Effects of saddle position on pedalling technique and methods to assess pedalling kinetics and kinematics of cyclists and triathletes

A thesis submitted to Auckland University of Technology
in fulfilment of the requirements for the degree of

Doctor of Philosophy

November 2011

Auckland University of Technology
Faculty of Health and Environmental Sciences

by

Rodrigo Rico Bini BPhysEdu & MHSc

Primary supervisor: Professor Patria Hume
Secondary supervisor: Dr. James Croft
Tertiary supervisor: Associate Professor Andrew Kilding

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

LIST OF EQUATIONS

ATTESTATION OF AUTHORSHIP

I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person (except where explicitly defined in the acknowledgements), nor material which to a substantial extent has been submitted for the award of any other degree or diploma of a university or other institution of higher learning.

Rodrigo Rico Bini

Rodrigo Rico Bini

August 2011

CANDIDATE CONTRIBUTIONS TO CO-AUTHORED WORKS

SPORTS PERFORMANCE
RESEARCH INSTITUTE NEW ZEALAND
AN INSTITUTE OF AUT UNIVERSITY

Mail address: AUT University. Private Bag 92009, Auckland 1020.
Courier address: Sport and Fitness Centre, Akoranga Drive, Northcote, Auckland
Phone: 64 9 921 9999. Fax: 64 9 921 9960. Web: www.sprinz.ac.nz.

8th August 2011

All co-authors on the chapters/papers indicated in the following table have approved these for inclusion in Rodrigo Bini's doctoral thesis.

Chapter publication reference	**Author %**
CHAPTER 2: Bini, R.R, Hume, P.A., Croft, J.L., & Kilding, A.E. (2011). Optimizing body position on the bicycle may help prevent overuse injury in cycling. Submitted to *Journal of Science and Medicine in Sport.*	RB = 80%; PA = 10%; JC = 5%; AK = 5%.
CHAPTER 3: Bini, R.R, Hume, P.A., & Croft, J.L. (2011). Body position on the bicycle for cyclists and triathletes. Submitted to *Measurement in Physical Education and Exercise Science*.	RB = 85%; PA = 10%; JC = 5%.
CHAPTER 4: Bini, R.R, & Hume, P.A. (2011). A comparison of static and dynamic measures of lower limb joint angles in cycling: Application to bike fitting. Submitted to *Sports Biomechanics.*	RB = 90%; PA = 10%.
CHAPTER 5: Bini, R.R, Hume, P.A., Croft, J.L., & Kilding, A.E. (2011). Pedal force effectiveness in cycling: a review of constraints and training effects. Submitted to *Journal of Applied Biomechanics.*	RB = 80%; PA = 10%; JC = 5%; AK = 5%.
CHAPTER 6: Bini, R.R, & Hume, P.A. (2011). Between-day reliability of pedal forces for cyclists during an incremental cycling test to exhaustion. Submitted to *Journal of Sports Sciences*.	RB = 85%; PA = 15%.
CHAPTER 7: Bini, R.R, Hume, P.A., & Croft, J.L. (2011). Pedalling technique changes with force feedback training in competitive cyclists and triathletes. Preliminary study. Submitted to *International Journal of Sports Physiology and Performance*.	RB = 80%; PA = 10%; JC = 10%.
CHAPTER 8: Bini, R.R, Hume, P.A., & Cervieri, A. (2011). A comparison of cycling SRM crank and strain gauge instrumented pedal measures of peak torque, angle of peak torque and power output. *Procedia Engineering, 13, 56-61.*	RB = 85%; PA = 10%; JC = 5%.

CHAPTER 9: Bini, R.R, & Hume, P.A. (2011). Bilateral asymmetry assessment in cycling using SRM cranks and instrumented pedals. Submitted to *Journal of Science and Medicine in Sport.*	RB = 90%; PA = 10%.
CHAPTER 10: Bini, R.R, & Hume, P.A. (2011). Knee joint modelling for cycling. Submitted to *The Knee.*	RB = 90%; PA = 10%.
CHAPTER 11: Bini, R.R, Hume, P.A., & Croft, J.L. (2011). Effects of bicycle saddle height on knee injury risk and cycling performance. *Sports Medicine, 41*(6), 463-476.	RB = 85%; PA = 10%; JC = 5%.
CHAPTER 12: Bini, R.R, & Hume, P.A. (2011). Effects of saddle height on knee forces of recreational cyclists with and without knee pain. Submitted to *Journal of Science and Medicine in Sport.*	RB = 90%; PA = 10%.
CHAPTER 13: Bini, R.R, Hume, P.A., & Croft, J.L. (2011). Effects of saddle height on pedal force effectiveness. *Procedia Engineering, 13*, 51-55.	RB = 85%; PA = 10%; JC = 5%.
CHAPTER 14: Bini, R.R, Hume, P.A., & Kilding, A.E. (2011). Saddle height effects on pedalling kinetics and kinematics in competitive cyclists and triathletes. Submitted to *Journal of Applied Biomechanics.*	RB = 85%; PA = 10%; AK = 5%.
CHAPTER 15: Bini, R.R, Hume, P.A., Lanferdini, F.J., & Vaz, M.A. (2011). Effects of moving forward or backward on the saddle on knee joint forces during cycling. Submitted to *Physical Therapy in Sport.*	RB = 80%; PA = 10%; FL = 5%; MV = 5%.
CHAPTER 16: Bini, R.R, Hume, P.A., Lanferdini, F.J., & Vaz, M.A. (2011). Effects of cycling at preferred, most forward and most backward body positions on the saddle on pedalling technique for cyclists and triathletes. Submitted to *Sports Biomechanics.*	RB = 80%; PA = 10%; FL = 5%; MV = 5%.
APPENDIX 1: Bini, R. R., Croft, J. L., Kilding, A. E., & Hume, P. A. (2009). Improving performance and preventing injuries using cycling biomechanics: Overcoming the challenges. *Proceedings of AUT Postgraduate Symposium 2009, 1: 18.*	RB = 80%; JC = 10%; AK = 5%; PA = 5%.
APPENDIX 2: Bini, R. R., Croft, J. L., Kilding, A. E., & Hume, P. A. (2009). Efectos del entrenamiento de la tecnica de pedaleo sobre la Economia y el rendimento en el ciclismo: una revision a la literatura. *4ta Jornadas de Actualizacion en Fisiologia del Entrenamiento* 2009.	RB = 80%; JC = 10%; AK = 5%; PA = 5%.
APPENDIX 3: Bini, R. R., Hume, P. A., Kilding, A. E., & Croft, J. L. (2010). Road cyclists overuse injury and cycling body position. *Proceedings of New Zealand Sports Medicine and Science conference 2010*, Wellington, New Zealand, 18 – 20 November, p 32.	RB = 80%; JC = 10%; AK = 5%; PA = 5%.
APPENDIX 4: Bini, R. R., & Hume, P. A. (2011). Pain from overuse injury in 104 competitive and recreational road cyclists, mountain bikers and triathletes. Submitted to *New Zealand Sports Medicine and Science conference 2011.*	RB = 90%; PA = 10%.

APPENDIX 5: Bini, R. R., & Hume, P. A. (2011). Moving forward or backward on the bicycle saddle does not change pedal force effectiveness. Submitted to *New Zealand Sports and Exercise Science conference 2011.*	RB = 90%; PA = 10%.

Mr. Rodrigo Bini (RB)

Prof. Patria Hume (PH)

Dr. James Croft (JC)

Assoc. Prof. Andrew Kilding (AK)

Dr. Andre Cervieri (AC)

Mr. Fabio Lanferdini (FL)

Assoc. Prof. Marco Vaz (MV)

ACKNOWLEDGEMENTS

I would like to thank the following people for the contribution they have made, and support they have given, to my PhD thesis.

I would like to thank my primary supervisor Professor Patria Hume for her extreme effort in helping me gain this achievement. I am really thankful for all the effort she has made even when she was not fully recovered from surgeries. I would certainly not have found anyone who would encourage me so much towards perfection as she has done.

Dr. James Croft was a superb supervisor during my first year and his help was very important for me to get through the most difficult time of my journey. Thanks a lot for your support. Thanks also to Associate Professor Andy Kilding, Associate Professor Marco Vaz, Dr. Andre Cervieri, and Mr. Fabio Lanferdini who each contributed to parts of the thesis.

Special thanks to Capes-Brazil for providing me the scholarship for my PhD. I will work hard to transfer the knowledge I acquired during my time in New Zealand to people from Brazil.

Thanks to the International Society of Biomechanics for the international grant for travel provided in 2009. That funding kept me alive during the first year of my PhD. Thanks to the AUT SPRINZ Running Mechanics Clinic for the Research Assistant scholarship provided during the first year of my PhD. Thanks also to the SPRINZ Sports Kinesiology, Injury Prevention and Performance research group for providing funding for my conference expenses.

Thanks to Professor John Cronin who helped throughout the way. Thanks to all personnel from AUT and the School of Sport and Recreation which has provided me an excellent support in various areas. Thanks to Joe and Natalia who were superb friends during these years in New Zealand.

Thanks to all cyclists who participated in my research projects. Their help was very important and I hope I have giving them good feedback and helped them to ride faster without injuries.

Last but not least, to my father, mother, brother and sister, for being extremely supportive. I would like to thank for all their help and understanding. I have been away for three years and I know they see this retreat as a necessary process for my improvement in various areas of my life. Unfortunately, I could not say good bye to my grand mothers who passed away during these three years. I have been keeping them on a very special place in my heart and I know they are looking after me all the time.

DEDICATION

I wish to dedicate this thesis to my lovely wife Alice. I could certainly have not achieved the completion of my thesis without her support. I would also like to dedicate this work to all people who did not stop believing in my potential during these three years and to the ones who could not follow my work in person but certainly observed it from a higher place.

ETHICAL APPROVAL

Ethical approval for this thesis research was granted by the Auckland University of Technology Ethics Committee (AUTEC). The AUTEC references were:

- 09/178 **Bicycle configuration survey: Anthropometric and kinematic measurements**. Approved on 24th August 2009.
- 09/119 **Effects of bike configuration on the knee joint forces and muscle activity of cyclists: Implications for injury prevention and rehabilitation.** Approved on 16th July 2009.
- 09/228 **Muscle mechanics and performance of cyclists and triatheletes: Kinematics, pedal forces and ultrasound evaluation.** Approved on 3rd November 2009.
- 10/56 **Technique training effects of performance of cyclists.** Approved on 14th March 2010.

ABSTRACT

Optimising body position on the bicycle may improve performance and reduce overuse injury risk. However, effectiveness of bicycle configuration in preventing overuse injuries is unknown as methods to assess body position on the bicycle (usually via static assessment) lack accuracy and are limited to road cyclists. The aims of this thesis were to: 1) assess validity and reliability of methods for determining body position (kinematics) and pedalling kinetics in cycling; and 2) to analyse effects from changes in saddle height and horizontal position on pedalling kinetics and kinematics of cyclists and triathletes. Pedal force effectiveness, crank torque, joint angles and joint mechanical work during cycling on an ergometer were measured for recreational and competitive road cyclists and competitive triathletes, with and without lower limb pain, in five experimental studies. Analysis of static 6 o'clock crank position was not valid therefore body position on the bicycle should be measured using video from cycling motion. Instrumented pedals should be used for pedalling kinetics assessment as bilateral asymmetry was underestimated using the SRM® torque analysis system. Reliability of pedal forces from instrumented pedals computed seven days apart was high. Technique training using feedback from instrumented pedal forces at training workloads and high pedalling cadence improved power output. Using improved knee joint modelling, cyclists with and without pain had similar adaptations to changes in saddle height. Saddle height changes of 5% of the self-selected preferred saddle height did not result in substantial changes in pedalling kinetics. Moving forward or backward on the saddle affected joint mechanical work but not knee joint forces. Saddle position changed pedalling technique similarly in cyclists and triathletes. To optimise cyclists and triathletes body position on the bicycle and their technique, instrumented pedals should be used to assess pedalling kinetics and video analysis should be used to assess kinematics.

CHAPTER 1: INTRODUCTION AND RATIONALISATION

Background

Cycling performance depends on physiological adaptations to training (Coyle, 2005) and avoiding unnecessary injuries (Burke & Pruitt, 2003). For acute improvements in performance, cyclists can manage workload (by changing gears), increase/decrease pedalling cadence and change their position on the bicycle. Optimizing the interaction between the components of the bicycle and the cyclist is important for complete transfer of force from the body to the bicycle. The way the cyclist applies force to the pedal is related to power output and speed of the bicycle, which suggests that improving technique to apply greater force may improve performance.

From November 2001 to October 2002, 33 million US residents rode a bicycle an average of six days a month, for an average of >1 hour on a typical day (Wanich, Hodgkins, Columbier, Muraski, & Kennedy, 2007). This statistical report indicates that cycling has increased in recent years as a sport and recreational activity. However, increasing the number of cyclists leads to increases in injuries (Asplund & St. Pierre, 2004; Wanich, et al., 2007) which are mainly overuse injuries due to the configuration of bicycle components (Holmes, Pruitt, & Whalen, 1994; Mellion, 1991), anatomical characteristics (Burke & Pruitt, 2003) and training aspects (Schwellnus & Derman, 2005). Preventive strategies via changes in bicycle configuration are available for road cyclists (De Vey Mestdagh, 1998; Silberman, Webner, Collina, & Shiple, 2005), however guidelines for configuration of bicycle components for triathlon bicycles are not yet available in published literature.

The methods for configuration of bicycle components assume that the dimensions of the bicycle should be adaptable to the cyclist's anthropometry (De Vey Mestdagh, 1998). However, for a given configuration (e.g. saddle height) joint angles are not similar for all cyclists (Peveler, Bishop, Smith, Richardson, & Whitehorn, 2005). When references to joint angles are reported they are limited to measures of the cyclist in static poses on the bicycle, which may not replicate the angles of cycling motion.

Methods to assess cycling kinetics are varied and pedal force effectiveness is one of the variables of interest in most research (Bini, Hume, & Croft, 2011b; Dorel, Couturier, & Hug, 2009; Leirdal & Ettema, in press; Sanderson, 1991). Theoretically power output can be improved via increases in pedal force effectiveness for a given pedal force application. However, conflicting relationships between pedal force effectiveness and cycling performance are observed (Bini & Diefenthaeler, 2010; Coyle, et al., 1991; Korff, Romer, Mayhew, & Martin, 2007; Leirdal & Ettema, 2011; Mornieux & Stapelfeldt, in press) along with unknown variability of this measure across different days. Evidence has suggested that to improve pedal force effectiveness, cyclists need to recruit "less efficient" muscles (e.g. hamstrings) (Korff, et al., 2007) and change muscle co-contraction patterns (Mornieux, Gollhofer, & Stapelfeldt, 2010). Experimental designs for the assessment of training interventions to improve performance via increases in pedal force effectiveness are mostly limited to low exercise intensity (up to 80% of maximal oxygen uptake) and pedaling cadence (<80 rpm) (Broker, Gregor, & Schmidt, 1993;

Holderbaum, Guimarães, & Petersen, 2007; Mornieux & Stapelfeldt, in press; Sanderson & Cavanagh, 1990) compared to cycling racing. These studies may not have provided sufficient time and/or training intensity to elicit adaptation of lower limb muscles to perform more efficiently (i.e. improve oxidative profile of knee and hip flexors). Therefore, a more realistic design is necessary for coaches to train cyclists to improve pedal force effectiveness during regular training sessions.

There has been increased use of power meters to monitor training and racing status (Abbiss, et al., 2006; Vogt, et al., 2006). Although the most popular power meter (i.e. SRM®) has been shown to be valid and reliable to measure power output (Abbiss, Quod, Levin, Martin, & Laursen, 2009), the additional device developed to compute crank torque (i.e. SRM® torque analysis system) and used to assess cycling asymmetries (Carpes, Rossato, Faria, & Mota, 2007a, 2007b) and pedaling technique (Edwards, Jobson, George, Day, & Nevill, 2009) has not been validated. The sensitivity of the SRM® torque analysis system to asymmetries of cyclists has not been shown compared to instrumented pedals developed to compute pedal forces and crank torque independently for right and left pedals and cranks. In theory, the SRM® should not be accurate to measure right and left crank torque but research evidence is needed to confirm this hypothesis.

Studies using biomechanical modelling of the knee joint have been conducted to evaluate effects of changes in saddle height (Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011) and pedalling directions (Bressel, 2001) on forces on the knee. Patellofemoral and tibiofemoral forces have been estimated with limitations related to assumptions of balance in quadriceps to patellar tendon force (Bressel, 2001) and analysis of only joint reaction forces (McCoy & Gregor, 1989). An improved model of the knee joint is required using freeware scripts, commercially available pedal force systems and high-speed video cameras for clinical assessment.

The position of the saddle is important for optimal performance and there seems to be a range of saddle heights that minimize oxygen uptake. However, authors disagree on an optimal saddle height. Shennum and DeVries (1976) and Nordeen-Snyder (1977) reported that a 5% reduction in saddle height resulted in a 5% change in oxygen uptake. For the horizontal position of the saddle, reduced oxygen uptake for a given workload was observed for 14 competitive cyclists using a more forward saddle position (Price & Donne, 1997). Therefore, in terms of oxygen uptake, the optimal saddle height seems to be in a range between 96-100% of the trochanteric leg length or 25° of knee angle at the bottom dead centre (Peveler, et al., 2005). For the horizontal position of the saddle, the optimal relationship between the joint angles (i.e. 25° for the knee angle at the bottom dead centre) and the body position on the bicycle (i.e. position of the knee relative to the pedal spindle) is not clear. Some pilot studies have resulted in changes in force profiles when the saddle was moved forward/backward (Diefenthaeler, et al., 2008; Rottenbacher, Bonandrini, Mimmi, & Buzzi, 2007; Rottenbacher, et al., 2009). No conclusive results could be found for an optimal configuration of the saddle (height and horizontal position) in terms of pedal and knee forces.

Questions addressed in this thesis

Given limitations in the literature, the overall question of this thesis is “What are the effects of saddle position on pedalling technique and what methods should be used to assess pedalling kinetics and kinematics of cyclists and triathletes?” Specific questions are:

- What is the validity and reliability of methods for determining body position (kinematics) and pedalling kinetics in cycling?
- What are the effects from changes in saddle height and horizontal position on pedalling kinetics and kinematics of cyclists and triathletes?

Structure of the PhD thesis

The thesis is structured into three thematic areas (see Figure 1.1). Theme 1 ‘Body position and injury risk factors’ determined the context of the overuse injury issue and what body positions are currently used by road cyclists and triathletes. Existing methods for bicycle configuration components reported in the literature are analysed along with body positions on the bicycle of cyclists of different disciplines via a cross-sectional experimental study. Theme 2 ‘Methods for cycling biomechanics assessment’ determined accuracy and validity of methods usually employed to evaluate pedalling kinetics and kinematics. A literature review, a reliability study, two validity studies, a cross-sectional study, a pilot training interventional study and the developement of a new knee model were conducted. Theme 3 ‘Saddle position changes and knee forces, force effectiveness and kinematics’ assessed the effects of saddle height on knee joint forces and pedalling technique. A literature review and two acute intervention studies were conducted. Given the use of similar methods for various chapters, there is some overlap in chapters when reporting methodological details.

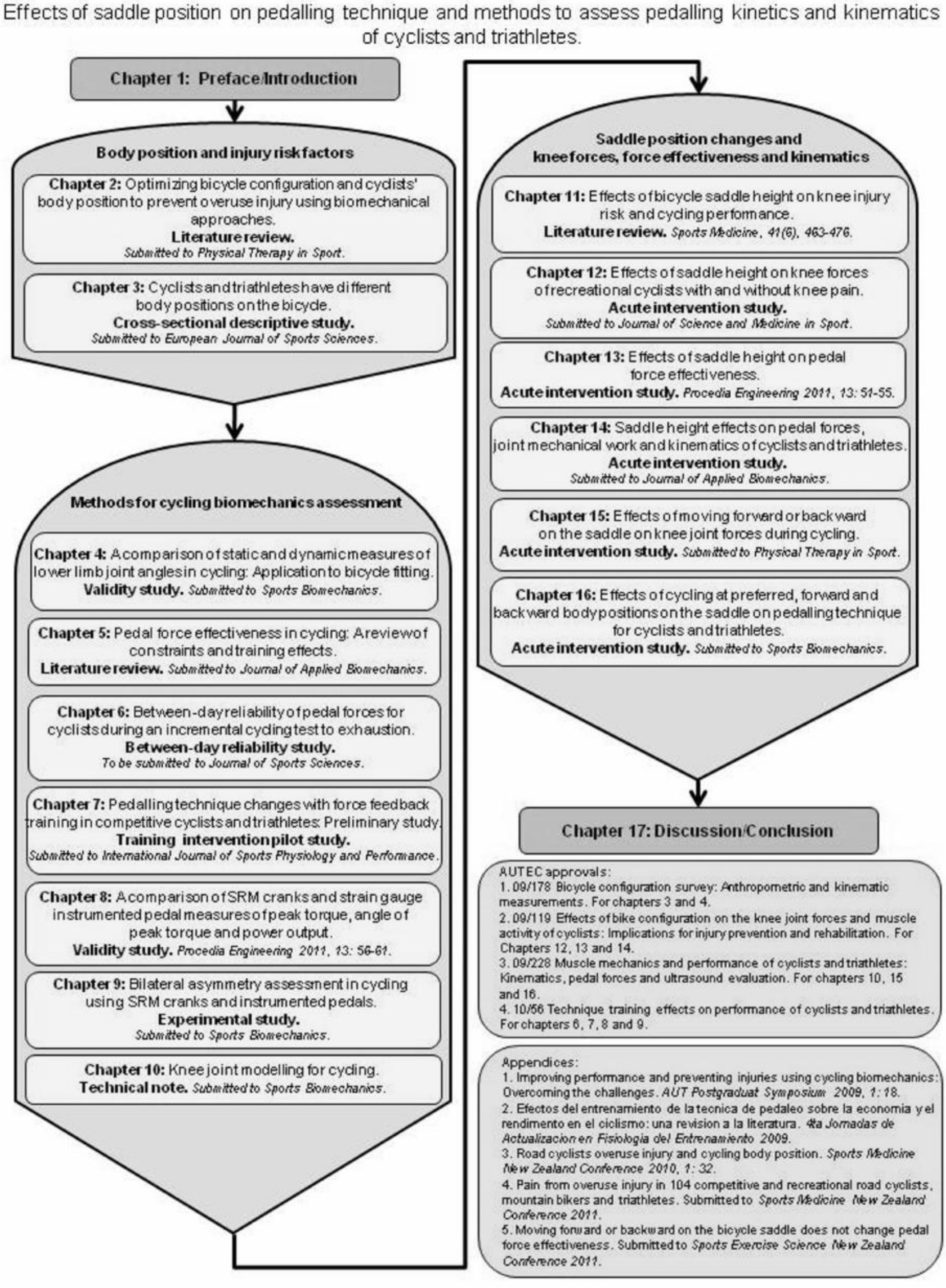


Figure 1.1: Overview of the structure of the thesis.

Chapter 2 is a review of the literature on methods and effectiveness of bicycle configuration components to prevent overuse injuries. There is currently little evidence that inappropriate bicycle configuration increases the risk of overuse injuries, or that optimization of bicycle configuration and cyclists' body position can prevent overuse injury. Optimizing body position on the bicycle theoretically has the potential to prevent overuse injury in cycling, however, biomechanical studies to date have not shown clear evidence that they can reduce overuse injury via optimization of bicycle configuration. This information formed the premise for Chapters 3, 4 and 10.

Chapter 3 consists of a cross-sectional comparison of body position on the bicycle of 36 recreational road cyclists, 16 compatitive road cyclists and 18 triathletes. Competitive triathletes had greater body forward projection (10% greater trunk flexion and 66% knee anterior position) and less frontal projected area (20%) than competitive road cyclists. Both recreational and competitive cyclists sat on their bicycles with their trunks in a more vertical position compared to triathletes. Guidelines for bicycle configuration for triathletes and road cyclists need to consider the body positions during events.

Chapter 4 consists of a validity study comparing lower limb joint angles in the sagittal plane taken statically and dynamically in 30 cyclists. Cyclists were not able to replicate in a static pose at the 6 o'clock crank position similar hip, knee and ankle joint angles as measured in dynamic cycling. To perform configuration of bicycle components using joint angles, measurements should be taken dynamically or with the cyclists in static poses at the 3 o'clock crank position, instead of the usually recommended 6 o'clock crank position.

Chapter 5 is a literature review on pedal force effectiveness in cycling. Pedal forces are often based on the computation of the index of effectiveness. Workload level and pedaling cadence affect pedal force effectiveness, but there are unclear effects of body position on the bicycle, fatigue state, cycling experience and ability on pedal force effectiveness. Technique training, using either augmented feedback of pedal forces or decoupled cranks, increases pedal force effectiveness in short duration studies but evidence of augmented feedback efficacy in long term studies is lacking. The effects of technique training trying to improve force effectiveness on economy/efficiency and performance are unclear. This information formed the premise for Chapters 6, 7 and 9.

Chapter 6 is a between-day reliability study comparing pedal force variables from different cycling sessions. Ten cyclists with competitive experience performed two incremental cycling tests to exhaustion separated by two to seven days. Differences between testing days in pedal force variables ranged from 5% to 14% with magnitudes from trivial to moderate. Greater repeatability was found for peak normal and total force applied on the pedals, with variability increasing for anterior-posterior force and pedal force effectiveness. Pedal force variables were highly reliable between two to seven days of testing with similar results compared to oxygen uptake assessed during an incremental step test to exhaustion.

Chapter 7 is a pilot training intervention study to compare effects of types of force feedback training in cycling performance. Six cyclists performed eight technique-training sessions on a cycle ergometer with visual feedback of bilateral pedal forces. Three cyclists in the “force effectiveness” group (FEG) received feedback during training sessions of pedal force effectiveness. Three cyclists in the “peak force” group (PFG) received feedback only for peak normal force. There were large increases in right normal and resultant pedal force application for the group receiving force effectiveness training (FEG) and for the group receiving peak force feedback (PFG). Large improvements in right pedal force effectiveness contrasted with large reductions in left pedal force effectiveness for FEG and PFG. Average power output was greater and pedalling cadence was slower in the FEG group and there was a small decrease in performance time. Preliminary results indicate that force effectiveness training can be translated into greater power output but not in better performance time during 4-km time trials.

Chapter 8 is a validity study comparing torque and power output measures from the SRM® torque analysis system and instrumented pedals. Seven competitive cyclists performed an incremental test to exhaustion on a stationary cycle ergometer equipped with an SRM® torque analysis system and a pair of strain gauge instrumented pedals system. The SRM® torque analysis system overestimated power output, underestimated peak torque and increased the angle of peak torque compared to the SGI pedals. Where possible a strain gauge instrumented pedals system should be used to measure variables contributing to cyclists performance rather than the SRM® torque analysis system. The findings from this study provided rationale for chapter 9.

Chapter 9 is a validity study comparing bilateral asymmetries in 11 cyclists using the SRM® torque analysis system and instrumented pedals. Greater peak torques for right (16-7%) and left (13-6%) cranks were observed for the instrumented pedals compared to the SRM® torque system between 100 W and 250 W. There was a trend for an increase in differences between right and left crank torque as workload increased using the SRM® torque system (8-33%) and the instrumented pedals (5-66%), but large differences were only found for the instrumented pedals at workloads higher than 200 W. Lower limb asymmetries in peak torque increased at higher workload levels in favour of the dominant leg. Limitations in design of the SRM® torque analysis system may preclude the use of this system to assess crank torque symmetry.

Chapter 10 is a technical note providing an improved type of modelling for the knee joint. The model enabled computation of patellofemoral compressive, tibiofemoral compressive and tibiofemoral anterior-posterior forces normalized by workload, using inverse dynamics based on sagital plane lower limb kinematics and pedal forces during cycling. The patellar tendon to quadriceps force ratio and the contribution of muscle forces to tibiofemoral joint forces have been additions to previous published models. The model improves assessment of knee loads for clinical assessment. The model was latter used in chapters 12 and 15.

Chapter 11 is a review of the literature on saddle height effects on knee injury risk and cycling performance. Methods for determining optimal saddle height are varied, not well established and have been based on relationships between saddle height and lower limb length or a reference range of knee joint flexion. There is limited information on the effects of saddle height on lower limb injury risk but more information on the effects of saddle height on cycling performance. The main limitations from the reported studies are that different methods have been employed for determining saddle height, small sample sizes have been used, cyclists with low level of expertise have mostly been evaluated, and different outcome variables have been measured. This information formed the premise for Chapters 12 and 13 and 14.

Chapter 12 is an acute intervention study with focus on effects of changes in saddle height on knee forces of eight recreational cyclists with knee pain and 16 recreational cyclists without knee pain. Effects of saddle height were not meaningfully different for patellofemoral and tibiofemoral forces when comparing cyclists with and without knee pain. Compared to the low saddle height there were large tibiofemoral anterior forces at optimal and high saddle heights. Bicycle saddle height can probably be set within a large range of knee motion (i.e., ~44-65° determined during dynamic cycling at the 3 o'clock position) to minimise possible detrimental effects of large patellofemoral and tibiofemoral forces.

Chapter 13 is an acute intervention study assessing effects of saddle height on pedal force of 11 competitive cyclists and 11 competitive triathletes. To elicit ±10° knee flexion, changes of ±3% of the preferred saddle height were required. Changes in average resultant force with saddle height were trivial to moderate. Changes in force effectiveness with saddle height were small to moderate. Lower saddle heights produced higher resultant force but lower force effectiveness. Saddle height changes resulted in moderate effects for pedal resultant force and force effectiveness for most saddle height comparisons. Unknown effects in cyclists' joint kinetics and kinematics provided the rationale for the study presented in Chapter 14.

Chapter 14 is an acute intervention study assessing effects of saddle hieght pedal forces, joint mechanical work and kinematics of 12 competitive cyclists and 12 compatitive triathletes. Changes in saddle height up to 5% of preferred saddle height for cyclists and 7% for triathletes affected hip and knee angles. In general, higher saddle height resulted in smaller knee angle and greater knee range of motion and hip mean angle. Cyclists presented improved pedal force effectiveness at the optimal saddle height compared to the preferred saddle height and triathletes presented greater ankle work and ankle range of motion for the optimal saddle height compared to the low saddle height. Triathletes presented greater mechanical work and range of motion, and small mean angle for the hip joint compared to cyclists.

Chapter 15 is an acute intervention study assessing effects of position on the saddle on knee joint forces. Twenty one competitive cyclists and triathletes performed an incremental cycling test to exhaustion to determine maximal aerobic workload and workload of second ventilatory threshold. In a second session, they performed a 1-minute cycling trial at maximal power output then three 2-minute trials at second ventilatory threshold workload at preferred, forward and backward saddle positions. Riding at the forward and backward saddle positions did not substantially affect patellofemoral compressive and tibiofemoral compressive forces. Tibiofemoral anterior shear force was greater for the backward saddle position compared to the forward and preferred saddle positions. Small increases in knee flexion angle for a constant workload level may explain the trivial differences in patellofemoral and tibiofemoral compressive forces. Tibiofemoral anterior shear force may be more sensitive to changes in knee joint angle compared to other knee force components.

Chapter 16 is an acute intervention study assessing effects of position on the saddle on pedal forces, joint kinetics and kinematics of 12 competitive cyclists and nine competitive triathletes. Riding at the most forward position on the saddle did not result in large effects for pedal force application and index of effectiveness. Increased knee mean angle and mechanical work of the knee, rectus femoris activation and smaller hip work occurred when riding at the most forward position on the saddle, without large changes in ankle joint angle. Differences between cyclists and triathletes were not substantial. Effects of changes in saddle positions were limited to the hip and knee joints.

Chapter 17 is an overall discussion of the key findings, implications and limitations of the preceding chapters and areas for further research.

The appendices contain material from chapters 2 and 16 that were presented as conference presentations (see Appendices 1-5). Sample subject information packs, questionnaires and consent form are provided in Appendix 6, 7 and 8. Appendices 9 are notifications from the Auckland University of Technology Ethics Committee (AUTEC) regarding ethical approval for the studies.

Copyright permission forms for chapters 8, 11 and 13 already published in journals are in Appendix 10. All other chapter papers are under review in the journals indicated in Figure 1.1.

CHAPTER 2: OPTIMIZING BICYCLE CONFIGURATION AND CYCLISTS' BODY POSITION TO PREVENT OVERUSE INJURY USING BIOMECHANICAL APPROACHES

Overview

This review evaluated evidence for the effectiveness of bicycle configuration and cyclists' body position to prevent overuse injuries in cycling using biomechanical approaches. Searches of data bases of peer-review journals using key words including 'overuse injury', 'body position', 'bike fitting', 'kinetics', 'kinematics' and 'electromyography/EMG' resulted in 60 articles being reviewed. Low back and knee joint injuries are the most common overuse injuries reported in cyclists. Joint angles and distances from body segments to bicycle components have been determined statically on the bicycle or during cycling motion. Joint kinematics, pedal forces, joint kinetics, muscle activation, and anthropometric dimensions of the body have been investigated with respect to bicycle configuration and body positions. There is currently little evidence that inappropriate bicycle configuration increases the risk of overuse injuries. Greater medial projection of the knee and large rotational moment on the vertical axis of the tibia have been linked to anterior knee pain. Optimizing body position on the bicycle theoretically has the potential to prevent overuse injury in cycling, however, biomechanical studies to date have not shown clear evidence that they can reduce overuse injury via optimization of bicycle configuration.

Introduction

Overuse injuries in cycling are common (Wanich, et al., 2007) with up to 85% of cyclists sustaining one or more overuse injuries during their lifetime (Dettori & Norvell, 2006). In one year, 33 million USA residents rode a bicycle an average of six days a month, for an average of >1 hour a day (Wanich, et al., 2007), suggesting that approximately 23 million cyclists may develop at least one overuse injury in their lifetime. Overuse injuries of the anterior knee joint (e.g. chondromalacea) are most common, affecting about 50% of injured cyclists (Dettori & Norvell, 2006). Cycling overuse injuries are not confined to the lower extremity as 46% of elite cyclists reported low back pain (Clarsen, Krosshaug, & Bahr, 2010).

A cyclist is in contact with the bicycle via the handlebars, saddle and the pedals. Consequently, the way a bicycle is configured can change a cyclists' body position. For example, if the saddle height is too low then the larger knee flexion angle (defined as the angle between the tibia and the forward projection line of the femur) close to peak pedal force may increase knee joint forces and lead to overuse injuries (Callaghan, 2005). A common strategy to reduce the risk of overuse injuries is to change bicycle configuration which leads to a different body position on the bicycle (Burke & Pruitt, 2003; Holmes, et al., 1994). However, there is concern that evidence on how to configure a bicycle to prevent overuse injuries is needed. For saddle height configuration, the methods to determine knee joint angles have been shown to be inaccurate (Peveler, et al., 2005), which indicates that existing guidelines may not be reliable for bicycle configuration towards knee injury prevention. Therefore, it is necessary to review existing guidelines for bicycle configuration that aim to help prevent overuse injuries, and to present potential benefits and limitations of biomechanical methods employed in cycling research to help prevent overuse injuries. The aims of this review were (1) to evaluate evidence for the effectiveness of bicycle configuration and cyclists' body position to prevent overuse injuries in cycling using current biomechanical approaches, and (2) to present strengths and limitations of biomechanical methods for the assessment of cyclists looking for injury prevention.

Methods

Peer-reviewed journals, books, theses, and conference proceedings published since 1960 and up to July 2011 were searched using Medline, Scopus, ISI Web of Knowledge, EBSCO, and Google Scholar data bases. Keywords searched included 'bicycle', 'overuse injury', 'body position', 'bike fitting', 'kinetics', 'kinematics' and 'electromyography/EMG'. Abstracts of each article were then analyzed for inclusion if they were related to bike fitting and/or to the use of biomechanical methods to assess cyclists' body position on the bicycle. Articles were excluded if they did not have at least an English abstract, or if they focused solely on acute cycling injuries. Sixty references were reviewed. Books, thesis and conference proceedings were analysed from their abstracts and included in the review if they reported research on the issues of interest of this review. We did not assess quality of the studies for inclusion because only one case control study was found presenting effectiveness of bicycle configuration to reduce overuse related pain. A total of fourteen articles were either case control studies or cross-sectional studies looking at least one biomechanics variable and its potential effect on the assessment of body position on the bicycle for overuse injury prevention. Biomechanical techniques selected for analysis in our review were based on their most frequent use in research on cycling biomechanics (i.e. anthropometry, kinetics, kinematics, and electromyography).

Results

The low back and knee were the main body sites of overuse injuries in cyclists (Clarsen, et al., 2010; Dettori & Norvell, 2006). No randomized controlled trials investigating risk factors for overuse injury in cycling were found. Although a relationship between improper bicycle configuration and injuries in cyclists has been suggested (Wanich, et al., 2007), evidence for the effectiveness of clinical biomechanical evaluation of body position and bicycle configuration in preventing the likelihood of overuse injuries in cycling was not found. Most preventive strategies for overuse injuries were based on optimizing the interaction between body position and bicycle components (Wanich, et al., 2007). One case control study reported that 70% of cyclists reduced low back pain when the saddle was tilted with the anterior portion downward by 10-15° (Salai, Brosh, Blankstein, Oran, & Chechik, 1999). Cross-sectional studies showed differences in kinematics and kinetics of injured versus uninjured cyclists (Hunt, Sanderson, Moffet, & Inglis, 2003, 2004) (see Table 1). No study provided evidence of likely thresholds for overuse injury risk for the various biomechanical variables analysed.

Optimization of configuration of bicycle components and cyclists' body position

The height of handlebars, saddle height and horizontal position, and position of the foot on the pedal (or position of the cleats) can all be changed (De Vey Mestdagh, 1998) but the distance from the saddle to the handlebars and crank length seem to be the most important for bicycle configuration components. Optimal muscle force production occurs within a certain range of muscle length, and consequently within a range of joint angles (Rassier, MacIntosh, & Herzog, 1999). Therefore, it may be expected that optimal joint angles would lead to maximal force production. However, most of the references for optimizing the configuration of bicycle components are based only on the length of body segments (De Vey Mestdagh, 1998) without concern for joint angles.

Handlebars vertical and horizontal position

The horizontal (distance from the handlebars to the saddle) and vertical (height) positions of the handlebars affect the upper body flexion angle (De Vey Mestdagh, 1998). The position of the handlebars has been empirically related to the sum of lengths of the torso and the arm (De Vey Mestdagh, 1998). The effects of the combined horizontal and vertical position of the handlebars will dictate the angles of the trunk and pelvis (see Figure 2.1).

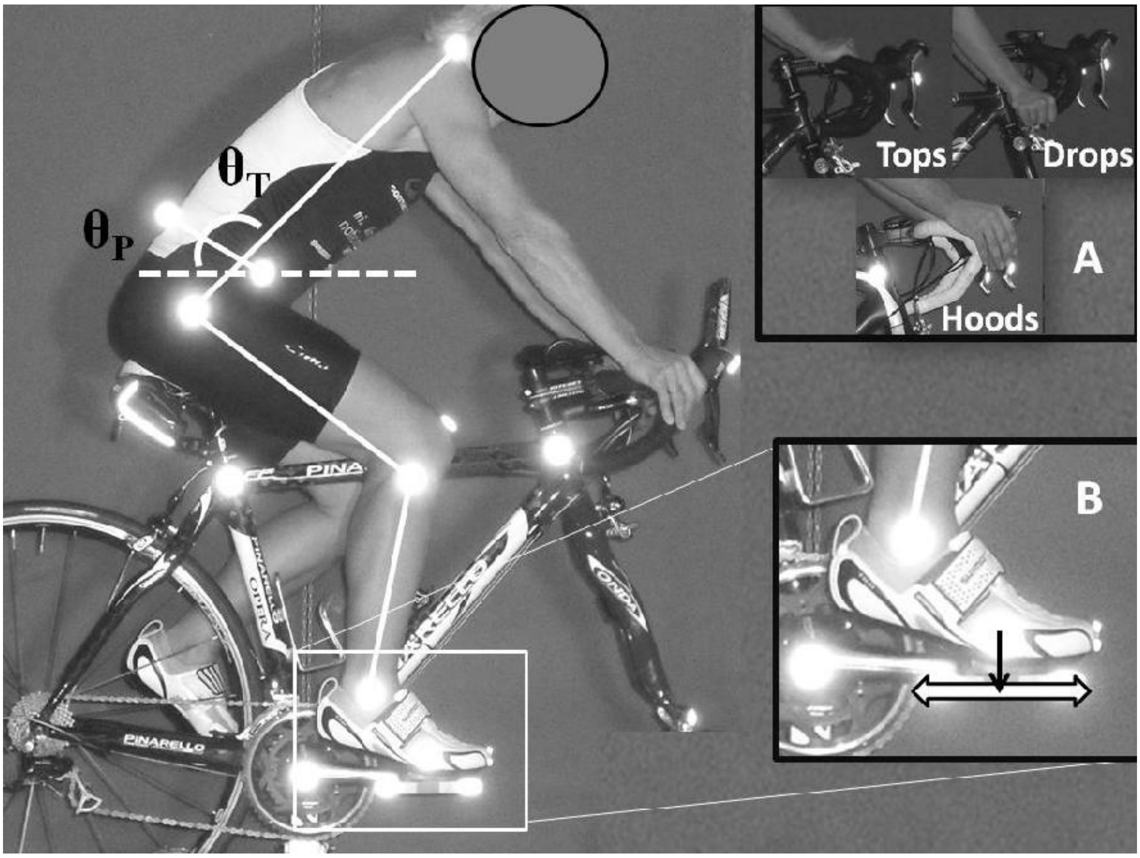


Figure 2.1. Flexion angle of the trunk (θT) and pelvis (θP) in the sagittal plane. Insert A: Three positions of the hands on the handlebars. Insert B: Forward-backward configuration of the shoe on the pedal.

To allow comparisons between studies, trunk and pelvis angles have been defined as presented in Figure 2.1. Trunk and pelvis angles were smaller with the hands on the top of the handlebars (141 ±0.9° for the trunk and 16 ±1.4° for the pelvis) than when the hands were on the brake hoods (156 ±0.7° for the trunk and 29 ±1.6° for the pelvis) for 20 female cyclists riding at 140 W of workload and 80 rpm of pedalling cadence (Bressel & Larson, 2003). This finding highlights that if the hands are further away in relation to the body, trunk and pelvic angles will be increased. Increased angles between the lumbar vertebral disks were observed when comparing hands on the top of the handlebars (~2.5°) with the hands on the drops of the handlebars (~3°) by radiologic analysis (Usabiaga, et al., 1997) of three professional cyclists positioned statically on their bicycles. This result indicates that a forward inclined upper body position of cyclists (compared to upright standing) will affect lumbar disk displacement with small effects from changes in position of the hands on the handlebars.

High occurrence of low back pain (up to 60%) (Burnett, Cornelius, Dankaerts, & O'Sullivan, 2004) have been reported for cyclists, which may result from forward trunk flexion and pronounced lumbar kyphosis on the bicycle compared to the standing upright position (Usabiaga, et al., 1997). Burnett et al. (2004) reported no significant difference between the pelvic angle for nine cyclists with (15 ±5.1°) and without (23 ±5.8°) low back pain during cycling with the hands on the drops of the handlebars. It is possible however that large between-subject variability in trunk and pelvic angles may hide any relationship between lower back pain and upper body position in cycling (Burnett, et al., 2004). Furthermore, both saddle shape (Bressel & Larson, 2003) and inclination angle (Salai, et al., 1999) have been found to affect the angle of the pelvis which may make it difficult to isolate the relationship between trunk angle and low back pain. The interaction effects between the horizontal and vertical positions of the handlebars, the angles of the trunk and pelvis and any relationships with overuse injuries are not yet understood.

Saddle height and horizontal position

The saddle can be adjusted in the vertical and horizontal directions without changing its inclination angle in the sagital plane. The position of the saddle can be measured relative to the bicycle components (e.g. saddle position relative to the bottom bracket) or relative to the cyclist (e.g. saddle height relative to the trochanteric length). When using the length of body segments to configure the saddle height, the distance from the greater trochanter to the floor, the distance from the pubis to the floor and the distance from the ischial tuberosity to the floor have all been used (Bini, Hume, & Croft, 2011a). However, there is concern that using length-based configuration methods may not result in similar joint kinematics (Peveler, et al., 2005). Therefore, the range of the knee flexion angle (e.g. 25-30° when the crank is at the 6 o'clock crank position) has been recommended as a more reliable method of configuring saddle height (Peveler, et al., 2005) to achieve consistent lower limb kinematics.

For the horizontal position of the saddle, it has been recommended that the vertical projection of the knee should intersect the pedal axis (Wanich, et al., 2007) based on the assumption that knee positions forward of the pedal axis may result in higher compressive forces on the patellofemoral joint (Wanich, et al., 2007). Triathletes often use a more forward position of the saddle compared to road cyclists (Ricard, Hills-Meyer, Miller, & Michael, 2006) which suggests that they should have a higher risk of knee injuries than road cyclists but prevalence of knee injuries in triathletes (14% to 63%) (Gosling, Gabbe, & Forbes, 2008) and cyclists (21% to 65%) are similar (Dettori & Norvell, 2006).

Position of the foot on the pedal

The position of the foot on the pedal can be changed in several ways. When cyclists use a cleat between the shoe and the pedal the position of the foot-shoe in relation to the pedal can be changed in the anterior-posterior direction and rotated about the longitudinal axis (see Figures 2.1 and 2.2, respectively).

The most common recommendation for optimal foot positioning is that the ball of the foot should lie over the pedal axis (see Figure 2.1) (Wanich, et al., 2007). However, when the horizontal position of the foot on the pedal is varied (forward or backward, see Figure 2.1), no significant effects on knee joint forces (Ericson & Nisell, 1987), muscle activity (Litzenberger, Illes, Hren, Reichel, & Sabo, 2008), pedal forces (Ericson & Nisell, 1988) or oxygen uptake

(Paton, 2009) were observed. Therefore, there is no current evidence to support that the horizontal position of the foot on the pedal can be optimised to reduce the risk of overuse injuries or improve performance.

Most cleats allow the shoe to rotate around the vertical axis (toes inward or outward) to a certain extent. It has been suggested that cleats that do not enable such rotation movements may increase forces on the knee joint and increase the likelihood of injuries (Gregor & Wheeler, 1994). It is likely that some rotational movement between the cleat and the shoe with respect to the vertical axis may improve force transfer from the legs to the pedals. However, the optimal range of motion around the vertical axis is currently unknown.

The position of the foot-shoe along the longitudinal axis of the pedal (medio-lateral displacement of the foot) has been related to the knee angle in the frontal plane (Sanderson, Black, & Montgomery, 1994). It is usually recommended that the knee should remain directly over the pedal axis in the frontal plane during the pedalling cycle (see Figure 2) (Sanner & O'Halloran, 2000). However, the knee moves medially between the 12 o'clock and 6 o'clock crank positions (propulsive phase) and externally between the 6 o'clock and 12 o'clock crank positions (recovery phase)(Bailey, Maillardet, & Messenger, 2003). The amount of medial knee movement that is tolerated is unknown but cyclists with anterior knee pain have been shown to have 320% greater medial displacement than cyclists without anterior knee pain (Bailey, et al., 2003) indicating that excessive medial movement may be associated with knee pain.

It is therefore important to consider the position of the foot-shoe relative to the pedal and its effect on the position of the knee (e.g. such as when using orthotics) to reduce knee injury risk and improve cycling economy.

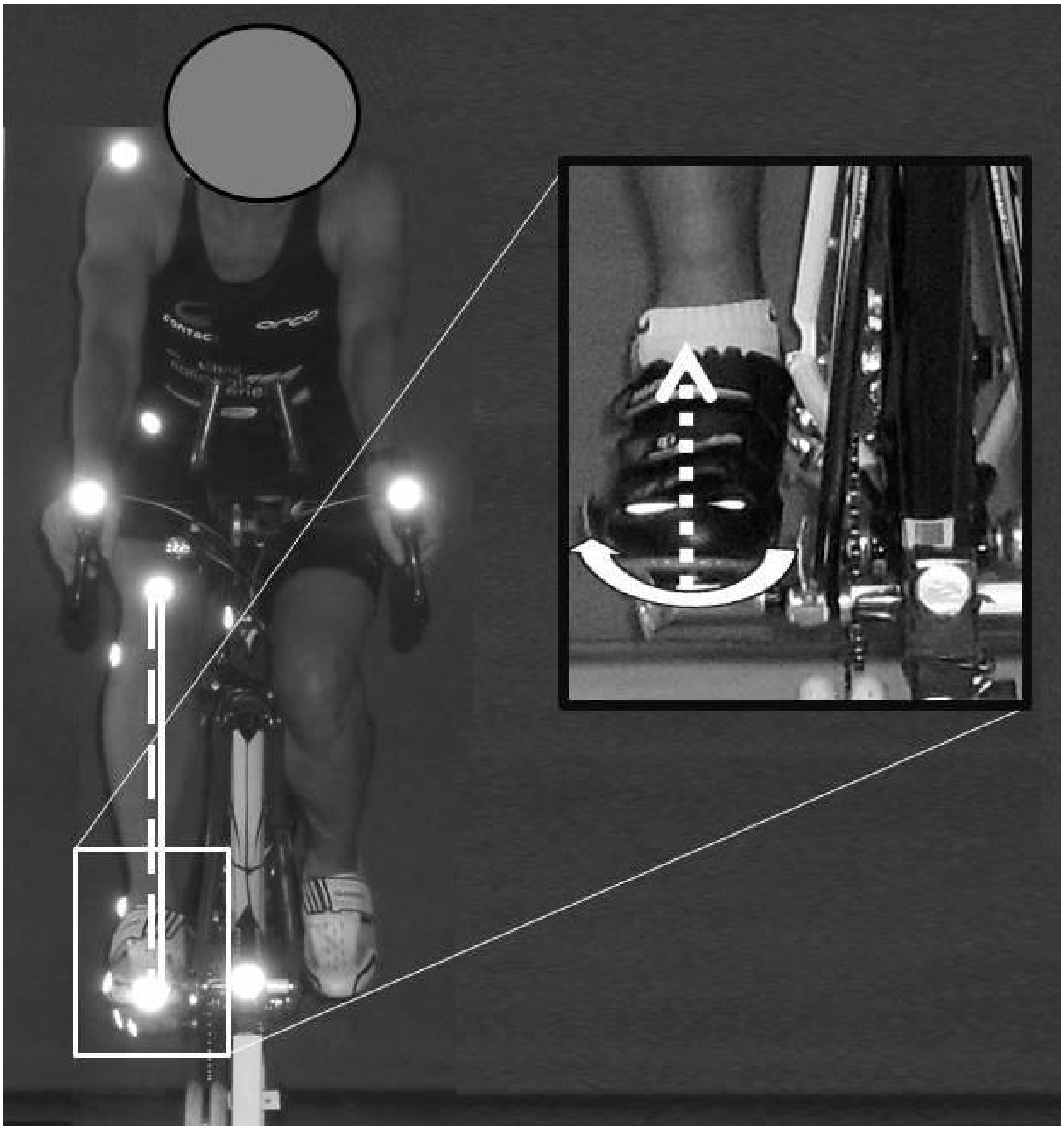


Figure 2.2. Medial projection of the knee (continuous line) in relation to the pedal axis (dashed line) in the frontal plane. Insert: Medio-lateral and rotational configuration of the shoe in relation to the vertical axis of the pedal.

Biomechanical approaches for optimizing bicycle configuration and cyclists' body position to prevent overuse injury

The most frequently suggested strategy to reduce injuries in cyclists is to optimize body position (Wanich, et al., 2007) by adjusting the bicycle configuration to fit the cyclist's anthropometry (De Vey Mestdagh, 1998). Additional approaches include altering pedalling technique using video analysis and force measurement, muscle activity and simulation models. Table 2.1 summarizes studies that have used biomechanical approaches to provide analyses of joint kinetics and kinematics and muscle activity in the scope of injury prevention and rehabilitation for overuse injuries in cyclists.

Table 2.1. Summary of studies using a biomechanical approach to provide analyses of joint kinematics, kinetics or muscle activity within the scope of injury prevention and rehabilitation for overuse injuries in cyclists.

Study	Biomechanical method; Participants; Injury in focus.	Main results and notes
Gregor and Wheeler (1994)	Review/case study; 1 cyclist with anterior knee pain, 27 uninjured cyclists. No gender or age given.	Increased moment about the vertical axis of the tibia for the symptomatic cyclist compared to the uninjured cyclists. Magnitude of difference not reported.
McLeod and Blackburn (1980)	Joint kinematics, muscle activity; 6 male uninjured non-cyclists aged 21 to 30 years. Injuries related to tibiofemoral (shear and compressive) and patellofemoral forces.	Proposed cyclists should activate hamstrings, hip flexors and tibialis anterior to prevent excessive compressive forces on the patella from quadriceps activation.
Semwal and Parker (1999)	Joint kinetics and kinematics; 4 elite uninjured cyclists with asymmetry on bilateral knee movement. No gender or age given. Knee injuries related to rotational moment on the vertical axis of the tibia.	Software graphical interface for analyses of joint moments and kinematics. Focus on provision of summarized feedback for cyclists and coaches. Descriptive analysis of possible increased risk of injuries due to greater rotational moment about the vertical axis of the tibia.
Farrell et al. (2003)	Joint kinematics; 6 male and 4 female uninjured non-cyclists aged 31 ±5.5 years. Iliotibial band.	Analysis of knee flexion angle to provide possible overload of iliotibial band. Diverse causes for iliotibial band impingement (anatomical, kinematics, forces). More time in the impingement zone (knee flexion < 30°) in cycling because of more repetitive action compared to running.
Bailey et al. (2003)	Joint kinematics; 24 active male cyclists (14 uninjured and 10 injured) aged 28 ±8.4 years. Patellar tendinitis and/or anterior knee pain.	A more medial position of the knee (~2.5° - 320%) in the injured group compared to the uninjured group.
Burnett et al. (2004)	Joint kinematics, muscle activity; 9 cyclists without pain (four male and five female aged 37 ±7.9 years) and 9 cyclists/triathletes with non-specific chronic low back pain	Low back pain group presented ~50% greater lumbar flexion compared to pain-free cyclists with ~50% increased rectus abdominis activity.

	(4 male and 5 female aged 42 ±9.7 years).	
Joseph et al. (2006)	Joint kinematics; 1 male 27 year old elite cyclist with non-specific chronic low back pain.	Descriptive increased hip movement in the frontal plane related to back pain.
Salai et al. (Salai, et al., 1999)	Joint kinematics; 40 uninjured recreational cyclists with non-specific chronic low back pain. No gender or age given.	70% reduced low back pain after 10-15° reduced saddle angle (lower forward portion compared to backward portion of the saddle). A prospective case control study.
Tamborind-eguy and Bini (2011)	Joint kinetics and kinematics; 9 male uninjured non-cyclists aged 22 and 36 years. Evaluated potential for injuries related to tibiofemoral compressive and shear, and patellofemoral compressive load.	No differences in peak knee force when reducing or increasing saddle height by 3 cm compared to the trochanteric height when cycling at low workload and pedalling cadence. Knee flexion angle at the 6 o'clock crank position was increased by 41% at the low saddle position compared to the reference and by 18% compared to the high saddle position. Optimising knee angle may reduce overload on the knee joint for low workload and pedalling cadence.
Ericson et. al. (1984)	Joint kinetics; 6 male uninjured non-cyclists aged 20 and 31 years. Evaluated potential for injuries related to medio-lateral moments on the knees.	Descriptive profile of medio-lateral moments on the knee. Higher medial knee forces (~120%) when knees close to the bicycle frame compared to the preferred position of the knees.
Ericson and Nisell (1986)	Joint kinetics; 6 male uninjured non-cyclists aged 20 and 31 years. Evaluated potential for injuries related to compressive and shear forces on the tibiofemoral joint.	Description of knee moment profile. Peak knee moment and compressive force between 60-100° knee angle. Peak anterior shear force close to 65° knee flexion angle. Compressive force increased with higher workload (~100%) or lower saddle height (~64%). No effects of pedalling cadence on tibiofemoral forces.
Ericson and Nisell (Ericson & Nisell, 1987)	Joint kinetics; 6 male uninjured non-cyclists aged 20 and 31 years. Evaluated potential for injuries related to compressive forces on the patellofemoral joint.	Peak patellofemoral force close to 80° knee flexion angle. Patellofemoral compressive force increased at higher workload (~75%) and lower saddle height (~25%). No effects of pedalling cadence on Patellofemoral compressive force.

Ruby et al. (1992)	Joint kinetics; 11 male uninjured cyclists aged 23 to 45 years. Evaluated potential for injuries related to rotational moments on the vertical axis and medio-lateral moments on the knees.	Descriptive profile of the three-dimensional forces and moments on the knee joint. Important to analyze the knee based on three-dimensional kinetics. Hypothesised that increased rotational moment on the vertical axis of the tibia may lead to greater injury risk.
Srinivasan and Balasubra-manian (2007)	Muscle activity; 14 male cyclists (7 with non-specific chronic low back pain, 7 without low back pain) aged 25.4 ±1.9 years.	Higher fatigue level based on muscle activity analysis for the right trapezius medialis and erector spinae for the low back pain group after 15 minutes of outdoor cycling covering 6 km.

Studies were mostly conducted using uninjured cyclists and/or non-cyclists. Cyclists with and without patellar tendonitis and/or anterior knee pain (Bailey, et al., 2003), or non-specific low back pain have been compared (Burnett, et al., 2004) and characteristics of cyclists with low back pain before and after changing saddle angle (Salai, et al., 1999) have been reported. However, no study has yet reported likely thresholds for biomechanical variables in predicting injury.

Anthropometrics

Several studies have highlighted the relationship between improper bicycle configuration and injuries in cyclists (Wanich, et al., 2007). However, there are various complications in applying optimal configurations. Appropriate joint angles and distances from body segments to specific components of the bicycle that have been suggested are not usually based on scientific research (Peveler, et al., 2005). In addition, various measures are inter-related and there is disagreement regarding which should be measured (Peveler, et al., 2005). Typical static analysis (De Vey Mestdagh, 1998) is inaccurate because it fails to account for the variability of joint dynamics observed while cycling, which is greater for less experienced cyclists (Chapman, Vicenzino, Blanch, & Hodges, 2009). Farrell et al. (2003) described that after setting the saddle height for a knee flexion angle of 30º (0° is full extension) with the crank at the 6 o'clock crank position prior to exercise, this angle changed during cycling to a range between 30-45º in ten non-cyclists. Defining optimal parameters based on knee and ankle angles measured dynamically should be better than using angles derived from static postures.

The relationship between anthropometric characteristics of the cyclist (e.g. inseam leg length) and the configuration of bicycle components (e.g. saddle height) has been related to comfort in 75 recreational cyclists (Christiaans & Bremner, 1998). However, the use of anthropometric methods based on segment length resulted in different knee flexion angles for different cyclists (Peveler, et al., 2005). Reduced accuracy of length-based methods for configuration of saddle height may increase the between-subject variability for joint kinematics.

Poor relationships between the horizontal configuration of the saddle and the upper-leg length have been presented (Christiaans & Bremner, 1998). The optimum saddle height for young cyclists with similar height (e.g. 145-149 cm) was calculated using length-based methods in participants 7-14 years old (Laios & Giannatsis, 2010) and found to have high variability (17%). Therefore, predicting the relative position of bicycle components from cyclists' anthropometric dimensions may not be as accurate as has been advocated (De Vey Mestdagh, 1998).

No published study has explored the possible relationship between injury and bicycle configuration by relating anthropometric characteristics and the history of injuries to the joint angles based on static or dynamic body positions on the bicycle. Limitations exist by assuming that the static position is representative of the joint position during riding and this must be taken in account.

Dynamic joint kinematics

The rapid development in technology has provided significant benefits for movement analysis (Liebermann, et al., 2002). For example, three-dimensional body kinematics measurements are now available without the need to attach skin markers (Goffredo, et al., 2009) and the result is more realistic tracking of movement. Unfortunately, these systems are expensive and unaffordable for cycling coaches and bicycle shops, where the configuration of bicycle components for an individual usually occurs. Fortunately, technological advancement and the reduced price of high-speed video cameras have been helpful for coaches using basic image analysis to provide feedback to cyclists.

Joint kinematics are sensitive to changes in body position on the bicycle (Sanderson & Amoroso, 2009) and to the presence of injury (Bailey, et al., 2003). Upper (Joseph, et al., 2006) and lower body injuries (Bailey, et al., 2003) have been related to altered kinematics which may lead to overload on soft tissues. The major resistive force in cycling, aerodynamic drag, can be reduced by increasing trunk angle to a more flexed position which decreases the frontal projected area (García-López, et al., 2008). However, for cyclists that suffer from low back pain, increasing upper body flexion may overload the posterior annulus of intervertebral discs (Burnett, et al., 2004) and may cause asymmetries in frontal plane kinematics of the low back (Joseph, et al., 2006). However, no study has reported trunk and pelvic flexion angles that could minimize overload on intervertebral discs.

The relationship between kinematics and injuries has been described for the knee joint (Bailey, et al., 2003) but not for the hip and the ankle joints. High variability in shank rotation about the vertical axis has been reported for international level uninjured cyclists (Semwal & Parker, 1999) and therefore may increase the risk of knee injuries. Increased frontal plane movement of the knees (Bailey, et al., 2003) and higher rotation moment about the vertical axis (Gregor & Wheeler, 1994) have been observed in cyclists with anterior knee pain compared to uninjured cyclists.

High variability of joint angles has been associated with poor movement skill in novice cyclists (Chapman, et al., 2009) and to the likelihood of injury occurrence in other sports

(Bartlett, Wheat, & Robins, 2007). For the ankle joint, kinematics were found to be substantially different among cyclists of similar performance levels (Kautz, Feltner, Coyle, & Bailey, 1991) and between road and mountain bike riders (Carpes, et al., 2006). However, it is unclear if variability in joint kinematics can affect injury risk in cyclists.

Pedal forces and joint kinetics

Studies have quantified pedal forces during controlled laboratory trials (Dorel, Drouet, Couturier, Champoux, & Hug, 2009). To increase ecological validity, Stapelfeldt et al. (2007) developed an adaptor to be used between the pedal and the crank that allowed cyclists to use their own pedal. With further development, portable systems that can be used in the field (Dorel, Drouet, Hug, Lepretre, & Champoux, 2008) may provide more ecological measurements of cycling mechanics. Unfortunately, these systems are not yet widely available for use by most coaches and cyclists, and usually only biomechanics laboratories provide this equipment. However, whilst wireless systems will no doubt become more available in the future, there is still a need to determine their reliability and accuracy.

The quantification of bilateral pedal forces may be useful in evaluating asymmetries of anterior cruciate ligament injured cyclists (Hunt, et al., 2003, 2004). The combined measurement of joint kinematics and pedal forces can provide an estimate of joint kinetics, using an inverse dynamics model, which has been used to describe changes in joint kinetics in injured cyclists (Hunt, et al., 2003). A higher rotational moment about the vertical axis of the tibia has been reported for one cyclist with anterior knee pain (Gregor & Wheeler, 1994), while significantly lower knee extensor moments have been reported for anterior cruciate ligament injured non-athletes during pedalling (Hunt, et al., 2003).

The uninjured leg is often used as a control when looking at the effects of injury on joint kinetics. However, asymmetries in hip (Smak, Neptune, & Hull, 1999) and knee joint kinetics (Semwal & Parker, 1999) have been reported in trained uninjured cyclists which make comparisons difficult to interpret. It is currently unclear what threshold of asymmetry can be defined as “functional” for uninjured cyclists for the purpose of injury prevention.

Cycling is commonly analyzed as a pure sagittal plane movement, which neglects medio-lateral forces that affect non-driving structures of the joints (Ruby, et al., 1992). Complete three-dimensional analysis of joint kinetics has rarely been reported (Ruby, et al., 1992). To give coaches an understanding of the various forces acting upon a cyclist, Semwal and Parker (1999) developed an animated computer software that provides selectable analyses of three-dimensional pedal forces and joint kinematics. Coach friendly computer software like this may enhance the use of joint kinetics as a tool for injury prevention. One example is the identification of rotational moments at the knees that are related to overload on the menisci and cruciate ligaments (Gregor & Wheeler, 1994). Once assessed using three-dimensional joint kinetics analysis, rotational knee moments can be minimized by using floating rather than fixed cleats at the shoe-pedal interface (Gregor, 2000).

Inverse dynamics analysis in the sagittal plane has been used to calculate the moments and resultant forces on the joints and such measures may be useful in preventing joint overload (Ericson & Nisell, 1986). Tibiofemoral compressive and shear force components in the sagittal plane have been compared using different bicycle configurations (e.g. saddle height), however, conflicting results were found (Ericson & Nisell, 1986; McCoy & Gregor, 1989; Tamborindeguy & Bini, 2011) potentially due to differences in cycling experience (i.e. cyclists versus non-athletes) or magnitude of changes in saddle height. Therefore, further research should determine the effects of bicycle configuration on sagital and three-dimensional forces of the lower limb joints.

Muscle activation

The profile of neural drive using surface electromyography has been reported in studies comparing cyclists of varying experience (Candotti, et al., 2009; Chapman, Vicenzino, Blanch, & Hodges, 2007), comparing effects of different bicycle configurations (Ericson, Nisell, Arborelius, & Ekholm, 1985), and in injured cyclists (Hunt, et al., 2003). Concern has been raised when using EMG over multiple days due to problems in repositioning electrodes (Merletti, Botter, Troiano, Merlo, & Minetto, 2009). However, a study (Malek, et al., 2006) that normalised electromyographic signals using a maximal power output cycling trial reported no significant differences in muscle activation when comparing sub-maximal workload levels using different between-electrodes distances.

McLeod and Blackburn (1980) conducted a qualitative analysis of muscle activation during cycling and suggested intervention based on muscle recruitment patterns (e.g. increasing activation of flexor group on the recovery phase) for rehabilitation during a single cycling bout. Experimental studies are yet to be conducted looking at the effectiveness of feedback from muscle activation while cycling motion.

A model for saddle height optimization has been proposed based on lowering the activation of vastus lateralis, vastus medialis, biceps femoris and gastrocnemius lateralis (Matsumoto, Tokuyasu, & Ohba, 2009). However, the case study approach and limitations in methods and analysis preclude any conclusions at this stage. A better definition of workload and cadence, the use of more comprehensive muscle activation amplitude analysis, assessment of more cyclists and the use of a reference condition to normalize the amplitude of muscle activation may improve the reliability of this technique for optimizing bicycle configuration.

The use of electromyography for biofeedback for patients after stroke (Jonsdottir, et al., 2007) and functional electrical stimulation for spinal cord injured patients (Triolo & Bogie, 1999) may be useful as a methodology for modifying activation patterns while cycling. No study was found using biofeedback or functional electrical stimulation to change muscle activation patterns in cyclists.

Simulation models of musculoskeletal mechanics incorporating muscle activation characteristics have shown ~33% lower patellofemoral compressive force during forward compared to reverse pedalling, which has been used in rehabilitation procedures and has the potential to prevent soft tissue overuse injuries (Neptune & Kautz, 2000). Further research using simulation models is needed to assess changes in bicycle configuration and its effect on joint forces.

Conclusion and practical application

The biomechanical approach for preventing overuse injuries in cyclists is still based on the optimization of body position on the bicycle. For this approach, static analysis of the cyclist on the bicycle is the most used procedure but this neglects variance in force and kinematics of the joints. The references for “optimum” configuration of bicycle components have mostly been based on empirical knowledge rather than experimental results. To date, few studies have shown that increased variability of force (Gregor & Wheeler, 1994) and joint kinematics (Bailey, et al., 2003) leads to more injuries in cyclists. Pedal force and joint kinetics analyses are still limited to biomechanics laboratories because of the complexity and high cost of purchasing these systems. The use of electromyography based biofeedback has not been reported for cycling, though successful results for other injured populations (e.g. stroke patients) suggests this could be a useful avenue for further research.

Sports equipment has been optimised based on muscle physiology characteristics (Neptune, McGowan, & Fiandt, 2009), but few studies have used mathematical models to optimize the configuration of bicycle components (Gonzalez & Hull, 1989). Unknown reliability of biomechanics techniques has compromised bicycle configuration effectiveness as a clinical tool to prevent injuries in cycling. The effectiveness of various strategies and biomechanical techniques to prevent injuries in cycling therefore need to be tested by prospective controlled trial studies using appropriate epidemiological analysis to provide risk ratios for various bicycle configurations, body positions and types of overuse injuries.

CHAPTER 3: CYCLISTS AND TRIATHLETES HAVE DIFFERENT BODY POSITIONS ON THE BICYCLE

Overview

Our study evaluated differences in body position on the bicycle for recreational cyclists, competitive cyclists and triathletes. Thirty six recreational cyclists, 17 competitive road cyclists and 18 competitive triathletes were assessed for body position on their bicycles on a cycle trainer. Images were taken of cyclists/triathletes in static poses with the crank at the 3 o'clock and 6 o'clock positions. Trunk, pelvis, hip, knee and ankle angles, anterior-posterior and medio-lateral positions of the knees in relation to the pedal axis and frontal projected area were measured using ImageJ. Comparison of body position between groups (recreational, competitive road cyclists and competitive triathletes) was conducted using effects sizes (ES). The greatest differences between groups in the measured variables were observed between the triathletes and the other two groups. Smaller differences were observed between competitive and recreational cyclists. Competitive triathletes had greater body forward projection (10% greater trunk flexion and 66% knee anterior position, ES = 2.5 and 1.2, respectively) and less frontal projected area (17%, ES = 1.3) than competitive road cyclists. Both recreational and competitive cyclists sat on their bicycles with their trunks in a more vertical position compared to triathletes. Guidelines for bicycle configuration for triathletes and road cyclists need to consider the body positions during events.

Introduction

Environmental concerns regarding the use of fossil fuels for transportation and increasing health awareness has resulted in more people riding bicycles for transportation and recreation (FPCUB, 2008). Bicycles have been used for cyclists and triathletes for racing and training. For efficient bicycle riding the components of the bicycle should be adjusted to match body dimensions of the cyclist (Burke & Pruitt, 2003) regardless of the goal of the cyclists (i.e. training, racing or commuting). In addition to different sizes of frames, modern bicycles offer a range of settings for the three contact points between the body and the bicycle (saddle, pedals and handlebars). The saddle position can usually be adjusted vertically (height) and horizontally. In some bicycles, there are settings for the height and horizontal position of the handlebars. The distance between the pedals and the centre of rotation (bottom bracket) is fixed by the length of the crank and cannot be adjusted except by replacing the crankset. Consequently, various adjustments can be made with a particular frame to accommodate different body dimensions and purpose of bicycle riding.

General guidelines have been used for configuration of road cycling bicycles, regardless the goal of the rider. The objectives of commuting by bicycle are different than road racing, partly being influenced by comfort. Thus, different configurations for recreation and racing may prove beneficial. Competitive cyclists are more likely to change their bicycle configuration to reduce frontal projected area and increase bicycle speed. Greater upper body flexion and increased medial projection of the knees in the frontal plane are amongst the potential changes in body position on the bicycle that competitive cyclists may do to reduce frontal area (Burke & Pruitt, 2003). Therefore, guidelines for bicycle configuration could be revised to allow for these differences between competitive and recreational cyclists.

Bicycles designed for triathlon tend to differ from the ones used by road cyclists. In triathlon, it is forbidden to pedal in groups or even behind another rider in some races (http://www.triathlon.org/images/uploads/ituevents_competition-rules-2010_2010-02-11.pdf) to decrease the effect of drag forces (McCole, Claney, Conte, Anderson, & Hagberg, 1990). Therefore, triathletes tend to seek improvements in their aerodynamic profile by reducing their frontal projected area (Moss, Cheryl, McKenzie, Ruby, & Heil, 2005). This is achieved by moving the saddle further forward relative to the bottom bracket compared to road cyclists (Ricard, et al., 2006) and moving the upper body further downwards to the handlebars (see Figure 2). Guidelines for bicycle configuration in the published literature have only focused on road cyclists. Given the different constraints on the triathlete and road racers it is likely that different configurations would be optimal for each event. To date, little attention has been giving to body position on the bicycle. Authors have suggested "optimal" or "appropriate" bicycle configuration (De Vey Mestdagh, 1998; Wanich, et al., 2007) but very little has been recommended in terms of optimal body position on the bicycle. Therefore it is important to ascertain differences in body position between road cyclists and triathletes.

The aim of the study was to evaluate differences in body position on the bicycle between competitive road cyclists and triathletes, and any differences between competitive and recreational road cyclists.

Methods

Participants

Seventy one male and female cyclists and triathletes volunteered to participate in the study (see Table 3.1). Thirty six recreational cyclists who commuted at least 30 km per week, 17 competitive road cyclists and 18 competitive triathletes who trained cycling at least 80 km per week and had competed for 2 years were included in the study.

