BICYCLE RIDER CONTROL:

A BALANCING ACT

By

Borut Fonda

A thesis submitted to The University of Birmingham for the degree of

DOCTOR OF PHILOSOPHY

College of Life and Environmental Sciences

School of Sport, Exercise and Rehabilitation Sciences

University of Birmingham

June 2015

ABSTRACT

Cycling is increasing in popularity which is accompanied with a higher rate of injuries sustained due to collisions, crashes or falls. A high proportion of these events happen when the bicycle rider loses control of the bicycle. In order to improve bicycle rider control, the skill of riding a bicycle needs to be understood. Therefore, the overall aim of this PhD work was to explore bicycle rider control skills and to examine the effects of different constraints on the control of a bicycle. The first part of this thesis focuses on developing a valid and reliable methodology that can be further used for studying bicycle rider control skill. Firstly, a protocol to determine knee angle during cycling is being developed. Secondly, some technical approaches when studying muscle activity during cycling are being questioned. Lastly, a portable device based on a single angular rate sensor to record steering rate and bicycle roll rate was tested for reliability in an outdoor setup. Second part of the thesis examines the effects on bicycle rider control of three different constraints: 1) expertise, 2) body position and 3) cycle lane design. Results overall showed that all three constraints significantly affect steering and bicycle roll rate.

To my beloved wife Nina

ACKNOWLEDGMENTS

I would like to gratefully and sincerely thank Dr François-Xavier Li for his guidance, understanding and patience during my studies at the University of Birmingham. His mentorship has been paramount in providing a wonderful experience and in supporting my long-term career goals. He has encouraged me to not only grow as a researcher and as a sports scientist, but also as a lecturer, mentor and, most importantly, an independent thinker. Special thanks to the Dutch team, a.k.a. team Matlab (Danique, Eline, Bastiaan, Carlijn, Mark), for introducing me to Matlab and providing me with an "infinite loop" of programming support. Special thanks also to Simon for proofreading basically every word I've written in the past three years.

Secondly, without the support from my boss, mentor and my friend, Dr Nejc Šarabon, I would never have even considered applying for a PhD position in Birmingham. Thank you Nejc and S2P for allowing me to extend my knowledge and culture abroad. In no particular order, thank you Ryan, Beni, Simon, Gareth, Laura, Marcus, Matt, Mike, Scott, Paul, Alex, Maria, Martina, Keejoon, Danique, Mark, Eline, Bastiaan and many, many others for being part of my life. I will never forget you.

Finally, and most importantly, I would like to thank my wife Nina. Her support, encouragement, quiet patience and love has unquestionably been the most significant contributing factor that has allowed me to develop as a person. I could not have wished for a better person to spend the rest of my life with. Massive thanks also to my family for supporting my ambitions in the last 27 years.

Table of Contents

List of Figures

List of Tables

LIST OF PUBLICATIONS

CHAPTER 2

Fonda B, Sarabon N, Li F-X. Validity and reliability of different kinematics methods used for bike fitting. Journal of Sport Sciences. 2014;32(10):940–6.

Fonda B, Sarabon N, Li F-X. Validity of methods to assess muscle activity during steady state cycling: a technical note. Journal of Electromyography and Kinesiology. 2015; under review

Fonda B, Sarabon N, Li F-X. Bicycle rider control assessment: a reliability study using an angular rate sensor. *Sports Biomechanics, 2015;* under review

CHAPTER 3

Fonda B, Sarabon N, Li F-X. Bicycle Rider Control Skills: expertise and assessment. Journal of Sport Sciences. 2015; in press.

CHAPTER 4

Fonda B, Sarabon N, Blacklock R, Li F-X. Changing the bicycle seat height: effects on rider control. *European Journal of Sport Science*. 2015; under review

CHAPTER 5

Fonda B, Sarabon N, Li F-X. Cycle lane design affects bicycle rider control. *Accident Analysis & Prevention*. 2015; in preparation.

ABBREVIATIONS

2D	Two dimensional
3D	Three dimensional
ANOVA	Analysis of variance
BDC	Bottom dead centre
BF	m. Biceps femoris
DEV	Deviation
EMG	Electromyography
GCL	m. Gastrocnemius lateralis
ICC	Intra-class coefficient of correlation
MAP	Maximal aerobic power
RF	m. Rectus femoris
RMS	Root mean square
ROLL	Roll
SO	m. Soleus
STE	Steering
TA	m. Tibialis anterior
TDC	Top dead centre
VL	m. Vastus lateralis
VM	m. Vastus medialis

CHAPTER 1. GENERAL INTRODUCTION

1.01 Cycling

Cycling is both a popular form of transport and a popular form of physical activity. As an active mode of commuting, cycling has become popular in order to avoid road congestion and traffic jams whilst allowing commuters to arrive at work on time and improve their health and well-being simultaneously. Cycling is increasing in popularity with more and more people cycling daily. The 2011 Census report (Prothero, 2014) revealed that approximately 43 % of the population in the United Kingdom have access to a bike and about 34 % of the population cycle at least once a year, with these numbers having increased in the past 10 years. It has been reported that the average distance cycled annually in the United Kingdom has increased by 20 % in the last 15 years with a greater increase in bike usage being observed in urban areas. For example, bike usage in London increased by 176 % from 2000 to 2012.

It is well known that cycling as a form of active commuting offers meaningful potential for improving well being and is recognised as an important factor in increasing public health (Bauman & Rissel, 2009; Oja et al., 2011). Commuting cycling can be seen as exercise that is relatively easy to incorporate into one's daily routine (Hendriksen, Simons, Garre, & Hildebrandt, 2010). Because cycling is mainly an aerobic activity, it positively affects cardiorespiratory, metabolic and musculoskeletal functions of the body which leads to many potential health benefits (Oja et al., 2011). In particular, previous studies have shown cycling to reduce the risk of cardiovascular diseases (Hamer & Chida, 2008), premature mortality (Andersen, Schnohr, Schroll, & Hein, 2000) and obesity (Lindström, 2008) whilst improving physical performance at the same time (de Geus, Joncheere, & Meeusen, 2009). Positive contributions of cycling towards health and well-being have been recognised in the majority of the world's biggest cities, which have started to design and implement interventions to further increase cycling participation (Pucher, Dill, & Handy, 2010). The London Cycling Campaign is one such example; striving to make the city a healthier and more cycling-friendly place through initiatives aimed at improving cycle safety and encouraging cycling over motorised methods of transport.

The increasing number of cyclists coincides with an increased absolute number of reported incidents (i.e. collisions, crashes and falls). However, when relativized, doubling the number of cyclists is linked with a 40 % increase in cycling incidents. Furthermore, a 34 % reduction in the relative risk per cyclist has been reported for the last 10 years (Jacobsen, 2003). Major research carried out in Belgium (Vandenbulcke et al., 2009) reported that places with a high proportion of cyclists are associated with lower risks of cycling incidents. Moreover, a decline in the overall number of cycling related incidents has been observed in London, which is probably due to the implementation of a "congestion tax", the construction of new cycling infrastructure (cycle lanes), cycling education and improved public transport (Bauman & Rissel, 2009).

Cycling incidents are often thought to outweigh the health benefits of cycling. However, when the benefits are compared to the risks, it is clear that the relative magnitude of the risk is low. Positive effects on chronic diseases, obesity and mental health are all reported to be higher than the risk of an injury during cycling (Bauman & Rissel, 2009; Hillman, 1993).

An advantage of active commuting, such as cycling, is the benefit to the environment through a significant reduction in air pollution. Cycling itself produces zero carbon pollution and is therefore an environmentally friendly solution (Chapman, 2007).

1.01.1 Road safety statistics

Scientific literature often uses the word 'accident' as a term to describe an event resulting in injuries. Accidents are normally understood to be unpredictable and unavoidable (Davis, 2001). However in cycling, most of these events are predictable and thus potentially avoidable (eg. hitting an obstacle could be avoided by spotting te obstale earlier). Henceforth, in this thesis, the following terminology will be used: 'Incidents', which describes an event that results in traumatic injury occurrence. This can either be a 'collision' (hitting a vehicle or other moving object), a 'crash' (hitting a stationary object) or a 'fall' (losing control without hitting another object).

According to the Statistical Release by the Department of Transport (Lloyd, 2013), every year 19,000 cyclists are killed or seriously injured in reported road incidents in the United Kingdom. The total number of injured cyclists has increased by 11 % from 2004 to 2008 (Knowls et al., 2009). Most of the incidents where cyclists are involved happen at or near junctions. The key factor for these incidents was reported to be a collision with motor vehicles where the drivers failed to look properly (Knowls et al., 2009). However, over 16 % of the reported incidents do not involve a collision with another vehicle and loss of control by the rider has been identified as the most significant contributing factor. A similar trend was observed in the late 1980's when Illingworth et al. (1981) examined the causes for cycling-related injuries. They reported that the majority of incidents happened due to cyclists losing control or crashing into obstacles. Whilst the majority of the cycling injuries result from falling, the largest contributing factor to fatal injuries is a collision with another motor vehicle (Noakes, 1995). It is worth noting that the total number of incidents due to loss of control is most likely much higher due to unreported incidents where cyclists do not seek medical care (Hillman & Morgan, 1992; Maimaris, Summer, Browning, & Palmer, 1994). Recent studies from Canada (Harris et al., 2011; Teschke et al., 2014a) included interviews with injured cyclists who were involved in road incidents, finding that approximately 60% of the reported incidents were crashes or falls (without involvement from a third party, i.e. motor vehicle, animal or pedestrian).

Reported statistical facts on cycling road safety clearly indicate the need for future research and development to prevent crashes and falls. As losing control is one of the most prominent contributing factors, a primary aim of this thesis is to examine some of the human and environmental factors that could potentially influence bicycle rider control.

1.01.2 Bicycle dynamics and stability

To examine bicycle rider control from a biological perspective, it must first be understood how a bicycle is stabilised without any control. From a mathematical modelling perspective, bicycles could be considered as multi-body systems. First attempts to examine the dynamics of a bicycle were made in 1869 by Rankine (1869), which is based on a model to study balance, steering and propulsion. These models have a historic value but were later found to have very little technical contribution in the area of single-track vehicle dynamics (Schwab & Meijaard, 2013). The first actual description of the general motion of a bicycle and a rider was done by Whipple (1899) with a set of nonlinear differential equations.[1]

A typical bicycle has two wheels which are in contact with the road surface. This is normally the only physical contact the rider-bicycle system has with the environment. As there are only two contact points, which are nominally placed directly behind each other, the bicycle becomes laterally unstable at low speeds and acts like an inverted pendulum (Schwab & Meijaard, 2013). This means that, in order to stabilise a bicycle, the support point needs to be shifted in the direction of the undesired or desired roll (side tilt of a bicycle). Whilst moving forward, shifting the support point can be achieved by steering towards the direction of the roll of the bicycle, either by providing human control or relying

[1] Sophisticated and detailed description of the mathematical modelling of a bicycle and cyclist is beyond the scope of this work and will not be discussed further.

on the dynamics of the bicycle. The later is commonly referred to as a phenomenon of bicycle self-stability.

Self-stability has been a popular topic in the literature on dynamics of two-wheeled single-track vehicles. Initially, engineers thought that a bicycle is self-stable due to the gyroscopic effect from the rotating wheels, the trail of the front wheel (caster effect), the steering head inclination and the mass distribution of the front fork. The importance of the gyroscopic and caster effect was also supported with theoretical models (Meijaard, Papadopoulos, Ruina, & Schwab, 2007). Bicycle stability was first described by Whipple (1899) and then furthered by the researchers from the Delft University of Technology in the Netherlands (Kooijman, Meijaard, Papadopoulos, Ruina, & Schwab, 2011; Meijaard et al., 2007), predicting that a bike does not need a gyroscopic and caster effect to reach a self-stable state. This has been experimentally confirmed by Kooijman et al. (2011) with a self-stable bike that contained counter rotating wheels to cancel the angular momentum and trail essentially a bicycle without the gyroscopic and caster effect (see Figure 1.1). They have shown that this bicycle achieves self-stability due to the two frames that are hinged together with a smaller mass at the front frame (fork and handlebars). A smaller frame falls faster than a bigger frame (rear frame), causing steering in the direction of the fall and maintaining stability. This can be illustrated with the example of vertically balancing a stick on your fingers; a smaller stick will be more difficult to balance than a longer stick (slower inverted pendulum) and thus a smaller stick falls quicker.

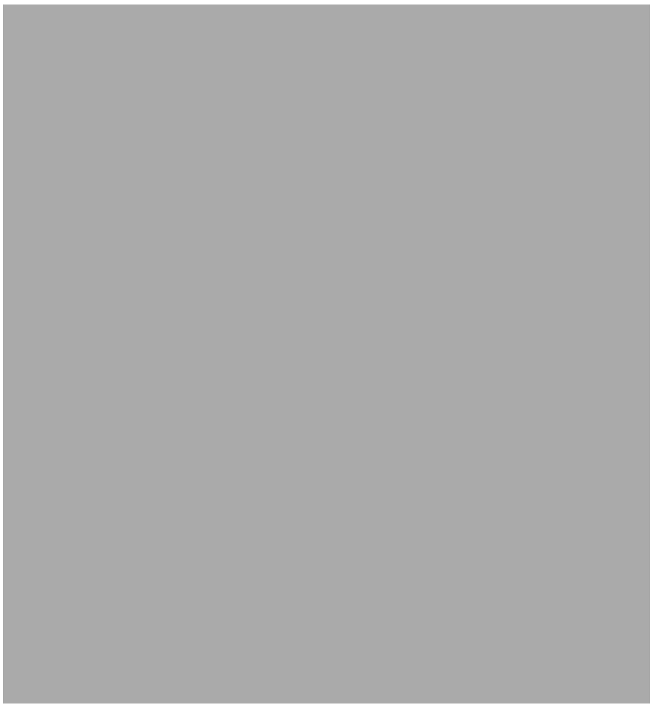


Figure 1.1: Self-stable bicycle without a trail and cancelled gyroscopic effect. Picture taken from bicycle.tudelft.nl.

Taking a bicycle to a certain speed and then releasing it can demonstrate the self-stability of a bicycle. When the velocity of the bicycle is low, the bicycle will start to laterally oscillate until the point when it will finally fall down. However, if the velocity is high enough, the bicycle will reach a stable motion without lateral oscillations and will reach a self-stability state (Schwab, 2012).

Understanding bicycle dynamics plays an important role in the controllability of a bicycle. However, the interaction of the bicycle dynamics and human control is what keeps the bicycle upright and headed in a desired direction. In essence, an improved understanding of the interaction between motor control and bike dynamics can lead to improved control of the bicycle and, consequently, safety. It is worth noting that the intention of this thesis is not to study mathematical models and dynamics of a bicycle, but to examine the less known human factor (i.e. motor control).

1.02 Motor control during cycling

1.02.1 Bicycle control and balance

From a balance point of view, riding a bike is a closed-loop motor control action with the eyes and vestibular apparatus as the main (but not the only) receptors for feedback. Closed-loop systems involve the processing of feedback against a reference of correct movement, determining errors in the movement and a subsequent correction of the movement. Vision, with its receptor the eyes, is certainly the most critical receptor for supplying information about the movement in respect to the environment (Schmidt & Lee, 1988). It has been shown that vision functions as an integral component of the control system for maintaining balance (Lee & Lishman, 1975). Furthermore, in the absence of visual information, one cannot make anticipatory actions to avoid collisions with moving or static objects and simply riding a bicycle on even ground without any obstacles would prove very difficult due to the lack of information on velocity and movement direction. As opposed to walking, where information about velocity can be provided from the leg velocity (Prokop, Schubert, & Berger, 1997), a cycling task does not have other sufficient means of providing information on the velocity of the movement.

The main motions during cycling to maintain balance and heading were found to be steering, bicycle roll, leaning, knee bouncing and spine twist (Kooijman, Schwab, & Moore, 2009; Moore, Kooijman, Schwab, & Hubbard, 2011). It has also been demonstrated that constant steering and roll motions are present to maintain heading in a straight line (Kooijman et al., 2009; van den Ouden, Ouden, & den Ouden, 2011). The main motions present during cycling are illustrated in Figure 1.2.

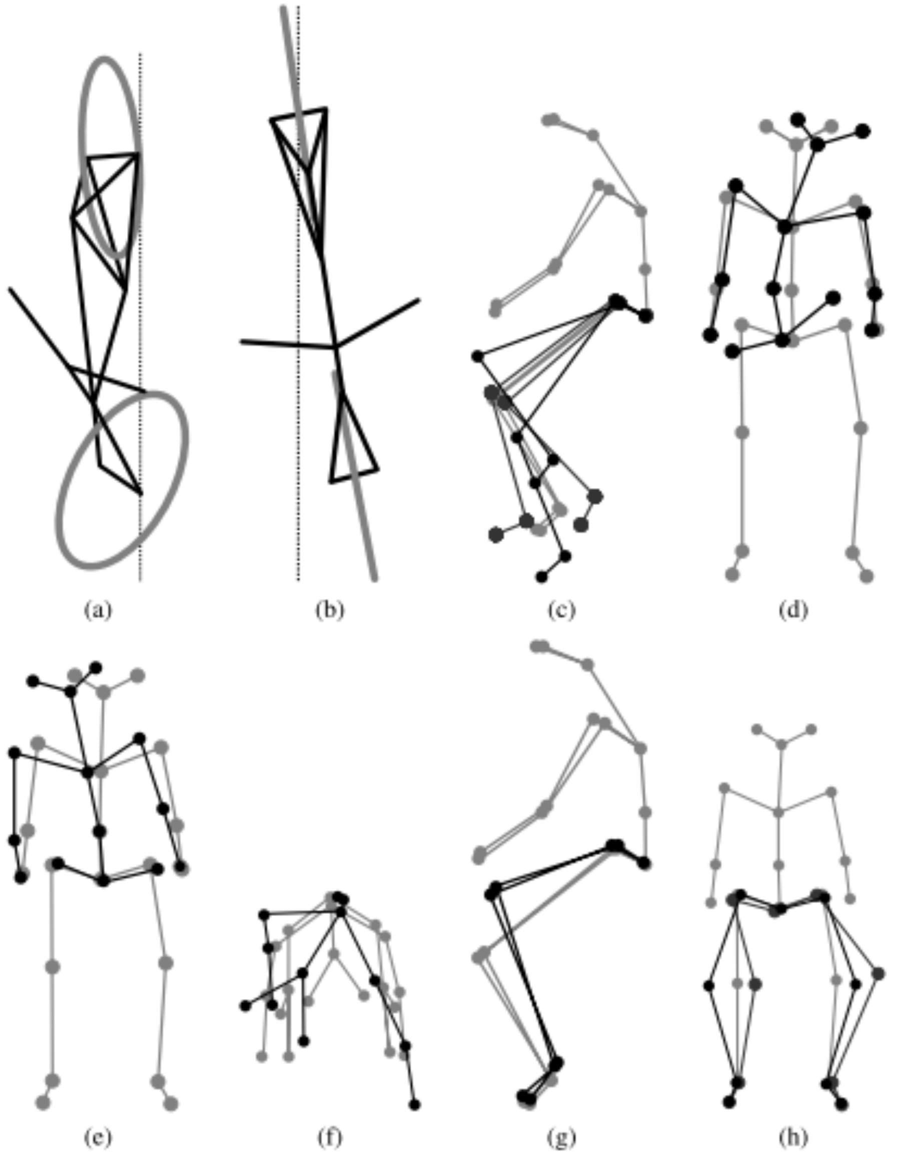


Figure 1.2: Main bicycle and rider motions to maintain balance. (a), steering bicycle roll; (b), bicycle yaw; (c), pedalling; (d), spine bend; (e), rider lean; (f), rider twist; (g), knee bounce; (h), lateral knee movements. Graphic taken from Moore, et al. (2011).

1.02.2 Bicycle rider control as a motor skill

Motor skills can be conceptualised in different ways. One can classify a motor skill as a specific task, e.g. a karate punch or driving a car. This leads to a number of dimensions and levels of motor skills, at least one for each task. Furthermore, a motor skill could also be conceptualised based on the proficiency one demonstrates whilst performing a certain movement, allowing classification of a motor skill as a task or an act. Hence, we can define a motor skill as a skill for which the primary determinant of success is the quality of the movement that the performer produces (Schmidt & Wrisberg, 2004).

Further classification can be done on the level of task organisation, which concerns the way the movement is organised. A brief duration and a clearly defined beginning and end describe discrete skills. Examples of discrete skills are jumps, kicking a ball etc. More complicated tasks, where discrete skills are linked together to form more complicated actions, are called serial skills. Examples in sport which describe serial skills are found in gymnastic routines, where a series of discrete elements form the final performance. The last category of task organisation involves the task where the movement has no definable beginning or end. These skills are called continuous skills and are repetitive or rhythmic and may last for a longer period of time. The duration of this kind of task is usually defined by the performer or an external object (e.g. the end of a running track).

Riding a bicycle along a path is a typical continuous skill; forcing riders to steer with the handlebar, lean the body, roll the bicycle and laterally move the knees to keep the bicycle balanced and on the correct trajectory whilst at the same time pedalling to ensure forward movement. Motor skills in general form an assortment of human experiences. Understanding how we learn the skill of riding a bicycle and how we become better in this particular skill is essential for the educational and safety setting (Clark, 1995). Becoming skilful in riding a bicycle also has other benefits, mainly when engaging in cycling as a form of physical activity or during the daily commute. Cyclists who possess sufficient skill at controlling their bicycle are likely to choose to cycle more often or use their bicycle as a means of transport more frequently.

Research on motor control during cycling has not yet established a collective variable that could capture cycling as a global pattern. If such a variable were to be established, it could then be studied for its improvements and changes. Hence, one of the goals of this PhD work was to narrow down the variables that describe bicycle rider control and establish a collective measure for this skill. Traditional approaches to research on motor behaviour, which can be observed in recent sports science (Bartlett, Wheat, & Robins, 2007), have used functional variability as a measure of variance in motor control performance. Variability is the most common feature in human movement (Latash, Scholz, & Schöner, 2002). From a dynamical system perspective, variability of human movement has been the central issue of research and is often studied in isolation. Variability has been related to the sensorimotor equivalence that arises from the abundance of motor system degrees of freedom, describing the human body (Davids, Glazier, Araujo, & Bartlett, 2003). Therefore, variability is not only expected, but is also an important part of the system.

1.02.3 Dynamical Systems Perspective for bicycle rider control

Dynamical system theory is a multidisciplinary system-led approach that considers different areas (mathematics, physics, psychology and chemistry) to describe the constantly changing system over different timescales (Davids et al., 2003). The dynamical system theory is using a concept to describe individual variability that is linked to constraints. Constraints are defined as boundaries interacting with the biological system to form a limitation and achieve optimal states of organisation (Clark, 1995; Davids et al., 2003). In essence, constraints reduce the number of configurations available to the system (Newell & van Emmerik, 1989).

Motor behaviour emerges from the constraints surrounding a particular task. For example, a constraint in cycling is body position. With three contact points between the body and the bicycle, there is a limited range of motion and a number of degrees of freedom to control the bicycle so the body position forms a boundary to limit the extent to which one can control the bicycle. However, to date, no studies have empirically examined the effects of changing body position on human control of the bicycle. As changing the body position on a bicycle is an easy but limited task, one part of this PhD work aimed to examine if there is an "optimal" position that can result in improved control of the bicycle. More about the body position constraint will be discussed in section 1.04 and is empirically addressed in CHAPTER 4.

Motor control and motor behaviour also changes with respect to experience gained. Learning new skills or improving existing ones occurs across a person's life span (Clark, 1995). The same is true for the skill of bicycle control, which is dependent on the cyclist's previous experience and biological status. Therefore, experience or expertise can present another constraint that in turn affects bicycle rider control. Expertise in relation to motor control will be further discussed in section 1.03 and examined in CHAPTER 3.

The last constraint that is within the scope of this thesis is the environment and, in particular, cycling facilities. The environment often presents a constraint in any task. For example, walking on a busy street forces people to continuously navigate between static and moving objects which constantly affects motor control. Another point of view could be perceived safety. Fear is a known factor that affects motor control. For example, a study (Osler, Tersteeg, Reynolds, & Loram, 2013) showed alterations in body sway during quiet standing on a balance beam positioned higher from the ground when compared to standing on a balance beam lower to the ground. Regardless of the task in this case, which was completely the same (quiet standing), the motor behaviour changed. How perception of safety affects bicycle rider control is one of the questions of this thesis and will be empirically addressed in CHAPTER 5.

1.02.4 Affordances

Affordances are defined as the opportunities for a given organism in a given environment and are closely linked to the physical characteristics of the organism (Gibson, 1979). Affordances can be met in everyday environments. For example, when walking on a busy street, one must constantly use visual information to accurately perceive possible affordances to plan and perform necessary adjustments to avoid possible collisions (Turvey, 2010).

Affordances during locomotion have been mainly studied during walking (Warren & Whang, 1987; Wilmut & Barnett, 2010). An interesting observation reported was that people rotate their shoulders before walking through a narrow gate, to always leave a safety margin of at least 1.3 times that of their body width (shoulders). This means that even if the gate/gap is wide enough to pass through without any posture modification, people will still rotate their body to leave that safety margin (Wilmut & Barnett, 2010). However, during cycling, shoulder rotation would have little effect as the widest point of the cyclist-bicycle system is often represented as the handlebar. In order to ensure smooth passage through a narrow (but still wide enough) gap, the only logical strategy would be to ride as straight as possible.

That, however, has not yet been empirically examined and will be partially discussed in section 1.05.2 and then later addressed in CHAPTER 5 of this thesis.

1.02.5 Neuromuscular control during cycling

In the past, studies on neuromuscular control during cycling were mainly focused on explaining cycling motion in terms of pedalling mechanics (Li, 2004; Raymond et al., 2005). A common method used to study activation patterns in movement sciences is electromyography (EMG) (Hug & Dorel, 2009). Despite its limitation, it can represent a valid method to determine temporal (timing) and amplitude characteristics of muscle coordination (Hug, 2011).

