance, athletes would also be more able to express their maximum force characteristics in this more-familiar movement pattern. Further, triple extension lifting patterns are most commonly applied for training cyclists [54], yet the unique characteristics of the ankle on the bike would suggest a somewhat distinct lower limb coordinative strategy. This would be further distinguished when considering the contributions of the biarticulate muscles in the integrated movement. Cadence, and hence FV conditions, can be easily manipulated on a bike and exaggerating the force demands at the pedals is easy to achieve. In such a case, the bike itself could be used to provide an effective overload as stimulus for strength gains.

Joint kinetic analysis provides an effective means of estimating net muscle action during movement, and this thesis suggests it may be a useful tool to inform training practices for elite sprint cyclists. It must be noted, however, that inverse dynamics fails to account for the activity of biarticulate muscles. Previous studies have suggested some unique complexities in the cycling action in this regard [93, 168]. Indeed calculations of average joint moments in Study Three revealed phases of conflicting moment and angular velocity direction suggesting the critical influence of biarticulate activity through particular regions of the pedal stroke and particular phases of movement in the resistance exercises. This thesis examined only the phases of extension where positive power was produced, permitting appropriate comparison between the different exercise modalities. However, the full extent of biarticulate muscle action across the pedal stroke would not be apparent using inverse dynamics. Approaches such as biarticular load compensation [256] or static optimisation [366], may offer alternative solutions. Additionally while the current testing system provided a novel solution to testing cycling biomechanics, the equipment only permitted assessment in a stationary cycling position. Prior study has, in fact, demonstrated compatibility of assessing power-cadence relationships on the SRM cycle ergometer used in this study to field-performance in elite track sprint cyclists [124]. Hence, while further study confirming this relationship is recommended, the current methodology can certainly be supported as providing a meaningful context for the assessment of athlete characteristics. By answering some key methodological questions and establishing fundamental joint-specific performance relationships, the

custom system and results of this thesis provide an increased understanding of joint kinetics in highly-trained sprint cyclists. The information presented helps improve the testing and training processes of the athletes and indeed, can be used to improve prescriptive practices. A number of novel outcomes were presented in this thesis benefiting advancement of the sport.

6.2 Directions for Future Research

The custom testing equipment provided the foundations for this research. However, a number of aspects of the design solution were challenging and present opportunity for future exploration. Unlike in-floor force platforms, the exact centre of pressure of the force vector could not be established and warrants a more accurate solution. A sensitivity analysis would also benefit understanding the impact of imprecise centre of pressure location on outcome measures. Comparisons of the 2D and 3D methodologies demonstrated the impact of distinct algorithms being applied in the background of different analysis packages with, for example, filters designed to the same specifications producing marginally different results. Improvements in the modelling process could also be affected through individualisation of the inertial parameters. Although possible using current medical scanning devices, such processes are highly expensive and an alternative solution is warranted. The 2D system is cheaper and more flexible than 3D systems. However, the inherent nature of manual synchronisation and manual or semi-automatic trajectory tracking creates a labour-intensive process. The accuracy of the 2D kinematics could also be improved through increased data resolution. Again, costs are currently a deterrent, since higher frame rates increase camera costs substantially. The pedal system itself also offers future research potential in both the sensor technology and opportunity for wireless transmission of data.

Recommendations from Study One suggested that the ability to quantify the inaccuracies of 2D processes may permit correction terms being applied to 2D outcomes. The reliability of the 2D testing protocol should also be established to support regular testing. Differences in reliability results through

for example, skill level, or degree of fatigue could also be established. Since the reliability study revealed that trial conditions impacted results, it would be similarly beneficial to assess the consistency of performance in resistance exercises and, indeed, establish differences in using 2D processes for assessing lifts. Testing of additional exercises, particularly jump squats and jump shrugs, would also be highly beneficial. Additionally, although execution of the clean exercise demands maximal velocity intent from the hang phase onwards, other lifts executed at non maximal loads may be executed with less than maximal intent. A comparison of joint-specific torque-angular velocity characteristics of a full spectrum of lifts across loads at maximal intent would augment current observations. With this information, it may then be that joint kinetics could be predicted from bar velocities. Given that gym testing often utilises linear position transducers attached to the bar in regular training sessions, this would provide an easy means of determining relative joint loading in any exercise condition.