Table 3.1. Characteristics (mean ±SD) of age, body mass, height, training hours per week and training volume per week of cyclists and triathletes (N = 71). Abbreviations used for effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

Groups	**Age (years)**	**Body mass (kg)**	**Height (cm)**	**Cycling time (hours/week)**	**Cycling volume (km/week)**
Recreational cyclists (n = 36)	41 ±9	79 ±13	178 ±7	7 ±3	73 ±57
Competitive cyclists (n = 17)	36 ±11	75 ±14	176 ±8	8 ±4	140 ±56
Competitive triathletes (n = 18)	38 ±11	77 ±24	169 ±32	6 ±2	93 ±50
Recreational cyclists vs. Competitive cyclists	12%; 0.4, S	5%; 0.3, S	1%; 0.2, T	12%; 0.3, S	***48%; 1.2, L***
Recreational cyclists vs. Competitive triathletes	7%; 0.3, S	2%; 0.1, T	6%; 0.4, S	21%; 0.4, S	22%; 0.4, S
Competitive cyclists vs. Competitive triathletes	4%; 0.1, T	2%; 0.1, T	5%; 0.4, S	37%; 0.7, M	49%; 0.9, M

Prior to the study, the participants were informed about possible risks and signed a consent form approved by the ethics committee of human research where the study was conducted.

Data collection

All participants attended one evaluation session where they provided information related to their cycling training history. Height and weight were measured according to International Society for Advancement of Kinanthropometry protocols (Marfell-Jones, Olds, Stewart, & Carter, 2006). Reflective markers attached to anatomical reference points allowed measurement of hip, knee and the ankle angles in the sagittal plane (see Figure 3.1a).

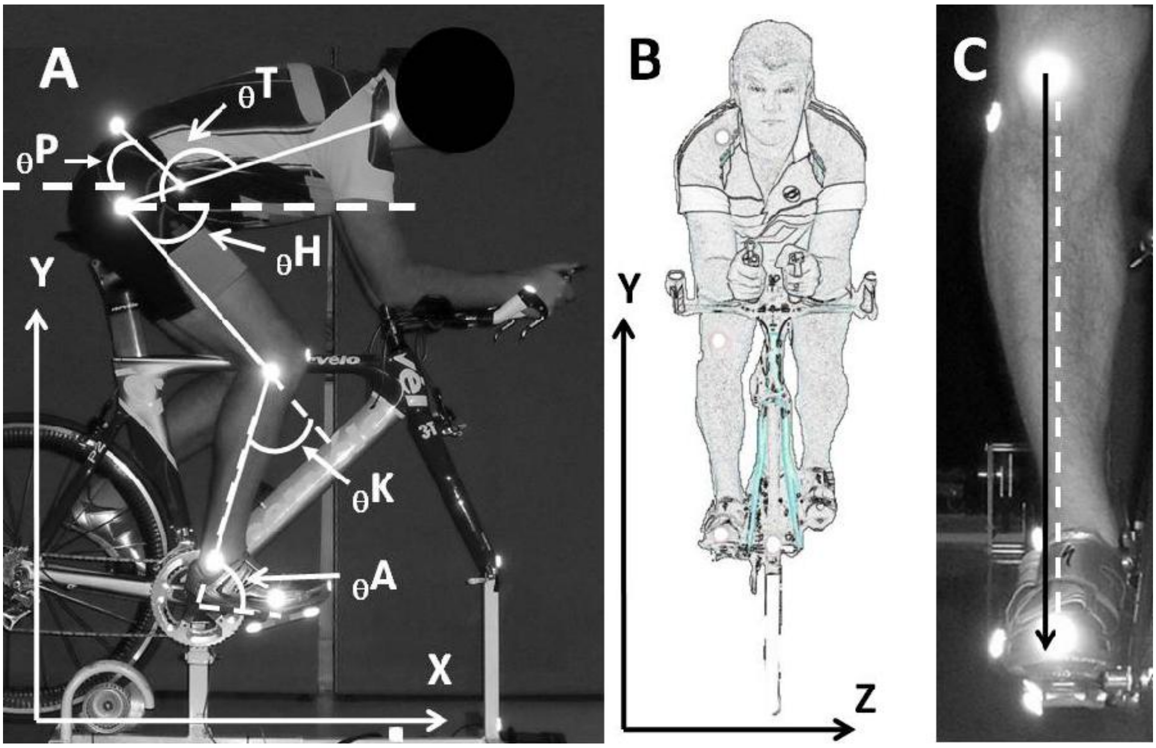


Figure 3.1. Representative photographs of the reflective marker locations. (A) In the sagittal plane: anterior superior iliac spine, posterior superior iliac spine, greater trochanter, lateral femoral epicondyle, and lateral malleolus, pedal (anterior and posterior aspects) and 5th metatarsal head. Definition of angles of the trunk, pelvis, hip, knee and ankle joints. (B) In the frontal plane: frontal projected area of one triathlete. (C) In the frontal plane: anterior surface of the patella and the most anterior surface of the shoe. Solid black line is vertical projection of the patella; dashed white line is vertical projection of the centre of the shoe. The horizontal difference between the two was used to measure medio-lateral position of the knee.

Cyclists had their own bicycles mounted on a cycle trainer (Kingcycle, Buckinghamshire, UK) and were asked to assume a position as similar as possible to outdoors cycling. A digital camera (Samsung ES15, Seoul, South Korea) recorded four high resolution images (3600 x 2400 pixels) from the sagittal plane with cyclists: 1) standing on the floor; 2) seated on the bicycle with the right crank in the most forward position (3 o'clock crank position); and 3) seated on the bicycle with the right crank in the lowest position on the crank cycle (6 o'clock crank position). The position of the hands on the handlebars was on the brake hoods (hands on the hoods close to the brakes) for recreational and competitive road cyclists. All triathletes laid their arms on the aerobars (see Figure 3.2). One image was also taken from the frontal plane with the right crank at the most forward position on the crank cycle (3 o'clock crank position).

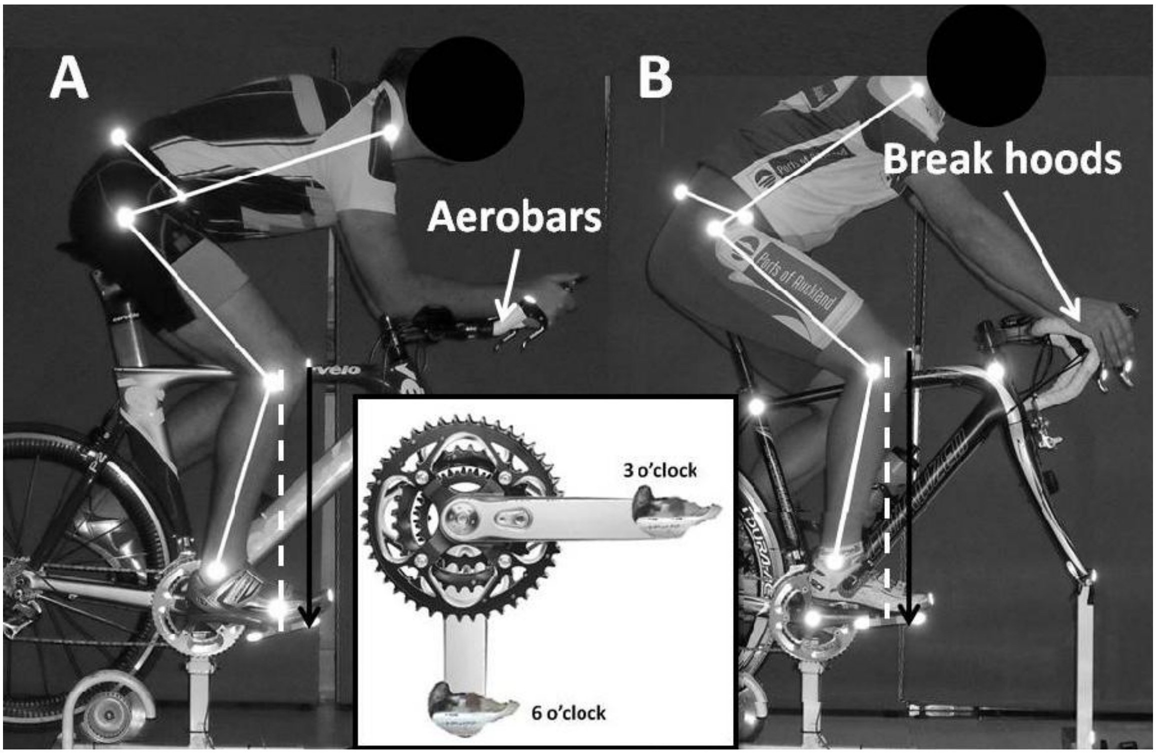


Figure 3.2. Representative photographs of a triathlete (A) and a road cyclist (B) in static poses at the 3 o'clock crank position. Indication of position of hands on the aerobars for triathletes and on the break hoods for road cyclists. Dashed line shows upward projection of the pedal axis and solid arrow shows projection on the knee over the pedal axis to illustrate anterior-posterior knee position.

Data analyses

Image files were digitized and markers were identified manually using ImageJ (National Institute of Health, USA) for x-y coordinates in the sagittal plane and z-y in the frontal plane. Images were scaled using two known distances within each image to convert measurements from pixels to metric units. From the image of the participants standing on the floor, the distance from the pedal axis to the top of the saddle when the crank was in line with the seat tube was defined as the saddle height (Bini, Hume, et al., 2011a). The lengths of the thigh and the shank were computed as the distance between the markers of the greater trochanter to the lateral femoral epicondyle and from the lateral femoral epicondyle to lateral malleolus, respectively. The lengths of the thigh, the shank and the distance from lateral malleolus to the floor were used to compute the lower limb length (Mellion, 1991).

Joint angles of the hip, knee, and ankle were calculated from the x-y coordinate data, as shown in Figure 2. Markers attached to the anterior superior iliac spine (ASIS) and the posterior superior iliac spine (PSIS) as reported by Bressel & Larson (2003) allowed measurement of pelvic and trunk angles in the sagittal plane at the 3 o'clock and 6 o'clock crank positions. From the image of the participant seated on the bicycle with the right pedal in the most forward position (3 o'clock crank position) the anterior-posterior position of the knee in relation to the pedal axis was measured to assess saddle anterior-posterior configuration. Markers attached to the anterior surface of the patella and on the anterior prominence of the cycling shoe in the frontal plane (see Figure 3.2b) (Bailey, et al., 2003) allowed measurement of knee medio-lateral position. From the image of the participants in the frontal plane, frontal projected area was manually computed using ImageJ by manual tracing and was normalized by body surface area (estimated from equations described by Du Bois & Du Bois (1916) to reduce effects of body dimensions on between-subjects comparisons. The rater's reliability in digitising was determined using images from static poses at the 3 o'clock and 6 o'clock crank positions of thirty randomly selected cyclists analysed seven days apart. Differences were trivial between days (<4.5%) for all variables.

Statistical analyses

Means and standard deviations were reported for saddle height relative to the lower limb length, saddle anterior-posterior configuration, knee medio-lateral position, trunk, pelvic, hip, knee and ankle angles at the 3 o'clock and 6 o'clock crank positions of the pedal and frontal area for the groups of cyclists. Normality of distribution and sphericity were evaluated via the Shapiro-Wilk and Mauchly tests respectively. Normality in distribution and sphericity were corrected for saddle height and knee flexion angle at the 3 o'clock crank position, a logarithmic transformation was applied.

Cohen's effect sizes (ES) were computed for the analysis of magnitudes of differences between groups (competitive cyclists and triathletes; recreational and competitive cyclists) and were rated as trivial (<0.25), small (0.25-0.5), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ensure non-overlap between means score greater than 55% (Cohen, 1988).

Results

Triathletes compared to competitive cyclists had greater anterior knee position and smaller frontal area. Competitive road cyclists did not substantially differ from recreational road cyclists for any variable related to body position on the bicycle. Competitive triathletes compared to recreational road cyclists had a larger pelvic angle at the 6 o'clock crank position and less frontal area (see Table 3.2).

Table 3.2. Means and standard deviations, percentage differences and magnitudes of effects for saddle height relative to lower limb length, saddle anterior-posterior configuration, knee medio-lateral position, trunk, pelvic, hip, knee and ankle angles at the 3 o'clock and 6 o'clock crank positions and frontal area comparing recreational cyclists, competitive cyclists and triathletes (N = 71). Abbreviations used are for competitive cyclists (Ccyc) and triathletes (Tri) and for effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

	Recreational cyclists n = 36	**Competitive cyclists n = 17**	**Competitive triathletes n = 18**
Standing position			
Saddle height (% of lower limb length)	96 ±3 Ccyc 2%; 0.7, M Tri <1%; 0.2, T	93 ±3 Tri 3%; 0.9, M	96 ±3
3 o'clock crank position			
Saddle anterior-posterior (cm)	2.6 ±3.1 Ccyc 33%; 0.2, T Tri 55%; 0.9, M	2.0 ±2.4 ***Tri 66%; 1.2, L***	5.9 ±4.1
Knee medio-lateral position (cm)	0.6 ±1.7 Ccyc 8%; 0.1, T Tri 57%; 0.5, M	0.7 ±1.0 Tri 53%; 0.6, M	1.4 ±1.5
Frontal area (% of body surface area)	21 ±2 Ccyc 2%; 0.2, T ***Tri 16%; 1.1, L***	21 ±1 ***Tri 17%; 1.3, L***	18 ±3
Trunk angle (°)	139 ±5 Ccyc 3%; 0.9, M ***Tri 12%; 3.2, L***	143 ±5 ***Tri 10%; 2.5, L***	158 ±7
Pelvic angle (°)	20 ±9 Ccyc 9%; 0.2, T ***Tri 30%; 1.2, L***	22 ±11 Tri 30%; 0.9, M	31 ±10
Hip angle (°)	39 ±4 Ccyc 1%; 0.1, T Tri 8%; 0.7, M	39 ±5 Tri 7%; 0.6, M	42 ±6
Knee angle (°)	62 ±6 Ccyc <1%; 0.1, T Tri <1%; 0.1, T	62 ±5 Tri 1%; 0.3, S	116±5
Ankle angle (°)	89 ±7 Ccyc <1%; 0.1, T Tri 4%; 0.4, S	86 ±7 Tri 3%; 0.3, S	85 ±7
6 o'clock crank position			

Trunk angle (°)	137 ±4 Ccyc 3%; 0.9, M ***Tri 9%; 3.4, L***	142 ±5 ***Tri 12%; 2.4, L***	156 ±6
Pelvic angle (°)	22 ±9 Ccyc 4%; 0.1, T ***Tri 34%; 1.1, L***	23 ±11 Tri 31%; 0.9, M	34 ±11
Hip angle (°)	68 ±4 Ccyc <1%; 0.1, T Tri 5%; 0.7, M	67 ±5 Tri 6%; 0.7, M	71 ±6
Knee angle (°)	30 ±7 Ccyc 2%; 0.1, T Tri 2%; 0.1, T	31 ±6 Tri 5%; 0.2, T	29 ±7
Ankle angle (°)	95 ±8 Ccyc <1%; 0.1, T Tri 3%; 0.4, S	96 ±6 Tri 3%; 0.4, S	93 ±6

Discussion

Existing guidelines for the configuration of bicycle components focus on road bicycle racing set-up (Burke & Pruitt, 2003; Silberman, et al., 2005) without information about triathlon bicycles or commuting road bicycles. The aim of our study was to compare body position on the bicycle of competitive road cyclists and triathletes and also recreational cyclists (commuters). Our results showed large differences in bicycle configuration and body position on the bicycle between competitive cyclists and competitive triathletes without substantial differences between recreational and competitive road cyclists. Triathletes presented lower frontal area, greater anterior projection of the knee, greater trunk and pelvic angles compared to recreational and competitive cyclists.

Triathletes are prohibited from riding behind another athlete during Ironman and other races, which would reduce drag forces (McCole, et al., 1990). Therefore, to reduce the power required to travel a set speed, triathletes seek to improve aerodynamic profile by reducing frontal projected area (Moss, et al., 2005). To reduce frontal area, triathletes have configured their bicycles to increase upper body flexion using aerobars to support their arms and hands and to ride with the knees close to the bicycle frame (Burke & Pruitt, 2003). Our triathletes presented 18-23% smaller frontal area compared to recreational and competitive cyclists which would be aligned with the 10-12% greater trunk flexion. Greater anterior projection of the knee in relation to the pedal axis (67-74%) for triathletes compared to recreational and competitive cyclists indicates a forward position of the body in relation to the bottom bracket for a similar saddle height configuration. Increasing the forward position of the body in relation to the bottom bracket can enhance running off the bike due to a greater hip flexion (as per defined in Figure 2.1) in triathlon bicycles (Garside & Doran, 2000). However, we did not observe substantial differences in hip flexion of triathletes compared to recreational cyclists and triathletes. For road cyclists, Price and Donne (1997) showed that a more anterior projection of the knee in relation to the pedal spindle may improve cycling efficiency. However, recreational and competitive cyclists from our study were in a "less efficient" horizontal projection of the body in relation to the bottom bracket.

We expected that competitive cyclists would adjust their bicycle components to elicit greater trunk flexion, thereby decreasing frontal area, to a greater extent than recreational cyclists who would opt for a more comfortable upright position. However, our data showed only moderate differences in trunk flexion (3%) for competitive cyclists compared to recreational cyclists resulting in only a small reduction in frontal area for competitive cyclists (2%). Trunk flexion angles may not be substantially changed because cyclists ride in groups in road cycling races which does not require the road cyclist to look for greater improvements in aerodynamic profile. However, time trial sections of multi-day races, when drafting is prohibited and cyclists are required to opt for a more aerodynamic position, should result in similar bicycle configuration and body positions to those observed for triathletes.

It has been suggested that the position of the saddle may play a role in the likelihood of overuse knee injuries (Holmes, et al., 1994). Our results of saddle height (~96% of lower limb length) and knee flexion angle for the 6 o'clock crank position (~30°) are in line with recommendations to prevent overuse knee injuries for cyclists and triathletes (Holmes, et al., 1994; Mellion, 1991). However, the projection of the knee in the sagittal plane has been recommended to lie over the pedal axis (Callaghan, 2005), which is contrary to what we observed, especially for triathletes. A greater projection of the knee (~6 cm) was found for the triathletes in our study, which suggests that this group may have a greater risk of developing overuse knee injuries compared to recreational and competitive cyclists. However, prevalence of knee injuries in triathletes (14% to 63%) (Gosling, et al., 2008) and cyclists (21% to 65%) are similar (Dettori & Norvell, 2006), which does not support the hypothesis of greater injury risk in triathletes using a more forward body position on the bicycle. Screening of cyclists and triathletes in terms of bicycle configuration and body positions, and then longitudinal tracking of their training and injury histories, is required before any clear associations can be ascertained.

A limitation of our study was the use of images of cyclists and triathletes taken in static poses, which may not be the same as images taken from video of cycling. Further research should compare joint kinematics of cyclists and triathletes during dynamic assessments. Future research with larger sample sizes may examine the effects of gender and joint range of motion and their effects on body position on the bicycle.

Conclusion

Competitive triathletes had greater body forward projection (greater trunk flexion and knee anterior position) than competitive road cyclists. Both recreational and competitive cyclists sat on their bicycles with their trunks in a more vertical position compared to triathletes. Guidelines for bicycle configuration for triathletes and road cyclists need to consider the body positions during events.

CHAPTER 4: A COMPARISON OF STATIC AND DYNAMIC MEASURES OF LOWER LIMB JOINT ANGLES IN CYCLING: APPLICATION TO BICYCLE FITTING

Overview

Configuration of bicycle components to the cyclist (bicycle fitting) commonly uses static poses of the cyclist on the bicycle at the 6 o'clock crank position to represent dynamic cycling positions. However, the validity of this approach has not been reported. Therefore, this study compared lower limb joint angles of cyclists in static poses compared to dynamic cycling. Using a digital camera, right sagittal images were taken of thirty cyclists seated on their own bicycles mounted on a stationary trainer with the crank at 3 o'clock and 6 o'clock positions. Video was then recorded during pedalling at a self-selected gear ratio and pedalling cadence. Sagittal plane hip, knee and ankle angles were digitised. Differences between static and dynamic angles were large at the 6 o'clock crank position with greater mean hip angle (4.9 ±3°), smaller knee angle (8.2 ±5°) and smaller ankle angle (8.2 ±5.3°) for static angles. Differences between static and dynamic angles (<1.4°) were trivial to small for the 3 o'clock crank position. To perform bicycle fitting, joint angles should be measured dynamically or with the cyclist in a static pose at the 3 o'clock crank position.

Introduction

Configuration of bicycle components to the cyclist (bicycle fitting) has been conducted using tape measures and plumb bobs (Burke & Pruitt, 2003) with the dimensions of bicycle components related to anthropometric dimensions of the cyclist (Christiaans & Bremner, 1998; Laios & Giannatsis, 2010). Photogrammetry methods have been applied to estimate body dimensions (Gavan, Washburn, & Lewis, 1952) with application to various research areas (Aksu, Kaya, & Kocadereli, 2010; Belli, Chaves, De Oliveira, & Grossi, 2009; Gruen, 1997). The frontal projected area of cyclists has been accurately estimated using digital photogrammetry compared to a weighting pictures method (Olds & Olive, 1999). For body segment lengths, Mellow et al. (2003) showed low technical error of manual measurement for digital photogrammetry (0.31%), which suggests that this technique may be valid to measure body segment dimensions in a photographic image. For the configuration of bicycle components, joint angles have been preferably recommended in comparison to anthropometric references (Peveler, et al., 2005). The reason is that length based references for saddle height configuration does not take into account particular differences in thigh, shank and foot length. The effectiveness of the "optimum" relationship between bicycle components and body dimensions have failed to result in similar body positions because joint angles have not been taken into account (Peveler, et al., 2005).

In cycling the use of video analysis to optimize the configuration of bicycle components is increasing. The decreasing price of regular and high speed cameras has enabled bicycle shops and clinics to afford video analysis systems. Free software packages are available to compute angles and distances in images. However, when using joint angles as a reference for bicycle configuration optimization, all guidelines are based on measurements of the cyclist in static poses. Burke and Pruitt (2003) suggested that knee flexion angle should be between 25-30° when the pedal is static at the bottom of the crank cycle (6 o'clock crank position) for an optimum saddle height configuration. Yet, it is unknown if cyclists reproduce the same joint angles dynamically as measured statically, which may compromise the use of existing guidelines.

Given a comparison between static and dynamic analyses of joint angles for optimization of bicycle components has not been reported in the literature, the aim of this study was to compare lower limb joint angles of cyclists in static postures compared to dynamic cycling. The hypothesis was that cyclists would replicate similar joint angles in static and dynamic cycling.

Methods

Participants

Thirty cyclists with experience ranging from recreational to competitive volunteered to participate in the study. The characteristics of the cyclists were (mean ±SD) 39 ±10 years old, 80 ±15 kg body mass, 177 ±8 cm height, 7.3 ±3.8 hours/week cycle training, and 8 ±7 years cycle experience. Prior to the study participants were informed about possible risks and signed a consent form approved by the Ethics Committee of Human Research where the study was conducted.

Data collection

All cyclists attended one evaluation session where anthropometric measures, images from static postures (photogrammetry) and dynamic cycling from video (videogrammetry) from their right sagittal plane were collected. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of the cyclists at the greater trochanter, lateral femoral condyle, and lateral malleolus (see Figure 4.1). Two markers were attached to the pedal to compute the pedal axis. Two markers were taped at a known distance on the bicycle frame to calibrate image coordinates in metric units.

Cyclists had their own bicycles mounted on a wind trainer (Kingcycle, Buckinghamshire, UK) and were asked to assume a position as similar as possible to outdoors cycling. A digital camera (Samsung ES15, Seoul, South Corea) recorded three high resolution images (3600 x 2400 pixels of resolution) from the sagittal plane with the cyclists standing on the floor (calibration image), cyclists seated on the bicycle with the right crank in the most forward position (3 o'clock) and the right crank in the lowest position on the crank cycle (6 o'clock). One image was recorded at each position to simulate common procedures used in bicycle fitting configuration when a cyclist's knee flexion angle is measured using a manual goniometer (Burke & Pruitt, 2003; Peveler, et al., 2005; Silberman, et al., 2005). Cyclists were then asked to select a gear ratio and assume pedalling cadence as similar as possible to steady state cruising road cycling for five minutes. After three minutes of riding, video was recorded for 20 s using a digital camera (30 Hz, 640 x 480 pixels of resolution) which was shown to provide reliable measurements of rearfoot timing variables (e.g. time of maximal eversion) during running in a previous study (Ferber, Sheerin, & Kendall, 2009).

Data analyses

Hip, knee and ankle joint angles were manually digitized from the static postures and video files using ImageJ (National Institute of Health, USA) by the same rater for the 30 cyclists. For dynamic cycling, five consecutive frames where the cyclists were at the 3 o'clock and at the 6 o'clock crank positions were visually selected to compute joint angles. The average of five revolutions of each joint angle was used for comparison with static poses. The rater's reliability in digitising was determined using images from static poses analysed on day one and day seven. Differences were trivial between days (<4.5%) for hip, knee and ankle angles.

Statistical analyses

Cyclists' means and standard deviations were reported for both static and dynamic hip, knee and ankle angles. To compare static and dynamic angles a multivariate analysis of variance was employed including the method of analysis (static and dynamic) as a fixed effect. SPSS for Windows 16.0 was employed for the analysis of Type I and II errors (Knudson, 2009). Cohen's effect sizes (ES) were computed for the analysis of magnitudes of the differences between the two methods and were rated as trivial (<0.25), small (0.25-0.5), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004).

Results

The multivariate model indicated that the method of analysis (static versus dynamic) of hip, knee and ankle joint angles significantly affected the dependent variables ($p<0.01$, observed power = 1.0). Differences between static and dynamic angles were large at the 6 o'clock crank position with greater hip angle (4.9 ±3°), smaller knee angle (8.2 ±5°) and smaller ankle angle (8.9 ±5°) for static angles. Differences between static and dynamic angles (<1.4°) were trivial to small for the 3 o'clock crank position (see Table 4.1 and Figure 4.1).

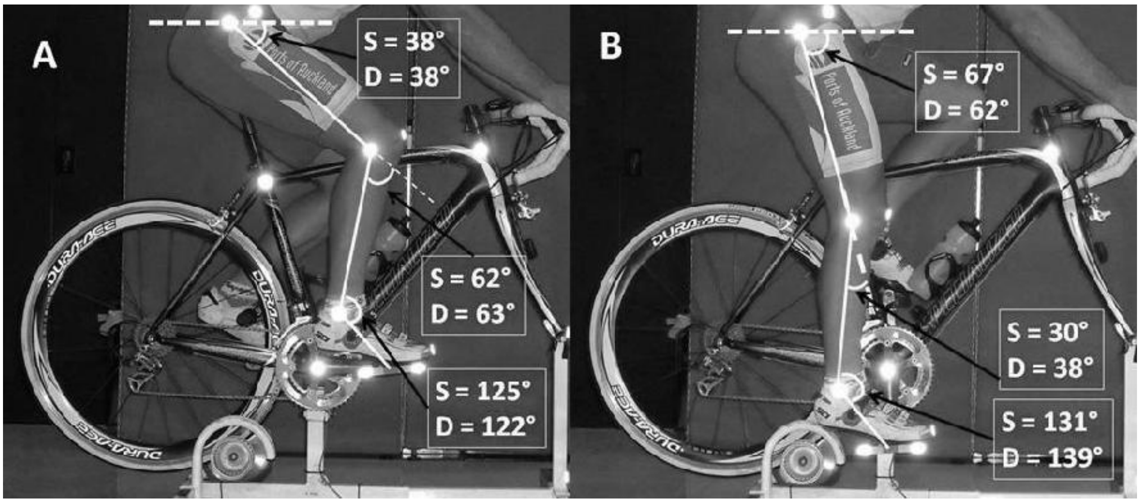


Figure 4.1. Illustration of reflective marker placement on the right side of the cyclist at the greater trochanter, lateral femoral condyle and lateral malleolus to indicate hip, knee and ankle joint angles. Markers attached to the pedal were used to compute the pedal axis for ankle joint measurement. Mean hip, knee and ankle joint angles are shown for the 30 cyclists for static (S) and dynamic (D) measurements at the 3 o'clock (A) and 6 o'clock (B) crank positions.

Table 4.1. Hip, knee and ankle angles (mean ±SD) at the 3 o'clock and 6 o'clock crank positions for 30 cyclists. Comparison of the angles determined by static and dynamic methods using univariate models (p value and effect sizes – ES).

			Difference between static and dynamic			
	Static angle (degrees)	**Dynamic angle (degrees)**	**Degrees**	**p value**	**ES**	**ES – magnitude inference**
3 o'clock crank position						
Hip angle	38 ±3.7	38 ±2.9	0.2 ±3	0.76	0.1	Trivial
Knee angle	62 ±4.6	63 ±4.1	1.1 ±4	0.32	0.3	Trivial
Ankle angle	125 ±6.7	122 ±6.1	2.4 ±7	0.12	0.4	Small
6 o'clock crank position						
Hip angle	67 ±4.9	62 ±3.9	4.9 ±3	<0.01	1.1	Large
Knee angle	30 ±6.7	38 ±4.3	8.2 ±5	<0.01	1.5	Large
Ankle angle	131 ±5.8	139 ±6.7	8.2 ±5	<0.01	1.4	Large

Discussion

At bicycle shops, clinics and in bicycle research, configuration of bicycle components to the cyclist (bicycle fitting) takes into account lower limb joint angles determined from a static position of cyclists at the 6 o'clock crank position measured once (Burke & Pruitt, 2003; Silberman, et al., 2005). However, cycling is a dynamic movement, so bicycle configuration should ideally be based on dynamic assessment looking at the average of consecutive pedal revolutions. Our study is the first to report differences between lower limb joint angles gathered from cyclists in static postures compared to dynamic cycling. Cyclists in our study were not able to replicate similar angles in static postures as those observed in video analysis when the crank was at the 6 o'clock crank position. Given the static 6 o'clock crank angle method is commonly used in bicycle shops and clinics, information needs to be provided on why the 3 o'clock position would be a better method to set-up a cyclist on a bicycle if dynamic cycling angles were not available.

The measurement of joint angles in images of cyclists on their own bicycles has the potential to improve the existing techniques for bicycle configuration components optimization (Peveler, et al., 2005). Joint angles are important variables for the configuration of bicycle components to help reduce injury risk and optimize performance (Bini, Hume, et al., 2011a; Peveler, 2008) but the assessment of joint angles of cyclists may depend on exercise conditions. Previous studies presented the dependence of joint angle on workload level (Black, Sanderson, & Hennig, 1993), pedalling cadence (Bini, Rossato, et al., 2010), fatigue state (Bini, Diefenthaeler, & Mota, 2010) and experience in cycling (Chapman, et al., 2009). Therefore, these factors should ideally be taken into account when providing a bicycle set-up.

Farrell et al. (2003) reported that configuring saddle height to elicit 25-30° of knee flexion using a goniometer with the cyclist in a static pose at the 6 o'clock crank position resulted in 30-45° knee flexion at the same 6 o'clock crank position in video analysis. The larger knee flexion angles (~10°) in dynamic cycling reported by Farrell using the goniometer method were also evident in our study (8.2 ±5°) using the digitisation of the static pose to determine knee flexion angle at the 6 o'clock crank position. Therefore, measurement of knee flexion angle of the cyclist using images of static poses or using the goniometer method to obtain static knee angles at the 6 o'clock crank position should be avoided.

Greater hip angle (smaller flexion), smaller knee angle (smaller flexion) and smaller ankle angle (greater flexion) were observed in static poses at the 6 o'clock crank position compared to dynamic assessment in our study. Looking at the main driving muscles of cycling (hip and knee joint extensors and ankle plantar flexors), hip and knee joint extensors may be shorter and ankle plantar flexors may be longer in the static pose at the 6 o'clock crank position compared to during dynamic cycling. These differences may affect muscle tendon-unit length and force production (Sanderson & Amoroso, 2009; Sanderson, Martin, Honeyman, & Keefer, 2006).

Differences in joint angles between static and dynamic analysis may be related to the lack of angular momentum at the 6 o'clock crank position during static poses, which is contrary to what is observed during dynamic cycling. For pedalling at 90 rpm, cyclists usually present ~27% greater angular velocity of the crank at the 12 o'clock and 6 o'clock crank positions compared to the average angular velocity of the revolution (Hull, Kautz, & Beard, 1991). Two reasons may explain the similarities of the static and dynamic joint angles at the 3 o'clock crank position: 1) There is ~28% lower angular velocity in dynamic cycling at the 3 o'clock crank position than the average angular velocity over the entire revolution of the crank (Hull, et al., 1991); and 2) To sustain the cranks horizontally at the 3 o'clock crank position, cyclists need to balance the mass of the ipsilateral and contralateral legs.

In terms of saddle height adjustment, a range of 25-30° of knee flexion has been recommended to improve efficiency and reduce the risk of injuries in cyclists (Bini, Hume, et al., 2011a). Reductions of ~8° may be expected for the knee flexion angle of cyclists assessed statically at the 6 'clock position in comparison to dynamic assessment.

The choice of using the same camera to acquire video and capture images of the cyclists in static poses had positive and negative effects in our study. One benefit was that there was no effect from different lenses on image distortion. However, the camera used in the present study was not capable of recording images and video at the same resolution (which would be similar to cameras used by bicycle shops providing bicycle configuration services). Video images had ~19% of the resolution of the static images, which may have reduced the precision of tracking markers in video images compared to images from static poses. However, the choice for analysis of mean results of joint angles over five crank revolutions would have increased the accuracy of crank angle determination for joint angle computation.

One source of error using sagittal plane video may be out of plane movements. In cycling, most movement can be assessed via sagittal plane analysis, but up to 10% of differences may be expected for the hip angle when measuring from the sagittal plane compared to 3D analysis (Umberger & Martin, 2001). Pelvic motion during cycling may also affect the comparison of images from static and dynamic analysis. Horizontal (±5 cm) and vertical (±2 cm) motion of the hip joint occurs during stationary cycling (Neptune & Hull, 1996) which may affect lower limb joint angles especially for cyclists using a higher saddle height. Therefore, bicycle set-up should ideally use images from both sagittal and frontal planes or 3D analyses.

Conclusion

Cyclists were not able to replicate in a static pose at the 6 o'clock crank position similar hip, knee and ankle joint angles as measured in dynamic cycling. To perform configuration of bicycle components using joint angles, measurements should be taken dynamically or with the cyclists in static poses at the 3 o'clock crank position, instead of the usually recommended 6 o'clock crank position.

CHAPTER 5: PEDAL FORCE EFFECTIVENESS IN CYCLING: A REVIEW OF CONSTRAINTS AND TRAINING EFFECTS

Overview

Pedal force effectiveness in cycling is usually measured by the ratio of force perpendicular to the crank (effective force) and total force applied to the pedal (resultant force). Most studies measuring pedal forces have been restricted to one leg but a few studies have reported bilateral asymmetry in pedal forces. Pedal force effectiveness is increased at higher power output and reduced at higher pedaling cadences. Changes in saddle position resulted in unclear effects in pedal force effectiveness, while lowering the upper body reduced pedal force effectiveness. Cycling experience and fatigue had unclear effects on pedal force effectiveness. Augmented feedback of pedal forces can improve pedal force effectiveness within a training session and after multiple sessions for cyclists and non-cyclists. No differences in pedal force effectiveness were evident between summarized and instantaneous feedback. Conversely, economy/efficiency seems to be reduced when cyclists are instructed to improve pedal force effectiveness during acute intervention studies involving one session. Decoupled crank systems effectively improved pedal force effectiveness with conflicting effects on economy/efficiency and performance.

Introduction

During cycling, lower limb movement in the sagital plane is constrained to a circular path by the geometry of the bicycle (i.e. cranks and pedals). Within these constraints the cyclist can vary pedaling technique by changing the kinematics of the three lower limb segments (thigh, shank and foot) and the activation of the lower limb muscles. Technique in cycling can be assessed through measurement of joint kinematics (Bini, Diefenthaeler, et al., 2010; Chapman, Vicenzino, Blanch, Dowlan, & Hodges, 2008; Hasson, Caldwell, & van Emmerik, 2008) and muscle activation patterns (Bini, Carpes, Diefenthaeler, Mota, & Guimarães, 2008; Candotti, et al., 2009; Dorel, Drouet, et al., 2009). Alternatively, pedal force effectiveness (ratio of the force perpendicular to the crank and the total force applied to the pedal) has also been used as a gold standard measure of technique in cycling (Dorel, Couturier, et al., 2009; Dorel, Drouet, et al., 2009; Rossato, Bini, Carpes, Diefenthaeler, & Moro, 2008). However, there has been criticism recently regarding using pedal force effectiveness exclusively for feedback as pedal force effectiveness may not provide a full representation of pedaling technique of cyclists (Bini & Diefenthaeler, 2010). Pedaling technique is probably too complex to be summarized by force effectiveness alone given that technique strategies may not be fully translated into better force effectiveness. However, cyclists can improve power output if they improve force effectiveness, but they cannot improve power output exclusively by improvements in pedaling technique. For a similar pedaling technique (e.g. focus on pushing down forces applied at the downstroke phase) power output can be improved by increasing magnitude of force application (assuming similar directions of the force). However, changing technique to a more circling action (i.e. greater force effectiveness for similar magnitude of forces) power output can be improved, but only because force effectiveness is improved.

Existing evidence is conflicting regarding the relationship between pedal force effectiveness and performance in cycling. Most research suggests that when the effectiveness of the force applied on the pedal is optimized, the economy/efficiency (i.e. ratio between mechanical energy produced and physiological energy demand) is reduced (Korff, et al., 2007; Mornieux, Stapelfeldt, Collhofer, & Belli, 2008). No research has been conducted to quantify the relationship between symmetry in pedal forces and performance. We chose to review the use of pedal force effectiveness during cycling as pedal force systems are now becoming commercially available for monitoring cycling training and performance. Therefore, it is important to analyze what we know and what we still need to learn in terms of pedal force effectiveness to better advise cyclists and coaches.

The purpose of this review was to summarize current knowledge on pedal force effectiveness during cycling and how it is affected by task constraints such as workload, pedaling cadence, body position, fatigue and cycling ability. Limitations and benefits of measuring and using pedal force effectiveness feedback exclusively are discussed throughout the article. Interventions to improve force effectiveness and cycling performance are also considered to identify interactions between technique training and performance.

Methods

Academic databases (MEDLINE, SCOPUS, ISI Web of Knowledge, EBSCO, and GOOGLE SCHOLAR) were searched for peer-review journals, books, theses, and conference proceedings since 1960 with the keywords pedal force effectiveness, workload, pedaling cadence, saddle position, cycling, fatigue, and symmetry. Articles were not included when they could not be retrieved without at least an English abstract. Journal articles (74), book chapters (5), and conference articles (14) were included in this review based on exclusion criteria of articles that were not related to pedal force measurements.

Results

Most studies measuring pedal forces have been restricted to one leg but a few studies have reported bilateral asymmetry in pedal forces. Pedal force effectiveness is increased at higher workload level and reduced at higher pedaling cadences. Changes in saddle position resulted in unclear effects in pedal force effectiveness, while lowering the upper body reduced pedal force effectiveness. Cycling experience and fatigue had unclear effects on pedal force effectiveness. Augmented feedback of pedal forces can improve pedal force effectiveness within a single training session and after multiple sessions for cyclists and non-cyclists. No differences in pedal force effectiveness were evident between summarized and instantaneous feedback. Conversely, economy/efficiency seems to be reduced when cyclists are instructed to improve pedal force effectiveness during acute intervention studies involving one session (Korff, et al., 2007; Mornieux, et al., 2008). Decoupled crank systems effectively improved pedal force effectiveness with conflicting effects on economy/efficiency and performance.

Discussion

Measuring pedal forces

Over the last 30 years technology has allowed pedal force measurement to advance to the stage where it is now possible to measure three components of force (Fx, Fy and Fz) and three associated moments (Mx, My and Mz) (Hull & Davis, 1981). Systems have been used during cycling on the road (Álvarez & Vinyolas, 1996; Dorel, et al., 2008) and off-road (Rowe, Hull, & Wang, 1998). A summary of the systems used to measure forces applied during cycling is provided in Table 5.1.

Table 5.1. Scientific papers reporting different systems to measure the force applied on the pedals during cycling.

Reference	Sensor type	Force components and moments	Cleats type	Application
Guye (1896)	Pressure[A]	Normal (Fz)	No cleats	Unknown[B]
Sharp (1896)	Pressure[A]	Normal (Fz)	No cleats	Laboratory
Hoes et al. (1968)	Strain gauge	Normal (Fz)	Toe clips	Laboratory
Sargeant & Davies (1977)	Strain gauge	Normal (Fz)	Toe clips	Laboratory
Dal Monte et al. (1973)	Strain gauge	Normal (Fz) and anterior-posterior (Fx)	Toe clips	Laboratory
Hull & Davis, (1981)	Strain gauge	Normal (Fz), anterior-posterior (Fx) and medio-lateral (Fy), and related moments	Toe clips	Laboratory
Harman et al. (1987)	Strain gauge	Normal (Fz) and anterior-posterior (Fx)	Unknown[B]	Laboratory
Newmiller et al. (1988)	Strain gauge	Normal (Fz) and anterior-posterior (Fx)	Clip in	Laboratory

Broker & Gregor (1990)	Piezoelectric	Normal (Fz), anterior-posterior (Fx) and medio-lateral (Fy), and related moments	Toe clips and Clip in	Laboratory
Álvarez & Vinyolas (1996)	Strain gauge	Normal (Fz) and anterior-posterior (Fx)	Clip in	Road
Boyd et al. (1996)	Strain gauge	Normal (Fz), anterior-posterior (Fx) and medio-lateral (Fy), and related moments	Clip in	Laboratory
Nabinger et al. (2002)	Strain gauge	Normal (Fz), anterior-posterior (Fx) and medio-lateral (Fy), and related moments	Clip in	Laboratory
Reiser Ii et al. (2003)	Strain gauge	Normal (Fz) and anterior-posterior (Fx)	Clip in	Laboratory
Chen et al. (2005)	Load cell	Normal (Fz) and anterior-posterior (Fx)	Unknown[B]	Road
Mornieux et al. (2006)	Cycle ergometer mounted on a force plate	Normal (Fz), anterior-posterior (Fx) and medio-lateral (Fy), and related moments	Toe clips	Laboratory
Stapelfeldt et al. (2007)	Hall effect sensor	Normal (Fz) and anterior-posterior (Fx)	Selectable	Laboratory
Valencia et al. (2007)	Piezoresistive force sensor attached to the pedal	Normal (Fz) and anterior-posterior (Fx)	Selectable	Laboratory
Dorel et al. (2008)	Strain gauge	Normal (Fz) and anterior-posterior (Fx)	Clip in and toe clips	Track
Chunfu (2009)	Strain gauge	Normal (Fz)	N.A[C]	N.A[C]

[A] No details about the measurement system characteristics.

[B] No details about pedal-shoe interface characteristics.

[C] The system was only analyzed using theoretical loads (finite elements).

The component of the force applied on the pedal in the frontal plane (medio-lateral) is presented in Figure 5.1.

The medio-lateral component (Fz), does not contribute to bicycle propulsion and is usually ignored despite suggestions that inter-segmental forces at the knee joint may be associated with injury (Ericson, et al., 1984; Ruby, et al., 1992).

The total force applied on the pedal in the sagital plane can be computed by the two components on the pedal surface (normal - Fy and anterior-posterior - Fx). A percentage of the total force on the pedal will be directed perpendicular to the crank (effective force). To compute the effective force, pedal angle in relation to the crank has been acquired using angular potentiometers (Hull & Davis, 1981), videography (Rossato, et al., 2008) or digital encoders (Martin & Brown, 2009). By trigonometry, normal and anterior-posterior forces were converted into components tangential to the crank. Effective force can produce propulsive or retarding force on the crank depending on the direction of the force applied on the pedal during the crank revolution (see Figure 5.2 that illustrates forces applied on the pedal during cycling in the sagital plane).

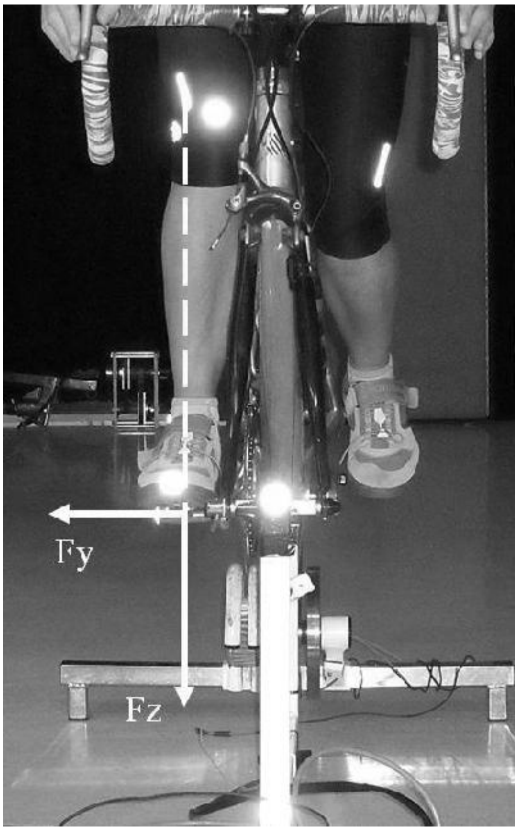


Figure 5.1. Frontal view image of one cyclist illustrating the normal and medio-lateral components of the force applied on the pedal. Dotted arrow shows the projection of the pedal in the frontal plane and highlights the medial-displacement of the knee.

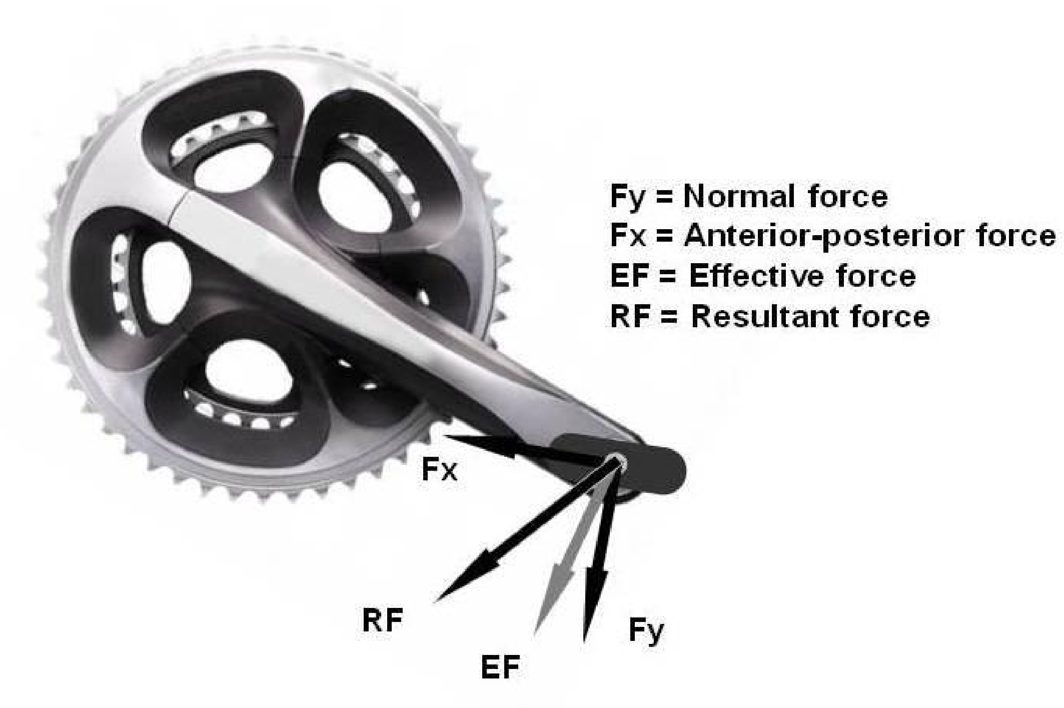


Figure 5.2. Diagram of the normal (Fz) and anterior-posterior (Fx) components of the total force applied on the pedal (resultant force – RF) in the sagital plane. The effective component (EF) of the resultant force applied on the sagital plane is also shown.

Pedal force effectiveness during cycling has been defined as the ratio of the force perpendicular to the crank (effective force) and the total force applied to the pedal (resultant force). This ratio has been defined as the index of effectiveness, which is the ratio of the impulse of the effective force to the impulse of the resultant force over a complete crank revolution (see equation 5.1) (LaFortune & Cavanagh, 1983a).

$$IE = \int_{0}^{360} EFdt \Big/ \int_{0}^{360} RFdt$$

Equation 5.1. Index of effectiveness (IE) is the ratio of the angular impulse of the effective force (EF) to the linear impulse of the resultant force (RF) over a complete crank revolution (LaFortune & Cavanagh, 1983a).

The index of effectiveness is the most used measure of technique in cycling because more skilled cyclists have a higher pedal force effectiveness (Bohm, Siebert, & Walsh, 2008; Hasson, et al., 2008; Holderbaum, et al., 2007). However, recent studies have reported that pedal force effectiveness may not fully represent joint kinetic and kinematic patterns associated with changes in pedaling technique (Bini & Diefenthaeler, 2010; Korff, et al., 2007; Mornieux, et al., 2008). The reason is that cyclists change joint kinetics and kinematics towards an improved technique (e.g. greater knee and hip joint flexor moments at the upward phase) but they do not necessarily convert these greater moments into better force effectiveness (Bini & Diefenthaeler, 2010).

The pedal cycle is usually divided into two phases (propulsive or downward and recovery or upward) or four quarters. Average normal and anterior-posterior forces from one male competitive cyclist (20 years old, 375 W of maximal aerobic power output and 65 $ml \cdot kg^{-1} \cdot min^{-1}$ of VO_{2Max}) and hypothetical ideal force application are presented in Figure 5.3. The ideal force direction is based on the assumption that all the force applied to the pedal should be converted into effective force in favor of crank motion.

Radial forces at the bottom (or top) dead centres of crank revolution, which are commonly observed in cyclists, do not create angular motion, therefore, do not help on crank motion. Inertial moment from the cyclist's leg may result in angular motion. Although related to inertial components of leg segments, the radial force applied on the pedal is not free of energy cost because energy is spent to convert potential energy at the top dead centre to kinetic energy towards the bottom dead centre. If the cyclist is riding with no resistance on the bicycle wheel, energy is still required to keep pedalling, i.e., there is an internal work problem (Minetti, 2011). Potential energy is stored in the muscles at the top and bottom dead centres and is converted to kinetic energy at the upstroke and downstroke phases. Changing the motion of the limb from downward to upward does not require energy from the ipsilateral leg. However, the connection with the contralateral leg (which will spend energy lifting the other leg) and the inertial effect (or potential to kinetic energy conversion) will create angular motion at the bottom dead centre transition. The reason for the extra metabolic energy to reduce radial forces and increase tangential forces on the crank is likely to depend on the recruitment of other muscles (i.e. knee and hip flexors) that are not normally used by cyclists for this particular task (Mornieux, et al., 2010).

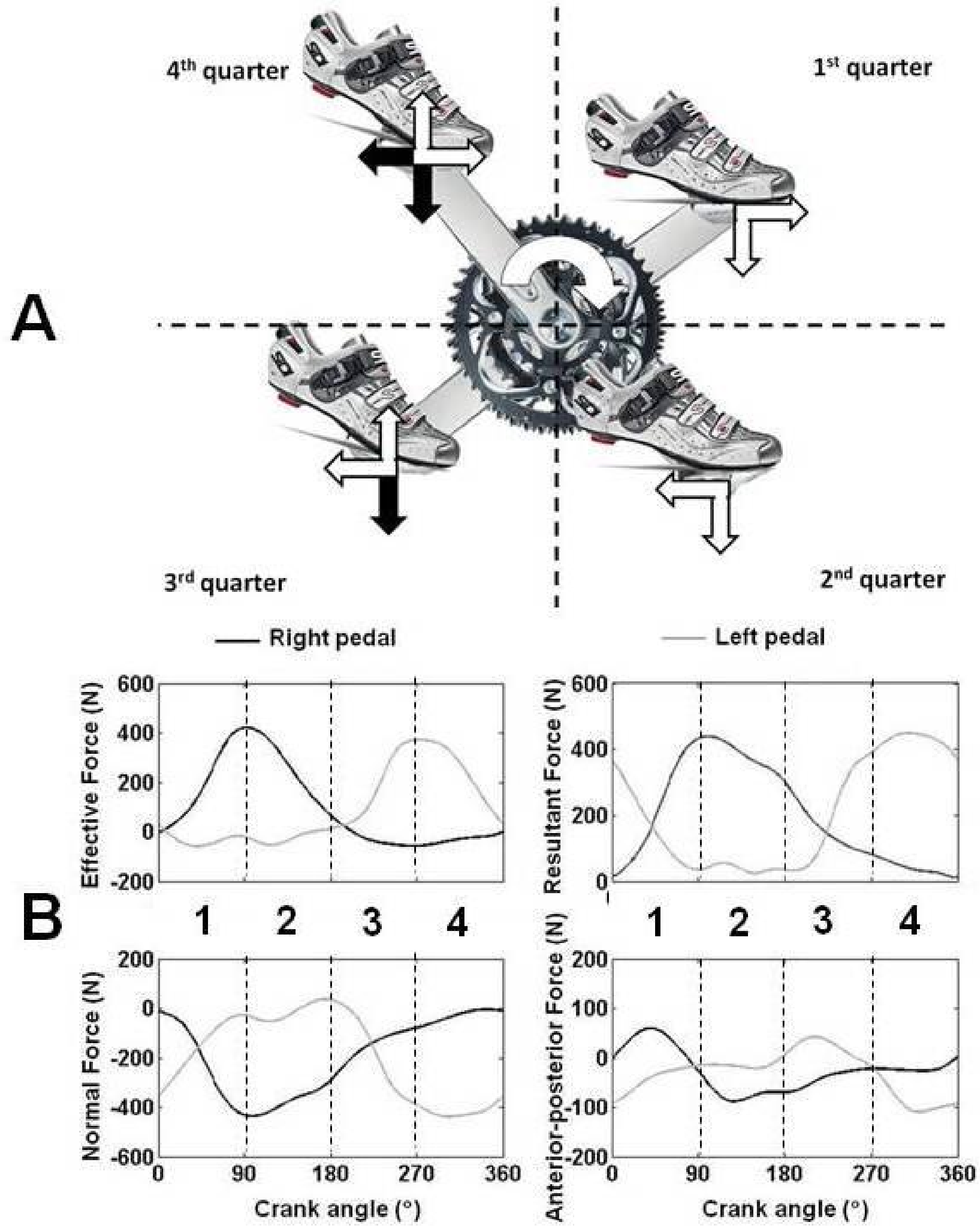


Figure 5.3. Representative diagram of pedal force directions at the four quarters of a pedal revolution. White arrows indicate ideal pedal force application to optimize force effectiveness and black arrows show normal and anterior-posterior pedal force application for one male competitive cyclist riding at 90 rpm and 350 W. Plots of right (black line) and left (grey line) normal and anterior-posterior force of one male competitive cyclist riding at 90 rpm and 350 W. Right and left effective (EF), resultant (RF), normal (Fy) and anterior-posterior (Fx) forces. For effective force, positive values indicate propulsive effective force. For normal force, positive values indicate force applied to pull the pedal, and for anterior-posterior force, positive values indicate a forward force applied to the pedal.

Pedal force application from the example cyclist was different from the hypothetical ideal force application presented in Figure 5.3. For normal force, propulsion is maximized by applying a downward force during the propulsive phase (from the top dead centre to bottom dead centre) and an upward force during the recovery phase (from the bottom dead centre to the top dead centre). Similarly for the anterior-posterior force, propulsion is maximized with anterior force during the first and the fourth quarters, and posterior force during the second and the third quarters. However, these idealized force profiles are not observed in cyclists.

In the propulsive phase the resultant force is consistent between cyclists (variance ratio = 0.063 [CV% = 10%]) but in the recovery phase normal force is more variable between cyclists (variance ratio = 0.204; CV% = 31%) (Hug, Drouet, Champoux, Couturier, & Dorel, 2008), possibly because some cyclists try to pull the pedal upward to improve force effectiveness (Mornieux, et al., 2008). Upward pulling of the pedal is possible during the recovery phase because most cyclists use a system (clipless or clip in) where the shoe is attached to the pedal by a cleat. Differences in anterior-posterior force between cyclists predominantly occur during the recovery phase, when some cyclists try to pull the pedal backwards (Coyle, et al., 1991; Kautz, et al., 1991).

An example of effective, resultant, normal and anterior-posterior forces on the pedal from the example cyclist are shown in Figure 5.3. Each plot shows the average forces of the right and left pedals from ten consecutive revolutions when the cyclist was riding at 350 W.

Using the right pedal as a reference, during the first quarter of the crank revolution the majority of normal force acted to push the right pedal downward and anterior-posterior force to push the right pedal forward. During the second quarter, the normal force still remained downward while the anterior-posterior force changed to a backward direction. A lower level of downward normal force was found during the third and fourth quarters and the anterior-posterior force remained negative (pulling backward). Comparing the force profiles of this cyclist to the ideal force profile presented in Figure 3 indicated that they were unable to apply a pulling normal force to the pedal during the recovery phase or a substantial anterior shear force during the fourth quarter of the crank revolution. Similar force patterns have been reported in other studies (Caldwell, Li, McCole, & Hagberg, 1998; Coyle, et al., 1991; Hull & Davis, 1981; Kautz, et al., 1991).

Most of the effective force is produced during the propulsive phase with the highest force generated at approximately 90° (Coyle, et al., 1991). Propulsive effective force is rarely observed during the recovery phase and most studies reported negative effective force during the recovery phase (Dorel, Drouet, et al., 2009; Rossato, et al., 2008; Sanderson & Black, 2003) which indicates that the effective component of pedal force is in the opposite direction to the crank movement, thereby resulting in resistive force for the contralateral leg (Coyle, et al., 1991). This resistive force can be seen in Figure 5.3 where the effective force is negative during the third and the fourth quarters of crank revolution.

Separate analyses of pedal force effectiveness during the propulsive and the recovery phases has been performed using the index of effectiveness for each phase (i.e. integral limits from the top dead centre to the bottom dead centre for the propulsive phase) (Rossato, et al., 2008). According to Mornieux et al. (2008), higher pedal force effectiveness is found during the propulsive phase, compared to the recovery phase, with lower effectiveness during the recovery phase possibly related to an inability of the cyclists to generate effective force from the knee and hip joint flexors at higher workloads similar to the those observed during racing.

Pedal force effectiveness can also be calculated over a complete pedal revolution by the instantaneous "ratio of effectiveness", which has been used to assess different parts of the pedal cycle (Sanderson, 1991). When the ratio of effectiveness is close to 1, a greater percentage of the resultant force is transferred as effective positive force. Conversely when the ratio of effectiveness is close to -1, most of the resultant force is transferred as effective force in the opposite direction of crank movement, resulting in resistive force for the contralateral leg. Typical values for the ratio of effectiveness are shown in Figure 5.4 using unpublished data from Rossato et al. (2008) for eight elite cyclists.

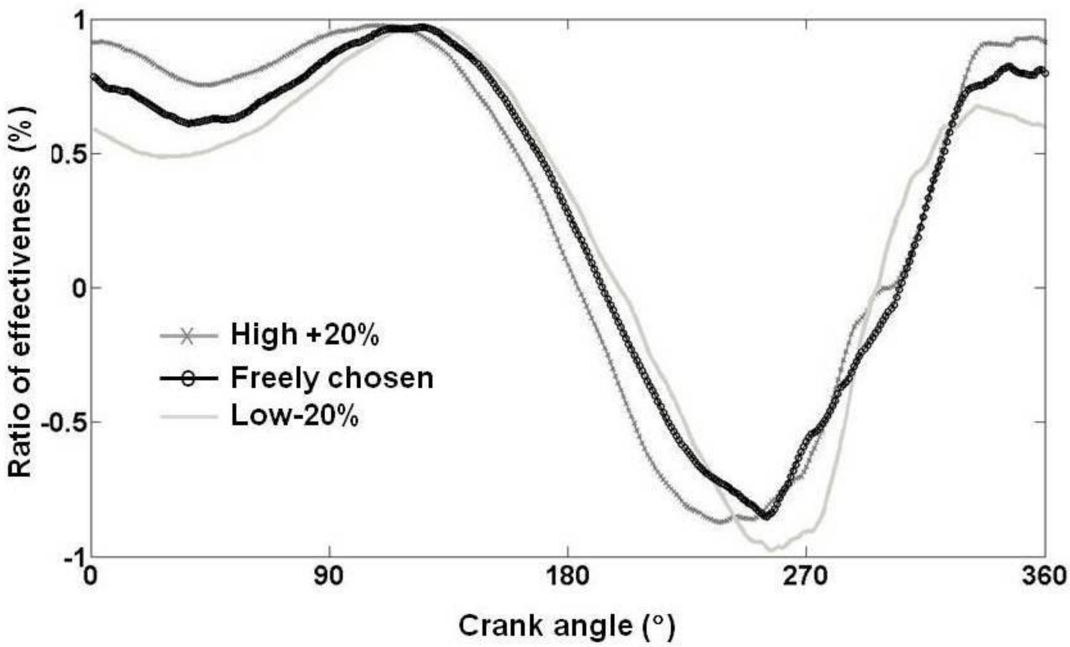


Figure 5.4. Average ratio of effectiveness for eight cyclists pedaling at 80% of their maximal power output. Freely chosen cadence was determined by the cyclists. "Low-20%" indicates pedaling cadence 20% lower than the freely chosen cadence and "High+20%" indicates pedaling cadence 20% higher than the freely chosen cadence. Unpublished data from previous research (Rossato, et al., 2008).

The ratio of effectiveness was close to 1 during the propulsive phase and close to -1 during the recovery phase indicating that the ipsilateral leg was directing most of the force applied on the pedal to generate propulsive torque on the crank (positive effective force). Conversely, during the recovery phase, most of the force applied on the pedal was creating resistive force for the crank (negative effective force). Similar ratios of effectiveness have been previously reported (Sanderson, 1991; Sanderson & Black, 2003).

Limitations on the exclusive use of force effectiveness analysis have been suggested because force effectiveness mixes muscular and non-muscular components of joint moments (Leirdal & Ettema, 2011) and does not fully represent cyclists pedaling technique (Bini & Diefenthaeler, 2010). An alternative analysis (decomposition method) separates the muscular and non-muscular components (mass and inertia) of pedal and intersegmental joint forces (Kautz & Hull, 1993). A limitation of this method is the mechanical dependence of non-muscular components on the muscular component pattern where muscular action will affect non-muscular components, and vice versa. For practical application, the decomposition method requires the analysis of joint kinematics to determine joint moments, which are not commonly measured. Net moments are also prone to errors due to limitations of the inverse dynamics technique (i.e. absence of co-contraction in the model and skeletal muscle modeling). Another approach (ratio between the mechanical work at the top and bottom dead centers by the overall mechanical work of crank revolution) has provided conflicting relationships with economy/efficiency in recent studies (Leirdal & Ettema, 2011, in press). Loras et al. (2009) assessed non-muscular component by measuring forces during unloaded cycling. However, this method is limited because a residual muscular component is still required to move the legs along with inertial components. Therefore, an ecologically valid, sensitive and reliable method of analysis of pedal force effectiveness to better represent cyclists pedaling technique is still required.

Most previous studies were conducted assuming symmetry between the right and left pedal forces. However, pedal force symmetry of non-injured athletes has ranged between ~2% (Smak, et al., 1999) to ~3% (Bini, Diefenthaeler, Carpes, & Mota, 2007). In injured non-athletes, pedal force asymmetry up to 400% has been reported between the non-injured and injured leg (Hunt, et al., 2003; Mimmi, Pennacchi, & Frosini, 2004). Further analysis should explore the degree of symmetry of each force component during the pedal cycle and whether the force symmetry is related to cycling ability, or other factors. Currently, few studies have presented asymmetries in crank torque for uninjured cyclists (Carpes, et al., 2007a; Daly & Cavanagh, 1976).

Constraints on force effectiveness

Pedal force effectiveness depends on workload level (Kautz, et al., 1991; Zameziati, Mornieux, Rouffet, & Belli, 2006), pedaling cadence (Candotti, et al., 2007; Patterson & Moreno, 1990), body position on the bicycle (Bini, Carpes, & Diefenthaeler, 2009; Diefenthaeler, et al., 2006; Diefenthaeler, et al., 2008; Dorel, Couturier, et al., 2009), fatigue (Diefenthaeler, Bini, Carpes, & Vaz, 2007; Dorel, Drouet, et al., 2009) and cycling experience/ability (Candotti, et al., 2007; Sanderson, 1991) (see Table 5.2).

On cycle ergometers, workload is measured by the average power output (in Watts) or the total mechanical work over time (in Joules) and calculated from the torque and angular velocity of the cranks. Crank torque depends on the mechanical characteristics of the bicycle (crank arm length) and on the effective force. The longer the crank arm length, the higher the torque for the same angular velocity and effective force.

Most studies assessed pedal force effectiveness during laboratory controlled trials at aerobic levels of workload (submaximal cycling). Pedal forces acquired during sprint cycling (5 s) conducted on a cycle ergometer were only reported by Dorel et al. (2010). Therefore, little is known about the effects of supramaximal (or anaerobic) workload levels for cycling variables (e.g. body position on the bicycle, fatigue and cycling experience/ability).

Figure 5.5 shows the normal, anterior-posterior, effective and resultant force components applied on the right pedal during three stages of an incremental maximal cycling test (75%, 90% and 100% of the maximal aerobic power output) from eleven competitive male cyclists (Bini, et al., 2007). At higher workload levels, the peak of the effective force was ~20% greater during the propulsive phase (between 0° and 180° of crank angle) and ~110% lower (less negative) during the recovery phase (between 180° and 360° of crank angle). Increases in the effective force are usually due to higher resultant and normal forces during the propulsive phase. At 100% of the maximal aerobic power output there was a ~58% reduction in the forward (positive) pedal force component and a ~175% increase in the backward (negative) pedal force component.

Table 5.2. Scientific papers related to effects of workload, pedaling cadence, body position, fatigue and cycling ability on pedal force effectiveness.

Reference	Independent variable	Subjects	Measurement System	Main results and notes
Dal Monte et al. (1973)[A]	Workload	Not defined.	Two pedals with tension transducer for recording normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane.	Qualitative increased pedal force application at higher workload and during standing cycling. Qualitative asymmetry results for pedal forces.
Daly & Cavanagh (1973)	Bilateral symmetry, workload and pedaling cadence	Twenty male non-cyclists with undefined age.	Strain gauge deformation based system to measure effective force in both cranks.	Higher within day (0.87) and lower between days (0.47) reliability in force symmetry. Undefined effects of workload and pedaling cadence in force symmetry because of high variability.
Ericson & Nisell (1988)	Workload, pedaling cadence and saddle height	Six male non-cyclists between 20 and 31 years.	Piezoeletric sensors attached to the left pedal for measurement of three pedal force components (normal-Fz, anterior-posterior -Fx and medio-lateral -Fy). Pedal and crank angle measurements from video images.	Improved pedal force effectiveness (three times) when workload was increased without effects of pedaling cadence or saddle height.
Patterson & Moreno (1990)	Pedaling cadence	Eleven recreational cyclists between 21 and 44 years.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right and left).	66% higher force effectiveness when changing from 100 to 200 watts of workload. 1.5 lower force effectiveness when cadence increased from 50 to 110 rpm.

Kautz et al. (1991)	Workload	Fourteen male trained cyclists with 23 ±3 years of age.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital (right pedal).	42 % higher pedal force effectiveness at the power phase and 3% lower pedal force effectiveness at the recovery phase when changing the workload from 60 to 90% of maximal oxygen uptake.
Coyle et al. (1991)	Cycling performance level	Fourteen male trained cyclists with 23 ±3 years of age.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	8% lower pedal force effectiveness for the best performance group compared to the ones who does not have the best performance.
Sanderson (1991)	Pedaling cadence and cycling expertise	Seven trained cyclists 30 ±11 years old and 38 male recreational cyclists 26 ±7 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	44% lower pedal force effectiveness changing from 60 to 100 rpm, 16% lower pedal force effectiveness when changing from 80 to 100 rpm. 56% greater pedal force effectiveness changing from 100 W to 235 W.
Black et al. (1993)	Workload	Five trained cyclists with undefined age and gender.	Piezoeletric system to measure the three pedal force components (normal-Fz, anterior-posterior -Fx and medio-lateral-Fy) and the three moments on the X, Y, and Z axis of the right pedal surface (Mx, My, and Mz). Pedal angle measured by potentiometer and optical sensors to calculate the crank angle.	100% increase in pedal force effectiveness in the end of the test.

Amoroso et al. (1993)	Fatigue	Eleven competitive cyclists with undefined age and gender.	Piezoeletric system to measure the three pedal force components (normal-Fz, anterior-posterior -Fx and medio-lateral-Fy) and the three moments on the X, Y, and Z axis of the right pedal surface (Mx, My, and Mz). Pedal angle measured by potentiometer and optical sensors to measure crank angle.	No significant difference was found in pedal force effectiveness.
Caldwell et al. (1998)	Seated vs. standing cycling	Eight elite male cyclists 28 ±5 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (undefined side).	30% higher pedal force effectiveness comparing uphill standing to uphill seated.
Sanderson & Black (2003)	Fatigue	Twelve competitive male cyclists 28 ±6 years old.	Piezoeletric system to measure right normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	No fatigue effects on pedal force effectiveness.
Zameziati et al. (2006)	Workload	Ten male non-cyclists 26 ±1 years old.	Monark cycle ergometer attached on a force plate to allow the measurement of the three pedal force components (normal-Fz, anterior-posterior -Fx and medio-lateral-Fy) and the three moments on the X, Y, and Z axis of the right and left pedals surface.	Positive relationship (r = 0.79) between pedal force effectiveness and economy/efficiency. Positive relationship between pedal force effectiveness during the recovery phase and economy/efficiency (r = 0.66).

Diefenthaeler et al. (2006)	Saddle height and horizontal position	Three male competitive cyclists 26 ±4 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	5-7% lower pedal force effectiveness when moving the saddle forward and 2-7% lower pedal force effectiveness when moving the saddle up or down.
Diefenthaeler et al.(2007)	Fatigue	Eight male elite competitive cyclists (31 ±6 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right and left).	No difference in pedal force effectiveness throughout the fatigue cycling test.
Candoti et al. (2007)	Pedaling cadence and cycling expertise	Nine male competitive cyclists 25 ±8 years old and eight male competitive triathletes 27 ±9 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	30% decrease in pedal force effectiveness when cadence changed from 60 to 105 rpm. 22% higher pedal force effectiveness for cyclists compared to triathletes. No significant differences between groups for higher cadence (90 and 105 rpm).
Korff et al. (2007)	Pedaling technique	Eight male competitive cyclists 35 ±6 years old.	Piezoeletric sensors attached on the right and left pedals for measurement of three pedal force components (normal-Fz, anterior-posterior -Fx and medio-lateral-Fy). Pedal and crank angle measurements by video images.	Two times higher pedal force effectiveness when changing from the preferred to the pulling technique.

Rossato et al. (2008)	Workload and pedaling cadence	Eight male competitive cyclists 24 ±3 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	13% higher pedal force effectiveness changing from 80% to 60% of peak power output. No significant changes in pedal force effectiveness between the preferred, the 20% faster and the 20% slower pedaling cadences.
Mornieux et al. (2008)	Pedaling technique	Eight elite cyclists and seven non-cyclists with undefined age.	Hall effect sensors attached in a custom made adaptor to measure normal (Fz) and anterior-posterior (Fx) pedal force components (right and left).	20% higher pedal force effectiveness on the recovery phase.
Bini et al. (2009)	Knee position relative to the bicycle frame	Three male competitive cyclists and three male competitive triathletes 29 ±9 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right pedal).	No significant effects of knee position on pedal force effectiveness.
Dorel et al. (2009)	Upper body position (trunk lean)	Twelve male competitive triathletes 31 ±8 years old.	Effective force measurement based on torque and crank arm measurement using the cycle ergometer system (right and left).	9.5% lower pedal force effectiveness at the recovery phase of crank revolution in the aero position, compared to the upright position.
Dorel et al. (2009)	Fatigue	Ten competitive cyclists 21 ±3 years old.	Strain gauge deformation based system to measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right and left).	No significant fatigue effects in pedal force effectiveness.
Emanuele et al.	Upper body	Six male amateur	Strain gauge deformation based system to	No significant differences between the upright

(2011)	position (trunk lean)	cyclists 28 ±3 years old.	measure normal (Fz) and anterior-posterior (Fx) pedal force components in the sagital plane (right and left).	position and the position with hands on the drops of the handlebars.

[A]These studies did not indicate the number or characteristics of the subjects.