A common concept in motor control is identifying functional muscle groups responsible for a certain movement. A functional muscle group is a group of muscles that work in synergy and that are activated at roughly the same time (Jacobs & Macpherson, 1996). Raasch & Zajac (1999) found that there are six functional groups responsible for continuous, smooth and efficient pedalling. These groups are: 1) uni-articular hip and knee extensors (vastus medialis, vastus lateralis, gluteus maximus), 2) uni-articular hip and knee flexors (iliopsoas, short head of biceps femoris), 3) hamstrings (semimebranosus, semitendinosus, long head of biceps femoris), 4) rectus femoris, 6) ankle plantar flexors (soleus, gastrocnemius) and 6) ankle dorsal flexors (tibialis anterior). All of these muscles were also found to be the main active muscles during cycling, albeit with different roles and activation times (Figure 1.3). Uni-articular muscles were identified as the main power producing muscles, whereas the bi-articular muscles serve as force direction control and energy transfer (Fonda & Sarabon, 2010; Gregor, Broker, & Ryan, 1991; Ryan & Gregor, 1992).

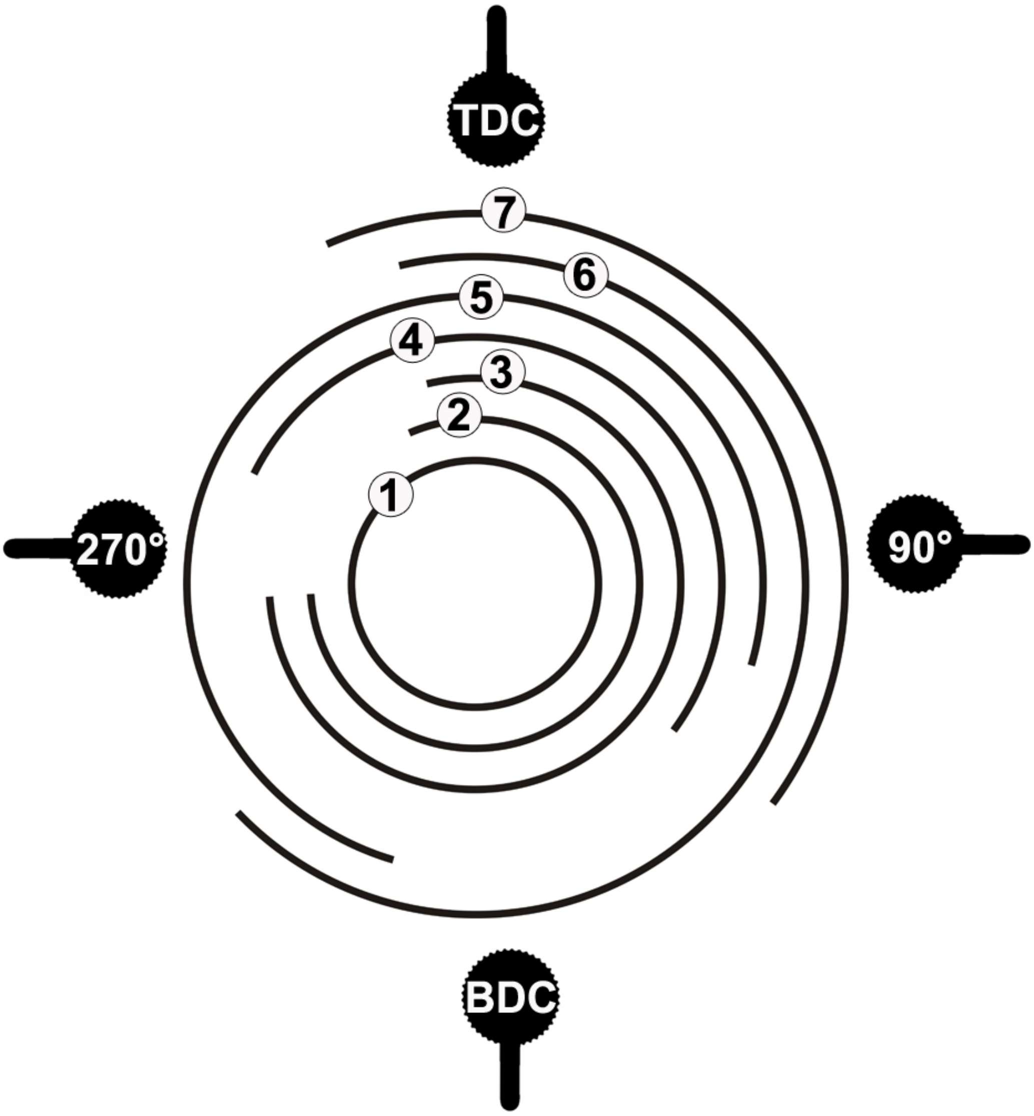


Figure 1.3: Schematic illustration of the muscle activation during cycling at fixed intensity and cadance. (1, tibialis anterior; 2. soleus; 3, gastrocnemius lateralis and medialis; 4, vastus lateralis and vastus emdialis; 5, recuts femoris; 6, biceps femoris; 7, gluteus maximus

Validity and reliability of the EMG patterns during cycling have previously been studied. Dorel, Couturier, & Hug (2008) reported good repeatability of temporal and amplitude characteristics in EMG activity of the leg muscles before and after cycling exercise, which was latter confirmed by other studies (Jobson, Hopker, Arkesteijn, & Passfield, 2013; Laplaud, Hug, & Grélot, 2006). To minimise the effects of variability and to improve the signal-noise ratio of the EMG pattern, a sufficient number of cycles need to be averaged (Bruce, Goldman, & Mead, 1977). In the current peer-reviewed literature, researchers used a range from 5 (Laplaud et al., 2006) to 50 (Fonda, Panjan, Markovic, & Sarabon, 2011) cycles to get an ensembled average of the muscle activity pattern during cycling. To date, there is no general consensus defining the required minimum number of cycles needed to create an ensembled average for a representative muscle activity pattern during cycling. Furthermore, to eliminate temporal shifts of the EMG patterns due to non-constant pedal velocity throughout the pedal cycle, it is recommended to use continuous measurement of the crank position for EMG rescaling (Hug & Dorel, 2009). However, that can sometimes be more technically demanding (eg. outdoor measurements) and it is therefore important to determine what the minimum number of crank detection points for EMG rescaling is in order to get a representative pattern.

The majority of the studies using EMG to study the effects of different constraints have been performed indoors on a fixed cycling ergometer which does not require active balancing. As one of the movement strategies to keep balance during cycling at lower speeds (< 10 km/h) includes lateral movement of the knees (Kooijman et al., 2009; Moore et al., 2011), the actual EMG pattern might differ to the one observed during ergometer cycling. To the author's best knowledge, no studies so far have examined muscle activation during cycling at lower speeds, requiring active balancing of the bicycle, and compared this to the activation patterns during ergometer cycling.

However, before such a study is conducted, a valid method and protocol to analyse EMG data need to be established. Hence, one of the goals of this thesis was to develop a method that would allow reliable and valid interpretation of the EMG data obtained during cycling.

1.03 Expertise

Expert performance is commonly defined as “consistently superior performance on a specific set of representative tasks for a domain” (Ericsson, 2014). Motor skills such as riding a bicycle are often associated with a certain level of expertise. In cycling terminology, the term “experts” often represent a cohort of people with superior physical fitness that is associated with their sporting achievements. However, the skill to control the bicycle does not necessary reflect one’s physical fitness. The first problem researchers encounter is the definition of the rider control skill and defining an objective measure to assess that skill. This skill could embrace reactions to dangerous situations, the ability to ride in a smooth and straight line or fast cornering for example. However, in this thesis, rider control skill will be considered as the ability to maintain balance and direction of the bicycle on a predefined trajectory.

The ability to maintain balance during cycling cannot be directly associated with static balance. The difference between the two tasks can be in the constant search for equilibrium with the intention of maintaining the body in an upright position, achieved by shifting the support point during cycling and using joint moments during quiet standing. This means that, even during quiet standing, oscillations are present (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). Similarly, during cycling in a straight line, lateral oscillations are present which are counterbalanced by steering and roll motions (van den Ouden et al., 2011). In the assessment of static balance, body sway parameters (centre of pressure displacement) describe the ability to maintain balance. The most reliable and sensitive parameters are linked to the amplitude, velocity and frequency of the centre of pressure movement.

When riding a bicycle, an essential skill to learn is that of steering in the direction of the roll. This becomes a difficult task for children using stabilisers as they experience less feeling for roll rate. Cain, Ulrich, & Perkins (2012) showed that children achieve a better cross-correlation between steering rate and roll rate throughout the process of learning riding a bicycle. To be exact, they (Cain et al., 2012) reported an increase in normalised peak cross-correlation coefficient from 0.22 to 0.75 during the learning process. However, once the balance becomes less of an issue, cross-correlation coefficient does not provide any more insight into the level of expertise. Cain (2013) also showed that experienced cyclists use different strategies to maintain balance while riding a bicycle on rollers mounted on a force platform. Rollers are a type of ergometer to ride a bicycle indoors without moving forward, however they are not attached to the bicycle and thus force the cyclist to control their balance. Rollers consist of three cylinders ("rollers"), one supporting the front wheel and two supporting the rear wheel. Cain (2013) showed that more experienced cyclists tend to use more rider lean than steering to maintain balance. This was further supported by the observation that more experienced cyclists exhibited better “balance performance” than less experienced cyclists, despite a lower cross-correlation between steering rate and bicycle roll rate. The main limitation of Cain’s study was that they used a testing protocol on rollers instead of actual riding with forward motion. Therefore, one of the aims of this PhD thesis was to examine the motor control strategies that are employed to maintain balance while riding a bicycle, and investigate how these differ between more and less experienced cyclists (see CHAPTER 3).

1.04 Body position

Cyclists often adjust their position on the bicycle to meet the demands of their riding. These demands can include comfort, performance, injury prevention, etc. Body position can be altered by either changing the bicycle set-up (seat, handlebar and pedals/cranks) or by simply changing the angle at various joints (e.g. bending elbows, trunk, etc.).

Most of the existing research addressing the body position constraint focused on two aspects: (1) performance (Peveler & Green, 2011; Peveler, Pounders, & Bishop, 2007; Peveler, 2008) and (2) non-traumatic injury prevention (Bini, Hume, & Croft, 2011; Bini, Hume, Lanferdini, & Vaz, 2013). The process of changing the body position on a bicycle with an aim to meet the desired goal is called bike fitting. Bike fitting has become a marketing niche, especially among bicycle retailers. However, when a commuting cyclist buys a new bicycle, often only the seat height is adjusted according to the cyclist’s body height.

Seat height affects the location of the centre of mass of the cyclist-bicycle system. This could potentially affect the control of a bicycle as the system can become more/less stable or easier/harder to stabilise. This has been observed during quiet standing, where researchers (Rosker, Markovic, & Sarabon, 2011) changed the vertical mass redistribution by putting weights above and below the body's normal centre of mass, as shown in Figure 1.4. An improvement in static balance was demonstrated when the mass was redistributed lower.

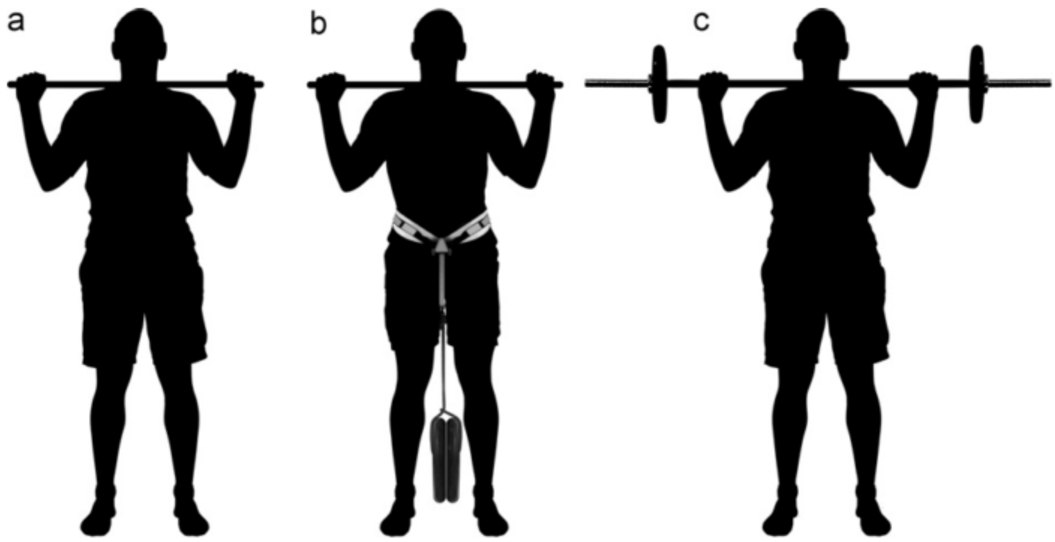


Figure 1.4: Changing the vertical mass redistribution during quiet standing. With permission from Rosker et al. (2011).

Based on the author's observations during cycling (e.g. cornering), it can be hypothesised that lowering the centre of mass would improve balance during cycling and would require less control of the bicycle. Therefore, one of the aims of this PhD work was to how seat height affects bicycle rider control.

1.05 Cycling infrastructure

1.05.1 Cycling infrastructure

Cyclists often share the road with other vehicles without a clear separation, making them vulnerable to collisions with vehicles. Therefore, cycle infrastructure availability (mainly cycle lanes) and design is an important consideration when evaluating the road safety of cyclists. A review of the existing literature suggests that the implementation of cycle lanes substantially reduces the risk of crashes and injuries sustained during cycling (Reynolds, Harris, Teschke, Cripton, & Winters, 2009; Teschke et al., 2012). At the same time, cycle infrastructure availability can encourage more people to use cycling as a means of transport or as a form of physical activity (Winters, Davidson, Kao, & Teschke, 2010; Winters & Teschke, 2010).

Chataway et al. (2014) compared the differences in safety perception between cyclists from Copenhagen and cyclists from Brisbane. Copenhagen was, and still is, an established cycling-friendly city whereas Brisbane was still an emerging cycling city at the time. In their research, they concluded that wider and standalone cycle pathways effectively reduced the fear of traffic and lead to an increase in the perception of safety. In particular, cyclists perceived infrastructure layouts safer when cycle lanes were painted. Moreover, the fear of traffic was found to negatively affect the perceived safety of the infrastructure (Chataway et al., 2014).

A situation that often puts cyclists in a dangerous position is when they ride in a straight trajectory on the side of a road and get struck by a motor vehicle (Cross & Fisher, 1977). It is a common belief among cyclists that implementation of cycle lanes would result in a clearer and greater separation of cyclists and motor traffic. However, research has demonstrated the opposite. Parkin & Meyers (2010) recorded smaller motor traffic passing distance and larger overtaking speeds of cyclists on roads with cycle lanes when compared to the roads without cycle lanes. Furthermore, wider roads and dual lane roads were associated with a larger passing distance of the motor traffic compared to narrower roads (Shackel & Parkin, 2014). An interesting observation of drivers' behaviour was recorded by Walker (2007), who found that drivers overtake females at a significantly greater distance than males. Moreover, cycling further away from the curb results in a closer overtaking. Walker's conclusion was that optimal distance of the bicycle lane from the edge of the road would be between 0.5 and 0.75 m. (Walker, 2007).

An interesting observation to come out of a study (Vansteenkiste, Zeuwts, Cardon, Philippaerts, & Lenoir, 2014) comparing gaze behaviour while cycling on a recently improved, 2 m wide cycle lane with cycling on an uneven (rough surface), 1 m wide cycle lane was that significantly more lateral eye movements were recorded when riding on a higher quality cycle lane. Further analysis revealed that there was an apparent shift of attention towards the cycle lane region on a lower quality cycle lane compared to the higher quality cycle lane. These observations indicate that, when the quality of the cycling infrastructure is higher, the cyclists can pay more attention to the surrounding environments which could potentially result in improved alertness and responsiveness to the environmental factors (pedestrians, traffic, etc.).

1.05.2 Cycle lane width

Guidelines in the UK suggest the minimum width of a cycle lane to be 1.2 m, but it is recommended that 1.5 m or 2 m width are employed on less busy or busier roads respectively (DfT, 2008). Despite these guidelines, the author of this thesis often observed narrower cycle lanes than recommended in the UK's second largest city (Birmingham), with the majority ranging between 0.8 – 1.2 m. Moreover, Frings, Parkin, & Ridley (2014) reported that the gap between the curb and the motor vehicle on different junctions was less than 1 m, including on sections with a designated cycle lane. Little is known on how different cycle lane widths affect cycling safety.

Most of the research on lane width has been performed for driving. Godley, Triggs, & Fildes (2004) reported that narrower lanes increase steering workload and reduce speed. Furthermore, McLean & Hoffmann (1972) used an instrumented car on three lane widths (2.4, 3.0, and 3.7 m) and showed an increase in the number of rapid steering movements with decreasing lane width, which was accompanied by less accurate steering (higher angular rate) at the narrowest lane compared to the two wider lanes.

Based on driving studies, one can expect that cycling on narrower lanes require more effort to control a bicycle. Recently, a study by Lee, Shin, Kang, & Lee (2015) examined the minimum desirable width of a one-way cycle lane. They used a global positioning system to measure the essential manoeuvring space and combined it with the handlebar width to estimate the minimum desirable width of a cycle lane at different cycling speeds. They concluded that a minimum desirable lane width in Korea is 2 m (Lee et al., 2015). An interesting observation was that the lateral deviation of the cyclists in their study was smaller on the narrower lanes compared to the wider ones.

Vansteenkiste et al. (2013) examined gaze behaviour during cycling and reported that cyclists are similar to pedestrians in that they direct their gaze to the near path and the goal area (end area). They also reported that gaze is directed nearer to the bicycle on narrow lanes and more towards irrelevant areas on wider lanes. This means that a narrow lane consumes the attention that could be directed to other environmental factors, such as traffic, pedestrians and obstacles.

To the author's knowledge, no studies thus far have examined bicycle rider control during cycling on cycle lanes of different widths. Hence, one of our aims was to examine if the width of a cycle lane elicits changes in motor control during cycling. At the same time, the aim was to test a realistic range of cycle lane widths (widths observed in the UK).

1.06 Thesis objective and outline

The ultimate goal of this thesis is to explore bicycle rider control skills and examine the effects of different constraints on the control of a bicycle.

The approach to accomplish this goal was two-fold with the following objectives:

1. Develop and validate the methodology used in studying bicycle rider control.
2. Investigate the effects of three different constraints on the bicycle rider control: 1) expertise, 2) body position and 3) environment. It was hypothesised that all three constraints affect bicycle rider control.

The work in this thesis is split into several chapters and sub-chapters with the summaries presented in this section[1].

- The methodology used in this work was firstly validated and tested for the reliability and is presented in **CHAPTER 2** with the following sub-sections:
    - A study presented in **section 2.01** aimed to examine the validity of different kinematics methods used for bike fitting. Participants rode a bicycle at different seat heights and their movement was recorded with four different motion capture systems.
    - The aim of the experiment presented in **section 2.02** was to determine the minimum number of cycles required need to form a valid ensembled average of the EMG activity. Furthermore, the aim

[1] The reader should note that all experimental chapters presented in this thesis are identical as published or submitted in the aforementioned scientific publications.

was also to determine how many crank detection points are required to rescale the EMG pattern. Participants' EMG activity of the lower extremity was recorded and analysed differently.

  - The technical note presented in **section 2.03** aimed to test the inter-session reliability of a single angular motion sensor used for assessment of steering and roll rate. Participants repeated the same testing session twice.

- The experiment presented in **CHAPTER 3** examined how more experienced cyclists control the bicycle compared to less experienced cyclists. It also aimed to define collective variables to describe bicycle rider control.
- The study in **CHAPTER 4** aimed to define an optimal seat height to improve bicycle rider control. Cyclists rode at different seat heights, while bicycle rider control was recorded.
- **CHAPTER 5** presents a study looking at the effects of cycle lane design on bicycle rider control. Using the validated single angular motion sensor, different cycle lane widths and infrastructures were examined.
- A general discussion is presented in **CHAPTER 6**.

CHAPTER 2. METHODOLOGY

2.01 Validity and Reliability of Different Kinematics Methods Used for Bike Fitting[1]

2.01.1 Abstract

The most common bike fitting method to set the seat height is based on the knee angle when the pedal is in its lowest position, i.e. bottom dead centre (BDC). However, there is no consensus on what method should be used to measure the knee angle. Therefore, the first aim of this study was to compare three dynamical methods to each other and against a static method. The second aim was to test the intra-session reliability of the knee angle at BDC measured by dynamic methods. Eleven cyclists performed five 3-minute cycling trials; three at different seat heights (25, 30 and 35° knee angle at BDC according to static measure) and two at preferred seat height. Thirteen infrared cameras (3D), a high-speed camera (2D), and an electrogoniometer were used to measure the knee angle during pedalling, when the pedal was at the BDC. Compared to 3D kinematics, all other methods statistically significantly underestimated the knee angle ($p = 0.00$; $\eta 2 = 0.73$). All three dynamic methods have been found to be substantially different compared to the static measure (effect sizes between 0.4 and 0.6). All dynamic methods achieved good intra-session reliability. 2D kinematics is a valid tool for knee angle assessment during bike fitting. However, for higher precision, one should use correction factor by adding 2.2° to the measured value.

[1] This chapter is published as: Fonda B, Sarabon N, Li F-X. Validity and reliability of different kinematics methods used for bike fitting. J Sport Sci; 2014;32(10):940–6

2.01.2 Introduction

Bike fitting is an important process to adjust the geometry of the bike to the needs of the cyclist. Optimal bicycle rider position may be considered as a position in which force application and comfort are maximised, whilst resistive forces and risk of injury are minimised, in order to maximise bicycle velocity (Iriberri, Muriel, & Larrazabal, 2008). The first scientific papers on bike fitting were published in the mid-sixties (Hamley & Thomas, 1967), in which, authors proposed an anthropometric-based method to set the seat height. Subsequently alterations in body position and its effect on the variables mentioned above were investigated. Based on these studies, numerous methodologies have been proposed to perform bike fitting (Burke, 2003; Holmes, Pruitt, & Whalen, 1994; Iriberri et al., 2008; Nordeen-Snyder, 1977).

Seat height is probably the most important parameter set in the procedure of bike fitting. Improper seat height can result in over-compression of the knee (Ericson & Nisell, 1987) and/or increased oxygen consumption (Nordeen-Snyder, 1977; Price & Donne, 1997). On the other hand, recent research suggests that changes in seat height within 4% of trochanter height do not affect cycling economy (Connick & Li, 2013). To avoid detrimental effects, various methods to set the seat height have been proposed. The Hamely and Thomas method (Hamley & Thomas, 1967) defines the optimal seat height (seat height defined as the distance between the pedal axle and top of the seat) at 109% of inseam or, as recently revised, at 108.6–110.4% of inseam (Ferrer-Roca, Roig, Galilea, & García-López, 2012). LeMond and Gordis (1990) suggested to set the seat height (seat height defined as the distance between the centre of the bottom bracket and top of the seat) at 88.3% of inseam. The heel method determines the seat height by placing the heel of the foot on the pedal and incompletely extending the knee. The pedal must be at the bottom of the stroke (crank angle 180°) and the cyclist must be sat on the seat (Burke, 1994). This method is not precise as the different cleat pedals have different heights and it is difficult to determine exactly where the heel is placed. To minimize the risk of injury, Holmes et al. (1994) proposed to set the seat height to the level where knee angle, when the pedal is in its lowest position, i.e. bottom dead centre (BDC), is between 25° and 35° (Figure 2.1). It has been shown that the aforementioned static methods to set the seat height vary significantly among each other and do not always yield the same results (Peveler, Bishop, Smith, Richardson, & Whitehorn, 2005). It is generally agreed, that dynamical methods provide more realistic results than static methods (Ferrer-Roca et al., 2012; Peveler, Shew, Johnson, & Palmer, 2012). A study by Ferrer-Roca et al. (2012) compared a static (anthropometric measurements) vs. a dynamic method (2D kinematics) to adjust the seat height and they concluded that seat height adjusted with static method (106–109% of inseam length) was outside of the recommended range in 56.5% of the participants. Therefore, in order to set the seat height according to knee angle, direct measurements of knee angles should be adopted instead of equations based on anthropometric data (Ferrer-Roca et al., 2012).

Figure 2.1: Knee angle measurement when pedal is in the bottom dead centre (BDC).

Based on the bike fitting recommendations, numerous studies on biomechanics of cycling at different seat heights used knee angle as standardization of the seat height (Bini, 2012; Bini, Hume, & Crofta, 2011; Bini, Hume, & Kilding, 2012; Peveler, Pounders, & Bishop, 2007). Knee angle in static conditions is usually measured with a manual goniometer, where the axis is centred to lateral femoral condyle, one arm pointing upward to the greater trochanter and the other arm pointing downward to the lateral malleolus of the ankle. Recently, several commercially available bike fitting systems are using knee angle as a parameter based on which the seat height is set.

Different kinematics systems do not necessary provide the same results as each system has different drawbacks and advantages (Vlasic et al., 2007). Umberger and Martin (2001) examined the differences in joint angles during cycling when analysed in 3D and 2D, and reported that no significant differences exist between the two techniques. They used the same setup and cameras for both analyses, but recruited only 4 participants and found no statistically significant differences. Further research should be carried out in this direction to examine if there are any differences between 2D and 3D method.

Many kinematic systems have been tested for their validity and/or within- and between-session reliability in other situations involving cyclic movements rather than during cycling. Studies on 3D kinematics during gait reported moderate to good within-session reliability of joint angles in the sagittal plane (McGinley, Baker, Wolfe, & Morris, 2009). Cyclic movements, such as cycling or gait, are assumed to have good within-session reliability in kinematics parameters. For gait, studies showed that an average of 5 trials achieved the level of confidence above 90% (Diss, 2001) for all kinematic patterns. To the authors' best knowledge, there is no published data on the within-session reliability of kinematics data during cycling.

Knee angle has been well defined in relation to seat height. However, there is a lack of systematic comparison of the methods used to assess knee angle. Therefore, the first aim of this study was to compare three dynamical methods for knee angle measurement to each other and against the static measure. The second aim was to test the intra-session reliability of these methods. Based on our pilot studies and practical experiences, our hypotheses were that 1) 2D kinematics will significantly underestimate the knee angle compared to 3D kinematics, 2) manual and electronic goniometers will not provide a valid knee angle measurements in BDC compared to 3D kinematics, and 3) all three dynamic methods will achieve excellent intra-session reliability.

2.01.3 Methods

Participants

According to a priori sample size calculations based on our pilot data, eleven participants ([mean ± SD] age 23.3 ± 2.8 years, body mass 71.6 ± 6.9 kg, body height 179.8 ± 6.1 cm) were recruited from the University's cycling club. 6 elite and 5 recreational cyclists reached ([mean ± SD; min-max] peak oxygen consumption of 60.0 ± 7.7; 48-68 ml/kg/min and maximal aerobic power of 342.6 ± 37.3; 300-420 W). Cyclists were training between 5 and 20 hours per week. Before the experiment each participant signed an informed consent form, which was approved by the local ethical committee.