The methodologies of this thesis provide the means to assess the effects of acute and chronic interventions on joint powers. Several areas of research would be of immediate benefit to the sport in this regard, including bike-position and the effects of fatigue. The outcomes of this thesis also provide the basis for a more extensive model of sprint cycling performance. Having built a profile of athlete characteristics across riding and lifting conditions, and with individualised inertial parameters in the model, the data could be used to provide predictive outcomes. Including the results of forward dynamics solutions and as well as terms involving the power demand side of the equation would support a full supply-demand model of sprint cycling. Some existing work from other laboratories has already established some of the methodology needed to this end [367].

6.3 Conclusion

In summary, this thesis aimed to investigate the joint kinetics of highly-trained sprint cyclists and provide information supporting the use of joint kinetic assessment as part of regular testing and training practices. A custom-designed force pedal system was established and evaluation was made of its use as

part of 2D as compared to 3D analysis processes. Having established the greater accuracy of the 3D system, the reliability of the 3D process was then analysed while simultaneously examining the impact of cadence and seating condition on joint kinetics. Finally, comparison was made of joint kinetics of sprint cycling with that of resistance exercises commonly used to improve the strength characteristics of athletes in order to help improve the specificity of programming practices.

This thesis concludes the following:

1) Differences are apparent in sagittal-plane joint kinetics assessed in 2D as compared to 3D, with magnitude of knee power lower and hip transfer power higher. Hip range of motion is significantly lower when assessed in 2D, while position of peak hip power is shifted to later in the pedal stroke. Such distinctions must be considered when comparing results of distinct analysis processes.
2) 3D processes are recommended where absolute values of joint kinetic profiles are important. The flexibility of 2D systems offer advantages in the training environment and 2D analysis still provide a valid assessment tool where relative changes in profile are of interest or where assessment is only concerned with the absolute values of unaffected measures.
3) A quadratic power-cadence relationship is observed at the crank and for total muscle power, with absolute values greater in standing as compared to seated cycling. However, contribution of each joint to total muscle power is uniquely related to both cadence and riding position. The hip displays a linear maximum power-cadence relationship in seated but quadratic in standing, while the knee displays the reverse. Ankle and hip transfer powers linearly declines with cadence both seated and standing. Underlying strategies and effort distribution cannot, therefore be directly inferred from power assessed at the crank, with optimal cadence for crank power representing a compromise in joint-specific power production.
4) Reliability of crank and muscle power is high, with coefficient of variation particularly low at the rider’s optimal cadence. While reliability of contributing joint powers is slightly lower, control strategies appear to maintain consistent external power despite variability in the

underlying distribution of muscular effort. Some variation in joint powers may be important in achieving dynamic optimisation of the task

5) While reliability at low to mid cadences is good in lower limb extension powers, high cadence pedalling (particularly in standing), lower limb flexion power and hip transfer power values are less consistent. Increased variation in these measures may be related both to the inherent challenge of the movement conditions and to the need for greater flexibility of the control domain in extreme performance conditions. Consideration must be given to unique performance variability associated with different joints and trial conditions when examining testing results.
6) Hip extensor torque is predominant in sprint cycling, and higher relative hip torques in the clean, deadlift, Romanian deadlift and single rack pull present these as effective choices for strength improvement at the hip for sprint cyclists.
7) The hip torque-angular velocity relationship of sprint performance in the saddle is equivalent to that of the 2nd pull in the clean, while that of sprinting out the saddle is equivalent to the hang pull, with differences largely related to the movement speed through each phase of the clean. The clean therefore provides a somewhat similar stimulus at the hip as sprint cycling.
8) Knee torque is consistently much greater than in any phase of the clean and technical complexities at the knee through the clean confound its relationship to sprint knee joint kinetics. In contrast knee torque in the step-up exceeds that of sprint performance and may provide an effective strength training stimulus for knee musculature.
9) Ankle torque-angular velocity relationships are distinct from all other joint-exercise conditions in showing a positive linear relationship. The ankle is seen to function in a distinct manner, benefitting high force production through ‘ankling’ action at low cadence, but becoming increasingly fixated to allow for efficient force transference and maintaining limb trajectory at high cadence. Optimisation of force production at the ankle for sprint cycling performance must, therefore, be considered distinctly from the other contributing joints.