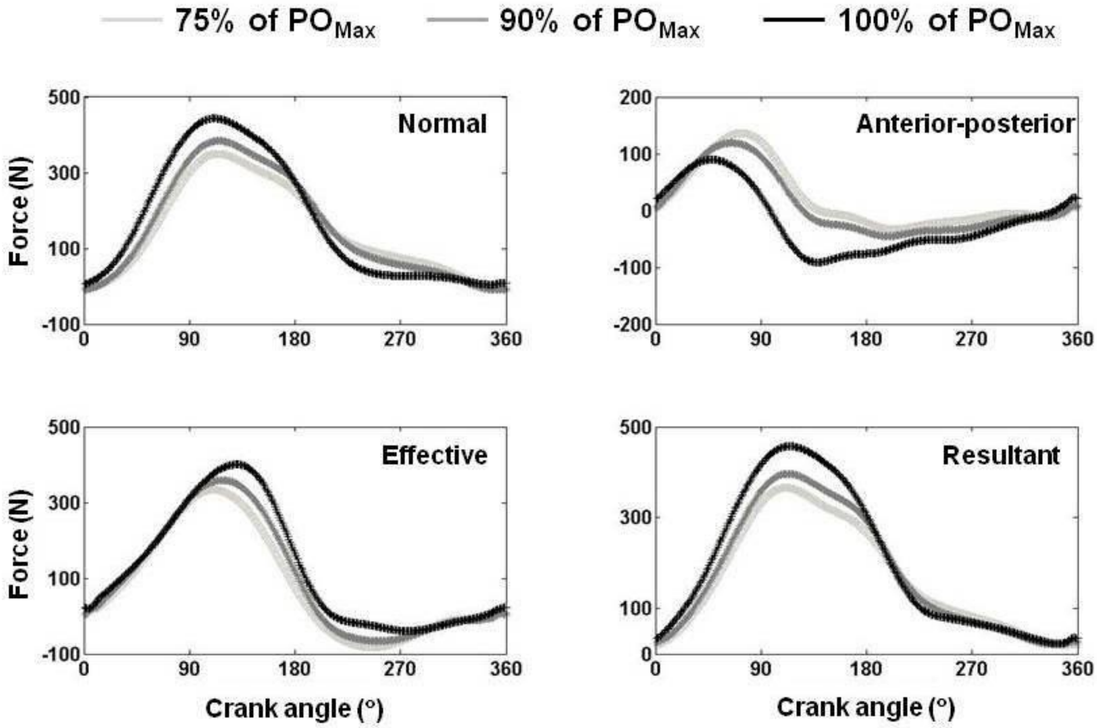


Figure 5.5. Average normal (Fz), anterior-posterior (Fx), effective (EF), and resultant (RF) forces applied to the right pedal from eleven cyclists during three stages of an incremental test (75, 90 and 100% of the maximal power output). Propulsive effective force is positive. Positive normal force is force applied to pull the pedal. Positive anterior-posterior force is forward force applied to the pedal. Unpublished data from previous research (Bini, et al., 2007).

It is unclear why cyclists present lower pedal force effectiveness at lower workload levels. Studies showed that wide increases in workload level (i.e. from 60% to 98% of maximal aerobic power output) led to higher force effectiveness (Black, et al., 1993; Zameziati, et al., 2006), which was not observed when smaller differences in workload level (i.e. from 75% to 100% of maximal aerobic power output) were assessed (Bini & Diefenthaeler, 2010). One possibility is that when improving pedal force effectiveness cyclists may increase activation of muscles that are less efficient (i.e. hip and knee flexors) which may increase energy expenditure and reduce economy/efficiency (Korff, et al., 2007; Mornieux, et al., 2008). Therefore, to maintain a lower oxygen uptake cyclists may postpone recruiting these less efficient muscles and rely on the knee and hip joint extensors to produce power (Fernandez-Pena, Lucertini, & Ditroilo, 2009).

The effect of pedaling cadence on pedal force effectiveness is uncertain (Ansley & Cangley, 2009). When cycling at constant workload in the laboratory, cyclists can minimize resultant force application by riding at approximately 90 rpm (Candotti, et al., 2007; Ericson & Nisell, 1988; LaFortune & Cavanagh, 1983a; Neptune & Herzog, 1999; Patterson & Moreno, 1990). Most studies have shown higher pedal force effectiveness at lower pedaling cadences (i.e. 60 rpm) when compared to self selected cadences (Candotti, et al., 2007; Ericson & Nisell, 1988; LaFortune & Cavanagh, 1983a). Improved pedal force effectiveness at low cadence may be due to lower overall muscle activation (MacIntosh, Neptune, & Horton, 2000), lower joint moments (Marsh, Martin, & Sanderson, 2000; Takaishi, Yamamoto, Ono, Ito, & Moritani, 1998) and reduced co-activation of extensor/flexor groups (Candotti, et al., 2009; Neptune & Herzog, 1999). In contrast, Rossato et al., (2008) reported that pedal force effectiveness of cyclists did not differ at a cadence 20% higher than the self selected cadence. Experienced cyclists typically pedal at high cadence (~100 rpm) resulting in reduced activation of the main driving muscles (i.e. vastus lateralis and gluteus maximus) (Lucia, et al., 2004), lower joint moments (i.e. reduced resultant moments) (Marsh, et al., 2000), hemodynamics (Ansley & Cangley, 2009) and effort perception (Ansley & Cangley, 2009). Cyclists may be able to sustain pedal force effectiveness while cycling at high pedaling cadences (Candotti, et al., 2007; Rossato, et al., 2008).

The configuration of bicycle components determines the position of the body on the bicycle, though it is acknowledged that different body positions can be obtained despite no change in bicycle geometry (e.g. by varying hand placement). Any change in body position resulting from a change in saddle height will affect knee angle (Nordeen-Snyder, 1977; Sanderson & Amoroso, 2009), muscle activation (Ericson, et al., 1985; Jorge & Hull, 1986), muscle length (Sanderson & Amoroso, 2009), and oxygen uptake (Nordeen-Snyder, 1977; Shennum & DeVries, 1976). Only two studies measured pedal forces at different seat heights. Ericson & Nisell (1988) found that seat height changes (±8% of the ischial tuberosity to the floor) did not affect pedal force effectiveness of non-cyclists. However, in trained cyclists, Diefenthaeler et al. (2006) found that increasing or decreasing saddle height from the preferred position by 1 cm reduced pedal force effectiveness by ~5.5%. It is likely that the experienced cyclists who were adapted to their bicycle configuration due to training were sensitive to the small changes in saddle height resulting in the acute effect on pedal force effectiveness, or it was simply a sub-optimal position. To ascertain the effect of adaptation to bike configuration, future research should repeat measures of pedal force effectiveness when cyclists are adapting to a new position to determine the magnitude and time course of adaptation.

In addition to the height of the saddle, the forward-backward position of the saddle may also affect pedal force effectiveness, based on changes in ankle joint kinematics (Price & Donne, 1997) and muscle activation (Ricard, et al., 2006). However, to our knowledge, no study has reported pedal force effectiveness at different forward-backward seat configurations.

Trunk angle (upright versus the most aerodynamic position) has an effect on effective force (Dorel, Couturier, et al., 2009). With the trunk in the most aerodynamic position the effective force was 9.5% lower during the recovery phase compared to the upright position (Dorel, Couturier, et al., 2009). In the aerodynamic position, the angle between the trunk and thigh was smaller which reduced the activation and possibly the length of hip joint flexor muscles, thereby decreasing the ability to generate pulling force during the recovery phase (Dorel, Couturier, et al., 2009). In contrast, Emanuele et al. (2011) observed no changes in effective force when cyclists used a position of the hands on the drops of the handlebars compared to the upright position (hands on the top of the handlebars). Increased hip power production and reduced knee joint power when the hands were on the drops were in contrast to findings from Dorel et al. (2009). Further research is required to assess to what extent upper body flexion compromises hip and knee muscle actions and pedal force effectiveness.

Cyclists usually stand up on the bicycle to ride uphill to benefit from using their upper body mass to apply force on the pedal in the downstroke phase (Caldwell, et al., 1998). Specifically, Caldwell et al. (1998) reported that the peak torque for the same workload level and pedaling cadence increased by ~30% and total pedal force increased by ~50% when standing compared to seated cycling uphill. Therefore any changes in torque profile would have come from changes in total pedal force with potential decreases in pedal force effectiveness. Consequently, the 30% higher (and delayed) peak torque and 50% greater total pedal force suggests reduced pedal force effectiveness when standing on the bicycle during uphill riding. Conversely, cycling at 75% of the workload of maximal oxygen uptake at 11% of incline has not changed pedal force effectiveness compared to level cycling for another study (Leirdal & Ettema, in press). Further research is needed on inclination effects on pedal forces.

Most studies failed to show a consistent change in pedal force effectiveness when cyclists were in a fatigued state (Diefenthaeler, et al., 2007; Sanderson & Black, 2003). Studies that did report changes with fatigue showed an increase in pushing down normal force during the propulsive phase (Amoroso, et al., 1993; Dorel, Drouet, et al., 2009), in resistive force during the recovery phase (Sanderson & Black, 2003), and in the pulling backward force on the pedal surface during the recovery phase (Dorel, Drouet, et al., 2009). For these studies, cyclists were either assessed at a fixed workload level of 300 W (Amoroso, et al., 1993) or at 80% of maximal aerobic power output (Dorel, Drouet, et al., 2009; Sanderson & Black, 2003) during time to exhaustion testing. These results suggested that lower limb mechanics change to balance for fatigue and sustain pedal force effectiveness. Increased ankle dorsi-flexion (Amoroso, et al., 1993; Sanderson & Black, 2003), higher range of motion for the ankle joint (Bini, Diefenthaeler, et al., 2010) and reduced knee flexion angle (Bini, Diefenthaeler, et al., 2010) have been found during fatigue. The increased activation of hip extensor muscle activation (Dorel, Drouet, et al., 2009) contrasted with the unchanged individual joint contribution to the absolute joint moments (Bini, Diefenthaeler, et al., 2010). Further research should shed light on how fatigue affects muscle activation and joint kinematics without substantial effects in joint moments.

It is unclear how experience in cycling affects pedal force effectiveness. From a cross-sectional perspective, differences were found between cyclists and non-cyclists (Mornieux, et al., 2008), cyclists and triathletes (Candotti, et al., 2007), but no differences were found between competitive and recreational cyclists (Sanderson, 1991). If pedal force effectiveness is important for performance it may be expected that pedal force effectiveness would be related to competitive results within a cohort of cyclists. However, a study of 14 competitive cyclists reported that the cyclists who achieved better performance indices where the ones who had lower pedal force effectiveness but were able to apply higher normal force on the pedal (Coyle, et al., 1991). Recent studies (Korff, et al., 2007; Mornieux, et al., 2008) have analyzed pedal force effectiveness and cycling efficiency with the aim of determining why there is a lack of relationship between pedal force effectiveness and performance in cycling. No relationship was found between economy/efficiency and pedal force effectiveness during sub maximal trials at constant aerobic power output, yet in the cycling community, it is advocated that better force effectiveness can be translated to higher economy/efficiency (Cavanagh & Sanderson, 1986). Further research is needed to increase our understanding of the implications of cycling experience on pedal force effectiveness.

Technique and performance

Improved pedal force effectiveness should theoretically result in an increase in economy/efficiency but this has not been the case (Korff, et al., 2007). However, cyclists still aim to improve pedaling technique via improving pedal force effectiveness. Research studies have provided visual feedback of pedal forces or have used assisting devices (e.g. decoupled cranks) to stimulate the cyclist to change their natural movement to improve pedal force effectiveness.

When cyclists are given feedback of pedal forces they can improve their force effectiveness (Broker, et al., 1993; Sanderson & Cavanagh, 1990; Ting, Raasch, Brown, Kautz, & Zajac, 1998). Visual (augmented) feedback of pedal force has been used in different phases of the pedal cycle (Hasson, et al., 2008; Henke, 1998; Holderbaum, et al., 2007) without differences between summarized and real time feedback (Broker, et al., 1993). Presentation of an ideal force diagram and the actual force (similar to the one presented in Figure 3) has been used as feedback (Hasson, et al., 2008; Holderbaum, et al., 2007). Cyclists were instructed to apply force on the pedal so their normal and anterior-posterior components of pedal force were closer to the ideal profile. Regardless of whether they focused only on the recovery phase or on specific quarters of pedal cycle, force effectiveness had similar improvements after training.

Changes in pedal force effectiveness with feedback occurs rapidly with one study reporting significant changes in novice cyclists after one session (Hasson, et al., 2008). Sanderson & Cavanagh (1990) showed that after the first two days of training, recreational cyclists improved pedal force effectiveness (lower resultant force during the recovery phase). No marked differences between the second and the 10th training sessions indicated that a plateau exists in pedal force effectiveness development. Retention of force effectiveness was similar one week and three months after cessation of the training period (Broker, et al., 1993).

Provision of visual feedback for trained (Henke, 1998) and recreational cyclists (Sanderson & Cavanagh, 1990) has resulted in improvements in force effectiveness ranging from 17% to 40%. Studies with non-cyclists (Broker, et al., 1993; Hasson, et al., 2008; Holderbaum, Bini, Nabinger, & Guimaraes, 2005; Holderbaum, et al., 2007; Nishiyama & Sato, 2005) have reported improvements in force effectiveness between 24% and 55%. However, Mornieux et al. (2008) compared pedal force effectiveness of cyclists and non-cyclists who were instructed to increase pulling upward forces during the recovery phase (one trial of feedback). Economy/efficiency reduced by 3% in non-cyclists and 10% in trained cyclists. Both groups reduced economy/efficiency by improving pedal force effectiveness, with worst results for trained cyclists. Long term adaptation to a specific motion (i.e. higher pushing forces during the propulsion phase) can result in neuromuscular adaptation for cyclists (Candotti, et al., 2009; Chapman, Vicenzino, Blanch, & Hodges, 2008), and changes in pedal force profile (Candotti, et al., 2007), which may limit their acute adaptation to changing motion (i.e. pedaling with higher force effectiveness). Physiological adaptation of highly trained cyclists (Coyle, et al., 1991) may support the hypothesis that cyclists are more efficient recruiting the quadriceps muscle group during a cycling task compared to non-cyclists (Takaishi, et al., 1998). When improving pulling upward forces during the recovery phase, cyclists recruited "less efficient" muscles, which resulted in a reduced economy/efficiency (Edwards, et al., 2009; Korff, et al., 2007; Mornieux, et al., 2008). However, Theurel et al. (in press) reported smaller reductions in sprint cycling power due to fatigue from 45 minutes of cycling at 75% maximal aerobic power output when cyclists received feedback to improve pedal force effectiveness. There was smaller economy/efficiency during the first 15 minutes of the test when using feedback, without differences in the following 30 minutes. Further research should be conducted using a control group (no feedback) to ascertain any learning effects, which were not addressed in the previous study.

To date only Mornieux & Stapelfeldt (in press) have assessed the effects of longer training (four weeks) using force effectiveness feedback for 12 sessions of 30 minutes at 60% maximal aerobic power output and 80 rpm pedalling cadence. No improvements in maximal aerobic power output occurred for the feedback group compared to the control group (no feedback during training). The feedback group did reduce force effectiveness during the propulsive phase of crank revolution and increased force effectiveness during the recovery phase. It is therefore unlikely that improving pedal force effectiveness with training may enhance performance in cycling. Further research at higher workload levels (>60% maximal aerobic power output) and pedaling cadence (>80 rpm) for training may provide evidence of whether force feedback training may (or may not) be useful in improving cycling performance.

On a normal bicycle the cranks are diametrically opposed (180°) and fixed which links the forces at each pedal. In an attempt to encourage higher force effectiveness, novel systems have been developed where the cranks are decoupled. These "Powercranks" (or Smartcranks) require a pulling force during the recovery phase of the crank cycle, and at the bottom dead centre, because the crank is attached to the chain ring via a free bearing system. This higher pulling force on the recovery phase was previously related to higher force effectiveness using decoupled cranks (Bohm, et al., 2008).

Only one study (Luttrell & Potteiger, 2003) reported benefits after training with decoupled cranks in cycling economy/efficiency. Six novice cyclists trained using Powercranks (Walnut Creek, CA) for six weeks at 70% of VO_{2Max} for one hour per day. After the training period, cyclists who trained using Powercranks improved economy/efficiency by 2.3% during a one hour constant load test, compared to the group who trained using normal cranks. Changes in economy/efficiency may have been caused by changes in muscle activation profiles of knee and hip flexor groups. A study showing decreased activity of vastus lateralis and increased biceps femoris after two weeks of training for 30-45 minutes per session at undefined workload using Powercranks provided some support for this suggestion (Fernandez-Pena, et al., 2009). In contrast, a similar study with five weeks training twice per week at 80% of the individual's anaerobic threshold found no changes in economy/efficiency for ten trained cyclists (Bohm, et al., 2008), even though force effectiveness did improve. It is unclear why the results from the two studies differed following such similar interventions. Possible lower levels of training of the "novice" cyclists from the study of Luttrell & Potteiger (2003) may explain the differences. Another study (Williams, et al., 2009) found no changes in power output at lactate threshold, economy/efficiency during steady state cycling, and time trial performance, of well-trained cyclists following training using decoupled crank systems. Until more evidence is available it is difficult to assess the potential benefit or harm of training with decoupled cranks.

There are several areas that research would contribute to improving the understanding of the relationship between optimal force effectiveness and performance. Establishing a "natural" range of symmetry of pedal forces should be the goal of future research and may explain the influence of symmetry in cycling performance and injury prevention. In addition, the effects of pedal force effectiveness training on economy/efficiency may be a focus of future research. Higher levels of workload (>60% maximal aerobic power output) and pedaling cadence (>80 rpm) for training may be used in future research, which may allow adaptation of the higher hip and knee flexors recruitment to pulling forces. Cycling experience may reduce adaptation to technique training. Comparison of competitive cyclists, triathletes and recreational cyclists may help identify populations likely to benefit from force effectiveness training. The use of decoupled crank systems should be investigated for longer training periods with different experience and ability levels in cycling.

Conclusions

Pedal forces are often based on the measurement of normal, anterior-posterior, effective and resultant force components, with analysis of pedal force effectiveness based on the computation of the index of effectiveness. Workload level and pedaling cadence affect pedal force effectiveness, but there are unclear effects of body position on the bicycle, fatigue state, cycling experience and ability on pedal force effectiveness.

Technique training, using either augmented feedback of pedal forces or decoupled cranks, increases pedal force effectiveness in short duration studies but evidence of augmented feedback efficacy in long term studies is lacking. The effects of technique training trying to improve force effectiveness on economy/efficiency and performance are unclear.

CHAPTER 6: BETWEEN-DAY RELIABILITY OF PEDAL FORCES FOR CYCLISTS DURING AN INCREMENTAL CYCLING TEST TO EXHAUSTION

Overview

Between-day variability in pedal force measurements has not yet been presented. Therefore, the aim of our study was to assess the variability of pedal force measures during incremental cycling tests to exhaustion performed on separate days. Ten cyclists with competitive experience performed two incremental cycling tests to exhaustion separated by two to seven days. Force and pedal-to-crank angles on the right and left pedals were measured during every step of 50 W of the test. Peak normal and anterior-posterior forces, average total force and pedal force effectiveness were computed for every step of the test, for every cyclist and compared using typical error, intraclass correlation and effect sizes for both test days. Differences between test days for pedal force variables ranged from 5% to 14% with magnitudes from trivial to moderate. Greater repeatability was found for peak normal and total force applied on the pedals, with variability increasing for anterior-posterior force and pedal force effectiveness. Pedal force variables were highly reliable between two to seven days of test with similar results compared to VO_2 assessed during an incremental step test to exhaustion.

Introduction

Cycling laboratory-based assessment has been extensively used to assess variables that can predict performance (Amann, et al., 2004). Physiological (VO_2 and heart rate) and biomechanical variables (power output) were amongst the main variables used to predict cycling performance and to assess training effects (Coyle, et al., 1991). Apart from that, pedal forces were also measured during laboratory tests using varying protocols (Bini & Diefenthaeler, 2010; Leirdal & Ettema, 2011; Mornieux, et al., 2008). There has been an increasing interest in monitoring training effects on the effectiveness of pedal forces (ratio between force driving the crank by the total force on the pedal) (Bohm, et al., 2008).

Reliability of cycling performance during laboratory based tests has been shown for peak power output during 30 seconds sprint (1.2-1.6%) (Watt, Hopkins, & Snow, 2002), 40-km time trial performance (3.5-4.5%) (Laursen, Shing, & Jenkins, 2003) and cycling efficiency (~0.6%) (Noordhof, et al., 2010). Hug et al. (2008) reported that variability in the effectiveness of pedal forces within a single session was 7.7-12.4% when competitive cyclists were assessed at workloads of 150 W and 250 W. However, no published study was found comparing effectiveness of pedal forces and other pedal force variables (e.g. total pedal force) in sessions conducted on separate days. This information is important for assessing the smallest worthwhile effect of training interventions for pedal force variables.

Incremental cycling exercise to exhaustion is a widely accepted cycling performance test which enables the measurement of physiological variables such as oxygen uptake (Lucía, Hoyos, Pérez, Santalla, & Chicharro, 2002) and ventilatory thresholds (Weston & Gabbett, 2001) during maximal effort. Recently, biomechanical variables such as muscle activity (Candotti, et al., 2008) and joint kinetics and kinematics (Bini & Diefenthaeler, 2010) have been measured during incremental tests to exhaustion. The rationale for linking biomechanical and physiological variables is that more realistic comparisons can be conducted between cyclists of different performance levels, instead of the definition of a set workload level. During the incremental test, workload is controlled and gradually increased until exhaustion and cyclists are instructed to control pedalling cadence using visual feedback (Bentley, Newell, & Bishop, 2007). Therefore, the incremental test is a well-controlled test providing the possibility to compare results across subjects of varying performance levels by percentage of maximal performance or by percentage of ventilatory threshold. The incremental test is a suitable test to assess pedal forces when workload is increased in a step profile, which secures a minimum time at each consistent workload.

Therefore, the aim of our study was to assess the reliability of pedal force measures during incremental cycling tests to exhaustion performed on separate days.

Methods

Participants

A quantitative repeated measures experimental design was used to collect data. Ten cyclists with competitive experience in cycling and triathlon were invited to participate in the study. Cyclists' (mean ±SD: 34 ±8 years, 72 ±13 kg, 177 ±12 cm, 59.6 ±7.4 $ml \cdot kg^{-1} \cdot min^{-1}$ maximal oxygen uptake, 372 ±80 W peak power output, 5.2 ±0.7 $W \cdot kg^{-1}$ peak power per body mass) signed an informed consent form in agreement with the committee of ethics in research of the institution where this study was conducted. No cyclist had an injury that would impact on test performance at the time of data collection.

Data collection

Pedal force components (normal and anterior-posterior) were calibrated using the regression between three static load points (0 kg, 5 kg and 10 kg) applied to the pedals and voltage output when R^2 was greater than 0.99. Mechanical coupling between anterior-posterior and normal loads were corrected using a gain matrix (Leirdal & Ettema, 2011). Potentiometers were calibrated using a manual goniometer set at four angles (0°, 90°, 180° and 270°) to compute the relationship between voltage output and the measured angle. The calibration factors were defined when mean differences in voltage were lower or equal to 1%.

Body mass and height were measured according to ISAK protocols (Marfell-Jones, et al., 2006). Cyclists completed the Waterloo inventory to allow the determination of lower limb dominance (Carpes, et al., 2011). Cyclists' bicycle saddle height and horizontal position were measured to set-up the stationary cycle ergometer (Velotron, Racemate, Inc). The cyclists performed an incremental cycling exercise on the cycle ergometer with three minutes of warm-up at 100 W and pedaling cadence visually controlled at 90 ±2 rpm. The workload was then increased to 150 W and remained increasing in a step profile of 25 W/min until cyclists' exhaustion (Lucía, et al., 2002). A script was configured in the Velotron CS2008 software (Velotron, Racemate, Inc) for automatic control of the cycle ergometer workload in a constant workload mode. This configuration enabled a constant workload with cycle ergometer resistance constantly changing to balance for fluctuations in pedalling cadence. Gas exchanges were continuously sampled from a mixing chamber where samples were drawn into the oxygen and carbon dioxide analyzers for continuous measurement using a metabolic cart (TrueOne 2400, Parvo Medics, Salt Lake City, UT, USA). Analyzers for oxygen and carbon dioxide were calibrated according to manufacturer recommendations. Maximal aerobic workload and maximal oxygen uptake were defined as the highest workload measured during the test and as the highest oxygen uptake value computed over a 15 s average of the data, respectively. After two to seven days, cyclists returned to the laboratory at the approximate same time of the day to perform the incremental test following the same procedures.

Normal and anterior-posterior forces were measured using a pair of strain gauge instrumented pedals (Candotti, et al., 2007), with pedal-to-crank angle measured using angular potentiometers attached to the pedal spindle. Pedal force data passed through an amplifier (Applied Measurements, Australia) and, along with potentiometers and reed switch signals were recorded using an analogue to digital board PCI-MIO-16XE-50 (National Instruments, USA) at 600 Hz per channel using a custom made script in Matlab (Mathworks Inc, MA). Analogue data were acquired between the 20th and the 40th s of each step of 50 W (i.e. 100 W, 150 W, 200 W, 250 W, etc).

Data analyses

Pedal-to-crank angle (see Figure 6.1) measured by the potentiometers were converted into sine and cosine to compute tangential and radial forces on the cranks. Low pass zero lag Butterworth digital filter with cut of frequency of 10 Hz was applied to the sine and cosine data from potentiometers to attenuate signal noise from gap in potentiometer voltage readings (Hull & Davis, 1981).

A reed switch attached to the bicycle frame detected the position of the crank in relation to the pedal revolution and enabled to separate pedal force data into every crank revolution. Peak normal and anterior-posterior force on the pedals were computed along with the average total force applied on the sagittal plane of the pedal surface. Pedal force effectiveness was assessed by the index of effectiveness computed as the ratio between the angular impulse of the tangential force on the crank and the linear impulse of the total force applied on the pedal (Rossato, et al., 2008). All force variables were averaged for each subject across five revolutions of the crank for each stage of the incremental test. Oxygen uptake was averaged for each stage of the incremental test.

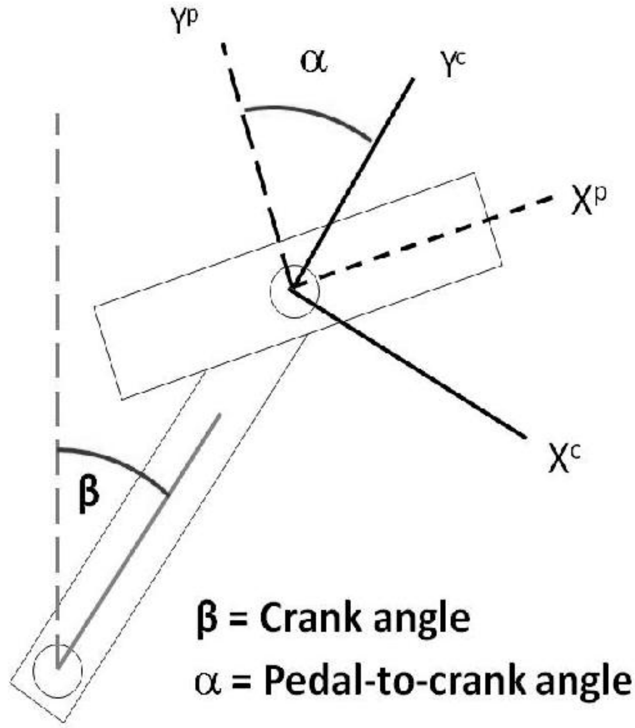


Figure 6.1. Definition of crank angle and pedal-to-crank angle for vertical and horizontal axis of the crank (X^C and Y^C) and the pedal (X^p and Y^p). Normal and anterior-posterior pedal forces were defined analogue to X^p and Y^p, respectively.

Statistical analyses

Errors of calibration of normal and anterior-posterior components and potentiometers of the pedals were computed as average percentage differences in voltage due to calibration load (or angle for potentiometer) in relation to the output voltage. As an example, for the normal force of the right pedal, the difference in voltage from 0 kg to 5 kg was 0.1547 V and the difference in voltage from 5 kg to 10 kg was 0.1544 V, resulting in 0.19% difference in voltage due to load application. Variation in pedalling cadence was computed by percentage differences across five crank revolutions.

Peak normal and anterior-posterior pedal forces, average total force applied on the pedal, index of effectiveness and oxygen uptake were compared between both days of evaluation session. All variables were analyzed for the 100 W, 150 W, 200 W, 250 W, 300 W and 350 W stages of the incremental test. Normality of distribution and sphericity were evaluated via the Shapiro-Wilk and Mauchly tests respectively. For oxygen uptake, right normal force, and anterior-posterior right and left force, a logarithm transform was applied.

Typical error of measurements (Hopkins, 2000) and intraclass correlation coefficients (ICC) were calculated for all variables. SPSS for Windows 16.0 was employed for the analysis of ICC. Finally, Cohen's effect sizes (ES) were computed for the analysis of the magnitude of the differences and subsequently rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Lower limb dominance were assessed by the Waterloo inventory indicating that all ten cyclists reported right leg dominance for more than 80% of the questions of the inventory.

Errors from calibration procedures were 0.19% and 0.68% for the normal force of the right and left pedals, respectively. Anterior-posterior force calibration resulted in errors of 0.68% and 0.56% for the right and left pedals, respectively. Error in pedal-to-crank angle of each potentiometer was 0.5%. Mean variation in pedalling cadence between cyclists was 1% resulting in an estimated error from equipment of ~1.37% and ~1.74% for index of effectiveness of the right and left pedals, respectively.

Between-day differences ranged from trivial to moderate for most pedal force variables. There were trivial differences between days for oxygen uptake for all stages. Percentage differences measured by typical error ranged from 5% to 14% for pedal force variables and typical error was 4% for oxygen uptake with trivial differences assessed by effect sizes (see Table 6.1).

Table 6.1. Mean and standard deviations, typical error of measurement (%) and effect sizes between dayss across different workload levels for oxygen uptake (VO2), peak normal force (NF), peak anterior-posterior force (APF), average total force on the pedal and index of effectiveness (IE) for right and left pedals. The number of cyclists completing each stage varied (n = 10 for 100 W to 250 W; n = 8 for 300 W; n = 6 for 350 W). Abbreviations used are for effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

Variables	100 W (n=10)	150 W (n=10)	200 W (n=10)	250 W (n=8)	300 W (n=10)	350 W (n=6)	Average across all stages typical error; ICC; ES, Magnitude inference
VO_2 ($ml^{\cdot}kg^{\cdot}min^{-1}$) Day 1	24 ±3	26 ±3	35 ±6	42 ±5	49 ±5	54 ±7	-
Day2	23 ±3	26 ±4	35 ±5	43 ±6	49 ±5	55 ±6	-
Day 1 vs. Day 2: typical error; effect size, magnitude inference	4%; 0.1, T	6%; 0.1, T	7%; 0.1, T	3%; 0.2, T	2%; 0.1, T	2%; 0.1, T	4%; 0.94; 0.1, T

Peak NF right (N) Day 1	230 ±39	264 ±37	312 ±50	359 ±52	404 ±59	464 ±63	-
Day2	228 ±41	272 ±48	312 ±52	356 ±51	408 ±59	472 ±49	-
Day 1 vs. Day 2: typical error; effect size, magnitude inference	5%; 0.1, T	11%; 0.2, T	5%; 0.1, T	5%; 0.1, T	3%; 0.1, T	5%; 0.1, T	6%; 0.98; 0.1, T
Peak NF left (N) Day 1	249 ±72	292 ±84	328 ±95	353 ±110	375 ±133	404 ±138	
Day2	245 ±81	292 ±95	334 ±91	349 ±142	372 ±142	397 ±114	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	11%; 0.1, T	13%; 0.1, T	10%; 0.1, T	21%; 0.1, T	7%; 0.1, T	7%; 0.1, T	12%; 0.98; 0.1, T
Peak APF right (N) Day 1	82 ±19	96 ±23	110 ±19	107 ±22	117 ± 29	109 ±30	
Day2	77 ±23	99 ±21	99 ±14	106 ±20	112 ±22	103 ±31	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	11%; 0.2, T	11%; 0.1, T	16%; 0.6, M	8%; 0.1, T	17%; 0.2, T	13%; 0.2, T	13%; 0.95; 0.2, T
Peak APF left (N) Day 1	65 ±18	81 ±26	97 ±20	96 ±20	98 ±15	90 ±27	
Day2	80 ±33	90 ±14	90 ±17	95 ±19	95 ±18	87 ±22	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	13%; 0.6, M	17%; 0.4, S	23%; 0.3, S	8%; 0.1, T	13%; 0.2, T	9%; 0.1, T	14%; 0.96; 0.1, T
Total force right (N) Day 1	129 ±25	138 ±26	153 ±25	164 ±25	180 ±29	201 ±29	

Day2	129 ±27	141 ±28	151 ±28	161 ±28	182 ±22	194 ±25	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	4%; 0.1, T	6%; 0.1, T	6%; 0.1, T	5%; 0.1, T	4%; 0.1, T	3%; 0.3, S	5%; 0.98; 0.1, T
Total force left (N) Day 1	114 ±32	127 ±32	136 ±38	145 ±52	153 ±63	162 ±59	
Day2	118 ±33	130 ±41	142 ±42	146 ±57	158 ±66	165 ±53	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	9%; 0.1, T	15%; 0.1, T	11%; 0.1, T	15%; 0.1, T	10%; 0.1, T	5%; 0.1, T	11%; 0.97; 0.1, T
IE right (%) Day 1	47 ±4	54 ±5	54 ±9	61 ±4	64 ±5	66 ±4	
Day2	49 ±6	55 ±8	59 ±7	61 ±11	64 ±6	67 ±7	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	4%; 0.4, S	13%; 0.2, T	14%; 0.6, M	17%; 0.1, T	6%; 0.1, T	5%; 0.1, T	10%; 0.94; 0.2, T
IE left (%) Day 1	36 ±9	41 ±11	42 ±8	47 ±10	49 ±13	51 ±6	
Day2	36 ±7	42 ±12	43 ±14	50 ±6	46 ±15	49 ±7	
Day 1 vs. Day 2: typical error; effect size, magnitude inference	10%; 0.1, T	16%; 0.1, T	15%; 0.1, T	12%; 0.4, S	22%; 0.2, T	10%; 0.3, S	14%; 0.91; 0.1, T

Discussion

The analyses of training effects in pedal forces can indicate if a training intervention has the potential to enhance (or reduce) pedal force effectiveness. To provide information on the biological error of measuring pedal forces, our study aimed to assess reliability of pedal force variables during incremental tests to exhaustion performed on separate days by competitive cyclists and triathletes. Between days difference ranged from 5% to 14% for pedal force variables (small to trivial) which were greater than the variability observed in oxygen uptake.

Biological error of measurement includes the technical noise in measuring analogue signals and converting them into digital signals along with cyclists' variability in performing the movement (Watt, et al., 2002). Therefore, for the evaluation of training effects, it is important to know the size of the biological error of various measures. In our study, we collected oxygen uptake during the incremental test to present a variable that has already been shown small variability in a similar designed study using the same metabolic cart (Crouter, Antczak, Hudak, DellaValle, & Haas, 2006). Peak normal and total force applied on the right pedal presented 6% and 5% of variation, respectively, which is similar to the results of oxygen uptake (4%). These results may be related to the consistent high pushing forces observed by cyclists during the propulsive phase of crank revolution (from 12 o'clock to 6 o'clock crank positions) (Coyle, et al., 1991). Greater variability in normal (12%) and total force applied on the left (11%) pedal may be due to all cyclists being right leg dominant. Further analysis of bilateral symmetry may shed light on dominance effects in pedal forces.

Compared to normal and total forces applied on the pedal, the anterior-posterior force components presented greater variability between days for right (13%) and left (14%) pedals. Hug et al. (2008) observed that total force applied on the pedal at the top (12 o'clock crank position) and bottom dead centres (6 o'clock crank position) were more variable than the force applied at the 3 o'clock crank position. In these two areas of crank revolution (12 o'clock crank position and 6 o'clock crank position), the anterior-posterior component has greater contribution than at the 3 o'clock crank position. It is also expected that anterior-posterior force would vary because some cyclists will try to pull the pedal backward at the 6 o'clock crank position (Diefenthaeler, Coyle, Bini, Carpes, & Vaz, in press). Therefore, the analysis of anterior-posterior force would be more variable then the normal and total force applied on the pedals.

The effectiveness of the force applied on the pedal was analysed by the index of effectiveness, which depends on the total and on the tangential force on the pedal and on the crank, respectively. To convert pedal force components (normal and anterior-posterior) into tangential crank force, the angle of the pedal must be taken into account. We would expect the greater variability on the index of effectiveness (10% and 14% for right and left pedals) compared to the total force applied on the pedal to be related to differences in pedal angle. Variability in lower limb kinematics within a single session would be expected to depend on cycling experience (Chapman, et al., 2009). However, no published study to date has presented data on variability of kinematics between days. It is possible that greater variability of the index of effectiveness may be related to variability in pedal and lower limb kinematics between days.

Further research is needed to assess variability in joint kinematics acquired in separate days to measure the biological error. To compute the index of effectiveness, information on normal and anterior-posterior force components is combined with pedal angle. In our study we estimated a combined error of right and left index of effectiveness of ~1.37% and ~1.74% due to calibration procedures of pedal forces and potentiometers. We infer that the differences observed between days in determination of the index of effectiveness from unknown sources (e.g. cyclists variability to perform the task) may be of ~8.34% and ~12.45% for right and left pedals. Bilateral symmetry in joint kinematics has been shown for non-cyclists, without reports on biological error between legs (Edeline, Polin, Tourny-Chollet, & Weber, 2004).

Between day variability on pedal forces did not present a trend depending on workload. For the index of effectiveness of the right pedal, moderate differences between days were observed for the 250 W (14%) and trivial differences were found for 350 W (5%). Within-cyclist variability may be reduced by assessing various levels of workload and accounting for average results across different workload levels, as conducted in our study. Therefore, conclusions regarding training effects drawn by a single assessment of pedal forces may be subject to greater (or smaller) biological error leading to overestimated (or underestimated) effects of training. It is also unknown if assessing pedal force during variable workload and pedalling cadence (e.g. time trial) may change variability of pedal force variables.

One limitation of our study was not providing a session for familiarisation on the incremental test before data were collected. The automatic control of workload by the cycle ergometer software and the visual control of pedalling cadence by the cyclists may have provided a consistent protocol with highly repetitive performance. We expected high repeatability in cycling motion due to the small variation in oxygen uptake (4%) and peak normal force (5%). Assessing lower limb kinematics along with pedal forces would have indicated if variability between days emerged from joint movements or from muscle force production via joint kinetics analyses.

Conclusions

Trivial differences in peak normal and anterior-posterior forces, total pedal force and index of effectiveness were observed between days. Greater reliability was found for peak normal and total force applied on the pedal, with variability increasing for anterior-posterior force and index of effectiveness. Pedal force variables were highly reliable between two to seven days of testing with similar results compared to oxygen uptake assessed during an incremental step test to exhaustion.

CHAPTER 7: PEDALLING TECHNIQUE CHANGES WITH FORCE FEEDBACK TRAINING IN COMPETITIVE CYCLISTS AND TRIATHLETES: PRELIMINARY STUDY

Overview

We compared training with different types of pedal force feedback on performance in cycling. Six cyclists performed eight technique-training sessions on a cycle ergometer with visual feedback of pedal forces from right then left (5 s each) pedals every 500 m in initial sessions, then less frequently in subsequent sessions. Three cyclists in the “force effectiveness” group (FEG) received feedback during training sessions of pedal force effectiveness. Three cyclists in the “peak force” group (PFG) received feedback only for peak normal force. Normal, anterior-posterior forces and pedal-to-crank angles were measured using instrumented pedals and potentiometers. Data were acquired during 4-km time trials to compute force effectiveness, peak normal force, resultant force, pedalling cadence and power output. Averages of the first two training sessions and averages of the last two training sessions were used for analysis of training effects and feedback type. There were large increases in right peak normal force for FEG (4%) and PFG (5%). Effectiveness of right pedal force was largely increased for right FEG (7%) and right PFG (9%). Conversely effectiveness of left pedal force was largely reduced for FEG (10%) and PFG (9%). Average power output substantially improved (10%, large difference) in FEG. Preliminary results indicate that force effectiveness training can be translated into increased right pedal force effectiveness and greater power output but not in better performance time during cycling time trials.

Introduction

Natural lower limb mechanics in cycling are characterised by coordinated flexion-extension of the hip, knee and ankle joints resulting in propulsive torque when the pedal is at the propulsive phase of crank revolution (from 12 o’clock to 6 o’clock) and resistive torque during the recovery phase of crank revolution (from 6 o’clock to 12 o’clock) (Kautz & Neptune, 2002). Pedal force effectiveness has been computed as the percentage of the total force applied to the pedal that results in propulsive torque (LaFortune & Cavanagh, 1983b). To maximise propulsive torque, cyclists are instructed to apply force in different directions during crank revolution. Figure 7.1 shows an example of “ideal” force application on the pedal to optimize pedal force effectiveness.

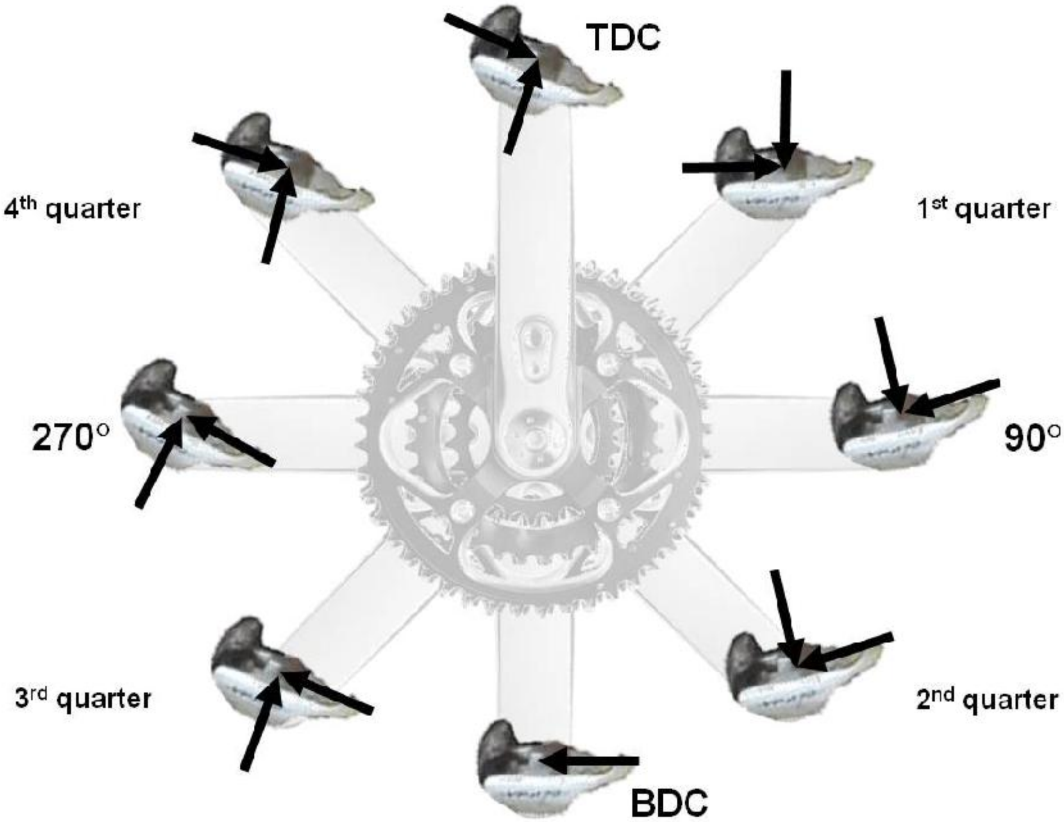


Figure 7.1. Illustration of "ideal" normal and anterior-posterior force components application on the pedal to optimize force effectiveness.

The success of pedal force feedback has commonly been evaluated using a reduction in resistive force at the 3rd and 4th quarters of pedal revolution (Sanderson & Cavanagh, 1990) and the difference between an "ideal" force and the measured force applied on the pedal (Hasson, et al., 2008).

Training can improve pedal force effectiveness and change the "natural" lower limb mechanics (Broker, et al., 1993; Hasson, et al., 2008; Holderbaum, et al., 2007; Mornieux & Stapelfeldt, in press; Sanderson & Cavanagh, 1990) by reducing the resistive crank torque and the total force applied to the pedal at the same workload. Greater activation of knee joint flexors and ankle dorsiflexors have been observed when pedal force effectiveness is improved (Mornieux, et al., 2008). Changes in muscle activation due to greater pedal force effectiveness have also been found to affect efficiency (Korff, et al., 2007; Mornieux, et al., 2008) and possibly performance in cycling.

Efficiency has been related to performance in endurance cycling (Hopker, et al., 2009) and theoretically reducing resistive force on the pedal may lead to higher efficiency. However, improving pedal force effectiveness has resulted in higher (Leirdal & Ettema, in press; Zameziati, et al., 2006) and lower efficiency in cycling (Korff, et al., 2007; Mornieux, et al., 2008). Studies that reported lower efficiency when cyclists aimed to improve force effectiveness during pedalling suggested that greater activation of hamstrings (Mornieux, et al., 2008) may result in additional energy expenditure compared to a "natural" pedalling action. Therefore, it is uncertain if cyclists should focus on improving pedal force effectiveness to reduce resistive force or if they should focus on producing high pushing force on the pedal when the crank is close to the 3 o'clock position due to the greatest moment arm. The latter would require greater force production from quadriceps muscle groups, which may be more efficient in trained cyclists (Coyle, et al., 1991).

Studies using pedal force feedback to improve pedal force effectiveness have been conducted using low exercise intensity (up to 80% of maximal oxygen uptake) and pedaling cadence (<80 rpm) (Broker, et al., 1993; Holderbaum, et al., 2007; Mornieux & Stapelfeldt, in press; Sanderson & Cavanagh, 1990). However, it is known that workloads higher than 75% of maximal aerobic does not affect pedal force effectiveness (Bini & Diefenthaeler, 2010) and higher pedaling cadence decreases pedal force effectiveness (Candotti, et al., 2007), which are the usual combination used by competitive cyclists (Lucia, et al., 2004). To date no studies have investigated the effects of training at race workload (>80% of maximal oxygen uptake) and cadence (>80 rpm) using visual feedback of pedal force effectiveness. To ascertain if using pedal force feedback would enhance force effectiveness and also result in better performance in cycling, we compared two types of pedal force feedback during training on 4-km cycling time trial performance. Two types of feedback were used: overall pedal force effectiveness during pedal revolution or peak normal force applied to the pedal. Our preliminary target was to assess the feasibility of using pedal force feedback during training at high intensity exercise (i.e. 4-km time trial) to improve pedal force effectiveness and performance in cycling.

Methods

Participants and allocation to pedal force feedback groups

Four male and two female athletes with competitive experience in cycling and in triathlon, ranked as "club riders" according to Ansley and Cangley (2009) were invited to participate in the study (mean ±SD: 28 ±7 years, 176 ±15 cm, 64.8 ±15 kg, 354 ±94 W maximal aerobic power output, and 60 ±6 $ml \cdot kg^{-1} \cdot min^{-1}$ of VO_{2Max}) and signed an informed consent form in agreement with the research ethics committee of the institution where the study was conducted.

Three cyclists were assigned to the "force effectiveness" group (FEG) that received visual feedback during training sessions of average pedal force effectiveness and a diagram indicating pedal force application in each quadrant. Three cyclists were assigned to the "peak force" group (PFG) (n = 3) that only received information on the average peak normal pedal force. Cyclists in each group were matched by similar performance (e.g. maximal aerobic power output and VO_{2Max}) and gender (e.g. one female to FEG and one female to PFG). The cyclists knew neither the nature of the other group's feedback nor the aims of the study. They were informed only that time trial training was going to be provided with use of pedal force feedback to help improve their performance and were asked to keep their regular cycling training. The choice for providing feedback of peak normal force applied on the pedal was to offer feedback that would not lead to substantial changes in pedal force effectiveness. The reason is that peak normal force occurs around 90° of crank revolution and would therefore be orientated perpendicular to the crank, influencing both effective and total forces proportionally (Coyle, et al., 1991).

Pre-training protocol

At the start of the first session body mass and height were measured according to ISAK protocols (Marfell-Jones, et al., 2006). Cyclists completed the Waterloo inventory to allow the determination of lower limb dominance (Carpes, et al., 2011). Cyclists' bicycle saddle height and horizontal position were measured to set up the stationary cycle ergometer (Velotron, Racemate, Inc). Cyclists performed an incremental cycling test to exhaustion with workload starting at 100 W for the first three minutes and increasing in steps of 25W each minute (Lucía, et al., 2002). Pedalling cadence was visually controlled by the cyclists at 90 ±2 rpm using the Velotron Coaching software 2008 (Velotron, Racemate, Inc). Gas exchanges were continuously sampled from a mixing chamber where samples were drawn into the oxygen and carbon dioxide analyzers for continuous measurement using a metabolic cart (TrueOne 2400, Parvo Medics, Salt Lake City, UT, USA). Prior to the test, the oxygen and carbon dioxide analyzers were calibrated according to the manufacturer's recommendations. Maximal aerobic power output and maximal oxygen uptake were defined as the highest power output measured during the test and as the highest oxygen uptake value computed over 15 s of data, respectively. After 10 minutes of rest, cyclists were familiarised with the 4-km time trial where they self-selected gear-ratio and pedalling cadence.

Training sessions

In the second session, the athletes performed two bouts of a 4-km time trial separated by 10 minutes of active rest on the bicycle. The 4-km time trial was used to elicit a maximal aerobic exercise effort under racing condition (i.e. change in pedalling cadence and gear ratio). During all time trials in the second session and the following seven sessions, normal and anterior-posterior forces were measured using a pair of strain gauge instrumented pedals (Candotti, et al., 2007), with pedal-to-crank angle measured using angular potentiometers. A reed switch attached to the bicycle frame detected the time that the crank passed the bicycle frame and was used to compute pedalling cadence. All data were acquired at 600 Hz by an analogue to digital converter (PCI-MIO-16XE-50, National Instruments, USA) using a custom Matlab (Mathworks Inc, MA) data acquisition script. Data were acquired for 10 s every 500 m of the time trial during the first two sessions. Feedback was provided for 5 s for the right pedal and then 5 s for the left pedal every 500 m in the first training session. Bilateral pedal force effectiveness, which is based on the ratio between the angular impulse of the effective (tangential) force on the crank and the resultant (total) force applied to the pedal surface, was computed along with bilateral average peak normal pedal force for every bout of feedback provided for five pedal revolutions.

During training sessions, feedback of forces on the right and left pedals was provided at distance points showed in Table 7.1 with a delay up to 15 s due to acquisition and processing of data using a custom made Matlab script. Frequency of feedback was reduced throughout the training period to minimize the participants dependence on the feedback and to improve the kinaesthetic sensory pathways responsible for fine-tuning pedal force application (Broker, et al., 1993) (see Table 7.1).

Table 7.1. Occurrence of force feedback during the 4-km time trial for the eight training sessions.

Distance points	**1st and 2nd sessions**	**3rd and 4th sessions**	**5th and 6th sessions**	**7th and 8th sessions**
500 m	**Yes**	**Yes**	No	No
1.0 km	**Yes**	No	**Yes**	No
1.5 km	**Yes**	**Yes**	**Yes**	No
2.0 km	**Yes**	**Yes**	No	**Yes**
2.5 km	**Yes**	**Yes**	**Yes**	No
3.0 km	**Yes**	No	No	**Yes**
3.5 km	**Yes**	**Yes**	**Yes**	No
3.8 km	**Yes**	**Yes**	No	No

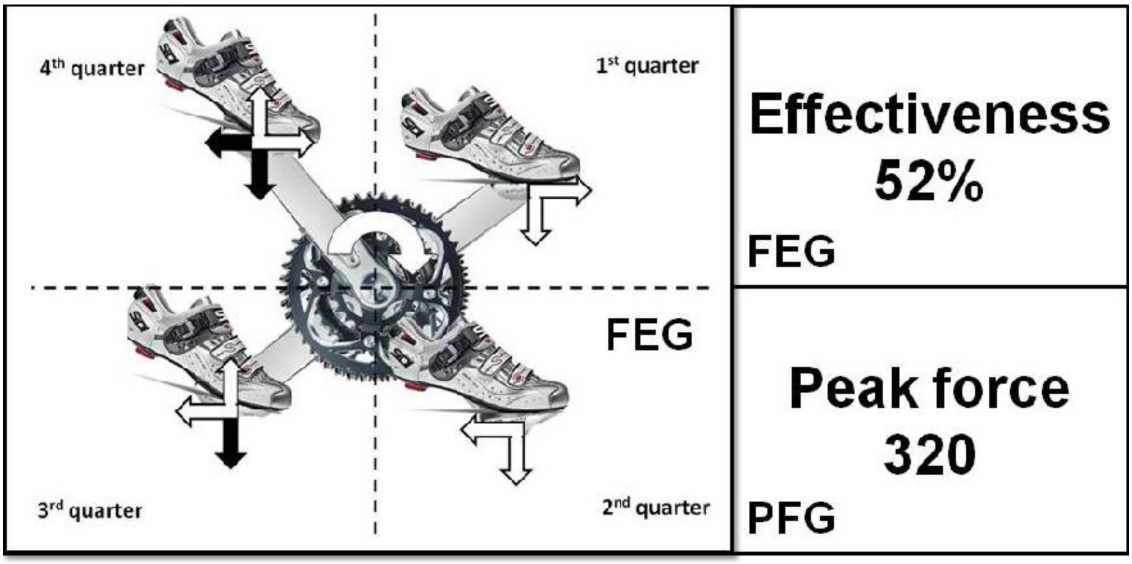


Figure 7.2. Example of the feedback screen shown to cyclists of the normal and anterior-posterior forces applied on the right pedal, effectiveness of pedal forces (%) and peak normal force (N). White arrows were presented when pedal force application resulted in propulsive torque, and black arrows were shown when pedal force resulted in resistive torque on the crank. Force diagram and force effectiveness value were shown only to the FEG and the peak force value was shown only to the PFG.

Data analyses

Off-line force data (peak normal force, resultant force and effectiveness of pedal forces) of both pedals were computed for five revolutions. Pedalling cadence was computed for each revolution from the time difference between each pulse of the reed switch signal. Force and pedalling cadence from each of the five revolutions were averaged for each of the eight distance points during the 4-km time trial. Force and pedaling cadence data from each revolution were then averaged for the first and second 4-km time trials of each session, resulting in eight values of force and pedaling cadence for each revolution. Average power output and 4-km time recorded by the Velotron Coaching software 2008 (Velotron, Racemate, Inc) for both of the 4-km time trials within each session were averaged.

Statistical analyses

Means and standard deviations were computed for each group and normalized by the results of the first training session for graphical presentation. To compare group responses to training, pedal force data were averaged for the first two training sessions (1st and 2nd) and for the last two training sessions (7th and 8th) for the right and left pedals and analysed using effect sizes. Cohen‘s effect sizes (ES) were computed for the analysis of magnitude of the differences between means and were rated as trivial (<0.25), small (0.25-0.5), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Lower limb symmetry assessed by the Waterloo inventory indicated only one cyclist from the FEG as left leg dominant (>60% of left leg preference in the inventory) with all other cyclists stating right leg dominance.

There was a trend for increases in right and left peak normal force, right and left resultant force and right pedal force effectiveness over the training sessions (see Figure 7.3).

There were large increases in right peak normal force for the FEG and PFG, followed by large changes in right resultant force for PFG and increases in right force effectiveness for both groups (see Table 7.2).

There was a large decrease in pedalling cadence for the FEG without substantial changes in power output or performance time (see Table 7.3).

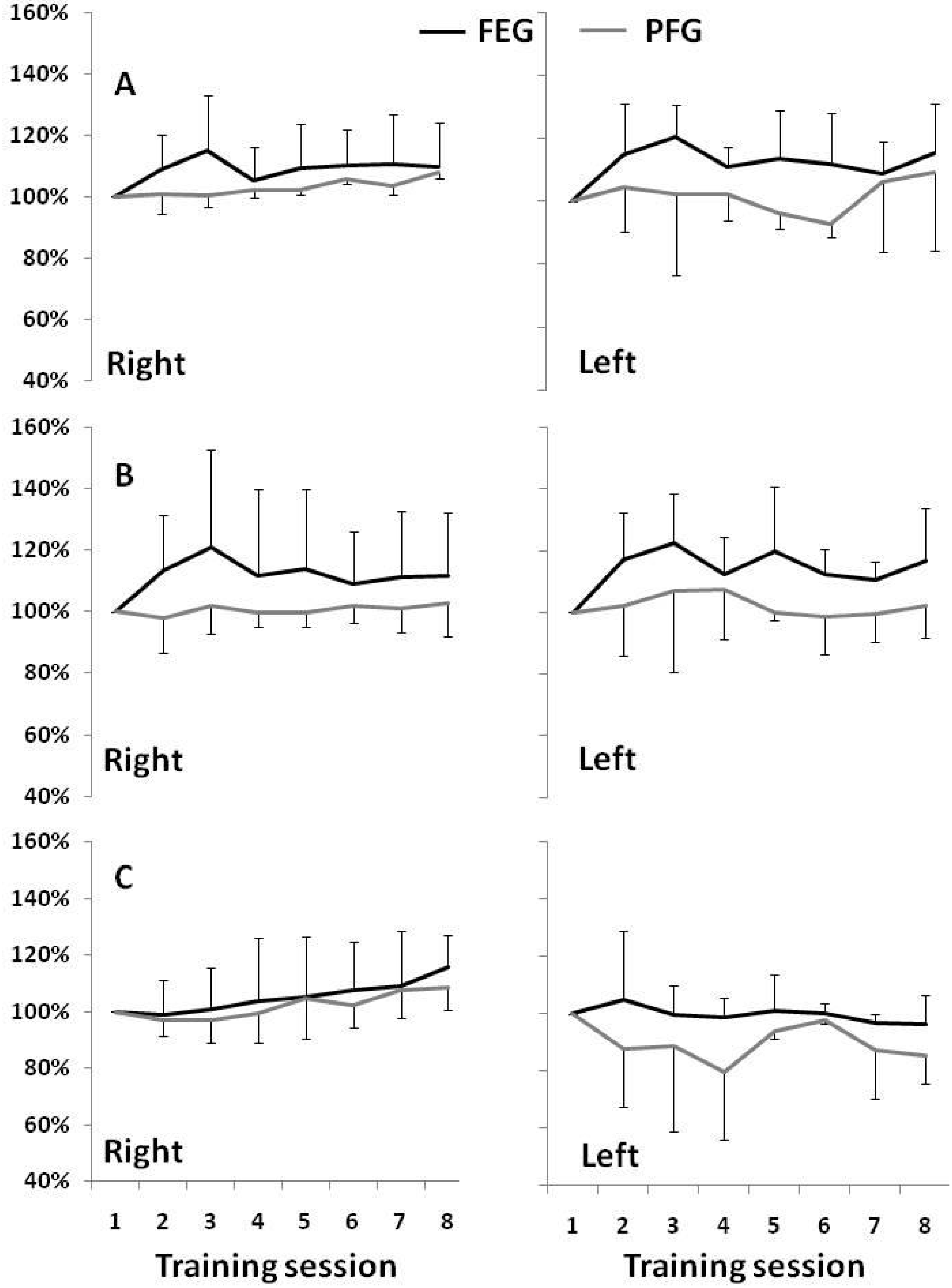


Figure 7.3. Means and standard deviations for force effectiveness group (FEG) and peak force group (PFG) for right and left peak normal force (A), resultant force (B) and effectiveness of pedal force (C) normalised by the results of the first training session. N = 3 for each group.

Table 7.2. Means and standard deviations and session change scores as percentages, and effect sizes and for peak normal force (NF), resultant force (RF) and force effectiveness (FE) for cyclists of FEG and PRG. Peak normal force and resultant force presented in Newtons and force effectiveness presented as % of linear impulse of resultant force. Abbreviations used are for forward (Fwd) and backward (back) positions on the saddle and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences are highlighted in bold italics.

	Force effectiveness group (FEG)			**Peak force group (PFG)**			**FEG vs. PFG**
	1&2 sessions	**7&8 sessions**	**Change (%; ES) between 1&2 and 7&8 sessions**	**1&2 sessions**	**7&8 sessions**	**Change (%; ES) between 1&2 and 7&8 sessions**	**Differences (%; ES) between 1&2 and 7&8 session changes for FEG versus PFG**
NF Right	338 ±77	352 ±58	***4%; 1.2, L***	349 ±123	367 ±129	***5%; 1.8, L***	1%; 0.4, S
Left	308 ±62	320 ±51	4%; 0.8, M	320 ±106	324 ±56	1%; 0.3, S	3%; 0.6, M
RF Right	152 ±43	156 ±32	2%; 0.5, M	159 ±58	161 ±50	***2%; 1.1, L***	<1%; 0.1, T
Left	139 ±22	145 ±16	4%; 0.7, M	148 ±72	145 ±67	1%; 0.5, M	***5%; 1.2, L***
FE Right	55 ±5	60 ±5	***7%; 2.2, L***	61 ±3	67 ±5	***9%; 2.1, L***	1%; 0.4, S
Left	48 ±13	44 ±7	***10%; 1.9, L***	44 ±5	40 ±3	***9%; 1.1, L***	1%; 0.2, T

Table 7.3. Means and standard deviations and session change scores as percentages, and effect sizes and for power output (W), pedalling cadence (rpm) and performance time (s) for participants between FEG and PRG. Abbreviations used are for forward (Fwd) and backward (back) positions on the saddle and effect sizes of trivial (T), small (S), moderate (M) and large (L).

Variables	**Force effectiveness group (FEG)**			**Peak force group (PFG)**			**FEG vs. PFG**
	1&2 sessions	**7&8 sessions**	**Change (%; ES) between 1&2 and 7&8 sessions**	**1&2 sessions**	**7&8 sessions**	**Change (%; ES) between 1&2 and 7&8 sessions**	**Differences (%; ES) between 1&2 and 7&8 session changes for FEG versus PFG**
Power output (W)	247 ±65	275 ±56	***10%; 1.0, L***	238 ±101	244 ±97	2%; 0.7, M	5%; 1.0, M
Pedaling cadence (rpm)	105 ±12	104 ±8	***1%; 1.0, L***	109 ±9	109 ±5	1%; 0.3, S	2%; 0.5, M
Performance time (s)	393 ±63	388 ±62	1%; 0.3, S	403 ±74	405 ±80	<1%; 0.3, S	2%; 0.4, S

Discussion

Our goal was to compare two types of pedal force feedback during 4-km time trial training on a bicycle ergometer. One group of cyclists received feedback of bilateral pedal force effectiveness and the other group received only feedback of bilateral peak normal pedal force. Our study provided preliminary evidence for the feasibility of using both types of feedback to enhance both effectiveness of pedal forces and peak normal force on the right pedal, but not on the left pedal during race type training (i.e. 4-km time trial). Differences in right and left pedal force variables may have been related to five of our six cyclists stating right leg dominance or to the order of feedback presentation. Our study improved previous designs (Mornieux & Stapelfeldt, in press) by conducting training with athletes with competitive experience in cycling and triathlon with a focus on improving cycling performance via pedal force feedback training during 4-km time trial training sessions. Only interventions with low exercise intensity (up to 80% of maximal oxygen uptake) and pedaling cadence (<80 rpm) have been reported previously (Broker, et al., 1993; Holderbaum, et al., 2007; Mornieux & Stapelfeldt, in press; Sanderson & Cavanagh, 1990), whereas our study provided four weeks (eight sessions) of training, high exercise intensity (maximal 4-km time trial) and high pedaling cadence (>100 rpm).

Both groups (FEG and PFG) presented large increases in peak normal force on the right pedal. However, only moderate and small increments in peak normal force on the left pedal were observed for the FEG and PFG, respectively. Both groups were aiming to improve performance during the training sessions, which may have required greater pedal force application to improve power output, as previously observed (Coyle, et al., 1991).

Surprisingly there were large improvements in pedal force effectiveness for the right pedal but large reductions for the left pedal for both groups (FEG and PFG). Cyclists of PFG were instructed to focus on normal downward force application on the pedal (peak normal force), which may have directed their attention to applying force at the most effective area of pedal revolution (i.e. close to 3 o'clock position of the crank). This hypothesis may explain the improvements in right pedal force effectiveness for the PFG. Right pedal feedback was provided first during all trials, which may have held the cyclists' attention to the right pedal and reduced their focus on the left pedal. Right pedal attention along with right lower limb dominance may explain the large improvement in force effectiveness for the right pedal. Training sessions were performed at maximal possible intensity for a 4-km time trial and high pedalling cadence, which may have reduced the cyclists' ability to concentrate on pedal force effectiveness and reduced the time to apply force on the pedal. Pedal force effectiveness is inversely related to pedalling cadence (Candotti, et al., 2007; Leirdal & Ettema, in press) and does not increase at workload levels higher than 75% of maximal power output (Bini & Diefenthaeler, 2010). Therefore, during a 4-km time trial, due to higher pedalling cadence and workload, cyclists are expected to present lower pedal force effectiveness compared to slower pedalling cadences and low workload levels.

Performance time was not substantially affected by the training program. Average power output increased over the eight sessions to a greater extent in the FEG compared to the PFG (5%). In addition, pedalling cadence decreased in the FEG across training sessions (1%) and increased in the PFG (1%). This suggests that cyclists from the FEG needed to reduce pedalling cadence in order to enhance power output via increased normal and resultant forces and via increases in pedal force effectiveness. Performance time was slightly reduced (<1%) in the PFG and slightly improved in the FEG (1%) which is contrary to prior evidence that improving pedal force effectiveness would reduce efficiency or performance in cycling (Korff, et al., 2007; Mornieux, et al., 2008). To improve pedal force effectiveness, cyclists need to direct the non-muscular component of pedal forces (e.g. inertial forces transferred to the pedals) into tangential crank force at the top and bottom dead centres (Korff, Fletcher, Brown, & Romer, in press). However, cyclists normally present large radial pedal force application during the 2nd and 3rd quarters of pedal revolution due to inertial and leg weight influences (Kautz & Hull, 1993). To change these components from radial to tangential force on the cranks, cyclists needed to recruit muscles from the flexor group (e.g. knee flexors and ankle dorsiflexors) which may be less efficient than the primary force generating muscles (e.g. hip and knee extensors) (Korff, et al., 2007). Cyclists from our study were capable of improving right pedal force effectiveness and power output (FEG only) through training, but these improvements were not substantial enough to translate into improvements in performance time. Differently, Theurel et al. (in press) observed that cyclists can sustain better performance during fatiguing cycling test when using pedal force effectiveness feedback. Better performance times may have been observed if cyclists had increased left pedal force effectiveness to the same extent as right pedal force effectiveness.

More training may have shown additional improvements given the trend over time for the right pedal force effectiveness profile to improve (increments in the final sessions compared to initial sessions). Single leg cycling has been positively used for training to improve force effectiveness (Hasson, et al., 2008), which could provide better results using a similar design to our study (e.g. high workload level and pedalling cadence).

Conclusion

Large increases in right normal and resultant pedal force application were observed for FEG and PFG. Large improvements in right pedal force effectiveness contrasted with large reductions in left pedal force effectiveness for FEG and PFG. Average power output was greater and pedalling cadence was slower in the FEG group and there was a small decrease in performance time. Preliminary results indicate that force effectiveness training can be translated into greater power output but not in better performance time during 4-km time trials. Future studies may give feedback from pedals presented in an alternating sequence.

CHAPTER 8: A COMPARISON OF SRM CRANKS AND STRAIN GAUGE INSTRUMENTED PEDAL MEASURES OF PEAK TORQUE, ANGLE OF PEAK TORQUE AND POWER OUTPUT

Overview

Our aim was to compare an SRM® torque analysis system with a strain gauge instrumented pedals system for right and left peak crank torque, crank angle of peak torque and power output. Seven competitive cyclists performed an incremental test to exhaustion on a stationary cycle ergometer equipped with an SRM® torque analysis system and a strain gauge instrumented pedals system (SGI pedals). The SRM® torque analysis system measured net torque while the SGI pedals measured the normal and anterior-posterior force applied on the pedal surface. Forces on the pedal surface were resolved into forces on the cranks (tangential and radial). Crank torque was measured by the pedals using the tangential force on the cranks and crank length. Power output was calculated from crank torque and angular velocity of the crank (calculated from pedalling cadence). All data were acquired between the 20th and the 40th seconds of each stage of the incremental test. Magnitudes of differences between outputs from the SGI pedals and the SRM® torque analysis system were assessed by effect sizes (ES). Power output was ~21% higher (ES = 1.0) for the SRM® torque analysis system than the SGI pedals. Peak torques were lower for the SRM® torque analysis system compared to the SGI pedals (right ~8%, ES = 1.5; left~7%, ES = 1.0). The angle of the right and left peak torque increased for the SRM® torque analysis system compared to the SGI pedals (right ~37□, ES = 3.5; left ~21□, ES = 1.7). The SRM® torque analysis system overestimated power output, underestimated peak torque and increased the angle of peak torque compared to the SGI pedals. Where possible a strain gauge instrumented pedals system should be used to measure variables contributing to cyclists performance rather than the SRM® torque analysis system.

Introduction

Cycling exercise intensity load monitoring has changed from the measurement of heart rate to power output because power output measurements are not affected by temperature or hemodynamic effects (Abbiss, et al., 2006; Vogt, et al., 2006). Therefore, the use of power meters in bicycles has become an important instrument during cycling training. Reliability of power output measurements of the most used device (SRM®) was confirmed by previous studies (Abbiss, et al., 2009; Gardner, et al., 2004). The SRM® power meter is a strain gauge instrumented crank set that measures the deformation on the shafts of the crank set resulting from the torque applied at the cranks. The SRM® power meter records net torque between right and left cranks and computes power output using reed switch triggers to measure pedalling cadence. The SRM® power meter has a head unit to measure power output, while the SRM® torque analysis system is an additional device that enables the acquisition of torque from the

SRM® power meter through an analog board with easier synchronization with other analog devices (e.g. reed switches for cadence measurement). The SRM® torque analysis system hardware converts the torque measures from Hz units used by the power meter controller to analog voltage. Aside from power output measurements, the SRM® torque analysis system has been used to compute peak torque on the crank for the analysis of bilateral symmetry (Carpes, et al., 2007a) and pedalling technique (Edwards, et al., 2009). However, no study has compared peak torque measurements from the SRM® torque analysis system to a strain gauge instrumented pedals system (SGI pedals). In this regard, pedals instrumented with strain gauges have been able to provide crank torque measurements independently for right and left legs, which are expected to offer a more reliable measure then the SRM® torque analysis system. Therefore a comparison of power output and peak crank torque between a SRM® torque analysis system and SGI pedals was our aim.

Methods

Participants

Four male and three female athletes with competitive experience in cycling and triathlon were invited to participate on the study (mean ±SD: 30 ±7 years old, 71.3 ±13 kg, 5.6 ±1.7 week training hours, 29.7 ± 9.4 $km \cdot h^{-1}$ average speed at self-selected racing events) and signed an informed consent form in agreement with the research ethics committee of the institution where the study was conducted.

Protocol

At the start of the evaluation session body mass and height were measured (Secca scales) and self-reported age, week training hours and average speed at self-selected racing events were recorded. Participants' saddle height and horizontal positions were measured to set up the stationary cycle ergometer (Velotron, Racemate, Inc). The athletes performed an incremental cycling exercise on the cycle ergometer with three minutes of warm-up at 100 W and pedaling cadence visually controlled at 90 ±2 rpm. Workload was then increased to 150 W and remained increasing in a step profile of 25 $W \cdot min^{-1}$ until athletes' exhaustion. A script was configured in the Velotron CS2008 software (Velotron, Racemate, Inc) for automatic control of the cycle ergometer workload in a constant load mode with cycle ergometer resistance changing to balance for fluctuations in pedalling cadence.

Data acquisition

Normal and anterior-posterior forces were measured using a pair of strain gauge instrumented pedals (SGI pedals) (Candotti, et al., 2007), with pedal-to-crank angle measured using angular potentiometers. The cycle ergometer was equipped with a science version of the SRM® power meter (Schoberer Rad Meßtechnik, Jülich, Germany) and the SRM® torque analysis system. The SRM® torque analysis system hardware converted the torque measures from Hz units used by the power meter controller to analog voltage. Pedal force data passed through an amplifier (Applied Measurements, Australia) and were recorded using an analog to digital board (PCI-MIO-16XE-50, National Instruments, USA) at 600 Hz per channel using Matlab (Mathworks Inc, MA). Data were acquired between the 20^{th} and the 40^{th} second of each step of 50 W (i.e. 100 W, 150 W, 200 W, 250 W and 300 W).