Protocol

Participants were required to visit the laboratory on two occasions. The first involved an incremental test to exhaustion on an electro-magnetically braked cycle ergometer (Lode Excalibur Sport, Lode, Groningen, NL) in order to determine maximal aerobic power. Gas analysis was constantly monitored with a breath-by-breath gas analyser (Jaeger Oxycon Pro, Erich Jaeger GmbH, Hoechberg, Germany). Participants started pedalling at 100 W at a self-selected cadence higher than 60 rpm. Resistance was increased by 25 W every 1 min, until the participant reached volitional exhaustion or cadence dropped below 60 rpm. Maximal aerobic power was noted as the highest power output at which pedalling was maintained for at least 30 s. This test was performed to standardize the intensity for the second session.

The second session was performed at least 48 h after the incremental test. After a warm up (5 min at 150 W, 80-90 rpm), each participant completed three trials at different seat heights, and two at their preferred seat height on an electro-magnetically braked cycle ergometer (Lode Excalibur Sport, Lode, Groningen, NL). All trials lasted 5 minutes and were carried out at 65 % of maximal aerobic power. Seat height was adjusted according to the knee angle when the pedal was at the BDC. Saddle heights corresponded to knee angle values of 25° (HIGH), 30° (MID) and 35° (LOW), measured with a standard manual goniometer in static conditions (Figure 1). Preferred seat height was adjusted according to the participant's bicycle setup. Participants were instructed to maintain a constant cadence (80 rpm) and to adapt a body position compared to real life conditions simulating long duration training. Knee angle measurements were performed three times for each seat height with 10 seconds of pedalling between the measurements. The handlebar position was adjusted according to the participant's individual bicycle setup, while the fore/aft position of the saddle was set according to bike fitting guidelines, where the patella is directly above the pedal spindle when the pedal is at a 90° position (Burke, 1994).

Setup

In the second test three systems were used to record simultaneously kinematics. 3D kinematics data were captured using a Vicon MX motion analysis system (Oxford Metrics Ltd., Oxford, UK) consisting of 13 cameras recording with a sampling rate of 250 Hz and which was calibrated with a residual error less than 1 mm. Retro-reflective markers were attached with double-sided adhesive tape over the greater trochanter, the lateral femoral condyle, and the lateral malleolus of the cyclist by the same tester to exclude inter-tester variability and. Furthermore, reflective markers were placed on the pedal spindle and crank centre of the bicycle ergometer to identify crank position. One high-speed camera (Casio Exilim Pro EX-F1, Dover, NJ, USA) recording with a sampling rate of 300 Hz, image resolution of 512x384 pixels and calibration resolution of 8 mm was positioned perpendicular to the participant at a distance of 4 m. An electrogoniometer (Biometrics Ltd., Newport, UK) operating at a sampling rate of 1000 Hz was attached with double sided adhesive tape to the right leg with one side attached to the middle of the shank (line between lateral malleolus and head of fibula) and the other to the middle of the thigh (line between lateral femoral condyle and greater trochanter). Synchronization between the Vicon system and electrogoniometer was established through an A/D card (National Instruments, Austin, TX, USA). The high-speed camera was not synchronised with the Vicon system and the electrogoniometer, but did record the same time period (± 3 seconds) as Vicon.

Data analysis

The first 15 cycles from the last 30 seconds of each trial were used for analysis. Analysis of the 3D kinematic data was performed using Matlab (MATLAB, MathWorks, Natick, MA, USA). 3D kinematic data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12 Hz. Knee angles for each trial were obtained by dividing data into individual crank cycles using the BDC pedal position determined as the point at which the pedal reflective marker reached its minimal vertical position, i.e. 180°. The knee angle from the 2D kinematic data was extracted (Kinovea 0.8.15) with the software’s function “angle” at a visually determined BDC crank position (with precision of 1.6°). Data from the goniometer was acquired and analysed with Labview (Labview, National Instruments, Austin, TX, USA). Crank angle from 3D kinematic data has been interpolated to 1000 Hz and merged with the knee angle data from the goniometer in order to calculate the knee angle when the pedal was in the BDC position.

Statistical analysis

For each trial and method, the average of 15 consecutive cycles was taken for further analysis from each kinematic method. All data is presented as a mean ± standard deviation. A two-way repeated measure ANOVA was used to test for method (3) x seat height (3) interaction and differences among methods at different seat heights. When sphericity was violated, Greenhouse-Geisser correction was applied. When the main effect was significant, *post hoc* comparisons with Bonferroni corrections were carried out. The two trials with preferred seat height were used to assess reliability. Differences between two trials were assessed using two-tailed paired t-test and 95% limits of agreement (Nevill, 1996; Nevill & Atkinson, 1997; Atkinson & Nevill, 1998). Statistical analysis was performed using SPSS V.20 (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$. Size effects ($\eta 2$) are reported as partial eta squared.

2.01.4 Results

Reliability is presented in Table 2.1. All dynamic methods showed good reliability with no statistically significant t-tests ($p = 0.37 – 0.51$) and absolute limits of agreement between 5.0 and 8.4. However inter-subject variability measured by SD is substantially larger to the electrogoniometer (13.45) than for the 2D (7.45) and 3D (7.8) kinematics.

Table 2.1: Intra-session reliability results for each method reported as mean ± SD for two trials (Trial 1 and Trial 2).

Method	Trial 1 (°)	Trial 2 (°)	Difference ± SD	Absolute Limits	p
2D camera	42.1 ± 7.4	43.8 ± 7.5	- 1.7 ± 4.3	- 1.7 ± 8.4	0.21
3D Vicon	42.9 ± 8.5	43.9 ± 6.7	- 1.1 ± 4.1	- 1.1 ± 8.0	0.41
Electrogoniometer	32.3 ± 22.3	32.0 ± 20.3	0.3 ± 2.5	0.3 ± 5.0	0.73

There was a significant main effect of methods at different seat heights ($F(1.039, 10.389) = 26.113$; $p = 0.000$; $\eta^2 = 0.731$) and a statistically significant interaction between Methods and Seat Height ($F(4, 40) = 4.449$; $p = 0.005$; $\eta^2 = 0.681$). Post hoc comparisons revealed that electrogoniometer substantially underestimated the knee angle and showed significantly smaller angles (ES = 0.7 – 0.8) compared to 2D and 3D kinematic methods at all seat heights (Figure 2.2). 2D significantly underestimated knee angle at the HIGH seat height ($p = 0.019$; ES = 0.2) compared to the 3D methods.

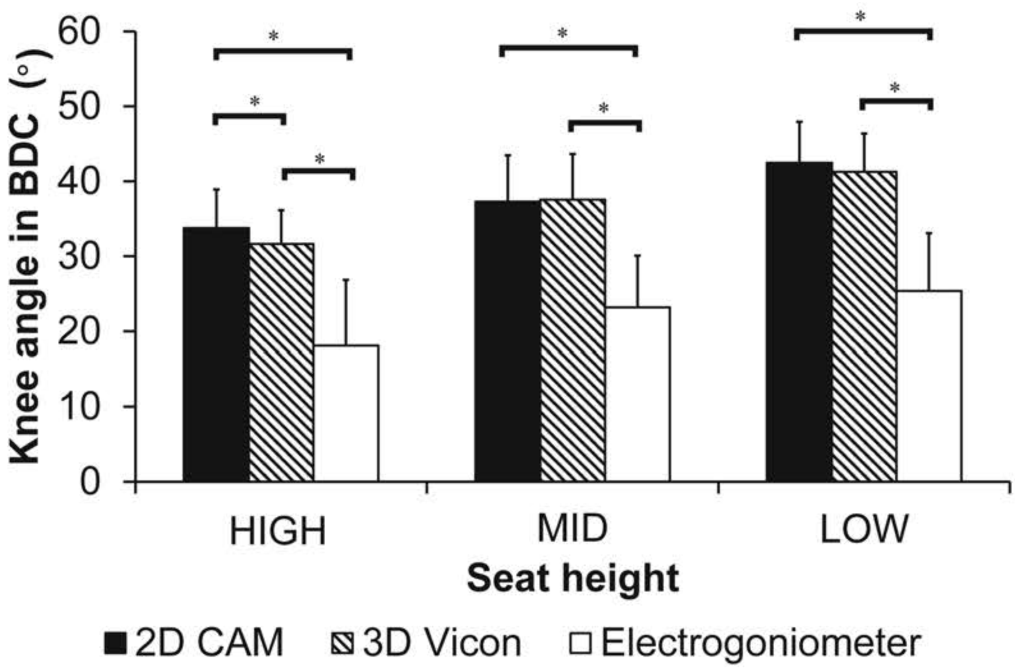


Figure 2.2: Knee angle values when the pedal is in the bottom dead centre (BDC) measured with a high-speed camera (black columns), Vicon system (dashed column) and electric goniometer (white columns). Dashed horizontal lines represent the angle based on which the seat height has been set with the static goniometer (ie. HIGH, MID, and LOW at 25°, 30°, and 35°, respectively) * = Bonferroni post hoc ($p < 0.05$).

All knee angle data measured with dynamic methods were different compared to static measures (between 15.2% and 38.2% and effect sizes between 0.4 and 0.6) which were used to set the seat height. The static measure underestimated the knee angle compared to 3D and 2D kinematics, while overestimating the knee angle compared to the electrogoniometer.

2.01.5 Discussion

The aim of this study was to compare three dynamical methods to assess knee angle during cycling at different seat height among each other and against a static method using a manual goniometer. We have confirmed our hypotheses by showing that 1) an electrogoniometer and 2D kinematics significantly underestimated the knee angle when compared to 3D kinematics, 2) a manual goniometer underestimated the knee angle at the BDC compared to 3D and 2D kinematics, and 3) all three dynamic methods achieved high intra-session reliability.

Various motion capturing systems do not necessarily provide the same results as they work on different basis (Vlasic et al., 2007), which has been partially confirmed in the present study. We have shown that knee angle at specific point assessed with two different passive kinematics systems and an electrogoniometer are substantially different. This is of practical importance when adjusting body position/technique. Overall for human motion analyses, kinematic systems that monitor movement in 3D space are more precise in assessing movement kinematics compared to 2D systems (Couto et al., 2008) and are therefore considered as gold standard. Moreover, a 2D method cannot be used to determine the external or internal rotations of the segments because this movement occurs in the transverse plane. Given the fact that pedalling is predominantly performed in the sagittal plane, 2D kinematics should provide the same results as 3D. Using a small sample size (n=4), a change of ~3° in knee angle, analysed with 2D and 3D kinematics using the same camera setup, confirms that 2D kinematics is valid for knee angle measurements during cycling (Umberger & Martin, 2001). In our study, two independent systems were used for 2D and 3D kinematics and were analysed using different software. Even though it resulted in significantly different knee angles at BDC, the difference was very similar (2.2°) to the one found by Umberger & Martin (2001). However, to achieve a higher level of precision of knee angle measurement with 2D kinematics, one should use a correction factor of 2.2° when assessing at higher seat heights.

Electrogoniometers are being regularly used in clinical practice and have been found to provide accurate and reliable measure of knee angles when a standardised protocol is used (Piriyaprasarth, Morris, Winter, & Bialocerkowski, 2008; Rowe, Myles, Hillmann, & Hazlewood, 2001). The disadvantage of the electrogoniometer is its susceptibility to skin movement artefacts as it is attached to a larger skin area compared to retro-reflective markers, which were to a certain extent susceptible for skin movement artefacts as well (Benoit et al., 2006). Another potential source of error is misalignment of the electrogoniometer to the anatomical axis of the knee joint, leading to difficulties in determining the zero position (Kettelkamp, Johnson, Smidt, Chao, & Walker, 1970). On the other hand, electrogoniometers provide immediate feedback on the measured knee angle, which is advantageous for bike fitting experts. It has been suggested (Petushek et al., 2012) that the electrogoniometer is a cost-effective and time-efficient alternative to video analysis for the assessment of knee flexion angle if the error is accounted for and the sensor is precisely attached. To the authors' knowledge, studies using electrogoniometers to measure knee angle during cycling have not been previously published. We have observed that electrogoniometers significantly underestimated the knee angle when the pedal was in the BDC, compared to 3D and 2D kinematics. Underestimation of the knee angle measured with electrogoniometer and large between-subject variation could be due to the difficulties of joint axis alignment and establishing zero offset, in line with the study of Kettelkamp et al. (1970). Future research should focus on standardizing the protocol for determining zero position in order to use electrogoniometers for cycling analysis.

2D analysis with high-speed cameras has been frequently used in human movement analysis even after the introduction of 3D systems. In fact, the vast majority of cycling studies on bike fitting have used 2D kinematics with a camera (Ericson, 1986; Nordeen-Snyder, 1977). The disadvantage of 2D kinematics is the phenomenon of parallax error, which occurs when objects are viewed away from the optical axis of the camera. Our results indicate that using a high-speed camera and open source software for *post-hoc* analysis without a correction factor do not provide the same results on knee angle measurement compared to Vicon 3D kinematic systems. We can assume that commercial software would not make any difference. Our results are similar to the finding by Umberger and Martin (2001) who found a difference of 3° between 2D and 3D method for knee joint angle measurement during cycling. In the present study two independent systems have been compared, whereas Umberger and Martin (2001) used the same cameras and setup. Another noteworthy observation is that in our study the differences between 2D and 3D results were not constant at all three seat heights. This could be the consequence of a fixed camera position for all three trials, which is normally the case in bike fitting setup. Our results indicate that 3D kinematics systems should be used for exact knee angle assessment (e.g. research), but 2D kinematics could be used for bike fittings. It is worth noting that despite observing significant differences between 2D and 3D analyses, recent research suggests that changes of 10° in a knee flexion angle do not substantially affect physiological and biomechanical markers (Connick & Li, 2013; Bini et al., 2012).

Bike fitting based on knee angle in static conditions is the most popular method among experts as it is the cheapest and the easiest method to use (de Vey Mestdagh, 1998). For these purposes bike fitters normally use manual goniometer. This method has several disadvantages, such as misalignment of the goniometer, dropping or raising the heel and absence of inertial momentum at the ipsilateral limb and force action in the contralateral limb. All these issues could potentially affect the knee angle. In our study we observed substantial underestimation of the knee angle measured with a manual goniometer compared to 3D kinematics and also other methods used in this study (2D and electrogoniometer). Furthermore, the large inter-subject variability observed for electrogoniometer suggests that it should be discouraged for bike fitting.

The reliability and validity of knee angle measurements should be known in order to be used appropriately. Intra-session reliability of knee angle measurement of all three methods in our study has been found to be excellent (ICC > 0.94). This concurs with the data on gait, where the highest reliability indices occurred in the hip and knee in the sagittal plane (McGinley et al., 2009). Future studies should be focused on inter-session reliability as it is critical for any intervention using bike fitting.

2.01.6 Conclusions

Each method has certain advantages and disadvantages, but all three methods tested in this study have shown high reliability. Experts should use 3D kinematics systems for knee angle assessment during cycling for the purpose of research, as this is the most valid way of knee angle assessment. Bike fitting experts using a high-speed camera for bike fitting, should make sure that the camera is positioned parallel to the captured motion of the cyclist. Additionally, to reach higher precision by using a high-speed camera, one should employ a correction factor by adding 2.2° to the measured value for knee angle measurement. In spite of the fact that electrogoniometers provides immediate feedback, due to its large inter-subject variability they are not suitable for bike fitting. Static measure of knee angle with manual goniometers should be discouraged in bike fitting.

2.02 Validity of methods to assess muscle activity during steady state cycling: a technical note[1]

2.02.1 Abstract

Valid protocols for electromyography (EMG) analyses during cycling are essential when examining the effects of mechanical or physiological constraints. The first aim of this study was to examine if the time dependant muscle activity characteristics would be different when rescaling the ensembled EMG pattern according to one crank detection point or interpolated between 4 and 360 crank detection points. The second aim was to define the minimum number of cycles required to form a representative ensembled average. Seventeen cyclists performed three 3-min trials at 50, 65 and 80% of maximal aerobic power. EMG was recorded from the tibialis anterior, soleus, lateral gastrocnemius, vastus lateralis, vastus medialis, rectus femoris and biceps femoris. EMG patterns were rescaled according to 1, 4 or 360 crank detection points. Backwards data elimination process was used to define the minimum number of cycles. Results showed that a statistically significant main effect of different rescaling methods was present for the offset, but not for onset or the peak activity occurrence. The minimum number of cycles required to get a representative ensembled average was 70 cycles. In conclusion, valid EMG analyses can be achieved by rescaling muscle activity patterns according to a minimum of four equally spaced crank detection points with at least 70 cycles averaged.

[1]This chapter is submitted as: Fonda B, Sarabon N, Li F-X. Validity of methods to assess muscle activity during steady state cycling: a technical note. J Elctromyogr Kinesiol. Under review

2.02.2 Introduction

In cycling biomechanics, the magnitude and timing of the muscle activity assessed by means of surface electromyography (EMG) are two of the most commonly studied variables. Temporal characteristics of muscle activity are typically expressed on a normalised crank angle position scale (from 0° to 360°) of one complete crank cycle, where 0 and 360° represent the highest position of the pedal (i.e. top dead centre (TDC)). It has been suggested that a continuous measurement of the crank position should be used for valid time-normalisation (Hug & Dorel, 2009), but this is sometimes technically difficult to achieve (e.g. outdoor measurements). In such case, it is important to know the minimum number of crank position detection points that are required to rescale the recorded signals. Additionally, the minimum number of cycles needed to form a valid ensembled average has not yet been established for cycling. Both of these findings could have an effect on the time dependent parameters of muscle activation (Hug & Dorel, 2009; Hug, 2011). Time dependent characteristics of the EMG patterns are important when identifying relationships between muscle coordination as well as changes in the mechanical and metabolic output (e.g. body position, workload, cadence, etc.). In order to design time-efficient and valid protocols for EMG measurements during cycling, the abovementioned questions need to be addressed.

To determine where in the crank cycle muscle activity occurs, the crank position detection method needs to be synchronised with the EMG acquisition. Researchers sometimes use optical or magnetic sensors (Connick & Li, 2013) to detect crank position and these are often placed on specific locations, such as the TDC. For a valid muscle activity pattern representation, (Hug & Dorel, 2009) recommended using continuous mechanical measurement of the crank position (e.g. rotational encoder (Fernández-Peña, Lucertini, & Ditroilo, 2009) or potentiometers (Sanderson & Amoroso, 2009) as the velocity of the crank varies throughout the crank cycle (Gregor et al., 1991; Hull, Kautz, & Beard, 1991) and is the greatest at a crank position of 90° and lowest at the TDC and 180° (Hull et al., 1991). Variations in the crank angular velocity could therefore affect the time scale normalisation in EMG analyses if one were not to use a sufficient number of crank detection points. This could, in turn, affect the calculations of the EMG time dependent characteristics. It can be speculated that using four crank detection points positioned in the middle of each crank cycle phase would suffice for the EMG time normalisation. However, to the authors` knowledge, the effect of crank velocity variation during steady state cycling on EMG pattern representation has not yet been empirically examined.

Dorel et al. (2008) reported good reliability of temporal and magnitude characteristics in EMG activity of the leg muscles before and after a simulated cycling exercise. Furthermore, Jobson et al. (2013) confirmed their (Dorel et al., 2008) results by showing similar intra- and inter-session reliability of the temporal and magnitude characteristics of the muscle activity during cycling. The main ''power producing'' muscles (vastus lateralis and vastus medialis), which exhibit single burst activity and are primarily mono-articular, displayed consistently more reliable temporal characteristics than bi-articular muscles (Dorel et al., 2008; Jobson et al., 2013). Despite inter-subject variations, muscle activity has also been found to be reliable during a progressive pedalling exercise performed to exhaustion (Laplaud et al., 2006).

Although temporal and magnitude characteristics were found to be reliable during submaximal cycling (Jobson et al., 2013), they can still be affected by the variability of the measurements. To minimise the effects of variability and improve signal-noise ratio of the EMG pattern, a sufficient number of cycles need to be averaged (Bruce et al., 1977). In the current peer-reviewed scientific literature, researchers used a range from 5 (Laplaud et al., 2006) to 50 (Fonda et al., 2011) cycles to get an ensembled average of the muscle activity pattern during cycling. To date, there is no general agreement defining the required number of cycles needed to create an ensembled average for a representative muscle activity pattern during cycling.

To avoid temporal shifts of the EMG patterns due to non-constant pedal velocity throughout the pedal cycle, it is recommended to use continuous measurement of the crank position for EMG rescaling (Hug & Dorel, 2009), however the minimum number of crank detection points for EMG rescaling to get a representative pattern is yet to be examined. Furthermore, the required number of cycles to form an ensembled average during steady state cycling has not yet been determined based on empirically supported data. Therefore, the first aim of this study was to test if methods using different number of crank detection points affect the calculation of the time dependent characteristics of the EMG patterns. The second aim was to establish the minimum number of cycles that need to be averaged to get a reliable muscle activity pattern during cycling. Based on the literature, we hypothesised that the time dependent characteristics of the EMG pattern would differ between using 1, 4 and 360 crank position detection points. Furthermore, based on the observations from our previous studies, we hypothesised that the required number of cycles to form a representative ensembled average will be over 50 cycles.

2.02.3 Methods

Participants

Seventeen recreational cyclists (13 male and 4 female) volunteered to take part in the study. Participants mean ± standard deviation age, height, body mass, and maximal aerobic power were 25.6 ± 6.1 years, 175.6 ± 7.4 cm, 67.7 ± 8.7 kg, and 342.6 ± 37.3 W respectively. The University's ethics committee approved the study. All participants signed an informed consent and were made fully aware of the purposes, protocols and procedures prior to taking part in the study.

Procedure

In the first session, participants completed an incremental test to volitional exhaustion to determine maximal aerobic power output (MAP). Prior to starting the incremental test, height and body mass were measured. For both sessions seat height was adjusted according to the participant's own bicycle setup. Participants then started cycling at 100 W for 3 min followed by incremental steps of 25 W every 1 min until either volitional exhaustion or cadence dropped below 60 rpm. MAP was noted as the highest power output at which pedalling was maintained for at least 30 s. This test was performed to standardise the intensity for the second session.

The second session was performed between 48 and 96 h after the incremental test. After the warm up (5 min at 150 W, self-selected cadence), each participant completed three trials at 50%, 65%, and 80% of their MAP. Participants were instructed to maintain a constant cadence (80 rpm) and to adapt their body position as they would in real life conditions.

Setup

Both test sessions were performed on the electro-magnetically braked cycle ergometer (Lode Excalibur Sport, Lode, Groningen, the Netherlands). In the second test, surface electromyography was recorded with single differential electrodes (DE-2.3, Delsys Inc., Boston, MA) which were placed on the skin over the belly of the tibialis anterior (TA), soleus (SO), gastrocnemius lateralis (GCL), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF) and the long head of biceps femoris (BF) according to the SENIAM standards (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). The skin at the site of each electrode site was shaved, abraded and cleaned with alcohol. To detect crank position, retro-reflective markers were placed on the pedal spindle and bottom bracket of the bicycle ergometer. These were then captured using a Vicon MX motion analysis system (Oxford Metrics Ltd., Oxford, England) with 4 cameras operating at a sampling rate of 250 Hz and calibrated with residual error less than 1 mm.

Muscle activity data capture was automatically synchronized with the kinematic data via a trigger with a 5-Volt rising edge pulse from the GPIO port of the Vicon MX system, and was recorded using a Delsys Myomonitor III EMG system (Delsys Inc., Boston, MA) at a sampling rate of 4000 Hz.

Data analysis

Kinematic data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12 Hz. Afterwards it was linearly interpolated to 4000 Hz to match the EMG data. Prior to this, each individual trial was checked for variations in crank angular velocity by comparing crank velocities for the two neighbouring samples (approximately every two degrees of the crank angle) with a t-test. All comparisons were found not to be statistically significant and therefore linear data interpolation was possible to reach the matching frequency. EMG data were band-pass filtered with cut-off frequencies of 20 and 450 Hz, zero aligned, full-wave rectified and further low-pass filtered using a second-order Butterworth low-pass filter with a cut-off frequency of 16 Hz to create a linear envelope.

The minimum number of cycles required to form a representative ensembled average was defined as the iteration in the data elimination process where the reduced muscle's ensembled average curve deviated from the 95% confidence interval of the original curve (ensembled average of 80 cycles) and remained outside for 3 consecutive occasions. Data elimination was performed backwards, meaning that the new ensembled average consisted of the first n cycles. This technique was previously successfully applied and validated in other tasks (Mathias, Barsi, van de Ruit, & Grey, 2014).

An average of 80 cycles (from 1 minute of recordings) for each muscle and each trial was assembled by rescaling the EMG patterns according to 360 (DP_{360}), four (DP_4) and one detection point (DP_1). Detection points were extrapolated from the kinematic data of the crank angle where 0°/ 360° represents the TDC. Onset and offset (threshold set at 20% of peak activity) and peak activity position were analysed on a rescaled EMG pattern according to 1 detection point (0°), interpolating between 4 detection points (0°, 90°. 180°, 270°) and between 360 detection points (every one degree of the crank angle). Differences between the three analyses using the different number of detection points were assessed separately for each intensity. Additionally, the area under the difference curve was calculated. This was determined by a subtraction of one signal from another. Three difference curves were calculated: $DIFF_1$ (difference between DP_{360} and DP_4), $DIFF_2$ (difference between DP_{360} and DP_1) and $DIFF_3$ (difference between DP_4 and DP_1). An example of calculating a difference curve is illustrated in Figure 2.3.