Overall findings of this thesis suggest joint kinetic analysis of highly-trained sprint cyclists can provide information about underlying performance strategies that cannot be assessed through conventional testing or through assessment solely at the crank. High reliability of the testing process for lower limb joint extension supports its incorporation into regular testing practices for athletes. Consistent power at the crank, in spite of variation in joint kinetics, suggests that the motor domain dynamically responds to the task conditions and that some variability of muscle contribution is inherent in skilled riders. Future research examining the extent to which this can be exploited to increase the global maximum power produced would be beneficial. Reliability statistics presented in this thesis support monitoring processes incorporating joint kinetic assessment, since the effects of both acute and chronic interventions can be assessed in greater detail than is provided by crank power alone. In this regard the reliability statistics presented provide critical information for accurate interpretation of results. The comparison of sprint cycling and resistance exercise performance provided evidence that can help inform exercise selection for strength development. The ability to monitor changes and target training intervention at joint level present highly beneficial outcomes that provide immediate practical benefit for training athletes to the highest levels of competition in sprint cycling.

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8 Appendices

Appendix A – Contribution of Work

STUDY 1

System Design and Construction:

Lynne A. Munro (69%), Geoff Wanders (25%), Tom Kepple (5%), Joan Charmant (1%)

Calibration System Design:

Assoc. Prof. Chris R. Abbiss (60%), Lynne A. Munro (30%), Nadija Vrdoljak (10%)

Calibration Process:

Lynne A. Munro (80%), Assoc. Prof. Chris R. Abbiss (20%)

Data Collection:

Lynne A. Munro (95%), Nicholas Flyger (5%)

Data Analysis:

Lynne A. Munro (85%), Nicholas Flyger (15%)

Statistical Analysis:

Lynne A. Munro (70%), Nicholas Flyger (30%)

STUDY 2

System Design and Construction:

Lynne A. Munro (69%), Geoff Wanders (25%), Tom Kepple (5%), Joan Charmant (1%)

Calibration System Design:

Assoc. Prof. Chris R. Abbiss (60%), Lynne A. Munro (30%), Nadija Vrdoljak (10%)

Calibration Process:

Lynne A. Munro (80%), Assoc. Prof. Chris R. Abbiss (20%)

Data Collection:

Lynne A. Munro (95%), Nicholas Flyger (5%)

Data Analysis:

Lynne A. Munro (80%), Nicholas Flyger (20%)

Statistical Analysis:

Lynne A. Munro (60%), Nicholas Flyger (23%), Dr. Andrew D. Govus (17%)

STUDY 3

System Design and Construction:

Lynne A. Munro (69%), Geoff Wanders (25%), Tom Kepple (5%), Joan Charmant (1%)

Calibration System Design:

Assoc. Prof. Chris R. Abbiss (60%), Lynne A. Munro (30%), Nadija Vrdoljak (10%)

Calibration Process:

Lynne A. Munro (80%), Assoc. Prof. Chris R. Abbiss (20%)

Data Collection:

Lynne A. Munro (90%), Nicholas Flyger (5%), Scott Baker (5%)

Data Analysis:

Lynne A. Munro (80%), Nicholas Flyger (15%), Scott Baker (5%)

Statistical Analysis:

Lynne A. Munro (80%), Nicholas Flyger (20%)