Data analysis

Forces on the pedal surface were resolved into forces on the cranks (tangential and radial) using angular potentiometers attached to pedal spindle for the measurement of the pedal-to-crank angle. Crank torque was measured by the SGI pedals using the tangential force on the cranks and crank length, and by the SRM® torque analysis system. The pedals were previously calibrated for normal and anterior-posterior force components applying known loads, and angular potentiometer were calibrated using a manual goniometer. The SRM® torque analysis system was also calibrated following procedures described by Wooles et al. (Wooles, Robinson, & Keen, 2006). A frequency to voltage conversion factor of 400e-6 and frequency to torque factor gathered at the calibration trial were used to convert torque measurements from voltage to N·m. A reed switch attached to the bicycle frame detected the position of the crank in relation to the pedal revolution and enabled separate torque data for every crank revolution. Power output was calculated from the effective force on the crank, crank arm length and angular velocity of the crank (calculated from pedaling cadence) for the SGI pedals and the SRM® torque analysis system. Peak crank torque and crank angle of peak torque were averaged over five complete pedal revolutions for both cranks for the SGI pedals and the SRM® torque analysis system.

Statistical analysis

Average and standard deviations for power output (W), peak crank torque (N·m) and the crank angle of the peak torque (°) for right and left cranks were calculated for the seven athletes and were compared between the SGI pedals and the SRM® torque analysis system as percentage differences for the five stages of the incremental test (100 W, 150 W, 200 W, 250 W and 300 W) using effect sizes (Hopkins, 2002; Knudson, 2009). Effect sizes were rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004).

Results

Ensemble results from the SGI pedals and the SRM® torque analysis system for five pedal revolutions for one representative cyclist showing smaller peak crank torque and delayed crank angle of the peak torque for the SRM® torque analysis system are shown in Figure 8.1

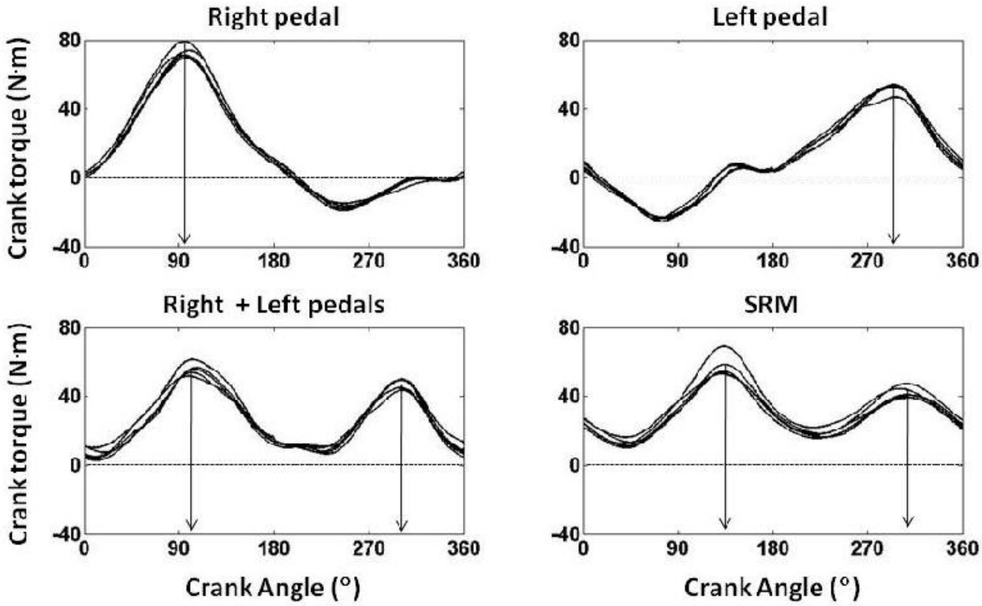


Figure 8.1. Crank torque measured by the pedals (right, left, and right + left) and by the SRM torque analysis system. Data are from five consecutive revolutions for one representative cyclist at 300 W of workload and 90 rpm of pedalling cadence. Arrows indicate peak torque and crank angle of peak torque for the strain gauge instrumented pedals and the SRM® torque analysis system.

Power output was ~21% higher (ES = 1.0) for the SRM® torque analysis system than the SGI pedals. Peak torques were lower for the SRM® torque analysis system (right crank ~8%, ES = 1.5; left crank ~7%, ES = 1.0) compared to the SGI pedals. The angle of the right and left peak torque were greater for the SRM® torque analysis system compared to the SGI pedals (right ~37°, ES = 3.5; left ~21°, ES = 1.7) (see Table 8.1).

Table 8.1. Average ±SD of power output (W), peak crank torque (N·m) and the crank angle of the peak torque (°) for the five stages of the incremental test (100 W, 150 W, 200 W, 250 W and 300 W) (N = 7). Percentage differences and effect sizes for comparisons between the SRM® torque analysis system and the SGI pedals.

Stages of the incremental test/ Variables	**100 W**	**150 W**	**200 W**	**250 W**	**300 W**
Power output – Pedals (W)	80 ±11	130 ±18	173 ±21	215 ±19	265 ±27
Power output – SRM (W)	111 ±55	161 ±61	205 ±62	259 ±69	343 ±73
Power output – Pedals vs. SRM (%; effect sizes)	31 ±6; 0.9, moderate	20 ±4; 0.8, moderate	16 ±4; 0.8, moderate	17 ±3; 0.9, moderate	26 ±2; 1.6, large
Right crank torque – Pedals (N.m)	32 ±4	41 ±6	49 ±7	56 ±9	67 ±4

Right crank torque – SRM (N.m)	16 ±7	25 ±10	35 ±13	45 ±16	65 ±17
Right crank torque – Pedals vs. SRM (%; effect sizes)	17 ±1; 2.9, large	10 ±7; 1.9, large	7 ±6; 1.5, large	4 ±6; 0.8, moderate	1 ±7; 0.2, trivial
Left crank torque – Pedals (N.m)	33 ±10	39 ±12	42 ±14	47 ±12	49 ±9
Left crank torque – SRM (N.m)	16 ±6	23 ±9	29 ±10	40 ±11	51 ±15
Left crank torque – Pedals vs. SRM (%; effect sizes)	16 ±15; 2.0, large	11 ±13; 1.5, large	6 ±11; 1.1, large	3 ±8; 0.6, moderate	1 ±7; 0.2, trivial
Right crank angle of peak torque – Pedals (°)	96 ±10	94 ±9	92 ±7	91 ±8	93 ±3
Right crank angle of peak torque – SRM (°)	138 ±22	136 ±13	123 ±20	127 ±10	127 ±11
Right crank angle of peak torque – Pedals vs. SRM (%; effect sizes)	29 ±12; 2.6, large	31 ±8; 3.9, large	24 ±10; 2.2, large	28 ±6; 3.9, large	26 ±6; 4.8, large
Left crank angle of peak torque – Pedals (°)	290 ±14	287 ±19	287 ±16	288 ±9	285 ±10
Left crank angle of peak torque – SRM (°)	317 ±16	313 ±13	307 ±15	306 ±10	304 ±8
Left crank angle of peak torque – Pedals vs. SRM (%; effect sizes)	8 ±3; 1.7, large	8 ±5; 1.6, large	6 ±4; 1.2, large	6 ±4; 1.8, large	6 ±2; 2.1, large

SRM = SRM® torque analysis system; Pedals = strain gauge instrumented pedals system.

Discussssion

Cycling training monitoring variables such as power output and peak crank torque have not been previously compared between instrumented pedals and the SRM® torque analysis system. We found ~21% greater power output for the SRM® torque analysis system compared to the strain gauge instrumented pedals. The benefit of the SRM® torque analysis system is it enables easier synchronization of the net torque between pedals measured by the SRM® power meter and any analog devices. However, lower results of peak torque for both cranks and delayed crank angle of the peak torque compared to instrumented pedals may suggest that the torque analysis system does not provide the most accurate measure of crank torque. One possible explanation is based on the hardware design of the SRM® torque analysis system, which converts torque measures from frequency to voltage unit, instead of the frequency measurements used by the power control unit from the SRM® power meter. This frequency to voltage conversion may be affected by aliasing effects from hardware set-up (i.e., low pass filter in voltage signal output), which may reduce peak voltage readings. A second explanation may be that the SRM® power meter is limited to measure the net torque applied by the right and left legs, differently from the instrumented pedals. A lower estimative in peak crank torque may affect the identification of torque symmetry or accurate peak crank torque in the evaluation of cycling performance. Force symmetry assessed by measuring the total force applied on the pedals cannot be computed using the SRM® torque analysis system. Delayed estimates of crank angle of the peak torque may lead to wrong assumptions in pedalling technique (i.e., delayed pedal force application).

Conclusion

Where possible a strain gauge instrumented pedals system should be used to measure variables contributing to cyclists performance rather than the SRM® torque analysis system.

CHAPTER 9: BILATERAL ASYMMETRY ASSESSMENT IN CYCLING USING SRM CRANKS AND INSTRUMENTED PEDALS

Overview

There is increasing use of the SRM® torque analysis system for symmetry assessment, but this device may underestimate peak crank torque and lower limb asymmetry. We compared peak torque symmetry from right and left cranks using the SRM® torque system and a pair of instrumented pedals during bilateral cycling. Ten competitive cyclists performed an incremental cycling test to exhaustion. Forces and pedal to crank angles were measured using right and left instrumented pedals along with crank torque using the SRM® torque analysis system. Lower limb dominance was assessed using the Waterloo inventory. Raw differences in right and left peak torque and asymmetry index were used to assess differences in peak torque from right and left legs. Greater peak torques for right (16-7%) and left (11-5%) cranks were observed for the instrumented pedals compared to the SRM® torque system between 100 W and 250 W. There was a trend for an increase in differences between right and left crank torque as workload increased using the SRM® torque system (7-33%) and the instrumented pedals (9-66%), but large differences were only found for the instrumented pedals at workloads higher than 200 W. Lower limb asymmetries in peak torque increased at higher workload levels in favour of the dominant right leg. Limitations in design of the SRM® torque analysis system may preclude the use of this system to assess crank torque symmetry.

Introduction

Bilateral cycling motion has usually been assessed assuming symmetry in force production and kinematics of lower limbs. However, differences in power output, and mechanical work of the legs have ranged from 5% to 20% in uninjured cyclists and non-cyclists (Carpes, Mota, & Faria, 2010). Conflicting results were reported comparing cyclists (Bini, et al., 2007; Carpes, et al., 2007a, 2007b; Smak, et al., 1999) and non-cyclists (Daly & Cavanagh, 1976) without clear relationships between pedalling cadence (Smak, et al., 1999) and workload level (Daly & Cavanagh, 1976) in bilateral symmetry.

Peak torque at the propulsive phase of crank revolution (i.e. from 12 o'clock to 6 o'clock crank positions) has been reported as one of the most important predictors of performance during 40-km time trials (Coyle, et al., 1991) given a large percentage of the force applied to the pedal in the sagittal plane can be translated into crank toque in this part of the crank revolution (Sanderson, 1991). Therefore, cyclists should aim for large crank torque application on both cranks to enhance power output for a given pedalling cadence. Using peak torque as a measure of pedalling symmetry, authors have reported that differences between legs were significant at lower workload levels (<90% of maximal oxygen uptake) and decreased at higher workload levels for six competitive cyclists (Carpes, et al., 2007a, 2007b). In contrast, another study did not show substantial differences in mean torque computed during full crank revolution for eleven cyclists at different workload levels (60-100% of maximal oxygen uptake) (Bini, et al., 2007). Therefore, it is unclear if torque symmetry is related to workload level. The potential reduction in asymmetries in torque for higher workload levels may be due to an increased bilateral neural input by inter-hemispheric cortical communication to facilitate the excitability of both legs (Carpes, Mota, et al., 2010).

Evaluation of bilateral asymmetry has increased because some commercial devices provide right to left crank comparisons for torque and power output. One example is the SRM® torque analysis system which enables the user to assess right and left crank torque during cycling (Barratt, 2008). However, a recent study showed that the measures of peak torque from the SRM® torque analysis system are only accurate for workloads greater than 80% of maximal aerobic power output (Bini, Hume, & Cervieri, 2011). The SRM® power meter (Schoberer Rad Meßtechnik, Jülich, Germany) measures the deformation on the shafts of the crank set resulting from the torque applied on both cranks (i.e. net crank torque). Therefore, separate measures of right and left crank torque are not possible using this device because the torque at right and left cranks are computed as a net torque (i.e. torque from the contralateral leg diminishes torque from the ipsilateral leg). Using the SRM® torque analysis system, it has been assumed that peak torque observed during the propulsive phase of crank revolution are exclusively affected by the ipsilateral leg (e.g. right leg) (Carpes, et al., 2007a), which may not be completely valid. Consequently, the accuracy of bilateral symmetry assessment using the SRM® torque analysis system may be compromised by the design of this device.

Pedals instrumented with strain gauges have been able to provide crank torque measurements independently for right and left legs (Hull & Davis, 1981), which are expected to offer a more accurate measure of peak crank torque then the SRM® torque analysis system. Our study compared peak torque for right and left lower limbs measured by the SRM® torque analysis system and by instrumented pedals.

Methods

Participants

Ten cyclists (three female and seven male) with competitive experience in cycling and/or triathlon were invited to participate in the study. Cyclists' (mean ±SD: 30 ±7 years, 72.8 ±13 kg, 175 ±12 cm, 55.6 ±8.8 $ml \cdot kg^{-1} \cdot min^{-1}$ maximal oxygen uptake, 336 ±77 W peak power output, 4.6 ±6 $W \cdot kg^{-1}$ peak power per body mass) signed an informed consent form in agreement with the committee of ethics in research of the institution where this study was conducted. No cyclist had an injury that would impact on test performance at the time of data collection.

Data collection

Pedal force components (normal and anterior-posterior) were computed using the regression between three static load points (0 kg, 5 kg and 10 kg) applied to the pedals and voltage output when R^2 was greater than 0.99. Mechanical coupling between anterior-posterior and normal loads were corrected using a gain matrix (Leirdal & Ettema, 2011). Potentiometers were calibrated using a manual goniometer set at four angles (0°, 90°, 180° and 270°) to compute the relationship between voltage output and the measured angle. The calibration factors were defined when mean differences in voltage were lower or equal to 1%.

Body mass and height were measured according to International Society for the Advancement of Kinanthropometry protocols (Marfell-Jones, et al., 2006). Right (or left) leg dominance was determined when frequency of leg preference was greater than 50% using the Waterloo inventory (Carpes, et al., 2011). Cyclists' bicycle saddle height and horizontal position were measured to set-up the stationary cycle ergometer (Velotron, Racemate, Inc). The cyclists performed an incremental cycling exercise on the cycle ergometer with three minutes of warm-up at 100 W and pedaling cadence visually controlled at 90 ±2 rpm. Workload was then increased to 150 W and remained increasing in a step profile of 25 $W \cdot min^{-1}$ until cyclists' exhaustion (Lucía, et al., 2002). A script was configured in the Velotron CS2008 software (Velotron, Racemate, Inc) for automatic control of the cycle ergometer workload in a constant workload mode. This configuration enabled a constant workload with cycle ergometer resistance changing to balance for fluctuations in pedalling cadence. Gas exchanges were continuously sampled from a mixing chamber where samples were drawn into the oxygen and carbon dioxide analyzers for continuous measurement using a metabolic cart (TrueOne 2400, Parvo Medics, Salt Lake City, UT, USA). Analyzers for oxygen and carbon dioxide were calibrated according to manufacturer recommendations. Maximal aerobic workload and maximal oxygen uptake were defined as the highest workload measured during the test and as the highest oxygen uptake value computed over a 15 s average of the data, respectively. After two to seven days, cyclists returned to the laboratory at the approximate same time of the day to perform the incremental test following the same procedures.

Normal and anterior-posterior forces were measured using a pair of strain gauge instrumented pedals (Candotti, et al., 2007), with pedal-to-crank angle measured using angular potentiometers attached to the pedal spindle. Pedal force data passed through an amplifier (Applied Measurements, Australia) and, along with potentiometers and reed switch signals were recorded using an analogue to digital board PCI-MIO-16XE-50 (National Instruments, USA) at 600 Hz per channel using a custom made script in Matlab (Mathworks Inc, MA). Analogue data were acquired between the 20^{th} and the 40^{th} s of each step of 50 W (i.e. 100 W, 150 W, 200 W, 250 W, etc).

Data analyses

Pedal-to-crank angle measured by the potentiometers were converted into sine and cosine to compute tangential and radial forces on the cranks. A low pass zero lag Butterworth digital filter with cut off frequency of 10 Hz was applied to the sine and cosine data from potentiometers to attenuate signal noise from the gap in potentiometer voltage readings (Hull & Davis, 1981). Crank torque was measured by the pedals using the tangential force on the cranks and crank length, and by the SRM® torque analysis system. A frequency to voltage conversion factor of 400e-6 and frequency to torque factor gathered at the calibration trial were used to convert torque measurements from voltage to N.m.

A reed switch attached to the bicycle frame detected the position of the crank in relation to the pedal revolution and enabled separate pedal forces and torque data for every crank revolution and for the propulsive (i.e. from 12 o'clock to 6 o'clock crank positions) and recovery phases (i.e. from 6 o'clock to 12 o'clock crank positions) for the right and left cranks. Peak crank torque of right and left cranks were determined when the crank was at the propulsive phase and at the recovery phase, respectively, using a clockwise motion of the crank as reference. Peak crank torque was averaged over five complete pedal revolutions for each crank on the instrumented pedals and SRM® torque analysis system. Asymmetry index was calculated as outlined by Robinson et al. (Robinson, Herzog, & Nigg, 1987) (see equation 9.1).

$$AI\% = [(R - L)/AVG(R, L)]$$

Equation 9.1. Asymmetry index (AI%) computed using measures from right (R) and left (L) legs normalized by the average (AVG) of right and left measures.

Statsitical analyses

Peak torque of right and left cranks and asymmetry index (mean and SD) were compared for instrumented pedals and the SRM® torque analysis system. All variables were analyzed for the 100 W, 150 W, 200 W, 250 W, 300 W and 350 W workloads of the incremental test. Normality of distribution and sphericity were confirmed for all variables via the Shapiro-Wilk and Mauchly tests, respectively, after application of a logarithmic transformation using SPSS for Windows 16.0 (SPSS, NY, USA).

Mean percentage differences between right and left peak torques and the asymmetry index from the SRM® torque analysis system and the instrumented pedals were computed and comparisons were conducted using Cohen's effect sizes (ES). Cohen's effect sizes were rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Lower limb symmetry assessed by the Waterloo inventory indicated all ten cyclists were right leg dominant.

Ensemble results from the instrumented pedals and the SRM® torque analysis system for five pedal revolutions for one representative cyclist are shown in Figure 9.1. Greater peak torque was observed comparing individual instrumented pedal measures for right (51 N.m) and left (41 N.m) pedals than from the SRM® torque analysis system (39 N.m for right crank and 32 N.m for left crank). Crank torque derived from right and left pedals (net torque) showed similar magnitudes to the measurements from the SRM® torque analysis system, illustrating ispilateral to contralateral legs effects in crank torque.

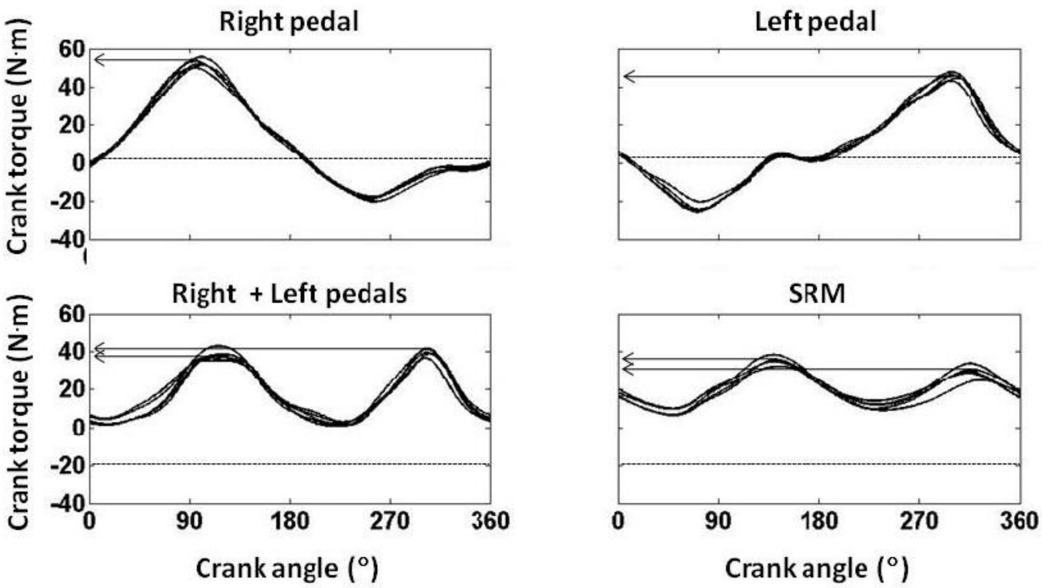


Figure 9.1. Crank torque measured by the pedals (right, left, and right + left) and by the SRM torque analysis system. Data from five consecutive revolutions of one cyclist at 200 W of workload and 90 rpm of pedalling cadence. Arrows indicate peak crank torque.

In general, large differences for right (16-7%) and left (11-5%) peak torque between the SRM® torque analysis system and the instrumented pedals were observed between 100 W and 250 W. At higher workloads (300 W and 350 W) there were moderate to trivial differences in peak crank torque. There was a trend for an increase in the difference between right and left crank torques using the SRM® torque analysis system (8-33%) and the instrumented pedals (5-66%), but large differences were only found for the instrumented pedals at workloads higher than 200 W. The instrumented pedals presented larger asymmetry indices compared to the SRM® torque analysis system at 250 W (see Table 9.1).

Table 9.1. Means and standard deviations, mean percentage differences and effect sizes of peak crank torque, differences between right and left crank torque and asymmetry index for the five workloads of the incremental test comparing both systems (SRM® torque analysis system and the instrumented pedals) for data from 10 cyclists. Abbreviations used are for right (R-pedal) and left pedals (L-pedal) and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences are highlighted in bold italics.

Workload level		**100 W (n = 10)**	**150 W (n = 10)**	**200 W (n = 10)**	**250 W (n = 10)**	**300 W (n = 7)**	**350 W (n = 6)**
Right peak torque (N.m)	SRM®	17 ±6	23 ±8	31 ±11	39 ±14	53 ±19	73 ±25
	R-Pedal	33 ±5	41 ±6	48 ±7	56 ±8	65 ±8	75 ±7
	SRM® vs.	***16%;***	***12%;***	***9%;***	***7%;***	4%;	1%
	R-Pedal	***3.0, L***	***2.4, L***	***2.0, L***	***1.6, L***	0.9, M	0.1, T
Left peak torque (N.m)	SRM®	18 ±8	22 ±7	28 ±8	36 ±9	45 ±11	55 ±18
	L-Pedal	30 ±9	34 ±10	39 ±8	43 ±12	46 ±12	46 ±14
	SRM® vs.	***11%;***	***8%;***	***5%;***	3%;	<1%;	3%;
	L-Pedal	***1.4, L***	***1.4, L***	***1.3, L***	0.7, M	0.1, T	0.6, M
Right vs. left peak torque	SRM	7%;	5%;	10%;	8%;	17%;	33%;
		0.2, T	0.1, T	0.3, S	0.3, S	0.5, M	0.8, M
	Pedals	9%;	19%;	***26%;***	***29%;***	***42%;***	***66%;***
		0.4, S	0.8, M	***1.3, L***	***1.3, L***	***1.9, L***	***2.8, L***
Asymmetry index (%)	SRM®	6 ±17	4 ±15	8 ±17	5 ±15	13 ±20	27 ±18
	R-Pedal	11 ±28	20 ±33	22 ±30	28 ±31	36 ±33	51 ±36
	SRM® vs.	280%;	418%;	181%;	***428%;***	189%;	93%;
	R-Pedal	0.8, M	0.7, M	0.6, S	***1.0, L***	0.9, M	0.9, M

Discussion

Our study compared peak crank torque of right and left legs measured by the SRM® torque analysis system and instrumented pedals during bilateral cycling. The reason for this comparison was based on the increasing use of the SRM® for symmetry assessment (Carpes, et al., 2007a, 2007b) and potentially because it would underestimate peak crank torque and asymmetry analysis (Bini, Hume, & Cervieri, 2011). Lower peak torque was observed for the SRM® torque analysis system compared to the instrumented pedals. Differences in right and left peak torque measured between systems varied from 1% to 16% and from <1% to 13%, respectively. Greater asymmetries were observed at higher workload levels but large differences between right and left crank torque were only observed using the instrumented pedals. The asymmetry index was greater than 20% (usually reported for uninjured cyclists (Carpes, Mota, et al., 2010)) for the SRM® torque analysis system only at 350 W and for the instrumented pedals at workloads greater than 250 W. Large standard deviations were observed for asymmetry indices using the SRM® torque analysis system and the instrumented pedals due to between-cyclists high variability in peak crank torque of right and left legs. The primary reason for these differences between systems is related to the electronic characteristics of each system. The SRM® torque analysis system, as outlined in Figure 9.2, is designed to measure the deformation on the shafts of the crank set due to the torque applied on both cranks.

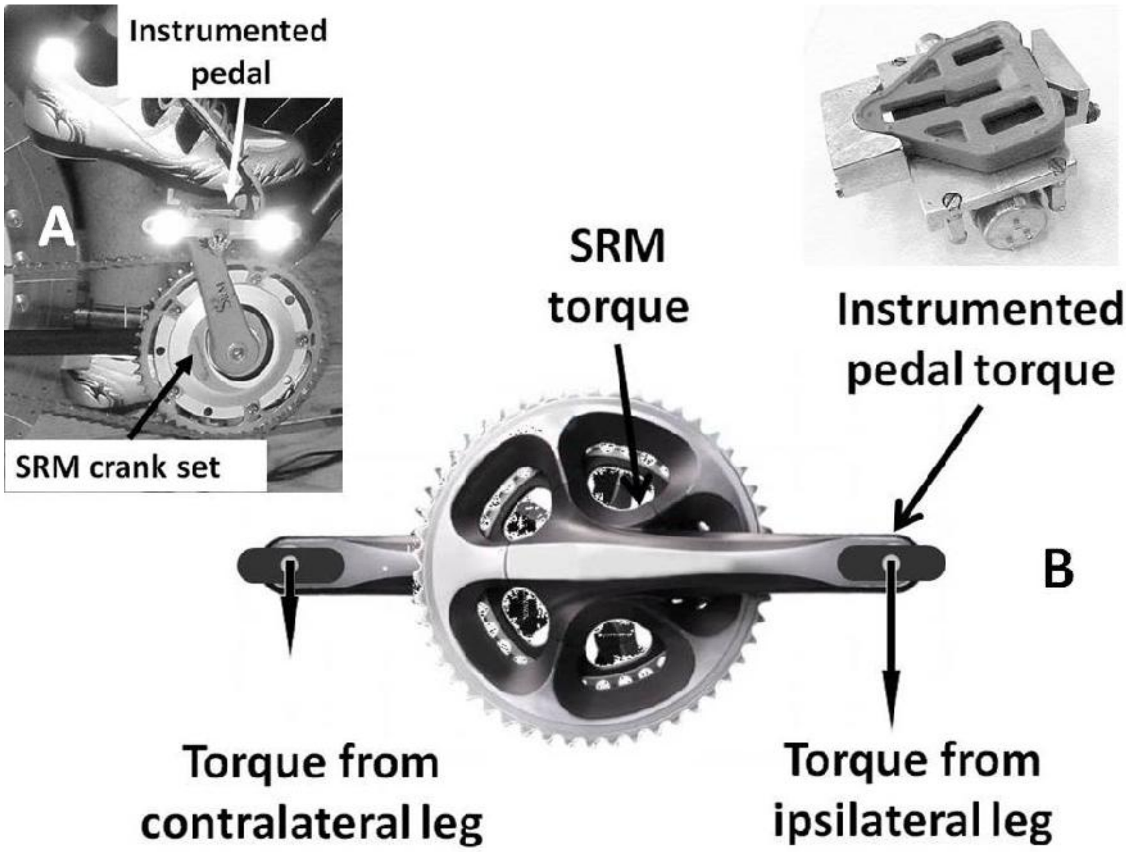


Figure 9.2. Image of the right instrumented pedal attached to the SRM® crank set (A). Illustration of the locations of sensors for crank torque measurement for the SRM® torque analysis system and instrumented pedals. Arrows indicate crank torque applied simultaneously by the ipsilateral and contralateral legs (B).

Using the example highlighted in Figure 9.2, if one cyclist applied 20 N.m of torque (clockwise) with the ipsilateral leg and 5 N.m of torque (anticlockwise) with the contralateral leg, the SRM® torque analysis system would record 15 N.m of torque. Therefore, the torque at right and left cranks is computed as a net torque (i.e. torque from the contralateral leg diminishes torque from the ipsilateral leg). Instrumented pedals measure the force on the pedal surface (e.g. normal and anterior-posterior components) and compute crank torque independently using pedal-to-crank angles (Hull & Davis, 1981). This is a more accurate approach because forces from each leg are measured separately, reducing contralateral to ipsilateral effects on crank torque measures.

Differences in peak crank torque measured by the SRM® torque analysis system and instrumented pedals were large for right and left legs. Large asymmetries were observed only using the instrumented pedals at workloads higher than 200 W. Therefore, assessments of bilateral asymmetries (Carpes, et al., 2007a) are not valid using the SRM® torque analysis system because this device was not capable of detecting substantial differences of 33% between right and left peak torques as measured by the instrumented pedals.

Higher workloads resulted in greater asymmetries which are contrary to previous findings (Bini, et al., 2007; Carpes, et al., 2007a, 2007b). Cyclists were observed to reduce differences in peak torque at higher workloads (Carpes, et al., 2007a, 2007b), however, peak torque was measured using the SRM® torque analysis system in these studies. Another study did not observe effects from workload level in crank torque symmetry in cyclists using instrumented pedals (Bini, et al., 2007). Reductions in asymmetries in torque for higher workloads have been hypothesized due to a potential increased bilateral neural input by inter-hemispheric cortical communication to facilitate the excitability of both legs (Carpes, Mota, et al., 2010). However, studies assessing muscle activation in cyclists during bilateral cycling exercise with increasing workloads did not report differences in lower limb muscle activation of both legs (Carpes, et al., 2011). Muscle activation during single leg cycling did not differ when cyclists right and left legs were assessed (Carpes, Diefenthaeler, et al., 2010), which suggests that lower limb neural drive may not differ between legs. Substantial differences in cycling efficiency have not been observed in cyclists during single leg cycling, suggesting that contributions from independent legs to efficiency are similar.

Lower limb dominance may have played a role in increasing asymmetries at higher workloads. Evidence suggests that the kicking dominant leg contributed significantly more to average crank power than the non-dominant leg (Smak, et al., 1999), which is in line with our results. The dominant leg may receive greater neural drive at higher workloads and/or fatigue states. Studies (Daly & Cavanagh, 1976; Smak, et al., 1999) have indicated that pedalling asymmetries are highly variable among cyclists and individual cyclists may exhibit different changes in asymmetry depending on exercise condition. Further research should assess bilateral muscle activation and joint kinetics in cyclists with similar levels of crank torque asymmetry to ascertain individual joint contributions to crank torque for right and left legs.

Some limitations may have affected the results of our study. Trials using single leg cycling could have been conducted to isolate the influence of the contralateral leg in torque readings from the SRM® torque analysis system. During single leg cycling, torque measured by the SRM® torque analysis system will not be affected by contralateral leg therefore crank torque measures would be isolated to the ipsilateral leg (Ting, et al., 1998). This would shed light on the effects of contralateral resistive torque on the SRM® torque analysis system data.

Conclusion

Greater peak torques for right and left cranks were observed for the instrumented pedals compared to the SRM® torque analysis system between 100 W and 250 W. Substantial differences in right and left peak torque could only be assessed using instrumented pedals, with increase in asymmetry observed at higher workloads in favour of the dominant leg. Limitations in the SRM® torque analysis system may preclude the use of this system to assess crank torque symmetry. Whenever possible, instrumented pedals should be used for torque and limb asymmetry assessment.

CHAPTER 10: KNEE JOINT MODELLING FOR CYCLING

Overview

Existing models to compute knee forces in cycling have not taken into account patellar ligament and quadriceps tendon imbalances and contributions of muscle forces to tibiofemoral forces. Therefore we proposed a model for computation of patellofemoral compressive, tibiofemoral compressive and tibiofemoral anterior-posterior forces including these parameters. Knee anatomic characteristics were gathered from literature. Right pedal force and lower limb joint kinematics were acquired for 12 competitive male cyclists during trials at two different levels of workload (maximal and ventilatory threshold) at 90 rpm and one trial at 70 rpm of pedalling cadence at the ventilatory threshold. Patellofemoral compressive force was smaller (15%) at the workload of the second ventilatory threshold compared to the maximal workload and at the 90 rpm trial (35%) compared to the 70 rpm trial. Compressive normal force at the tibiofemoral joint and anterior force at the tibiofemoral joint were not largely affected by changes in workload or pedalling cadence. Compared to previous studies, force magnitudes normalized by workload were larger for patellofemoral (mean = 19 N/J; percentage differences to other studies = 20-45%), tibiofemoral compressive (7.4 N/J; 20-572%) and tibiofemoral anterior (0.5 N/J; 60-200%) forces. Differences in model design, testing condition and experience in cycling may affect prediction of knee joint forces. Patellar tendon to quadriceps force ratio and the contribution of muscle forces to tibiofemoral joint forces have been our additions compared to previous published models.

Introduction

Forces acting on the knee joint have been researched for different kinds of movements. Closed and open kinetic chain movements involving the knee joint have been analyzed (Ericson & Nisell, 1986; Escamilla, 2001; Matthews, Sonstegard, & Henke, 1977). During closed kinetic chain movements, ground reaction forces and joint kinematics have been used to understand the contact forces on the tibiofemoral and patellofemoral joints (Escamilla, 2001). Concern regarding overload of the soft tissue of these joints (e.g. patellofemoral cartilage) has given support to this type of research.

In cycling, low levels of force were found at the tibiofemoral (0.5-1.14 times body weight) and the patellofemoral joints (~2.35 times body weight), compared to walking (2 times body weight) and running (>2 times body weight), because the body is aligned over the bicycle saddle during seated cycling (Ericson, Bratt, & Nisell, 1986; Ericson & Nisell, 1987). However, the great number of repetitions performed during bicycle riding may increase the risk of knee joint overuse injuries due to repetitive stress on the soft tissues of the tibiofemoral and patellofemoral joints (Schwellnus & Derman, 2005). Clinical reports indicate that anterior knee pain, potentially from condromalacea patellofemoral, is the most common non-traumatic injury in cyclist (Holmes, et al., 1994). Therefore, measuring forces acting on the knee is important during cycling to help predict risk factors of overuse injuries in cyclists (Neptune & Kautz, 2000). Compressive forces on the patellofemoral joint due to excessive knee flexion resulting from low saddle heights may result in anterior knee pain, which is the mostly reported injury in cyclists (Bailey, et al., 2003).

Measurements of joint contact forces on the knee are rare during cycling (Fleming, et al., 1998; Kutzner, et al., 2008). Most studies have used mathematical models of the knee to predict joint forces (Bressel, 2001; Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011). Most models have used pedal reaction forces and joint kinematics to compute patellofemoral compressive forces using inverse dynamics (Bressel, 2001; Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011). Others have included muscle mechanics (e.g. force-velocity-length relationship) and muscle activation profiles to estimate muscle forces (Neptune & Kautz, 2000). For a clinical perspective, pedal force components can be measured using commercial instrumented pedals (Stapelfeldt, et al., 2007) along with video from the sagittal plane using high speed cameras commercially available. Inverse dynamics can be employed using models provided by van den Bogert (http://isbweb.org/data/invdyn/index.html). The computation of joint moments is possible and individual assessment of cyclists may provide important information. However, a step-by-step approach to gather pedal forces and joint kinematics to provide patellofemoral and tibiofemoral joint forces for clinical application has not been presented. Patellofemoral and tibiofemoral forces have been modelled without sufficient information on knee anatomy. Teaching may be improved once friendly user models are presented with enough details. Existing models available at the International Society of Biomechanics website (van den Bogert & de Koning, 1996) are limited to the computation of joint moments. Previous models to compute patellofemoral force have not accounted for imbalances in quadriceps to patellar tendon force during knee extension (Bressel, 2001) and models to calculate tibiofemoral force have not accounted for muscle force effects on joint forces (Tamborindeguy & Bini, 2011). Consequently, a method addressing the aforementioned limitations in calculations of patellofemoral and tibiofemoral forces during cycling is needed.

We have proposed a method to compute patellofemoral and tibiofemoral joint forces in cycling using inverse dynamics based on pedal forces and joint kinematics data. A step-by-step procedure is presented accounting for quadriceps to patellar tendon force imbalance in the patellofemoral force and accounting for muscle force effects on tibiofemoral joint forces, thereby improving previous models used in cycling (Bressel, 2001; Ericson, et al., 1986; Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011). The effects of changes in workload and pedalling cadence on patellofemoral and tibiofemoral joint forces were assessed to illustrate the sensitivity of the model.

Methods

Participant's characteristics

Twelve competitive road cyclists (28 ±6.6 years, 71 ±6.8 kg of body mass, 177 ±9.7 cm of stature, 372 ±71 km/week of cycling training, 377 ±30 W of maximal aerobic workload, 64 ±5.2 ml/kg/min of maximal oxygen uptake) gave informed consent to participate in the study as per the requirement of the institution where the study was undertaken. Cyclists did not have any history of injury or pain in their knees.

Data collection

During the first session, anthropometrics (height and body mass) were measured according to ISAK protocols (Marfell-Jones, et al., 2006). Cyclists warmed up at 150 W for 10 minutes before the test began. They then performed an incremental maximal step test to exhaustion with initial work load of 100 W and increments of 25 W/min (Lucía, et al., 2002) using their own bicycles on a Computrainer cycle trainer (RaceMate, USA). Power output was measured throughout the incremental test along with gas exchanges by the breath-by-breath method using an open-circuit gas exchange system (MGC CPX/D, Medical Graphics Corp., St Louis, MO, USA). Pedalling cadence was controlled close to 90 ±2 rpm using visual feedback of the cycle trainer head unit. The test was stopped by the cyclists' voluntary exhaustion or when they were unable to maintain pedalling cadence. The oxygen and carbon dioxide analysers were calibrated using medical grade gases that spanned air in the physiological range. The workload of the second ventilatory threshold was determined as reported elsewhere (Weston & Gabbett, 2001).

After 48 hours cyclists warmed up for 10 minutes at 150 W. They rode one minute with 90 rpm of pedalling cadence at the maximal workload from the incremental test in a preferred position on the saddle and then at the workload set to the second ventilatory threshold in the same position on the saddle. An additional trial at 70 rpm of pedalling cadence and workload of the second ventilatory threshold was performed. The order of the 90 rpm and the 70 rpm pedalling cadence trials were randomly selected.

Force applied on the right pedal and right lower limb kinematics were recorded for the last 20 s of each trial. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of the cyclists at the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, anterior and posterior pedal stick. Two markers were taped to the bicycle frame and used as the reference for image calibration. A 2D pedal dynamometer custom developed for Look® type cleats (see Figure 10.1) (Nabinger, et al., 2002) and one high speed camera positioned perpendicular to the motion plane (AVT PIKE F-032, Allied Vision Technologies GmbH, Germany) were synchronized by an external trigger. Kinematics were recorded at 60 Hz using AVT ActiveCam viewer software (Allied Vision Technologies GmbH, Germany) and force data were recorded at 600 Hz per channel employing a 16-bit analogical to digital converter (DI220, Dataq Instruments, USA) using WINDAQ® software (WINDAQ, DataQ Instruments Inc., USA).

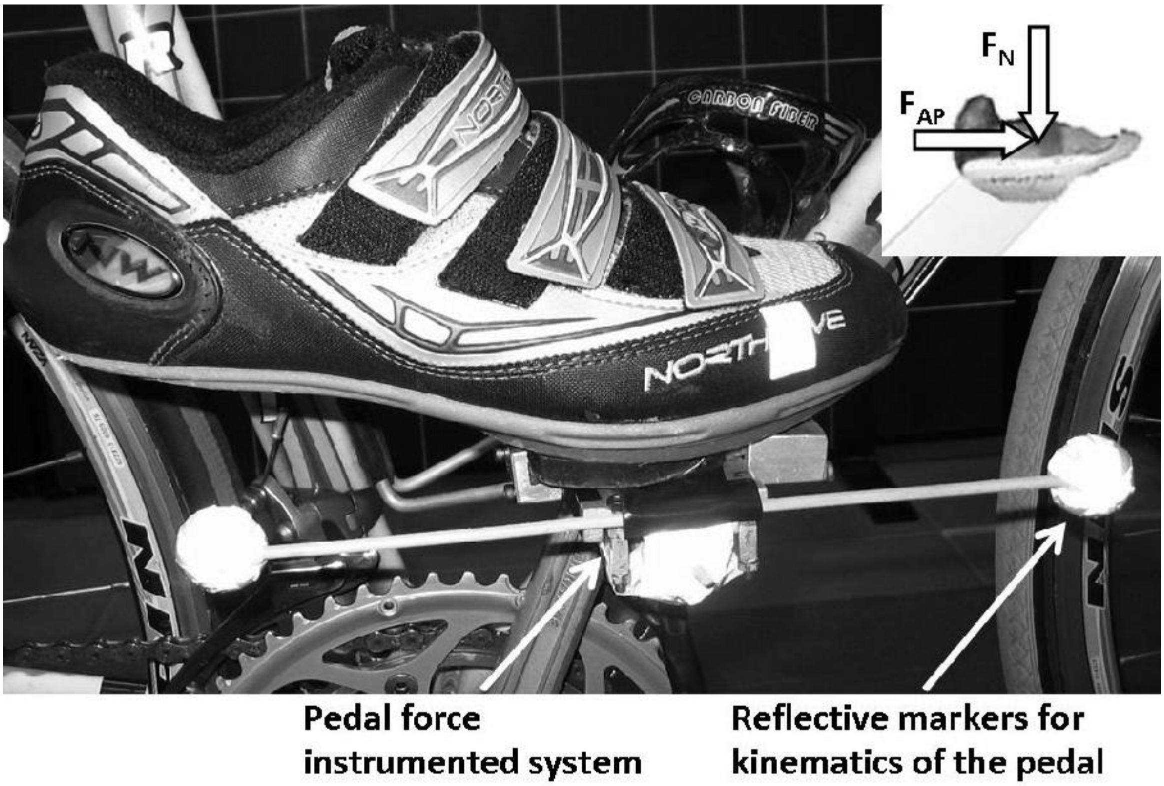


Figure 10.1. Image of the instrumented pedal force system to compute normal force (F_N) and anterior-posterior force (F_{AP}) components of the total force applied on the pedal in the sagittal plane. Reflective markers attached to the pedal were used to compute kinematics of the pedal.

Data analyses

Video files were digitized using automatic tracking of markers in DgeeMe software (Video4Coach, Denmark) for x-y coordinates over time. Kinematic data were smoothed with a digital second order zero lag low pass Butterworth filter with cut-off frequency optimized to reduce signal residual (Winter, 2005). Segment kinematics of the hip, knee, and ankle joints during pedalling movement were calculated from the smoothed x-y coordinate data. Correction of the hip joint center was based on the average coordinate between the marker on the anterior superior iliac spine and the greater trochanter (Neptune & Hull, 1995) (see Figure 10.2). Knee flexion angle (α_K) was defined as shown in Figure 10.2.

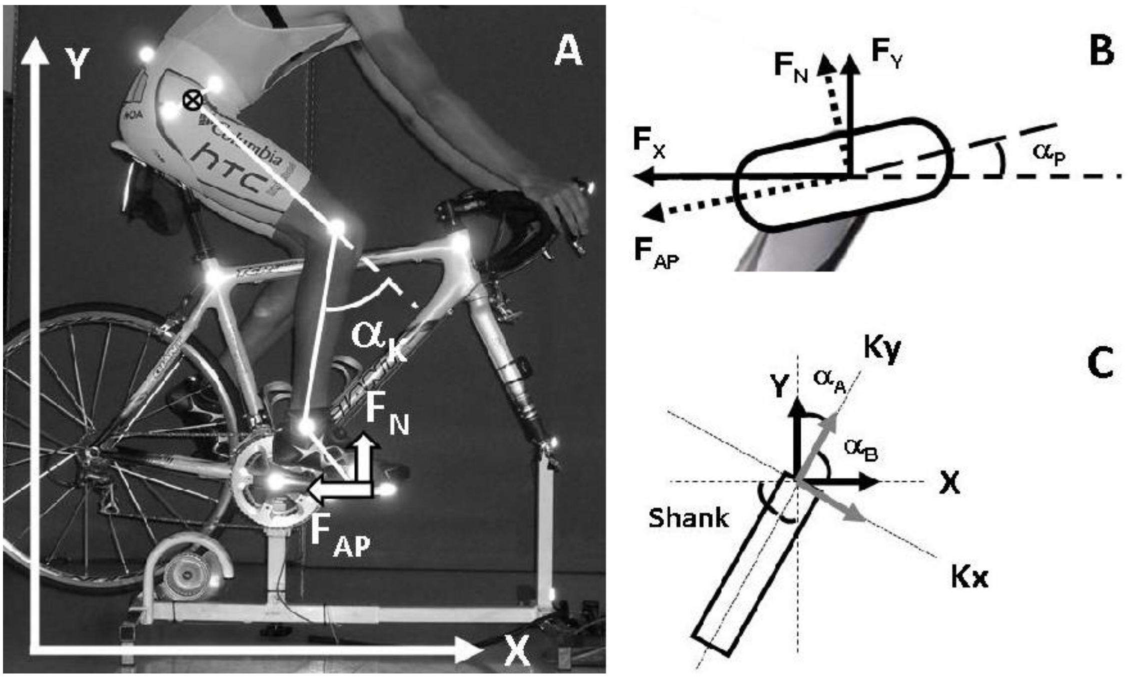


Figure 10.2. (A) Illustration of the anatomic sites used to create the sagittal plane model of cycling and estimate of the hip joint centre using markers at the anterior superior iliac spine and the greater trochanter. Knee joint flexion angle (αK) and reaction analogs of the normal (FN) and anterior-posterior (FAP) pedal force components. (B) Pedal reaction force components converted into forces in the vertical (FY) and horizontal (FX) coordinates using pedal angle (αP). (C) Knee joint normal (KY) and anterior-posterior (KX) reaction forces computed using knee joint resultant forces in global coordinates (X and Y) and shank to vertical (αA) and horizontal (αB) angles.

Force signals were synchronized with kinematics data by selecting the section of the force signal data that presented the change in voltage input from the synchronization unit. Kinematic signals were then interpolated to the sample rate of the force signals using the interpft script in Matlab (Mathworks Inc, MA). Linear and angular velocities and accelerations were computed from smoothed kinematic data by a three points derivative method (Winter, 2005). Pedal angle in relation to the global coordinate system was calculated to convert the forces on the pedal reference system to forces in the global reference system by means of trigonometric procedures (see Figure 10.2) using equations 10.1 and 10.2.

$$F_Y = (F_N \cdot \cos\alpha_P) + (F_{AP} \cdot \sin\alpha_P)$$

Equation 10.1. Vertical force on the global coordinate system (F_Y) computed using the normal (F_N) and anterior-posterior (F_{AP}) forces on the pedal surface and pedal angle (α_P).

$$F_X = (F_N \cdot (-\sin\alpha_P)) + (F_{AP} \cdot \cos\alpha_P)$$

Equation 10.2. Horizontal force on the global coordinate system (F_X) computed using the normal (F_N) and anterior-posterior (F_{AP}) forces on the pedal surface and pedal angle (α_P).

The right lower limb was modeled as a three-segment rigid body system (thigh, shank and foot-pedal) with segment mass and center of mass estimated according to De Leva (1996). Conventional inverse dynamics were conducted to calculate the net joint moments at the knee and ankle (Redfield & Hull, 1986) using adapted scripts of van den Bogert and de Koning (1996), using equations 10.3 to 10.8.

$$F_{AY} = (m_f \cdot a_{fY}) - (m_f \cdot G) - F_Y$$

Equation 10.3. Vertical force on the ankle joint (F_{AY}) computed using the mass of the foot (m_f), acceleration of the foot in the vertical axis (a_{fY}), gravity acceleration (G) and the pedal reaction force in the vertical axis (F_Y).

$$F_{AX} = (m_f \cdot a_{fx}) - F_X$$

Equation 10.4. Horizontal force on the ankle joint (F_{AX}) computed using the mass of the foot (m_f), acceleration of the foot in the horizontal axis (a_{fx}) and the pedal reaction force in the horizontal axis (F_X).

$$M_A = (I_f \cdot \alpha_f) - (F_X \cdot d_{Xd}) - (F_Y \cdot d_{Yd}) - (F_{AX} \cdot d_{Xp}) - (F_{AY} \cdot d_{Yp})$$

Equation 10.5. Net moment in the ankle joint (M_A) computed using the inertia of the foot (I_f), the angular acceleration of the foot in relation to the coordinate origin (α_f), pedal reaction force in the horizontal axis (F_X), moment-arm of the horizontal pedal reaction force (d_{Xd}), pedal reaction force in the vertical axis (F_Y), moment-arm of the vertical pedal reaction force (d_{Yd}), horizontal force on the ankle joint (F_{AX}), moment-arm of the horizontal ankle force (d_{Xp}), vertical force on the ankle joint (F_{AY}), and moment-arm of the vertical ankle force (d_{Yp}).

$$F_{KY} = (m_s \cdot a_{sY}) - (m_s \cdot G) - F_{AY}$$

Equation 10.6. Vertical force on the knee joint (F_{KY}) computed using the mass of the shank (m_s), acceleration of the shank in the vertical axis (a_{sY}), gravity acceleration (G) and the ankle reaction force in the vertical axis (F_{AY}).

$$F_{KX} = (m_s \cdot a_{sx}) - F_{AX}$$

Equation 10.7. Horizontal force on the knee joint (F_{KX}) computed using the mass of the shank (m_s), acceleration of the shank in the horizontal axis (a_{sx}) and the ankle reaction force in the horizontal axis (F_{AX}).

$$M_K = (I_s \cdot \alpha_s) - M_A - (F_{AX} \cdot d_{Xd}) - (F_{AY} \cdot d_{Yd}) - (F_{KX} \cdot d_{Xp}) - (F_{KY} \cdot d_{Yp})$$

Equation 10.8. Net moment in the knee joint (M_K) computed using the inertia of the shank (I_s), the angular acceleration of the shank in relation to the coordinate origin (α_s), net reaction moment in the ankle joint (M_A), ankle reaction force in the horizontal axis (F_{AX}), moment-arm of the horizontal ankle reaction force (d_{Xd}), ankle reaction force in the vertical axis (F_{AY}), moment-arm of the vertical ankle reaction force (d_{Yd}), horizontal force on the knee joint (F_{KX}), moment-arm of the horizontal knee force (d_{Xp}), vertical force on the knee joint (F_{KY}), and moment-arm of the vertical knee force (d_{Yp}).

Patellofemoral (compressive force) was computed adapted from Bressel (2001) (Figure 10.3-A) including corrections for the quadriceps-patellar tendon force ratio (Sharma, et al., 2008) (Figure 10.3-B).

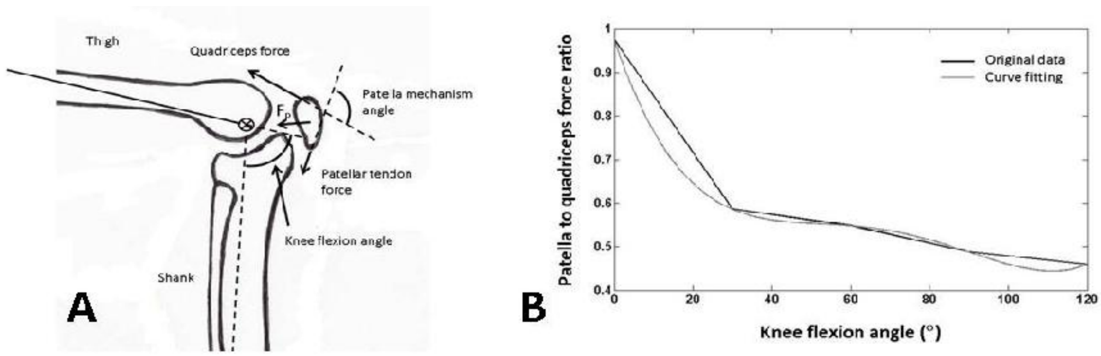


Figure 10.3. (A) Model for knee flexion angle and patella mechanism angle adapted from Bressel (2001), where FP is the compressive force at the patellofemoral joint. (B) Curve fitting of patellar to quadriceps force ratio using data from Sharma et al. (2008).

Patella mechanism angle was estimated using equation 9 presented by Matthews et al. (1977).

$$\alpha_{Pat} = 30.46 + 0.53 \cdot \alpha_K$$

Equation 10.9. Patellar mechanism angle (α_{Pat}) as a function of knee flexion angle (α_K) (Matthews, et al., 1977).

Imbalances in quadriceps to patellar tendon force are observed due to differences in moment arms of the quadriceps tendon and the patellar tendon in relation to the centre of the patella (Sharma, et al., 2008). The correction of patellar to quadriceps force ratio used a function that represented the original data of Sharma et al. (2008). Equation 10.10 was derived from the curve fitting analysis presented in Figure 3-B with normalized residuals of 2.10e-15 and Person correlation to the data of r = 0.99.

$$F_{PQ} = 2.1972e\text{-}8 \cdot (\alpha_K^4) - 6.2597e\text{-}6 \cdot (\alpha_K^3) + 6.20e\text{-}4 \cdot (\alpha_K^2) - 2.6534e\text{-}2 \cdot (\alpha_K) + 0.97502$$

Equation 10.10. Patellar to quadriceps force ratio (F_{PQ}) computed using knee flexion angle (α_K).

Quadriceps muscle force was estimated by the ratio between the knee extensor moment and the moment arm of the patellar tendon from Herzog and Read (1993), exclusively when the net knee moment was extensor. When knee moment was flexor, quadriceps force was assumed to be zero. Patellar tendon force was computed applying the correction of Sharma et al. (2008) to the quadriceps muscle force. Patellofemoral compressive force was then computed employing equation 10.11 adapted from Bressel (2001).

$$F_P = F_Q \cdot \sin(\alpha_{Pat} \cdot 2^{-1}) + F_{PT} \cdot \sin(\alpha_{Pat} \cdot 2^{-1})$$

Equation 10.11. Patellofemoral compressive force (F_P) as a function of quadriceps force (F_Q) and patellar tendon force (F_{PT}), where α_{Pat} is the patellar mechanism angle.

To compute anterior shear and compressive force components at the tibiofemoral joint, knee joint reaction forces were initially converted from the global coordinate system to the tibial plateau coordinate system (see Figure 10.2-C). Quadriceps and hamstrings forces were estimated when the knee moment was extensor or flexor, respectively, using moment arms and muscles lines of action from Herzog and Read (1993). Patellar tendon force was computed applying the correction of Sharma et al. (2008) to the quadriceps muscle force. Normal and anterior-posterior force components on the tibiofemoral joint were computed using equations 10.12 and 10.13, respectively, adapted from Thambyah et al. (2005).

$$F_{TN} = F_{KY} + F_{PT} \cdot \cos\alpha_{PT} + F_{Ham} \cdot \cos\alpha_{Ham}$$

Equation 10.11. Force normal on the tibial plateau (F_{TN}) calculated using the vertical reaction force at the tibial plateau (F_{KY}), patellar tendon force (F_{PT}), patellar tendon angle (α_{PT}), hamstrings muscle force (F_{Ham}), and hamstrings tendon angle (α_{Ham}).

$$F_{TAP} = F_{KX} + F_{PT} \cdot \sin\alpha_{PT} - F_{Ham} \cdot \sin\alpha_{Ham}$$

Equation 10.12. Force anterior-posterior on the tibial plateau (F_{TAP}) calculated using the horizontal reaction force at the tibiofemoral joint (F_{KX}), patellar tendon force (F_{PT}), patellar tendon angle (α_{PT}), hamstrings muscle force (F_{Ham}), and hamstrings tendon angle (α_{Ham}).

Peak patellofemoral, compressive and anterior tibiofemoral forces were used to contrast knee forces with previous studies and to compare the effects of changing workload and pedalling cadence. Knee forces and flexion angle were computed using a custom written program in Matlab (Mathworks Inc, MA) available at http://sites.google.com/site/binirodrigo/about-me/data-analysis-scripts for ten consecutive crank revolutions to determine the average and standard deviation. For comparison to other studies in cycling using different workload levels, peak knee forces were normalized by cyclists' individual mechanical workload at the cycle ergometer.

Statistical analyses

Data normality distribution and sphericity were confirmed for all variables by the Shapiro-Wilk and Mauchly tests, respectively using SPSS for Windows 16.0 (SPSS, NY, USA).

The effects of changes in workload and pedalling cadence on the patellofemoral compressive force, tibiofemoral anterior shear and compressive force were evaluated using Cohen‘s effect sizes (ES). Differences were rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Knee joint net moment of one representative cyclist is presented in Figure 10.4 as a function of the crank angle (A) and the knee flexion angle (B). Arrows indicate the propulsive (from 0° to 180° of the crank angle) and recovery (from 180° to 360° of the crank angle) phases.

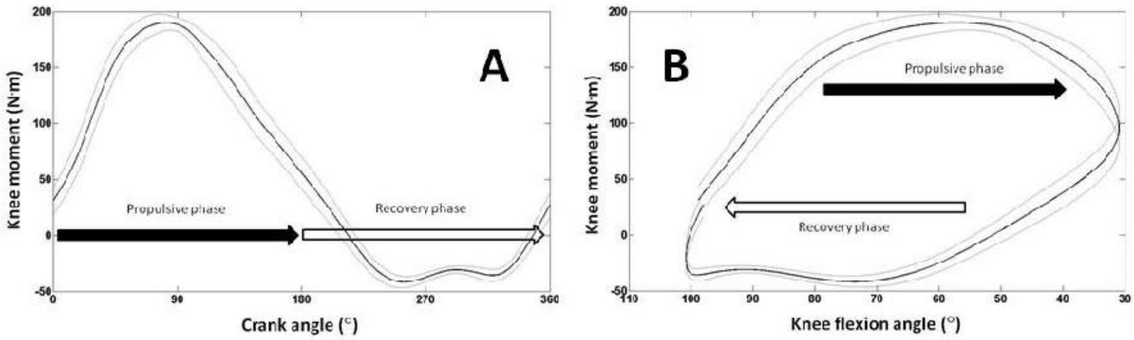


Figure 10.4. Mean ±SD of ten consecutive crank revolutions for the knee joint net moment as a function of the crank angle (A) and the knee flexion angle (B). Arrows indicate the propulsive (from 0° to 180° of the crank angle) and recovery (from 180° to 360° of the crank angle) phases.

In Figure 10.5, patellofemoral compressive force is presented as a function of the crank angle (A) and the knee flexion angle (B). Note that the force is zero between crank angles 200-340° as co-contraction is not considered in the model. When the knee moment is flexor the model assumed no force is being applied at the patellofemoral joint.

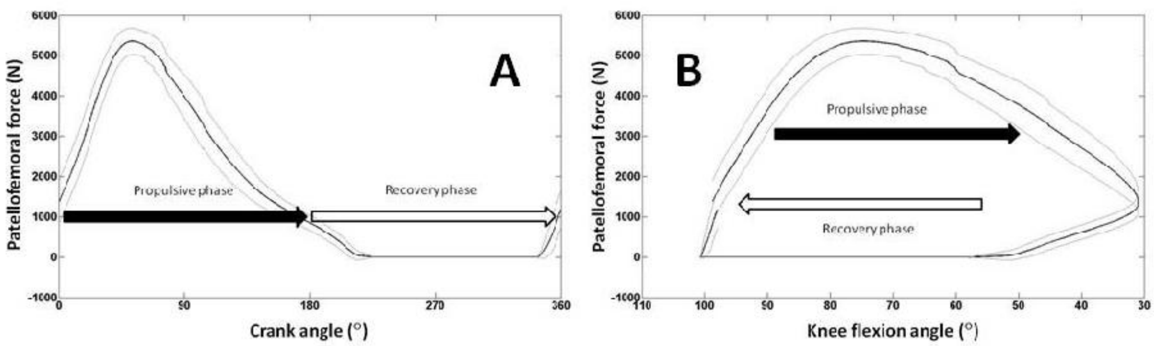


Figure 10.5. Mean ±SD of the patellofemoral compressive force as a function of the crank angle (A) and the knee flexion angle (B). Arrows indicate the propulsive (from 0° to 180° of the crank angle) and recovery (from 180° to 360° of the crank angle) phases.

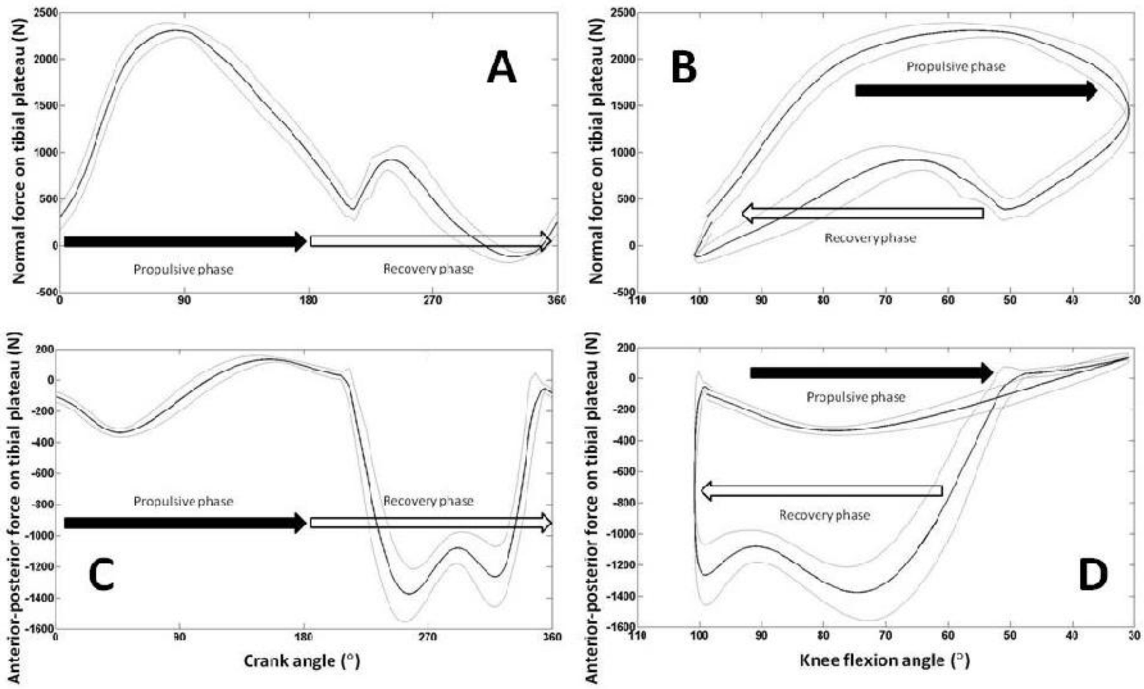


Figure 10.6. Mean ±SD of the normal force on the tibial plateau as a function of the crank angle (A) and the knee flexion angle (B). Mean ±SD of the anterior-posterior force force on the tibial plateau as a function of the crank angle (C) and the knee flexion angle (D). Arrows indicate the propulsive (from 0° to 180° of the crank angle) and recovery (from 180° to 360° of the crank angle) phases.

In Figure 10.6, results of force normal (A and B) and anterior-posterior (C and D) at the tibiofemoral joint (tibial plateau) are presented. Large normal compressive force at the tibiofemoral joint during the propulsive phase was associated with greater pedal force application. There was less anterior force than posterior force at the tibiofemoral joint.

Peak patellofemoral compressive force was smaller (15%) at the workload of the second ventilatory threshold compared to the maximal workload and at the 90 rpm trial (35%) compared to the 70 rpm trial. Peak compressive normal force at the tibiofemoral joint and peak anterior force at the tibiofemoral joint were not largely affected by changes in workload or pedalling cadence (see Table 10.1).

Table 10.1. Mean ±SD results of peak patellofemoral compressive (PFC) and tibiofemoral normal compressive (TFC) force and peak tibiofemoral anterior (TFA) forces for two levels of workload (maximal and ventilatory threshold - VT) and two pedalling cadences (90 and 70 rpm). Differences between workloads and pedalling cadences are reported as mean difference percentages along with effect size magnitudes.

Abbreviations used for ventilatory threshold workload (VT) and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

	Maximal workload	**VT workload**	**70 rpm**
Patellofemoral compressive force (N)	4625 ±1779 ***VT 15%; 1.1, L***	3952 ±1618 ***70 rpm 35%; 2.3, L***	5554 ±1980
Tibiofemoral anterior force (N)	138 ±66 VT 1%; 0.1, T	138 ±63 70 rpm 4%; 0.3, S	136 ±61
Tibiofemoral compressive force (N)	1885 ±755 VT 1%; 0.1, T	1842 ±707 70 rpm 22%; 0.8, M	2203 ±812

Peak patellofemoral compressive force (mean ±SD: 4625 ±1779 N) was as expected larger than peak compressive normal force at the tibiofemoral joint (1885 ±755 N) and peak anterior force at the tibiofemoral joint (138 ±66 N) for the group of cyclists (see Table 10.2) for the maximal workload trial. Results of peak patellofemoral, peak compressive normal force and peak anterior forces on the tibial plateau from different studies are shown in Table 10.2 normalized by workload to enable between-subjects comparison across the different studies. Our study showed larger forces than those reported using other models.

Table 10.2. Mean results of peak patellofemoral compressive (PFC) and tibiofemoral normal compressive (TFC) force and peak tibiofemoral anterior (TFA) forces from different studies in cycling in Newtons and normalized by mechanical work (% mechanical work).

Study; subjects	Power output; pedalling cadence; mechanical work	Knee model	Knee joint forces (N, % mechanical work)
Ericson & Nisell (1986); 6 non-cyclists (20 to 31 years)	120 W; 60 rpm; 120 J	Inverse dynamics without patellar force correction.	PFC: -; TFC: ~810, ~6.7%; TFA: ~356, ~3%.
Ericson & Nisell (1987); 6 non-cyclists (20 to 31 years)	120 W; 60 rpm; 120 J	Inverse dynamics without patellar force correction.	PFC: ~1675, ~14%; TFC: -; TFA: -.
McCoy & Gregor (1989); 10 non-cyclists (~29 years)	200 W; 80 rpm; 150 J	Inverse dynamics without muscle forces.	PFC: -; TFC: ~205, 1.4%; TFA: 112.8, 0.8%.
Neptune & Kautz (2000); 16 non-cyclists (24 ±7 years)	~150 W; 60 rpm; 150 J	Musculoskeletal modelling including activation dynamics.	PFC: 1294, 8.6%; TFC: 1343; 8.9%; TFA: 1526; 10.2%.
Bressel (2001); 21 non-cyclists (28 ±6 years)	157 W; 80 rpm; 118 J	Inverse dynamics without patellar force correction.	PFC: 1542, 13.1%; TFC: -; TFA: -.
Tamborindeguy & Bini (2011); 9 non-cyclists (22 to 36 years)	70 W; 70 rpm; 60 J	Inverse dynamics without patellar force correction. No muscle force in the tibiofemoral model.	PFC: 721; 12.0%; TFC: 69; 1.1%; TFA: 84; 1.4%.
Present study; 1 competitive male cyclist (26 years)	377 W; 90 rpm; 249 J	Inverse dynamics including patellar force correction and muscle forces.	PFC: 4625, 19%; TFC: 1885, 7.4%; TFA: 138, 0.5%.

Discussion

We have presented a method to compute patellofemoral and tibiofemoral joint forces in cycling using inverse dynamics based on pedal forces and joint kinematics data. A step-by-step procedure has accounted for patellar tendon to quadriceps force ratio in the patellofemoral force and accounted for muscle force effects on tibiofemoral joint forces. We illustrated increases in peak patellofemoral compressive forces at higher workload level and lower pedalling cadences. Our results presenting patellofemoral force normalized by workload level were higher compared to other studies (20-45%) (Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011) with differences varying compared to other results for tibiofemoral compressive force (20-572%) (Ericson & Nisell, 1986; McCoy & Gregor, 1989) and tibiofemoral anterior force (60-200%) (Ericson & Nisell, 1986; McCoy & Gregor, 1989). Workload and pedalling cadence are potential factors on differences from knee forces measured in our study and previous studies.

Previous studies measured knee forces at lower power output levels (from 70 W to 200 W) and lower pedalling cadence (60-70 rpm) (Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011) compared to the present study and to what cyclists are submitted during racing (377 W of workload and 90 rpm of pedalling cadence). Workload level affects predicted tibiofemoral,(Ericson & Nisell, 1986) and patellofemoral forces (Ericson & Nisell, 1987) as shown in our study and by others. Higher pedal reaction forces are observed for higher levels of workload (Patterson & Moreno, 1990). The choice of using higher pedalling cadence compared to previous studies may add an effect from inertial components of joint forces. For a target power output, the choice for a higher pedalling cadence will result in reduced pedal reaction forces, which reduces concentric forces and increases eccentric forces during cycling (Neptune & Herzog, 1999). The increase in pedalling cadence from 70 rpm to 90 rpm resulted in large increase in peak patellofemoral compressive force in our study. Therefore, the testing condition is a primary factor in knee joint predicted forces, with preference for replicating effort level similar to that observed in cycling training, and using pedalling cadence close to 90 rpm which tends to be the most efficient in endurance cycling (Marsh, et al., 2000; Neptune & Herzog, 1999).

The models employed to estimate patellofemoral compressive force were limited to the assumption that the patella is a concentric pulley and that the force from the quadriceps tendon and the patellar tendon have the same effect in patellofemoral compressive force. Mason et al. (2008) observed that various studies measuring forces in vitro reported a reduction of patellar tendon to quadriceps tendon force ratio with the increase in knee flexion angle. Therefore, we applied a correction in the patellar tendon force based on the results of Sharma et al. (2008) to take into account the reduced moment-arm of the patellar tendon compared to the quadriceps tendon. We also improved previous models for tibiofemoral forces (McCoy & Gregor, 1989; Tamborindeguy & Bini, 2011) that did not include anatomy of the knee in the model. Lower tibiofemoral forces were reported by McCoy & Gregor (1989) and Tamborindeguy and Bini (2011) compared to ours because they did not include in their models information from lines of action or moment-arms of the patellar tendon and hamstrings. These limitations also highlight the large contribution from muscle forces to tibiofemoral compressive and anterior-posterior force components (Kellis, 2001). Therefore, the inclusion of patellar tendon to quadriceps force ratio and information on the moment-arms and lines of action of the main muscle groups is important for the prediction of patellofemoral and tibiofemoral forces.

Experience and ability in cycling is likely to affect knee forces. Cyclists and non-cyclists have been observed to differ in terms of pedal force application (Mornieux, et al., 2008), which may affect comparison of knee forces between these groups. Hasson et al. (2008) observed that non-cyclists improve posterior force applied on the pedal surface when knee flexor moment is increased. Greater knee flexor moment and posterior pedal force application are expected for experienced cyclists (Bini & Diefenthaeler, 2010; Diefenthaeler, et al., in press). In our results, greater posterior tibiofemoral force was observed in the recovery phase (see Figure 6-C), which may be linked to greater posterior force on the pedal force from activation of knee joint flexor muscles. Further research would elucidate differences between knee forces of cyclists and non-cyclists and relationships to pedal force application.

Predicting knee forces in cycling depends on measuring the output force on the pedal, capturing lower limb motion and on modelling the knee anatomy. Limitations from capturing lower limb motion and relating that to tibiofemoral and patellofemoral motion are linked to skin motion artefacts (Morton, Maletsky, Pal, & Laz, 2007) and the definition of tibiofemoral centre of rotation (Kellis & Baltzopoulos, 1999). Modelling the knee depends on characteristics of the muscles crossing the knee joints (e.g. patellar tendon moment arm) which usually need to be retrieved from the literature. Measuring muscle forces also depends on assuming that there is no co-contraction between antagonist muscles (e.g. quadriceps and hamstrings group). Forward dynamics simulation should reduce these errors once it accounts for muscle co-activation and muscle mechanical characteristics (force-length and force-velocity relationships) (Neptune & Van Den Bogert, 1998). Muscle co-contration and mechanical properties may be improved in our model.

Progress in technology related to motion analysis and pedal force measurements will enable the wide use of knee models to predict knee forces for clinical use.

Conclusion

The model enables computation of patellofemoral compressive, tibiofemoral compressive and tibiofemoral anterior-posterior forces normalized by workload, using inverse dynamics based on sagittal plane lower limb kinematics and pedal forces during cycling. The patellar tendon to quadriceps force ratio and the contribution of muscle forces to tibiofemoral joint forces have been additions to previous published models. The model improves assessment of patellofemoral compressive force for clinical assessment because it was sensitive to changes in workload and pedalling cadence.