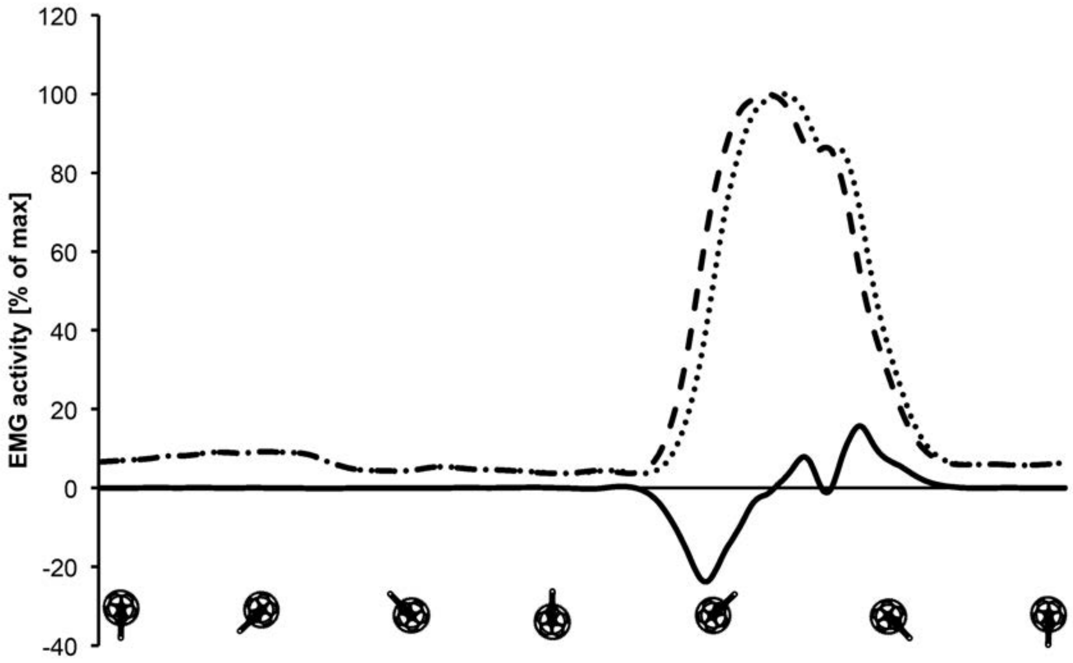


Figure 2.3: Graphical representation of the difference curve calculation. Dashed line represents EMG signal of one muscle when rescaled according to one detection point and the dotted line represents the EMG signal of the same muscle rescaled according to 360 detection points. Solid line is the difference curve of the two EMG signals.

Statistical analysis

All data are presented as a mean ± SD and were first tested for normality with a Shapiro-Wilk test. A 3-way mixed analysis of variance (ANOVA) (*muscle (7) x detection points (3) x intensity (3)*) with *muscle* as a between-subject and *detection points x intensity* as the within-subject factor) was used to test for differences in onset, offset, peak activity position and difference curves among analyses using a different number of detection points at different intensities and muscles. A 2-way mixed ANOVA *(muscle (7) x intensity (3))* with *muscle* as a between-subject and *intensity* as the within-subject factor) was used to test for differences in the required number of cycles among the muscles. When sphericity was violated, the Greenhouse-Geisser correction was applied. Post hoc pairwise comparisons were done with Bonferroni correction. Statistical analyses were performed using SPSS V.20 (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$. Size effects (η^2) are reported as partial eta squared.

2.02.4 Results

The EMG patterns rescaled according to 1, 4, and 360 crank detection points for each recorded muscle of one participant are illustrated in Figure 2.4.

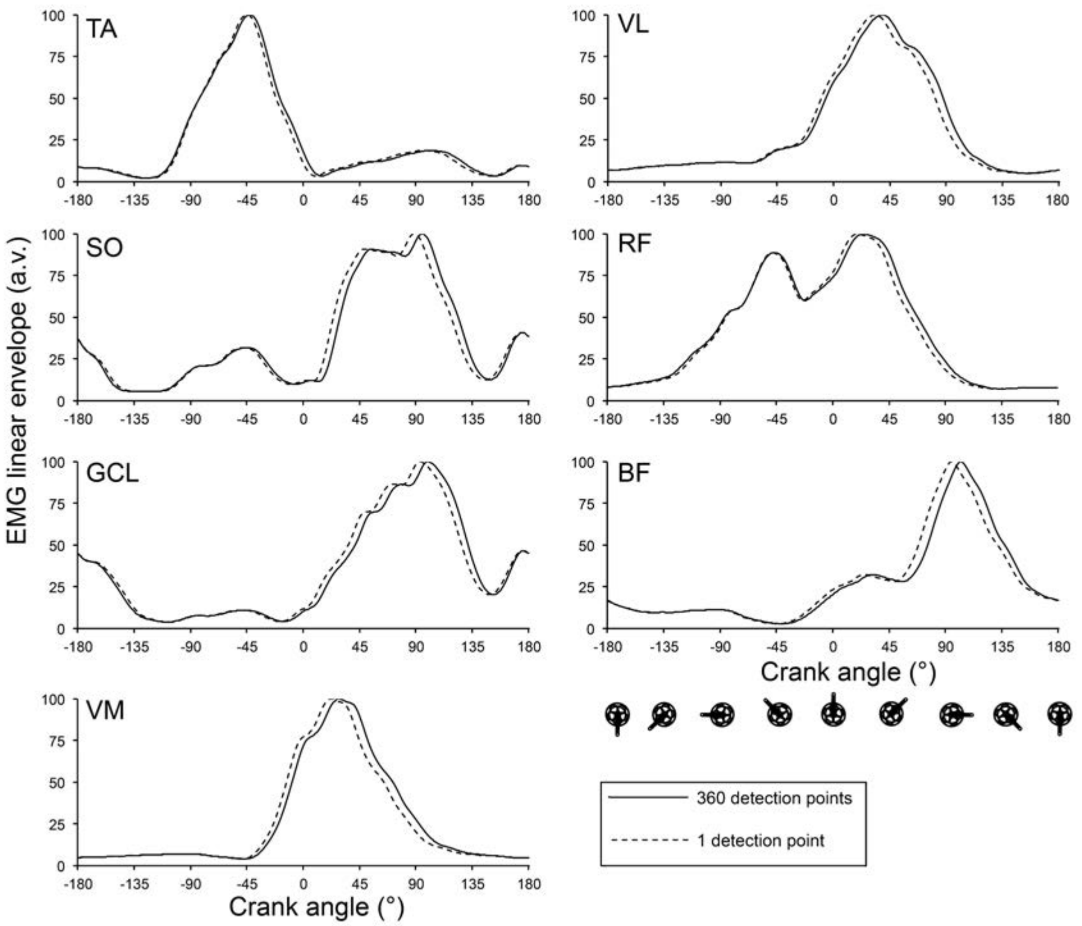


Figure 2.4: Graphical representation of EMG patterns rescaled according to 360 (solid line) and 1 (dashed line) crank detection points for one participant at 80% of maximal aerobic power. EMG signal rescaled according to 4 detection points has not been included as it almost perfectly fits the signal scaled according to 360 detection points. TA, tibialis anterior; SO, soleus; GCL, gastrocnemius lateralis; VM, vastus medialis; VL, vastus lateralis; RF, rectus femoris; BF, biceps femoris.

Figure 2.5 illustrates the mean values of the onset, offset and peak position respectively for the trial at 80% of MAP. The *muscle (7) x detection points (3) x intensity (3)* ANOVA interaction was not statistically significant for the onset ($F(24, 448) = 1.10$, $p = .340$, $\eta^2 = .056$), offset ($F(24, 448) = .97$, $p = .504$, $\eta^2 = .049$) or the peak activity position ($F(24, 448) = .91$, $p = .588$, $\eta^2 = .046$). In addition, a significant *main effect* of detection point was observed for the offset ($F(1.5, 164) = 5.95$, $p = .007$, $\eta^2 = .050$) but not for the onset ($F(1.4, 156) = 1.69$, $p = .196$, $\eta^2 = .015$) or peak position ($F(1.4, 153) = .52$, $p = .521$, $\eta^2 = .005$)). *Pairwise comparisons* for offset were completed to determine if differences exist between the number of detection points. They revealed that statistically significant differences lie between 360 and 1 detection point at 50 ($p < .001$), 65 ($p < .001$) and 80% ($p = .009$) of MAP. On the other hand, comparisons between the methods using 360 and 4 detection points showed statistically significant differences only at 65 ($p = .008$) and 80% ($p = .019$) of MAP.

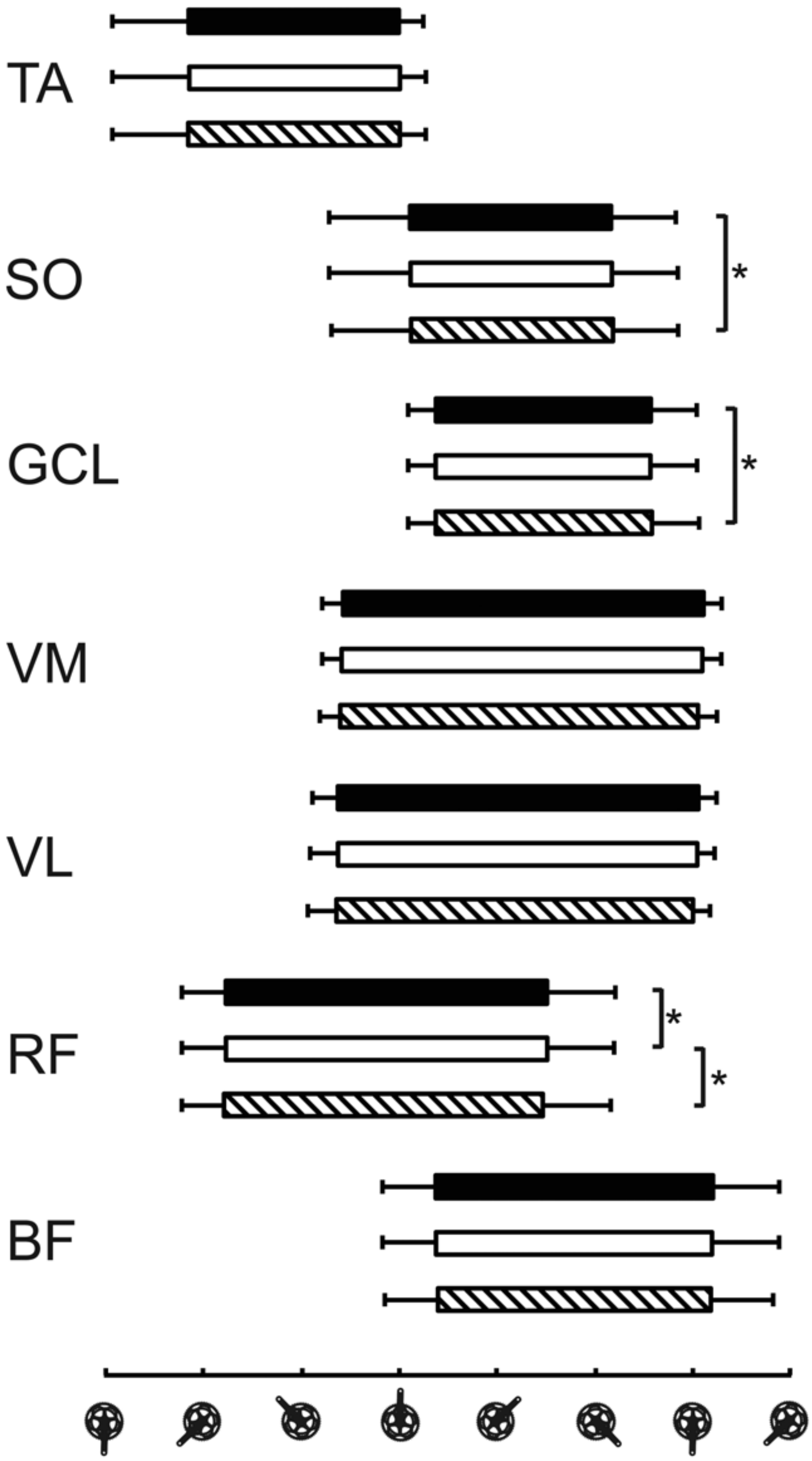


Figure 2.5: Mean and SD for onset (left part of the bars) and offset (right part of the bars) at 80% of maximal aerobic power from EMG patterns rescaled according to 1 (black), 4 (white) and 360 (striped) crank detection points. * indicates post-hoc differences with $p < 0.05$. TA, tibialis anterior; SO, soleus; GCL, gastrocnemius lateralis; VM, vastus medialis; VL, vastus lateralis; RF, rectus femoris; BF, biceps femoris.

The area under the subtraction curve was found to be statistically significantly different between $DIFF_1$, $DIFF_2$ and $DIFF_3$ ($F(2,224) = 195.7$; $p < .001$; $\eta^2 = .636$). A statistically significant interaction was observed for *muscle (7) x detection points (3)* ($F(12,224) = 2.49$; $p = .004$; $\eta^2 = .117$), but not for *intensity (3) x detection points (3)* ($F(4,448) = .87$; $p = .481$, $\eta^2 = .008$). *Post hoc* pairwise comparisons showed statistically significant differences between $DIFF_1$ and $DIFF_2$, and between $DIFF_1$ and $DIFF_3$ (Figure 2.6).

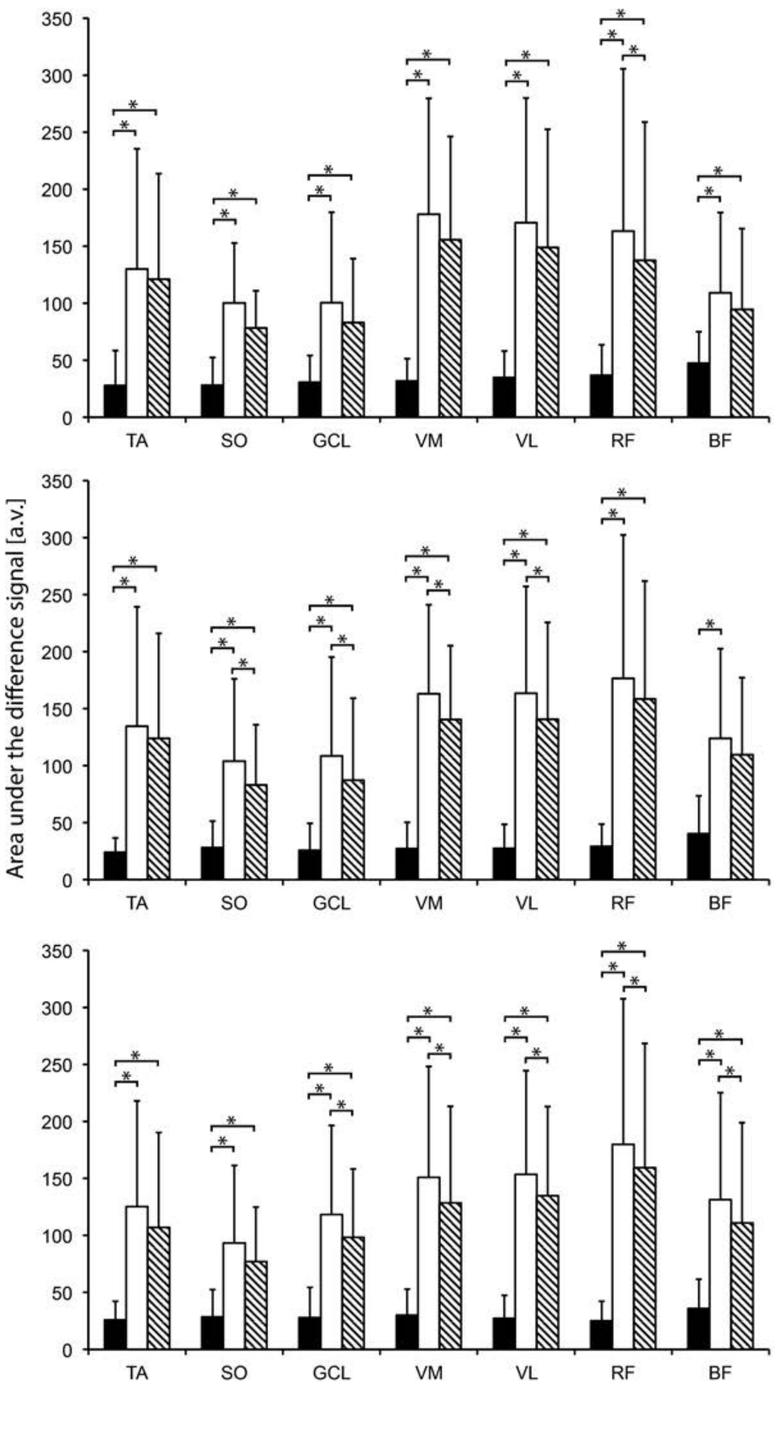


Figure 2.6: Areas under the difference signals for 50 % (top graph), 65 % (middle graph) and 80 % (bottom graph) of maximal aerobic power output. Area is calculated for the difference signal between 360 and 4 detection points (black bars), 360 and 1 detection point (white bars), and 4 and 1 detection point (stripped bars). * indicates post-hoc differences with $p < 0.05$. TA, tibialis anterior; SO, soleus; GCL, gastrocnemius lateralis; VM, vastus medialis; VL, vastus lateralis; RF, rectus femoris; BF, biceps femoris.

-

Mean values for the minimum number of cycles are illustrated in Figure 2.7. *The 7 x 3 ANOVA* revealed no significant *muscle x intensity* output interaction for the minimum number of cycles required to get a representative ensembled average ($F(12, 224) = 1.292$, $p = .22$, $\eta^2 = .065$). In addition, there were no statistically significant differences among the muscles ($F(6, 112) = 197$, $p = .977$, $\eta^2 = .010$) but there were between intensities ($F(2, 224) = 4.29$, $p = .015$, $\eta^2 = .037$). It was observed that significantly more cycles are necessary for the intensity at 80% (61 ± 2 cycles) compared to 65% (59 ± 2 cycles) and 50% (56 ± 2 cycles) of MAP.

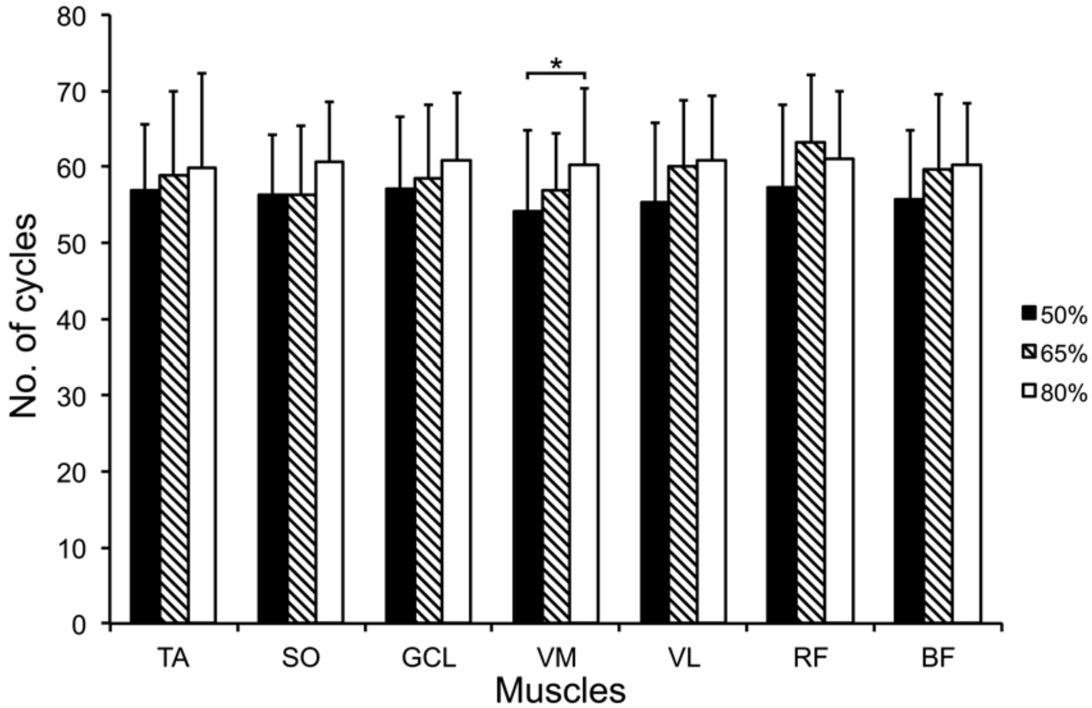


Figure 2.7: The minimum number of cycles for each muscle (mean and SD) at 50% (black), 65% (striped), and 80% (white) of MAP. * indicates post-hoc differences with p < 0.05. TA, tibialis anterior; SO, soleus; GCL, gastrocnemius lateralis; VM, vastus medialis; VL, vastus lateralis; RF, rectus femoris; BF, biceps femoris.

2.02.5 Discussion

The first aim of this study was to test if the number of crank position detection points elicits differences in the calculation of the temporal characteristics of muscle activity during steady state cycling. The second aim was to establish the minimum number of cycles that need to be averaged to get a reliable muscle activity pattern during cycling. Results showed that four crank detection points are sufficient to rescale EMG patterns if 70 cycles are used to form an ensembled average. The first hypothesis is supported as the number of detection points significantly affected the EMG pattern of all muscles when rescaled according to 1 crank detection point, but did not differ between 4 and 360 crank detection points. The required number of averaged cycles was 70, which confirms the second hypothesis. The minimum required number of cycles to form a valid ensembled average was not substantially different among the muscles.

Number of detection points

The results from the present study show that the majority of the muscle activity temporal parameters are not significantly affected when using four crank detection points for EMG rescaling compared to 360, but are significantly different when using only one crank detection point. Offset of the muscle activation significantly varied when comparing methods using 360 and one detection point. Differences were observed for offset of VL, VM, SO and GCL. These changes occurred in the propulsive or power phase, i.e. between 10° and 170° (Raymond et al., 2005). During this phase, total forces exerted on the pedal (Hull et al., 1991; Hull, Wootten, Boyd, & Hull, 1996) and the crank angular velocity (Gregor et al., 1991; Hull et al., 1991) are at their highest. These changes were not present when comparing 4 and 360 detection points. This could be due to the fact that the four detection points used in the present study were in the middle of each phase, minimising the temporal shift caused by non-constant crank angular velocity.

To analyse the overall shift in muscle activity, the difference curve method was used. This method takes into account the whole signal and not just specific time points of the signal, hence providing an overall representation of the difference between two similar signals. Overall, there was no difference in temporal characteristics of muscle activity when using four crank detections points compared to continuous crank position detection. This was shown by a significantly lower difference curve between the signal rescaled according to 4 and 360 crank detection points compared to the other difference curves. Signals rescaled according to four detections points compared to the signals from 360 detection points were very similar. Therefore, it can be concluded that four is the minimum number of crank detection points for EMG rescaling.

The findings partially support the recommendations provided in the review by Hug & Dorel (2009) who suggested synchronizing the EMG signal with a continuous mechanical measurement of the crank position and interpolating the EMG signal according to 360 detection points, particularly if there are variations in the crank velocity. This is especially true during cycling sprints or cycling with one leg where crank velocity varies to a higher extent than during steady state cycling. The crank angular velocity at a constant cadence varies (Gregor et al., 1991) and affects the temporal characteristics of the EMG pattern if it is linearly rescaled according to only one detection point. We support the recommendation from Hug & Dorel (2009) that, if possible, one should use a continuous measurement of the crank position but if that is not possible, at least four equally spaced crank detection points should be used.

Minimum number

A representative EMG profile is obtained by averaging the linear envelopes for a number of consecutive cycles or trials (Hug & Dorel, 2009; Shiavi, Frigo, & Pedotti, 1998). We have observed that the minimum number of cycles to get a representative ensembled average is approximately 70 cycles (mean + one standard deviation). This is substantially more than recommended for gait by Shiavi et al. (1998) who concluded that 6-10 cycles for cyclic movements is enough. It is also more than what was used in the majority of published papers examining muscle activity during cycling (e.g. Chapman, Vicenzino, Blanch, & Hodges (2007) and Dorel et al. (2008)).

Our findings do not completely support previous studies (Dorel et al., 2008; Jobson et al., 2013) that examined the reliability of the EMG activity during cycling, where they used smaller number of cycles to form an ensembled average. However, in their experiments, parameters focusing on specific time points of the muscle activity (e.g. onset, offset, peak occurrence, etc.) were analysed and not the pattern itself. On the other hand, analysis in the present experiment used the data elimination process where the reduced muscle's ensembled average curve deviated from the 95% confidence interval of the original curve. This technique was previously successfully applied and validated in brain stimulation research (Mathias et al., 2014) and is sensitive for temporal and magnitude changes of the EMG pattern.

An ensembled average with more cycles improves signal-noise ratio, becomes smoother and is more representative (Bruce et al., 1977). A valid and representative EMG pattern is essential to eliminate the variability when comparing muscle activation under different mechanical or physiological constraints, e.g. uphill versus flat terrain cycling (Sarabon, Fonda, & Markovic, 2012). If the number of averaged cycles is not sufficient, the differences could be due to the variability and/or noise, and are less reflective of the actual constraints one is facing. Recording and analysing more cycles during cycling activity is relatively simple and fast; at a cycling cadence of 80 rpms, less than a minute of recording is necessary. However, when intermittent sprints of short duration are being examined and only a few cycles can be taken into analyses, this may become more challenging (Billaut, Basset, & Falgairette, 2005). However, at higher intensities (e.g. all out sprints), other parameters of the EMG signal may be more important (e.g. frequency spectrum). Further research on this topic should focus on the validity of the methodological approaches of the EMG analyses.

Smoothness of the EMG pattern is affected by the cut-off frequency. In the present study, we only used one frequency (16 Hz). This is relatively high but was chosen based on visual inspection of the data and is similar to other studies that examined reliability of the EMG during cycling (e.g. Jobson et al. (2013)). Lower cut-off frequencies may, in turn, provide smoother signals and consequently require fewer cycles to form an ensembled average. However, they risk losing some parts of the muscle activity pattern. Agreement on the optimal cut-off frequency to smooth the signals remains to be examined.

2.02.6 Conclusion

The present study demonstrated two important findings about muscle activity analyses during steady state cycling. Firstly, we have shown that EMG rescaling can be based on four equally spaced crank detection points in a cycle. However, when possible, a continuous measurement of the crank should be used to provide even higher validity of the EMG pattern. Secondly, the required number of cycles to form an ensembled EMG average during cycling is approximately 70, which is more than previous studies have used. Further research is required to address the methodological approaches for EMG analyses during cycling under different physiological and mechanical constraints (e.g. all out sprints).