CHAPTER 11: EFFECTS OF BICYCLE SADDLE HEIGHT ON KNEE INJURY RISK AND CYCLING PERFORMANCE

Overview

Incorrect bicycle configuration may predispose athletes to injury and reduce their cycling performance. There is disagreement within scientific and coaching communities regarding optimal configuration of bicycles for athletes.This review summarizes literature on methods for determining bicycle saddle height and the effects of bicycle saddle height on measures of cycling performance and lower limb injury risk. Peer-reviewed journals, books, theses, and conference proceedings published since 1960 were searched using Medline, Scopus, ISI Web of Knowledge, EBSCO, and Google Scholar data bases resulting in 62 references being reviewed. Keywords searched included 'body positioning', 'saddle', 'posture, 'cycling' and 'injury'. Methods for determining optimal saddle height are varied and not well established, which have been based on relationships between saddle height and lower limb length (Hamley and Thomas, trochanteric length, length from ischial tuberosity to floor, LeMond, heel methods) or a reference range of knee joint flexion. There is limited information on the effects of saddle height on lower limb injury risk (lower limb kinematics, knee joint forces and moments and muscle mechanics), but more information on the effects of saddle height on cycling performance (performance time, energy expenditure/oxygen uptake, power output, pedal force application). Increasing saddle height can cause increased shortening of the vastii group, but no change in hamstrings length. Length and velocity of contraction in *soleus* seems to be more affected by saddle height than *gastrocnemius*. The majority of evidence suggested that a 5% change in saddle height affected knee joint kinematics by 35% and moments by 16%. Patellofemoral compressive force seems to be inversely related to saddle height but the effects on tibiofemoral forces are uncertain. Changes less than 4% of trochanteric length do not seem to affect injury risk or performance. The main limitations from the reported studies are that different methods have been employed for determining saddle height, small sample sizes have been used, cyclists with low level of expertise have mostly been evaluated, and different outcome variables have been measured. Given that the occurrence overuse knee joint pain is 50% in cyclists, future studies may focus on how saddle height can be optimized to improve cycling performance and reduce knee joint forces to reduce lower limb injury risk. Given the conflicting evidence on the effects of saddle height changes on performance and lower limb injury risk in cycling, we suggest the saddle height may be set using the knee flexion angle method (25-30°) to reduce the risk of knee injuries and to minimize oxygen uptake.

Introduction

The increased popularity of cycling as a sport and recreational activity has led to a higher incidence of acute (Harvey, Bonning, & Cave, 2008; Kim, et al., 2006) and overuse (Asplund & St. Pierre, 2004; Wanich, et al., 2007) (90% and 85% respectively) injuries. Anterior knee pain will occur in 25% of the population sometime during their life (Ward, Terk, & Powers, 2005) and for cyclists, the knee joint is one of the most affected by overuse injuries (Asplund & St. Pierre, 2004). Overuse injuries can be a result of poor positioning on the bicycle (Burke & Pruitt, 2003). However, there is disagreement within scientific and coaching communities regarding optimal configuration of bicycles for athletes (Burke & Pruitt, 2003).

The most controversial aspect of configuration of the bicycle is saddle height, and consequently this has been the focus of most studies regarding body position on the bicycle (De Vey Mestdagh, 1998; Peveler, 2008; Peveler, et al., 2005; Price & Donne, 1997; Sanderson & Amoroso, 2009). Nevertheless, cyclists often select the saddle position relative to the pedals (and therefore crank) by comfort rather than scientific knowledge. There is concern that an improper position could lead to joint overuse injuries (Gregor, 2000), mainly those affecting the knee joint (Wanich, et al., 2007). On the other hand, most of the strategies to prevent knee injuries based on the configuration of bicycle components have not been assessed by scientific research (Dettori & Norvell, 2006).

Wishv-Roth (2009) recently indicated that understanding the geometry and research around optimal configuration of the bicycle components is vital to maximize performance and minimize injury both for recreation and elite cyclists. Most guidelines reported in magazines are based on empirical data, without guidance from scientific experimental research. Sports medicine practitioners need to be able to advise their athletes on ways to reduce knee injury risk in cycling whilst maintaining or improving cycling performance. Therefore an understanding of how saddle height may be configured and the effects on knee injury risk and cycling performance are important for better prescription by the sports medicine practitioner for bicycle configuration.

This review summarizes, for the sports medicine practioner, literature on methods for determining bicycle saddle height configuration and the effects of saddle height on cycling performance (measured via performance time, energy expenditure/oxygen uptake, power output, and pedal force application) and knee injury risk measures (measured via lower limb kinematics, knee joint forces and moments and muscle mechanics).

Methods

Peer-reviewed journals, books, theses, and conference proceedings published since 1960 were searched using Medline, Scopus, ISI Web of Knowledge, EBSCO, and Google Scholar data bases. Keywords searched included 'body positioning', 'saddle', 'posture, 'cycling', and 'injury'. Results were searched for the keyword 'knee joint' to locate studies regarding the effects of saddle position on the knee joint. Articles were excluded if they did not have at least an English abstract, or if they were concerned with the analysis of different bicycle saddles, saddle pressure, and/or the effects on erectile dysfunction. Fifty four references were reviewed.

Results

The first findings thematic section outlines methods for configuring saddle height. Knowledge of the various methods available is needed by the reader to be able to interpret the later two thematic sections on the effects of bicycle saddle height configuration on cycling performance (thematic section 2) and knee injury risk (thematic section 3). Sports medicine practitioners, coaches, and cyclists need to be aware of how changing seat height for performance may influence injury risk and vice versa.

Since initial investigations of saddle height on physiology and performance,(1967) sports scientists have been searching for the “optimal” configuration of bicycle components to increase performance and prevent injuries (Peveler, 2008). A variety of methods have been proposed, some of which are based upon scientific studies and others on anecdotal experience. Some methods for determining saddle height are based on lower limb length: Hamley and Thomas (1967); trochanteric length (Nordeen-Snyder, 1977); length from ischial tuberosity to floor (1976); Greg LeMond (Burke, 2002); and the heel method (Burke & Pruitt, 2003). A reference range of knee joint flexion has been also used to set saddle height (Burke & Pruitt, 2003; Holmes, et al., 1994). Experimental studies (see Table 11.1) and reviews and empirical based articles (see Table 11.2) examining effects of saddle configuration have showed that “optimal” saddle height depends on the outcome variable measured: cycling performance time (Hamley & Thomas, 1967); energy expenditure/oxygen uptake (Nordeen-Snyder, 1977; Shennum & DeVries, 1976); power output (Peveler, Pounders, & Bishop, 2007); lower limb kinematics (Desipres, 1974; Diefenthaeler, et al., 2006; Nordeen-Snyder, 1977; Price & Donne, 1997; Sanderson & Amoroso, 2009); pedal force application (Diefenthaeler, et al., 2006; Ericson & Nisell, 1988); knee joint forces and moments (Ericson, et al., 1986; Gonzalez & Hull, 1989); and muscle mechanics (Ericson, et al., 1985; Jorge & Hull, 1986).

Table 11.1. Summary of experimental studies examining effects of saddle configuration.

Study	Method of setting saddle height	Outcome measures	Subjects[A]	Main results and notes
Hamley and Thomas (1967)	Percentage of inseam leg length	Time to exhaustion during constant load cycling exercise	100 undefined subjects	109% of inseam leg minimized time to exhaustion during constant workload cycling exercise. No additional information on how different saddle heights were compared.
Desipres (1974)	Percentage of inseam leg length	Muscle activity and joint kinematics	Three male junior cyclists	No significant effects of saddle height (95% and 105% of the inseam leg length) on quadriceps and hamstrings activation. Ankle joint kinematics were most affected when raising saddle height.
Shennum and DeVries (1976)	Percentage of inseam leg length	VO_2	Five undefined subjects between 16 and 18 years	Between 100% and 103% of inseam leg length minimized VO_2. Between 103% and 104% of inseam leg length could minimize power output.
Rugg and Gregor (1987)	Percentage of inseam leg length	Muscle estimated length, shortening velocity, moment-arm of lower limb muscles	Five male cyclists	102% of the trochanteric length (high saddle height) increased shortening of the vastii group, while the hamstrings group was not affected due to its bi-articular attachment.
Peveler et al. (2005)	Hamley and Thomas(1967) method and LeMond method(2002)	Knee angle when pedal was at the bottom dead centre	Fourteen male and five female cyclists	No difference between Hamley and Thomas(1967) and Greg LeMond methods. Both methods did not ensure that the knee angle was between 25-30° for minimizing knee joint load.
Peveler et al. (2007)	Degree of knee angle, percentage of inseam	Anaerobic power	Nine male trained cyclists, three male	25º knee angle resulted in significantly higher mean power compared with 109% inseam leg length in those that fell outside

	leg length		non-cyclists, 15 female non-cyclists	the recommended range on the anaerobic test.
Peveler (2008)	Degree of knee angle, percentage of inseam leg length	VO_2	Five male cyclists, two male non-cyclists, eight female non-cyclists	VO_2 was significantly lower at a saddle height set using 25° knee angle compared to 35° knee angle or 109% of inseam leg length.
Nordeen Snyder (1977)	Percentage of trochanteric length	VO_2, joint kinematics	Ten females non-cyclists between 18 and 31 years	100% of trochanteric length minimized VO_2 compared to 95% and 105%. Major adaptations for knee and ankle joint kinematics when shifting saddle height.
Price and Donne (1997)	Percentage of trochanteric length	VO_2, joint kinematics	Fourteen competitive road cyclists with mean age of 22.9 ±4.1 years	Reduced efficiency at 104% of trochanteric length (higher saddle height) compared to 100% and 96%. Optimal range of saddle height for minimal VO_2 was between 96-100% of trochanteric height.
Jorge and Hull (1986)	Percentage of trochanteric length	Muscle activity	Six cyclists of different training levels	Higher quadriceps and hamstring activation for saddle height at 95% of trochanteric length compared to 100%.
Sanderson and Amoroso (2009)	Percentage of trochanteric length	Muscle activity, estimated muscle length and joint kinematics	Thirteen female cyclists with mean age of 25.6 ±5.9 years	Increased activation of *gastrocnemius medialis* with greater saddle height (107%) compared to the preferred (102%) and low (92%) saddle height. Both muscles of triceps surae do not operate on the same length range when saddle height is modified. *Soleus* was more affected by saddle height in relation to length and velocity of contraction than *gastrocnemius medialis*, mainly when saddle height was raised 5% of the preferred position. *Gastrocnemius medialis* length seems affected by the combination of ankle and knee joint kinematics.

Gonzalez and Hull (1989)	Percentage of trochanteric length	Average absolute hip and knee joint moments	Three male trained cyclists	97% of trochanteric length minimized the average absolute hip and knee moments.
McCoy and Gregor (1989)	Percentage of trochanteric length	Compressive and anterior-posterior force of the tibiofemoral joint	Ten male non-athletes (average age 29 years)	No effects of saddle height (94%, 100% and 106%) on the compressive force of the tibiofemoral joint for ten male subjects riding at 200 W of power output and 80 rpm of pedaling cadence.
Ericson et al. (1985)	Percentage of the ischial tuberosity to the floor	Muscle activity	Six healthy non-cyclists aged between 20 and 31 years	Increased activation of *gluteus medius, semimembranosus, soleus* and *gastrocnemius medialis* for 120% ischial tuberosity to the floor (higher saddle height) compared to 102% and 113%).
Ericson and Nisell (1988)	Percentage of ischial tuberosity to floor	Pedal force effectiveness[C]	Six healthy non-cyclists aged between 20 and 31 years	Saddle heights (102%, 113% and 120% of the ischial tuberosity to the floor) did not affect force effectiveness.
Diefenthaeler, et al. (2006)	1 cm relative to preferred saddle height	Pedal force, muscle activity and joint kinematics	Three elite cyclists aged between 23 and 30 years	Saddle height altered pedaling technique and muscle activity with optimal results for preferred saddle height.
Rankin and Neptune (2008)	Saddle position relative to the bottom bracket	Power output	Computational simulation	Small changes in saddle height (1 cm) affected power output. Ankle joint compensates for most changes in saddle height.
Houtz and Fischer (1959)	Lowest possible on the ergometer[B]	Muscle activity	Three healthy females non-cyclists	Reduced muscle activation in high saddle heights associated with less perceived effort.

[A]Subject characteristics were not always specified in the papers. Where possible the age, gender and cycling level are reported.

[B]Saddle height configuration relative to subject anthropometry was not reported.

[C]Ratio of the force perpendicular to the crank (effective force) to the total force applied to the pedal (resultant force).

Table 11.2. Summary of review or empirical based articles examining effects of saddle configuration.

Study	Method of setting saddle height	Outcome measures	Paper type	Main results and notes
Burke and Pruitt; (2003) and Burke (2002)	Heel method, inseam leg length method, LeMond method, degree of knee joint angle	Optimize power output and reduce the risk of injuries	Book chapter	Knee joint range method used 25-30°. No recommendation for any of the four methods.
Silberman et al. (2005)	LeMond (Burke, 2002) and Holmes et al. (1994) methods	Optimize power output and reduce the risk of injuries	Review	Greg LeMond (Burke, 2002) and Holmes et al. (1994) methods as possibilities for saddle height configuration.
Mellion (1991)	Percentage of inseam leg length	Overview on overuse problems and cycling injuries	Review	109% of inseam leg to fit saddle height. 96% of the sum of shank and thigh length as an alternative set for saddle height. Saddle fore-aft adjust by the knee to pedal axis (see Figure 1-b).
Wanich et al. (2007)	Percentage of inseam leg length	Overview of overuse problems and cycling injuries	Review	109% of inseam leg method for optimal fitting of the saddle height.
Holmes et al. (1994)	Degree of knee joint angle	Clinical based analysis of the common overuse problems and cycling injuries	Review	Minimal knee joint range between 25-30° for minimizing knee joint load.
Moore (2008)	Degree of knee joint	Body positioning for	Magazine article	Holmes et al. (1994) method but with knee joint range

	angle	cycling		between 20-30º.
Borysewicz (1985)	Percentage of trochanteric length	VO_2	Book chapter	Cyclists could minimize VO_2 setting saddle height at 96% of trochanteric length.
De Vey Mestdagh (1998)	Percentage of trochanteric length or inseam leg length	Optimize power output and reduce the risk of injuries	Review	Nordeen Snyder (1977) method optimal to set the saddle height - use 100% of trochanteric length or 107% of the inseam leg.
Gregor (2000)	Percentage of trochanteric length or inseam leg	Biomechanical variables related to cycling	Review	Saddle height affects knee joint resultant force, muscle activity, joint kinematics and muscle length.

Discussion

Methods for configuring saddle height

This section outlines the various methods for configuring saddle height. All measurements for lower leg length of the cyclist have been taken in a standing position unless otherwise indicated. For a proper configuration, the saddle height measurement must be completed with the crank in line with the seat tube and the measurement taken from the pedal surface to the top of the saddle. The use of various saddle height methods and the effects on performance or injury risk outcomes are contained in subsequent sections.

Percentage of lower leg length methods

The inseam leg length, ischial and trochanteric methods are all based on anthropometric length measurements of the lower leg for configuration of saddle height.

The Hamley and Thomas (1967) method was probably the first research-based method. For a proper set-up using this method (see Figures 11.1 and 11.2-A), the saddle height must be set at 109% of inseam leg length measurement.

The trochanteric length method (see Figure 11.1) uses the length from the most prominent bony surface of the greater trochanter to the floor (Nordeen-Snyder, 1977). Settings of 100% of trochanteric length have been reported (Nordeen-Snyder, 1977; Price & Donne, 1997).

The length from the ischial tuberosity to the floor method (see Figure 11.1) is measured with the cyclist standing and the distance taken between the most prominent bony surface of the ischial tuberosity to the floor (Shennum & DeVries, 1976). Settings of 113% of ischial tuberosity to floor length have been reported (Ericson, et al., 1986).

The Greg LeMond method (Burke & Pruitt, 2003) involves the measurement of the inseam leg length and the configuration of the saddle height based on 88.3% of the distance between the top of the saddle and the centre of the bottom bracket. This method (see Figures 11.1 and 11.2-B) is based on the empirical experience of three times Tour de France winner Greg LeMond. It is important to note that this method does not consider differences in the crank length dimensions. Longer crank length (i.e. 5 mm) result in lower pedalling cadence and smaller knee flexion angle (MacDermid & Edwards, 2010). Further research may look at the effects of crank length on performance variables and on variables related to the risk of injuries.

The empirical heel method (see Figure 11.3-A) is commonly used (Burke, 2002). When the cyclist is seated on the saddle, the knee must be fully extended when the heel is on the pedal and the crank is in line with the seat tube.

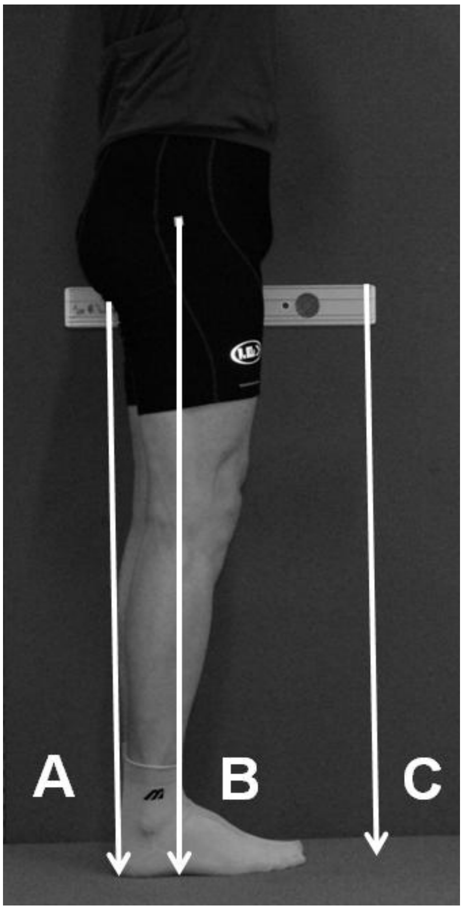


Figure 11.1. Examples of lower leg length measurements (A – ischial tuberosity; B – trochanteric length; C – inseam leg length).

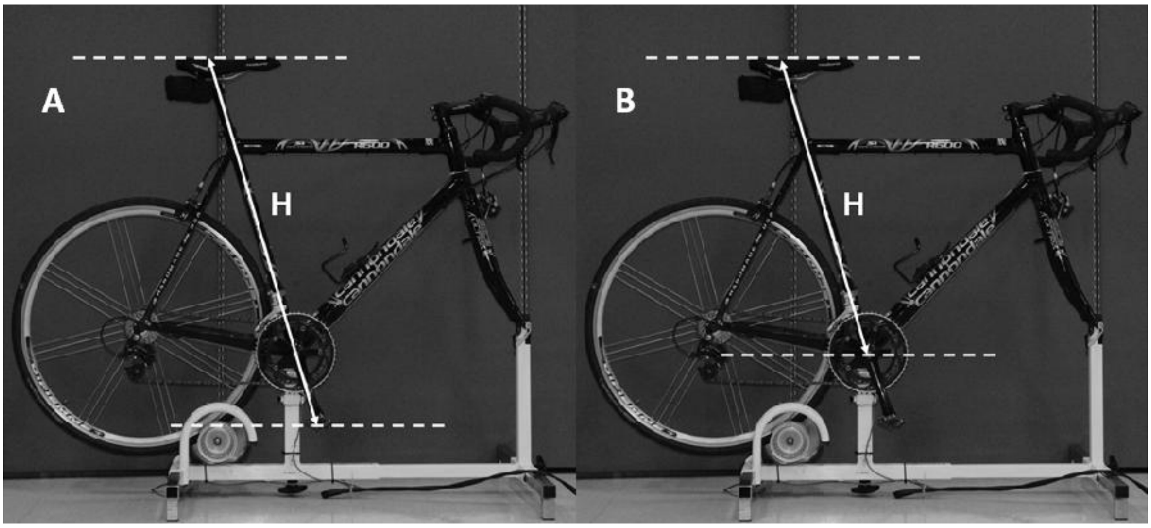


Figure 11.2. Saddle to pedal axis distance (A), used for setting the saddle height by Hamley and Thomas (Hamley & Thomas, 1967), trochanteric length (Nordeen-Snyder, 1977), and length from the ischial tuberosity to the floor (Shennum & DeVries, 1976) methods. Saddle to the centre of the bottom bracket distance (B), used for setting the saddle height by LeMond method (Burke & Pruitt, 2003).

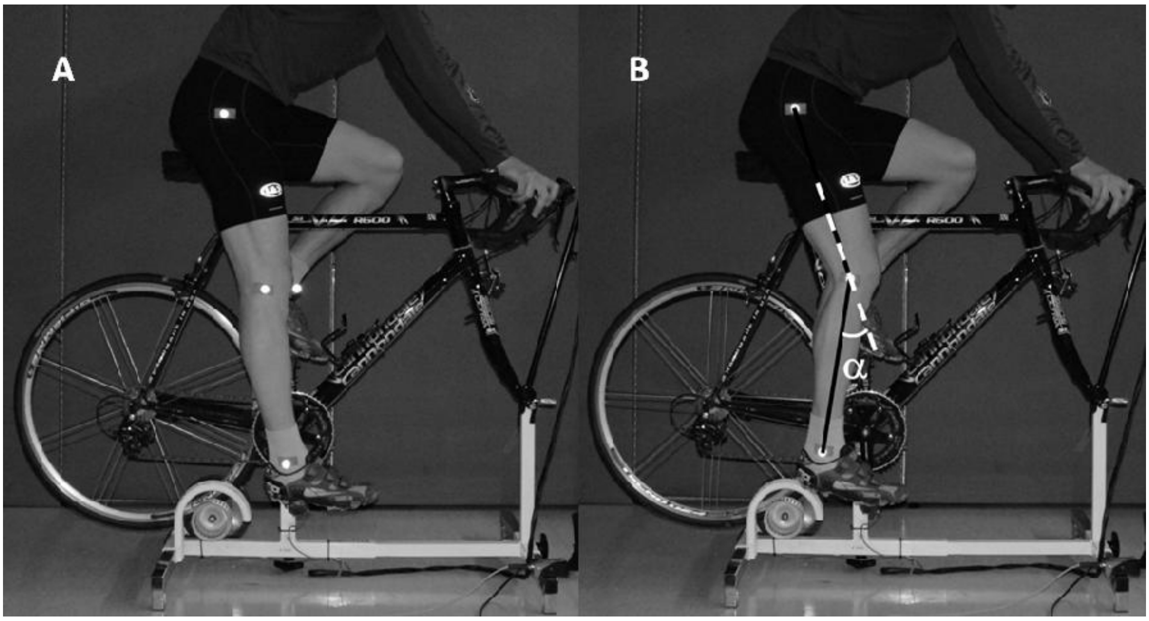


Figure 11.3. Saddle height configuration based on the Heel method (A) and on the Holmes et al. (1994) and on the Howard (2002) methods (B).

Knee angle methods

The Holmes et al. (1994) method (see Figure 11.3-B) involves measurement of the knee angle flexed when the pedal is at bottom dead centre and the cyclist is seated on the saddle, for 25º of flexion for chondromalacea and patellar tendinitis, between 25-30º of flexion for quadriceps tendinitis and medial plica/medial patellofemoral ligament injury, and between 30-35º of flexion for iliotibial band syndrome and biceps tendinitis.

A variation of the Holmes et al. (1994) method was reported by Burke (2002) as the Howard method, for a knee angle of 30º with the pedal at bottom dead centre and the cyclist seated on the saddle. Similar to the Holmes et al. (1994) method, the knee angle measurement will depend on the ankle angle. Increasing ankle plantar flexion will result in higher knee flexion angle.

Comparing methods

Peveler et al. (2005) compared the knee angle when the pedal was in the bottom dead centre using different methods. They observed that length based measures (Hamley and Thomas (1967) and LeMond (Burke, 2002) methods) did not ensure the same knee joint angle range. Only 13 of 19 cyclists reached the desired knee angle range (25-35°) using either method. The reason is possibly because the length based methods do not take into account individual variations in femur, tibia, and foot length (Peveler & Green, 2011).

Review papers by De Vey Mestdagh (1998), Silberman et al. (2005) and Wanich et al. (2007) reported a series of cycling posture adjustments for performance improvement and injury prevention during cycling based on measuring joint angles and segment lengths in relation to optimal references from experimental research (Hamley & Thomas, 1967; Nordeen-Snyder, 1977; Shennum & DeVries, 1976) and from empirical knowledge. As reported by Peveler

(2008), most of the references for posture optimization on the bicycle were based on empirical data and therefore we still do not have enough valid or reliable scientific studies to determine which method is the best. The knee flexion angle method seems more reasonable compared to the length methods because it may standardize the kinematics of the knee, which is one of the most affected joints in terms of injuries in cycling (Asplund & St. Pierre, 2004) and one of the most important for power production (Gonzalez & Hull, 1989), reducing the risk of injuries and improving performance (Peveler, 2008).

Effects of bicycle saddle height configuration on cycling performance

Since Hamley and Thomas (1967) reported that bicycle saddle height affected time to exhaustion during constant workload cycling trials, studies have investigated the effect of saddle height on other parameters. We review in this section studies that have examined the effects of saddle height on cycling performance measures (cycling performance time, energy expenditure/oxygen uptake, power output, and pedal force application).

Cycling performance time

There was only one study that investigated the effects of saddle height on cycling performance time. Hamley and Thomas (1967) measured time to exhaustion during constant load trials in the laboratory for 100 non-specified performers. A longer time to exhaustion could be achieved when setting the saddle height by 109% of the inseam leg length.

Energy expenditure/Oxygen uptake (VO_2)

There seems to be an optimal range of saddle heights to minimize oxygen consumption but studies differ on the optimal saddle height configuration (Hamley & Thomas, 1967; Nordeen-Snyder, 1977; Shennum & DeVries, 1976). Shennum and DeVries (1976) and Nordeen-Snyder (1977) reported that a 5% reduction in saddle height resulted in a 5% increase in VO2. Oxygen uptake was minimized with saddle height set between 100% and 103% of inseam leg length during steady state cycling for five healthy subjects (Shennum & DeVries, 1976) and when set to 100% of trochanteric length (about 107% of inseam leg length) for ten healthy females (Nordeen-Snyder, 1977). Borysewicz (1985) reported lowest oxygen uptake during 45 minutes of steady state cycling when the saddle height was set at 96% of trochanteric length. VO2 during steady state cycling has been reported as significantly lower for 25° knee angle at bottom dead centre compared to 35° knee angle at bottom dead centre and 109% of inseam leg length conditions (Peveler, 2008).

Power output

The effects of saddle height on power output and subsequent increased cycling performance have been observed in anaerobic exercises (Peveler, et al., 2007) with suggested increased power output at higher saddle positions, compared to aerobic cycling (Price & Donne, 1997). Few studies on saddle height changes could be included for this topic because power output was set as an independent variable with focus on the measurement of physiological variables (i.e. VO_2) (Hamley & Thomas, 1967; Nordeen-Snyder, 1977; Shennum & DeVries, 1976).

Cycling economy (power output to VO_2 ratio)

Cycling economy is an important performance predictor because it indicates the ratio between power output and oxygen uptake (Lucía, et al., 2002). The majority of research on saddle height evaluated economy based on steady-state cycling (i.e., fixed power output) and effects on VO_2 (Hamley & Thomas, 1967; Nordeen-Snyder, 1977; Shennum & DeVries, 1976). Peveler (2008; 2011) observed the effects of different saddle height configuration on cycling economy based on oxygen uptake measurement during steady-state cycling, with optimal results when setting the saddle height as 25º of knee angle. Price and Donne (1997) reported that with power output fixed at 200 W, economy was better with seat height at either 96% or 100% trochanteric length compared to 104%.

Pedal force application

Any relationship between maximal performance and saddle height depends on the optimization of pedal force application (Price & Donne, 1997). Changing saddle height can affect the ankle angle (Diefenthaeler, et al., 2006; Nordeen-Snyder, 1977; Peveler, et al., 2007; Rankin & Neptune, 2008), which works as a link between the force produced in the hip and knee joints and the crank (Bini, Diefenthaeler, et al., 2010; Mornieux, Guenette, Sheel, & Sanderson, 2007). Ericson and Nisell (1988) found no significant effects on the force transferred from the pedal to the crank generating propulsive torque for pedal forces from six recreational cyclists at different saddle heights. Pedaling technique, based on effective pedal force application of trained cyclists, compared to recreational cyclists, may be more sensitive to changes in saddle height (Diefenthaeler, et al., 2006; Ericson & Nisell, 1988).

In summary, when the saddle height is set between 96-100% of the trochanteric leg length (Nordeen-Snyder, 1977; Shennum & DeVries, 1976) or using the knee flexion angle (25°) (Peveler, 2008), reduced oxygen uptake and higher economy were observed. Moreover, when the saddle height is set to 109% of the inseam leg length (~102% of the trochanteric leg length), performance time during time to exhaustion test is optimized. (Hamley & Thomas, 1967) In the other hand, no substantial effects in pedal force were found when changing the saddle height (Ericson & Nisell, 1988).

Effects of bicycle saddle height configuration on knee injury risk

One of the main reasons for the prevalence of knee injuries in cyclists is due to the relation between knee joint forces and kinematics (Bressel, 2001). In this section were included studies that have examined the effects of saddle height on knee injury risk measures (lower limb kinematics, knee joint forces and moments, and muscle mechanics).

Lower limb kinematics

Most studies regarding cycling lower limb kinematics have focused on sagittal plane movement (Nordeen-Snyder, 1977; Sanderson & Amoroso, 2009; Shennum & DeVries, 1976). Typical range of motion of these joints in the sagittal plane are 45º for hip angle (from the thigh parallel to the horizontal axis), 75º for knee angle (between 25 and 100º of knee flexion angle), and 20º for ankle angle (about ±10º from the neutral ankle position) (Faria & Cavanagh, 1978). Saddle height affects lower limb kinematics of the ankle (Desipres, 1974; Diefenthaeler, et al., 2006; Nordeen-Snyder, 1977; Rankin & Neptune, 2008), the knee (Gregor, 2000; Price & Donne, 1997) or both the ankle and knee (Sanderson & Amoroso, 2009; Shennum & DeVries, 1976). Hip and ankle joint angles are most affected by the kinematic method of measurement (i.e., 2-D versus 3-D) (Umberger & Martin, 2001). The lower limb also moves inward in the frontal plane and this movement is affected by saddle height (Ruby, et al., 1992).

Between 4-5% change (increase or decrease) in saddle height resulted in 25% (Sanderson & Amoroso, 2009) change in knee range of motion and 40% (Faria & Cavanagh, 1978) reduction in knee joint angle when the pedal was at the bottom dead centre, and 25% (Price & Donne, 1997) to 51% (Sanderson & Amoroso, 2009) change in the maximal ankle angle. Changes in joint range of motion cause changes in muscle length (Sanderson & Amoroso, 2009) and in moment arms (Rugg & Gregor, 1987) of the active muscles and force production.

Knee joint forces and moments

During stationary cycling, maximal compressive force on the patellofemoral joint has been estimated to be between 800 N (riding at 75 W and 70 rpm) and 1500 N (riding at 157 W and 80 rpm) (Bressel, 2001; Neptune & Kautz, 2000; Tamborindeguy & Bini, 2011). Assuming a contact area between the patella and the femur of 0.026 m2 (Wolchok, Hull, & Howell, 1998) and a peak force of 1500 N on the patellofemoral joint (Bressel, 2001), we can achieve 30 MPa of pressure at the cartilage, which is above the reported physiological load (Cohen, et al., 2001).

Three studies have reported compressive forces on the patellofemoral joint during cycling (Bressel, 2001; Ericson & Nisell, 1987; Neptune & Kautz, 2000). Ericson and Nisell (1987) developed a kinetic model that estimated from trigonometric procedures the patellofemoral compressive forces during cycling. Using three saddle heights (102%, 113%, and 120% from the ischial tuberosity to the floor) they showed that compressive force was inversely related to saddle height. Bressel (2001) showed that backward pedaling resulted in a shift in the location of peak pedal force to a more flexed knee angle which increased patellofemoral compressive force. Neptune and Kautz (2000) described that reverse cycling has been used in rehabilitation. However, Bressel (2001) reported that it can increase patellofemoral compressive force by producing higher knee flexion angles when peak force is applied on the pedal. This example highlights the relationship between joint kinematics and patellofemoral compressive load.

Neptune and Kautz's (2000) muscle-skeletal model results agreed with Bressel's (2001) results of increased patellofemoral compressive force during backward pedaling. However, for a very similar workload (~150 W), Neptune and Kautz (2000) observed lower peak patellofemoral compressive force. This result suggested that a muscle-skeletal model improves the analysis of knee joint forces, compared to the kinetic model, because it included the effects of knee joint muscles' co-contraction. During cycling, the knee joint flexors have an important contribution to knee extension, which could reduce the compressive patellofemoral force by co-contraction (Van Ingen Schenau, Boots, De Groot, Snackers, & Van Woensel, 1992).

Tibiofemoral forces are important because compressive forces on the menisci and the shear forces on the anterior and the posterior ligaments of the knee have been linked with injury (Neptune & Kautz, 2000). Ruby et al. (1992) used a 3D kinetic model of the knee to report compressive tibiofemoral forces and anterior shear forces on the knee throughout the crank cycle. This was the first study to report medio-lateral forces on the knee and rotational moments around the long axis of the tibia and their results led to analysis of cycling as a 3D movement. However, we could not find studies reporting the effects of different saddle heights to the 3D forces and moments on the knee joint.

Ericson and Nisell (1986) reported that saddle height followed an inverse relationship with tibiofemoral compressive force and shear force for six healthy subjects riding in a constant load trial. McCoy and Gregor (1989) reported no effects of saddle height on the compressive force of the tibiofemoral joint for ten male subjects riding at 200 W of power output and 80 rpm of pedaling cadence. When in vivo forces on the anterior cruciate ligament were compared at three levels of workload (75, 125, and 175 W) and two pedaling cadences for eight subjects (Fleming, et al., 1998), there were no significant differences in peak anterior cruciate ligament strain in any situation. Therefore cycling can be useful in rehabilitation exercise programs because of the low strain imposed on the anterior cruciate ligament. One study found that backward pedaling can increase the shear component and reduce the compressive component at the tibiofemoral joint (Neptune & Kautz, 2000). Patients with menisci damage may be better off pedaling backwards, while patients with patellofemoral disorders or ligaments (anterior and/or posterior cruciate ligaments) injuries should avoid pedaling backwards.

The complex relationship between joints affects changes in lower limb joint moments. Increased extensor moments and reduced flexor moments were observed when saddle height was at a low position (102% of ischial tuberosity to the medial malleolus (Ericson, et al., 1986) or 94% of the leg length) (McCoy & Gregor, 1989). The opposite behavior was observed with a high saddle (120% of the ischial tuberosity) compared to the average position (113% of the ischial tuberosity) (Ericson, et al., 1986) and for the 106% of the leg length compared to the average position (100%) (McCoy & Gregor, 1989). For the ankle joint, Sanderson and Amoroso (2009) reported increased peak extensor moment with a low saddle and decreased peak extensor moment when the saddle was raised from the reference position.

Regardless of some discrepancies between studies, it seems that a 5% change in saddle height affects force production and joint moments, joint angles, and muscle length. Knee joint angle and moment are strongly affected by saddle height but the optimal saddle height is still unclear because different methods have been used to measure angles and moments. Moreover, Umberger and Martin (2001) and Sanderson and Amoroso (2009) reported that cyclists chose an average of 104% and 102%, respectively, of the trochanteric length as the saddle height, which suggests that cyclists in these studies would have adapted to a different position than one that could minimize joint moments (97% of trochanteric length for Gonzalez and Hull (1989)) or VO2 (Price & Donne, 1997). As previously observed by Herzog et al. (1991) and Savelberg and Meijer (2003), long term adaptations of training can affect the muscle force-length relationship. These adaptations increase the variability of the results and make it difficult to assess the contribution of adapted position. Only Umberger and Martin (2001) and Sanderson and Amoroso (2009) reported the preferred saddle heights of their cyclists.

Few studies have estimated knee joint forces during cycling with changes in saddle height, and some controversial results have emerged from the reviewed research (Ericson & Nisell, 1986; McCoy & Gregor, 1989). For the patellofemoral joint, an inverse relationship was observed in one study (Ericson & Nisell, 1987) while for the tibiofemoral joint controversial results have been reported (Ericson & Nisell, 1986; McCoy & Gregor, 1989). Joint kinematics and moments results have had different outcomes (Desipres, 1974; Diefenthaeler, et al., 2006; Nordeen-Snyder, 1977; Price & Donne, 1997). Joint kinematics and moments also seem to depend on cycling expertise, which compromises comparison between studies (Desipres, 1974; Diefenthaeler, et al., 2006; Nordeen-Snyder, 1977; Price & Donne, 1997). Therefore, we do not have enough evidence to define “optimal” saddle height based on the results of knee joint forces or joint kinematics. If the aim is to minimize the risk of patellofemoral joint injuries, the inverse relationship between saddle height and patellofemoral compressive force may be used as a reference.

Muscle mechanics and activation patterns

The effects of muscle length on force production have been a focus of much sports science research (Rassier, et al., 1999). Direct measurement of muscle length is usually not possible for ethical reasons, but indirect measurements using ultrasound (Austin, Keren, Wieland, & Herzog, 2008; Muraoka, Kawakami, Tachi, & Fukunaga, 2001), or anthropometric models (Grieve, Pheasant, & P.R., 1978) have been used to estimate fascicle length and its effect on force production in sports (Sanderson, et al., 2006). Grieve et al. (1978) proposed anthropometric methods based on cadaver measurements of the muscle-tendon unit length while Frigo and Pedotti (1978) reported a model to estimate muscle-tendon unit length based on the line of action of lower limb muscles. Both studies reported relationships between predicted muscle length and kinematics, which allow the estimation of muscle length during dynamic situations.

Some studies (Rugg & Gregor, 1987; Sanderson & Amoroso, 2009; Sanderson, et al., 2006) have proposed measuring kinematics to infer muscle length during cycling. The force production and the magnitude of joint load depend on muscle length. Rugg and Gregor (1987) observed in five cyclists pedaling at 90 rpm of cadence, that increasing saddle height resulted in increased shortening of the vastii group, but no significant change in the hamstrings group, possibly due to its bi-articulate attachment. Sanderson and Amoroso (2009) applied the model of Grieve et al. (1978) to evaluate the effects of three different saddle heights on gastrocnemius and soleus. Gastrocnemius and soleus muscles operated in different length ranges when saddle height was raised 5% and lowered 10% from the preferred saddle height. Length and velocity of contraction in soleus was more affected by saddle height than gastrocnemius, with greatest changes occurring when the saddle height was raised 5% from the preferred position. Gastrocnemius length seemed to be affected by the combination of ankle and knee joint kinematics. These results extend previous data with similar experimental design (Gregor, 2000).

There is inconclusive information on muscle-length behaviour during dynamic situations (Austin, et al., 2008). Computational models have been used to estimate muscle force production and length of shortening during cycling (Hawkins & Hull, 1990; Neptune, Kautz, & Zajac, 2000) but these models have not been used to investigate saddle height changes. Future simulations of muscle-length force production during cycling at different saddle heights would add important information regarding the best saddle height for muscle force production.

Given the changes in muscle length with changes in saddle height it is likely that neural drive to control muscle force would also change. Muscle force and joint load will also depend on neural drive. The first report of changes in muscle activation at different saddle heights was by Houtz and Fischer (1959), who observed increased muscle activation when the saddle was reduced 15% from the reference position. Houtz and Fischer's study (1959), and later studies (Desipres, 1974; Diefenthaeler, et al., 2006) have been conducted with limited numbers of subjects which does not allow results to be generalized to the population. Jorge and Hull (1986) found increased quadriceps and hamstring activation with saddle height set at 95% of trochanteric length compared to 100%. For high saddle height based on the ischial tuberosity to floor height, Ericson et al. (1985) found that semimembranosus and gastrocnemius medialis had increased activation for six healthy subjects, which was subsequently confirmed by Sanderson and Amoroso (2009) for 13 female trained cyclists. The differences may be related to preparation of surface electromyography or to pedaling skills of the subjects (i.e., trained cyclists or healthy subjects). There was also a report that muscle timing (i.e., onset and offset) would be modified by saddle height (Gregor, 2000), however, there is still no conclusive evidence. Currently, we cannot define an optimal saddle height for improving performance or preventing injuries using evidence from muscle activity studies.

Limitations of the cited studies

There are many limitations in the research studies reviewed. The different approaches for setting saddle height made it difficult to compare results between studies. Only Shennum and DeVries (1976) reported their results of oxygen uptake with comparison to other methods while Peveler et al. (2005) highlighted the differences in the knee joint angle using different methods to configure the saddle height.

Sample size ranged from three (Houtz & Fischer, 1959) to 100 (Hamley & Thomas, 1967). Expertise of the subjects ranged from trained road cyclists (Price & Donne, 1997) to healthy non-cyclists (Ericson & Nisell, 1988; Ericson, et al., 1985; Nordeen-Snyder, 1977; Shennum & DeVries, 1976) to mixed levels of cycling experience (Peveler, 2008). It is possible that experienced cyclists have adapted to a specific position due to the time they spend training. Such adaptation may be less marked for recreational cyclists or those that ride in multiple positions (e.g. triathletes). However, we could not find any studies that had a focus on training cyclists to ride at different saddle heights and measured the differences in performance.

Different outcome variables were analysed to indicate the effects of optimal saddle height for injury prevention and performance optimization. Most studies did not report the sensitivity or variability of these variables to changes in saddle height. It is possible that different positions are optimal for performance versus injury prevention. The magnitude of the differences in some studies (Peveler, 2008; Peveler & Green, 2011) were too small (effects sizes 0.07-0.20) so it was unclear how substantial the changes were in the studies.

If we consider previously reported optimal settings for saddle height (96-100% of trochanteric length) and we use the "optimal" setting of the saddle to the bottom bracket length (0.773 m) and crank length (0.191 m) reported by Gonzalez and Hull (1989) (resulting in a saddle height of 0.964 m) for a subject 177.8 cm high cycling at 90 rpm, our "optimal range" for the saddle height will be between ~ 0.925 m and 0.964 m. This difference of ~ 4 cm is more than any experienced cyclist would consider and a 4% difference in saddle height would result in ~5% change in oxygen uptake (Nordeen-Snyder, 1977).

Most methods of setting saddle height resulted in different joint kinematics which would affect joint forces and increase risk of injury (Peveler, et al., 2005).

Practical implications and recommendations

The configuration of the bicycle saddle height is not standardized in relation to the methods that can be used for this configuration. The optimal reference for each method is not well defined and a wide-range (i.e., 96-100% of the trochanteric length to the floor) used for performance optimization has been proposed. Evidence for performance improvements has led to using the knee joint angle method from Holmes et al. (1994) compared to the leg length methods (Peveler, 2008). Future studies may focus on the effects of previous training adaptation on the optimal reference for the knee angle for setting the bicycle saddle height.

Given the limitations of the research studies reviewed, sports medicine practitioners are encouraged to advise their cycling athletes to configure their bicycle using the Holmes et al. (1994) method, which involves the measurement of the knee angle when the pedal is at bottom dead centre. For proper configuration of the saddle height using this method, the knee must be flexed between 25-30 degrees which has been related to lowering knee joint load (De Vey Mestdagh, 1998) and improving cycling economy (Peveler, 2008).

Conclusion

Methods for determining optimal saddle height are varied and have not been comprehensively compared using experimental research studies. There is limited information on the effects of saddle height on lower limb injury risk, but more information on the effects of saddle height and cycling performance. The range between 25-30 degrees of knee flexion has been advocated to reduce the risk of knee injuries and minimize oxygen uptake. Given that overuse knee joint injury is common in cyclists, future studies should determine how saddle height can be optimized to improve cycling performance and reduce knee joint forces to reduce lower limb injury risk.

CHAPTER 12: EFFECTS OF SADDLE HEIGHT ON KNEE FORCES OF RECREATIONAL CYCLISTS WITH AND WITHOUT KNEE PAIN.

Overview

Our study compared the effects of different saddle heights on patellofemoral and tibiofemoral forces of cyclists with and without pain. Cross-sectional. Sixteen cyclists without knee pain and eight cyclists with knee pain performed four 2-minute submaximal cycling trials using their preferred saddle height, two saddle heights (high and low) eliciting ±10° change in knee flexion angle when the crank was at the 6 o'clock crank position and a saddle height (advocated optimal) eliciting 25° knee flexion. Power output was consistent and pedalling cadence was visually controlled at 90 rpm for every trial. Right pedal forces and joint kinematics recorded during all trials enabled calculation of patellofemoral and tibiofemoral forces using a musculoskeletal model. Effects of saddle height were not meaningfully different for patellofemoral and tibiofemoral forces when comparing cyclists with and without knee pain. Compared to the low saddle height there were large tibiofemoral anterior forces at optimal (35% without pain, 51% pain) and high saddle heights (76% without pain, 92% pain). Bicycle saddle height can probably be set within a large range of knee motion (i.e., ~44-65° determined during dynamic cycling at the 3 o'clock position) to minimise possible detrimental effects of large patellofemoral and tibiofemoral forces.

Introduction

Overuse injuries affect up to 85% of cyclists with the rates of knee joint injury ranging between 21% to 65% (Dettori & Norvell, 2006). The position of the body on the bicycle has been suggested to play an important role in the rates of overuse injuries (Callaghan, 2005). Knee joint kinematics assessment has indicated that lower saddle height leads to greater knee flexion angle in cycling (Nordeen-Snyder, 1977). Repetitive forces applied at an improper joint angle (i.e. greater knee flexion angle – see Figure 12.1) may result in soft tissue damage (Callaghan, 2005), where knee flexion angle is defined as the angle between the tibia and the forward projection line of the femur.

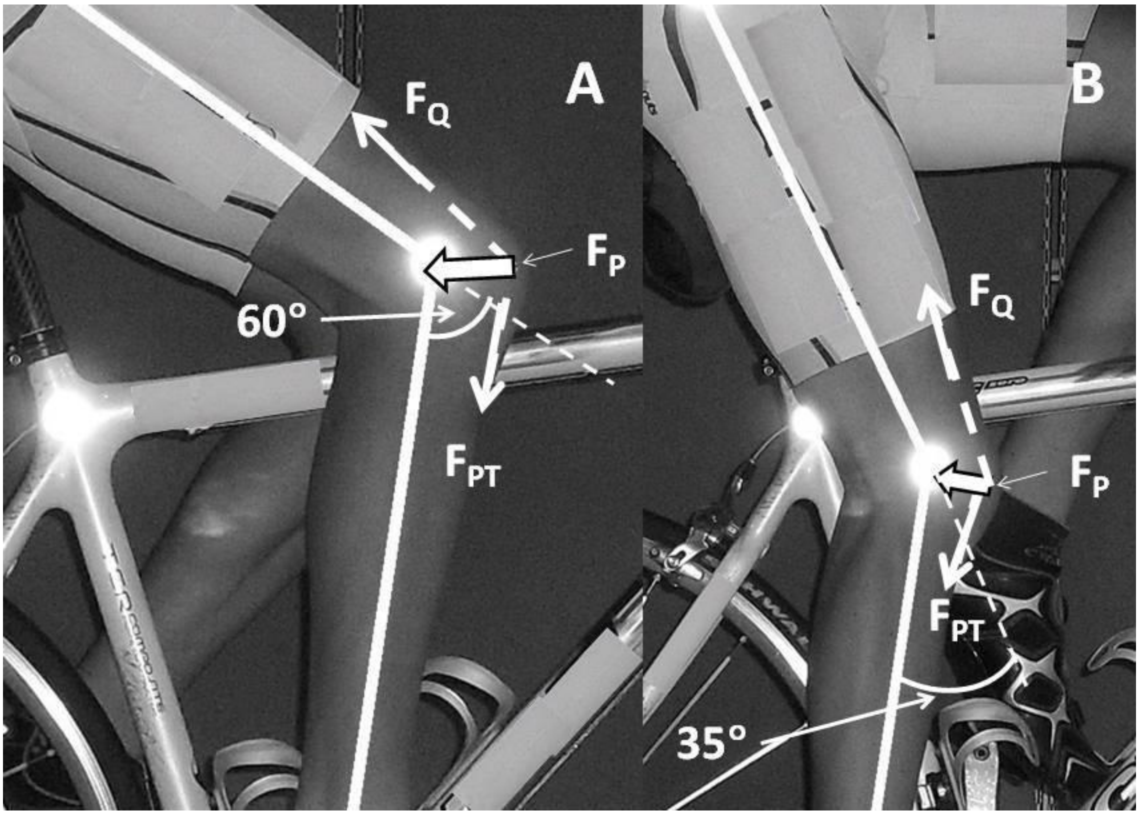


Figure 12.1. Schematic illustration of 25° of change in the knee flexion angle from 60° (A) to 35° (B) that should theoretically decrease patellofemoral compressive force (F_P). Arrows indicate quadriceps muscle force (F_Q) and patellar tendon force (F_{PT}).

Previous studies have given attention to the effects of saddle height on knee forces (Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011). Patellofemoral joint compressive force and tibiofemoral compressive and shear forces were the focus of previous investigations because they may be linked to overload in patellofemoral cartilage, menisci and anterior-posterior cruciate ligaments (Neptune & Kautz, 2000). However, results from previous studies on knee forces are conflicting. For patellofemoral compressive forces, Ericson and Nisell (1987) found increases using a 10% lower saddle height while Tamborindeguy and Bini (2011) did not observe significant changes via 3% changes in saddle height. Ericson and Nisell (1986) reported that saddle height (~10% changes) followed an inverse relationship with tibiofemoral compressive force and anterior shear force, however, no effects of saddle height on tibiofemoral compressive force were found for two other studies with 3-6% changes in saddle height (McCoy & Gregor, 1989; Tamborindeguy & Bini, 2011).

Most studies to date have assessed a limited number (i.e. 6-10) of non-athletes using leg length based methods to set the height of the saddle. Knee joint kinematics are not similar between subjects when leg length methods are used compared to knee flexion angle methods (Peveler, et al., 2005). None of the previous studies provided evidence for knee joint forces of cyclists who reported pain and/or injury in their knees using different saddle heights. This information is important because cyclists with knee pain present different knee joint kinematics compared to healthy cyclists (Bailey, et al., 2003). If there is an inverse relationship between knee forces and saddle height, cyclists with knee pain would benefit from using a higher saddle height compared to healthy cyclists. However, there is no experimental evidence to support this hypothesis.

Therefore, the aim of our study was to compare the effects of different saddle heights on patellofemoral and tibiofemoral forces of recreational cyclists with and without knee pain.

Methods

Participants

Twenty four cyclists without competitive experience in cycling participated in the study. Cyclists reporting pain and/or injury in the right knee as a result of overuse (i.e. non-traumatic injury) and those who remained riding a bicycle for a minimum of 10 km each day three times per week were included in the study. Cyclists were interviewed to assess training and existing injuries related to bicycle riding. They were separated into groups with and without existing knee pain or injury on the right or both knees. The characteristics of the cyclists are shown in Table 12.1.

Table 12.1. Mean (SD) age, body mass, height, time of training and training volume of 16 cyclists without pain and 8 cyclists with pain. Differences between cyclists with and without pain are reported as mean difference percentages along with effect size magnitudes. Abbreviations used for effect sizes of trivial (T), small (S), moderate (M) and large (L).

Groups	Age (years)	Body mass (kg)	Height (cm)	Time of cycling training (hours/week)	Training volume (km/week)
Pain (n = 8)	40 (11)	82 (13)	181 (5)	7 (3)	105 (60)
No pain (n = 16)	43 (9)	78 (18)	178 (7)	7 (3)	80 (60)
Pain vs. No pain	6%; 0.3, S	6%; 0.3, S	2%; 0.4, S	1%; 0.1, T	32%; 0.4, S

Prior to the study, the cyclists were informed about possible risks and signed a consent form approved by the Ethics Committee of Human Research where the study was conducted.

Data collection

Anthropometrics (height and weight) were measured according to ISAK protocols (Marfell-Jones, et al., 2006). Cyclists' current bicycle saddle height and horizontal position were measured to set up the stationary cycle ergometer (Velotron, Racemate, Inc) at the cyclists self-selected “preferred height” configuration. Knee joint flexion angle was measured using a goniometer while the cyclists held the crank at the 6 o’clock crank position. Saddle height was recorded when the saddle was changed from the preferred position to high (-10° of knee flexion with respect to the preferred height), low (+10° of knee flexion with respect to the preferred height), and to the theoretical optimal (25° of knee flexion). The optimal saddle height was chosen because it was supposed to reduce the risk of knee injuries (Holmes, et al., 1994).

Cyclists then performed 10 minutes of warm-up cycling at 90 ±2 rpm of pedalling cadence on the stationary Velotron cycle ergometer using their preferred saddle height. Workload was then increased to match 3.04 (SD 0.78) $W \cdot kg^{-1}$ (243 SD 78 W or 154 SD 47 J) and pedalling cadence was visually controlled at 90 (SD 2) rpm for two minutes. Data were recorded during the first 20 s of the second minute for each saddle height trial. One minute of static rest was completed between trials with different saddle heights.

Force applied on the right pedal and right lower limb kinematics were recorded for the last 20 s during all saddle height conditions. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of the cyclists at the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, anterior and posterior pedal stick. One marker was attached to the sacrum to measure the position of the cyclists when they were evaluated at the different saddle heights. Two markers were taped to the bicycle frame and used as a reference for image calibration. A 2D pedal dynamometer (Candotti, et al., 2007) and one high speed camera positioned perpendicular to the motion plane (AVT PIKE F-032, Allied Vision Technologies GmbH, Germany) were synchronized by an external trigger. Kinematics were recorded at 60 Hz using AVT ActiveCam viewer software (Allied Vision Technologies GmbH, Germany) and force data were recorded at 600 Hz per channel employing a 16-bit analogical to digital converter (PCI-MIO-16XE-50, National Instruments, USA) using a custom Matlab (Mathworks Inc, MA) data acquisition script.

Data analyses

Video files were digitized and automatic tracking of markers were conducted in DgeeMe software (Video4Coach, Denmark) for x-y coordinates over time. Kinematic data were smoothed with a digital second order zero lag band pass Butterworth filter with cut-off frequency optimized to reduce signal residual (Winter, 2005). Segment kinematics of the hip, knee, and ankle joints during pedalling movement were calculated from the smoothed x-y coordinate data. Knee flexion angle was defined as the angle between the tibia and the forward projection line of the femur. Correction of the hip joint center based on the average coordinate between the marker on the anterior superior iliac spine and the greater trochanter was completed (Neptune & Hull, 1995). The average relative horizontal position of the marker on the sacrum to the bottom dead centre was computed over time across ten pedal revolutions for the analysis of body position on the saddle at the four saddle heights.

Linear and angular velocities and accelerations were computed from smoothed kinematic data by a three points derivative method (Winter, 2005). Pedal angle in relation to the global coordinate system was calculated to convert the forces on the pedal reference system to forces in the global reference system by means of trigonometric procedures. The right lower limb was modeled as a three-segment rigid body system (thigh, shank and foot-pedal) with segment mass and center of mass estimated according to De Leva (1996). Conventional inverse dynamics were conducted to calculate the net joint moments at the hip, knee and ankle (Redfield & Hull, 1986) using adapted scripts of van den Bogert and de Koning (1996). Patellofemoral compressive force was computed as described by Bressel (2001) including corrections for quadriceps-patellar tendon force ratio (Sharma, et al., 2008). Tibiofemoral compressive and anterior-posterior forces were computed as described by Thambyah et al. (2005). Peak patellofemoral force and tibiofemoral components (anterior and compressive) were calculated. Knee flexion angles at 3 o'clock and 6 o'clock crank positions were determined. Knee forces and flexion angle were computed using a custom written program in Matlab (Mathworks Inc, MA) for ten consecutive crank revolutions to determine means and standard deviations for each cyclist. Knee joint forces were normalised by workload level (in Joules) for each cyclist.

Statistical analyses

Data were reported as means and standard deviations for each group (with and without pain) and saddle height (preferred, high, low and optimal) for the tibiofemoral (anterior and compressive) and patellofemoral (compressive) peak forces and for the knee flexion angles at 3 o'clock and 6 o'clock crank positions. Data normality distribution and sphericity were evaluated by the Shapiro-Wilk and Mauchly tests, respectively using SPSS for Windows 16.0 (SPSS, NY, USA). Data normality of tibiofemoral compressive force was corrected using a logarithm transform.

The effects of changes in saddle height on the patellofemoral compressive force, tibiofemoral anterior and compressive force, and knee angles at 3 o'clock and 6 o'clock crank positions were evaluated using Cohen's effect sizes (ES). Differences were rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Of the eight cyclists in the knee pain group, three had non-specific knee pain, one had medial knee pain, one had lateral knee pain, two had received surgery and one had arthritis in the patella.

Knee flexion angles at the 3 o'clock and 6 o'clock crank positions derived during dynamic cycling were largely different amongst the four saddle heights (as expected due to the saddle height set-up based on static knee angles) with lower saddle heights leading to greater knee flexion angles. There were no differences in saddle heights for cyclists with and without knee pain. There were no differences between cyclists with and without knee pain from changes in saddle height for any of the force variables.

Tibiofemoral anterior force was larger at high and optimal saddle heights compared to the low saddle height for cyclists with and without pain. Large changes in knee joint angle occurred for the 3 o'clock and 6 o'clock crank positions for both groups when saddle height was changed (see Table 12.2). Changes in height of the saddle were 3-6% for cyclists without pain and 2-7% for cyclists with knee pain to elicit ±10° of change in knee flexion angle measured statically with the crank at the 6 o'clock position in our study. Cyclists with and without existing knee pain presented large differences for the knee flexion angle at the 3 o'clock crank position and 6 o'clock crank position when saddle height was changed.

Table 12.2. Mean (SD) results of saddle height, patellofemoral compressive force, tibiofemoral anterior and compressive force, knee flexion angle at 3 o'clock and 6 o'clock crank positions presented for four saddle heights (preferred, high, low and optimal) for cyclists with and without knee pain. Differences between cyclists with and without pain, and differences between saddle heights within a group, are reported as mean difference percentages along with effect size magnitudes. Percentage and magnitude of differences presented between groups and for different saddle heights. Abbreviations used for comparisons are no-pain (NP), preferred saddle height (Pref) and effect sizes of trivial (T), small (S), moderate (M) and large (L).

	Pain (n = 8)				**No pain (n = 16)**			
	Optimal	**High**	**Preferred**	**Low**	**Optimal**	**High**	**Preferred**	**Low**
Saddle height (cm)	89 (4.7) High 3%; 1.1, L Pref 5%; 1.8, L Low 7%; 2.4, L *NP 1%; 0.1, T*	88 (6.8) Pref 2%; 0.7, M Low 6%; 1.3, L *NP 2%;0.4, S*	85 (4.7) Low 2%; 0.6, M *NP 1%; 0.3, S*	84 (3.8) *NP 1%; 0.1, T*	89 (4.7) High 1%; 0.1, T Pref 3%; 2.2, L Low 6%; 4.6, L	89 (4.1) Pref 3%; 2.1, L Low 6%; 4.4 L	87 (3.9) Low 3%; 2.4 L	84 (4.5)
Patellofemoral compressive force (%Workload)	1812 (861) High 4%; 0.1, T Pref 3%; 0.1, T Low 7%; 0.1, T *NP 11%; 0.3, S*	1736 (657) Pref 7%; 0.2, T Low 11%; 0.3, S *NP 6%; 0.2, T*	1864 (783) Low 4%; 0.1, T *NP 8%; 0.3, S*	1944 (882) *NP 1%; 0.1, T*	1616 (521) High 1%; 0.1, T Pref 6%; 0.2, T Low 21%; 0.8, M	1639 (381) Pref 4%; 0.2, T Low 16%; 0.8, M	1713 (331) Low 12%; 0.7, M	1954 (374)
Tibiofemoral anterior force (%Workload)	71 (20) High 7%; 0.2, T Pref 21%; 0.6, M Low 51%; 1.7, L *NP 6%; 0.2, T*	66 (26) Pref 15%; 0.3, S Low 92%; 1.3, L *NP 30%; 0.7, M*	56 (31) Low 62%; 0.8, M *NP 22%; 0.4, S*	35 (21) *NP 41%; 0.6, M*	75 (23) High 14%; 0.4, S Pref 9%; 0.3, S Low 35%; 1.1, L	86 (26) Pref 20%; 0.7, M Low 76%; 1.4, L	69 (27) Low 40%; 0.7, M	49 (25)

Tibiofemoral compressive force (%Workload)	1111 (443) High 3%; 0.1, T Pref 1%; 0.1, T Low 2%; 0.1, T *NP 10%; 0.4, S*	1072 (364) Pref 3%; 0.1, T Low 6%; 0.2, T *NP1%; 0.1, T*	1106 (381) Low 3%; 0.1, T *NP 7%; 0.1, T*	1139 (406) *NP 7%; 0.3, S*	995 (200) High 6%; 0.2, T Pref 4%; 0.2, T Low 6%; 0.3, S	1059 (311) Pref 6%; 0.1, T Low 1%; 0.1, T	1031 (203) Low 2%; 0.1, T	1055 (167)
Knee flexion angle 3 o'clock crank position (°)	47 (6) High 6%; 0.5, M Pref 20%; 1.4, L Low 26%; 1.9, L *NP 5%; 0.4, S*	50 (5) Pref 13%; 1.1, L Low 16%; 1.6, L *NP 5%; 0.3, S*	56 (7) Low 5%; 0.4, S *NP 5%; 0.5, M*	59 (7) *NP 1%; 0.1, T*	48 (5) High 4%; 0.4, S Pref 9%; 0.9 M Low 19%; 1.7, L	48 (6) Pref 13%; 1.5, L Low 20%; 2.2, L	52 (3) Low 9%; 1.5, L	58 (6)
Knee flexion angle 6 o'clock crank position (°)	31 (11) High 12%; 0.4, S Pref 31%; 0.9, M Low 51%; 1.9, L *NP 1%; 0.1, T*	35 (8) Pref 17%; 0.7, M Low 26%; 1.7, L *NP 10%; 0.5, M*	41 (9) Low 13%; 0.8, M *NP 4%; 0.2, T*	47 (6) *NP 6%; 0.4, S*	32 (7) High 1%; 0.1, T Pref 25%; 1.4, L Low 40% 2.1, L	31 (6) Pref 25%; 1.6, L Low 30%; 2.3, L	40 (5) Low 11%; 0.9, M	45 (6)

Discussion

We compared patellofemoral compressive, tibiofemoral anterior and compressive force along with the knee flexion angle at two positions of the crank (3 o'clock and 6 o'clock) in cyclists with and without knee pain using four different heights of the saddle (preferred, low, high and optimal). Comparisons to the advocated optimal saddle height that resulted in 25° of knee flexion angle when the crank is at the 6 o'clock crank position were conducted. Apart from similarities between the high and optimal saddle heights for both groups of cyclists, changes in saddle height from the preferred to high, low and optimal heights were large and resulted in changes in anterior tibiofemoral force and knee flexion angles.

Reducing saddle height has been reported to increase knee flexion angles (Tamborindeguy & Bini, 2011). In general, lower saddle heights resulted in greater knee flexion angle in our study, which is in agreement to previous studies (Tamborindeguy & Bini, 2011). Knee flexion angle was similarly affected by changes in saddle height for cyclists with knee pain compared to pain-free cyclists, which may refute that existing pain limits knee motion to minimize forces on the knees (Madsen, Noer, Carstensen, & Normark, 2000). Cyclists with knee pain would be expected to prevent knee overload via changes in kinematics of the hip and ankle. Previous studies have reported that when saddle height is changed the ankle joint kinematics are largely changed towards greater plantar flexion for cyclists (Price & Donne, 1997) and non-cyclists (Bini, Tamborindeguy, & Mota, 2010).

Although large changes were observed in knee flexion angle, only tibiofemoral anterior force was largely affected by changes in saddle height (smaller force for low height compared to high saddle height) for cyclists with and without knee pain. These findings indicate that changes in knee flexion angle from various combinations of saddle heights were not fully translated into large changes in patellofemoral and tibiofemoral compressive force. Previous studies reported either an increase (Ericson & Nisell, 1987) or no changes in patellofemoral compressive force (Tamborindeguy & Bini, 2011) when saddle height was reduced by 10% and 3%, respectively for non-athletes without knee pain. We may expect that patellofemoral compressive force is only largely affected with greater changes in saddle height than the ones conducted in our study (i.e. changes in saddle height greater than those caused by changing knee flexion angle by 10°).

Cyclists are advised to configure the saddle height to elicit 25-30° of knee flexion when the crank is static at the 6 o'clock position (Holmes, et al., 1994). This configuration has been empirically chosen with reports of optimal cycling efficiency (Peveler, 2008). However, no study to date has assessed if this configuration of saddle height would reduce patellofemoral compressive force compared to other configurations used by cyclists. Cyclists may not be able to replicate statically (e.g. knee flexion angle at the 6 o'clock crank position) similar joint angles as taken dynamically (Farrell, et al., 2003). Our cyclists without knee pain presented greater knee flexion angles dynamically compared to the results measured statically (i.e. 25°) at the 6 o'clock crank position at the optimal saddle height (~32 SD 7° - no pain group; 31 SD 10° - pain group). Therefore, statically measured knee flexion angle at the 6 o'clock crank position should not be used to configure saddle height. Dynamic measures of joint angles should provide more accurate assessment of effects of saddle height configuration for body position on the bicycle. Preferably, knee flexion angle at the 3 o'clock crank position should be used because it is a position of crank revolution closer to peak knee forces (Ericson & Nisell, 1987). In terms of knee joint flexion angle measured dynamically, our cyclists presented ~39 (SD 5)° (without pain) and ~41 (SD 9)° (with pain) for the 6 o'clock crank position at their preferred saddle height, which indicates a lower saddle height than the advocated optimal (Holmes, et al., 1994). However, patellofemoral compressive force was not largely affected by changes in saddle height. Only trivial to small changes were observed for tibiofemoral compressive force across the different saddle heights.

Anterior tibiofemoral force was greater at the high (76-92% change) and optimal (35-51% change) compared to the low height for cyclists with and without knee pain. One explanation for this result is that the anterior tibiofemoral force depends on the patellar moment arm, which is expected to increase at the high and optimal saddle heights, using Herzog and Read's model (1993). However, the importance of the anterior tibiofemoral force has been neglected because low levels of strain on the anterior cruciate ligament were found with *in vivo* measures (Fleming, et al., 1998). No reports of anterior cruciate ligament injuries have been found in cyclists (Dettori & Norvell, 2006). These possibilities may explain why cyclists with and without knee pain have not been affected by changes in saddle height in terms of anterior tibiofemoral force.

Some limitations may have affected the results of the present study. We did not control forward-backward body position on the saddle when saddle height was changed across trials. However, we measured body position on the saddle during all trials by video tracking of the position of the sacrum and observed trivial differences due to saddle height changes. Modelling joint kinetics (e.g. knee forces) without information of muscle activation and muscle anatomy (e.g. muscle moment arm) may not be as accurate as forward dynamics simulations (Neptune & Kautz, 2000). However, most studies have used only the kinetic-kinematic model (Ericson & Nisell, 1987; Tamborindeguy & Bini, 2011) which enables the comparison of our results to results from other studies of similar design.

Conclusion

Effects of saddle height were not meaningfully different for patellofemoral and tibiofemoral forces when comparing cyclists with and without knee pain. Compared to the low saddle height there were large tibiofemoral anterior forces at optimal (35% without pain, 51% pain) and high saddle heights (76% without pain, 92% pain). Bicycle saddle height can probably be set within a large range of knee motion (i.e., ~44-65° determined during dynamic cycling at the 3 o'clock position) to minimise possible detrimental effects of large patellofemoral and tibiofemoral forces.

CHAPTER 13: EFFECTS OF SADDLE HEIGHT ON PEDAL FORCE EFFECTIVENESS

Overview

Bicycle saddle height configuration may affect pedal force application. Our aim was to compare pedal force effectiveness for different saddle height configurations. Eleven cyclists (38 ±12 years) and eleven triathletes (44 ±8 years) with competitive experience performed 2-min trials at four different saddle heights (preferred, high, low, theoretical optimal) each separated by one minute of rest. Workload was normalized by body weight and pedaling cadence was visually controlled by the athletes at 90 ±2 rpm for all trials. The preferred saddle height replicated the horizontal and vertical configuration of each athlete's bicycle. High and low saddle heights were selected to elicit ±10° knee flexion from knee flexion at preferred saddle height. Guidelines from Peveler (2008) were used to set the theoretical optimal saddle height based on 25° knee flexion when the pedal crank was at the 6 o'clock position. Knee joint angles were measured with a goniometer prior to each trial. Normal and shear forces were measured using an instrumented right pedal and pedal-to-crank angle was measured using an angular potentiometer. A reed switch attached to the bicycle frame detected the position of the crank in relation to the pedal revolution. Forces on the pedal surface were resolved into the tangential force on the crank to compute force effectiveness (ratio between tangential and resultant force applied on the pedal). Magnitudes of differences between the saddle heights were assessed by effect sizes (ES) for the average total (resultant) force and force effectiveness. To elicit ±10° knee flexion, changes of ±3% of the preferred saddle height were required. Changes in average resultant force with saddle height were trivial (1% for preferred versus optimal; ES = 0.2) to moderate (5% for high versus low; ES = 0.8). Changes in force effectiveness with saddle height were small (2% for preferred versus optimal; ES = 0.3) to moderate (6% for high versus low; ES = 1.0). Lower saddle heights produced higher resultant force but lower force effectiveness. Saddle height changes resulted in moderate effects for pedal resultant force and force effectiveness for most saddle height comparisons.

Introduction

The position of the saddle is important for optimal performance because there seems to be a range of saddle heights that minimize oxygen consumption. When saddle height is equivalent to 100% of the height from the greater trochanter to the floor VO_2 can be minimized (1977). When knee flexion is between 25-30° as a result of saddle height VO_2 can also be minimized (Peveler, 2008). No studies have investigated effects on pedal forces as a result of saddle height changes based on the knee flexion angle. We would expect that force applied on the pedal would be minimized when saddle height was set to elicit 25-30° of knee flexion. Pedal force effectiveness has been used to express the percentage of the force applied on the pedal that creates propulsive torque on the crank. The greater the force effectiveness, the lower would be the waste of energy during cycling. The aim of the study was to compare pedal force effectiveness for different saddle height configurations (preferred, high, low, theoretical optimal).

Methods

Participants

Eleven cyclists and eleven triathletes with competitive experience were invited to participate in the study (mean ±SD: 41 ±11 years old, 74 ±14 kg, 8 ±4 week training hours, 31.5 ± 8 $km \cdot h^{-1}$ average speed at self-selected racing events) and signed an informed consent form in agreement with the research ethics committee of the institution where the study was conducted.

Protocol

At the start of the evaluation session body mass was measured (Secca scales) and self-reported age, week training hours and average speed at self-selected racing events were recorded. Participants' bicycle saddle height and horizontal position were measured to set up the stationary cycle ergometer (Velotron, Racemate, Inc) at the "preferred height" configuration. Knee joint flexion angle was measured using a goniometer while the participants held the pedal crank at the 6 o'clock position. Saddle height was recorded when the saddle was changed from the preferred position to high (-10° of knee flexion with respect to the preferred height), low (+10° of knee flexion with respect to the preferred height), and to the theoretical optimal (25° of knee flexion). Participants then performed 10 minutes of warm-up cycling at a self-selected cadence on the stationary Velotron cycle ergometer using their preferred saddle height. Workload was then increased to match 3.4 ±0.4 $W \cdot kg^{-1}$ (247 ±45 W) and pedalling cadence was visually controlled at 90 ±2 rpm for two minutes. Data were recorded during the first 20 s of the second minute for each saddle height trial. One minute of static rest was completed between trials with different saddle heights.

Data collection

Normal and anterior-posterior forces were measured using a strain gauge instrumented right pedal (Candotti, et al., 2007) and pedal-to-crank angle was measured using an angular potentiometer (Hull & Davis, 1981). A reed switch attached to the bicycle frame detected the position of the crank in relation to the pedal revolution (Bini, et al., 2008). All data were acquired at 600 Hz by an analog to digital converter (PCI-MIO-16XE-50, National Instruments, USA) using a custom Matlab (Mathworks Inc, MA) data acquisition script.

Data analysis

Forces on the pedal surface were resolved into the tangential force on the crank to compute force effectiveness (ratio between tangential resultant pedal forces) (Rossato, et al., 2008). Tangential (effective) force and resultant pedal force were averaged over ten complete pedal revolutions. Data analysis was conducted offline using a custom Matlab (Mathworks Inc, MA) analysis package.

Statistical analysis

Magnitudes of differences between the saddle heights were assessed by effect sizes (ES) for the average resultant pedal force and force effectiveness (Hopkins, 2002; Knudson, 2009). Effect sizes were rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004).

Results

In Table 13.1, means and standard deviations for knee flexion angle, saddle height, resultant force and force effectiveness of the 22 athletes for four saddle heights (preferred, high, low and optimal) are presented. Average preferred saddle height among cyclists and triathletes was 86.5 ±5.1 cm. To elicit -10° of knee flexion (high saddle height), changes of +3 ±1% of the preferred saddle height were conducted, while to achieve +10° of knee flexion (low saddle height) changes of -2 ±1% of the preferred height were conducted. For the optimal saddle height (25° of knee flexion) changes of +3 ±2% of the preferred height were conducted.

Table 13.1. Means and standard deviations for knee flexion angle (°), saddle height (% of preferred height), resultant force (N) and force effectiveness (%) of the 22 athletes for four saddle heights (preferred, high, low and optimal).

Variables	Preferred saddle height	High saddle height	Low saddle height	Optimal saddle height
Knee flexion angle at 6 o'clock pedal crank position (°)	37 ±3.6	26 ±3.6	46 ±3.5	25
Saddle height (%)	100	103 ±1.4	98 ±0.8	103 ±1.5
Resultant force (N)	160 ±33	156 ±35	163 ±36	159 ±36
Force effectiveness (%)	54 ±5	57 ±6	54 ±6	56 ±7

Changes in average resultant force with saddle height were trivial (1% for preferred versus optimal; ES = 0.2) to moderate (5% for high versus low; ES = 0.8). Changes in force effectiveness with saddle height were small (2% for preferred versus optimal; ES = 0.3) to moderate (6% for high versus low; ES = 1.0). The low saddle height condition produced higher resultant force but similar force effectiveness compared to the preferred height (see Table 13.2).

Table 13.2. Percentage differences and effect sizes for knee flexion angle, saddle height, resultant force and force effectiveness of the 22 athletes for four saddle heights (preferred, high, low and optimal).

Saddle height/ Variables	Preferred vs. High	Preferred vs. Low	Preferred vs. Optimal	High vs. Low	High vs. Optimal	Low vs. Optimal
Knee flexion angle at 6 o'clock pedal crank position (% change; ES)	38%; 2.4, large	22%; 2.4, large	45%; 2.7, large	43%; 4.8, large	5%; 0.3, small	85%; 5.2, large
Saddle height (% change; ES)	3%; 2.1, large	2%; 2.8, large	4%; 2.1, large	6%; 4.8, large	1%; 0.4, small	6%; 4.6, large
Resultant force (% change; ES)	3%; 0.4, small	2%; 0.4 small	1%; 0.2, trivial	5%; 0.8, moderate	2%; 0.2, trivial	3%; 0.5, moderate
Force effectiveness (% change; ES)	4%; 0.6, moderate	2%; 0.4, small	2%; 0.3, small	6%; 1.0, moderate	2%; 0.3, small	4%; 0.6, moderate

Discussion

Previous research suggested that optimal efficiency in cycling may be achieved when the saddle height is set to elicit a knee flexion angle of 25° when the pedal crank is at the 6 o'clock position (Peveler, 2008). We expected that pedal force application would also be optimized using this configuration for the saddle height. However, we found that preferred saddle height used by competitive cyclists and triathletes resulted in greater knee flexion (~37°) then that recommended to optimize efficiency. Only trivial changes in resultant force (-1%) and small improvements in force effectiveness (+2%) were found when the optimal saddle height was compared to the preferred saddle height.

Lowering the saddle height by approximately 2% of the current preferred saddle height elicited only small increases (2%) in resultant force applied on the pedal for cyclists and triathletes, which may not be expected to improve cycling efficiency. Hip and ankle joints would be expected to compensate for changes in saddle height to achieve similar power production of the knee joint muscles. Given the varied results for comparisons between the various saddle heights, further work is required to determine how best to set saddle height to improve both force effectiveness and resultant force.

Conclusion

Saddle height changes resulted in small to large effects for pedal resultant force and force effectiveness for most saddle height comparisons.

CHAPTER 14: SADDLE HEIGHT EFFECTS ON PEDAL FORCES, JOINT MECHANICAL WORK AND KINEMATICS OF CYCLISTS AND TRIATHLETES

Overview

Our study assessed the effects of saddle height on pedal forces, joint mechanical work and kinematics in 12 cyclists and 12 triathletes. Four sub-maximal 2-min cycling trials (3.4 W/kg and 90 rpm) were completed using preferred saddle height, low and high (±10° knee flexion at 6 o'clock crank position) saddle heights and an advocated optimal saddle height for cycling efficiency (25° knee flexion at 6 o'clock crank position). Right pedal forces via one instrumented pedal and lower limb kinematics via one high speed camera were recorded at each saddle height. Increases in saddle height (5% of preferred height) resulted in large increases in index of effectiveness (7%) at the optimal saddle height compared to the preferred saddle height for cyclists. Greater knee (9-15%) and smaller hip (5-8%) angles were observed for cyclists and triathletes at the low and preferred saddle heights (triathletes only) compared to high and optimal saddle heights. Smaller hip angle (5%) and greater hip range of motion (9%) were observed at the preferred saddle height for triathletes compared to cyclists. Changes in saddle height up to 5% of preferred saddle height for cyclists and 7% for triathletes affected hip and knee angles but not joint mechanical work.

Introduction

Optimising bicycle set-up may improve performance and decrease risk of overuse injuries (Burke & Pruitt, 2003). Saddle height has been reported as the most important characteristic of bicycle configuration (Silberman, et al., 2005) as it affects lower limb joint kinematics (Desipres, 1974; Diefenthaeler, et al., 2006; Nordeen-Snyder, 1977; Rankin & Neptune, 2008), muscle length (Rugg & Gregor, 1987), and muscle activation (Sanderson & Amoroso, 2009). However, the effects of saddle height on pedal forces and joint kinetics (e.g. mechanical work) are unclear (Bini, Tamborindeguy, et al., 2010; Ericson & Nisell, 1988; Horscroft, Davidson, McDaniel, Wagner, & Martin, 2003), precluding a definition of an optimal saddle height for enhancing performance.

In non-athletes, pedal forces (Ericson & Nisell, 1988) and joint mechanical work (Bini, Tamborindeguy, et al., 2010; Horscroft, et al., 2003) have not significantly changed when saddle height was varied. Changes in saddle height smaller than ±4% of trochanteric leg length may not result in substantial differences in pedal forces and joint mechanical work and therefore cycling performance may not be affected (Bini, Hume, et al., 2011a). However, different methods of saddle height configuration (e.g. inseam leg length vs. knee flexion angle method) have been used in studies to date, which may result in different joint kinematics (Peveler, et al., 2005). Consequently, direct comparisons between studies are problematic and it is not clear if the lack of variation in pedal forces and joint mechanical work is due to inconsistencies in joint kinematics, based on using different methods, or if a change in saddle height of less than ±4% of the trochanteric leg length does not result in substantial changes in pedal forces and joint mechanical work.

Changes in saddle height will affect hip, knee and ankle joint angles and therefore muscle force-length, force-velocity (Sanderson & Amoroso, 2009) and power (or mechanical work) should be affected. Bini et al. (2010) and Horscroft et al. (2003) showed that power produced by the hip, knee and ankle joints dictated power output during seated cycling and that power from the individual joints may not be sensitive to saddle height effects. However, these studies involved few cyclists (Horscroft, et al., 2003) or non-athletes (Bini, Tamborindeguy, et al., 2010) and so further analysis of these variables, using a larger sample of competitive cyclists, is required.

The majority of studies have focussed on the effect of saddle height in road cyclists. However, triathletes have been shown to differ from road cyclists in terms of pedal force effectiveness (Candotti, et al., 2007) and muscle activation (Candotti, et al., 2009; Chapman, et al., 2007). Therefore, it is expected that cyclists and triathletes may also differ in their joint kinematics and that each may have a particular adaptation using different configurations for saddle height. No comparison between cyclists and triathletes has been conducted in terms of joint mechanical work.

The purpose of our study was to assess the effects of saddle height on pedal forces, joint mechanical work, and kinematics in cyclists and triathletes. It was hypothesised that saddle height would have a large influence on joint kinematics, but not on joint mechanical work or pedal forces. The reason for this is that when saddle height is varied, joint mechanical work may be balanced among the hip, knee and ankle joints, without a specific effect at a single joint. In other words, even with large changes in joint kinematics, either muscle capacity to generate power at the hip, knee and ankle joints may not be substantially affected or individual changes in muscle capacity to generate power (e.g. lower knee joint extensors power) may be balanced by hip and/or ankle joint muscles when saddle height is changed (e.g. increase hip and/or ankle joint power).

Methods

Participants

Twelve cyclists and twelve triathletes with competitive experience participated in our study. Cyclists had more cycling training volume (hours and distance of training per week) compared to triathletes (see Table 14.1). Large differences were highlighted in bold italics.

Table 14.1. Characteristics (mean ±SD) of age, body mass, height, time of training and training volume of 12 cyclists and 12 triathletes. Differences between cyclists and triathletes are reported as mean difference percentages along with effect size magnitudes.

Groups	Age (years)	Body mass (kg)	Height (cm)	Time of cycling training (hours/week)	Training volume (km/week)
Cyclists	36 ±14	77 ±14	179 ±5	9 ±5	180 ±22
Triathletes	42 ±8	74 ±16	176 ±10	6 ±2	112 ±56
Cyclists vs. triathletes	15%; 0.6 moderate	5%; 0.3 small	1%; 0.3 small	***57%; 1.0 large***	***61%; 1.7 large***

Prior to the study, the participants were informed about possible risks and signed a consent form approved by the ethics committee of human research at the institution where the study was conducted.

Data collection

Upon arriving at the laboratory height and body mass measures were taken following ISAK protocols (Marfell-Jones, et al., 2006). Each athlete's bicycle vertical and horizontal position of the handlebars were measured to set up the stationary cycle ergometer (Velotron, Racemate, Inc) at their "preferred height" configuration. Saddle height was measured from the central portion of the top of the saddle to the pedal spindle with the crank in line with the seat tube angle (Bini, Hume, et al., 2011a) in each athlete's bicycle along with the horizontal position of the saddle to the bottom bracket. These measures were replicated in the cycle ergometer to simulate the configurations used for cyclists and triatheltes in their bicycles. Knee joint flexion angle was measured using a goniometer with the crank held at the 6 o'clock position. Saddle height was recorded when the saddle was changed from the preferred position to high (-10° knee flexion with respect to the preferred height), low (+10° knee flexion with respect to the preferred height), and to the theoretical optimal (25° knee flexion). The latter saddle height was included in the study as it has been previously reported to optimize cycling efficiency (Peveler, 2008).

Data analyses

Video files were digitized and automatic tracking of markers was conducted in DgeeMe software (Video4Coach, Denmark) for x-y coordinates over time. Kinematics and force data were smoothed with a digital second order zero lag low pass Butterworth filter, with cut-off frequency optimized to reduce signal residuals (Winter, 2005). Joint angles of the hip, knee and ankle during pedalling movement were calculated from the smoothed x-y coordinate data, as per the spatial model shown in Figure 14.1. From each pedal revolution, the mean value and range of motion of the hip, knee and ankle joints were calculated over time (Bini, Diefenthaeler, et al., 2010).

Correction of the hip joint centre based on the average coordinate between the marker on the anterior superior iliac spine and the greater trochanter was performed (Neptune & Hull, 1995). The average relative horizontal position of the marker on the sacrum to the bottom bracket was computed over ten pedal revolutions, for the analysis of body position on the saddle at the four saddle heights.

Linear and angular velocities and accelerations were computed from smoothed kinematic data by a three points derivative method (Winter, 2005). Pedal angle in relation to the global coordinate system was calculated to convert the forces on the pedal reference system to forces in the global reference system by means of trigonometric procedures (Marsh, et al., 2000). The right lower limb was modelled as a three-segment rigid body system (thigh, shank and foot-pedal) with segment mass and center of mass estimated according to De Leva (1996). Conventional inverse dynamics were used to calculate the net joint moments at the hip, knee and ankle (Redfield & Hull, 1986), using adapted scripts of van den Bogert and de Koning (1996). Net joint mechanical work was calculated by integrating joint power (moment x angular velocity at the joints) with respect to time, and relative contributions of the ankle, knee and hip joints were calculated as a percentage of total mechanical work at three joints (Bini, Rossato, et al., 2010). From each pedal revolution, the mean value and range of motion of the hip, knee and ankle joints were calculated over time. Pedal force effectiveness and pedal force application were computed from the overall index of effectiveness (ratio between the tangential force on the crank and the total force on the pedal surface) and the average total force applied on the pedal respectively (Rossato, et al., 2008). All variables were processed using a custom written program in MATLAB® (MathWorks Inc, USA) for ten consecutive crank revolutions to determine means and standard deviations for each cyclist and triathlete.

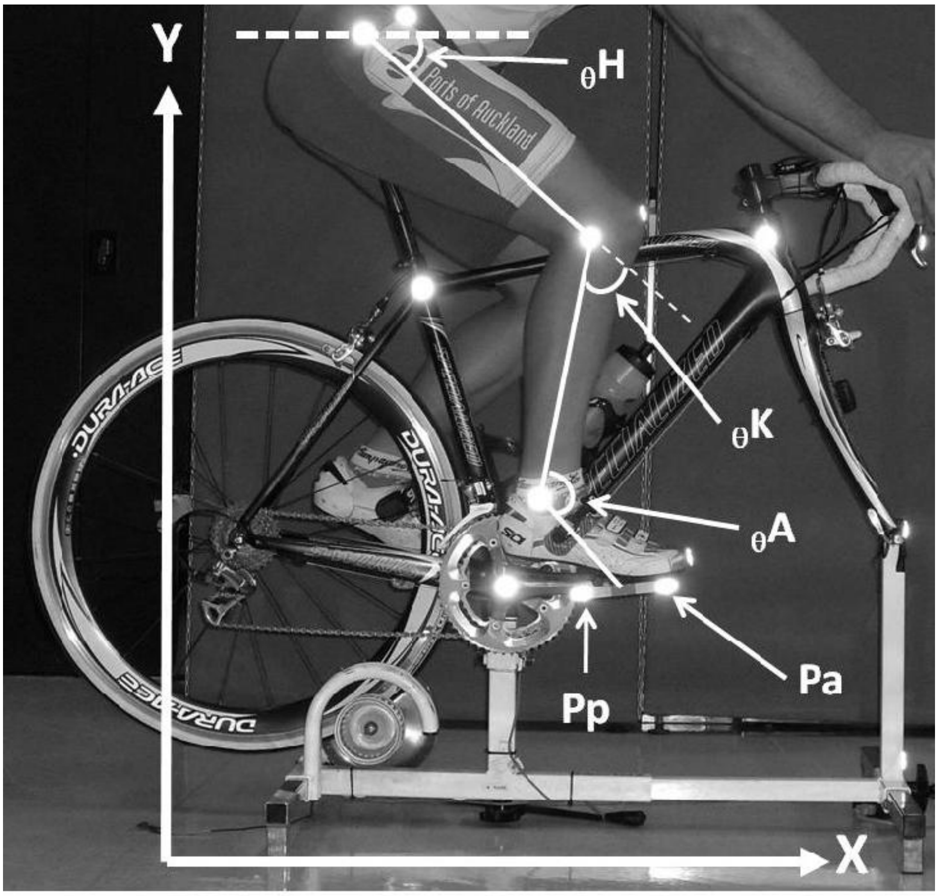


Figure 14.1. Illustration of reflective marker placement on the right side of the cyclist at the anterior superior iliac spine, sacrum, greater trochanter, lateral femoral condyle and lateral malleolus to measure hip ($_{\theta}$H), knee ($_{\theta}$K) and ankle ($_{\theta}$A) joint angles. Reflective markers were attached to the anterior (Pa) and posterior (Pp) extremities of the reference stick attached to the pedal axis for computation of pedal force components into the global coordinate system.

Statistical analyses

Cyclists and triathletes' characteristics (body mass, height, age, time of training and training volume) were grouped as means and standard deviations and compared using Cohen's effect sizes (ES). Means and standard deviations were calculated for the average total force applied on the pedal, the index of effectiveness, the mean angle, range of motion and mechanical work at the hip, knee and ankle joints for cyclists and triathletes. Normality of distribution and sphericity were evaluated via the Shapiro-Wilk and Mauchly tests respectively. When the assumption of data normality was violated, a logarithmic transformation was applied for the index of effectiveness, and relative contributions of the ankle, knee and hip joints to the total mechanical work. Force variables and joint mechanical work were normalized by individual workload level (in Joules).

To compare the effects of saddle height for cyclists and triathletes on the dependent variables, Cohen's effect sizes (ES) were computed for the analysis of the magnitude of the differences and subsequently rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Advocated optimal saddle height resulted in increased index of effectiveness compared to the preferred saddle height for cyclists. No substantial changes were observed in total pedal force or index of effectiveness (triathletes only) when saddle height was changed or when comparing cyclists and triathletes (see Table 14.2).

Large decreases in ankle range of motion and mechanical work were observed for triathletes at the low saddle height compared to the optimal saddle height. Increases in knee mean angles and decreases in hip mean angles were observed for cyclists and triathletes at the low and preferred (triathletes only) compared to high and optimal saddle heights. Smaller hip mean angle and greater hip range of motion at the preferred saddle height were observed for triathletes compared to cyclists (see Table 3).

Table 14.2. Means and standard deviations for saddle height, total force applied on the pedal and index of effectiveness for four saddle heights (preferred, high, low and optimal) for cyclists and triathletes. Differences between cyclists and triathletes (in italics), and differences between saddle heights within a group, are reported as mean difference percentages along with effect size magnitudes. Abbreviations used for comparisons are for triathletes (Tri), preferred saddle height (Pref) and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

	Cyclists (n = 12)				**Triathletes (n = 12)**			
	Optimal	**High**	**Preferred**	**Low**	**Optimal**	**High**	**Preferred**	**Low**
Saddle height (cm)	88 ±2.9 High 1%, 0.1, T ***Pref 3%; 2.2, L*** ***Low 5%; 4.6, L*** *Tri 1%; 0.2, T*	88 ±3.3 ***Pref 3%; 2.0, L*** ***Low 5%; 4.4, L*** *Tri 1%; 0.1, T*	86 ±3.1 ***Low 2%; 2.4, L*** *Tri 1%; 0.1, T*	84 ±3.3 *Tri 1%; 0.2, T*	89 ±6.5 High 1%; 0.8, M ***Pref 4%; 2.3, L*** ***Low 7%; 5.3, L***	88 ±7.3 ***Pref 3%; 2.0, L*** ***Low 3%; 5.1, L***	85 ±6.8 ***Low 3%; 3.6, L***	83 ±6.9
Total pedal force (% of workload)	101 ±19 High 3%; 0.1, T Pref 1%; 0.1, T Low 5%; 0.3, S *Tri 2%; 0.1, T*	99 ±16 Pref 2%; 0.2, T Low 7%; 0.4, S *Tri 4%; 0.2, T*	101 ±13 Low 5%; 0.3, S *Tri <1%; 0.1, T*	106 ±18 *Tri 7%; 0.4, T*	100 ±17 High 5%; 0.3, S Pref 2%; 0.1, T Low 1%; 0.1, T	95 ±16 Pref 6%; 0.4, S Low 4%; 0.3, S	101 ±16 Low 2%; 0.1, T	99 ±15
Index of effectiveness (%)	63 ±7 High 1%; 0.1, T ***Pref 7%; 1.2, L*** Low 9%; 0.8, M *Tri 3%; 0.2, T*	63 ±6 Pref 8%; 0.7, M Low 6%; 0.7, M *Tri <1%; 0.1, T*	59 ±6 Low 2%; 0.3, S *Tri 5%; 0.4, S*	60 ±5 *Tri 3%; 0.3, S*	65 ±9 High 3%; 0.3, S Pref 5%. 0.5, M Low 6%; 0.7, M	63 ±6 Pref 2%; 0.4, S Low 3%; 0.4, S	62 ±7 Low 1%; 0.1, T	62 ±9

Table 14.3. Means and standard deviations for mean angle, range of motion and mechanical work of the hip, knee and ankle joints for four saddle heights (preferred, high, low and optimal) for cyclists and triathletes. Differences between cyclists and triathletes (in italics), and differences between saddle heights within a group, are reported as mean difference percentages along with effect size magnitudes. Abbreviations used for comparisons are for triathletes (Tri), preferred saddle height (Pref) and effect sizes of trivial (T), small (S), moderate (M) and large (L).

		Cyclists (n = 12)				**Triathletes (n = 12)**			
		Optimal	**High**	**Preferred**	**Low**	**Optimal**	**High**	**Preferred**	**Low**
Hip	mean angle (°)	66 ±4 High 1%; 0.1, T Pref 3%; 0.5, M ***Low 6%; 1.1, L*** *Tri 2%; 0.4, S*	66 ±4 Pref 3%; 0.5, M ***Low 6%; 1.1, L*** *Tri 3%; 0.4, S*	64 ±3 Low 4%; 0.7, M *Tri 5%; 1.0, L*	62 ±3 *Tri 5%; 0.9, M*	65 ±3 High 1%; 0.1, T ***Pref 5%; 1.1, L*** ***Low 8%; 1.7, L***	64 ±3 ***Pref 5%; 1.0, L*** ***Low 9%; 1.6, L***	61 ±3 Low 3%; 0.7, M	59 ±3
	range of motion (°)	45 ±4 High 1%; 0.1, T Pref 5%; 0.2, T Low 5%; 0.6, M *Tri 7%; 0.6, M*	45 ±5 Pref 1%; 0.2, T Low 5%; 0.5, M *Tri 6%; 0.6, M*	45 ±3 Low 2%; 0.4, S ***Tri 9%; 1.0, L***	47 ±4 *Tri 8%; 0.9, M*	48 ±5 High 1%; 0.1, T Pref 4%; 0.4, S Low 7%; 0.6, M	48 ±5 Pref 5%; 0.4, S Low 6%; 0.6, M	50 ±5 Low 2%; 0.2, T	51 ±5
	work (% of workload)	30 ±5 High 1%; 0.1, T Pref 8%; 0.3, S Low 12%; 0.3, S *Tri 11%; 0.3, S*	30 ±6 Pref 8%; 0.3, S Low 11%; 0.3, S *Tri 28%; 0.6, M*	33 ±5 Low 3%; 0.1, T *Tri 44%; 0.9, M*	33 ±7 *Tri 43%; 0.7, M*	28 ±8 High 12%; 0.4, S Pref 11%; 0.4, S Low 12%; 0.3, S	27 ±9 Pref 2%; 0.1, T Low 6%; 0.1, T	28 ±10 Low 6%; 0.1, T	29 ±11

	mean angle (°)	64 ±4 High 1%; 0.1, T Pref 6%; 0.9, M ***Low 11%; 1.6, L*** *Tri 1%; 0.1, T*	64 ±5 Pref 7%; 0.9, M ***Low 11%; 1.5, L*** *Tri 1%; 0.2, T*	68 ±4 Low 5%; 0.8, M *Tri 3%; 0.5, M*	72 ±5 *Tri 3%; 0.6, M*	64 ±4 High 1%; 0.1, T ***Pref 9%; 1.6, L*** ***Low 15%; 2.7, L***	65 ±3 ***Pref 8%; 1.5, L*** ***Low 12%; 2.6, L***	70 ±3 ***Low 5%; 1.1, L***	74 ±3
Knee	range of motion (°)	78 ±3 High 1%; 0.1, T Pref 8%; 0.9, M ***Low 7%; 1.7, L*** *Tri 3%; 0.6, M*	78 ±3 Pref 4%; 0.9, M ***Low 8%; 1.7, L*** *Tri 2%; 0.4, S*	75 ±3 Low 4%; 0.8, M *Tri 1%; 0.1, T*	73 ±3 *Tri 1%; 0.1, T*	80 ±5 High 1%; 0.1, T ***Pref 6%; 1.0, L*** ***Low 10%; 1.7, L***	80 ±5 Pref 6%; 0.9, M ***Low 11%; 1.6, L***	75 ±5 Low 4%; 0.6, M	72 ±5
	work (% of workload)	61 ±4 High 1%; 0.1, T Pref 2%; 0.1, T Low 4%; 0.1, T *Tri 7%; 0.4, S*	61 ±5 Pref 1%; 0.1, T Low 3%; 0.1, T *Tri 12%; 0.7, M*	61 ±11 Low 2%; 0.1, T *Tri 16%; 0.8, M*	59 ±6 *Tri 17%; 0.9, M*	64 ±7 High 1%; 0.3, S Pref 3%; 0.6, M Low 3%; 0.8, M	65 ±8 Pref 3%; 0.3, S Low 3%; 0.5, M	65 ±9 Low 3%; 0.2, T	64 ±9

	mean angle (°)	119 ±5 High 1%; 0.1, T Pref 3%; 0.6, M Low 4%; 0.8, M *Tri 2%; 0.5, M*	120 ±5 Pref 3%; 0.6, M Low 4%; 0.8, M *Tri 2%; 0.4, S*	116 ±5 Low 1%; 0.2, T *Tri 4%; 0.8, M*	115 ±6 *Tri 3%, 0.5, M*	122 ±5 High 1%; 0.1, T Pref 1%; 0.2, T Low 3%; 0.7, M	122 ±6 Pref 1%; 0.2, T Low 4%; 0.7, M	121 ±6 Low 2%; 0.4, S	118 ±6
Ankle	range of motion (°)	24 ±6 High 1%; 0.1, T Pref 12%; 0.4, S Low 12%; 0.4, S *Tri 8%; 0.3, S*	24 ±6 Pref 8%; 0.3, S Low 12%; 0.4, S *Tri 11%; 0.4, S*	22 ±5 Low 9%; 0.1, T *Tri 24%; 0.8, M*	21 ±8 *Tri 33%; 0.8, M*	22 ±6 High 4%; 0.1, T Pref 21%; 0.7, M ***Low 29%; 1.0, L***	21 ±7 Pref 18%; 0.6, M Low 35%; 0.8, M	17 ±6 Low 11%; 0.3, S	16 ±6
	work (% of workload)	9 ±3 High 2%; 0.1, T Pref 12%; 0.3, S Low 10%; 0.3, S *Tri 29%; 0.7, M*	9 ±3 Pref 10%; 0.3, S Low 9%; 0.2, T *Tri 29%; 0.9, M*	8 ±2 Low 2%; 0.1, T *Tri 38%; 1.1, L*	8 ±3 *Tri 63%; 1.1, L*	7 ±2 High 2%; 0.1, T Pref 17%; 0.7, M ***Low 28%; 1.1, L***	7 ±2 Pref 16%, 0.7, M ***Low 37%; 1.1, L***	6 ±1 Low 16%; 0.4, S	5 ±2

Discussion

We compared total pedal force, index of effectiveness and hip, knee and ankle kinematics and mechanical work of cyclists and triathletes using different saddle heights. Our hypothesis was that saddle height would have a large influence on joint kinematics, but not on joint mechanical work or pedal forces. Due to joint mechanical work at different saddle heights potentially being balanced among the three lower limb joints, without specific effects on a single joint. Our results partially supported this hypothesis because we observed substantial changes in hip and knee joint angles, particularly for triathletes who also presented changes in ankle joint mechanical work (reduced at lower saddle heights).

Our changes in saddle height were up to 5% for cyclists and 7% for triathletes, which resulted in greater knee range of motion for cyclists (7%) and triathletes (10%). These results are in line with, but of smaller magnitude, to the work of Sanderson and Amoroso (2009) who reported that a 5% increase in saddle height resulted in a 25% greater knee range of motion for cyclists. Cyclists only presented large differences between optimal saddle height compared to the low saddle height for knee mean angle and range of motion and for hip mean angle in our study. Sanderson and Amoroso (2009) reported substantial effects of saddle height in knee joint kinematics for competitive cyclists, which our results also showed. Triathletes presented differences for hip and knee joints mean angle and knee range of motion. These results are contrary to those of previous studies which showed that the ankle (Bini, Tamborindeguy, et al., 2010; Nordeen-Snyder, 1977; Price & Donne, 1997) was the most affected joint when changing saddle height for cyclists and non-athletes. Differences in joint kinematics were more evident comparing low to high and optimal saddle height for cyclists, rather than changes from preferred to other saddle height, which were only observed for triathletes. The preferred saddle height resulted in greater knee mean angle and smaller knee range of motion, and smaller hip mean angle for triathletes than cyclists. Cyclists' knee and hip angles were sensitive to changes of ~5% of preferred saddle height and triathletes presented large differences when changes of 3-4% were conducted in saddle height.

It is unclear how previous studies have shown significantly better efficiency when cyclists used a saddle height that elicited 25° knee flexion (optimal height in our study), compared to a saddle height that elicited 35° knee flexion (similar to preferred height in our study) (Peveler, 2008; Peveler & Green, 2011). Differences from using a saddle height that elicited 25° knee flexion compared to the saddle height that elicited 35° knee flexion were trivial (effects sizes 0.07-0.20) (Peveler, 2008; Peveler & Green, 2011) so it was unclear how substantial the changes could be from a practical perspective.

Joint mechanical work would be expected to change due to changes in knee and hip joint angles, but that was not the case. Triathletes presented reduced ankle mechanical work at the low saddle height compared to the high and optimal saddle heights possibly due to smaller ankle range of motion at the low saddle height. These results are partially contrary to previous findings from cyclists (Horscroft, et al., 2003) and non-cyclists (Bini, Tamborindeguy, et al., 2010), who presented greater knee mechanical work at lower saddle heights (6% change in saddle height for both studies). For non-athletes, ankle work was lower at the low saddle height compared to high saddle height (6% change in saddle height) (Bini, Tamborindeguy, et al., 2010), similar to findings for triathletes in our study. Adaptation to changes in saddle height may be similar comparing triathletes and non-athletes in relation to cyclists.

Although effects were observed in joint kinematics when saddle height was changed, pedal forces were only affected in cyclists (lower index of effectiveness for preferred than optimal saddle height). Cyclists and triathletes seem to adapt to changes in saddle height to sustain similar pedal force application. Muscle tendon unit length has been reported to change depending on saddle height (Rugg & Gregor, 1987), which was then expected to affect muscle force production and pedal force application. However, our results did not provide evidence for this association. A previous study on pedal forces of non-athletes assessed using different saddle heights (Ericson & Nisell, 1988) supported our results of no changes in pedal force application using different saddle heights. We can infer that, even with large changes in joint kinematics, either muscle capacity to generate power at the hip, knee and ankle joints may not be substantially affected or individual changes in muscle capacity to generate power (e.g. lower knee joint extensors power) may be balanced by hip and/or ankle joint muscles when saddle height is changed.

Cyclists and triathletes differ in terms of pedal forces (Candotti, et al., 2007) and muscle activation (Candotti, et al., 2009; Chapman, et al., 2007), however, it has been unclear how these groups of athletes differ in relation to joint kinetics and kinematics. Triathletes presented smaller ankle work and hip mean angle, and greater hip range of motion compared to cyclists at the preferred saddle height in our study. Interestingly, triathletes were less sensitive to changes in saddle height than cyclists. For example, when saddle height was changed from preferred to optimal (3%) only cyclists' index of effectiveness was affected (7% increase). This apparent position sensitivity of cyclists could be due to several factors. Firstly, cyclists may tend to change their position on the bicycle more often than triathletes. Cycling races are varied in terms of incline and distance, which results in greater changes in body position on the bicycle in different stages of the race (e.g. standing pedalling during uphill). In contrast, triathletes perform time trials of varying distances (from 20 km to 180 km) during competition mostly seated on the bicycle with the arms lying on the aerobars. This position is chosen to reduce drag forces because, different from cyclists, triathletes are usually not allowed to ride in groups during long racing (i.e Ironman). Secondly, triathletes from our study presented less weekly volume of cycling training compared to cyclists potentially because they share their training time between swimming, cycling and running, which offer different load profiles to lower limb muscles (Savelberg & Meijer, 2003; Suriano & Bishop, 2010). Therefore, triathletes would be expected to present increases in muscle force across a larger range of muscle lengths and potentially greater adjustment to different muscle lengths (e.g. when changing saddle height) because they complete running and swimming training at different muscle lengths for force production (e.g. shorter knee extensors) compared to cycling. For that reason, triathletes from our study presented greater adaptation to changes in saddle height compared to cyclists, with changes observed in pedal forces (i.e. index of effectiveness) only being observed for cyclists.

Conclusion

Changes in saddle height up to 5% of preferred saddle height for cyclists and 7% for triathletes affected hip and knee angles. In general, higher saddle height resulted in smaller knee angle and greater knee range of motion and hip mean angle. Cyclists presented improved index of effectiveness at the optimal saddle height compared to the preferred saddle height and triathletes presented greater ankle work and ankle range of motion for the optimal saddle height compared to the low saddle height. Triathletes presented greater mechanical work and range of motion, and small mean angle for the hip joint compared to cyclists. There was greater adaptation of triathletes to changes in saddle height compared to cyclists leading to similar pedal forces.

CHAPTER 15: EFFECTS OF MOVING FORWARD OR BACKWARD ON THE SADDLE ON KNEE JOINT FORCES DURING CYCLING

Overview

Effects of cycling at preferred, forward and backward saddle positions on patellofemoral compressive and tibiofemoral compressive and shear forces were compared. Twenty-one competitive cyclists (28 ±7 years) performed an incremental cycling test to exhaustion determined cyclists' maximal power output and second ventilatory threshold. In a second session, 1-minute cycling trial at maximal power output then three 2-minute trials at second ventilatory threshold workload at preferred, forward and backward saddle positions. Right pedal force via instrumented pedals, lower limb joint kinematics via video and inverse dynamics calculated knee joint forces. Patellofemoral compressive, tibiofemoral compressive and shear forces, and knee flexion angle. Changes to forward/backward saddle positions did not substantially affect compressive forces for patellofemoral (1-4%) or tibiofemoral (1-3%) joints. Tibiofemoral shear force increased the backward compared with preferred or forward saddle positions (19-45%). Knee flexion angle at 3 o'clock (22%) and 6 o'clock crank positions (36%) increased at the forward compared to the backward saddle position. Small increases in knee flexion angle (5-6°) explained the trivial differences in patellofemoral and tibiofemoral compressive forces. Tibiofemoral shear force may be more sensitive to changes in knee joint angle compared to other knee force components.

Introduction

The existing rationale for the link between body position on the bicycle and overuse knee injuries is based on the application of repetitive load at improper greater knee flexion angles resulting in soft tissue damage (Callaghan, 2005). In this regard, attention has been given to the position of the cyclist on the bicycle saddle and changes in knee joint forces (Ericson & Nisell, 1986, 1987) and kinematics (Nordeen-Snyder, 1977; Tamborindeguy & Bini, 2011). Greater knee joint force may cause overload at the patellofemoral cartilage, menisci, and at the anterior and posterior cruciate ligaments (Neptune & Kautz, 2000). Theoretically, with a reduced moment arm (smaller patellar tendon angle from a greater knee flexion angle), greater force should be required for the quadriceps muscles to overcome the same moment at the knee joint, which may lead to higher contact forces at the tibiofemoral and patellofemoral joints (see Figure 15.1).

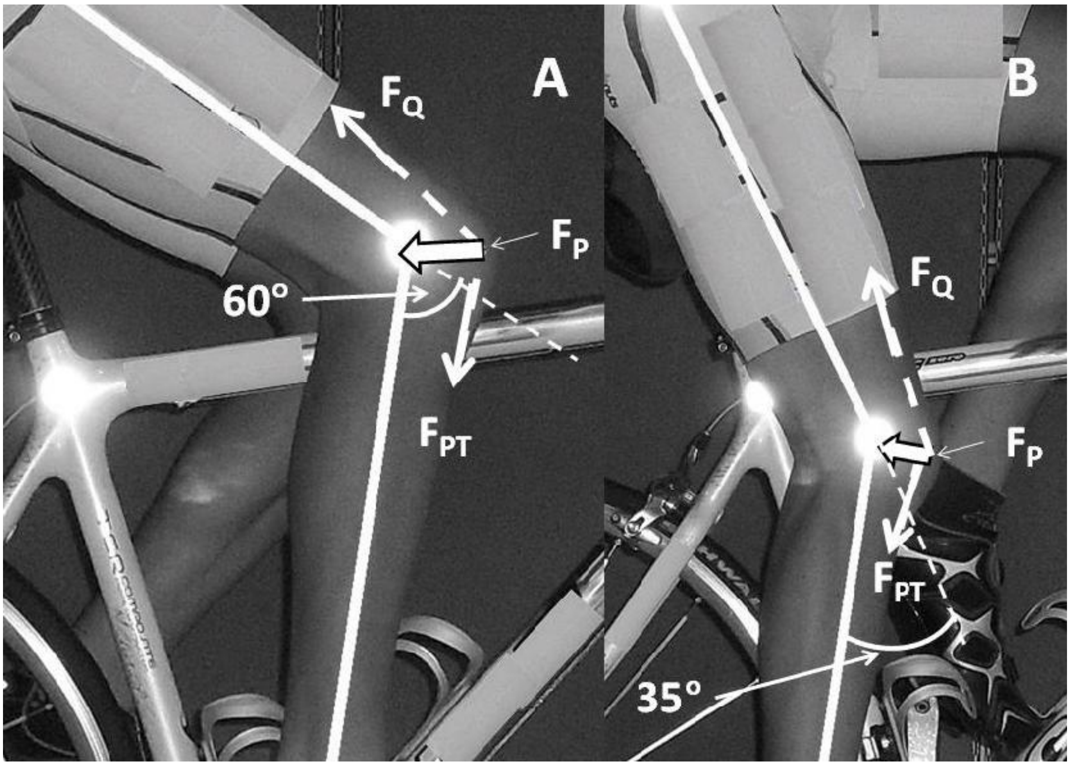


Figure 15.1. Schematic illustration of 25° of change in the knee flexion angle from 60° (A) to 35° (B) that should theoretically decrease patellofemoral compressive force (F_P). Arrows indicate quadriceps muscle force (F_Q) and patellar tendon force (F_{PT}).

Using the prediction model of Herzog and Reid (1993), the change in knee angle illustrated in Figure 15.1 (60° to 35°) would result in an approximately 7° greater patellar tendon angle to the axis of the tibia and 5% greater moment arm for the patellar tendon at a smaller knee flexion angle. High quadriceps force would be predicted during cycling at a lower saddle height and has been linked to large patellofemoral compressive force (Ericson & Nisell, 1987) but not with larger tibiofemoral compressive force (Ericson & Nisell, 1986; Tamborindeguy & Bini, 2011). A reduced moment arm (via smaller patellar tendon angle) would reduce the anterior shear force on the tibiofemoral joint and increase the compressive tibiofemoral force, which should minimize the tension at the anterior cruciate ligament.

Cyclists and triathletes usually move on the saddle depending on the characteristics of the race. When sprinting, cyclists tend to project the body forward on the bicycle or even sit in the most forward position possible on the saddle, which is commonly observed during time trial events and during the cycling leg of triathlon (Ricard, et al., 2006). However, to reduce the risk of patellofemoral overuse injuries it has been recommended that the vertical projection of the knee should not be ahead of the pedal spindle when the crank is at the 3 o'clock position (Callaghan, 2005; Silberman, et al., 2005; Wanich, et al., 2007). Therefore, the forward projection of the knee ahead of the pedal spindle resulting from a forward position of the body on the saddle for triathletes could result in risk of overuse patellofemoral injuries. Greater compressive tibiofemoral and patellofemoral forces and lower anterior force on the tibiofemoral joint would be expected when sitting forward on the saddle for a given workload (i.e. similar pedal reaction force). Patellofemoral contact area would be predicted to increase at the more forward position on the saddle due to greater knee flexion angle (Salsich, Ward, Terk, & Powers, 2003), which would reduce the pressure (force per unit of area) on the patellofemoral joint and minimize the detrimental effects of compressive force to the cartilage matrix (Cohen, et al., 2001). Therefore, the magnitude of changes in patellofemoral compressive force and tibiofemoral force components (compressive and shear) should be assessed by experimental research examining cyclists in a forward position on the saddle.

The aim of this study was to compare patellofemoral compressive force and tibiofemoral compressive and anterior shear forces at three positions on the saddle: 1) Preferred (self-selected); 2) Most forward; 3) Most backward. It was predicted that a forward saddle position would increase patellofemoral and tibiofemoral compressive forces and would reduce tibiofemoral anterior shear force. Opposite changes were predicted for the backward saddle position.

Methods

Participants

Twenty one cyclists (average ± SD 28 ±7 years, 73 ±8.2 kg, 179 ±8.8 cm, 61.4 ±5.8 $ml \cdot kg^{-1} \cdot min^{-1}$ maximal oxygen uptake, 381.7 ±37.0 W peak power output, 5.3 ±0.4 $W \cdot kg^{-1}$ peak power to mass ratio, 316 ± 85 $km \cdot week^{-1}$ cycling training volume) with competitive experience in cycling or triathlon participated in the study. Cyclists were informed about possible risks and signed a consent form approved by the ethics committee of human research where the study was conducted.

Data collection

During the first session height and body mass were measured according to ISAK protocols (Marfell-Jones, et al., 2006). Cyclists warmed up at 150 W for 10 minutes followed by an incremental maximal step test to exhaustion with initial work load of 100 W and increments of 25 $W \cdot min^{-1}$ (Lucía, et al., 2002) using their own bicycles on a Computrainer cycle simulator (RaceMate, USA). Power output was measured throughout the incremental test along with gas exchange by the breath-by-breath method using an open-circuit gas exchange system (MGC CPX/D, Medical Graphics Corp., St Louis, MO, USA). Pedalling cadence was controlled close to 90 ±2 rpm for all cyclists and triathletes using visual feedback from the cycle simulator head unit. The test was stopped by the cyclists' voluntary exhaustion or when they were unable to maintain pedalling cadence. The oxygen and carbon dioxide analysers were calibrated using calibrated medical grade gases that spanned air in the physiological range. Gas exchange data were analyzed to define the ventilatory threshold based on the ventilatory equivalent method (Weston & Gabbett, 2001).

After 48 hours the cyclists warmed up for 10 min at 150 W and then rode for one minute with 90 rpm pedalling cadence maximal power output from the incremental test in their preferred saddle position, then at a workload set to the second ventilatory threshold in the three saddle positions: preferred, most forward position, most backward. For the forward and backward saddle positions cyclists were instructed to simulate a position similar to time-trial and hill climb cycling, respectively, without substantial changes in the upper body lean position. The order of the forward and the backward saddle positions was allocated randomly.

Forces applied on the right pedal and right lower limb kinematics were recorded for the last 20 s during all conditions. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of the cyclists at the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, anterior and posterior pedal stick. One marker was attached to the sacrum to measure the position of the cyclists on the bicycle when they were evaluated at the forward and backward saddle positions. Two markers were taped to the bicycle frame and used as reference for image calibration. A 2D pedal dynamometer (Nabinger, et al., 2002) and one high speed camera positioned perpendicular to the motion plane (AVT PIKE F-032, Allied Vision Technologies GmbH, Germany) were synchronized by an external trigger. Kinematics were recorded at 60 Hz using AVT ActiveCam viewer software (Allied Vision Technologies GmbH, Germany) and force data were recorded at 600 Hz per channel employing a 16-bit analogical to digital converter (DI220, Dataq Instruments, USA) using WINDAQ® software (WINDAQ, DataQ Instruments Inc., USA).

Data analyses

Video files were digitized and automatic tracking of markers were conducted in DgeeMe software (Video4Coach, Denmark) for x-y coordinates over time. Kinematic data were smoothed with a digital second order zero lag band pass Butterworth filter with cut-off frequency optimized to reduce signal residual (Winter, 2005). Segment kinematics of the hip, knee, and ankle joints during pedalling movement were calculated from the smoothed x-y coordinate data. Relative angle between the thigh and the shank segments was defined as the knee angle and converted to flexion angle (full leg extension = zero knee flexion). Correction of the hip joint centre based on the average coordinate between the marker on the anterior superior iliac spine and the greater trochanter was conducted (Neptune & Hull, 1995). The average relative horizontal position of the marker on the sacrum to the bottom dead centre was computed over time across ten pedal revolutions for the analysis of body position on the saddle at the three saddle positions (preferred, forward and backward). In Figure 2 the representative displacement of the markers is shown for one cyclist for the three saddle positions.

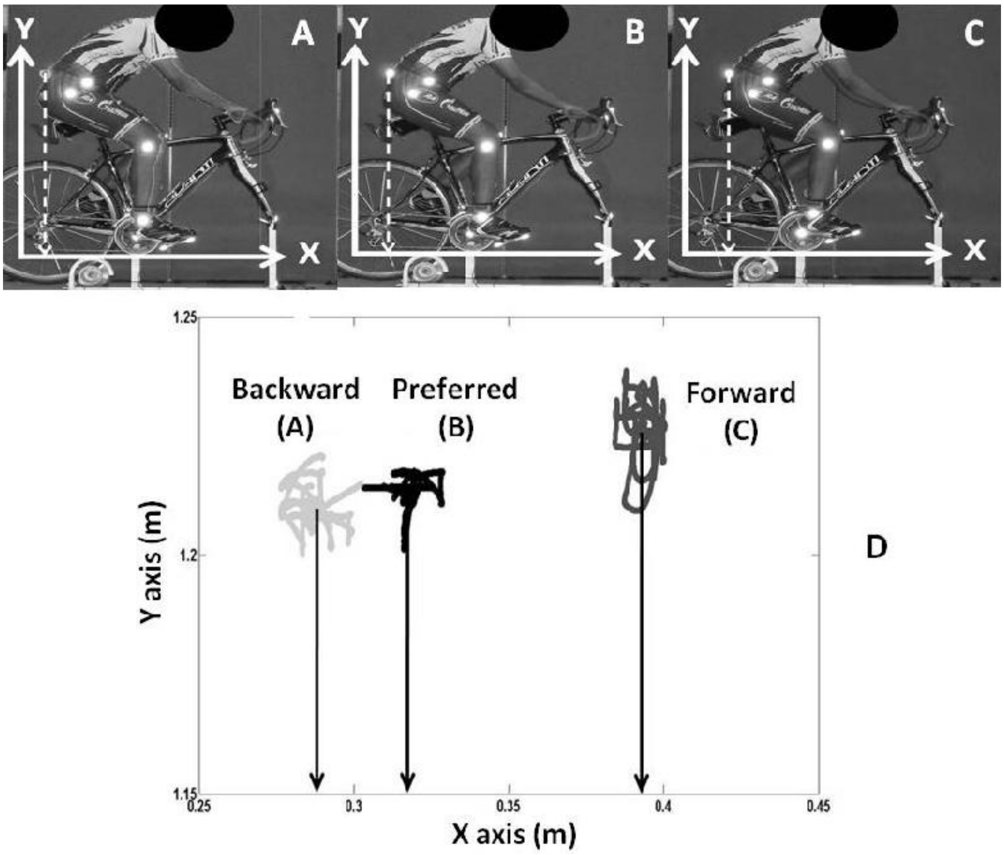


Figure 15.2. Ensemble displacement of the reflective markers at the preferred (A), the most forward (B), and the most backward (C) position on the saddle. The arrows illustrate the projection of the marker on the sacrum at the X axis. The displacement of the marker of the sacrum in the X and Y axis is presented (D) to highlight the differences between the three positions on the saddle across 10 consecutive crank revolutions.

Linear and angular velocities and accelerations were computed from smoothed kinematic data by a three points derivative method (Winter, 2005). Pedal angle in relation to the global coordinate system was calculated to convert the forces on the pedal reference system to forces in the global reference system by means of trigonometric procedures (Marsh, et al., 2000). The right lower limb was modelled as a three-segment rigid body system (thigh, shank and foot-pedal) with segment mass and centre of mass estimated according to De Leva (1996). Conventional inverse dynamics were conducted to calculate the net joint moments at the hip, knee and ankle (Redfield & Hull, 1986) using adapted scripts of van den Bogert and de Koning (1996). Patellofemoral compressive force was computed as described by Bressel (2001) including corrections for quadriceps-patellar tendon force ratio (Sharma, et al., 2008). Tibiofemoral compressive and shear forces were computed as described by Thambyah et al. (2005). Peak patellofemoral compressive force and tibiofemoral anterior shear and compressive forces were calculated. Knee flexion angles at 3 o'clock and 6 o'clock crank positions were determined. Knee forces and flexion angle were computed using a custom written program in Matlab (Mathworks Inc, MA) for ten consecutive crank revolutions to determine average and standard deviation for each cyclist.

Statistical analyses

Means and standard deviations were reported for the tibiofemoral anterior shear and compressive peak forces and patellofemoral compressive peak forces and for the knee angle at 3 o'clock and 6 o'clock crank positions. Individual cyclist's peak knee forces were normalized by their result at the maximal aerobic workload trial to reduce effects of workload and anthropometrical differences. We used the 1-min cycling trial at workload of maximal aerobic workload for normalisation to minimize workload effects for between-subjects analyses of knee forces given that Ericson and Nisell (1987) reported workload plays a major role in knee joint forces. Data normality distribution and sphericity were confirmed for all variables by the Shapiro-Wilk and Mauchly tests, respectively using SPSS for Windows 16.0 (SPSS, NY, USA).

The effects of changes in cyclists' position on the saddle on the patellofemoral compressive force, tibiofemoral anterior shear and compressive force, and knee angles at 3 o'clock and 6 o'clock crank positions were evaluated using Cohen's effect sizes (ES). Differences were rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Maximal workload levels used to normalize knee forces and workload of the second ventilatory threshold were 378 ±42W and 309 ±48W, respectively. Distance from the sacrum to the bicycle bottom bracket was 0.31 ±0.06 m at the maximal aerobic workload trial, 0.32 ±0.05 m at the preferred saddle position, 0.26 ±0.05 m at the forward saddle position and 0.35 ±0.05 m at the backward saddle position.

The differences in cyclists' position on the saddle between the preferred and forward or backward saddle positions were substantial as shown by the large effect sizes in Table 15.1. As a result of the forward saddle position there were large reductions in tibiofemoral anterior shear force. Large increases in knee flexion angle occurred when comparing forward to backward saddle positions (22% - 3 o'clock crank position and 36% - 6 o'clock crank position). Riding at the most forward or backward saddle positions did not substantially affect patellofemoral compressive and tibiofemoral compressive forces.

Table 15.1. Mean ±SD results of body position on the saddle, patellofemoral compressive force, tibiofemoral anterior shear and compressive force, knee flexion angle at 3 o'clock and 6 o'clock crank positions presented for three positions on the saddle (forward, preferred, backward) for cyclists. All variables (except knee flexion angle) presented as % of the results of the maximal power output trial. Differences between positions on the saddle are reported as mean difference percentages along with effect size magnitudes. Abbreviations used for forward (Fwd) and backward (back) positions on the saddle and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

		Forward	Preferred	Backward
Body position on the saddle (% of maximal workload trial)		84 ±8 ***Back 30%; 3.2, L***	102 ±6 ***Fwd 17%; 2.5, L*** ***Back 13%; 1.3, L***	114 ±11
Patellofemoral compressive force (% of maximal workload trial)		90 ±15 Back 4%; 0.4, S	86 ±12 Fwd 4%; 0.2, T Back 1%; 0.1, T	84 ±12
Tibiofemoral anterior force (% of maximal workload trial)		75 ±17 ***Back 45%; 2.3, L***	99 ±14 ***Fwd 26%; 1.6, L*** ***Back 19%; 1.1, L***	117 ±21
Tibiofemoral compressive force (% of maximal workload trial)		95 ±14 Back 3%; 0.1, T	96 ±13 Fwd 1%; 0.1, T Back 2%; 0.1, T	96 ±9
Knee flexion angle	**3 o'clock crank position (°)**	62 ±6 ***Back 22%; 1.7, L***	55 ±6 Fwd 9%; 0.9, M Back 10%; 0.8, M	50 ±6
	6 o'clock crank position (°)	38 ±6 ***Back 36%; 1.5, L***	33 ±5 Fwd 14%; 0.7, M Back 18%; 0.8, M	28 ±5

Discussion

Concern about the configuration of bicycle components has been raised based on reports of high rates of knee joint pain and overuse injuries in cyclists (Dettori & Norvell, 2006; Holmes, et al., 1994). Overuse knee pain for triathletes and other cyclists who tend to use a more forward configuration of the saddle compared to the conventional road bicycle set-up would be predicted. Recommendations for horizontal saddle position are based on the suggestion that moving forward on the saddle may increase patellofemoral compressive force (Callaghan, 2005). However, only trivial differences in patellofemoral compressive force between the preferred and forward saddle positions were observed in our study. The existing assumption of knee joint overload when riding at a most forward saddle position probably comes from the higher knee flexion angle, which was partially confirmed by our results. Large (9-12°) increases in the knee flexion angle were found for the forward saddle position compared to the backward saddle position, without large changes in knee flexion angle when the forward and the backward positions were compared to the preferred saddle position. However, changes in the knee flexion angle from movement on the saddle did not substantially affect patellofemoral or tibiofemoral compressive forces.

An explanation for the trivial effects of the forward and backward saddle positions on the patellofemoral compressive force is that the increase in knee flexion angle was smaller than predicted, which may have resulted in small changes in the patellar tendon angle and moment arm. However, an increase of 5-6°in knee flexion angle in our study resulted in a 9-10% greater patellar tendon angle between the preferred and forward positions. Yet, patellofemoral compressive force was changed by only 4% in our study. Compressive force at the patellofemoral joint depends on the magnitude of the force produced by the quadriceps muscle and the angle of the quadriceps tendon and the patellar tendon in relation to the longitudinal axis of the patella (Ericson & Nisell, 1987). Five times greater flexion angle for the knee joint at the peak pedal reaction force was related to 109% greater patellofemoral compressive force for similar pedal reaction force (Bressel, 2001). Therefore, an additional explanation may be related to differences in the magnitude of pedal reaction forces, which were not predicted in the present study based on the control of the workload level between trials.

Some studies have focussed on patellofemoral joint stress (compressive force per unit of area) because that should dictate the level of degeneration of the patellofemoral cartilage (Brechter & Powers, 2002). Based on the magnitude of increase for peak patellofemoral compressive force (~4%) and knee flexion angle (5-6°) when changing from the preferred to the forward saddle position, an increase of ~1% in the contact area and ~4% on the patellofemoral stress would be predicted using the model of Salsich et al. (2003). However, this change may not be substantial enough to affect patellofemoral cartilage matrix.

For the tibiofemoral joint, moving forward on the saddle was predicted to increase the compressive and reduce the anterior shear force components, which was partially observed in our study. In a previous study, greater flexion angle could not be linked to greater tibiofemoral forces during cycling (Tamborindeguy & Bini, 2011). The greater flexion angle found in our study elicited changes of ~1.3% on the moment arm of the patellar tendon. Therefore, we infer that based on the different magnitude of the anterior and compressive force on the tibiofemoral joint (13 times greater for the compressive component in our study), the compressive tibiofemoral force seems to have a smaller dependence on the knee angle resulting from changes in saddle position compared to the anterior shear force.

The link between greater knee flexion angle, excessive forward position of the saddle and knee joint load has been made without supportive evidence from experimental research (Callaghan, 2005). Our study results suggest that sitting in the forward or backward saddle positions may not elicit substantial changes in patellofemoral and tibiofemoral compressive force components. Greater anterior shear force on the tibiofemoral joint was observed for the backward position in our study. The possible reason was an increase in patellar tendon angle and patellar tendon moment arm due to smaller knee flexion angle. However the importance of this finding may be minimal because no reports of anterior cruciate ligament injuries have been observed for cycling (Dettori & Norvell, 2006). Low magnitudes of anterior shear force on the tibiofemoral joint (~0.2 times the body weight for present study) and tension on the anterior cruciate ligament during cycling (Fleming, et al., 1998) may explain the unlikely risk of anterior cruciate ligament ruptures in cycling.

Some limitations may have affected the results of our study. The use of a kinetic model for the knee joint including information from the literature related to anatomical characteristics of the knee (e.g. patellar tendon angle) and inverse dynamics techniques (e.g. assumption of purely rotational joints without translation movement between bones) must be noted. Knee joint modelling should be improved with inclusion of muscle activation and muscle mechanics as inputs (e.g. muscle force-length profile).

In summary, our study determined whether the forward saddle position adopted by race cyclists influenced the forces associated with the two main articulations of the knee joint. The only significant change with knee flexion angle was associated with the tibiofemoral shear force.

Conclusion

Riding at the forward and backward saddle positions did not substantially affect patellofemoral compressive and tibiofemoral compressive forces. Tibiofemoral anterior shear force was greater for the backward saddle position compared to the forward and preferred saddle positions. Small increases in knee flexion angle (5-6°) for a constant workload level may explain the trivial differences in patellofemoral and tibiofemoral compressive forces. Tibiofemoral anterior shear force may be more sensitive to changes in knee joint angle compared to other knee force components.

CHAPTER 16: EFFECTS OF CYCLING AT PREFERRED, FORWARD AND BACKWARD BODY POSITIONS ON THE SADDLE ON PEDALLING TECHNIQUE FOR CYCLISTS AND TRIATHLETES

Overview

Cyclists usually change their body position on the saddle depending on the characteristics of the race. We compared the effects of cycling at three body positions on the saddle (preferred/self-selected, most forward, most backward) on pedalling technique for cyclists and triathletes. Twelve cyclists and nine triathletes performed four trials starting with the maximal aerobic workload, followed by three trials at the workload of their ventilatory threshold. Force applied on the right pedal via an instrumented pedal, lower limb kinematics via video and muscle activation via electromyography were recorded during all trials. Pedalling technique was quantified using total force applied on the pedal, pedal force effectiveness, activation of six lower limb muscles, joint angles and mechanical work at the ankle, knee and hip joints. Analyses using effect sizes showed no large effects from changes in position on the saddle were observed for pedal forces, ankle joint work and ankle kinematics. There were large increases in knee joint angle and mechanical work and rectus femoris activation along with smaller hip work at the forward position on the saddle. Differences between cyclists and triathletes were not substantial. Effects of changes in saddle positions were limited to the hip and knee joints.

Introduction

The configuration of bicycle saddle position will dictate muscle activation (Ricard, et al., 2006) and joint kinematics (Price & Donne, 1997), which may affect oxygen uptake (Nordeen-Snyder, 1977) and performance in cycling (Hamley & Thomas, 1967).

Cyclists usually change their body position on the saddle depending on the characteristics of the race (e.g. uphill climbing or sprinting). For hill climbing, some cyclists tend to sit in the most backward position on the saddle, which differs from the most forward position on the saddle found during sprint or time trial races (Ricard, et al., 2006). Lower oxygen uptake was found when the saddle position was moved forward compared to the preferred position and to a more backward position of the saddle (Price & Donne, 1997). This result was related to reduced range of motion for the hip joint, which may have reduced the range of hip muscles' length for force production (Price & Donne, 1997). For triathlon, running after cycling using a more forward position on the saddle improved the adaptation to running and performance at the first 5 km of a 10-km time trial racing (Garside & Doran, 2000) because rectus femoris can operate in a longer length and possibly produce more force.

Moving forward on the saddle was found to reduce the activation of biceps femoris muscle during Wingate sprint tests (Ricard, et al., 2006). This result may be related to a backward shift in the pedal reaction force vector in relation to the knee joint, which will possibly change the knee joint moment from extensor to flexor. Therefore, moving forward on the saddle during a race can be associated with two acute adaptations. The first is based on a larger knee flexion angle, which will increase the length of knee extensor muscles for force production (Rugg & Gregor, 1987). The second is based on potentially smaller moment arms of the hip joint extensor muscles. Therefore forward and backward changes on the position on the saddle practiced by cyclists and triathletes during racing will possibly result in opposite effects in joint coordination, muscle recruitment and pedalling technique. Even though the rationale for moving forward on the saddle may be related to the lower contribution of hamstrings muscle group (Ricard, et al., 2006) due to greater knee flexion angle and better running performance after cycling (Garside & Doran, 2000), it is not clear how pedalling technique changes in cyclists and triathletes. Pedal force effectiveness is part of the technique profile of cyclists (Bini & Diefenthaeler, 2010), however,, pedalling technique changes while cycling at the most forward and backward position on the saddle are unclear. Leirdal & Ettema (in press) observed no changes in pedal force effectiveness when cyclists were assessed moving 4 cm forward their saddle.

Cyclists and triathletes can differ in pedal force effectiveness (Candotti, et al., 2007) and muscle activation (Candotti, et al., 2009; Chapman, et al., 2007). Therefore, pedalling technique adaptation to positions on the saddle may not be similar for cyclists and triathletes. However, there are no published studies that assessed pedalling technique and muscle activation of cyclists and triathletes while cycling at various positions on the saddle.

The aim of our study was to compare the effects of cycling at three positions on the saddle (preferred/self-selected, most forward, most backward) on pedalling technique via pedal force effectiveness, joint mechanical work, kinematics, and muscle activation for cyclists and triathletes.

Methods

Participants

Twelve cyclists and nine triathletes with competitive experience participated in the study (see Table 16.1 for participant characteristics).

Table 16.1. Characteristics (mean ±SD) of age, body mass, height, maximal aerobic workload and maximal oxygen uptake of 12 cyclists and 9 triathletes. Percentage differences and effect sizes are for comparisons of cyclists and triathletes. Abbreviations used are for effect sizes of trivial (T), small (S), moderate (M) and large (L).

Groups	**Cyclists (n=12)**	**Triathletes (n=9)**	**Cyclists vs. Triathletes**
Age (years)	28 ±6	28 ±8	<1%; 0.1, T
Body mass (kg)	71 ±7	75 ±10	6%; 0.5, M
Height (cm)	177 ±10	180 ±8	2%; 0.4, S
Maximal aerobic workload (W)	377 ±30	386 ±45	2%; 0.2, T
Maximal oxygen uptake (ml/kg/min)	64 ±5	59 ±5	9%; 1.0, L
Workload of the ventilatory threshold (W)	325 ±35	320 ±38	1%; 0.1, T
Oxygen uptake at ventilatory threshold (ml/kg/min)	53 ±6	49 ±4	9%; 0.8, M

Before the study, participants were informed about possible risks and signed a consent form approved by the ethics committee of human research where the study was conducted.

Data collection

During the first session height and body mass were measured according to International Society for the Advancement of Kinanthropometry protocols (Marfell-Jones, et al., 2006). Cyclists and triathletes warmed up at 150 W for 10 minutes followed by an incremental maximal step test to exhaustion with initial workload of 100 W and increments of 25 W/min (Lucía, et al., 2002) using their own bicycles on a Computrainer cycle simulator (Racemate, Inc, Seattle, USA). Workload was recorded throughout the incremental test along with gas exchanges by the breath-by-breath method using an open-circuit gas exchange system (MGC CPX/D, Medical Graphics Corp., St Louis, MO, USA). Before the incremental test, the oxygen and carbon dioxide analysers were calibrated using medical grade gases that spanned air in the physiological range. Gas exchange data were analyzed to define the workload related to the ventilatory threshold based on the ventilatory equivalent method (Weston & Gabbett, 2001). Pedalling cadence was controlled close to 90 ±2 rpm using feedback from the cycle trainer head unit. The test was stopped by voluntary exhaustion or when cyclists and triathletes were unable to maintain pedalling cadence.

After 48 hours the cyclists and triathletes warmed up for 10 min at 150 W and then rode for one minute with 90 rpm pedalling cadence at the highest workload recorded during the incremental test in their preferred saddle position. Additional 1-minute tests were completed at a workload set to the second ventilatory threshold and in the three saddle positions: preferred, most forward position, most backward. For the forward and backward saddle positions cyclists and triathletes were instructed to simulate a position similar to time-trial and hill climb cycling, respectively, without substantial changes in the upper body lean position. The order of the forward and the backward saddle positions was allocated randomly.

Force applied on the right pedal and right lower limb kinematics were recorded for the last 20 seconds during all saddle position conditions. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of cyclists and triathletes at the anterior superior iliac spine, greater trochanter, lateral femoral condyle, lateral malleolus, anterior and posterior pedal stick. One marker was attached to the sacrum to measure the position of cyclists and triathletes in relation to the bicycle frame at the saddle positions. Two markers were taped to the bicycle frame and used as reference for image calibration. A 2D pedal dynamometer (Nabinger, et al., 2002) and one high speed camera positioned perpendicular to the motion plane (AVT PIKE F-032, Allied Vision Technologies GmbH, Germany) were synchronized by an external trigger. Kinematics were recorded at 60 Hz using AVT ActiveCam viewer software (Allied Vision Technologies GmbH, Germany) and force data were recorded at 600 Hz per channel employing a 16-bit analogical to digital converter (DI220, Dataq Instruments, USA) using WINDAQ® software (WINDAQ, DataQ Instruments Inc., USA).

Muscle activation was measured by surface electromyography for the tibialis anterior, the medial head of gastrocnemius, soleus, the long head of biceps femoris, rectus femoris, and the vastus medialis muscles using a Bortec electromyography system (AMT-8, Bortec Electronics Inc., Calgary, Canada). Pairs of Ag/AgCl electrodes (bipolar configuration, 22 mm diameter) were positioned on the skin after carefully shaving and cleaning the area using an abrasive cleaner and alcohol swabs to reduce the skin impedance as recommended by the International Society of Electrophysiology and Kinesiology (De Luca, 1997; Merletti, et al., 2009). Electrodes were placed over the belly of the muscles, one third of the muscle length from the midpoint (to avoid the musculotendinous junction), parallel with the muscle fibres and taped to the skin using micropore tape (3M Company, USA). A bandage was wrapped around the electrode to minimize sweat interference. The reference electrode was placed over the anterior surface of the tibia. The electrodes' wires were then taped to the skin to reduce movement artifact.

Data analyses

Video files were digitized and automatic tracking of markers were conducted in DgeeMe software (Video4Coach, Denmark) for x-y coordinates over time. Kinematics and force data were smoothed with a digital second order zero lag low-pass Butterworth filter with cut-off frequency optimized to reduce signal residuals (Winter, 2005). Joint angles of the hip, knee, and ankle during pedalling movement were calculated from the smoothed x-y coordinate data, as per the spatial model shown in Figure 1. From each pedal revolution, the mean value and range of motion of the hip, knee and ankle joints were calculated over time (Bini, Diefenthaeler, et al., 2010).

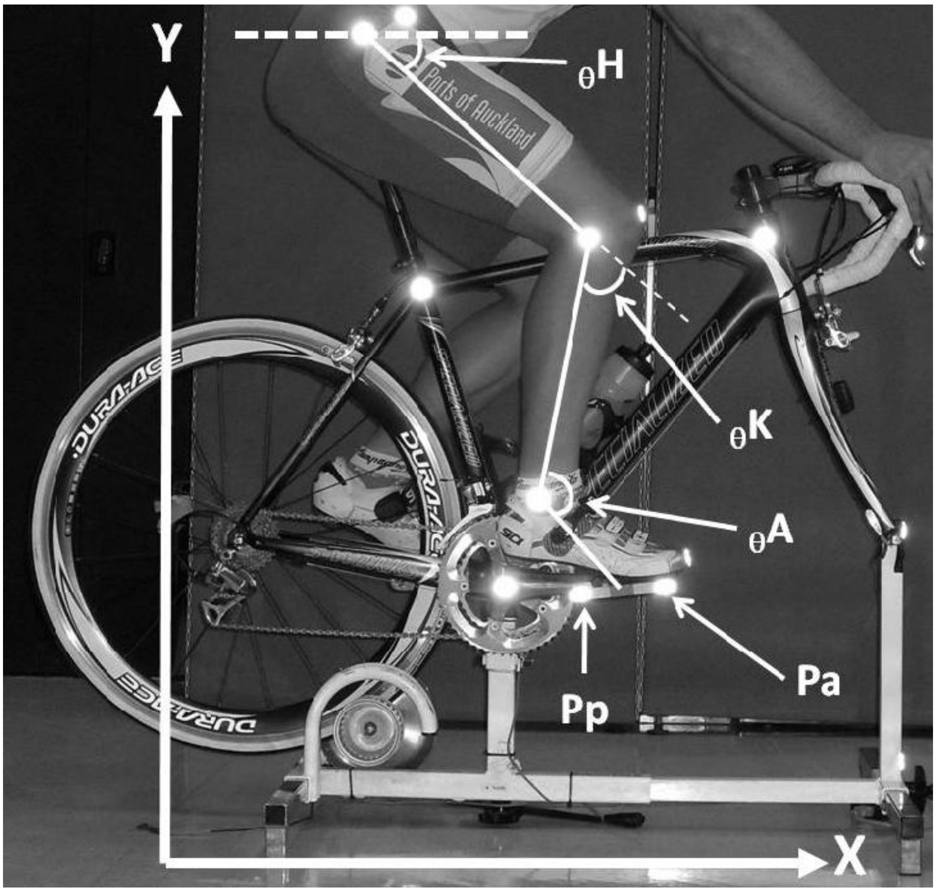


Figure 16.1. Illustration of reflective marker placement on the right side of a cyclist at the anterior superior iliac spine, sacrum, greater trochanter, lateral femoral condyle and lateral malleolus to measure hip ($_{\theta}$H), knee ($_{\theta}$K) and ankle ($_{\theta}$A) joint angles. Reflective markers were attached to the anterior (Pa) and posterior (Pp) aspects of the reference stick attached to the pedal axis for computation of pedal force components into the global coordinate system.

Determination of the hip joint centre based on the average coordinate between the marker on the anterior superior iliac spine and the greater trochanter was conducted (Neptune & Hull, 1995). The average relative horizontal position of the marker on the sacrum to the bottom dead centre was computed over time across ten pedal revolutions for the analysis of body position on the saddle at the three positions on the saddle (preferred, most forward and most backward). In Figure 16.2 the representative displacement of the markers is shown for one cyclist for the three positions on the saddle.

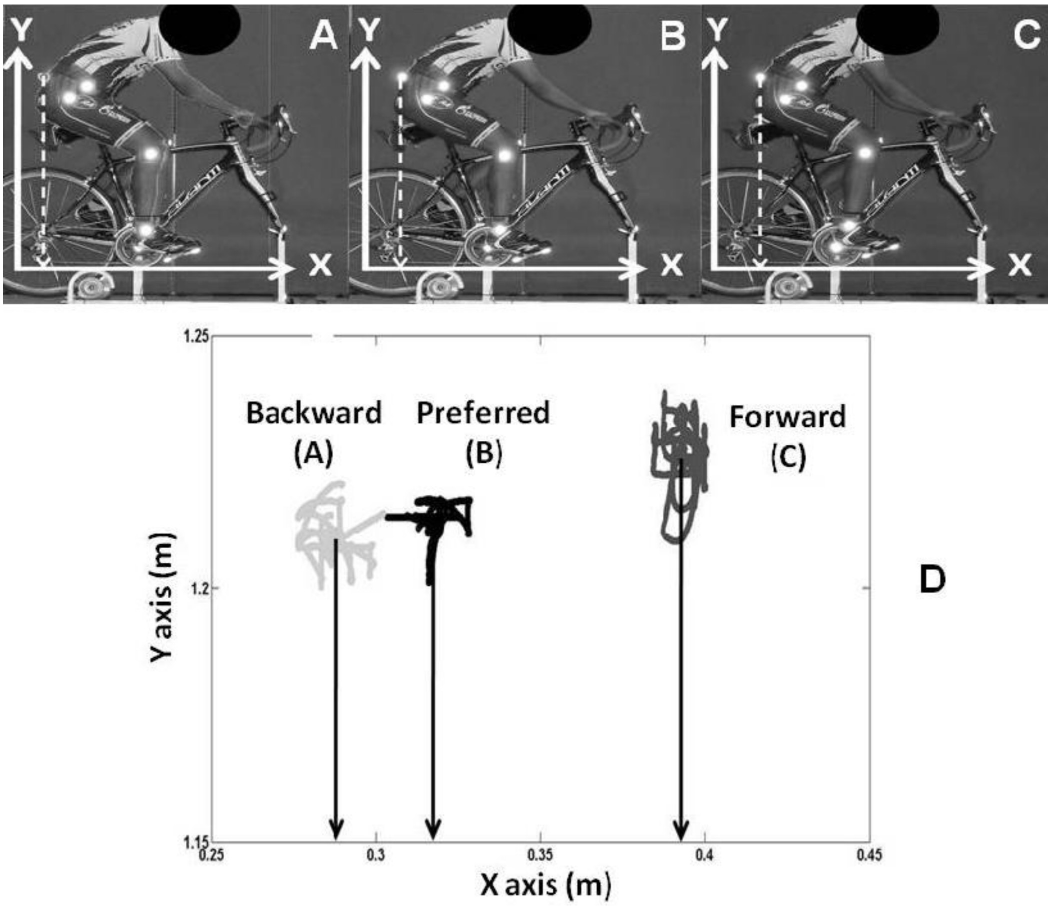


Figure 16.2. Ensemble displacement of the reflective markers at the preferred (A), the most forward (B), and the most backward (C) positions on the saddle. The arrows illustrate the projection of the marker on the sacrum at the X axis. The displacement of the marker of the sacrum in the X and Y axis is presented (D) to highlight the differences between the three body positions on the saddle across 10 consecutive crank revolutions.

Linear and angular velocities and accelerations were computed from smoothed kinematic data by a three points derivative method (Winter, 2005). Pedal angle in relation to the global coordinate system was calculated to convert the forces on the pedal reference system to forces in the global reference system by means of trigonometric procedures (Marsh, et al., 2000). The right lower limb was modelled as a three-segment rigid body system (thigh, shank and foot-pedal) with segment mass and centre of mass estimated according to De Leva (1996). Conventional inverse dynamics were conducted to calculate the net joint moments at the hip, knee and ankle (Redfield & Hull, 1986) using adapted scripts of van den Bogert and de Koning (1996). Net joint mechanical work was calculated by integrating joint power (moment x angular velocity at the joints) with respect to time, and relative contributions of the ankle, knee, and hip joints were calculated by their percentage contribution to total mechanical work at three joints (Bini, Rossato, et al., 2010). Pedal force effectiveness and pedal force application were computed by the overall index of effectiveness (ratio between the tangential force on the crank and the total force on the pedal surface) and by the average total force applied on the pedal, respectively (Rossato, et al., 2008).