2.03 Bicycle rider control assessment: a reliability study using an angular rate sensor[1]

2.03.1 Abstract

Bicycle rider control is often described as a steering and roll motion. The rate of these two motions is linked with the level of bicycle control expertise, which was always examined in a laboratory setting. The aim of the present study was to test the reliability of steering and roll rate related parameters when cycling outdoors. We hypothesised that steering and roll rate parameters would be highly reliable. 11 participants completed two identical cycling test protocols on two separate occasions. The first protocol was riding in a straight line and the second was riding around the perimeter of a square. A single, 3-axis angular rate sensor was mounted on the bicycle's stem to record steering and roll rate. Parameters calculated were root mean square and standard deviation of steering and roll rate. Results showed good-to-excellent reliability of all calculated parameters (ICC > 0.78). The main finding of the present study was that cyclists perform steering and bicycle roll motions in a reliable manner when riding outdoors in a straight line or around the perimeter of a square. A 3-axis gyro sensor can, therefore, be used in future studies examining the effects of different constraints on bicycle control in outdoor conditions.

[1] [1]This chapter is submitted as: Fonda B, Sarabon N, Li F-X. Bicycle rider control assessment: a reliability study using an angular rate sensor. Sport Biomechanics. Under review

2.03.2 Introduction

A bicycle has only two contact points with the ground and in order to retain balance, support points need to be shifted to the falling side of the bicycle (Meijaard et al., 2007). During cycling, that is normally achieved by steering and bicycle roll motions (Kooijman et al., 2009). It has been demonstrated that more experienced riders perform these two motions at a smaller amplitude and rate. The rate of steering and bicycle roll motions were found to be the best at describing the level of bicycle rider control expertise (Cain, Ulrich, & Perkins, 2012; Fonda, Sarabon, & Li, 2015).

The skill of riding a bicycle is expected to improve with the amount of rider experience, which would concurrently change a cyclist's motor behaviour. In our previous experiments, we examined both more and less experienced cyclists' motor behaviour when cycling in a straight line using a camera-based 3D motion capturing system (Fonda et al., 2015). Less lateral movement of the bicycle in more experienced cyclists compared to less experienced cyclists was recorded. This means that more experienced cyclists can follow a pre-defined trajectory better than less experienced riders. Furthermore, experienced cyclists perform significantly less steering and roll of the bicycle at a significantly lower rate. The observed motor behaviour during cycling was also found to be highly repeatable. Based on these observations, it can be concluded that steering and roll related parameters could reliably describe the level of expertise in control of the bicycle.

However, all of our previous experiments were performed on a limited length (7 m at average speed of 14 km/h) and inside a lab facility, which may lack ecological validity (Fonda et al., 2015). Similarly, the majority of other studies (e.g. Kooijman et al. (2009)) that examined rider control during cycling were performed inside (e.g. using a treadmill), which affects perception of the environment. Perception is a known factor that affects the consequent motor action. For example, when walking through narrow apertures, people seem to rotate their shoulders to while passing through the aperture even if it is wide enough to safely pass through without collision (Warren & Whang, 1987). Thus, riding outdoors compared to inside a laboratory facility could have an effect on the control of a bicycle.

Cain & Perkins (2012) attempted to quantify bicycle rider control changes in outdoor conditions using three wireless inertial motion sensors mounted on the stem, the frame and the wheel to record steering rate, roll rate and speed respectively. They recorded children throughout the process of learning how to ride a bicycle during a 5-day training course. By analysing the steering and roll rate, changes in rider control and skill level were assessed. Their main finding revealed that peak cross-correlation coefficient between the steering and roll rate increases together with the skill. However, to the best of our knowledge, there are no studies published in peer-reviewed periodicals that would examine the reliability of steering and roll motions whilst riding in outdoor conditions.

It is clear that the main changes in bicycle rider control happen in the steering and roll of the bicycle, but so far only one study (Fonda et al., 2015) validated the test protocol in terms of the reliability of bicycle rider control by using a camera-based 3D motion capturing system (Vicon Nexus) in a laboratory setting on a limited length. Henceforth, the aim of the present study was to examine a between-day reliability of the steering and roll rate measured with a 3-axis angular rate sensor. Our hypothesis was that all steering and roll rate related parameters would be highly reliable.

2.03.3 Methods

Eleven participants (10 males, and 2 females, 33.1 ± 14.6 years, 76.0 ± 7.6 kg, 176.4 ± 7.6 cm) volunteered to take part in the study. Participants were not competitively involved in cycling but were, on average, riding a bicycle 3 times per week. Anthropometric data and information on the amount of bicycle riding was obtained through a questionnaire before the participation. Before the experiment, each participant signed an informed consent form, which was approved by the National medical ethics committee.

Participants were asked to attend two identical testing sessions, separated by at least 24 hours. Participants rode their own bike (a commuting or mountain bike) on a secured 50 x 50 metre car park, free from traffic or pedestrians. The first condition involved riding in a straight line for 40 m. Participants were instructed to ride as straight as possible towards the end point, which was clearly marked with a coloured adhesive tape. Speed was controlled by setting the gear ratio and verbally correcting the participants with instructions to go faster or slower during the practice trials. There was no other line on the ground to follow, just the end point. The second condition was riding on a 1-metre wide cycle lane around a 20 x 20 m square. The cycle lane was painted with white paint. Participants were instructed to stay in the middle of the right lane, riding counter-clockwise. Each condition was repeated 5 times.

A custom-built portable device with a local data logger and 3-axis gyro (LSM9DS0 IMU Breakout) was fixed to a custom-made mount on the bicycle's stem (Figure 2.8). All data were transferred through an Arduino board (Seeeduino Stalker v2, Seeed Technology Inc., Shenzhen, China) and stored on an SD card. The SD card was connected through an extension for easier removal and download of the data.

Figure 2.8: Box with a data logger and sensor to record steering and roll rate. On the surface there are two spirit levels for the initial setup to make the device as level as possible, a button to start and stop the recordings and an LED light indication, which is switched on during the recordings.

A push button was mounted on top of the device, which started and stopped the data recording. Participants were asked to press the button, stand still for three seconds, perform the trail and press the button again once they had stopped safely.

Before analysing the data, the roll rate and steering rate were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 5 Hz, which is comparable with previous studies (Cain & Perkins, 2012; Cain, 2013). Parameters calculated for the straight line condition are split into three sections. Firstly, parameters were calculated for the entire trial. Secondly, parameters were calculated only for the first three seconds and, lastly, parameters were calculated for the middle part (after 3 s of the start of the trial when riders acquired steady state speed and before the last 3 s). Detailed descriptions of all calculated variables are presented in Table 2.2. These variables were chosen to cover different aspects of bicycle control: the entire trial, the steady-state speed part and the initial part of the trial (start). Data analyses were performed with a bespoke script in MATLAB (The MathWorks, Inc., Natick, MA, USA).

Table 2.2: List of variables and its descriptions calculated from the recorded data.

Variable	Variable description
STE_{RMS}	Root mean square of the steer rate
STE_{SD}	Standard deviation of the steer rate
ROL_{RMS}	Root mean square of the roll rate
ROL_{SD}	Standard deviation of the roll rate
CC	Cross-correlation coefficient between the steer and roll rate

For each dependent variable a mean of five trials from each condition and each testing session were taken for further statistical analysis. Both absolute and relative components of the reliability analysis were computed using Absolute limits and Inter-class coefficient of correlation ($ICC_{(2,1)}$ respectively (Atkinson & Nevill, 1998). ICC was scored as follows: fair (0.40–0.59), moderate (0.60–0.74), good (0.75– .89) and excellent (> .90) (Fleiss, 1999). Furthermore, we calculated the absolute difference to compare the means from each visit. Statistical analyses were performed using MATLAB (The MathWorks, Inc., Natick, MA, USA).

2.03.4 Results

The reliability analysis results for the line conditions are presented in Table 2.3. Overall, all conditions achieved good-to-excellent reliability. When the whole trial was analysed, ICC reached values above 0.90 for all five parameters. When the trial was analysed in the first three seconds after the start, ICC was found to be between 0.78 and 0.90 for all five parameters. Good-to-excellent reliability was also observed in the middle section of the trial, with ICC between 0.79 and 0.95 for all five parameters.

Table 2.3: Reliability of the steer and roll rate results when riding in a straight line

Variable	Visit 1 (mean ± SD)	Visit 2 (mean ± SD)	Absolute Difference	Absolute Limits	ICC
Whole Trial					
STE_{RMS}	11.2 ± 2.42	12.0 ± 2.47	1.37	± 2.88	0.90
STE_{SD}	5.27 ± 0.87	5.01 ± 1.04	0.33	± 1.73	0.94
ROL_{RMS}	5.60 ± 1.04	5.20 ± 0.97	0.42	± 0.83	0.95
ROL_{SD}	10.3 ± 1.74	9.83 ± 1.98	0.75	± 1.73	0.94
CC	0.78 ± 0.14	0.72 ± 0.16	0.06	± 0.10	0.97
Start of the trial					
STE_{RMS}	15.8 ± 2.73	16.1 ± 2.93	1.72	± 4.36	0.82
STE_{SD}	15.1 ± 2.26	14.31 ± 3.25	1.96	± 4.64	0.78
ROL_{RMS}	7.40 ± 0.99	7.03 ± 1.38	0.77	± 1.70	0.85
ROL_{SD}	7.16 ± 0.89	6.84 ± 1.43	0.76	± 1.68	0.85
CC	0.83 ± 0.08	0.78 ± 0.13	0.06	± 0.12	0.90
Middle part of the trial					
STE_{RMS}	8.44 ± 2.93	9.22 ± 2.90	2.08	± 4.74	0.79
STE_{SD}	7.32 ± 2.31	6.57 ± 1.37	0.88	± 2.77	0.88
ROL_{RMS}	4.71 ± 1.41	4.16 ± 0.86	0.69	± 1.80	0.82
ROL_{SD}	4.32 ± 1.24	3.96 ± 0.75	0.44	± 1.34	0.88
CC	0.77 ± 0.14	0.68 ± 0.14	0.10	± 0.12	0.95

STE_{RMS}, root mean square of the steer rate; STE_{SD}, standard deviation of the steer rate; ROL_{RMS}, root mean square of the roll rate; ROL_{SD}, standard deviation of the roll rate; CC, normalized cross-correlation coefficient between the steer and roll rate.

The results of the reliability analysis for the square condition are presented in Table 2.4. All five parameters were found to have good or excellent reliability (ICC between 0.79 and 0.96).

Table 2.4: Reliability of the steer and roll rate results when riding around a perimeter of a square.

Variable	Visit 1 (mean ± SD)	Visit 2 (mean ± SD)	Absolute Difference	Absolute Limits	ICC
STE_{RMS}	22.8 ± 2.26	22.97 ± 1.98	0.99	± 2.82	0.87
STE_{SD}	10.5 ± 1.48	10.6 ± 1.39	0.42	± 1.05	0.96
ROL_{RMS}	10.7 ± 1.39	11.2 ± 1.75	0.78	± 2.59	0.79
ROL_{SD}	20.7 ± 1.68	21.1 ± 1.86	0.69	± 1.67	0.94
CC	0.65 ± 0.07	0.65 ± 0.06	0.03	± 0.08	0.94

STE_{RMS}, root mean square of the steer rate; STE_{SD}, standard deviation of the steer rate; ROL_{RMS}, root mean square of the roll rate; ROL_{SD}, standard deviation of the roll rate; CC, normalized cross-correlation coefficient between the steer and roll rate.

2.03.5 Discussion

The aim of the present study was to examine the between-day reliability of the steering and roll rate related parameters during outdoor cycling. The results support the hypotheses that both steering and roll rate parameters exhibit good-to-excellent between-day reliability. Hence, a testing protocol outdoors and an angular rate sensor mounted on a stem provide a reliable way to assess one aspect of bicycle rider control and can be used in future research.

Steering and roll rate related parameters have been frequently used to assess one of the most important aspects of bicycle rider control, i.e. steering and roll (Cain, 2013; Kooijman et al., 2009; Moore, 2009), but never tested for reliability when recorded outdoors. This is the first study to examine if a single, 3-axis gyro can reliably assess steering and roll rate at two different testing sessions. It must be noted that a single, angular rate sensor mounted on a stem cannot accurately measure roll but can provide a close approximation, especially when assessing changes in roll rate. In order to measure the actual roll motion, one should use a similar setup as Cain et al. (2012) with an additional angular rate sensor mounted on the frame.

In the present study, we used the same parameters as some previous studies (Cain et al., 2012; Fonda et al., 2015) to assess one aspect of bicycle rider control but, to the authors' knowledge, only our previous experiment tested the reliability of the measurement which was performed on a limited length and in a laboratory setting (Fonda et al., 2015). In the present study, reliability of the steering and roll rate parameters when riding around the perimeter of a square were measured and resulted in a good reliability. This condition is more similar to what Cain et al. (2012) had used in their study. Unlike cycling in a straight line, this condition provides some insight into the rider control when navigating a bicycle around corners. The main difference to the study by Cain et al. (2012) is that they allowed their participants to ride the bicycle freely, where as the present study used a controlled protocol. Although the ecological validity is increased in the former protocol, a higher between-subject variability may be observed as participants can ride at various speeds and trajectories and therefore cannot serve as a tool to compare the level of expertise.

Normalised peak cross-correlation coefficients between the steering and roll rate have been suggested as a measure of a rider control skill, with higher values describing better skills (Cain et al., 2012). However, despite our findings that this parameter achieved good reliability in both tasks used in the present study, we believe it is not a valid measure when comparing riders who already know how to ride a bicycle. Based on basic dynamics to maintain balance during cycling, a cyclist should always steer in the direction of the roll (Meijaard et al., 2007), which in turn results in a relatively high cross-correlation. In the present study, slightly smaller normalised peak cross-correlation coefficients than the study by Cain et al. (2012) was observed. A possible explanation could be the fact that only one angular rate sensor mounted on the stem was used, whereas Cain et al. (2012) used two sensors mounted on the stem and the frame respectively meaning that, in their case, they were able to measure the roll rate of the bicycle frame. Moreover, the riding protocol used in the present study was different to their study and therefore no direct comparisons are possible.

The main limitation of the present study was that the methodology was not tested for sensitivity to detect differences in expertise. However, based on the findings from our previous work (Fonda et al., 2015), it is known that cyclists' motor behaviour is consistent, but differentiates according to the cyclist's expertise (Cain, Ashton-Miller, & Perkins, 2013; Fonda et al., 2015). In the present study, similar values for steering rate were observed when compared to our previous work with a different recording methodology and a shorter duration of the task. That means that an angular motion sensor mounted on a bicycle's stem provides a valid way to measure steering rate. Furthermore, roll rate is also measured but, due to the location of the sensor, it is measured with a smaller accuracy. However, reliability is still good-to-excellent.

The main finding of the present study is that steering and roll rate related parameters are reliable when recorded on different testing sessions. Furthermore, a single 3-axis angular rate sensor mounted on the stem provides a valid way to assess steering rate and roll rate. This has practical implications for future research and development. However, the reader should keep in mind that to ensure accuracy, one should use a separate angular rate sensor mounted on the bicycle's frame to measure the actual roll.

CHAPTER 3. BICYCLE RIDER CONTROL SKILLS: EXPERTISE AND ASSESSMENT

3.01 Abstract[1]

Research on how human balance and control bicycles is inconclusive, largely due to the small number of participants in the previous studies. Therefore, the aim of this study was to test the hypotheses that 1) cycling lateral deviation amplitude will reliably show differences between more and less experienced cyclists and 2) more experienced will exhibit slower and smaller steering motions compared to the less experienced cyclists. Twenty eight experienced and inexperienced cyclists rode a bicycle in a straight line. Lateral deviation, steering and roll were measured. Inter-session reliability of the deviation was high with Cronbach's alpha values higher than 0.75. The amplitude, variability and rate of steering and roll parameters showed statistically significant differences between the groups. The test used in this study is sensitive to detect differences between more and less experienced cyclists and can be used for further research that aims to test the effect of a specific intervention addressing rider control. We also showed that steering and roll angle, which were described before as two of the main motor control actions in bicycle control differ in the variability, amplitude and rate between more and less experienced cyclists. Results of the present study have practical implications for improving bicycle rider control and increasing the safety of cyclists.

[1] This chapter is published as: Fonda B, Sarabon N, Li F-X. Bicycle Rider Control Skills: expertise and assessment. J Sport Sci. 2015;in press. DOI: 10.1080/02640414.2015.1039049

3.02 Introduction

A bicycle is laterally unstable at low speeds and stable, or easier to stabilize, at higher speeds (Kooijman et al., 2009). Bicycle rider control to maintain balance during cycling is important to prevent falling, crashing into obstacles and consequentially to prevent injury occurrence. According to Statistical Release by the Department for Transport, 19,000 cyclists are killed or seriously injured every year in the United Kingdom. Moreover, losing control is reported to be one of the reasons accounting for over 16 % of all the reported accidents (Lloyd, 2013). This clearly indicates that more research is necessary on bicycle rider control skills and interventions for improving them.

A bicycle is unstable as it only has two contact points with the ground and therefore acts as an inverted pendulum. In order to retain balance, support points need to be shifted to the falling side of the bicycle (Moore, 2009; Schwab & Meijaard, 2013). This can be achieved by steering, lateral movements of the knees, and to some extent through body leaning (Kooijman et al., 2009). Investigations on bicycle rider control (Moore et al., 2011) revealed that the most common body movements to maintain cycling in a straight line are bicycle steering, spine bend, rider lean, rider twist and knee movements. Even though some clear patterns have been observed, all aforementioned studies examined rider control on a small number of participants (maximum of three). This makes it difficult to generalise rider control skills for different types of cyclists (road, mountain bike, etc.) and skill levels.

In one of their experiments, (Kooijman et al., 2009) examined a town bicycle ride with motion sensors to gather information about situations one encounters during a normal bicycle ride. They observed that mainly steering motions were used when maintaining a straight line. In their follow up experiments, they looked at cyclists' (n=2) kinematic behaviour while riding in a straight line on a large treadmill. Their experiments were furthered by Moore et al. (2011) who demonstrated that lateral control is mainly accomplished by steering.

From a cognitive point of view riding a bicycle becomes a complex task that can be managed at three levels: strategic, manoeuvring and control level (Wierda & Brookhuis, 1991). Maintaining balance, speed and heading depends upon the control level. This means that more and less experienced cyclists will perform differently when cycling in traffic. More experienced cyclist will control the bicycle mainly through automatic action patterns, while a less experienced cyclist will actively control the bicycle. This was practically demonstrated by Wierda & Brookhuis (1991) who compared reaction times during riding a bicycle among participants of different age groups. They showed that younger (age 6-8 years), less experienced cyclists devote more attention to the primary task (riding a bicycle) and exhibit a longer reaction time compared to the older, more experienced cyclists. This could indicate that rider control of more experienced is less affected by another cognitive task.

It has been suggested that motor control changes when a cyclist progresses from less to more skilled (Cain, 2013). Wierda & Brookhuis (1991) showed higher standard deviation of the steering angle for younger cyclists compared to older ones. However, as the amount of steering during cycling depends on the speed and as the youngest group rode at speeds lower than their older peers, they concluded that these data are unreliable. The skill to control a bicycle is expected to increase in accordance with the power law of practice (Newell, Rosenbloom, & Anderson, 1981; Wierda & Brookhuis, 1991). Cain et al. (2013) demonstrated that the correlation between steer and roll angle is increasing with rider control skills. This suggests that more skilled riders steer in the direction of the roll whereas less skilled riders perform additional steering motions that are unrelated to the roll of the bicycle. Cain (2013) also examined the main differences in rider control of experienced versus inexperienced riders and concluded that experienced cyclists exhibit higher correlation between the lateral position of the center of pressure and center of mass compared to inexperienced cyclists. Experienced cyclists also employ more body lean control (independently from the roll of the bicycle), less steer control, and use less control effort than inexperienced riders. Expertise in movement control is often reflected in variability of the movement (Scott, Li, & Davids, 1997). To the authors' knowledge, no studies yet empirically examined this aspect during cycling.

Most of the research dealing with bicycle rider control skills examined riding at steady state speeds (e.g. Kooijman et al. (2009).). Our observations indoors and outdoors revealed that the motion amplitude (steer and roll) is substantially higher at the start (~ 3m) of riding a bicycle. Furthermore, this is also practically important when considering road safety as cyclists often have to stop and start (e.g. traffic lights) during their commute. Understanding bicycle rider control as a skill is a necessity for developing appropriate actions/interventions in order to improve control of a bicycle. The problem researchers encounter is that there is no objective and scientifically accepted test battery to assess rider control skills. Studies examining interventions (e.g. training) to improve rider control or constraints (e.g. school bag) that could affect rider control have mainly used errors on an obstacle course rather than measurements of rider control (Legg, Laurs, & Hedderley, 2003). Most common variables used in rider control evaluation are related to steer and roll of the bicycle; however, they have not been tested yet for reliability and validity (Cain et al., 2012) and are not clearly linked to the expertise of a cyclist. In order to better understand how experienced cyclists control their bicycles it is crucial to assess their skill level first. Therefore the aims of the present study are 1) to examine which variables are reliable and distinguish between more and less experienced cyclists and 2) how motor control during cycling differs between more and less experienced cyclists. The hypotheses are that 1) cycling deviation amplitude is a reliable test of cycling skill level and 2) all cyclists use similar type of control; however more experienced cyclists display less variability than less experienced cyclists.

3.03 Methods

3.03.1 Participants

According to a priori sample size calculations based on our pilot data, fourteen experienced cyclists (10 males, and 4 females, 24.1 ± 4.0 years, 70.1 ± 8.7 kg, 179.1 ± 8.3 cm) and fourteen inexperienced cyclists (8 males, and 6 females, 24.1 ± 6.6 years, 67.8 ± 15.2 kg, 170.7 ± 11.3 cm) volunteered to take part in this experiment. Before the experiment each participant signed an informed consent form, which was approved by the University's ethical committee. Information on amount of bicycle riding was obtained through a questionnaire before the participation. Experienced cyclists (experts) had over 5 years of cycling experience and were riding over 3000 km per year. Inexperienced cyclists (novices) had not ridden a bicycle for more than 100 km per year for the last 5 years and had no history of regular cycling activity. Similarity in the amount of cycling activity was also an inclusion criteria for participation in the study.

3.03.2 Protocol

Participants were asked to visit the laboratory on two occasions separated by at least 24 hours. Both sessions were identical. The test involved riding a commuting bicycle in the laboratory (30x10 m size) in a straight line in three different conditions. The first condition (Lane (Moving)) involved cycling in the centre of a cycle lane for 7 m with prior speed, achieving steady state speed while riding on the cycle lane. The second condition (Lane (Stationary)) consisted of cycling in the centre of a lane for 7 m with no prior speed, starting just before the cycle lane. The cycling lane was 60 cm wide and marked with a coloured adhesive tape. The third condition (Line) consisted of cycling for 7 m following a 3 cm wide line with prior speed, achieving steady state speed while riding on the line. Participants were instructed for Lane (Moving) and Lane (Stationary) to ride as straight as possible in the centre of the lane and for the Line condition to follow the line as precisely as possible. Centre of the lane was marked at the beginning and end of the lane. After the end of the lane, an area of 7 m provided enough space to stop the bicycle safely so participants did not have to anticipate stopping before the end of the lane. Speed was not specifically regulated but restricted by setting the gearing system to a ratio of 34x16, which at average cadence of 60 rpm resulted in a speed of 10 km/h. The same speed was also observed at the final one meter of the lane condition from a standing start.

Participants rode under each condition 10 times, with the first 5 being practice trials, which were not included in the analysis. Conditions were performed in a random order to avoid any learning or fatigue effect.

3.03.3 Setup

Two commuting bicycles (Transfer Hybrid and Excelle, Apollo Ltd., Victoria, Australia) with the same wheel size (29”), the same handlebar width and type, but different frame types and sizes were available to the participants. Both bicycles had the same tire type and were inflated to 400 kPa. Participants were assigned to one of the two bicycles based on their body height. For both sessions, seat height was set according to bicycle fitting recommendations using the heel method.

For kinematic recordings a 13 camera Vicon MX motion analysis system (Oxford Metrics Ltd., Oxford, UK) with a sampling rate of 200 Hz and calibration residual error less than 1 mm was used. Retro-reflective markers were attached with double-sided adhesive tape to the bicycle on a custom made frame fitted to the front of the bicycle and two on the stem (Figure 3.1). Extremities of the cycling lane were marked with markers prior to each testing to define the centre of the lane and the line.

Figure 3.1 Bicycle setup with all the markers. Two markers were positioned on the stem and three on the T-frame.

3.03.4 Variables

Deviation from the centre of the lane or line was calculated as an absolute mean of the deviation (DEV_{MEAN}), standard deviation of the deviation (DEV_{SD}), absolute maximal deviation (DEV_{MAX}) and range of the deviation (DEV_{RANGE}). Deviation was calculated as the distance of the front tyre contact point from the centre of the lane (or line). The contact point was calculated from the projection of two markers on the T-frame that were positioned directly above the contact point when the bicycle was completely upright. In case of a roll, the projection would still be towards the contact point. Steering parameters were defined as the mean absolute steering angle (STE_{MEAN}), standard deviation of the steering angle (STE_{SD}), range of the steering angle (STE_{RANGE}) and root mean square of steering angular velocity (STE_{RATE}). Bicycle roll was assessed as the mean absolute roll angle ($ROLL_{MEAN}$), standard deviation of the roll angle ($ROLL_{SD}$), range of the roll angle ($ROLL_{RANGE}$) and root mean square of the roll velocity ($ROLL_{RATE}$).

3.03.5 Statistics

The mean of the five trials in each condition was used for statistical analysis. A two-way mixed ANOVA (*condition (3) x group (2)*) with *condition* as a within-subject and *group* as a between-subject factor was applied to test for differences between the groups. *Post-hoc* comparisons between the groups in each condition were calculated with a Bonferroni correction. Intra-class coefficient of the correlation (ICC), Cronbach's alpha, absolute limits and the repeated measure ANOVA were used to assess the between-visit reliability of the deviation parameters. Statistical analyses were performed using SPSS V.20 for Mac (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$.

3.04 Results

3.04.1 Reliability

Reliability results are displayed in Table 3.1Table 3.1: Inter-session reliability results for the deviation parameters reported between two trials.. The most reliable variable was found to be $DEV_{SD,}$ with Cronbach's alpha above 0.77 and ICC above 0.73 in all three conditions. Smaller reliability was found for DEV_{MEAN} in the Line condition with Cronbach's alpha 0.55 and ICC 0.4. Lane (Stationary) had the most consistent reliability for all deviation variables with Cronbach's alpha above 0.76 and ICC above 0.65. The only statistically significant difference between the visits was found in DEV_{MEAN} in Lane (Stationary) condition and DEV_{RANGE} in the Line condition.

Table 3.1: Inter-session reliability results for the deviation parameters reported between two trials.