Raw electromyographic signals were filtered using a band-pass Butterworth filter with cut-off frequencies optimized to reduce signal residuals (Winter, 2005). The root mean square envelope averaged data in 40 ms moving windows across the entire raw signal (Neptune, Kautz, & Hull, 1997) and then cut data into ten consecutive crank revolutions to determine the average and standard deviation of each muscle for each cyclist and triathlete. The root mean square envelopes were normalized by the mean root mean square value from the condition where the participants rode at the maximal workload trial in a preferred saddle position. All variables were processed using a custom written program in MATLAB® (MathWorks Inc, USA) for ten consecutive crank revolutions to determine average and standard deviation for each participant.

Statistical analyses

Means and standard deviations were reported for cyclists and triathletes' body position on the saddle, total force applied on the pedal, index of effectiveness, mean value and range of motion of the hip, knee and ankle joints, individual contribution of the hip, knee and ankle joints to the total mechanical work, and mean value of the root mean square envelope of the tibialis anterior, the medial head of gastrocnemius, soleus, the long head of biceps femoris, rectus femoris, and the vastus medialis muscles. Data normality distribution and sphericity was evaluated by the Shapiro-Wilk and Mauchly tests, respectively. For hip joint mechanical work and range of motion, ankle mean angle and mechanical work, activation of rectus femoris, tibialis anterior and gastrocnemius medialis, a logarithm transform was applied. Results from the preferred, forward and backward saddle positions gathered at the maximal workload trial were used to normalize pedal force, muscle activation and joint mechanical work variables for each cyclist and triathlete.

To compare the effects of saddle position on the dependent variables (position on the saddle and groups), Cohen's effect sizes (ES) were computed for the analysis of magnitude of the differences, and rated as trivial (<0.25), small (0.25-0.49), moderate (0.5-1.0), and large (>1.0) (Rhea, 2004). We chose large effect sizes for discussion of results to ascertain non-overlap between mean scores greater than 55% (Cohen, 1988).

Results

Changes in position on the saddle were 5 ±1 cm for cyclists and 6 ±1 cm for triathletes for the forward saddle position and 5 ±1 cm for cyclists and 3 ±1 cm for triathletes for the backward saddle position. Large changes in body position on the saddle were observed for cyclists and triathletes without large differences between groups. No large differences in total pedal force or index of effectiveness were observed for changes in position on the saddle or between cyclists and triathletes (see Table 16.2).

Table 16.2. Means and standard deviations of body position on the saddle, total force applied on the pedal and index of effectiveness presented for the three saddle positions (preferred, forward and backward) for cyclists and triathletes. Comparisons of position on the saddle and groups (cyclist versus triathletes in italics) are shown in percentage differences and effects sizes. Abbreviations used are for forward (Fwd) and backward (Back) positions on the saddle, triathletes (Tri) and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

	Cyclists			Triathletes		
	Forward	**Preferred**	**Backward**	**Forward**	**Preferred**	**Backward**
Body position on the saddle (% of maximal workload)	83 ±9 ***Back 26%; 3.7, L*** *Tri 2 %; 0.2, T*	100 ±4 ***Fwd 17%; 2.8, L*** ***Back 13%; 2.8, L*** *Tri 4 %; 0.8, M*	113 ±7 *Tri 1 %; 0.1, T*	85 ±6 ***Back 29%; 2.8, L***	104 ±9 ***Fwd 19%; 2.8, L*** Back 10%; 0.8, M	114 ±15
Total pedal force (% of maximal workload)	93 ±12 Back 2%; 0.2, T *Tri <1%; 0.1, T*	92 ±8 Fwd 1%; 0.1, T Back 1%; 0.1, T *Tri 1%; 0.1, T*	91 ±10 *Tri 1%; 0.1, T*	93 ±8 Back 1%; 0.1, T	93 ±9 Fwd <1%; 0.1, T Back 1%; 0.1, T	92 ±9
Index of effectiveness (% of maximal workload)	101 ±12 Back <1%; 0.1, T *Tri 6%; 0.5, M*	95 ±10 Fwd 4%; 0.3, S Back 1% ; 0.4, S *Tri 1%; 0.1 T*	101 ±9 *Tri 6%; 0.8, M*	95 ±10 Back <1%; 0.1, T	96 ±8 Fwd 1%; 0.1, T Back 1%; 0.1 T	95 ±6

Large increases in knee mean angle were observed for cyclists and triathletes moving forward on the saddle, without substantial differences between groups. Reduced knee range of motion and increased knee work were found for triathletes at the forward position on the saddle compared to the backward and preferred saddle positions. Cyclists' knee work was increased at the forward position compared to the backward position on the saddle. Hip mean angle was smaller for cyclists than for triathletes at the backward position on the saddle. Hip work of cyclists was smaller at the forward compared to the preferred and backward positions on the saddle (see Table 16.3).

Table 16.3. Means and standard deviations of mean angle, range of motion and mechanical work of the hip, knee and ankle joints presented for the three positions on the saddle (preferred, forward and backward) for cyclists and triathletes. Comparisons of position on the saddle and groups (cyclist versus triathletes in italics) are shown in percentage differences and effects sizes. Abbreviations used are for forward (Fwd) and backward (Back) position on the saddle, triathletes (Tri) and effect sizes of trivial (T), small (S), moderate (M) and large (L).

		Cyclists			Triathletes		
		Forward	Preferred	Backward	Forward	Preferred	Backward
Ankle	mean angle (°)	116 ±8 Back 1%; 0.2, T *Tri 4%; 0.7, M*	115 ±8 Fwd 1%; 0.1, T Back 2%; 0.3, S *Tri 4%; 0.7, M*	117 ±8 *Tri 3%; 0.7, M*	120 ±4 Back 1%; 0.4, S	119 ±3 Fwd 1%; 0.1, T Back 2%; 0.6, M	121 ±3
	Range of motion (°)	30 ±10 Back 11%; 0.4, S *Tri 31%; 0.8, M*	29 ±8 Fwd 2%; 0.1, T Back 13%; 0.5, M *Tri 10%; 0.3, S*	34 ±9 *Tri 17%; 0.6, M*	23 ±7 Back 21%; 0.8, M	27 ±8 Fwd 7%; 0.5, M Back 7%; 0.3, S	29 ±7
	Work (% of maximal workload)	116 ±37 Back 13%; 0.4, S *Tri 13%; 0.4, S*	104 ±17 Fwd 2%; 0.4, S Back 1%; 0.1, T *Tri 6%; 0.3, S*	103 ±20 *Tri 4%; 0.2, T*	103 ±20 Back 4%; 0.2, T	110 ±19 Fwd 7%; 0.4, S Back 3%; 0.2, T	107 ±20
Knee	Mean angle (°)	72 ±3 ***Back 13%; 3.0 ,L*** *Tri 4%; 0.8, M*	68 ±3 ***Fwd 6%; 1.4, L*** ***Back 7%; 1.4, L*** *Tri 2%; 0.6, M*	64 ±3 *Tri 3%; 0.6, M*	70 ±3 ***Back 12%; 2.6, L***	66 ±2 ***Fwd 5%; 1.2, L*** ***Back 7%; 2.6, L***	62 ±3

	Range of motion (°)	72 ±6 Back 7%; 0.9, M *Tri 5%;* *0.7, M*	75 ±6 Fwd 3%; 0.4, S Back 4%; 0.5, M *Tri 3%;* *0.4, S*	78 ±5 *Tri 4%;* *0.6, M*	69 ±5 ***Back 8%;*** ***1.5, L***	73 ±4 Fwd 6%; 0.8, M Back 3%; 0.6, M	75 ±4
	Work (% of workload)	102 ±6 ***Back 6%;*** ***1.0, L*** *Tri 1%;* *0.3, S*	97 ±7 Fwd 5%; 0.8, M Back 1%; 0.2, T *Tri 2%;* *0.4, S*	96 ±6 *Tri 1%;* *0.2, T*	101 ±7 Back 6%; 0.8, M	95 ±3 ***Fwd 6%;*** ***1.1, L*** Back 1%; 0.1, T	95 ±8
Hip	Mean angle (°)	64 ±8 Back 2%; 0.4, S *Tri 5%;* *0.8, M*	63 ±5 Fwd 2%; 0.2, T Back 1%; 0.1, T *Tri 5%;* *0.8, M*	63 ±5 ***Tri 6%;*** ***1.1, L***	67 ±5 Back 2%; 0.4, S	66 ±4 Fwd 1%; 0.3, S Back 1%; 0.1, T	66 ±3
	Range of motion (°)	44 ±9 Back 3%; 0.2, T *Tri 13%;* *0.8, M*	45 ±7 Fwd 2%; 0.1, T Back 1%; 0.1, T *Tri 10%;* *0.6, M*	46 ±7 *Tri 10%;* *0.7, M*	39 ±3 Back 6%; 0.6, M	41 ±6 Fwd 4%; 0.4, S Back 1%; 0.1, T	41 ±5
	Work (% of workload)	87 ±20 ***Back 28%;*** ***1.4, L*** *Tri 18%;* *0.6, M*	109 ±19 ***Fwd 6%;*** ***1.1, L*** Back 6%; 0.3, S *Tri 29%;* *0.7, M*	115 ±20 *Tri 38%;* *0.7, M*	105 ±40 Back 48%; 0.7, M	138 ±71 Fwd 2%; 0.6, M Back 11%; 0.2, T	153 ±95

Increased rectus femoris activation was observed for cyclists at the preferred compared to the backward saddle positions. Gastrocnemius medialis activation was increased for triathletes changing from forward to backward positions on the saddle. Soleus activation was higher for cyclists than triathletes at the preferred and backward saddle positions (see Table 16.4).

Table 16.4. Means and standard deviations for activity of biceps femoris, rectus femoris, vastus medialis, tibialis anterior, gastrocnemius medialis and soleus presented for the three positions on the saddle (preferred, forward and backward) for cyclists and triathletes. Comparisons of position on the saddle and groups (cyclist versus triathletes in italics) are shown in percentage differences and effects sizes.. Abbreviations used are for forward (fwd) and backward (back) position on the saddle, triathletes (Tri) and effect sizes of trivial (T), small (S), moderate (M) and large (L). Large differences were highlighted in bold italics.

	Cyclists			**Triathletes**		
	Forward	**Preferred**	**Backward**	**Forward**	**Preferred**	**Backward**
Biceps femoris (% of maximal workload)	75 ±30 Back 9%; 0.4, S *Tri 2%;* *0.1, T*	88 ±20 Fwd 13%; 0.5, M Back 4%; 0.2, T *Tri 1%;* *0.1, T*	84 ±21 *Tri 4%;* *0.2, T*	77 ±22 Back 11%; 0.6, M	87 ±9 Fwd 10%; 0.7, M Back 1%; 0.1, T	88 ±19
Rectus femoris (% of maximal workload)	75 ±31 Back 10%; 0.4, S *Tri 5%;* *0.2, T*	84 ±17 Fwd 9%; 0.4, S ***Back 19%;*** ***1.1, L*** *Tri 3%;* *0.2, T*	65 ±18 *Tri 21%;* *0.7, M*	80 ±20 Back 6%; 0.2, T	81 ±18 Fwd 1%; 0.1, T Back 5%; 0.2, T	86 ±44
Vastus medialis (% of maximal workload)	81 ±12 Back 4%; 0.3, S *Tri 4%;* *0.2, T*	80 ±14 Fwd 1%; 0.1, T Back 5%; 0.3, S *Tri 3%;* *0.2, T*	85 ±14 *Tri 7%;* *0.4, S*	85 ±22 Back 8%; 0.3, S	83 ±11 Fwd 2%; 0.1, T Back 9%; 0.6, M	92 ±22
Tibialis anterior (% of maximal workload)	100 ±33 Back 5%; 0.2, T *Tri 13%;*	100 ±35 Fwd 1%; 0.1, T Back 5%; 0.2, T *Tri 6%;*	95 ±25 *Tri 2%;*	87 ±9 Back 6%; 0.3, S	94 ±18 Fwd 7%; 0.4, S Back 1%; 0.1, T	93 ±29

	0.6, M	*0.2, T*	*0.1, T*			
Gastrocne- mius medialis (% of maximal workload)	92 ±15 Back 7%; 0.4, S *Tri 2%;* *0.2, T*	94 ±15 Fwd 2%; 0.1, T Back 5%; 0.3, S *Tri 3%;* *0.2, T*	99 ±13 *Tri 12%;* *0.6, M*	90 ±12 ***Back 21%;*** ***1.1, L***	97 ±13 Fwd 7%; 0.6, M Back 14%; 0.7, M	111 ±29
Soleus (% of maximal aerobic workload)	96 ±21 Back 3%; 0.1, T *Tri 16%;* *0.9, M*	98 ±20 Fwd 2%; 0.1, T Back 5%; 0.2, S ***Tri 19%;*** ***1.2, L***	93 ±24 ***Tri 20%;*** ***1.2, L***	80 ±11 Back 7%; 0.7, M	79 ±11 Fwd 1%; 0.1, T Back 6%; 0.6, M	73 ±8

Discussion

Our study compared pedal forces, joint mechanical work, kinematics and muscle activation of lower limb muscles of cyclists and triathletes riding at three positions on the saddle. From a self-selected preferred position, cyclists and triathletes were asked to choose the most forward and most backward possible positions on the saddle. The main findings were that pedal forces were not substantially affected by changes in body position on the saddle, however, knee and hip angles and mechanical work showed large changes for cyclists and triathletes. Activation of rectus femoris decreased for cyclists and gastrocnemius medialis increased for triathletes at the backward saddle position. Soleus activation was increased for cyclists compared to triathletes, without large differences for other variables.

Cyclists and triathletes change their horizontal position on the saddle depending on geographic characteristics of the race, which has been expected to affect pedalling technique and muscle activation (Clarys, Alewaeters, & Zinzen, 2001). During uphill cycling, cyclists tend to move backward on the saddle to enhance the hip joint moment-arm in relation the pedal axis (Burke & Pruitt, 2003). There is a more forward position on the saddle during time trial cycling (Ricard, et al., 2006). Triathletes choose a forward body position on the bicycle to increase hip extension and improve running after the bike section of the triathlon (Garside & Doran, 2000). Reduced hip range of motion has been reported for more forward body positions (greater seat tube angle) of cyclists at 200 W and 85 rpm (Price & Donne, 1997). During Wingate tests, cyclists have had reduced biceps femoris activation using a more forward position on the saddle (greater seat tube angle) (Ricard, et al., 2006). These results are different from ours because we did not observe large changes in hip joint angle or biceps femoris activation when body position on the saddle was changed. One explanation is the potentially greater changes in body position due to larger seat tube angles conducted in previous studies (6-18°) compared to our study (~3°). Saddle height was corrected when seat tube angle was changed in previous studies, which does not replicate racing and training when cyclists are not able to reduce saddle height when moving backward on the saddle during uphill riding. Therefore, our study provided a more ecological assessment of changes in body position during cycling. Cyclists from our study presented reduced rectus femoris activation at the backward position on the saddle compared to the preferred position which can be associated with greater hip mean angle for triathletes (less hip flexion) compared to cyclists at the backward saddle position. Cyclists may have presented a smaller hip flexion at the backward position than triathletes due to shorter rectus femoris muscle length (Herzog, et al., 1991), if triathletes' rectus femoris adapts to produce force at longer muscle lengths due to parallel running training.

Changes in body position on the saddle and hip joint kinetics and kinematics may have affected the knee joint. Large changes in mean knee angle were observed for cyclists and triathletes. In general, the forward position on the saddle resulted in greater knee flexion angle, which is contrary to previous findings (Diefenthaeler, et al., 2006; Price & Donne, 1997; Rottenbacher, et al., 2009) where no substantial differences were reported for knee joint kinematics. In our study, greater mechanical work of the knee occurred for the forward position compared to the preferred and backward positions, which may be related to the greater rectus femoris activation at the preferred compared to the backward position in cyclists and longer knee extensors muscles length (greater knee flexion angle). These results suggest that cyclists may vary joint kinematics in different ways to sustain power output when riding at various positions on the saddle.

The ankle joint plays a role in transferring mechanical work produced at hip and knee joints to the pedal (Kautz & Neptune, 2002). When changing position on the saddle, cyclists and triathletes in our study presented stability in ankle kinetics and kinematics, without large changes for mean angle, range of motion or mechanical work of the ankle. Conversely, Rottenbacher et al. (2009) found greater ankle angle (increased plantar flexion) when moving the saddle backward from the preferred position. Activation of gastrocnemius medialis was reduced in triathletes when the most forward position on the saddle was compared to the most backward in our study. The stability in ankle kinematics at positions on the saddle was translated into similar total pedal force application and index of effectiveness, which indicates similar ability of cyclists and triathletes to apply force on the pedal and to convert this force into torque on the crank. Likewise, only trivial differences were found for the total force applied on the pedal and for the index of effectiveness in previous studies (Leirdal & Ettema, in press; Rottenbacher, et al., 2009). However, Korff et al. (in press) showed that using a combination of lower and more forward positions of the saddle resulted in greater ankle power, without changes in pedal force effectiveness or hip and knee joint power. Merging changes in saddle height and horizontal position may mask particular effects of each component of saddle position on lower limb kinetics and kinematics.

Cyclists and triathletes with similar performance in the incremental cycling test to exhaustion participated in our study (maximal aerobic workload and workload of the ventilatory threshold), which would minimize differences between groups. Differences between groups were only observed for mean hip angle at the most backward position and soleus activation at the preferred and backward positions. Cyclists presented smaller hip flexion angle and greater soleus activation than triathletes. These results do not fully support previous findings of differences between cyclists and triathletes in knee (Candotti, et al., 2009) and ankle muscles activation (Chapman, et al., 2007). Chapman et al. (2007) indicated that cyclists presented lower variability in calf muscles activation compared to triathletes, without comparison of the magnitude of the muscle activation for similar workloads of exercise. Cyclists from the study of Candotti et al. (2007) presented similar indices of effectiveness compared to triathletes at 90 rpm of pedaling cadence, which is in line with our findings.

Tracking body position on the saddle by video analysis may be less reliable then using pressure sensors on the saddle surface, which could have limited our findings. However, the changes in saddle position were substantial enough to be detected by video analysis.

Conclusion

Riding at the most forward position on the saddle did not result in large effects for pedal force application and index of effectiveness. Increased knee mean angle and mechanical work of the knee, rectus femoris activation and smaller hip work occurred when riding at the most forward position on the saddle, without large changes in ankle joint angle. Differences between cyclists and triathletes were not substantial. Effects of changes in saddle positions were limited to the hip and knee joints.

CHAPTER 17: DISCUSSION / CONCLUSIONS

Body position on the bicycle has been suggested to play a role in maximizing performance and reducing the risk of overuse injuries (Peveler, 2008; Peveler, et al., 2005). A more complete transfer of force from the body to the bicycle may be achieved when body position on the bicycle is optimized. However, effectiveness of bicycle configuration in preventing overuse injuries is unknown as methods to assess body position on the bicycle (usually via static assessment) lack accuracy and are limited to road cyclists. Recommendations for bicycle configuration are mostly based on empirical knowledge (Peveler, 2008) partially because the methods to assess cyclists body position on the bicycle and cycling kinetics need to be assessed for validity and reliability. Methods to evaluate cycling kinetics, to enable changes aimed at improving performance, have not been experimentally assessed. Therefore, precision of measures taken using some devices commercially available may compromise decisions by coaches and clinicians (Bini, Hume, & Cervieri, 2011).

The bicycle saddle has been suggested as the most important variable in bicycle configuration because it will affect lower limb kinematics (Wanich, et al., 2007). However, effects of changing saddle position were uncertain for applications of force on the pedals and forces on the lower limb joints. Knee forces and pedal force effectiveness can be affected when saddle height or horizontal position of the body on the saddle are modified during cycling (Ericson & Nisell, 1987).

Given limitations in the literature, the overall question of this thesis was “What are the effects of saddle position on pedalling technique and what methods should be used to assess pedalling kinetics and kinematics of cyclists and triathletes?” Specific questions were “What is the validity and reliability of methods for determining body position (kinematics) and pedalling kinetics in cycling?” and “What are the effects from changes in saddle height and horizontal position on pedalling kinetics and kinematics of cyclists and triathletes?”

The series of studies conducted to address these questions provided novel perspectives for the evaluation of body position on the bicycle via assessment of joint kinetics and kinematics and force applied to the pedals. Effects of changes in saddle height were highlighted for recreational cyclists with and without knee pain and for cyclists compared with triathletes. The following discussion summarises the results and inferences of these studies for the main paradigms of interest: body position on the bicycle, methods for cycling biomechanics assessment and effects of changing saddle position on pedalling kinetics and kinematics.

Body position on the bicycle

The evidence for the effectiveness of bicycle configuration and cyclists’ body position to prevent overuse injuries in cycling using current biomechanical approaches (Chapter 2) and the comparison of body position on the bicycle for cyclists and triathletes (Chapter 3) were completed. Optimizing body position on the bicycle theoretically has the potential to prevent overuse injury in cycling, however, biomechanical studies to date have not shown clear evidence that they can reduce overuse injury via optimization of bicycle configuration. In the assessment of cyclists and triathletes body position on the bicycle, competitive triathletes had greater body forward projection (greater trunk flexion and knee anterior position) than competitive road cyclists. Both recreational and competitive cyclists sat on their bicycles with their trunks in a more vertical position compared to triathletes. Guidelines for bicycle configuration for triathletes and road cyclists need to consider the body positions used during events. If attempting to optimize bicycle configuration and cyclists' body position to prevent overuse injury then it is suggested that dynamic assessment of joint kinetics and/or kinematics is undertaken. The information from the literature review and the descriptions of cyclists and triathletes body positions on the bicycles provided rationale for the studies in the thesis.

Methods for cycling biomechanics assessment

When lower limb joint angles of cyclists in static poses or during cycling motion were compared, cyclists were not able to replicate in a static pose at the 6 o'clock crank position similar hip, knee and ankle joint angles as measured in dynamic cycling. To perform configuration of bicycle components using joint angles, measurements should be taken dynamically or with the cyclists in static poses at the 3 o'clock crank position, instead of the usually recommended 6 o'clock crank position.

Changes in cycling kinetics are not limited to changes in body position on the bicycle. Therefore, we conducted a literature review (Chapter 5) to assess how pedal force effectiveness would be affected by constraints (e.g. workload, pedalling cadence, etc) and what is the response of this variable to technique training. Most studies measuring pedal forces have been restricted to one leg but a few studies have reported bilateral asymmetry in pedal forces. Pedal force effectiveness is increased at higher workloads and reduced at higher pedaling cadences. Changes in saddle position resulted in unclear effects in pedal force effectiveness, while lowering the upper body reduced pedal force effectiveness. Cycling experience and fatigue had unclear effects on pedal force effectiveness. Augmented feedback of pedal forces can improve pedal force effectiveness within a training session and after multiple sessions for cyclists and non-cyclists. No differences in pedal force effectiveness were evident between summarized and instantaneous feedback. Conversely, economy/efficiency seems to be reduced when cyclists are instructed to improve pedal force effectiveness during acute intervention studies involving one session. Decoupled crank systems effectively improved pedal force effectiveness with conflicting effects on economy/efficiency and performance. The unclear reliability of pedal force effectiveness across different days provided motivation to compare pedal force variables measured on different days (Chapter 6). Trivial differences in peak normal and anterior-posterior forces, total pedal force and index of effectiveness were observed between days. Greater reliability was found for peak normal and total force applied on the pedal, with variability increasing for anterior-posterior force and index of effectiveness. Pedal force variables were highly reliable between two to seven days of testing with similar results compared to oxygen uptake assessed during an incremental step test to exhaustion. Another issue that emerged from the literature review was that the designs for training interventions were limited to low exercise intensity (up to 80% of maximal oxygen uptake) and pedaling cadence (<80 rpm) compared to racing conditions. A pilot study that examined the feasibility of training cyclists to improve pedal force effectiveness showed large increases in right normal and resultant pedal force application for the group receiving pedal force effectiveness feedback (FEG) and for the group receiving peak normal force feedback (PFG). Large improvements in right pedal force effectiveness contrasted with large reductions in left pedal force effectiveness for FEG and PFG. Average power output was greater and pedalling cadence was slower in the FEG group and there was a small decrease in performance time. Preliminary results indicate that force effectiveness training can be translated into greater power output but not in better performance time during 4-km time trials.

Power meters such as the SRM® are becoming popular in cycling communities because they can provide easy measures of power output during training and racing. However, the use of the SRM® to assess bilateral crank torque has not been assessed in terms of validity. One study (Chapter 8) reported in this thesis highlighted a lower estimate of peak crank torque for the SRM® torque analysis system compared to instrumented pedals. The reduced peak torque also affected the assessment of bilateral asymmetries using the SRM® torque analysis system (Chapter 9). During incremental cycling tests to exhaustion, lower limb asymmetries in peak torque increased at higher workload levels in favour of the dominant leg. Limitations in design of the SRM® torque analysis system may preclude the use of this system to assess crank torque symmetry because substantial asymmetries were observed only using the instrumented pedals.

Predicting knee forces is important to ascertain what movements could lead to overuse injury and what preventive strategies may help to reduce overuse injury risk. In cycling, methods to estimate knee forces (i.e. experimental design and models of knee joint) were limited to low workloads, low pedalling cadence, and used models with incomplete information on knee joint anatomy. The technical note presented using a case study (Chapter 10) enabled computation of patellofemoral compressive, tibiofemoral compressive and tibiofemoral anterior-posterior forces normalized by workload, using inverse dynamics based on sagital plane lower limb kinematics and pedal forces during cycling. The patellar tendon to quadriceps force ratio and the contribution of muscle forces to tibiofemoral joint forces were additions to previous published models. The model improves assessment of knee loads for clinical assessment and is to be provided online as part of the AUT Bicycle Mechanics Clinic.

Saddle position changes and knee forces, force effectiveness and kinematics

Given vertical (height) and horizontal positions of the saddle affect joint kinematics, knee joint forces would change because of changes in muscle length for force production. With this in mind, a literature review was conducted looking at methods for determining bicycle saddle height and the effects of bicycle saddle height on measures of cycling performance and lower limb injury risk (Chapter 11). Methods for determining optimal saddle height are varied and not well established, which have been based on relationships between saddle height and lower limb length (Hamley and Thomas, trochanteric length, length from ischial tuberosity to floor, LeMond, heel methods) or a reference range of knee joint flexion. There is limited information on the effects of saddle height on lower limb injury risk (lower limb kinematics, knee joint forces and moments and muscle mechanics), but more information on the effects of saddle height on cycling performance (performance time, energy expenditure/oxygen uptake, power output, pedal force application). Increasing saddle height can cause increased shortening of the vastii group, but no change in hamstrings length. Length and velocity of contraction in *soleus* seems to be more affected by saddle height than *gastrocnemius*. The majority of evidence suggested that a 5% change in saddle height affected knee joint kinematics by 35% and moments by 16%. Patellofemoral compressive force seems to be inversely related to saddle height but the effects on tibiofemoral forces are uncertain. Changes less than 4% of trochanteric length do not seem to affect injury risk or performance. The main limitations from the reported studies are that different methods have been employed for determining saddle height, small sample sizes have been used, cyclists with low level of expertise have mostly been evaluated, and different outcome variables have been measured. Given that the occurrence of overuse knee joint pain is 50% in cyclists, future studies should focus on how saddle height can be optimized to improve cycling performance and reduce knee joint forces to reduce lower limb injury risk. Given the conflicting evidence on the effects of saddle height changes on performance and lower limb injury risk in cycling, we suggest the saddle height may be set using the knee flexion angle method (25-30°) to reduce the risk of knee injuries and to minimize oxygen uptake. However, this latter recommendation was based on theoretical assumptions that greater knee flexion angle (due to lower saddle height) would lead to increased knee joint forces. The analysis of responses of cyclists with and without knee pain when saddle height was changed was reported in Chapter 12. Effects of saddle height were not meaningfully different for patellofemoral and tibiofemoral forces when comparing cyclists with and without knee pain. Compared to the low saddle height there were large tibiofemoral anterior tibiofemoral forces at optimal (35% without pain, 51% pain) and high saddle heights (76% without pain, 92% pain). Bicycle saddle height can probably be set within a large range of knee motion (i.e., ~44-65° determined during dynamic cycling at the 3 o'clock position) to minimise possible detrimental effects of large patellofemoral and tibiofemoral forces.

Although effects of changes in saddle height on knee force were clear from Chapter 12, effects on pedal force effectiveness of competitive cyclists and triathletes were conflicting, as observed in the literature review. Therefore, Chapters 13 and 14 provided evidence that increases in saddle height (5% of preferred height) resulted in large increases in index of effectiveness (7%) at the optimal saddle height compared to the preferred saddle height for cyclists. Greater knee (9-15%) and smaller hip (5-8%) angles were observed for cyclists and triathletes at the low and preferred saddle heights (triathletes only) compared to high and optimal saddle heights. Smaller hip angle (5%) and greater hip range of motion (9%) were observed at the preferred saddle height for triathletes compared to cyclists. Changes in saddle height up to 5% of preferred saddle height for cyclists and 7% for triathletes affected hip and knee angles but not joint mechanical work.

Changes in horizontal position on the saddle were unclear because most studies merged changes in horizontal and vertical positions on the saddle but cyclists do not change the vertical position on the saddle when moving forward or backward on the saddle during racing. In Chapter 15 effects from forward and backward changes in saddle position on knee forces and kinematics were assessed. Changes to forward/backward saddle positions did not substantially affect compressive forces for patellofemoral (1-4%) or tibiofemoral (1-3%) joints. Tibiofemoral shear force increased the backward compared with preferred or forward saddle positions (19-45%). Knee flexion angle at 3 o'clock (22%) and 6 o'clock crank positions (36%) increased at the forward compared to the backward saddle position. Small increases in knee flexion angle (5-6°) explained the trivial differences in patellofemoral and tibiofemoral compressive forces. Tibiofemoral shear force may be more sensitive to changes in knee joint angle compared to other knee force components. In Chapter 16 we observed that any large effects from changes in position on the saddle occurred for pedal forces, ankle joint work and ankle kinematics. There were large increases in knee joint angle and mechanical work and rectus femoris activation along with smaller hip work at the forward position on the saddle. Differences between cyclists and triathletes were not substantial. Effects of changes in saddle positions were limited to the hip and knee joints.

Thesis limitations

Some methodological limitations occurred in the thesis studies and were discussed where relevant in each chapter. In summary, these limitations mainly surrounded the use of sagital plane assessment of cycling kinetics and kinematics and recruitmentment of cyclists.

- Thoughout the research sagittal plane joint angles were detemined which does not provide information on medio-lateral and rotational movements of the lower limb. This limitation was also observed in pedal force and joint kinetics assessment.

- Recuitment of competitive cyclists and triathletes for studies limited the sample size. Recruitment of more elite cyclists would have improved the possibility of providing practical implications for future training of elite road cyclists.

- Assessment of lower limb angles were conducted in most of the studies which does not provide information on effects of changes in saddle position in spine and upper body joints.

- The knee joint was modelled using sagittal plane forces only, which limits the model to extrapolate for effects from rotational and medio-lateral forces on knee injury risk.

- Pedal force monitoring was conducted using an instrumented pedal developed for Look® compatible cleats. Most cyclists in New Zealand use Shimano® cleats and during data collection sessions, shoes of similar size were provided. Effects of cleat type on pedal forces need to be determined.

- Assessment of bilateral joint kinematics and pedal forces would have provided information on asymmetries in joint kinetics and kinematics when saddle height was changed.

- Movement on the saddle during changes in saddle height and horizontal position was measured using one reflective marker attached to the cyclists' sacrum. It may have been more valid and reliable to monitor the body position on the saddle using pressure sensors on the surface of the saddle. However, changes in horizontal position of the saddle were substantial enough to be detected by video analysis and any substantial changes were observed when saddle height was changed.

While this thesis has demonstrated substantial outcomes that further broaden the body of knowledge surrounding the body position on the bicycle, methods for cycling biomechanics assessment and saddle height effects on lower limb kinetics and kinematics paradigms, the results need to be interpreted with caution given the aforementioned thesis limitations.

Recommendations for future research

The findings of this thesis have lead to the following recommendations for future research:

- Three dimensional joint kinetics and kinematics may provide more realistic analysis of cycling biomechanics. The use of 3D pedals and multi-camera system will enhance the ability to determine any relationships between injuries and body position on the bicycle.

- Upper body position should be a focus for future research given lumbar pain is common in cycling.

- Force feedback training interventions should use high intensity workload and pedalling cadence to enable cyclists to better transfer improvements in pedal force effectiveness to racing exercise intensity. The use of single leg cycling for training may improve attention of cyclists to each leg for acquisition of new motor control skills to improve pedal force effectiveness.

- Training effects associated with body position on the bicycle, particularly the saddle height, may provide a better picture of the role of bicycle configuration to prevent injuries. Large changes in saddle height after adaptation training may reduce joint forces and decrease injury risk.

- Reliability of joint kinetics and kinematics taken on different days may highlight the smallest worthwhile effect. Clinical use of the smallest worthwhile effect for individual assessment of cyclists is a potential avenue for research.

- Validity of other devices for force measurement may enhance knowledge of strengths and limitations of various systems for monitoring pedalling kinetics.

Effects of saddle height in anaerobic exercise conditions (i.e. sprints) measured in the field may shed light on joint kinetics and kinematics adaptation and performance factors.

Conclusions

This thesis demonstrated substantial outcomes that further broaden the body of knowledge surrounding body position on the bicycle, methods for cycling biomechanics assessment and saddle height effects on lower limb kinetics and kinematics paradigms. The main question this thesis addressed was “What are the effects of saddle position on pedalling technique and what methods should be used to assess pedalling kinetics and kinematics of cyclists and triathletes?” Specific questions were “What is the validity and reliability of methods for determining body position (kinematics) and pedalling kinetics in cycling?” and “What are the effects from changes in saddle height and horizontal position on pedalling kinetics and kinematics of cyclists and triathletes?” The effectiveness of changing body position on the bicycle to reduce the risk of injuries is uncertain particularly because guidelines for bicycle configuration are limited to road cyclists, without application to other disciplines. The validity of using static positions of cyclists on the bicycle to infer dynamic joint kinematics was limited to the measurements of lower limb angles at the 3 o’clock crank position. Reliability of pedal force variables across seven days of testing was small, which supports the use of these measures for monitoring training. Training cyclists to improve pedal force effectiveness and peak normal pedal force resulted in improvements in pedal force effectiveness and average power output during 4-km time trial training sessions without substantial changes in performance time. Using the SRM® torque analyses system is not recommended for analysis of peak torque and asymmetries of torque in cyclists. The model developed to estimate knee joint forces improved existing models by including the patellar tendon to quadriceps force ratio and the contribution of muscle forces to tibiofemoral joint forces. The effects of saddle height on knee forces were similar when comparing cyclists with and without knee pain. Changes in saddle height greater than the existing recommendations may minimise detrimental effects of large patellofemoral and tibiofemoral forces. Greater changes in saddle height (~7%) affected cyclists but not triathletes pedal force effectiveness with both groups balancing effects from saddle height via changes in hip and knee joint kinematics. Moving forward or backward on the saddle affected joint kinematics and mechanical work but not lower limb joint forces of cyclists and triathletes.

The information gained from this PhD has been used to develop protocols for the AUT Bicycle Mechanics Clinic. This clinic will provide on-going research, education and public services on bicycle configuration, cyclists' kinematics and kinetics and performance and injury risk in cycling. The following table summarises the variables assessed in this thesis that may or may not affect cycling performance and/or injury risk (see table 17.1).

Table 17.1. Summary of main variables assessed in the various studies of this thesis that may and may not affect cycling performance and/or overuse injury risk.

Variable	Performance related	Injury prevention related
Trunk flexion	Yes – reduced frontal area	Unclear effect on low back pain
Saddle inclination	Unknown	Yes – reduced low back pain
Knee medial projection	Unknown	Yes – reduced medial projection decreased injury risk
Tibial internal rotation	Unknown	Yes – reduced internal rotation decreases injury risk
Frontal projected area	Yes – reduced power output	Unknown
Saddle height	Unclear. Up to 3 cm change did not affect joint mechanical work	No. Up to 3 cm change did not affect knee forces
Horizontal body position on the bicycle	Unclear. Up to 6 cm change did not affect joint mechanical work	No. Up to 6 cm change in horizontal body position did not affect knee forces
Knee flexion angle	Unclear. Up to 20° change when the crank is at the 3 o'clock position did not affect joint mechanical work	No. Up to 20° change when the crank was at the 3 o'clock position did not affect knee forces
Force effectiveness	Unclear	Unclear

REFERENCES

Abbiss, C. R., Quod, M. J., Levin, G., Martin, D. T., & Laursen, P. B. (2009). Accuracy of the velotron ergometer and SRM power meter. *International Journal of Sports Medicine, 30*(2), 107-112.

Abbiss, C. R., Quod, M. J., Martin, D. T., Netto, K. J., Nosaka, K., Lee, H., et al. (2006). Dynamic pacing strategies during the cycle phase of an ironman triathlon. *Medicine and Science in Sports and Exercise, 38*(4), 726-734.

Aksu, M., Kaya, D., & Kocadereli, I. (2010). Reliability of reference distances used in photogrammetry. *Angle Orthodontist, 80*(4), 670-677.

Álvarez, G., & Vinyolas, J. (1996). A new bicycle pedal design for on-road measurements of cycling forces. *Journal of Applied Biomechanics, 12*(1), 130-142.

Amann, M., Subudhi, A. W., Walker, J., Eisenman, P., Shultz, B., & Foster, C. (2004). An evaluation of the predictive validity and reliability of ventilatory threshold. *Medicine and Science in Sports and Exercise, 36*(10), 1716-1722.

Amoroso, A., Sanderson, D. J., & Henning, E. M. (1993). *Kinematic and kinetic changes in cycling resulting from fatigue*. Paper presented at the 14th ISB Congress in Biomechanics, Paris, France.

Ansley, L., & Cangley, P. (2009). Determinants of "optimal" cadence during cycling. *European Journal of Sport Science, 9*(2), 61-85.

Asplund, C., & St. Pierre, P. (2004). Knee Pain and Bicycling. *Physician and Sportsmedicine, 32*(4), 23-30.

Austin, N., Keren, T., Wieland, C., & Herzog, W. (2008). *In vivo skeletal muscle fibre function during cycling*. Paper presented at the North American Congress on Biomechanics, Michigan, U.S.A.

Bailey, M. P., Maillardet, F. J., & Messenger, N. (2003). Kinematics of cycling in relation to anterior knee pain and patellar tendinitis. *Journal of Sports Sciences, 21*(8), 649-657.

Barratt, P. (2008). SRM Torque Analysis of Standing Starts in Track Cycling (P85). In M. Estivalet & P. Brisson (Eds.), *The Engineering of Sport 7* (pp. 443-448): Springer Paris.

Bartlett, R., Wheat, J., & Robins, M. (2007). Is movement variability important for sports biomechanists? *Sports Biomechanics, 6*(2), 224-243.

Belli, J. F. C., Chaves, T. C., De Oliveira, A. S., & Grossi, D. B. (2009). Analysis of body posture in children with mild to moderate asthma. *European Journal of Pediatrics, 168*(10), 1207-1216.

Bentley, D. J., Newell, J., & Bishop, D. (2007). Incremental exercise test design and analysis: Implications for performance diagnostics in endurance athletes. *Sports Medicine, 37*(7), 575-586.

Bini, R. R., Carpes, F. P., & Diefenthaeler, F. (2009). Effects of knee frontal plane position on pedal forces during cycling: A preliminary study. *Brazilian Journal of Kineantropometry and Human Performance, 11*(2), 142-150.

Bini, R. R., Carpes, F. P., Diefenthaeler, F., Mota, C. B., & Guimarães, A. C. S. (2008). Physiological and electromyographic responses during 40-km cycling time trial: Relationship to muscle coordination and performance. *Journal of Science and Medicine in Sport, 11*(4), 363-370.

Bini, R. R., & Diefenthaeler, F. (2010). Kinetics and kinematics analysis of incremental cycling to exhaustion. *Sports Biomechanics, 9*(4), 223-235.

Bini, R. R., Diefenthaeler, F., Carpes, F. P., & Mota, C. B. (2007). *External work bilateral symmetry during incremental cycling exercise*. Paper presented at the 25 International Symposium on Biomechanics in Sports, Ouro Preto, Brazil.

Bini, R. R., Diefenthaeler, F., & Mota, C. B. (2010). Fatigue effects on the coordinative pattern during cycling: Kinetics and kinematics evaluation. *Journal of Electromyography and Kinesiology, 20*(1), 102-107.

Bini, R. R., Hume, P. A., & Cervieri, A. (2011). A comparison of cycling SRM crank and strain gauge instrumented pedal measures of peak torque, crank angle at peak torque and power output. *Procedia Engineering, 13*, 56-61.

Bini, R. R., Hume, P. A., & Croft, J. L. (2011a). Effects of bicycle saddle height on knee injury risk and cycling performance. *Sports Medicine, 41*(6), 463-476.

Bini, R. R., Hume, P. A., & Croft, J. L. (2011b). Effects of saddle height on pedal force effectiveness. *Procedia Engineering, 13*, 51-55.

Bini, R. R., Rossato, M., Diefenthaeler, F., Carpes, F. P., Dos Reis, D. C., & Moro, A. R. P. (2010). Pedaling cadence effects on joint mechanical work during cycling. *Isokinetics and Exercise Science, 18*(1), 7-13.

Bini, R. R., Tamborindeguy, A. C., & Mota, C. B. (2010). Effects of saddle height, pedaling cadence, and workload on joint kinetics and kinematics during cycling. *Journal of Sport Rehabilitation, 19*(3), 301-314.

Black, A. H., Sanderson, D. J., & Hennig, E. M. (1993). *Kinematic and kinetic changes during an incremental exercise test on a bicycle ergometer*. Paper presented at the 14th ISB Congress in Biomechanics, Paris, France.

Bohm, H., Siebert, S., & Walsh, M. (2008). Effects of short-term training using SmartCranks on cycle work distribution and power output during cycling. *European Journal of Applied Physiology, 103*(2), 225-232.

Borysewicz, E. (1985). *Bicycle Road Racing Complete Program for Training and Competition* Brattleboro: Velo-new Corporation.

Boyd, T., Hull, M. L., & Wootten, D. (1996). An improved accuracy six-load component pedal dynamometer for cycling. *Journal of Biomechanics, 29*(8), 1105-1110.

Brechter, J. H., & Powers, C. M. (2002). Patellofemoral joint stress during stair ascent and descent in persons with and without patellofemoral pain. *Gait and Posture, 16*(2), 115-123.

Bressel, E. (2001). The influence of ergometer pedaling direction on peak patellofemoral joint forces. *Clinical Biomechanics, 16*(5), 431-437.

Bressel, E., & Larson, B. J. (2003). Bicycle seat designs and their effect on pelvic angle, trunk angle, and comfort. *Medicine and Science in Sports and Exercise, 35*(2), 327-332.

Broker, J. P., & Gregor, R. J. (1990). A dual piezoelectric element force pedal for kinetic analysis of cycling. *International Journal of Sports Biomechanics, 6*, 394-403.

Broker, J. P., Gregor, R. J., & Schmidt, R. A. (1993). Extrinsic feedback and the learning of kinetic patterns in cycling. *Journal of Applied Biomechanics, 9*, 111-123.

Burke, E. R. (2002). Perfect positioning. In P. A. Burke ER (Ed.), *Serious Cycling* (pp. 235-245). Champaign: Human Kinetics.

Burke, E. R., & Pruitt, A. L. (2003). Body positioning for cycling. In E. R. Burke (Ed.), *High-Tech Cycling* (pp. 69-92). Champaign, IL: Human Kinetics.

Burnett, A. F., Cornelius, M. W., Dankaerts, W., & O'Sullivan, P. B. (2004). Spinal kinematics and trunk muscle activity in cyclists: A comparison between healthy controls and non-specific chronic low back pain subjects - A pilot investigation. *Manual Therapy, 9*(4), 211-219.

Caldwell, G. E., Li, L., McCole, S. D., & Hagberg, J. M. (1998). Pedal and crank kinetics in uphill cycling. *Journal of Applied Biomechanics, 14*(3), 245-259.

Callaghan, M. J. (2005). Lower body problems and injury in cycling. *Journal of Bodywork and Movement Therapies, 9*(3), 226-236.

Candotti, C. T., Loss, J. F., Bagatini, D., Soares, D. P., da Rocha, E. K., de Oliveira, A. R., et al. (2009). Cocontraction and economy of triathletes and cyclists at different cadences during cycling motion. *Journal of Electromyography and Kinesiology, 19*(5), 915-921.

Candotti, C. T., Loss, J. F., Melo, M. D. O., La Torre, M., Pasini, M., Dutra, L. A., et al. (2008). Comparing the lactate and EMG thresholds of recreational cyclists during incremental pedaling exercise. *Canadian Journal of Physiology and Pharmacology, 86*(5), 272-278.

Candotti, C. T., Ribeiro, J., Soares, D. P., De Oliveira, A. R., Loss, J. F., & Guimarães, A. C. S. (2007). Effective force and economy of triathletes and cyclists. *Sports Biomechanics, 6*(1), 31-43.

Carpes, F. P., Dagnese, F., Bini, R. R., Diefenthaeler, F., Rossato, M., Mota, C. B., et al. (2006). Pedaling kinematics characteristics of competitive cyclists of different disciplines. *Portuguese Journal of Sports Sciences, 6*(1), 7-14.

Carpes, F. P., Diefenthaeler, F., Bini, R. R., Stefanyshyn, D., Faria, I. E., & Mota, C. B. (2010). Does leg preference affect muscle activation and efficiency? *Journal of Electromyography and Kinesiology, 20*(6), 1230-1236.

Carpes, F. P., Diefenthaeler, F., Bini, R. R., Stefanyshyn, D. J., Faria, I. E., & Mota, C. B. (2011). Influence of leg preference on bilateral muscle activation during cycling. *Journal of Sports Sciences, 29*(2), 151-159.

Carpes, F. P., Mota, C. B., & Faria, I. E. (2010). On the bilateral asymmetry during running and cycling - A review considering leg preference. *Physical Therapy in Sport, 11*(4), 136-142.

Carpes, F. P., Rossato, M., Faria, I. E., & Mota, C. B. (2007a). Bilateral pedaling asymmetry during a simulated 40-km cycling time-trial. *Journal of Sports Medicine and Physical Fitness, 47*(1), 51-57.

Carpes, F. P., Rossato, M., Faria, I. E., & Mota, C. B. (2007b). Influence of exercise intensity on bilateral pedaling symmetry. *Progress in motor control IV, 11*, S54-S55.

Cavanagh, P. R., & Sanderson, D. J. (1986). The biomechanics of cycling: Studies of the pedaling mechanics of elite pursuit riders. In E. R. Burke (Ed.), *Science of Cycling* (Vol. 1, pp. 91-122). Champaign, IL: Human Kinetics Publishers.

Chapman, A., Vicenzino, B., Blanch, P., & Hodges, P. (2009). Do differences in muscle recruitment between novice and elite cyclists reflect different movement patterns or less skilled muscle recruitment? *Journal of Science and Medicine in Sport, 12*(1), 31-34.

Chapman, A., Vicenzino, B., Blanch, P., & Hodges, P. W. (2008). Patterns of leg muscle recruitment vary between novice and highly trained cyclists. *Journal of Electromyography and Kinesiology, 18*(3), 359-371.

Chapman, A. R., Vicenzino, B., Blanch, P., Dowlan, S., & Hodges, P. W. (2008). Does cycling effect motor coordination of the leg during running in elite triathletes? *Journal of Science and Medicine in Sport, 11*(4), 371-380.

Chapman, A. R., Vicenzino, B., Blanch, P., & Hodges, P. W. (2007). Leg muscle recruitment during cycling is less developed in triathletes than cyclists despite matched cycling training loads. *Experimental Brain Research, 181*(3), 503-518.

Chen, J. Y., Wen, F., Lin, C. H., & Lin, D. F. (2005). A new automatic measurement system for a bicycle. *Proceedings of the 2005 IEEE International Conference on Mechatronics, ICM '05, 2005*(1), 514-519.

Christiaans, H. H. C. M., & Bremner, A. (1998). Comfort on bicycles and the validity of a commercial bicycle fitting system. *Applied Ergonomics, 29*(3), 201-211.

Chunfu, G. (2009). Study on the sensor for the pedal push force of bicycle. *ICEMI 2009 - Proceedings of 9th International Conference on Electronic Measurement and Instruments, 1*(1), 2977-2980.

Clarsen, B., Krosshaug, T., & Bahr, R. (2010). Overuse injuries in professional road cyclists. *American Journal of Sports Medicine, 38*(12), 2494-2501.

Clarys, J. P., Alewaeters, K., & Zinzen, E. (2001). The influence of geographic variations on the muscular activity in selected sports movements. *Journal of Electromyography and Kinesiology, 11*(6), 451-457.

Cohen, J. (1988). *Statistical power analysis for the behavioral sciences* (2nd ed ed.). Hillsdale, NJ: Lawrence Earlbaum Associates.

Cohen, Z. A., Roglic, H., Grelsamer, R. P., Henry, J. H., Levine, W. N., Mow, V. C., et al. (2001). Patellofemoral stresses during open and closed kinetic chain exercises : An analysis using computer simulation. *American Journal of Sports Medicine, 29*(4), 480-487.

Coyle, E. F. (2005). Improved muscular efficiency displayed as Tour de France champion matures. *Journal of Applied Physiology, 98*(6), 2191-2196.

Coyle, E. F., Feltner, M. E., Kautz, S. A., Hamilton, M. T., Montain, S. J., Baylor, A. M., et al. (1991). Physiological and biomechanical factors associated with elite endurance cycling performance. *Medicine and Science in Sports and Exercise, 23*(1), 93-107.

Crouter, S. E., Antczak, A., Hudak, J. R., DellaValle, D. M., & Haas, J. D. (2006). Accuracy and reliability of the ParvoMedics TrueOne 2400 and MedGraphics VO2000 metabolic systems. *European Journal of Applied Physiology, 98*(2), 139-151.

Dal Monte, A., Manoni, A., & Fucci, S. (1973). Biomechanical study of competitive cycling: The forces exercised on the pedals. In S. Venerando & A. Wartenweiler (Eds.), *Medicine and Sport: Vol. 8. Biomechanics III* (Vol. 8, pp. 434-439). Georgia, USA: University Park Press.

Daly, D. J., & Cavanagh, P. R. (1976). Asymmetry in bicycle ergometer pedalling. *Medicine and Science in Sports and Exercise, 8*(3), 204-208.

De Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics, 29*(9), 1223-1230.

De Luca, C. J. (1997). The use of surface electromyography in biomechanics. *Journal of Applied Biomechanics, 13*(2), 135-163.

De Vey Mestdagh, K. (1998). Personal perspective: In search of an optimum cycling posture. *Applied Ergonomics, 29*(5), 325-334.

Desipres, M. (1974). An electromyographic study of competitive road cycling conditions simulated on a treadmill. In R. C. Nelson & C. Morehouse (Eds.), *Biomechanics IV* (pp. 349–355). Baltimore: University Park Press.

Dettori, N. J., & Norvell, D. C. (2006). Non-traumatic bicycle injuries: A review of the literature. *Sports Medicine, 36*(1), 7-18.

Diefenthaeler, F., Bini, R. R., Carpes, F. P., & Vaz, M. A. (2007). *Analysis of pedaling technique during a maximal cycling exercise*. Paper presented at the 25 International Symposium on Biomechanics in Sports, Ouro Preto, Brazil.

Diefenthaeler, F., Bini, R. R., Laitano, O., Guimarães, A. C. S., Nabinger, E., Carpes, F. P., et al. (2006). Assessment of the effects of saddle position on cyclists pedaling technique. *Medicine and Science in Sports and Exercise, 38*(5), S181.

Diefenthaeler, F., Bini, R. R., Nabinger, E., Laitano, O., Carpes, F. P., Mota, C. B., et al. (2008). Methodological proposal for evaluation of the pedaling technique of cyclists: A case study. *Brazilian Journal of Sports Medicine, 14*(2), 155-158.

Diefenthaeler, F., Coyle, E. F., Bini, R. R., Carpes, F. P., & Vaz, M. A. (in press). Muscle activity and pedal force profile of triathletes during cycling to exhaustion. *Sports Biomechanics*.

Dorel, S., Couturier, A., & Hug, F. (2009). Influence of different racing positions on mechanical and electromyographic patterns during pedalling. *Scandinavian Journal of Medicine and Science in Sports, 19*(1), 44-54.

Dorel, S., Couturier, A., Lacour, J. R., Vandewalle, H., Hautier, C., & Hug, F. (2010). Force-velocity relationship in cycling revisited: Benefit of two-dimensional pedal forces analysis. *Medicine and Science in Sports and Exercise, 42*(6), 1174-1183.

Dorel, S., Drouet, J.-M., Hug, F., Lepretre, P.-M., & Champoux, Y. (2008). New instrumented pedals to quantify 2D forces at the shoe-pedal interface in ecological conditions: Preliminary study in elite track cyclists. *Computer Methods in Biomechanics and Biomedical Engineering, 11*(S1), 89-90.

Dorel, S., Drouet, J. M., Couturier, A., Champoux, Y., & Hug, F. (2009). Changes of pedaling technique and muscle coordination during an exhaustive exercise. *Medicine and Science in Sports and Exercise, 41*(6), 1277-1286.

Du Bois, D., & Du Bois, E. F. (1916). A formula to estimate the approximate surface area if height and weight be known. *Archives of Internal Medicine, 17*, 863-871.

Edeline, O., Polin, D., Tourny-Chollet, C., & Weber, J. (2004). Effect of workload on bilateral pedaling kinematics in non-trained cyclists. *Journal of Human Movement Studies, 46*(6), 493-517.

Edwards, L. M., Jobson, S. A., George, S. R., Day, S. H., & Nevill, A. M. (2009). Whole-body efficiency is negatively correlated with minimum torque per duty cycle in trained cyclists. *Journal of Sports Sciences, 27*(4), 319-325.

Emanuele, U., Horn, T., & Denoth, J. (2011). *Influence of racing position on cycling patterns*. Paper presented at the 29 International Conference on Biomechanics in Sports, Porto, Portugal.

Ericson, M. O., Bratt, A., & Nisell, R. (1986). Load moments about the hip and knee joints during ergometer cycling. *Scandinavian Journal of Rehabilitation Medicine, 18*(4), 165-172.

Ericson, M. O., & Nisell, R. (1986). Tibiofemoral joint forces during ergometer cycling. *American Journal of Sports Medicine, 14*(4), 285-290.

Ericson, M. O., & Nisell, R. (1987). Patellofemoral joint forces during ergometric cycling. *Physical Therapy, 67*(9), 1365-1369.

Ericson, M. O., & Nisell, R. (1988). Efficiency of pedal forces during ergometer cycling. *International Journal of Sports Medicine, 9*(2), 118-122.

Ericson, M. O., Nisell, R., Arborelius, U. P., & Ekholm, J. (1985). Muscular activity during ergometer cycling. *Scandinavian Journal of Rehabilitation Medicine, 17*(2), 53-61.

Ericson, M. O., Nisell, R., & Ekholm, J. (1984). Varus and valgus loads on the knee joint during ergometer cycling. *Scandinavian Journal of Sports Sciences, 6*(2), 39-45.

Escamilla, R. F. (2001). Knee biomechanics of the dynamic squat exercise. *Medicine and Science in Sports and Exercise, 33*(1), 127-141.

Faria, I. E., & Cavanagh, P. R. (1978). *The Physiology and Biomechanics of Cycling*. New York: John Wiley.

Farrell, K. C., Reisinger, K. D., & Tillman, M. D. (2003). Force and repetition in cycling: Possible implications for iliotibial band friction syndrome. *Knee, 10*(1), 103-109.

Ferber, R., Sheerin, K., & Kendall, K. D. (2009). Measurement error of rearfoot kinematics during running between a 100Hz and 30Hz camera. *International SportMed Journal, 10*(3).

Fernandez-Pena, E., Lucertini, F., & Ditroilo, M. (2009). Training with independent cranks alters muscle coordination pattern in cyclists. *Journal of Strength and Conditioning Research, 23*(6), 1764-1772.

Fleming, B. C., Beynnon, B. D., Renstrom, P. A., Peura, G. D., Nichols, C. E., & Johnson, R. J. (1998). The strain behavior of the anterior cruciate ligament during bicycling. An in vivo study. *American Journal of Sports Medicine, 26*(1), 109-118.

FPCUB. (2008). Federação Portuguesa de Cicloturismo e Utilizadores de Bicicleta. O papel da bicicleta na mobilidade urbana. *www.transportesemrevista.com, 62*(abril 2008), 18-24.

Frigo, C., & Pedotti, A. (1978). Determination of muscle length during locomotion. In E. Assmussen & K. Jorgensen (Eds.), *International series on biomechanics* (Vol. 2A, pp. 355-360). Baltimore: University Park Press.

García-López, J., Rodríguez-Marroyo, J. A., Juneau, C. E., Peleteiro, J., Martínez, A. C., & Villa, J. G. (2008). Reference values and improvement of aerodynamic drag in professional cyclists. *Journal of Sports Sciences, 26*(3), 277-286.

Gardner, A. S., Stephens, S., Martin, D. T., Lawton, E., Lee, H., & Jenkins, D. (2004). Accuracy of SRM and Power Tap power monitoring systems for bicycling. *Medicine and Science in Sports and Exercise, 36*(7), 1252-1258.

Garside, I., & Doran, D. A. (2000). Effects of bicycle frame ergonomics on triathlon 10-km running performance. *Journal of Sports Sciences, 18*(10), 825-833.

Gavan, J. A., Washburn, S. L., & Lewis, P. H. (1952). Photography: an anthropometric tool. *American journal of physical anthropology, 10*(3), 331-353.

Goffredo, M., Schmid, M., Conforto, S., Carli, M., Neri, A., & D'Alessio, T. (2009). Markerless human motion analysis in Gauss-Laguerre transform domain: An application to sit-to-stand in young and elderly people. *IEEE Transactions on Information Technology in Biomedicine, 13*(2), 207-216.

Gonzalez, H., & Hull, M. L. (1989). Multivariable optimization of cycling biomechanics. *Journal of Biomechanics, 22*(11-12), 1151-1161.

Gosling, C. M., Gabbe, B. J., & Forbes, A. B. (2008). Triathlon related musculoskeletal injuries: The status of injury prevention knowledge. *Journal of Science and Medicine in Sport, 11*(4), 396-406.

Gregor, R. J. (2000). Biomechanics of cycling. In W. E. Garret & D. T. Kirkendall (Eds.), *Exercise and Sports Science* (pp. 549-571). Philadelphia: Lippincott Williams & Wilkings.

Gregor, R. J., & Wheeler, J. B. (1994). Biomechanical factors associated with shoe/pedal interfaces: Implications for injury. *Sports Medicine, 17*(2), 117-131.

Grieve, D., Pheasant, S., & P.R., C. (1978). Prediction of gastrocnemius length from knee and ankle joint posture. In E. Assmussen & K. Jorgensen (Eds.), *International series on biomechanics* (Vol. 2A, pp. 405-412). Baltimore: University Park Press.

Gruen, A. (1997). Fundamentals of videogrammetry - A review. *Human Movement Science, 16*(2-3), 155-187.

Guye, M. (1896). Analyse dynamometrique de la locomotive a bicyclette. *La Nature, 1212*, 177.

Hamley, E. J., & Thomas, V. (1967). Physiological and postural factors in the calibration of the bicycle ergometer. *Journal of Physiology, 191*(2), 55P-56P.

Harman, E., Knuttgen, H. G., & Frykman, P. (1987). Automated data collection and processing for a cycle ergometer. *Journal of Applied Physiology, 62*(2), 831-836.

Harvey, M., Bonning, J., & Cave, G. (2008). Injury severity in mountain bike competitions. *International SportMed Journal, 9*(4), 182-183.

Hasson, C. J., Caldwell, G. E., & van Emmerik, R. E. A. (2008). Changes in muscle and joint coordination in learning to direct forces. *Human Movement Science, 27*(4), 590-609.

Hawkins, D., & Hull, M. L. (1990). A method for determining lower extremity muscle-tendon lengths during flexion/extension movements. *Journal of Biomechanics, 23*(5), 487-494.

Henke, T. (1998). *Real-time feedback of pedal forces for the optimization of pedaling technique in competitive cycling*. Paper presented at the 16 International Symposium on Biomechanics in Sports, Konstanz - Germany.

Herzog, W., Guimaraes, A. C., Anton, M. G., & Carter-Erdman, K. A. (1991). Moment-length relations of rectus femoris muscles of speed skaters/cyclists and runners. *Medicine and Science in Sports and Exercise, 23*(11), 1289-1296.

Herzog, W., & Read, L. J. (1993). Lines of action and moment arms of the major force-carrying structures crossing the human knee joint. *Journal of Anatomy, 182*(2), 213-230.

Hoes, M. J. A. J. M., Binkhorst, R. A., Smeekes-Kuyl, A. E. M. C., & Vissers, A. C. A. (1968). Measurement of forces exerted on pedal and crank during work on a bicycle ergometer at different loads. *European Journal of Applied Physiology and Occupational Physiology, 26*(1), 33-42.

Holderbaum, G., Bini, R. R., Nabinger, E., & Guimaraes, A. C. S. (2005). *Methodological approach for learning pedaling technique based on augmented visual feedback*. Paper presented at the XI Brazilian Conference of Biomechanics, João Pessoa, Brazil.

Holderbaum, G. G., Guimarães, A. C. S., & Petersen, R. D. S. (2007). *Analysis of the recovering phase after the cycling practice using augmented visual feedback*. Paper presented at the 25 International Symposium on Biomechanics in Sports, Ouro Preto, Brazil.

Holmes, J. C., Pruitt, A. L., & Whalen, N. J. (1994). Lower extremity overuse in bicycling. *Clinics in Sports Medicine, 13*(1), 187-203.

Hopker, J., Passfield, L., Coleman, D., Jobson, S., Edwards, L., & Carter, H. (2009). The effects of training on gross efficiency in cycling: a review. *International Journal of Sports Medicine, 30*(12), 845-850.

Hopkins, W. G. (2000). Measures of reliability in sports medicine and science. *Sports Medicine, 30*(1), 1-15.

Hopkins, W. G. (2002). A Scale of Magnitudes for Effect Statistics. *A New View of Statistics* Retrieved 26th January, 2009, from http://www.sportsci.org/resource/stats/effectmag.html

Horscroft, R. D., Davidson, C. J., McDaniel, J., Wagner, B. M., & Martin, J. C. (2003). Effects of saddle height on joint power distribution. *Medicine and Science in Sports and Exercise, 35*(5), S16.

Houtz, S. J., & Fischer, F. J. (1959). An analysis of muscle action and joint excursion during exercise on a stationary bicycle. *J. Bone Joint Surg., 41 A*(1), 123-131.

Hug, F., Drouet, J. M., Champoux, Y., Couturier, A., & Dorel, S. (2008). Interindividual variability of electromyographic patterns and pedal force profiles in trained cyclists. *European Journal of Applied Physiology, 104*(4), 667-678.

Hull, M. L., & Davis, R. R. (1981). Measurement of pedal loading in bicycling: I. Instrumentation. *Journal of Biomechanics, 14*(12), 843-856.

Hull, M. L., Kautz, S., & Beard, A. (1991). An angular velocity profile in cycling derived from mechanical energy analysis. *Journal of Biomechanics, 24*(7), 577-586.

Hunt, M. A., Sanderson, D. J., Moffet, H., & Inglis, J. T. (2003). Biomechanical changes elicited by an anterior cruciate ligament deficiency during steady rate cycling. *Clinical Biomechanics, 18*(5), 393-400.

Hunt, M. A., Sanderson, D. J., Moffet, H., & Inglis, J. T. (2004). Interlimb asymmetry in persons with and without an anterior cruciate ligament deficiency during stationary cycling. *Archives of Physical Medicine and Rehabilitation, 85*(9), 1475-1478.

Jonsdottir, J., Cattaneo, D., Regola, A., Crippa, A., Recalcati, M., Rabuffetti, M., et al. (2007). Concepts of motor learning applied to a rehabilitation protocol using biofeedback to improve gait in a chronic stroke patient: An A-B system study with multiple gait analyses. *Neurorehabilitation and Neural Repair, 21*(2), 190-194.

Jorge, M., & Hull, M. L. (1986). Analysis of EMG measurements during bicycle pedalling. *Journal of Biomechanics, 19*(9), 683-694.

Joseph, S., Ganason, R., Jalil, A. P., Aizam, Z. S., & Wilson, B. (2006). *A functional biomechanical analysis of an elite cyclist.* Paper presented at the International Symposium on Biomechanics in Sports Salzburg - Austria

Kautz, S., Feltner, M. E., Coyle, E. F., & Bailey, M. P. (1991). The pedaling technique of elite endurance cyclists: changes with increasing workload at constant cadence. *International Journal of Sports Biomechanics, 7*, 29-53.

Kautz, S. A., & Hull, M. L. (1993). A theoretical basis for interpreting the force applied to the pedal in cycling. *Journal of Biomechanics, 26*(2), 155-165.

Kautz, S. A., & Neptune, R. R. (2002). Biomechanical determinants of pedaling energetics: Internal and external work are not independent. *Exercise and Sport Sciences Reviews, 30*(4), 159-165.

Kellis, E. (2001). Tibiofemoral joint forces during maximal isokinetic eccentric and concentric efforts of the knee flexors. *Clinical Biomechanics, 16*(3), 229-236.

Kellis, E., & Baltzopoulos, V. (1999). In vivo determination of the patella tendon and hamstrings moment arms in adult males using videofluoroscopy during submaximal knee extension and flexion. *Clinical Biomechanics, 14*(2), 118-124.

Kim, P. T., Jangra, D., Ritchie, A. H., Lower, M. E., Kasic, S., Brown, D. R., et al. (2006). Mountain biking injuries requiring trauma center admission: a 10-year regional trauma system experience. *The Journal of trauma., 60*(2), 312-318.

Knudson, D. (2009). Significant and meaningful effects in sports biomechanics research. *Sports Biomechanics, 8*(1), 96-104.

Korff, T., Fletcher, G., Brown, D., & Romer, L. M. (in press). Effect of ''Pose'' cycling on efficiency and pedaling mechanics. *European Journal of Applied Physiology*.

Korff, T., Romer, L. M., Mayhew, I., & Martin, J. C. (2007). Effect of pedaling technique on mechanical effectiveness and efficiency in cyclists. *Medicine and Science in Sports and Exercise, 39*(6), 991-995.

Kutzner, I., Heinlein, B., Graichen, F., Rohlmann, A., Bender, A., & Bergmann, G. (2008). In vivo measurements of loads during ergometer cycling 6 months post-operatively. *Journal of Biomechanics, 41*(S1), 323.

LaFortune, M. A., & Cavanagh, P. R. (1983a). *Effectiveness and efficiency during bicycle riding.* Paper presented at the International Series on Biomechanics

LaFortune, M. A., & Cavanagh, P. R. (1983b). Effectiveness and efficiency during bicycle riding. In H. K. Matsui & K. Kobayashi (Eds.), *International Series on Biomechanics* (Vol. VIII-B, pp. 928-936). Champaign: Human Kinetics.

Laios, L., & Giannatsis, J. (2010). Ergonomic evaluation and redesign of children bicycles based on anthropometric data. *Applied Ergonomics, 41*(3), 428-435.

Laursen, P. B., Shing, C. M., & Jenkins, D. G. (2003). Reproducibility of a laboratory-based 40-km cycle time-trial on a stationary wind-trainer in highly trained cyclists. *International Journal of Sports Medicine, 24*(7), 481-485.

Leirdal, S., & Ettema, G. (2011). Pedaling technique and energy cost in cycling. *Medicine and Science in Sports and Exercise, 43*(4), 701-705.

Leirdal, S., & Ettema, G. (in press). The relationship between cadence, pedalling technique and gross efficiency in cycling. *European Journal of Applied Physiology*.

Liebermann, D. G., Katz, L., Hughes, M. D., Bartlett, R. M., McClements, J., & Franks, I. M. (2002). Advances in the application of information technology to sport performance. *Journal of Sports Sciences, 20*(10), 755-769.

Litzenberger, S., Illes, S., Hren, M., Reichel, M., & Sabo, M. (2008). Influence of pedal foot position on muscular activity during ergometer cycling (P242). In M. Estivalet & P. Brisson (Eds.), *The Engineering of Sport 7* (Vol. 1, pp. 215-222): Springer Paris.

Loras, H., Ettema, G., & Leirdal, S. (2009). The muscle force component in pedaling retains constant direction across pedaling rates. *Journal of Applied Biomechanics, 25*(1), 85-92.

Lucía, A., Hoyos, J., Pérez, M., Santalla, A., & Chicharro, J. L. (2002). Inverse relationship between VO2max and economy/efficiency in world-class cyclists. *Medicine and Science in Sports and Exercise, 34*(12), 2079-2084.

Lucia, A., San Juan, A. F., Montilla, M., Canete, S., Santalla, A., Earnest, C., et al. (2004). In professional road cyclists, low pedaling cadences are less efficient. *Medicine and Science in Sports and Exercise, 36*(6), 1048-1054.

Luttrell, M. D., & Potteiger, J. A. (2003). Effects of Short-Term Training Using Powercranks on Cardiovascular Fitness and Cycling Efficiency. *Journal of Strength and Conditioning Research, 17*(4), 785-791.

MacDermid, P. W., & Edwards, A. M. (2010). Influence of crank length on cycle ergometry performance of well-trained female cross-country mountain bike athletes. *European Journal of Applied Physiology, 108*(1), 177-182.

MacIntosh, B. R., Neptune, R. R., & Horton, J. F. (2000). Cadence, power, and muscle activation in cycle ergometry. *Medicine and Science in Sports and Exercise, 32*(7), 1281-1287.

Madsen, B. L., Noer, H. H., Carstensen, J. P., & Normark, F. (2000). Long-term results of periosteal transplantation in osteochondritis dissecans of the knee. *Orthopedics, 23*(3), 223-226.

Malek, M. H., Housh, T. J., Coburn, J. W., Weir, J. P., Schmidt, R. J., & Beck, T. W. (2006). The effects of interelectrode distance on electromyographic amplitude and mean power frequency during incremental cycle ergometry. *Journal of Neuroscience Methods, 151*(2), 139-147.

Marfell-Jones, M., Olds, T., Stewart, A., & Carter, L. (2006). *International standards for anthropometric assessment*. Potchefstroom, South Africa: ISAK.

Marsh, A. P., Martin, P. E., & Sanderson, D. J. (2000). Is a joint moment-based cost function associated with preferred cycling cadence? *Journal of Biomechanics, 33*(2), 173-180.

Martin, J. C., & Brown, N. A. T. (2009). Joint-specific power production and fatigue during maximal cycling. *Journal of Biomechanics, 42*(4), 474-479.

Mason, J. J., Leszko, F., Johnson, T., & Komistek, R. D. (2008). Patellofemoral joint forces. *Journal of Biomechanics, 41*(11), 2337-2348.

Matsumoto, S., Tokuyasu, T., & Ohba, K. (2009). A study on postural optimization for bicycle exercise based on electromyography. *Artificial Life and Robotics, 14*(2), 144-149.

Matthews, L. S., Sonstegard, D. A., & Henke, J. A. (1977). Load bearing characteristics of the patello-femoral joint. *Acta Orthopaedica Scandinavica, 48*(5), 511-516.

McCole, S. D., Claney, K., Conte, J. C., Anderson, R., & Hagberg, J. M. (1990). Energy expenditure during bicycling. *Journal of Applied Physiology, 68*(2), 748-753.

McCoy, R. W., & Gregor, R. J. (1989). The effects of varying seat position on knee loads during cycling. *Medicine and Science in Sports and Exercise, 21*(S(2)), S79.

McLeod, W. D., & Blackburn, T. A. (1980). Biomechanics of knee rehabilitation with cycling. *American Journal of Sports Medicine, 8*(3), 175-180.

Mellion, M. B. (1991). Common cycling injuries. Management and prevention. *Sports Medicine, 11*(1), 52-70.

Mellow, P., Hume, P., Keogh, J., & Pearson, S. (2003). Digital photogrammetry versus ISAK anthropometry in determining body lengths of powerlifters. *Journal of Science and Medicine in Sport, 6(Supplement)*, 25-25.

Merletti, R., Botter, A., Troiano, A., Merlo, E., & Minetto, M. A. (2009). Technology and instrumentation for detection and conditioning of the surface electromyographic signal: State of the art. *Clinical Biomechanics, 24*(2), 122-134.