Variable	Difference (cm) (mean (SD))	Difference (Coefficient of variation)	Absolute limits	Cronbach's alpha	ICC	F-test
Deviation Absolute Mean						
Lane (Stationary)	.806(1.413)	1.75	± 2.77	.769	.673	.010*
Lane (Moving)	.085(1.42)	17.7	± 2.79	.712	.711	.589
Line	.738(2.978)	4.03	± 5.84	.551	.400	.236
Deviation SD						
Lane (Stationary)	.299(1.34)	4.48	± 2.63	.788	.734	.287
Lane (Moving)	.078(1.12)	14.4	± 2.20	.773	.777	.737
Line	.353(.942)	2.67	± 1.85	.903	.867	.079
Deviation Range						
Lane (Stationary)	1.00(4.28)	4.28	± 8.39	.817	.657	.339
Lane (Moving)	.093(3.74)	40.2	± 7.34	.744	.673	.795
Line	1.59(3.07)	1.93	± 6.02	.899	.459	.018*
Deviation Max						
Lane (Stationary)	1.54(3.08)	2.00	± 6.03	.824	.723	.054
Lane (Moving)	.013(2.63)	202	± 5.16	.695	.703	.974
Line	1.39(4.52)	3.25	± 8.85	.712	.577	.143

3.04.2 Deviation

When compared to the experienced, inexperienced cyclists had statistically significantly higher DEV_{MEAN}, DEV_{SD}, DEV_{MAX} and DEV_{RANGE} ($F(1,26) = 10.26$; $p = 0.004$; $\eta^2 = 0.283$, $F(1,26) = 11.17$; $p = 0.003$; $\eta^2 = 0.304$, $F(1,26) = 9.91$; $p =$

0.004; η^2 = 0.276 and F(1,26) = 11.37; p = 0.002; η^2 = 0.304, respectively). Furthermore, *post hoc* pairwise comparisons revealed that there were statistically significant between-group differences for all parameters under all conditions except for DEV_{MAX} in Line condition (Figure 3.2).

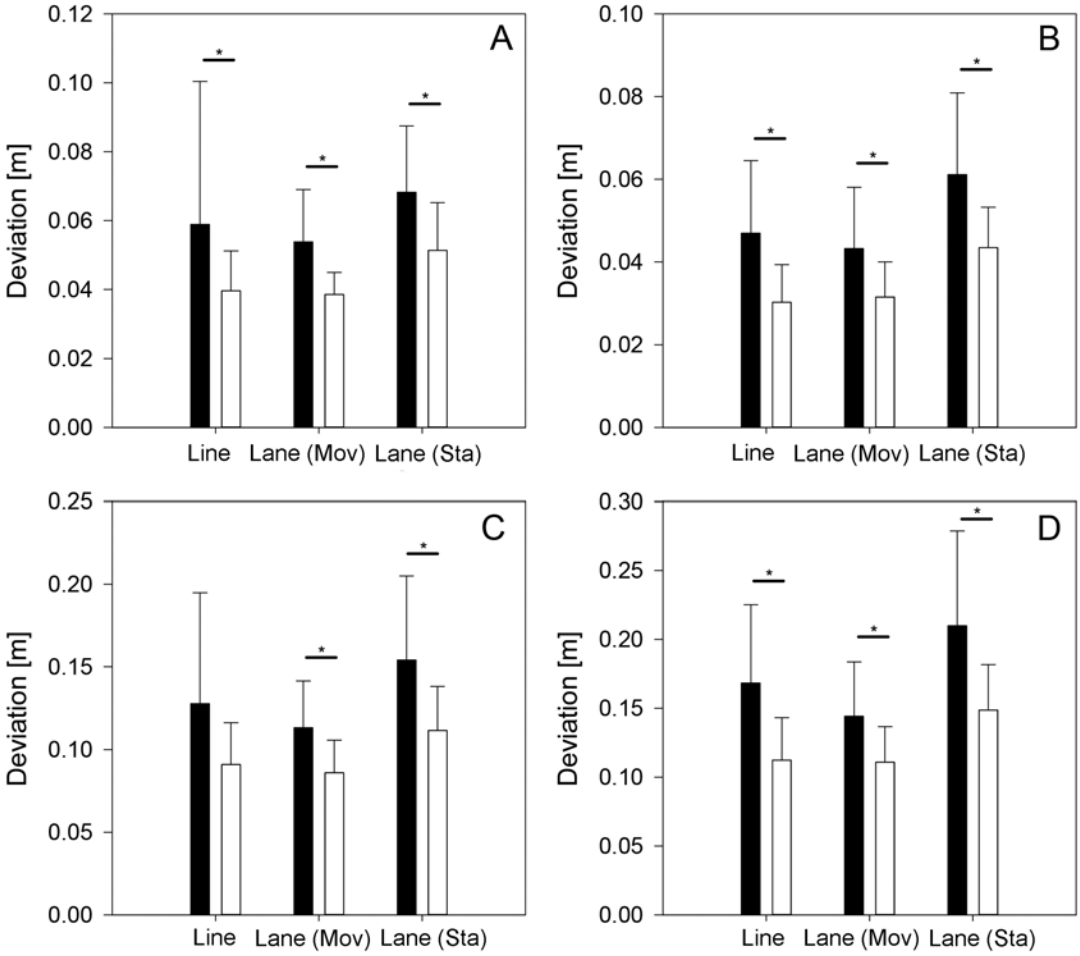


Figure 3.2: Absolute mean deviation (A), standard deviation of the deviation (B), absolute maximum of the deviation (C) and range of the deviation (D) for novice (black bars) and expert (white bars) cyclists at three different conditions. Line, following a line with prior speed; Lane (Mov), in the middle of a cycle lane with prior speed; Lane (Sta), in the middle of the cycle lane from a standing start; *, p < 0.05 after the Bonferroni correction.

3.04.3 Roll and Steering

Steering parameters are illustrated in Figure 3.3. 2-way ANOVA revealed statistically significant between-group differences for STE_{SD}, STE_{RANGE} and STE_{RATE} ($F(1,26) = 10.21$; $p = 0.004$; $\eta^2 = 0.282$, $F(1,26) = 12.03$; $p = 0.002$; $\eta^2 = 0.316$ and $F(1,26) = 7.94$; $p = 0.009$; $\eta^2 = 0.234$, respectively). Post hoc pairwise comparisons showed statistically significantly higher values for the inexperienced cyclists compared to the experienced cyclists for STE_{SD}, STE_{RANGE} and STE_{RATE} at all three conditions ($p < 0.05$). No statistically significant between-group differences were present for STE_{MEAN} ($F(1,26) = 0.60$; $p = 0.444$; $\eta^2 = 0.023$).

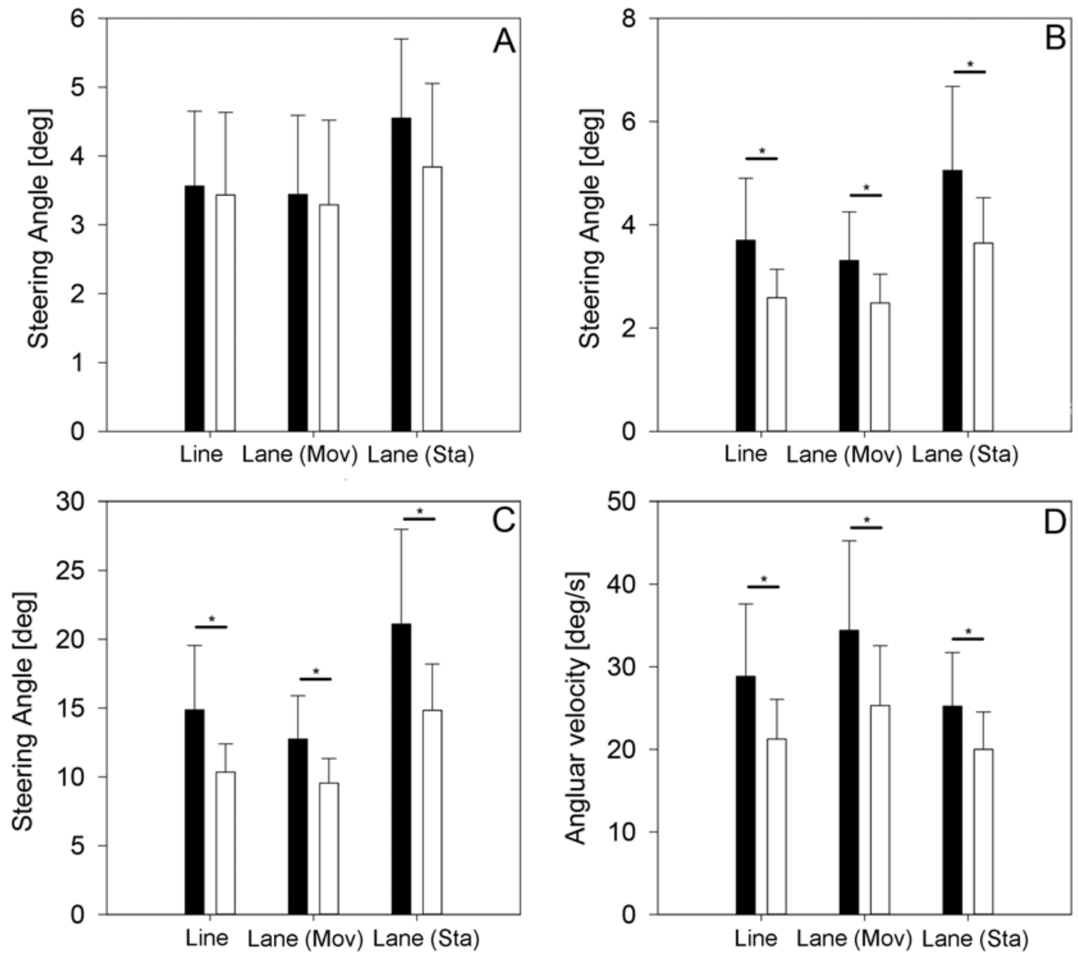


Figure 3.3: Absolute mean steering angle (A), standard deviation of the steering angle (B), range of the steering angle (C) and root mean square of the steering velocity (D) for novice (black bars) and expert (white bars) cyclists at three different conditions. Line, following a line with prior speed; Lane (Mov), in the middle of a cycle lane with prior speed; Lane (Sta), in the middle of the cycle lane from a standing start; *, p < 0.05 after the Bonferroni correction.

Roll parameters are illustrated in Figure 3.4. Inexperienced cyclists had statistically significant higher values of $ROLL_{SD}$, $ROLL_{RANGE}$ and $ROLL_{RATE}$ ($F(1,26) = 5.21$; $p = 0.031$; $\eta^2 = 0.167$, $F(1,26) = 5.45$; $p = 0.028$; $\eta^2 = 0.173$ and $F(1,26) = 5.00$; $p = 0.034$; $\eta^2 = 0.161$, respectively). Furthermore, it was found that inexperienced cyclists had statistically significantly higher values for $ROLL_{RATE}$ in Line and Lane (Stationary) conditions, $ROLL_{SD}$ in Line condition and $ROLL_{RANGE}$ in Line and Lane (Moving) conditions. No statistically significant differences between the groups were found for $ROLL_{MEAN}$ ($F(1,26) = 0.51$; $p = 0.478$; $\eta^2 = 0.020$).

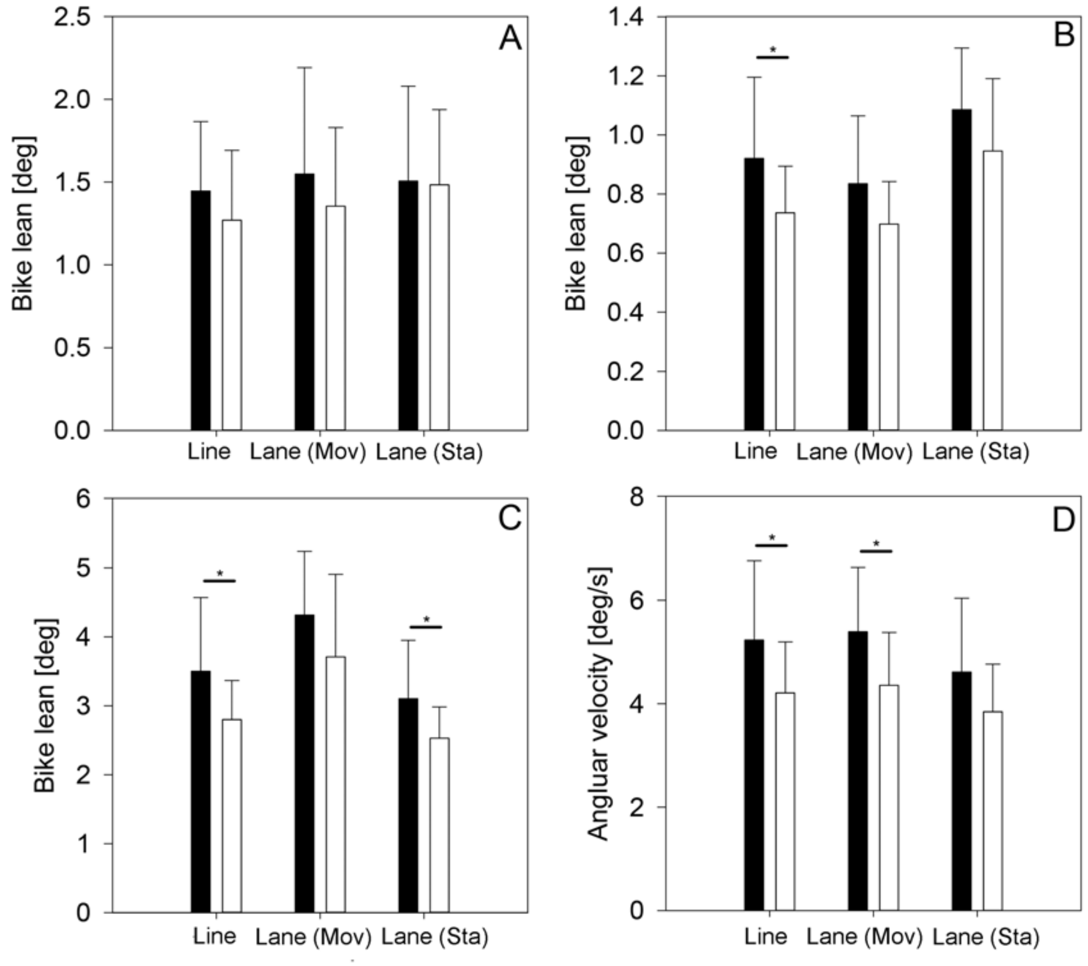


Figure 3.4: Absolute mean roll angle (A), standard deviation of the roll angle (B), range of the roll angle (C) and root mean square of the roll velocity (D) for novice (black bars) and expert (white bars) cyclists at three different conditions. Line, following a line with prior speed; Lane (Mov), in the middle of a cycle lane with prior speed; Lane (Sta), in the middle of the cycle lane from a standing start; *, p < 0.05 after the Bonferroni correction.

3.05 Discussion

The overall aim of the present study was to examine which variables reliably distinguish between more and less experienced cyclists, and to design a protocol that would serve as a valid and reliable tool to test the effects of any future interventions to improve bicycle rider control. Results show that deviation, steering and roll parameters reliably differentiate between cyclists with more and less experience. Motor control mechanisms in the two groups tested differ in the variability, amplitude and velocity of the control. Using gender-mixed design allows us to generalize the results of the present study to both genders.

Research assessing bicycle rider control has sometimes focused on errors while cycling on an obstacle course (Legg et al., 2003). Van den Ouden (2011) examined the trajectory of riding but has not performed any direct statistical comparisons between more and less experienced cyclists. Thus, this is the first peer-reviewed study that examined the actual trajectory of cycling, revealing that cycling in a straight line reliably describes some rider control skills. The results show that the standard deviation (DEV_{SD}) and range (DEV_{RANGE}) of the deviation achieved high inter-session reliability, indicating that lateral deviation is a reliable measure, which detects differences between more and less experienced cyclists. The advantage of DEV_{SD} over DEV_{RANGE} to assess rider control skills is that it takes the whole trial into an account, whereas DEV_{RANGE} takes only the two most extreme points. DEV_{SD} therefore gives an overview of the variability of how the task was performed throughout the entire distance. Variability itself has been previously shown as a good indicator of expertise in gait regulation (Scott et al., 1997).

Understanding rider control as a skill is necessary to design appropriate interventions for increasing cycling safety. This is the first study that examined one aspect of rider control skills and compares the motor behaviour of differently experienced cyclists in the same age group. We used a similar protocol and variables as Wierda & Brookhuis (1991) with the main difference in the length of the track. In addition, we also measured the actual performance of the task (lateral deviation) whereas they observed only the steering angle and speed. They found that younger cyclists (6–8 years old) performed with substantially more variability in the steering angle compared to their older peers. No significant changes occurred among the older age groups. Similar observations were found in a series of experiments by Cain (2013) who reported higher correlation between steer and roll angle in more experienced riders. That led to suggestions that more experienced riders steer in the direction of the roll of the bicycle, whereas less experienced riders perform additional steering motions that are not related to the roll of the bicycle. More experienced riders employed more rider's lean control, less steer control, and used less control effort than less experienced riders. These observations support our findings about the variability in all aspects of human control of a bicycle. We found significant differences between the experienced and inexperienced group for the standard deviation of steering, roll as well as the lateral deviation. Moreover, the amplitude of these movements was significantly higher for the inexperienced group compared to the experienced group. This indicates that less experienced cyclists maintain balance in a similar way as more experienced, but with a higher degree of variability. Similar pattern of skill level differences has previously been shown in other tasks (e.g. Scott et al., 1997; Wilson, Simpson, Van Emmerik, & Hamill, 2008). In the experiment presented here, the speed of cycling did not differ between the experienced and inexperienced group, which could indicate that differences between age groups observed in the study by Wierda & Brookhuis (1991) could also be a consequence of a skill level and not just lower speed, which has an effect on bicycle stability (Kooijman et al., 2009).

In static balance, the main measures that were shown to describe the performance of a certain task relate to the centre of pressure measurements (Panjan & Sarabon, 2010). Some of the most sensitive parameters are linked to body sway velocity and amplitude (Sarabon, Rosker, Loefler, & Kern, 2013). In the present study we observed that rate and amplitude of steering and bicycle roll are the most sensitive variables when comparing between more and less experienced cyclists. This means that experts perform steer and roll motions not only with a smaller amplitude and variability but also at a slower rate.

The length of the lane in the present experiment was chosen mainly due to the limitations of the size of the laboratory. One could argue that this is not enough to provide a clear picture of rider control. However, after a visual inspection of the raw data, all changes in steering and roll angle in both conditions with prior speed achieve steady state without any large deviation. That indicates that riders' control was constant and can reflect actual motor behaviour. On the other hand, participants in the present study were instructed to ride with a prescribed gear ratio, which in both conditions with prior speed resulted in an average speed of approximately 10 km/h, which is probably less than an average commute (Kooijman et al., 2009). At this speed, participants relied more on active control and less on self-stability, which allowed us to compare control actions.

The limitation of the present study was that bicycle rider control was examined in a controlled and safe environment, on a relatively short cycle lane, which could lack ecological validity. For example, traffic could represent a factor in one's behaviour during cycling. On the other hand, Wierda & Brookhuis (1991) showed that an additional cognitive task during cycling does not have an effect on steering variability and amplitude. However, they reported that there is an effect on a cognitive component (reaction time) during cycling, where younger cyclists seem to be more sensitive to the concurrency of the detection task and cycling (Wierda & Brookhuis, 1991). The data from our experiment could present a good starting point for further experiments implemented outdoors through observation of steer and roll of the bicycle. Another limitation of the present study is that four participants in the inexperienced group had to use the smaller geometry bicycle, whereas only two participants in the experienced group used that bicycle. Bicycle geometry might potentially have an effect on manoeuvrability and could consequently affect rider control. However, as the number of participants that used the large sized bicycle is still rather balanced between the groups, the difference can be neglected when making conclusions.

An interesting observation from the present study is that lateral deviation can reach the total amplitude over 30 cm even within a controlled environment, void of environmental hazard such as wind and potholes. According to London

Cycling Design Standards, in some cases, a cycling lane can be as narrow as 0.8 m. A novice cyclist on a standard commuting bicycle with a typical handlebar width of 0.6 m and a deviation of 0.3 m would exceed the width of a cycling lane, and therefore be exposed to a higher risk of having a collision with another vehicle. This would suggest that a cycling lane of 0.8 m in width is not a safe environment for novice cyclists. The width of the lane in the present study (0.6 m) was selected based on some personal observations of actual cycle lanes' widths in the United Kingdom, even though the suggested minimum width of the lane is 0.8 m. Therefore, it is crucial for further research to focus on improving bicycle rider control skills in order to reduce the risk of collisions and consequentially increase the safety of cyclists. Further research in the field is also important to understand the interaction between cycle lane designs and bicycle rider control (Schepers & den Brinker, 2011).

3.06 Conclusions

Our main findings are that riding a bicycle in a straight line in the centre of the lane reliably describes one aspect of bicycle rider control. The test is sensitive to detect differences between experienced and inexperienced cyclists and can be used for further research that aims to test the effect of a specific intervention (e.g. training) to affect control of a bicycle. We also showed that the main motions, which were described before as the main motor control actions in controlling a bicycle in a very limited number of participants (Kooijman et al., 2009; Moore et al., 2011) differ in the variability, amplitude and velocity between the experienced and inexperienced cyclists. This has practical implications for improving bicycle rider control and consequently increasing the safety of cyclists.

CHAPTER 4. CHANGING THE BICYCLE SEAT HEIGHT: EFFECTS ON RIDER CONTROL

4.01 Abstract[1]

The control of balance whilst riding a bicycle is essential for safety. The most common adjustment made to bike position is seat height. The purpose of this study was to examine how seat height affects bicycle rider control. It was hypothesized that seat heights set lower than currently recommended would improve control of the bicycle. Forty two cyclists completed five trials riding a bicycle in a straight line at four different seat heights. The initial seat height was set as the inner leg length multiplied by 1.09 for males and 1.06 for females. Other seat heights were 3% and 6% lower, and 3% higher than the initial seat height. Lateral deviation, steering, bicycle roll and knee angle were measured. Results revealed a statistically significant main effect for all dependent variables at the different seat heights. In particular, a decrease in steering and roll parameters was observed at seat heights set lower than the initial seat height. To improve the control of a bicycle, it can be recommended to set the seat height to 106% and 103% of the inner leg length for males and females, respectively. Knee angle, when the pedal is in its lowest position, should be between 43-57°.

[1] This chapter is submitted as: Fonda B, Sarabon N, Li F-X. Changing the bicycle seat height: effects on rider control. Eur J Sport Sci. under review

4.02 Introduction

Bicycle rider control skills for maintaining balance and bicycle heading are important to prevent falling off the bicycle, crashing into obstacles and avoiding subsequent traumatic injury. Lack of skill to control the bicycle is one of the most common reasons for traumatic injury occurrence and accounts for between 16% and 40% of the reported events resulting in injuries (Lloyd, 2013; Teschke et al., 2014b).

A bicycle itself is laterally unstable at low speed and stable, or easier to stabilise, at higher speeds (Kooijman et al., 2009; Moore, 2009). A bicycle is unstable at low speeds as it only has two contact points with the ground and therefore acts as an inverted pendulum. In order to retain balance, support points need to be shifted by means of steering to the falling side of the bicycle (Kooijman & Schwab, 2013; Moore, 2009). Schwab & Meijaard (2013) explored the main concepts of bicycle rider control and manoeuvrability. The vast majority of research on bicycle rider control (Hubbard et al., 2011; Kooijman et al., 2009; Schwab et al., 2012) has been focused on studying human motion and dynamics of the bicycle. Investigations on a limited number of participants (n = 3) revealed that the most common control action during cycling in a straight line is steering. This action is accompanied with several reactions, such as bicycle roll, spine bend, rider lean, rider twist and knee movements (Moore et al., 2011).

Recently, a study (Fonda, Sarabon, & Li, 2015) demonstrated how expertise affects rider control and collective variables that reliably describe bicycle rider control as a skill have been defined. They observed that less experienced cyclists exhibit higher variability in steering and bicycle roll compared to their more experienced peers. Additionally, more experienced cyclists were found to use slower steering and roll motions at smaller amplitudes. Hence, improvements in bicycle rider control could be reflected in smaller amplitude and slower rate of steering and bicycle roll motions.

To the authors' best knowledge, training is the only empirical intervention aimed at improving rider control published in peer-review literature. However, findings have been conflicting (Ducheyne et al., 2014; Ducheyne et al., 2013; Macarthur et al., 1998). To evaluate the level of rider control, Macarthur et al. (1998) used an obstacle course as a test, however this might lack sensitivity to detect small changes. In their study, Fonda et al. (2015) developed and validated a test battery that is reliable and sensitive to detect differences between cyclists with more and less experience. They found that the standard deviation of the steering and bicycle roll angle and root mean square of the steering and bicycle roll rate provide a reliable measure of rider control, with higher values linked to poorer skill.

Adjustments of body position on a bicycle have previously been explored to establish improvements in physical performance (Ashe et al., 2003; Ferrer-Roca et al., 2012; Peveler and Green, 2011; Peveler, 2008) and to reduce the risk of non-traumatic (overuse) injury occurrence (Bini et al., 2011; Holmes et al., 1994). Guidelines based on kinematics and/or morphological measures to properly set up a bicycle have been proposed (Burke, 1994; Pruitt, 2006). Seat height is usually the first control parameter a cyclist will adjust after buying a new bicycle and it has been extensively studied from a performance perspective, but never from a balance or rider control perspective (Gámez et al., 2008; Nordeen-Snyder, 1977; Peveler et al., 2005; Shennum and DeVries, 1976). Alterations in seat height have a direct effect on ankle, knee and hip biomechanics (Bini et al., 2012) which consequently affects cycling performance (Nordeen-Snyder, 1977) and comfort (Gámez et al., 2008). Changing the body position also affects the location of the centre of mass. The redistribution of mass may have an effect on rider control and balance. It has been shown in other tasks (e.g. sit-to-stand), that moving the centre of mass by changing the seat height affects postural control and balance (Janssen et al., 2002). Moreover, lowering the centre of mass improves static balance during quiet standing (Rosker et al., 2011). However, this effect could not be directly translated to cycling, as the mechanism to maintain balance can only be achieved by shifting the support points. Hence, from the mechanical point of view, centre of mass positioned higher would be easier to balance. That is due to a larger time constant for a longer inverted pendulum compared to a shorter one. On the other hand, sitting lower on a bicycle could affect the perception of fear and ease the control of a bicycle.

Bicycle rider control is essential to prevent falls and subsequent traumatic injuries. Adjustment of seat height affects the centre of mass of the cyclist-bicycle system and, in turn, this change may affect the ability to balance and control the bicycle. The aim of the present study was to test the effect of changing the seat height on rider control during cycling. As lowering the centre of mass would probably reduce the bicycle-plus-rider stability, but lower the perception of fear, it was hypothesized that there would be a trade-off between the physical stability and fear at a certain seat height described with a decreased steering and roll amplitude and rate while riding in a straight line.

4.03 Methods

4.03.1 Participants

According to a priori sample size calculations based on the pilot data, 18 male and 24 female cyclists ([mean ± SD] age 26.1 ± 7.3 years, body mass 65.5 ± 10.3 kg, and body height 169.5 ± 7.3 cm) were recruited. Exclusion criteria prevented cyclists taller than 185 cm and smaller than 150 cm from participating in the study due to the limited bicycle sizes available. Participants reported that, on average, they ride a bicycle 1735 ± 2449 km per year, ranging between 10 and 10,000 km. This large range of experience covers most of the cycling population and should enable conclusions to be made across a wide population and in both sexes. Before the experiment, each participant signed an informed consent form, which was approved by the University's ethical committee.

4.03.2 Protocol

The test involved riding a commuting bicycle in the laboratory (30x10 m size) for 7 m in the centre of a 60 cm wide cycle lane from a standing start. The instructions were to ride as straight as possible in the centre of the cycle lane. This task was completed 5 times for practice followed by 5 times at each seat height. The test protocol has been shown to reliably describe bicycle rider control (Fonda et al., 2015). The speed was partially controlled for by setting a fixed gear ratio and verbally instructing the participants to go faster or slower during the practice trials.

For all male participants, the initial seat height was set according to the Hamley & Thomas (1967) method, by measuring the inseam leg length (distance from the pubic symphysis to the floor) and multiplying it by 1.09. This length was then used to set the seat height as a distance between the top of the saddle and the top of the pedal when the pedal is in its lowest position. A pilot study revealed that female participants could not reach the pedals when setting the seat height 3% higher than the initial seat height. Therefore a correction factor multiplying the inseam length by 1.06 was used to set the initial seat height for female participants only. From the initial seat height (100%), the other seat heights were 3% higher (103%), 3% lower (97%) and 6% lower (94%). These values were chosen as they represent a realistic range of seat adjustments. Trials at different seat heights were performed in a random order to minimize any order effects.

4.03.3 Setup

Two commuting bicycles (Transfer Hybrid (Seat tube angle: 73°, crank length: 175 mm, bottom bracket height: 280 mm) and Excelle Apollo Ltd., Victoria, Australia Hybrid (Seat tube angle: 74°, crank length: 175 mm, bottom bracket height: 275 mm)) with the same wheel size (29”), handlebar width and type but different frame types and sizes were available to the participants. Both bicycles had the same tyre type, inflated to 400 kPa. Participants were assigned to one of the two bicycles based on their body height.

Vicon MX motion analysis system (Oxford Metrics Ltd., Oxford, UK), consisting of 13 cameras with a sampling rate of 200 Hz and calibration residual error of less than 1 mm, was used to capture 3D kinematics. Retro-reflective markers were attached with double-sided adhesive tape on a custom made frame fitted to the front of the bicycle (three markers), whilst two were attached to the stem (Figure 4.1). The extremities of the cycling lane were marked with markers prior to each test to define the centre of the lane. Additional markers were attached unilaterally on the greater trochanter, lateral condyle, lateral malleolus, heel and laterally in line with the metatarsophalangeal joints.

Figure 4.1: Bicycle and participant setup. Reflective markers were put on the main anatomical points of the body and on the bicycle with a custom designed T-frame in front allowing us to calculate steering and leaning parameters.

4.03.4 Variables

The main bicycle control parameters were calculated from the lateral deviation, steering and bicycle roll kinematics. Lateral deviation was calculated as the lateral distance of the bicycle's frame, calculated from the projection of two markers on the T-frame that were positioned directly above the contact point when the bicycle was completely upright. In case of bicycle roll, the calculated point would be the projection through both markers towards the ground. With that, we compensated for the lateral movement due to bicycle roll only. Lateral deviation was quantified as a standard deviation of the lateral deviation (DEV_{SD}). Furthermore, standard deviation of the steering and roll angle (STE_{SD} and $ROLL_{SD}$, respectively) and root mean square of the steering and roll rate (STE_{RATE} and $ROLL_{RATE}$, respectively) were calculated. All of the listed variables were found to be sufficiently sensitive and reliable to detect differences in rider control expertise (Fonda et al., 2015). Increases in lateral deviation, steering and roll amplitude and steering and roll rate are linked to poorer control of the bicycle. We also calculated the knee angle when the pedal was in its lowest position.

4.03.5 Statistics

The mean of the five trials was used for statistical analysis. The Shapiro-Wilk test showed that all data had a normal distribution. Multivariate ANOVA was used to test for differences among the seat heights for 5 dependent variables (DEV_{SD}, STE_{SD}, STE_{RATE}, $ROLL_{SD}$, $ROLL_{RATE}$). *Seat height (4)* was set as a repeated measure factor and *dependent variables (5)* as an independent factor. Additional one-way repeated measures ANOVA was performed for each dependent variable, including the knee angle. Mauchly's test of sphericity was conducted a priori and if it was violated, an appropriate correction was applied. *Post-hoc* pairwise comparisons between the seat heights were calculated using a Bonferroni correction. Statistical analyses were performed using SPSS V.22 for Mac (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$.

4.04 Results

Changing the seat height resulted in a statistically significant *main effect* for all dependent variables ($F_{2.37,94.89} = 7.05$; $p = 0.001$; $\eta^2 = 0.150$).

Results for the DEV_{SD} are illustrated in Figure 4.2. There was no statistically significant difference for DEV_{SD} among different seat heights ($F_{3,120} = 1.24$; $p = 0.299$; $\eta^2 = 0.030$).

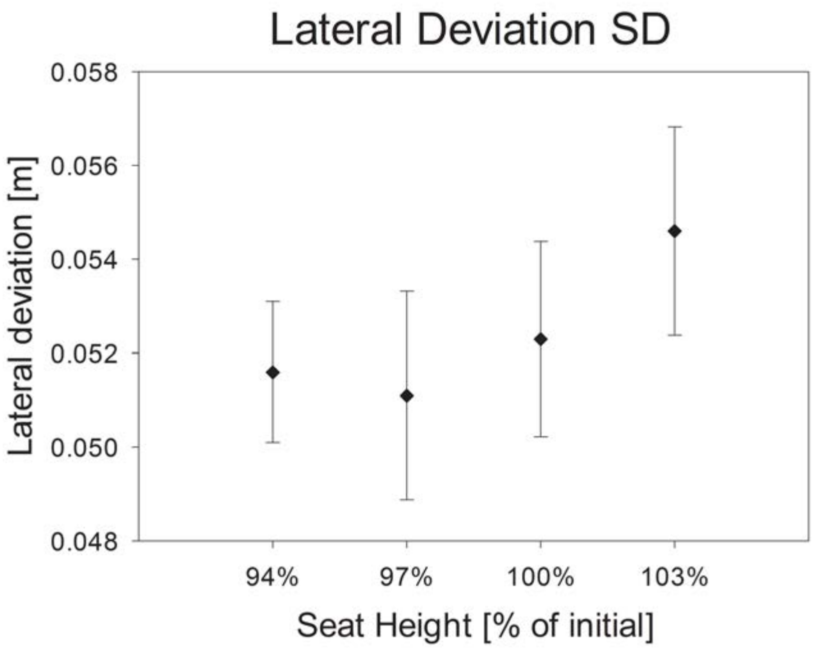


Figure 4.2: Deviation parameters at different seat heights. 94%, seat height set 6 % lower as the initial; 97%, seat height set 3% lower as the initial; 100%, initial seat height; 103%, seat height set 3% higher as the initial seat height. There were no statistically significant differences among the seat heights for any of the variables.

Bicycle roll results are illustrated in Figure 4.3. There was a statistically significant difference at different seat heights for $ROLL_{SD}$ and $ROLL_{RATE}$ ($F_{3,120}$ =

7.76; $p = 0.000$; $\eta^2 = 0.162$ and $F_{2.41,96.39} = 4.83$; $p = 0.000$; $\eta^2 = 0.309$, respectively). Furthermore, pairwise *post-hoc* comparisons revealed that $ROLL_{SD}$ was statistically significantly lower at 94% compared to 100% ($p = 0.004$) and 103% ($p = 0.000$). $ROLL_{RATE}$ was statistically significantly lower at the seat height set to 94% compared to 100% ($p = 0.000$) and 103% ($p = 0.000$) and at the seat height set to 97% compared to 100% ($p = 0.001$) and 103% ($p = 0.000$).

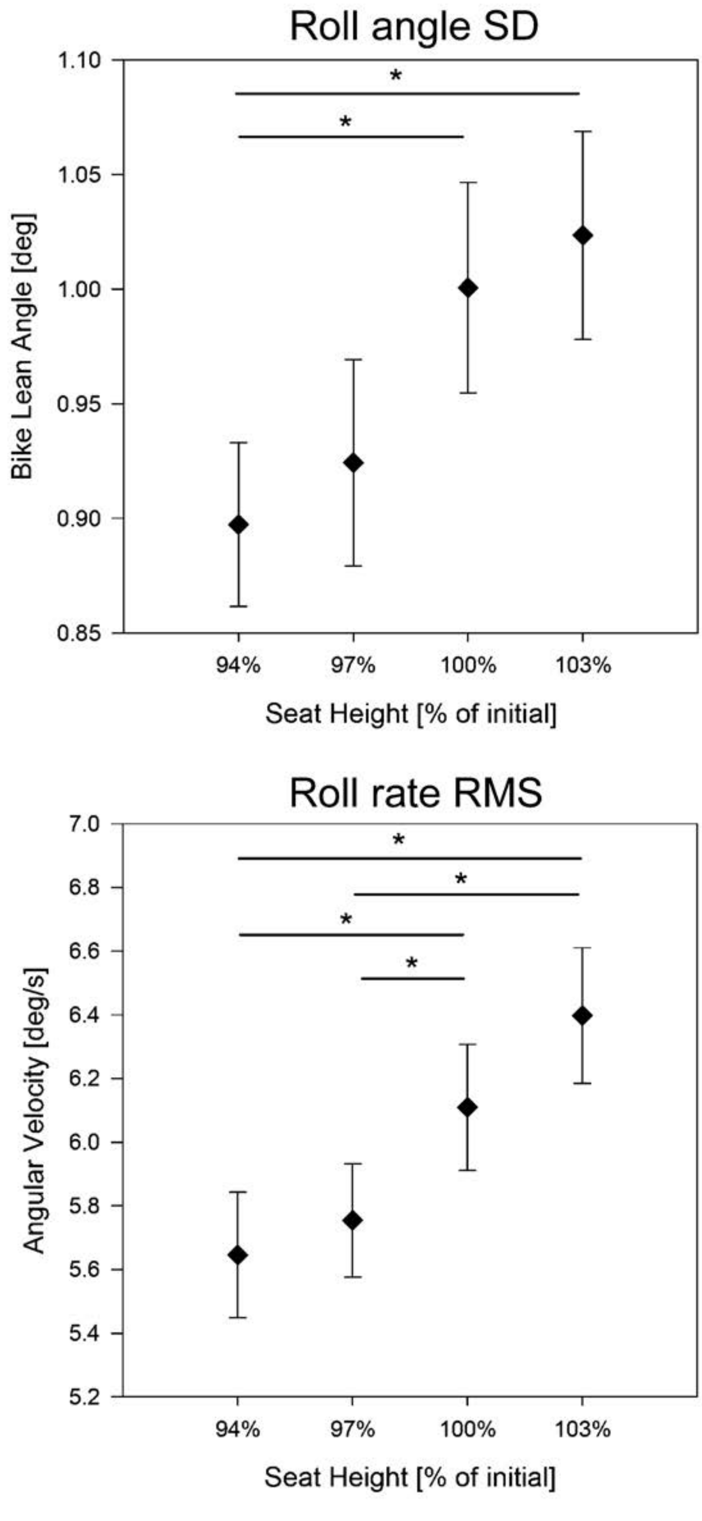
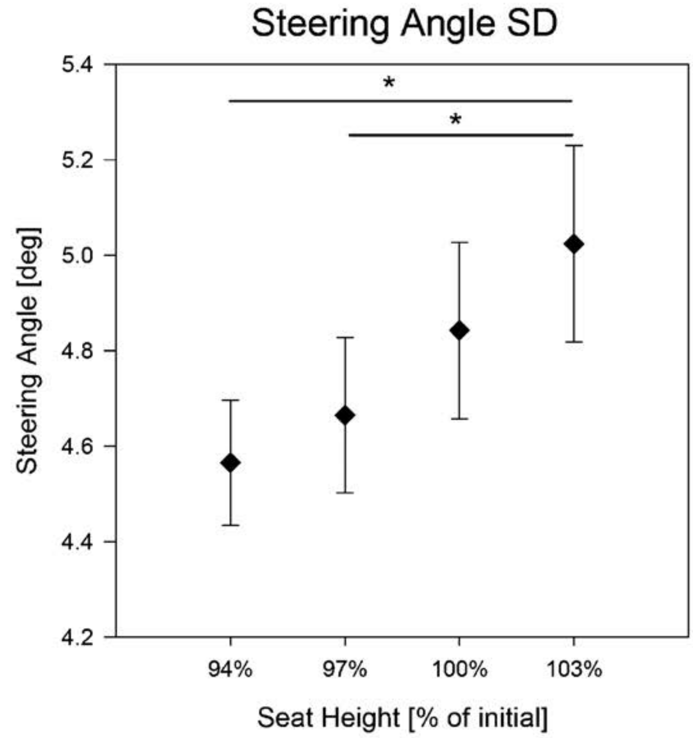


Figure 4.3: Roll angle parameters at different seat heights. 94%, seat height set 6 % lower as the initial; 97%, seat height set 3% lower as the initial; 100%, initial seat height; 103%, seat height set 3% higher as the initial; *, statistically significant ($p < 0.05$) difference after Bonferroni correction between two seat heights.

Results for steering parameters are illustrated in Figure 4.4. A statistically significant difference at different seat heights was found for STE_{SD} and STE_{RATE} ($F_{3,120} = 3.33$; $p = 0.022$; $\eta^2 = 0.077$ and $F_{2.38,95.06} = 5.44$; $p = 0.004$; $\eta^2 = 0.120$, respectively). *Post hoc* comparisons revealed that STE_{SD} was statistically significantly lower at the seat height set to 94% compared 103% ($p = 0.010$). Additionally, it was significantly lower at 97% compared to 103% ($p = 0.026$). STE_{RATE} was found to be statistically significantly lower only for 97% compared to 103% ($p = 0.009$).



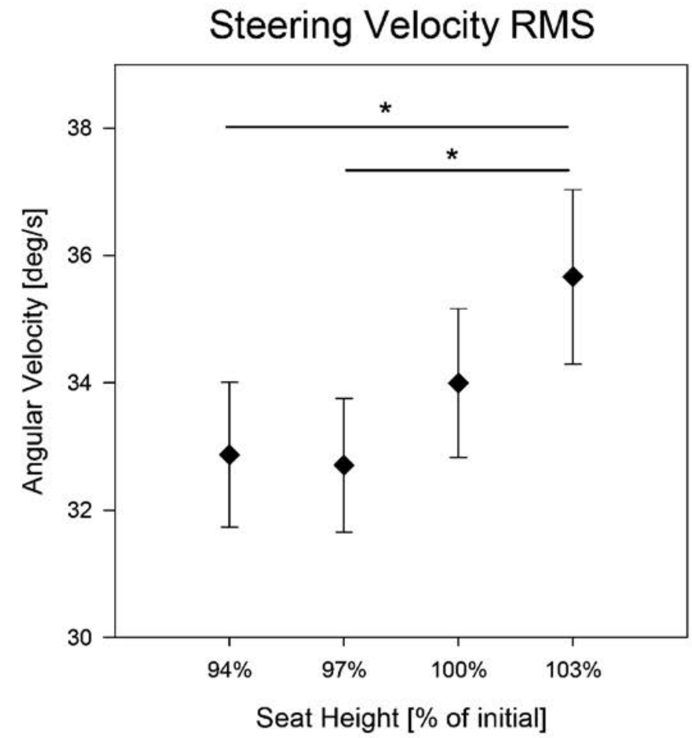


Figure 4.4: Steering parameters at different seat heights. 94%, seat height set 6% lower than the initial; 97%, seat height set 3% lower than the initial; 100%, initial seat height; 103%, seat height set 3% higher than the initial; *, statistically significant ($p < 0.05$) difference after Bonferroni correction between two seat heights.

Knee angle *main effect* for different seat heights was found to be statistically significant ($F_{2.41,91.59} = 155.14$; $p = 0.000$; $\eta^2 = 0.803$). The highest knee angle was observed at 94% (57.5 ± 6.5°), followed by 97% (50 ± 7.7°), 100% (43.3 ± 7.7°) and 103% (35.5 ± 8.6°). *Post hoc* tests showed statistically significant differences between the four seat heights ($p < 0.016$).

4.05 Discussion

The aim of the present study was to examine how changes in seat height affect the control of the bicycle. Changes in bicycle control at different seat heights were observed, but not in the trajectory of riding. Results support the hypothesis that seat height affects variability and rate of steering and bicycle roll motions.

Seat height has been a popular subject of research in sport science, aiming to find the position that would improve mechanical efficiency (de Vey Mestdagh, 1998; Nordeen-Snyder, 1977; Peveler and Green, 2011) and prevent the occurrence of non-traumatic injuries (Bini et al., 2011). The present study is the first to examine how changes in seat height affect bicycle rider control. It was observed that a decrease in the amplitude and rate of the two main motor control actions to maintain balance during cycling (steering and roll) is present when seat height is set lower than currently recommended (Hamley and Thomas, 1967). As originally hypothesized, this behaviour could be the result of a trade off between the physical stability when the centre of mass is higher and a reduced perception of fear when the centre of mass is lower.

Similarity in maintaining balance during standing and cycling can be found in the mechanism of maintaining equilibrium. The bicycle and the human body are both inherently unstable due to the two contact points between the bicycle/feet and the supporting surface (Schwab and Meijaard, 2013; Winter, 1995). However, during cycling, balance can only be maintained by shifting the support points by steering, whereas during quiet standing, balance can be maintained by exerting torques on certain joints and keeping the centre of mass over the centre of pressure. In the present study the task was not only to maintain balance, but also to control the direction of the bicycle as accurately as possible. In terms of the accuracy (lateral deviation), no significant differences at different seat heights were observed. This could be explained by the redundancy and availability of degrees of freedom (Bernstein, 1967; Latash et al., 2002). In essence, cyclists have more than one way to achieve the same outcome. For example, cyclists can use different strategies to maintain balance (Kooijman et al., 2009), or even a combination of these strategies, with each option potentially leading to a similar lateral deviation. It seems that changing seat height only changes the level of constraint in one control variable and participants were able to adjust their behaviour using other degrees of freedom.

Hamley & Thomas (1967) first suggested setting the seat height as a function of inner leg length (inseam). Their method was also used in the present study to set the initial (i.e. currently recommended) seat height. However, during the pilot testing, it was observed that female participants could not reach the pedals when the seat height was set 3% higher than the initial height, whilst male participants did not experience any difficulty. It was found that setting the initial seat height for females approximately 3% lower than for males (inseam multiplied by 1.06) led to similar knee angles for both genders when the pedal was in its lowest position and, consequently, this correction factor was applied. The gender differences are probably related to differences in pelvis anatomy and orientation during cycling (Potter et al., 2008; Sauer et al., 2007) and this warrants further research to identify if the factor applied in this experiment can be generalized with a larger sample of female cyclists.

Based on the results of the present study, it can be suggested that commuting cyclists and cyclists who are seeking to improve control of the bicycle should set their seat height using the inseam leg length multiplied by 1.06 and 1.03 for males and females respectively. This is approximately 3% lower than currently recommended (Ferrer-Roca et al., 2012). For a seat height set to 95 cm (measured from the top of the seat to the pedal when it's in its lowest position), that means lowering the saddle by 2.85 cm. Ferrer-Roca et al. (2014) have shown that changes in seat height of this magnitude have a significant effect on gross efficiency. As the vast majority of the cycling literature has tested male participants, it would be interesting to investigate further the effect of seat height on female cyclists performance**,** although there was no significant difference in steering and roll parameters between the seat height set to 3% and 6% lower than the initial height.

Although not described in the methods and results section, each participant got asked which of the four tested seat height appeared as the most comfortable and stable. All participants uniformly answered that it was the one set to 3% lower than the initial seat height. Based on the results of the present study and subjective observations from the participants, it can be recommended that the seat height should be set to 1.06 and 1.03 multiplied by the inseam leg length for males and females, respectively, in order to improve comfort and control of the bicycle.

It was observed that the knee joint is significantly more flexed when lowering the seat height, which confirms the findings of previous studies (Bini et al., 2010). However, when comparing the knee angle values at different seat heights from the present study to other studies (Ferrer-Roca et al., 2012), a slight difference can be observed despite using the same method to set the initial seat height. These differences could be the result of the kinematic method used (Fonda, Sarabon, & Li, 2014) and the fact that participants in the present study had to maintain balance, for which lateral knee movements can be used as well (Kooijman et al., 2009). Bicycle fitting experts often set the position to reach knee angle values within the range of 25° to 35° when measured with a manual goniometer (Holmes et al., 1994). However, manual goniometer underestimates the knee angle measurement and should be discouraged for the purposes of bicycle fitting (Fonda et al., 2014). In the present study, the knee angle values measured with a 3D motion capturing system at the optimal seat height reached values, on average, of 50 ± 7°. Based on these measurements it can be suggested that the seat height should be set to reach knee angle values between 43° and 57° in order to improve the control of the bicycle.

Although it has been shown (Fonda et al., 2015) that the protocol used in the present study is valid and reliable to describe rider control skills, it only examines one aspect of riding (i.e. riding in a straight line from a standing start) which is a limitation of this study. Further research should be carried out to examine how changing the position on a bicycle affects control of the bicycle in a more varied environment and during a larger range of tasks.

4.06 Conclusion

Based on the result of the present study it can be concluded that alterations in seat height affect bicycle rider control. This should be taken into account by bicycle fitting experts when setting a bicycle for a commuting cyclist or a cyclist whose main objective is to ride safely. Furthermore, it seems that seat height should be set differently for female cyclists compared to their male peers. It can be recommended that the seat height should be set based on the inseam leg length, multiplied by 1.06 and 1.03 for males and females respectively. Knee angle, when the pedal is in its lowest position, should be between 43° and 57° when a 3D motion capturing system is used. Further research on altering the body position to improve rider control is required.

CHAPTER 5. CYCLE LANE DESIGN AFFECTS BICYCLE RIDER CONTROL

5.01 Abstract[1]

One of the important factors to increase cyclists' road safety is infrastructure, which mainly includes cycle lanes and specifically designated cycle paths. Despite guidelines in the UK stipulating the minimum width of a cycle lane to be 1.2 m, narrower cycle lanes are often observed, with widths ranging between 0.8 – 1.2 m. It is unknown how the design of a cycle lane, more specifically the width and surrounding infrastructure of a cycle lane, affects rider control. Therefore, the aim of the present study was to examine the effects of cycle lane width on bicycle rider control. 18 cyclists rode a bicycle with an angular rate sensor mounted on the stem along a 20 m cycle lane at semi-controlled speed (fixed gear ratio). The cycle lane had three designs: 1) 'just lane' where the lane was marked with a red tape on each side, 2) 'infrastructure' with road barriers on the right side and a curb on the left side and 3) 'gap passing' with two vertical poles at each side. All designs had three widths: 80, 100 and 120 cm. The average steering rate root mean square (RMS) and roll rate RMS were calculated. Steering rate and roll rate at 80 cm wide infrastructure condition were statistically significantly higher compared to the other two widths in this design. Steering rate and roll rate were found increased just prior to passing the gap and decreased after the gap. In conclusion, the results showed that the width of a cycle lane affects steering and roll motions when the design physically restricts the manoeuvring area (curb and barriers).

---

[1] This chapter will be submitted as Fonda B, Sarabon N, Li F-X. Cycle lane design affects bicycle rider control. *Accident Analysis & Prevention*; in preparation.

5.02 Introduction

According to the Department for Transport, 19,000 cyclists are killed or seriously injured every year in the United Kingdom (Lloyd, 2013). One of the most common reasons for accident occurrence is losing control of the bicycle. It accounts for over 16 % of the reported accidents in the UK (Lloyd, 2013) and over 40% in Canada (Teschke et al., 2014a). It is clear that bicycle rider control for maintaining balance plays an important role in the prevention of falling off the bicycle, crashing into obstacles and avoiding subsequent traumatic injury.

One of the important factors to increase cyclists' road safety is infrastructure, which mainly includes cycle lanes and, specifically, designated cycle paths. The existing literature suggests that implementation of cycle lanes substantially reduces the risk of crashes and injuries sustained during cycling (Reynolds et al., 2009; Teschke et al., 2012). Moreover, cyclists' perception of safety is greatly increased when better cycle lane facilities are available. For example, a study (Chataway et al., 2014) revealed that the presence of wider cycle lanes effectively reduces the fear of traffic. With an increased perception of safety, one can expect that more people will cycle daily and choose a bicycle as their preferred mode of transport.

At the moment, guidelines in the UK suggest a minimum width for a cycle lane to be 1.2 m (DfT, 2008). Despite these guidelines, the authors often observed narrower cycle lanes in one of the UK's major cities, with the majority ranging between 0.8 – 1.2 m. Furthermore, Frings, Parkin, & Ridley (2014) reported that in their study, the gap between the curb and the motor vehicle on different junctions with a cycle lane was often less than 1m. Wider cycle lanes do not only offer more manoeuvring area, but also affect perception which, in turn, could change the cyclist's motor behaviour and control of the bicycle. This effect has been observed during driving at different lane widths. Godley, Triggs, & Fildes (2004) reported that narrower lanes increased steering workload and reduced the speed of driving. Furthermore, McLean & Hoffmann (1972) showed an increase in the number of rapid steering movements and less accurate steering (higher angular rate) with decreasing lane width.