Mimmi, G., Pennacchi, P., & Frosini, L. (2004). Biomechanical analysis of pedalling for rehabilitation purposes: Experimental results on two pathological subjects and comparison with non-pathological findings. *Computer Methods in Biomechanics and Biomedical Engineering, 7*(6), 339-345.

Minetti, A. E. (2011). Bioenergetics and biomechanics of cycling: The role of 'internal work'. *European Journal of Applied Physiology, 111*(3), 323-329.

Moore, F. (2008). Practical guide - Bike set-up in cyclilng. *SportEX medicine, 37*(6), 4.

Mornieux, G., Gollhofer, A., & Stapelfeldt, B. (2010). Muscle coordination while pulling up during cycling. *International Journal of Sports Medicine, 31*(12), 843-846.

Mornieux, G., Guenette, J. A., Sheel, A. W., & Sanderson, D. J. (2007). Influence of cadence, power output and hypoxia on the joint moment distribution during cycling. *European Journal of Applied Physiology, 102*(1), 11-18.

Mornieux, G., & Stapelfeldt, B. (in press). Modification of the pedaling technique after a training based on the pedal forces feedback. *Science et Motricité*.

Mornieux, G., Stapelfeldt, B., Collhofer, A., & Belli, A. (2008). Effects of pedal type and pull-up action during cycling. *International Journal of Sports Medicine, 29*(10), 817-822.

Mornieux, G., Zameziati, K., Mutter, E., Bonnefoy, R., & Belli, A. (2006). A cycle ergometer mounted on a standard force platform for three-dimensional pedal forces measurement during cycling. *Journal of Biomechanics, 39*(7), 1296-1303.

Morton, N. A., Maletsky, L. P., Pal, S., & Laz, P. J. (2007). Effect of variability in anatomical landmark location on knee kinematic description. *Journal of Orthopaedic Research, 25*(9), 1221-1230.

Moss, A., Cheryl, A. J., McKenzie, J. M., Ruby, B. C., & Heil, D. P. (2005). Predicting Projected Frontal Area For Cycling In Women Competing At The 2004 Hawaiian Ironman Triathlon. *Medicine & Science in Sports & Exercise, 37*(5), S105.

Muraoka, T., Kawakami, Y., Tachi, M., & Fukunaga, T. (2001). Muscle fiber and tendon length changes in the human vastus lateralis during slow pedaling. *Journal of Applied Physiology, 91*(5), 2035-2040.

Nabinger, E., Iturrioz, I., & Zaro, M. A. (2002). *Development of a triaxial force platform for the measurement of force at a bicycle pedal.* Paper presented at the International Symposium of Biomechanics in Sports, Caceres - Extremadura - Spain

Neptune, R. R., & Herzog, W. (1999). The association between negative muscle work and pedaling rate. *Journal of Biomechanics, 32*(10), 1021-1026.

Neptune, R. R., & Hull, M. L. (1995). Accuracy assessment of methods for determining hip movement in seated cycling. *Journal of Biomechanics, 28*(4), 423-437.

Neptune, R. R., & Hull, M. L. (1996). Methods for determining hip movement in seated cycling and their effect on kinematics and kinetics. *Journal of Applied Biomechanics, 12*(4), 493-507.

Neptune, R. R., & Kautz, S. A. (2000). Knee joint loading in forward versus backward pedaling: Implications for rehabilitation strategies. *Clinical Biomechanics, 15*(7), 528-535.

Neptune, R. R., Kautz, S. A., & Hull, M. L. (1997). The effect of pedaling rate on coordination in cycling. *Journal of Biomechanics, 30*(10), 1051-1058.

Neptune, R. R., Kautz, S. A., & Zajac, F. E. (2000). Muscle contributions to specific biomechanical functions do not change in forward versus backward pedaling. *Journal of Biomechanics, 33*(2), 155-164.

Neptune, R. R., McGowan, C. P., & Fiandt, J. M. (2009). The influence of muscle physiology and advanced technology on sports performance. *Annual Review of Biomedical Engineering, 11*, 81-107.

Neptune, R. R., & Van Den Bogert, A. J. (1998). Standard mechanical energy analyses do not correlate with muscle work in cycling. *Journal of Biomechanics, 31*(3), 239-245.

Newmiller, J., Hull, M. L., & Zajac, F. E. (1988). A mechanically decoupled two force component bicycle pedal dynamometer. *Journal of Biomechanics, 21*(5), 375-386.

Nishiyama, T., & Sato, T. (2005). *The modification of pedaling skill with real-time representation of pedaling force in non-cyclists*. Paper presented at the 29th Annual Meeting of the American Society of Biomechanics, Cleveland, OH.

Noordhof, D. A., De Koning, J. J., Van Erp, T., Van Keimpema, B., De Ridder, D., Otter, R., et al. (2010). The between and within day variation in gross efficiency. *European Journal of Applied Physiology, 109*(6), 1209-1218.

Nordeen-Snyder, K. S. (1977). The effect of bicycle seat height variation upon oxygen consumption and lower limb kinematics. *Medicine and Science in Sports and Exercise, 9*(2), 113-117.

Olds, T., & Olive, S. (1999). Methodological considerations in the determination of projected frontal area in cyclists. *Journal of Sports Sciences, 17*(4), 335-345.

Paton, C. D. (2009). Effects of shoe cleat position on physiology and performance of competitive cyclists. *International Journal of Sports Physiology and Performance, 4*(4), 517-523.

Patterson, R. P., & Moreno, M. I. (1990). Bicycle pedalling forces as a function of pedalling rate and power output. *Medicine and Science in Sports and Exercise, 22*(4), 512-516.

Peveler, W. W. (2008). Effects of saddle height on economy in cycling. *Journal of Strength and Conditioning Research, 22*(4), 1355-1359.

Peveler, W. W., Bishop, P., Smith, J., Richardson, M., & Whitehorn, E. (2005). Comparing methods for setting saddle height in trained cyclists. *Journal of Exercise Physiology Online, 8*(1), 51-55.

Peveler, W. W., & Green, J. M. (2011). Effects of saddle height on economy and anaerobic power in well-trained cyclists. *Journal of Strength and Conditioning Research, 25*(3), 629-633.

Peveler, W. W., Pounders, J. D., & Bishop, P. A. (2007). Effects of saddle height on anaerobic power production in cycling. *Journal of Strength and Conditioning Research, 21*(4), 1023-1027.

Price, D., & Donne, B. (1997). Effect of variation in seat tube angle at different seat heights on submaximal cycling performance in man. *Journal of Sports Sciences, 15*(4), 395-402.

Rankin, J. W., & Neptune, R. R. (2008). *Determination of the optimal seat position that maximizes average crank power: A theoretical study*. Paper presented at the North American Congress on Biomechanics, Michigan, U.S.A.

Rassier, D. E., MacIntosh, B. R., & Herzog, W. (1999). Length dependence of active force production in skeletal muscle. *Journal of Applied Physiology, 86*(5), 1445-1457.

Redfield, R., & Hull, M. L. (1986). On the relation between joint moments and pedalling rates at constant power in bicycling. *Journal of Biomechanics, 19*(4), 317-329.

Reiser Ii, R. F., Peterson, M. L., & Broker, J. P. (2003). Instrumented bicycle pedals for dynamic measurement of propulsive cycling loads. *Sports Engineering, 6*, 41-48.

Rhea, M. R. (2004). Determining the magnitude of treatment effects in strength training research through the use of the effect size. *Journal of Strength and Conditioning Research, 18*(4), 918-920.

Ricard, M. D., Hills-Meyer, P., Miller, M. G., & Michael, T. J. (2006). The effects of bicycle frame geometry on muscle activation and power during a Wingate anaerobic test. *Journal of Sports Science and Medicine, 5*(1), 25-32.

Robinson, R. O., Herzog, W., & Nigg, B. M. (1987). Use of force platform variables to quantify the effects of chiropractic manipulation on gait symmetry. *Journal of Manipulative and Physiological Therapeutics, 10*(4), 172-176.

Rossato, M., Bini, R. R., Carpes, F. P., Diefenthaeler, F., & Moro, A. R. P. (2008). Cadence and workload effects on pedaling technique of well-trained cyclists. *International Journal of Sports Medicine, 29*(9), 746-752.

Rottenbacher, C., Bonandrini, G., Mimmi, G., & Buzzi, E. (2007). *Special Cycle-Ergometer to Optimize the Rider Position through Force Measurement*. Paper presented at the XVIII Congress AIMETA Italian Association for Theoretical and Applied Mechanics, Brescia.

Rottenbacher, C., Zaccaria, D., Gualea, M. R., Mimmi, G., Bonandrini, G., & Buzzi, E. (2009). *A Study on the Biomechanical Efficiency of Different Cycling Positions*. Paper presented at the XIX Congress AIMETA Italian Association for Theoretical and Applied Mechanics, Ancona.

Rowe, T., Hull, M. L., & Wang, E. L. (1998). A pedal dynamometer for off-road bicycling. *Journal of Biomechanical Engineering, 120*(1), 160-164.

Ruby, P., Hull, M. L., & Hawkins, D. (1992). Three-dimensional knee joint loading during seated cycling. *Journal of Biomechanics, 25*(1), 41-53.

Rugg, S. G., & Gregor, R. J. (1987). The effect of seat height on muscle lengths, velocities and moment arms lengths during cycling. *Journal of Biomechanics, S20*, 899.

Salai, M., Brosh, T., Blankstein, A., Oran, A., & Chechik, A. (1999). Effect of changing the saddle angle on the incidence of low back pain in recreational bicyclists. *British Journal of Sports Medicine, 33*(6), 398-400.

Salsich, G. B., Ward, S. R., Terk, M. R., & Powers, C. M. (2003). In Vivo Assessment of Patellofemoral Joint Contact Area in Individuals Who are Pain Free. *Clinical Orthopaedics and Related Research*(417), 277-284.

Sanderson, D. J. (1991). The influence of cadence and power output on the biomechanics of force application during steady-rate cycling in competitive and recreational cyclists. *Journal of Sports Sciences, 9*(2), 191-203.

Sanderson, D. J., & Amoroso, A. T. (2009). The influence of seat height on the mechanical function of the triceps surae muscles during steady-rate cycling. *Journal of Electromyography and Kinesiology, 19*(6), e465-e471.

Sanderson, D. J., & Black, A. (2003). The effect of prolonged cycling on pedal forces. *Journal of Sports Sciences, 21*(3), 191-199.

Sanderson, D. J., Black, A. H., & Montgomery, J. (1994). The effect of varus and valgus wedges on coronal plane knee motion during steady-rate cycling. *Clinical Journal of Sport Medicine, 4*(2), 120-124.

Sanderson, D. J., & Cavanagh, P. R. (1990). Use of augmented feedback for the modification of the pedaling mechanics of cycling. *Canadian Journal of Sports Sciences, 15*(1), 38-42.

Sanderson, D. J., Martin, P. E., Honeyman, G., & Keefer, J. (2006). Gastrocnemius and soleus muscle length, velocity, and EMG responses to changes in pedalling cadence. *Journal of Electromyography and Kinesiology, 16*(6), 642-649.

Sanner, W. H., & O'Halloran, W. D. (2000). The biomechanics, etiology, and treatment of cycling injuries. *Journal of the American Podiatric Medical Association, 90*(7), 354-376.

Sargeant, A. J., & Davies, C. T. M. (1977). Forces applied to cranks of a bicycle ergometer during one- and two leg cycling. *Journal of Applied Physiology: Respiratory, Environmental and Exercise Physiology, 42*(4), 514-518.

Savelberg, H. H. C. M., & Meijer, K. (2003). Contribution of mono- and biarticular muscles to extending knee joint moments in runners and cyclists. *Journal of Applied Physiology, 94*(6), 2241-2248.

Schwellnus, M. P., & Derman, E. W. (2005). Common injuries in cycling: Prevention, diagnosis and management. *South African Family Practice, 47*(7), 14-19.

Semwal, S. K., & Parker, M. J. (1999). An animation system for biomechanical analysis of leg motion and predicting injuries during cycling. *Real-Time Imaging, 5*(2), 109-123.

Sharma, A., Leszko, F., Komistek, R. D., Scuderi, G. R., Cates Jr, H. E., & Liu, F. (2008). In vivo patellofemoral forces in high flexion total knee arthroplasty. *Journal of Biomechanics, 41*(3), 642-648.

Sharp, A. (1896). *Bicycles and Tricycles*. London: MIT Press (reprinted 1977).

Shennum, P. L., & DeVries, H. A. (1976). The effect of saddle height on oxygen consumption during bicycle ergometer work. *Medicine and Science in Sports and Exercise, 8*(2), 119-121.

Silberman, M. R., Webner, D., Collina, S., & Shiple, B. J. (2005). Road bicycle fit. *Clinical Journal of Sport Medicine, 15*(4), 271-276.

Smak, W., Neptune, R. R., & Hull, M. L. (1999). The influence of pedaling rate on bilateral asymmetry in cycling. *Journal of Biomechanics, 32*(9), 899-906.

Srinivasan, J., & Balasubramanian, V. (2007). Low back pain and muscle fatigue due to road cycling-An sEMG study. *Journal of Bodywork and Movement Therapies, 11*(3), 260-266.

Stapelfeldt, B., Mornieux, G., Oberheim, R., Belli, A., & Gollhofer, A. (2007). Development and evaluation of a new bicycle instrument for measurements of pedal forces and power output in cycling. *International Journal of Sports Medicine, 28*(4), 326-332.

Suriano, R., & Bishop, D. (2010). Physiological attributes of triathletes. *Journal of Science and Medicine in Sport, 13*(3), 340-347.

Takaishi, T., Yamamoto, T., Ono, T., Ito, T., & Moritani, T. (1998). Neuromuscular, metabolic, and kinetic adaptations for skilled pedaling performance in cyclists. *Medicine and Science in Sports and Exercise, 30*(3), 442-449.

Tamborindeguy, A. C., & Bini, R. R. (2011). Does saddle height affect patellofemoral and tibiofemoral forces during bicycling for rehabilitation? *Journal of Bodywork and Movement Therapies, 15* (2), 186-191.

Thambyah, A., Pereira, B. P., & Wyss, U. (2005). Estimation of bone-on-bone contact forces in the tibiofemoral joint during walking. *Knee, 12*(5), 383-388.

Theurel, J., Crepin, M., Foissac, M., & Temprado, J. J. (in press). Effects of different pedalling techniques on muscle fatigue and mechanical efficiency during prolonged cycling. *Scandinavian Journal of Medicine and Science in Sports*.

Ting, L. H., Raasch, C. C., Brown, D. A., Kautz, S. A., & Zajac, F. E. (1998). Sensorimotor state of the contralateral leg affects ipsilateral muscle coordination of pedaling. *Journal of Neurophysiology, 80*(3), 1341-1351.

Triolo, R. J., & Bogie, K. (1999). Lower extremity applications of functional neuromuscular stimulation after spinal cord injury. *Topics in Spinal Cord Injury Rehabilitation, 5*(1), 44-65.

Umberger, B. R., & Martin, P. E. (2001). Testing the planar assumption during ergometer cycling. *Journal of Applied Biomechanics, 17*(1), 55-62.

Usabiaga, J., Crespo, R., Iza, I., Aramendi, J., Terrados, N., & Poza, J. J. (1997). Adaptation of the lumbar spine to different positions in bicycle racing. *Spine, 22*(17), 1965-1969.

Valencia, J. S. O., Cardona, S. G., & Mejía, S. R. (2007). Diseño y construcción de un pedal de bicicleta basado en sensores piezorresistivos para determinar la fuerza resultante. *Revista Ingeniería Biomédica, 1*(2), 55-60.

van den Bogert, A. J., & de Koning, J. J. (1996). *On optimal filtering for inverse dynamics.* Paper presented at the IXth Biennial Conference of the Canadian Society for Biomechanics, Vancouver

Van Ingen Schenau, G. J., Boots, P. J. M., De Groot, G., Snackers, R. J., & Van Woensel, W. W. L. M. (1992). The constrained control of force and position in multi-joint movements. *Neuroscience, 46*(1), 197-207.

Vogt, S., Heinrich, L., Schumacher, Y. O., Blum, A., Roecker, K., Dickhuth, H. H., et al. (2006). Power output during stage racing in professional road cycling. *Medicine and Science in Sports and Exercise, 38*(1), 147-151.

Wanich, T., Hodgkins, C., Columbier, J. A., Muraski, E., & Kennedy, J. G. (2007). Cycling injuries of the lower extremity. *Journal of the American Academy of Orthopaedic Surgeons, 15*(12), 748-756.

Ward, S. R., Terk, M. R., & Powers, C. M. (2005). Influence of patella alta on knee extensor mechanics. *Journal of Biomechanics, 38*(12), 2415-2422.

Watt, K. K. O., Hopkins, W. G., & Snow, R. J. (2002). Reliability of performance in repeated sprint cycling tests. *Journal of Science and Medicine in Sport, 5*(4), 354-361.

Weston, S. B., & Gabbett, T. J. (2001). Reproducibility of ventilation of thresholds in trained cyclists during ramp cycle exercise. *Journal of Science and Medicine in Sport, 4*(3), 357-366.

Williams, A. D., Raj, I. S., Stucas, K. L., Fell, J. W., Dickenson, D., & Gregory, J. R. (2009). Cycling efficiency and performance following short-term training using uncoupled cranks. *International Journal of Sports Physiology and Performance, 4*(1), 18-28.

Winter, D. A. (2005). *Biomechanics and motor control of human movement*. New Jersey: John Wiley & Sons.

Wishv-Roth, T. (2009). Assessment of cycling biomechanics to optimise performance and minimise injury *Journal of Science and Medicine in Sport, 12*(S(1)), S51.

Wolchok, J. C., Hull, M. L., & Howell, S. M. (1998). The effect of intersegmental knee moments on patellofemoral contact mechanics in cycling. *Journal of Biomechanics, 31*(8), 677-683.

Wooles, A. L., Robinson, A. J., & Keen, P. (2006). A static method for obtaining a calibration factor for SRM bicycle power cranks. *Sports Eng, 8*, 137-144.

Zameziati, K., Mornieux, G., Rouffet, D., & Belli, A. (2006). Relationship between the increase of effectiveness indexes and the increase of muscular efficiency with cycling power. *European Journal of Applied Physiology, 96*(3), 274-281.

APPENDIX 1: IMPROVING PERFORMANCE AND PREVENTING INJURIES USING CYCLING BIOMECHANICS: OVERCOMING THE CHALLENGES.

Bini Rodrigo, Croft James, Kilding Andrew, Hume Patria
Institute of Sport and Recreation Research New Zealand (ISRRNZ), AUT University, Auckland, New Zealand.

Background: Bicycles are used for a range of everyday tasks including transportation, rehabilitation, leisure, sport and improving fitness. Research on cycling has been conducted by engineers, kinesiologists, physiologists, and physiotherapists with a focus on the human-bicycle interaction. **Aim of PhD initial studies**: For my PhD we have started a series of studies on cycling biomechanics with a focus on injury prevention and performance: (1) Bike configuration survey - Anthropometric and kinematic measurements; (2) Effects of bike configuration on the knee joint forces and muscle activity of cyclists; (3) Body position effects on knee forces and performance of cyclists and triathletes; (4) Pedaling technique changes with training in elite cyclists and triathletes and effects on performance. **Methods:** The first study is survey-based with a focus on bicycle configuration for cyclists with different levels of expertise. The other three studies are laboratory-based during which several data acquisition systems will operate in synchrony. Specifically, custom-built instrumented pedals will measure the percentage of force produced by the cyclist to move the bicycle. Movement analysis in two planes will be captured by high speed video cameras to determine body position information (kinematics) during cycling. Muscle activity will be determined during cycling using electromyography. **Challenges:** The most challenging aspects of our research on cycling biomechanics are related to equipment given the accuracy of measurement required and the need to develop systems that can be used for data collection on the road rather than in the laboratory. **Overcoming the challenges**: Initial investigations and pilot work in the laboratory aim to develop systems and protocols that will allow accurate measurement of pedaling force, muscle activity and movement patterns during cycling on the road. Such systems will be assessed for validity and reliability in the laboratory and then tested on the road.

APPENDIX 2: EFECTOS DEL ENTRENAMIENTO DE LA TECNICA DE PEDALEO SOBRE LA ECONOMIA Y EL RENDIMENTO EN EL CICLISMO: UNA REVISION A LA LITERATURA.

Bini Rodrigo, Croft James, Kilding Andrew, Hume Patria
Institute of Sport and Recreation Research New Zealand (ISRRNZ), AUT University, Auckland, New Zealand.

Objetivos: To determine the effect of pedalling technique training on cycling economy and performance. **Methods:** Critical literature review of the existing body of knowledge on cycling biomechanics, physiology and performance. **Resultados:** Pedalling technique training, using augmented feedback of pedal forces, has been demonstrated to improve force effectiveness (2), increase activation of the flexor group (i.e. biceps femoris) and increase dorsi flexion of ankle joint moments after one pedalling technique training session (3). However, in contrast to positive gains in force production, the effects of pedalling technique training on economy are conflicting, with most studies reporting reduced economy when pedalling technique training is optimized (4). Decoupled crank systems have been used to improve pedalling technique training, but their impact on cycling performance or economy has yet to be clearly determined (1, 5). **Conclusión**: Pedalling technique training, involving coupled or decoupled crank systems, appears to have a positive effect on improving technique and force effectiveness of cyclists. However, the ability of pedalling technique training to simultaneously enhance economy and performance, regardless of crank system, is questionable.

References

1. Bohm H, Siebert S, and Walsh M. Effects of short-term training using SmartCranks on cycle work distribution and power output during cycling. European Journal of Applied Physiology 103: 225-32, 2008.
2. Broker JP, Gregor RJ, and Schmidt RA. Extrinsic feedback and the learning of cycling kinetic patterns. Journal of Biomechanics 22: 991, 1989.
3. Hasson CJ, Caldwell GE, and van Emmerik REA. Changes in muscle and joint coordination in learning to direct forces. Human Movement Science 27: 590-609, 2008.
4. Korff T, Romer LM, Mayhew I, and Martin JC. Effect of pedaling technique on mechanical effectiveness and efficiency in cyclists. Medicine and Science in Sports and Exercise 39: 991-5, 2007.
5. Luttrell MD, and Potteiger JA. Effects of Short-Term Training Using Powercranks on Cardiovascular Fitness and Cycling Efficiency. Journal of Strength and Conditioning Research 17: 785-91, 2003.

APPENDIX 3: ROAD CYCLISTS OVERUSE INJURY AND CYCLING BODY POSITION SMNZ 2010.

[12]***Bini, R.R.***, [1]*Hume, P.A.*, [1]*Kilding, A.E.*, [1]*Croft, J.L.*
[1]Sport Performance Research Institute New Zealand (SPRINZ), AUT University, Auckland, New Zealand; [2]Capes Foundation, Ministry of Education of Brazil

Background: Incorrect body position on the bicycle of road cyclists may lead to an increased risk of overuse injury (Holmes, et al., 1994). **Aim:** To describe the association between prior overuse injury and current body position on the bicycle in cyclists. **Methods:** A retrospective survey was conducted, along with quantitative biomechanical analyses of body position on the bicycle for 44 road cyclists ranging in ability from recreational to highly competitive (age: 40 ±10 years; height: 180 ±6 cm; weight: 79 ±13 kg). Occurrence of injuries or pain during cycling and the site of injury were recorded. Cyclists had their bicycle mounted on a stationary cycle trainer and then assumed a position similar to that used during normal cycling training. Two static, digital camera sagittal plane images from the right leg, one with the right pedal at bottom dead centre (180° of pedal revolution), the other at right pedal at 90° of pedal revolution) were digitized and knee flexion angle and projection of the knee in relation to the pedal spindle were measured. The numbers of cyclists with knee flexion angle at bottom dead centre between 25-30° of flexion (Holmes, et al., 1994), and with the projection of the knee behind or ahead of the pedal spindle (Silberman, et al., 2005b) were calculated. **Results:** Of the 44 cyclists, 16% had a knee flexion angle within the recommended 25-30° flexion for injury prevention, with 36% presenting a smaller flexion angle and 52% presenting a greater flexion angle. For the majority (73%) of cyclists the knee was forward of the pedal spindle at 90° of pedal revolution. Of the 50% of cyclists who reported injury or pain during cycling, the knee was the most common site of injury (50%). Of the 11 cyclists who reported knee injury or pain nine (82%) had a knee angle different from the expected 25-30° flexion; four had a knee flexion angle <25° and five had a knee flexion angle >30° at bottom dead centre. **Discussion:** The high occurrence of knee joint overuse injury or pain in our sample of cyclists is consistent with previous findings (Dettori & Norvell, 2006; Holmes, et al., 1994). Knee flexion angle can indicate if the seat height has an appropriate configuration (Holmes, et al., 1994). The forward position of the knee in relation to the pedal spindle has been related to a higher risk of knee injuries (Silberman, et al., 2005b). Given the high percentage of injured cyclists who did not meet recommended joint angle ranges, we believe that analysis of bicycle configuration may be useful in reducing the risk of cycling overuse injury for all levels of cyclist. **Conclusion:** Knee flexion angles outside of the recommended 25-30° flexion, and forward position of the knee in relation to the pedal spindle, were both associated with reports of knee pain or overuse injury.

References:

1. Holmes, J.C., A.L. Pruitt, and N.J. Whalen, *Lower extremity overuse in bicycling.* Clinics in Sports Medicine, 1994. **13**(1): 187-203.
2. Silberman, M.R., et al., *Road bicycle fit.* Clinical Journal of Sport Medicine, 2005. **15**(4): 271-276.
3. Dettori, N.J. and D.C. Norvell, *Non-traumatic bicycle injuries: A review of the literature.* Sports Medicine, 2006. 36(1): 7-18.

APPENDIX 4: PAIN FROM OVERUSE INJURY IN 104 COMPETITIVE AND RECREATIONAL ROAD CYCLISTS, MOUNTAIN BIKERS AND TRIATHLETES.

[12]**Bini, R.R.**, [1]*Hume, P.A.*
[1]Sport Performance Research Institute New Zealand (SPRINZ), AUT University, Auckland, New Zealand; [2]Capes Foundation, Ministry of Education of Brazil

Background: Occurrence of overuse injuries and pain in cycling may differ among cyclists of different disciplines [1]. The knee and low back have been reported as the main injury sites in elite cyclists, but data have not been provided for other cycling disciplines and experience levels. **Aim**: To describe the occurrence of pain in cyclists of different disciplines and experience level. **Methods:** A survey of 104 cyclists (49 recreational non-competitive road cyclists, 21 competitive road cyclists, 13 non-competitive mountain bike riders and 21 competitive triathletes) provided information on characteristics of existing pain during cycling, perceived comfort on the bicycle (yes or no) and cycling training volume (hrs/week). **Results and discussion**: Percentage values in Table I are presented within each cycling discipline and for all cyclists. For all cyclists surveyed, 38% had existing pain even though 85% reported they felt comfortable on the bicycle. Fifty six percent reported both pain and comfort, therefore, comfort on the bicycle is probably not a good indicator for overuse injury risk. The knee joint and low back were frequently reported sites of pain for all the cycling disciplines and levels which supports findings from other studies on road cyclists [1]. A larger percentage of triathletes reported pain compared to the other cycling disciplines. The larger percentage of ankle-foot pain in triathletes during cycling may be a cross-over effect due to additional running training. **Conclusion:** Triathletes report more ankle-foot overuse pain than other cycling disciplines. Comfort on the bicycle and cycling training hours per week are probably not good indicators of overuse injury risk. Training intensity and total training volume of triathletes (including running and swimming) should be measured to gain further understanding of risk factors for overuse injuries

Table I. Cyclists' age and training volume, comfort during cycling, percentage of cyclists with pain, and body sites of pain for recreational cyclists, mountain bikers, competitive cyclists and triathletes.

Cyclist discipline	Age (years)	Training volume (hrs/wk)	Comfort (%)	Pain (%)	Upper body (%)	Low back (%)	Butt - hips (%)	Knee (%)	Ankle - foot (%)
Competitive cyclists (n = 21)	37 ±11	8 ±4	86	33	29	43	43	50	7
Recreational cyclists (n = 49)	41 ±9	7 ±3	85	41	52	48	43	87	9
Mountain bikers (n = 13)	39 ±10	5 ±3	85	38	25	88	0	75	0
Triathletes (n = 21)	38 ±10	6 ±2	79	58	13	25	25	75	38
Total (N = 104)	40 ±10	7 ±3	85	38	30	41	28	61	9

References

1. Dettori, N.J. and D.C. Norvell, Non-traumatic bicycle injuries: A review of the literature. Sports Medicine, 2006. 36(1): p. 7-18.

APPENDIX 5: MOVING FORWARD OR BACKWARD ON THE BICYCLE SADDLE DOES NOT CHANGE PEDAL FORCE EFFECTIVENESS.

[12]**Bini, *R.R***., [1]Hume, *P.A.*
[1]Sport Performance Research Institute New Zealand (SPRINZ), AUT University, Auckland, New Zealand; [2]Capes Foundation, Ministry of Education of Brazil

Background: Cyclists change their position on a saddle in races in an attempt to improve performance. The effects of these changes on pedal forces are unknown. **Aim**: To compare forward and backward saddle position for cyclists on pedal forces. **Methods:** Twelve competitive cyclists (28 ±6 years) performed an incremental cycling test to exhaustion to determine maximal power output and ventilatory threshold. At a second session they performed one 2-minute maximal trial then three 2-minute sub-maximal trials (load of the ventilatory threshold) at different positions on the saddle (preferred, most forward, most backward). Body position on the saddle was analyzed by the position of one reflective marker attached to the cyclists' sacrum in relation to the bicycle frame. All cycling tests were performed using the cyclists' bicycle on a cycle trainer. Force on the right pedal was measured using a strain gauge instrumented pedal and right sagittal plane kinematics were acquired using a high speed camera. Pedal force effectiveness was computed as the ratio between the tangential and total force applied on the pedal and was normalized by maximal power output. Magnitudes of changes between the preferred, most forward and most backward positions on the saddle were assessed by effect sizes for force effectiveness and body position on the saddle. **Results and discussion**: Moving forward on the saddle would be expected to improve pedal force application due to hypothetically greater contributions from knee joint extensor muscles with opposite effects for a backward saddle position. However, in our study, no substantial changes in force effectiveness were observed with forward or backward body position changes on the saddle. Price and Donne [1] reported no clear effects on pedalling technique even though position on the saddle was changed by varying the saddle-tube angle by 12° which did reduce hip range of motion. Ricard et al., [2] reported smaller activation of biceps femoris for 12 cyclists when saddle-tube angle was increased by 10°. Changes in lower limb joint kinematics may balance the effects of moving forward or backward on the saddle resulting in similar muscle force production.

Table 1. Means and standard deviations for body position on the saddle and pedal force effectiveness normalized by maximal power output at forward, preferred and backward saddle positions with differences expressed as effect sizes.

Body position			Force effectiveness		
Forward	**Preferred**	**Backward**	**Forward**	**Preferred**	**Backward**
83 ±9	100 ±3	113 ±7	101 ±12	97 ±10	101 ±9
← 7%; large →	← 13%; large →		← 4%; small →	← 4%; small →	
← 20%; large →			← <1%; trivial →		

Conclusion

Moving forward or backward on the saddle did not affect cyclists' pedal force effectiveness

References

1. Price, D. and B. Donne, Effect of variation in seat tube angle at different seat heights on submaximal cycling performance in man. Journal of Sports Sciences, 1997. 15(4): p. 395-402.
2. Ricard, M.D., et al., The effects of bicycle frame geometry on muscle activation and power during a Wingate anaerobic test. Journal of Sports Science and Medicine, 2006. 5(1): p. 25-32.

APPENDIX 6: BIKE SURVEY QUESTIONNARIES

Bicycle Configuration Survey

The aim of the proposed research is to report how cyclists configure their bicycle components and the methods they choose for bicycle configuration.

Project supervisor: **Dr. James Croft**

Researcher: **Rodrigo Bini**

Cyclists name: ____

Date: ____ / _____ / ____ Age: _________ Gender: ___________

Cyclist's contact details:

Address: ____

Phone: ____________________________ Email: ______________________

Questions about bicycle components and bicycle configuration.

Most of the questions only need one answer, but in questions with multiple choices (i.e. A or B etc) indicate with a tick all relevant information. Add any comments that you think are important in the spaces provided. For example, in Table 1 put a tick in each table cell that applies. Note that MTB means mountain bicycle.

1- What do you mostly use your bicycle for?

Bike use	MTB for Women	MTB for Men	Road bike	Triathlon bike	Hybrid
Leisure					
Locomotion					
Training					

Other ( ) please specify: ____

2- How many hours per week do you ride each bicycle?

	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
Hours of use					

Comment: ____________________

3- How many years have you been riding each bicycle?

	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
Years of use					

Comment: ____________________

4- Rank which kind of bicycle you mostly ride? Use the following scale:

0=never, 1=rarely, 2=occasionally, 3=sometimes, 4=normally, 5=most often.

	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
Rank of bicycle use					

Comment: ____________________

5- State the brand of each bicycle you have.

	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
Brands					

Comment: ____________________

6- What kind of saddle do you have for each of your bicycles?

Saddle	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
Conventional					
With relief/cutaway area					
Noose-less					
With forward/backward settings					
No forward/backward settings					
Foam					
Gel					

Comment: ____________________

7- What kind of handlebars do you have for each of your bicycles?

Handlebars	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
MTB					
Road cycling					
Triathlon					

Comment: ____

8- What kind of pedal do you have for each of your bicycles? If you use clips please indicate the brand you use.

Pedal	MTB for Women	MTB for Men	Road bicycle	Triathlon bicycle	Hybrid
Clip less					
Toe clips					
No clip					

Comment: ____

9- Do you know how to set the position of the seat on your bicycle?
( ) Yes ( ) No

10- Do you know how to set the position of the handlebars on your bicycle?
( ) Yes ( ) No

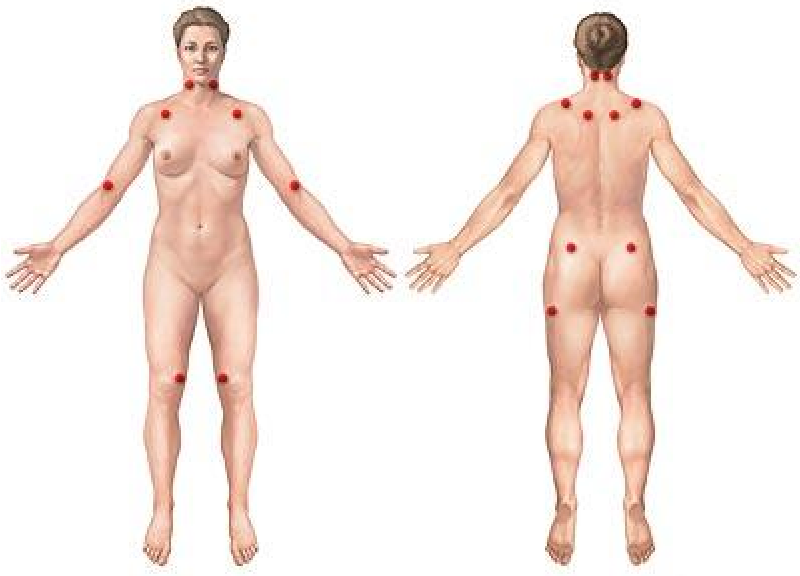
11- Have you been evaluated in relation to bike positioning?
( ) Yes ( ) No
By whom? ____

Questions about your injury history.

1- Do you currently have any kind of <u>injury</u> or disease that may affect your bicycle riding?
( ) Yes ( ) No
What kind? ____

How long have you been feeling pain? ________________________________

2- Pain while cycling.
Please indicate previous pain with a "O" and current pain with a "X".



3- Do you feel comfortable when you ride your bicycle? (   ) Yes   (   ) No
If not, please describe what feels uncomfortable.

___

___

4- Do you participate in other sports? If yes, please describe your training frequency.

___

___

Cyclist signature: ....
Cyclist's contact details:
……………………………………………………………………………………..
……………………………………………………………………………………..

Project Supervisor Contact Details:
Dr. James Croft
Institute of Sport & Recreation Research New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006
Auckland 1020
Ph 921 9999 ext. 7685
james.croft@aut.ac.nz

Approved by the Auckland University of Technology Ethics Committee on 24 August 2009.
AUTEC reference number 09/178

ADDITIONAL QUESTIONAIRY OF TRAINING AND INJURY HISTORY

Date Information Sheet Produced: 30th April 2009

Date: ___/___/______

Subject name:

Training and bicycling history questionnaire

1- What is your main purpose of riding your bicycle (i.e. recreation, training, locomotion)?____

2a- How many days in a week do you usually ride? _________

2b- How many hours and/or distance do you usually ride in each day? _________

3- What kind of bicycle do you mostly use to ride (i.e. road bike, mountain bike, other)?____

4- Do you currently have any kind of injury or disease that may affect your riding? What kind?

____

5- Have previously had joint or muscle pain during bicycle exercise? If yes, please indicate if you still feel pain during bicycle exercise.

____

7- Do you feel comfortable when you ride your bike? If not, please describe what feels uncomfortable.

____

8- Do you participate in other sports? If yes, please describe your training frequency.

____

Approved by the Auckland University of Technology Ethics Committee on 16 July 2009. AUTEC reference number 09/119

APPENDIX 7: SUBJECT INFORMATION PACKS

Participant Information Sheet

DATE INFORMATION SHEET PRODUCED:

27th July 2009.

PROJECT TITLE

Bicycle configuration survey: Anthropometric and kinematic measurements.

AN INVITATION

Hi, my name is Rodrigo Bini and I am a PhD student at AUT University. Along with my supervisors Dr. James Croft, Assoc. Prof. Andrew Kilding and Prof. Patria Hume (all from AUT University) I am inviting you to help with a project that looks at body position on the bicycle. The project will involve a survey regarding your bicycle's components and how you set them up.

You should decide whether or not you would like to be involved. You don't have to be involved, and you can stop being involved in the project at any time.

WHAT IS THE PURPOSE OF THIS RESEARCH?

The aim of the proposed research is to report how cyclists set their bicycle components and the methods they choose for bicycle configuration.

HOW WAS I CHOSEN FOR THIS INVITATION?

Cyclists with different level of cycling expertise were chosen for this invitation.

WHAT WILL HAPPEN IN THIS RESEARCH?

You will receive a survey to fill in regarding your cycling expertise, characteristics of your bicycle's components and how you set up your bicycle components. Photographs on static poses on the bicycle will be used to measure angles of trunk, hip, knee and ankle.

Using the collected information we will determine the most common method for setting bicycle components and if it is related to cycling experience. We also will determine the differences between your position on the bicycle and existing guidelines.

WHAT ARE THE DISCOMFORTS AND RISKS?

You will not suffer any discomfort or risk during this project.

WHAT ARE THE BENEFITS?

You may choose to receive recommendations on optimal methods for configuration of your bicycle components.

WHAT COMPENSATION IS AVAILABLE FOR INJURY OR NEGLIGENCE?

There is no risk of developing any injury during this project.

HOW WILL MY PRIVACY BE PROTECTED?

All information related to you will be coded in order to ensure that you cannot be identified. The information will remain in locked storage and will only be accessible to the researchers of the bicycle project. Photographs will be used only to measure joint angles and bicycle components and included on the individual report offered to you. They will be coded after digitized and storage by the principal responsible of this project.

WHAT ARE THE COSTS OF PARTICIPATING IN THIS RESEARCH?

The only cost to you is that of time. The duration of the data collection session will be about five minutes for completing questionnaires, plus 10 minutes for anthropometric body measurements and 10 minutes for bicycle configuration image capture. Therefore you will need to provide 25 minutes for the project data collection.

WHAT OPPORTUNITY DO I HAVE TO CONSIDER THIS INVITATION?

- You may take the time you need and decide whether or not you would like to be involved in the project.
- You can stop being involved in the project at any point.
- Your involvement in this project is voluntary.

HOW DO I AGREE TO PARTICIPATE IN THIS RESEARCH?

If you agree to participate please fill in the attached consent form and return to Rodrigo Bini at the Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7685, rbini@aut.ac.nz.

WILL I RECEIVE FEEDBACK ON THE RESULTS OF THIS RESEARCH?

Yes, a summary report will be provided to you if you request it.

WHAT DO I DO IF I HAVE CONCERNS ABOUT THIS RESEARCH?

Any concerns regarding the nature of this project should be notified in the first instance to the project primary supervisor:

Dr. James Croft, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7685, james.croft@aut.ac.nz
Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEC, Madeline Banda, *madeline.banda@aut.ac.nz*, Ph 921 9999 ext 8044.

WHOM DO I CONTACT FOR FURTHER INFORMATION ABOUT THIS RESEARCH?

Researcher Contact Details:
Rodrigo Rico Bini, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Auckland 0637, Ph 921 9999 ext. 7295, rbini@aut.ac.nz

Project Primary Supervisor Contact Details:
Dr. James Croft, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7685, james.croft@aut.ac.nz

Approved by the Auckland University of Technology Ethics Committee on 24th August 2009.
AUTEC Reference number 09/178.

Participant Information Sheet

DATE INFORMATION SHEET PRODUCED:

1st July 2009.

PROJECT TITLE

Effects of bike configuration on the knee joint forces and muscle activity of cyclists: Implications for injury prevention and rehabilitation.

AN INVITATION

Hi, my name is Rodrigo Bini and I am a PhD student at AUT University. Along with my supervisors Dr. James Croft, Assoc. Prof. Andrew Kilding and Prof. Patria Hume I am inviting you to help with a project that looks at saddle position and its effects on knee joint load. You should decide whether or not you would like to be involved. You don't have to be involved, and you can stop being involved in the study at any time.

WHAT IS THE PURPOSE OF THIS RESEARCH?

The purpose of the research is to compare knee joint forces and muscle activity during pedalling with different body positions, in an attempt to achieve a better understanding of the bike configuration and its effects on joint overload.

HOW WAS I CHOSEN FOR THIS INVITATION?

Injured and uninjured recreational cyclists are being chosen for this invitation.

WHAT WILL HAPPEN IN THIS RESEARCH?

You will come to the University lab at the Millennium Institute for Sport and Health in Albany four times on different days (day 1, day 3, day 14 and Day 24). On the first day the testing will take two hours. On the subsequent days testing will take 70 minutes. On the first day we will measure your maximal aerobic capability and your anthropometry (body dimensions). On the three following sessions we will measure the force you apply to the pedal while you cycle and your movements using captured video. Data will be analyzed to calculate force on the knee. Results will be used in scientific journal articles and conference presentations. Media articles will be written based on the results of the present study.

WHAT ARE THE DISCOMFORTS AND RISKS?

If you often feel pain during cycling, this may also occur during the maximal aerobic test. However, if you have any discomfort you will be able to stop the test. You will also experience temporary exertion after the incremental test used to measure your maximal aerobic capability. You will be able to interrupt the maximal aerobic test whenever you feel uncomfortable.

HOW WILL THESE DISCOMFORTS AND RISKS BE ALLEVIATED?

Ice will be available for you to put on your knees if you feel pain. Water and sugar will be offered if you feel dizzy from over-exertion.

WHAT ARE THE BENEFITS?

You will benefit from this study by understanding bicycle set-up for your body type. This should enable you to cycle with a reduced risk of knee overload.

WHAT COMPENSATION IS AVAILABLE FOR INJURY OR NEGLIGENCE?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

HOW WILL MY PRIVACY BE PROTECTED?

All information related to you will be coded in order to ensure that you cannot be identified. The information will remain in locked storage and will only be accessible to the people of the bike project. No-one will be able to identify you from any of the summary findings for the report of the project.

WHAT ARE THE COSTS OF PARTICIPATING IN THIS RESEARCH?

The only cost to you is that of time. On the first day, the consent form should not take more than five minutes, the anthropometric evaluation takes 30 minutes, 15 minutes for the maximal incremental test, 30 minutes of rest and 30 minutes for familiarization test. The second evaluation session (day 3) will be composed of 15 minutes for pedal force, kinematics and EMG preparation, 10 minutes of warm-up, 15 minutes of steady state cycling, 15 minutes for bike fit, retest of 15 minutes steady state cycling (70 minutes total). The third and fourth evaluation sessions (day 14 and day 24 respectively), will take the same time as the second evaluation session (70 minutes). We will provide you with a $20 petrol voucher to help cover costs of transport to the testing sessions.

WHAT OPPORTUNITY DO I HAVE TO CONSIDER THIS INVITATION?

- You may take the time you need and decide whether or not you would like to be involved
- You can stop being involved in the project at any point.

HOW DO I AGREE TO PARTICIPATE IN THIS RESEARCH?

If you agree to participate please fill in the attached consent form and return to myself.

WILL I RECEIVE FEEDBACK ON THE RESULTS OF THIS RESEARCH?

Yes, feedback will be provided to you, if you request it,

WHAT DO I DO IF I HAVE CONCERNS ABOUT THIS RESEARCH?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor:

Dr. James Croft, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7685, james.croft@aut.ac.nz
Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEC, Madeline Banda, *madeline.banda@aut.ac.nz*, Ph 921 9999 ext 8044.

WHOM DO I CONTACT FOR FURTHER INFORMATION ABOUT THIS RESEARCH?

Researcher Contact Details:
Rodrigo Rico Bini, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Auckland 0637, Ph 921 9999 ext. 7295, rbini@aut.ac.nz

Project Supervisor Contact Details:
Dr. James Croft, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7685, james.croft@aut.ac.nz

Approved by the Auckland University of Technology Ethics Committee on 16th July 2009 approval was granted, AUTEC Reference number 09/119.

Participant Information Sheet

DATE INFORMATION SHEET PRODUCED:

25th September 2009.

PROJECT TITLE

Muscle mechanics and performance of cyclists and triathletes: Kinematics, pedal forces and ultrasound evaluation.

AN INVITATION

Hi, my name is Rodrigo Bini and I am a PhD student at AUT University in New Zealand. Along with my supervisors Dr. James Croft, Assoc. Prof. Andrew Kilding, Prof. Patria Hume and Assoc. Prof. Marco Vaz, I am inviting you to help with a project that looks at saddle position and its effects on knee joint load.

You should decide whether or not you would like to be involved. You don't have to be involved, and you can stop being involved in the study at any time.

WHAT IS THE PURPOSE OF THIS RESEARCH?

The purpose of the research is to compare pedal force, muscle activity, and lower limb dynamics with different body positions, in an attempt to achieve a better understanding of the bike configuration and its effects on performance.

HOW WAS I CHOSEN FOR THIS INVITATION?

You have been identified as being a cyclists or triathlete with regular training and competitive experience.

WHAT WILL HAPPEN IN THIS RESEARCH?

You will come to the University lab two times on different days (day 1 and day 3). On day 1, you will complete questionnaires about your cycling and training history, and we will measure your body composition using scales and skin fold callipers and your body dimensions using tape measures and rulers. You will also perform a maximal incremental cycle test which is performed on an indoor ergometer. The exercise intensity starts very easy and it progressively gets harder until you can no longer continue. During the test we will monitor your heart rate and oxygen consumption. To do this you will wear a heart rate monitor around your chest and a sealed face mask (or mouthpiece) so we can measure the volume of oxygen you use when cycling. The second evaluation session on day 3 will be composed of videoing you during steady state cycling. We will determine pedal forces and joint angles whilst you pedal with a variety of saddle positions. We will also capture images of your muscles using an ultrasound machine which requires a propbe to be placed on the surface of your skin whilst you exercise. In addition, we will attach some EMG wires to your legs to determine which muscles are contracting. Both ultrasound and EMG are completely painless.

Data will be analyzed to calculate pedalling technique, muscle activity and lower limb dynamics. Results will be used in scientific journal articles and conference presentations. Media articles will be written based on the results of the present study.

WHAT ARE THE DISCOMFORTS AND RISKS?

You will also experience temporary exhaustion after the incremental test used to measure your maximal aerobic capability. You will be able to interrupt the maximal aerobic test whenever you feel uncomfortable.

HOW WILL THESE DISCOMFORTS AND RISKS BE ALLEVIATED?

Water and sugar will be offered if you feel dizzy from over-exertion.

WHAT ARE THE BENEFITS?

You will benefit from this study by understanding bicycle set-up for your body type. This should enable you to improve your cycling performance.

WHAT COMPENSATION IS AVAILABLE FOR INJURY OR NEGLIGENCE?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by Sistema Unico de Saúde (SUS – Brazil).

HOW WILL MY PRIVACY BE PROTECTED?

All information related to you will be coded in order to ensure that you cannot be identified. The information will remain in locked storage and will only be accessible to the people of the bike project. No-one will be able to identify you from any of the summary findings for the report of the project.

WHAT ARE THE COSTS OF PARTICIPATING IN THIS RESEARCH?

The only cost to you is that of time. On the first day, the consent form should not take more than five minutes, the anthropometric evaluation takes 30 minutes, 15 minutes for the maximal incremental test, 30 minutes of rest and 30 minutes for familiarization test. The second evaluation session (day 3) will be composed of 15 minutes for pedal force, kinematics, ultrasound and EMG preparation, 10 minutes of warm-up, 10 minutes of steady state cycling in five different configurations of the saddle (60 minutes total). We will provide you with a $20 petrol voucher to help cover costs of transport to the testing sessions.

WHAT OPPORTUNITY DO I HAVE TO CONSIDER THIS INVITATION?

- You may take the time you need and decide whether or not you would like to be involved
- You can stop being involved in the project at any point.

HOW DO I AGREE TO PARTICIPATE IN THIS RESEARCH?

If you agree to participate please fill in the attached consent form and return to myself.

WILL I RECEIVE FEEDBACK ON THE RESULTS OF THIS RESEARCH?

Yes, feedback will be provided to you, if you request it,

WHAT DO I DO IF I HAVE CONCERNS ABOUT THIS RESEARCH?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor:

Prof. Marco Aurélio Vaz, Laboratório de Pesquisa do Exercício, Escola de Educação Física, Universidade Federal do Rio Grande do Sul, Rua Felizardo 750, Ph 3308 5860, marco.vaz@esef.ufrgs.br
Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEC, Madeline Banda, *madeline.banda@aut.ac.nz*, Ph 921 9999 ext 8044.

WHOM DO I CONTACT FOR FURTHER INFORMATION ABOUT THIS RESEARCH?

Researcher Contact Details:
Rodrigo Rico Bini, Laboratório de Pesquisa do Exercício, Escola de Educação Física, Universidade Federal do Rio Grande do Sul, Rua Felizardo 750, Ph 3308 5859, rbini@aut.ac.nz

Project Supervisor Contact Details in New Zealand:
Dr. James Croft, Institute of Sport & Recreation Research New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7685, james.croft@aut.ac.nz

Project Supervisor Contact Details in Brazil:

Dr. Marco Aurélio Vaz, Associate Professor, Exercise Research Laboratory, School of Physical Education, Federal University of Rio Grande do Sul, Rua Felizardo, 750, Bairro Jardim Botânico, Porto Alegre, RS, Brazil, 90690-200, Phone: 55-51-33085860, Fax: 55-51-33085858, E-mail: marcovaz@esef.ufrgs.br

Approved by the Auckland University of Technology Ethics Committee on 3rd November 2009 approval was granted, AUTEC Reference number 09/228.

Participant Information Sheet

DATE INFORMATION SHEET PRODUCED:
12^{th} March 2010

PROJECT TITLE
Technique training effects on performance of cyclists.

AN INVITATION
Hi, my name is Rodrigo Bini and I am a PhD student at AUT University. Along with my supervisors Prof. Patria Hume, Assoc. Prof. Andrew Kilding, Dr. James Croft, and my PhD sport advisors Craig Palmer and Joe McQuillan, who are providing supervision of the project and guidance for the design of the training intervention and analyses of data, I am inviting you to help with a project that looks at the effects of technique training on performance of cyclists and triathletes.
You should decide whether or not you would like to be involved. You do not have to be involved, and you can stop being involved in the study at any time.

WHAT IS THE PURPOSE OF THIS RESEARCH?
The purpose of the research is to compare two types of training intervention feedback for cycling performance. You will be in either the cycling economy feedback group or the force feedback group.

HOW WAS I CHOSEN FOR THIS INVITATION?
You have been identified as being a cyclist with regular training and competitive experience.

WHAT WILL HAPPEN IN THIS RESEARCH?
You will come to the AUT University laboratory at the AUT Millennium Institute of Sport and Health campus for a first session. In this session, you will complete questionnaires about your cycling and training history, and we will measure your body dimensions using anthropometry equipment (similar to rulers and measuring tapes). You will also perform a maximal incremental cycle test on an indoor cycle ergometer. The exercise intensity starts very easy and progressively gets harder until you can no longer continue. During the test we will monitor your heart rate with a heart rate monitor you will wear around your chest. You will be allocated (randomly) to either the economy feedback group or the force feedback group. The session will be repeated after seven days.

From the third to the eighteenth session (eight weeks) you will be trained on an indoor ergometer receiving either economy feedback from oxygen uptake measures or force feedback from visual force profiles of your pedal forces. Workload and cadence will be progressively increased over the training period.

Forty-eight hours after the last training session you will return to the laboratory for a second incremental test. After two weeks, another session of incremental testing will be performed. During the study we will record your reported amount of cycling training (hours and average intensity) you do outside of the laboratory testing.

Data will be used to calculate changes in pedalling technique and the peak power output. Results will be used in scientific journal articles and conference presentations. Media articles will be written based on the results of the present study. You will not be individually identified in any of the articles or presentations.

WHAT ARE THE DISCOMFORTS AND RISKS?

You will experience temporary exhaustion after the incremental tests which measure your maximal aerobic capability. These are of similar intensity to a hard cycle race. You will be able to stop the maximal aerobic test whenever you feel uncomfortable.

HOW WILL THESE DISCOMFORTS AND RISKS BE ALLEVIATED?

A sweet drink or water will be offered if you feel dizzy from over-exertion.

WHAT ARE THE BENEFITS?

You will benefit from this study by improving your power output and pedalling technique and you may improve your cycling performance.

WHAT COMPENSATION IS AVAILABLE FOR INJURY OR NEGLIGENCE?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

HOW WILL MY PRIVACY BE PROTECTED?

All information related to you will be coded in order to ensure that you cannot be identified. The information will remain in locked storage and will only be accessible to the researchers involved in this project. No-one will be able to identify you from any of the summary findings that are reported.

WHAT ARE THE COSTS OF PARTICIPATING IN THIS RESEARCH?

The only cost to you is that of time. The duration of the incremental test sessions will probably last about one hour (10 minutes for filling in questionnaires, 30 minutes for anthropometric protocols and 15 minutes for the maximal incremental test). The training sessions will be composed of 10 minutes of warm-up and 30 minutes of training (40 minutes total). We will provide you with a $20 petrol voucher to help cover costs of transport to the testing sessions.

WHAT OPPORTUNITY DO I HAVE TO CONSIDER THIS INVITATION?

- You may take the time you need and decide whether you would like to be involved.
- You can stop being involved in the project at any point.

HOW DO I AGREE TO PARTICIPATE IN THIS RESEARCH?

If you agree to participate please fill in the attached consent form and return to myself.

WILL I RECEIVE FEEDBACK ON THE RESULTS OF THIS RESEARCH?

Yes, feedback will be provided to you, if you request it.

WHAT DO I DO IF I HAVE CONCERNS ABOUT THIS RESEARCH?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor:

Professor Patria Hume, Sport Performance Research Institute New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7306, patria.hume@aut.ac.nz

Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEC, Madeline Banda, *madeline.banda@aut.ac.nz*, Ph 921 9999 ext 8044.

WHOM DO I CONTACT FOR FURTHER INFORMATION ABOUT THIS RESEARCH?

Researcher Contact Details:
Rodrigo Rico Bini, Sports Performance Research Institute, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 477 2052, rbini@aut.ac.nz

Project Supervisor Contact Details:
Professor Patria Hume, Sport Performance Research Institute New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Ph 921 9999 ext. 7306, patria.hume@aut.ac.nz

Approved by the Auckland University of Technology Ethics Committee on 14th May 2010, AUTEC Reference number 10/56.

APPENDIX 8: SUBJECT CONSENT FORMS

Consent to Participate in Research

Title of project: **"Bicycle configuration survey: Anthropometric and kinematic measurements."**

Project supervisor: **Dr. James Croft**

Researcher: **Rodrigo Bini**

- I have read and understood the information provided about this research project (Information Sheet dated 27th July 2009). Yes/No
- I have had an opportunity to ask questions and to have them answered. Yes/No
- I am not suffering from any injury or illness which may impair my physical performance Yes/No
- I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way. Yes/No
- If I withdraw, I understand that all relevant information will be destroyed. Yes/No
- I understand that the camera images collected will be used for academic/feedback purposes only and will not be published in any form outside of this project without my written permission.
- I agree to take part in this research. Yes/No
- I wish to receive a copy of the report from the research: tick one: Yes O No O

Participant signature: ....
Participant name: ……………………………………………………………
Date: ……………………………………………………………
Participant's contact details:

Project Supervisor Contact Details:
Dr. James Croft
Institute of Sport & Recreation Research New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006
Auckland 1020
Ph 921 9999 ext. 7685
james.croft@aut.ac.nz

Approved by the Auckland University of Technology Ethics Committee on 24th August 2009. AUTEC reference number 09/178

Consent to Participation in Research

Title of Project: **"Effects of bike configuration on the knee joint forces and muscle activity of cyclists: Implications for injury prevention and rehabilitation."**

Project Supervisor: **Dr. James Croft**

Researcher: **Rodrigo Bini**

- I have read and understood the information provided about this research project (Information Sheet dated 1st July 2009). Yes/No
- I have had an opportunity to ask questions and to have them answered. Yes/No
- I am not suffering from any injury or illness which may impair my physical performance Yes/No
- I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way. Yes/No
- If I withdraw, I understand that all relevant information will be destroyed. Yes/No
- I understand that the video collected will be used for academic/feedback purposes only and will not be published in any form outside of this project without my written permission.
- I agree to take part in this research. Yes/No
- I agree to my general practitioner providing medical details to the researcher.

  Yes/No
- I wish to receive a copy of the report from the research: tick one: Yes O No O

Participant signature: ....
Participant name: ....
Date: ....
Participant's Contact Details:

Project Supervisor Contact Details:
Dr. James Croft
Institute of Sport & Recreation Research New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006
Auckland 1020
Ph 921 9999 ext. 7685
james.croft@aut.ac.nz

Approved by the Auckland University of Technology Ethics Committee on 16th July 2009. AUTEC reference number 09/119

Consent to Participation in Research

Title of Project: **"Muscle mechanics and performance of cyclists and triathletes: Kinematics, pedal forces and ultrasound evaluation."**

Project Supervisors: **Dr. James Croft, Dr. Marco Aurélio Vaz**

Researcher: **Rodrigo Bini**

- I have read and understood the information provided about this research project (Information Sheet dated 1st July 2009). Yes/No
- I have had an opportunity to ask questions and to have them answered. Yes/No
- I am not suffering from any injury or illness which may impair my physical performance Yes/No
- I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way. Yes/No
- If I withdraw, I understand that all relevant information will be destroyed. Yes/No
- I understand that the video collected will be used for academic/feedback purposes only and will not be published in any form outside of this project without my written permission.
- I give permission for my results to be shared with my coach Yes/No
- I agree to take part in this research. Yes/No
- I wish to receive a copy of the report from the research: tick one: Yes O No O

Participant signature: ....
Participant name: ....
Date:..
Participant's Contact Details:

Project Supervisor Contact Details in New Zealand:
Dr. James Croft
Institute of Sport & Recreation Research New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006
Auckland 1020
Ph 921 9999 ext. 7685
james.croft@aut.ac.nz

Project Supervisor Contact Details in Brazil:
Dr. Marco Aurélio Vaz
Associate Professor
Exercise Research Laboratory
School of Physical Education
Federal University of Rio Grande do Sul
Rua Felizardo, 750
Bairro Jardim Botânico
Porto Alegre, RS, Brazil, 90690-200
Phone: 55-51-33085860
Fax: 55-51-33085858
E-mail: marcovaz@esef.ufrgs.br

***Approved by the Auckland University of Technology Ethics Committee on 3rd November 2009.* AUTEC Reference number 09/228.**

Consent to Participation in Research

Title of Project: **Technique training effects on performance of cyclists.**

Project Supervisors: Prof. Patria Hume, Assoc. Prof. Andrew Kilding, Dr. James Croft, Mr. Craig Palmer, Mr. Joe McQuillan

Researcher: **Rodrigo Bini**

* I have read and understood the information provided about this research project (Information Sheet dated 12th March 2010). Yes/No

* I have had an opportunity to ask questions and to have them answered. Yes/No

* I am not suffering from any injury or illness which may impair my physical performance Yes/No

* I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way. Yes/No

* If I withdraw, I understand that all relevant information will be destroyed. Yes/No

* I give permission for my results to be shared with my coach Yes/No

* I agree to take part in this research. Yes/No

* I wish to receive a copy of the report from the research: tick one: Yes O No O

Participant signature: ....
Participant name: ....
Date: ....
Participant's Contact Details:
....
....

Project Supervisor Contact Details:
Professor Patria Hume
Institute of Sport & Recreation Research New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006
Auckland 1020
Ph 921 9999 ext. 7306
patria.hume@aut.ac.nz

***Approved by the Auckland University of Technology Ethics Committe* on *14th May 2010*, AUTEC Reference number** 10/56*.*

APPENDIX 9: ETHICS APPROVAL

MEMORANDUM

Auckland University of Technology Ethics Committee (AUTEC)

To: James Croft
From: **Madeline Banda** Executive Secretary, AUTEC
Date: 24 August 2009
Subject: Ethics Application Number 09/178 **Bicycle configuration survey: Anthropometric and kinematic measurements.**

Dear James

Thank you for providing written evidence as requested. I am pleased to advise that it satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC) at their meeting on 10 August 2009 and that on 21 August 2009, I approved your ethics application. This delegated approval is made in accordance with section 5.3.2.3 of AUTEC's *Applying for Ethics Approval: Guidelines and Procedures* and is subject to endorsement at AUTEC's meeting on 14 September 2009.

Your ethics application is approved for a period of three years until 21 August 2012.

I advise that as part of the ethics approval process, you are required to submit the following to AUTEC:

- A brief annual progress report using form EA2, which is available online through *http://www.aut.ac.nz/research/research-ethics*. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 21 August 2012;

- A brief report on the status of the project using form EA3, which is available online through *http://www.aut.ac.nz/research/research-ethics*. This report is to be submitted either when the approval expires on 21 August 2012 or on completion of the project, whichever comes sooner;

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are reminded that, as applicant, you are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

Please note that AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to make the arrangements necessary to obtain this. Also, if your research is undertaken within a jurisdiction outside New Zealand, you will need to make the arrangements necessary to meet the legal and ethical requirements that apply within that jurisdiction.

When communicating with us about this application, we ask that you use the application number and study title to enable us to provide you with prompt service. Should you have any further enquiries regarding this matter, you are welcome to contact Charles Grinter, Ethics Coordinator, by email at ethics@aut.ac.nz or by telephone on 921 9999 at extension 8860.

On behalf of the AUTEC and myself, I wish you success with your research and look forward to reading about it in your reports.

Yours sincerely

Madeline Banda
Executive Secretary
Auckland University of Technology Ethics Committee
Cc: Rodgigo Rico Bini rbini@aut.ac.nz, Andrew Kilding, Patria Hume

AUT UNIVERSITY

MEMORANDUM

Auckland University of Technology Ethics Committee (AUTEC)

To: James Croft
From: **Madeline Banda** Executive Secretary, AUTEC
Date: 16 July 2009
Subject: Ethics Application Number 09/119 **Effects of bike configuration on the knee joint forces and muscle activity of cyclists: Implications for injury prevention and rehabilitation.**

Dear James

Thank you for providing written evidence as requested. I am pleased to advise that it satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC) at their meeting on 15 June 2009 and that I have approved your ethics application. This delegated approval is made in accordance with section 5.3.2.3 of AUTEC's *Applying for Ethics Approval: Guidelines and Procedures* and is subject to endorsement at AUTEC's meeting on 10 August 2009.

Your ethics application is approved for a period of three years until 16 July 2012.

I advise that as part of the ethics approval process, you are required to submit the following to AUTEC:

- A brief annual progress report using form EA2, which is available online through http://www.aut.ac.nz/about/ethics. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 16 July 2012;
- A brief report on the status of the project using form EA3, which is available online through http://www.aut.ac.nz/about/ethics. This report is to be submitted either when the approval expires on 16 July 2012 or on completion of the project, whichever comes sooner;

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are reminded that, as applicant, you are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

Please note that AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to make the arrangements necessary to obtain this.

When communicating with us about this application, we ask that you use the application number and study title to enable us to provide you with prompt service. Should you have any further enquiries regarding this matter, you are welcome to contact Charles Grinter, Ethics Coordinator, by email at charles.grinter@aut.ac.nz or by telephone on 921 9999 at extension 8860.

On behalf of the AUTEC and myself, I wish you success with your research and look forward to reading about it in your reports.

Yours sincerely

Madeline Banda
Executive Secretary
Auckland University of Technology Ethics Committee
Cc: Andrew Kilding, Patria Hume

MEMORANDUM

Auckland University of Technology Ethics Committee (AUTEC)

To: James Croft
From: **Madeline Banda** Executive Secretary, AUTEC
Date: 3 November 2009
Subject: Ethics Application Number 09/228 **Muscle mechanics and performance of cyclists and triatheletes: kinematics, pedal forces and ultrasound evaluation.**

Dear James

Thank you for providing written evidence as requested. I am pleased to advise that it satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC) at their meeting on 12 October 2009 and that I have approved the first stage of your ethics application. This delegated approval is made in accordance with section 5.3.2.3 of AUTEC's *Applying for Ethics Approval: Guidelines and Procedures* and is subject to endorsement at AUTEC's meeting on 14 December 2009.

Your ethics application is approved for a period of three years until 3 November 2012.

This approval is only for the Brazilian component of the research. Full information about the other stages needs to be approved by AUTEC before any data collection for those stages commences.

I advise that as part of the ethics approval process, you are required to submit the following to AUTEC:

- A brief annual progress report using form EA2, which is available online through http://www.aut.ac.nz/research/research-ethics. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 3 November 2012;
- A brief report on the status of the project using form EA3, which is available online through http://www.aut.ac.nz/research/research-ethics. This report is to be submitted either when the approval expires on 3 November 2012 or on completion of the project, whichever comes sooner;

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are reminded that, as applicant, you are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

Please note that AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to make the arrangements necessary to obtain this. Also, if your research is undertaken within a jurisdiction outside New Zealand, you will need to make the arrangements necessary to meet the legal and ethical requirements that apply within that jurisdiction.

When communicating with us about this application, we ask that you use the application number and study title to enable us to provide you with prompt service. Should you have any further enquiries regarding this matter, you are welcome to contact Charles Grinter, Ethics Coordinator, by email at ethics@aut.ac.nz or by telephone on 921 9999 at extension 8860.

On behalf of the AUTEC and myself, I wish you success with your research and look forward to reading about it in your reports.

Yours sincerely

Madeline Banda
Executive Secretary
Auckland University of Technology Ethics Committee

Cc: Rodrigo Rico Bini rbini@aut.ac.nz, Andrew Kilding, Patria Hume

MEMORANDUM

Auckland University of Technology Ethics Committee (AUTEC)

To: Patria Hume
From: **Madeline Banda** Executive Secretary, AUTEC
Date: 14 May 2010
Subject: Ethics Application Number 10/56 **Technique training effects of performance of cyclists.**

Dear Patria

Thank you for providing written evidence as requested. I am pleased to advise that it satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC) at their meeting on 12 April 2010 and that I have approved your ethics application. This delegated approval is made in accordance with section 5.3.2.3 of AUTEC's *Applying for Ethics Approval: Guidelines and Procedures* and is subject to endorsement at AUTEC's meeting on 14 June 2010.

Your ethics application is approved for a period of three years until 14 May 2013.

I advise that as part of the ethics approval process, you are required to submit the following to AUTEC:

- A brief annual progress report using form EA2, which is available online through http://www.aut.ac.nz/research/research-ethics. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 14 May 2013;

- A brief report on the status of the project using form EA3, which is available online through http://www.aut.ac.nz/research/research-ethics. This report is to be submitted either when the approval expires on 14 May 2013 or on completion of the project, whichever comes sooner;

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are reminded that, as applicant, you are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

Please note that AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to make the arrangements necessary to obtain this. Also, if your research is undertaken within a jurisdiction outside New Zealand, you will need to make the arrangements necessary to meet the legal and ethical requirements that apply within that jurisdiction.

When communicating with us about this application, we ask that you use the application number and study title to enable us to provide you with prompt service. Should you have any further enquiries regarding this matter, you are welcome to contact Charles Grinter, Ethics Coordinator, by email at ethics@aut.ac.nz or by telephone on 921 9999 at extension 8860.

On behalf of the AUTEC and myself, I wish you success with your research and look forward to reading about it in your reports.

Yours sincerely

Madeline Banda
Executive Secretary
Auckland University of Technology Ethics Committee

Cc: Rodrigo Bini rbini@aut.ac.nz, Andrew Kilding, James Croft

APPENDIX 10: COPYRIGHT PERMISSION FORMS

LETTER SEEKING PERMISSION FROM PUBLISHERS/THIRD PARTY COPYRIGHT MATERIAL

Rodrigo Bini
Sport Performance Research Institute New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006, Auckland, New Zealand
Phone (649) 921 9999 ext 7295
Fax (649) 921 9960
Email: rbini@aut.ac.nz

7th August 2011

Adis Data Information BV
Dear Jenny Thomson
I am the author/co-author of "**Effects of Bicycle Saddle Height on Knee Injury Risk and Cycling Performance**" ("the Work") which was published by **Adis, a Wolters Kluwer business** in **Sports Medicine** and for which the copyright was assigned to **Adis, a Wolters Kluwer business** by an agreement dated **8th of June 2010**.
I am a doctoral/master's student at Auckland University of Technology and would like to include the Work in my doctoral thesis '**Effects of saddle position on pedalling technique and methods to assess pedalling kinetics and kinematics of cyclists and triathletes'**. The Work would be fully and correctly referenced in this thesis. A print copy of this thesis when completed will be deposited in the Auckland University of Technology Library, and a digital copy will also be made available online via the University's digital repository, ScholarlyCommons@AUT http://autresearchgateway.ac.nz/ . This is a not-for-profit research repository for scholarly work which is intended to make research undertaken in the University available to as wide an audience as possible. I would be grateful if you, or the company you represent, could grant me permission to include the Work in my thesis and to use the Work, as set out above, royalty free in perpetuity.
If you agree, I should be very grateful if you would sign the form below and return a copy to me. If you do not agree would you please notify me of this. I can most quickly be reached by email at rbini@aut.ac.nz. Thank you for your assistance. I look forward to hearing from you.
Yours sincerely
Rodrigo Rico Bini

I/~~We~~ J L Thomson hereby grant permission for use of the Work for the purposes and the terms identified above.

For and on behalf of [*name of company*] Adis, a Wolters Kluwer business (Sports Medicine journal)

Date: 8/8/2011

LETTER SEEKING PERMISSION FROM PUBLISHERS/THIRD PARTY COPYRIGHT MATERIAL

Sample letter from author to publisher seeking permission to include in a thesis previously published material written by the author in which copyright has been assigned to the publisher

Rodrigo Bini
Sport Performance Research Institute New Zealand
School of Sport and Recreation
Auckland University of Technology
Private Bag 92006, Auckland, New Zealand
Phone (649) 921 9999 ext 7295
Fax (649) 921 9960
Email: rbini@aut.ac.nz

7th August 2011

Elsevier Ltd
Dear Patrick Clifton
I am the author/co-author of "**A comparison of cycling SRM crank and strain gauge instrumented pedal measures of peak torque, crank angle at peak torque and power output"** and "**Effects of saddle height on pedal force effectiveness**" ("the Work") which was published by **ELSEVIER Ltd** in **Procedia Engineering** and for which the copyright was assigned to **ELSEVIER Ltd** by an agreement dated **19th of March 2011**.
I am a doctoral/master's student at Auckland University of Technology and would like to include the Work in my doctoral thesis '**Effects of saddle position on pedalling technique and methods to assess pedalling kinetics and kinematics of cyclists and triathletes'**. The Work would be fully and correctly referenced in this thesis. A print copy of this thesis when completed will be deposited in the Auckland University of Technology Library, and a digital copy will also be made available online via the University's digital repository, ScholarlyCommons@AUT http://autresearchgateway.ac.nz/ . This is a not-for-profit research repository for scholarly work which is intended to make research undertaken in the University available to as wide an audience as possible. I would be grateful if you, or the company you represent, could grant me permission to include the Work in my thesis and to use the Work, as set out above, royalty free in perpetuity.
If you agree, I should be very grateful if you would sign the form below and return a copy to me. If you do not agree would you please notify me of this. I can most quickly be reached by email at rbini@aut.ac.nz. Thank you for your assistance. I look forward to hearing from you.
Yours sincerely
Rodrigo Rico Bini

I/We ______________________ hereby grant permission for use of the Work for the purposes and the terms identified above.

For and on behalf of [*name of company*] ELSEVIER LTD.

Date: 8 AUGUST 2011

SIGNED BY DAN LOVEGROVE