Recently, Lee et al. (2015) aimed to design guidelines for minimum cycle lane widths based on the cyclists' trajectories while riding on a cycle lane of various widths. By measuring real-time kinematics with a global positioning system they calculated an essential manoeuvring space, which is defined as the maximum lateral movement of the bicycle (extremities of the handlebar). The study reported that the minimum width of a one-way cycle lane should be at least 2 m. To date, this is the only attempt to examine the effects of a cycle lane design on a bicycle control (trajectory of riding). Even though the results of their study (Lee et al., 2015) demonstrated a desirable minimum width for a cycle lane, this might not be realistic due to space restrictions in urban environments. Therefore, research should focus also on narrower widths for a cycle lane that are more realistic to implement on limited space.

A study by Vansteenkiste et al. (2013) examined gaze behaviour during cycling at various lane widths and speeds of cycling. They found that cyclists adjust their speed of cycling according to the lane width (riding faster in wider lanes) and direct more visual attention towards the end area of the path at higher speeds, towards the near pathway on narrow lanes and more towards irrelevant areas on wider lanes. A similar pattern was observed in their follow-up experiment (Vansteenkiste et al., 2014) where they examined visual behaviour during cycling on cycle paths of different quality. They observed a shift of attention to more proximate path regions when cycling on a low quality bicycle path. Hence, cyclists' attention to environmental hazards may be affected when cycling on a lower quality cycle path.

Although Vansteenkiste et al. (2013, 2014) did not record any of the bicycle rider control parameters, such as steering and bicycle roll, one could expect that motor behaviour during cycling is affected by the cycle lane width or design (e.g. quality). This would probably be even more obvious in cases where a cycle path or a gap that a cyclist would like to pass through is very narrow. Humans perceive the environment and objects in body-scaled dimension (Gibson, 1979). For a cyclist to afford cycling between a curb and a car, the gap must be perceived in relation to the cyclist's widest frontal dimension, which is likely to be the handlebar. In gait, Warren & Whang (1987) demonstrated that when people walk through an aperture, they modify motor behaviour by rotating their shoulders to leave a safety margin of at least 1.3 times their shoulder width. That means if the gap is narrower than 1.3 times the shoulder width, people will rotate their shoulders to ensure perceived safe passage.

Rotating the shoulders would have little effect during cycling as the handlebar is the widest part of the cyclist-bicycle system. The adjustment in motor behaviour would probably happen in steering and bicycle roll motions to ensure a smaller amount of lateral movement. Based on the aforementioned driving studies (Godley et al., 2004; McLean & Hoffmann, 1972), the rate of steering and bicycle roll motions would be increased with a reduction in the manoeuvring area. This hypothesis remains to be tested. Therefore, the first aim of the present study was to examine the effect of cycle lane width and design on bicycle rider control. It has been hypothesised that narrower cycle lanes would result in a higher steering and bicycle roll rate. Also, changing the design of a cycle lane by physically restricting the manoeuvring area would result in an increase in steering and roll rate. The second aim was to assess the amount of adjustments in bicycle rider control prior to and after passing a gap of different widths. It has been hypothesised that steering and rolling rate would be increased just prior to the gap and decreased after passing the gap.

5.03 Methods

5.03.1 Participants

After a priori sample size calculations based on our pilot data, 16 male and 2 female cyclists ([mean ± SD] age 24 ± 4.8 years, body mass 74.3 ± 11.8 kg, and body height 177.5 ± 5.9 cm) were recruited. Exclusion criteria prevented cyclists taller than 185 cm and smaller than 170 cm to participate in the study to ensure that all cyclists could comfortably use the same bicycle. Participants had different levels of expertise (riding between 10 and 10000 km per year). Before the experiment each participant signed an informed consent form, which was approved by the University's ethical committee.

5.03.2 Apparatus

A commuting bicycle (Transfer Hybrid, Apollo Ltd., Victoria, Australia) was used with the seat height adjusted for each participant according to the current recommendations (Fonda, Sarabon, Blacklock, & Li, 2014). A custom-built portable device with a data logger and 3-axis angular rate sensor was fixed to a custom-made mount on the bicycle's stem. The device consisted of one 3-axis angular rate sensor with a scale set to 500 degrees per second and sensitivity set to 131 deg/s. All data was transferred through an Arduino board (Seeeduino Stalker v2, Seeed Technology Inc., Shenzhen, China) and stored at a sampling rate of 200 Hz on a SD card.

A hall-effect sensor was mounted on the front fork to record TTL pulses from four magnets on the front wheel. The magnets were evenly allocated around the wheel at a distance of 0.55 m of the wheel circumference. The sensor was connected to the Arduino and synchronised with the angular rate sensor.

5.03.3 Set up and procedure

Participants were asked to ride the instrumented bicycle outdoors in a secure car park, which was free from traffic and pedestrians. Three different cycle lane designs with a length of 20 m and starting line 15 m prior to the lane were used. In the first design, '*Just Lane*', the cycle lane was marked with a red tape on each side without any physical restrictions on either side. The second design, '*Curb & Barriers*', had longitudinally placed road barriers on the right side and a polystyrene curb on the left side. The third design, '*Gap Passing*', used the same cycle lane as in the Just Lane design but with the inclusion of two parallel vertical poles at the 15 m mark from the beginning of the cycle lane. Each design used three different widths (0.8, 1 and 1.2 m), measured from the inner side of the lane, curb/barriers and poles. These designs were selected as they represent a realistic range of cycling infrastructure.

Participants rode the bicycle five times at each design (45 trials in total). They were instructed to maintain pedalling both before and while riding in the cycle lane. Speed was semi-controlled using a fixed gear ratio (39x19), but no other instruction regarding speed was given, allowing us to observe changes in speed under different designs.

5.03.4 Data analysis

An angular rate sensor directly recorded steering and roll rate. It is worth noting that the roll was not the actual roll rate of the bicycle frame as the sensor was mounted on the stem and not directly on the frame, hence this needs to be interpreted with caution. Recorded steering and roll rates were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 5 Hz (Cain & Perkins, 2012; Cain, 2013).

For the *Just Lane* and *Curb & Barriers* conditions, the absolute data from the entire 20 m cycle lane was taken into further analysis. Root mean square (RMS) of the steering and roll rate, number of steering reversals and average speed were calculated. Additionally, for the *Gap Passing* conditions, steering rate RMS and roll rate RMS were calculated in two parts: for the length of 5 m before the vertical poles and for the length of 5m after the vertical poles. The data from these two sections was normalised to the first 10 m in each condition. All data analyses were performed with a bespoke script in MATLAB (The MathWorks, Inc., Natick, MA, USA).

At the end of the five trials in each condition, participants were asked to grade their perception of safety on a scale from 1 to 10, with 1 as unsafe to ride and 10 as completely safe to ride.

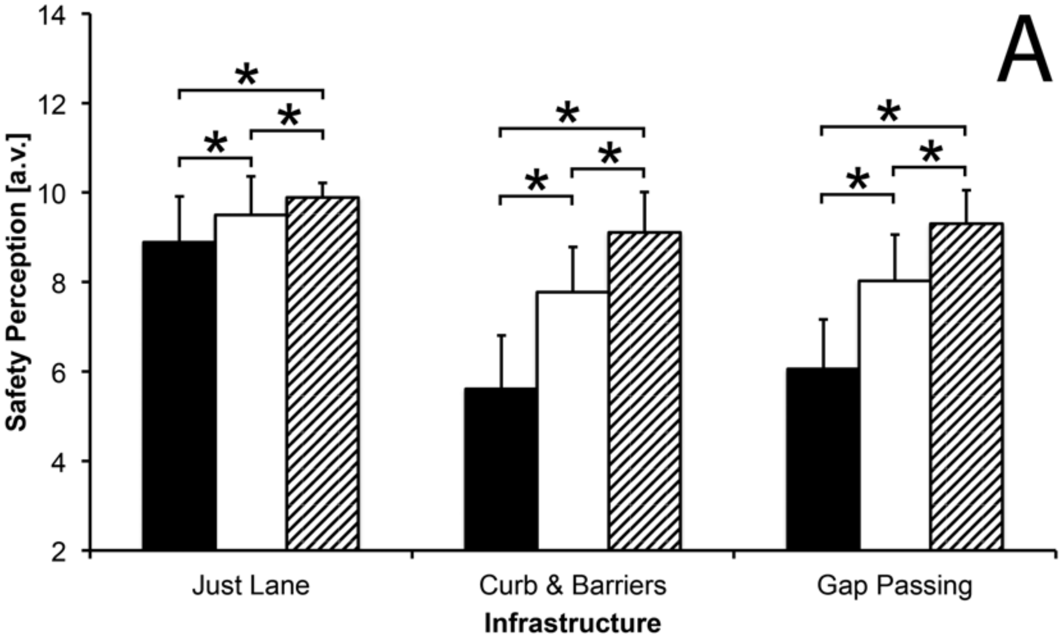
5.03.5 Statistics

The mean of the five trials was used for statistical analysis. The Shapiro-Wilk test showed that all data had a normal distribution. Initially, both the average speed and perceived safety were compared across all designs with a 2-way *infrastructure(3) x width(3)* repeated measures analysis of variance (RM ANOVA). Furthermore, steering rate RMS, roll rate RMS and number of steering reversals in the *Just Lane* and *Curb & Barriers* conditions were analysed with a 2-way *infrastructure(2) x width(3)* RM ANOVA. *Gap passing* condition was tested with a 2-way RM ANOVA with a *segment(2) x width(3)* interaction for steering rate RMS and roll rate RMS. Mauchly's test of sphericity was conducted a priori and if it was violated, appropriate correction was applied. *Post-hoc* pairwise comparisons between the conditions were calculated using a Bonferroni correction. Statistical analyses were performed using SPSS V.22 for Mac (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$.

5.04 Results

Results for average speed and perceived safety across all conditions are illustrated in Figure 5.1. There was a statistically significant *main effect* of the width for average speed ($F_{2,34} = 10.73$; $p = 0.000$; $\eta^2 = 0.387$). However, the difference in average speed at different infrastructures was not statistically significant ($F_{2,34} = 2.61$; $p = 0.089$; $\eta^2 = 0.133$) and neither was the *width x infrastructure* interaction ($F_{4,68} = 0.83$; $p = 0.512$; $\eta^2 = 0.46$). Perceived safety was statistically different between the widths ($F_{2,34} = 171.22$; $p = 0.000$; $\eta^2 = 0.910$) and infrastructures ($F_{2,34} = 68.97$; $p = 0.000$; $\eta^2 = 0.802$). There was also a statistically significant *width x infrastructure* interaction ($F_{4,68} = 23.8$; $p = 0.000$; $\eta^2 = 0.583$).

Average speed was statistically significantly decreased for the 0.8 m wide *Just Lane* compared to the 1 m Just Lane condition ($p = 0.014$). Similarly, 0.8 m wide *Gap Passing* conditions exhibited statistically significantly lower speed as the 1 m wide gap passing condition ($p = 0.031$). Perceived safety was statistically significantly different between all conditions ($p < 0.05$) with lower marks in the narrower conditions.



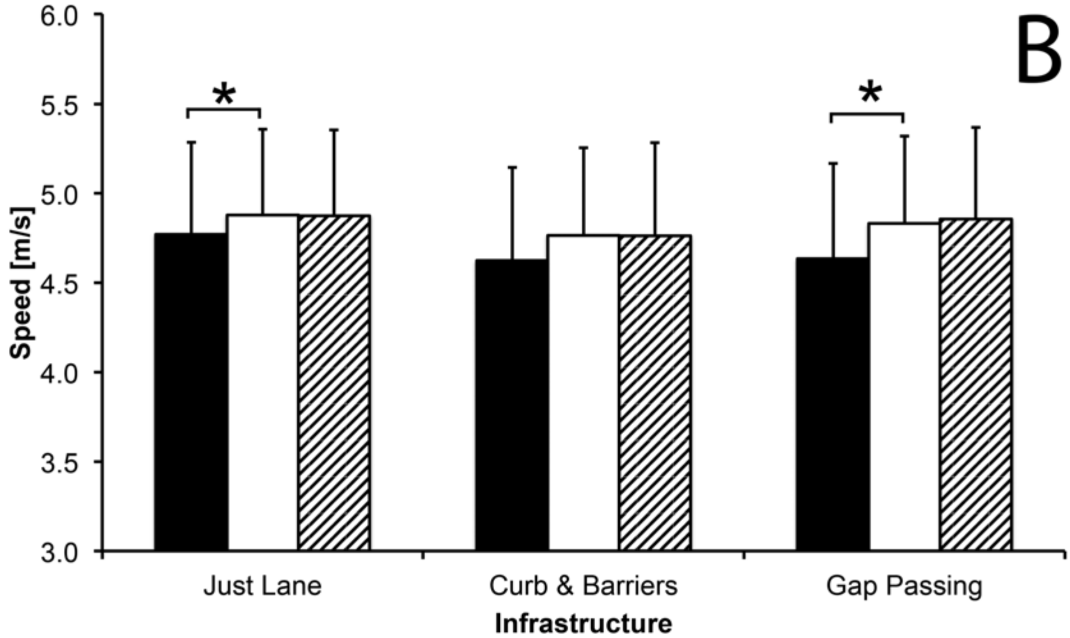


Figure 5.1: Perceived safety (A) and average speed (B) at different infrastructures and lane width (black column, 80 centimetres; white column, 100 centimetres; patterned column, 120 centimetres). *, statistically significant change ($p < 0.05$)

5.04.1 Infrastructure

The effect of infrastructure and width for steering rate RMS, roll rate RMS and number of steering reversals is illustrated in Figure 5.2. There was a statistically significant *main effect* of the infrastructure for steering rate RMS ($F_{1,17} = 16.22$; $p = 0.001$; $\eta^2 = 0.488$), roll rate RMS ($F_{1,17} = 10.50$; $p = 0.005$; $\eta^2 = 0.382$) and number of steering reversals ($F_{1,17} = 5.26$; $p = 0.035$; $\eta^2 = 0.236$). Similarly, a statistically significant *main effect* of the width was observed for steering rate RMS ($F_{2,34} = 13.13$; $p = 0.000$; $\eta^2 = 0.436$), roll rate RMS ($F_{2,34} = 9.80$; $p = 0.000$; $\eta^2 = 0.366$) and number of steering reversals ($F_{2,34} = 6.29$; $p = 0.005$; $\eta^2 = 0.270$). An *infrastructure x width* interaction was found statistically significant for steering rate RMS ($F_{2,34} = 4.55$; $p = 0.018$; $\eta^2 = 0.211$) and roll rate RMS ($F_{2,34} = 4.10$; $p = 0.025$; $\eta^2 = 0.194$), but not for number of steering reversals ($F_{2,34} = 0.59$; $p = 0.559$; $\eta^2 = 0.034$).

Post-hoc pairwise comparisons revealed that the 0.8 m wide *Curb & Barrier* condition exhibited a statistically significantly higher ($p < 0.05$) steering rate RMS and roll rate RMS compared to the 1 m and 1.2 m *Curb & Barrier* conditions.

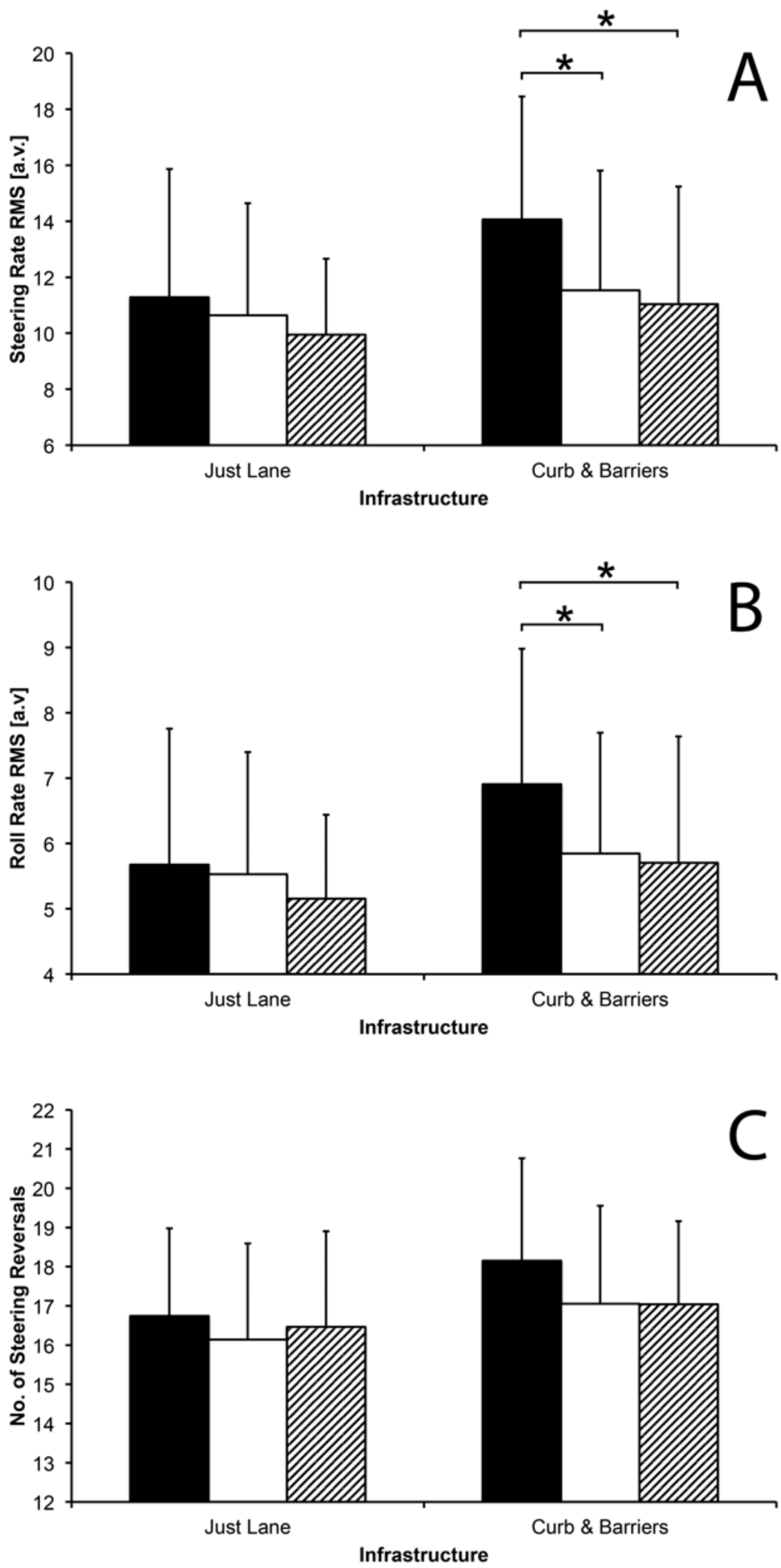


Figure 5.2: Steering rate root mean square (A), roll rate root mean square (B) and number of steering reversals (C) at different infrastructures and lane widths (black column, 80 centimetres; white column, 100 centimetres; patterned column, 120 centimetres). *, statistically significant change (p < 0.05)

5.04.2 Gap passing

Results for the *Gap passing* conditions are graphically presented in Figure 5.3. There was a statistically significant *main effect* of the segment for normalised steering rate RMS ($F_{1,17} = 31.36$; $p = 0.000$; $\eta^2 = 0.648$) and normalised roll rate RMS ($F_{1,17} = 9.69$; $p = 0.006$; $\eta^2 = 0.363$). Both, normalised steering rate RMS and normalised roll rate RMS exhibited a statistically significant *main effect* of the width ($F_{2,34} = 4.732$; $p = 0.015$; $\eta^2 = 0.218$ and $F_{2,34} = 12.38$; $p = 0.000$; $\eta^2 = 0.421$, respectively). *Segment x width* interaction was not statistically significant for normalised steering rate RMS ($F_{2,34} = 2.10$; $p = 0.991$; $\eta^2 = 0.001$) or normalised roll rate RMS ($F_{2,34} = 1.26$; $p = 0.296$; $\eta^2 = 0.069$).

Normalised steering rate RMS was statistically significantly increased at the segment before compared to the segment after the 0.8, 1 or 1.2 m wide gap ($p = 0.000$, $p = 0.009$ and $p = 0.011$, respectively). Normalised roll rate RMS was statistically significantly increased only when the gap was 1.2 m wide ($p = 0.001$), but not when it was 0.8 and 1 m wide ($p = 0.361$ and 0.125, respectively).

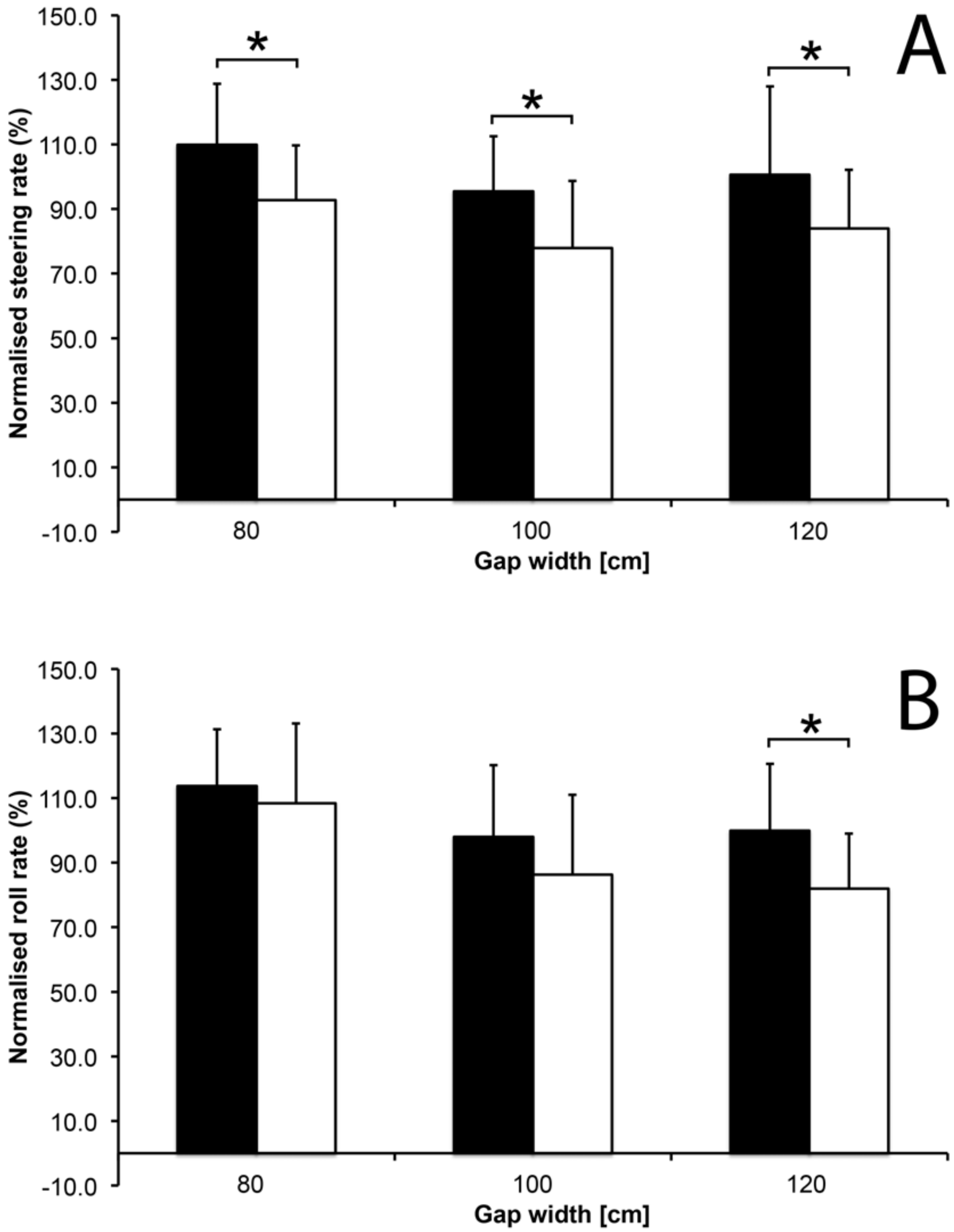


Figure 5.3: Normalised steering rate root mean square (A) and normalised roll rate root mean square (B) at different lane widths before the gap (black column) and after the gap (white column). *, statistically significant change ($p < 0.05$)

5.05 Discussion

The aim of the present study was to examine how different cycle lane designs affect one aspect of bicycle rider control. The main findings are that 1) narrower cycle lanes resulted in decreased perceived safety associated with an increase in steering and roll rate, 2) physical restriction of the manoeuvring area within the cycle lane infrastructure increased the steering and roll rate and decreased perceived safety and 3) steering and roll rate were found to be increased before passing a gap, which returned back to the baseline after the gap. Results of the present study confirm the overall hypothesis that cycle lane design affects bicycle rider control.

It has been observed that participants perceived narrower cycle lanes to be less safe than wider ones. In the present study, the narrowest width (80 cm) in some cases (*Curb & Barriers* infrastructure) resulted in a significantly reduced average speed. This phenomenon was previously also observed in children who were tested indoors using very narrow lanes (between 10 and 40 centimetres) (Vansteenkiste, Cardon, & Lenoir, 2015). Reduced speed in more dangerous road scenarios (e.g. narrow lanes) is also a common behaviour during driving (Godley et al., 2004). However, results on the average speed in the present study need to be interpreted with caution as the experimental protocol partially controlled the speed (fixed gear ratio). This could present a limitation of the present study, as it would also be interesting to see how cycle lane design affects bicycle rider control at different speeds.

The results demonstrated a decrease in perceived safety in the conditions where the manoeuvring area was physically restricted by the cycle lane infrastructure (*Curb & Barriers* and *Gap Passing*), with the highest perceived risk at the narrowest width. This supports previous findings on cycle lane research, where cyclists perceive cycling closer to the traffic without a clear separation significantly less safe (Chataway et al., 2014; Thomas & DeRobertis, 2013). Furthermore, safety perception was found to be associated with steering and roll rate. It is known that perception of the environment presents a common mechanism, which affects motor control. For example, standing on a beam higher from the grounds increases the perception of fear, which is consequently associated with an increase in body sway (Osler et al., 2013). The results of the present study suggest a similar behaviour with an increase in steering and rolling rate when cycling in the conditions that were perceived to be less safe.

In terms of bicycle control, no significant differences were found between the widths within the *Just Lane* condition. This suggests that, even though the lane was clearly marked, the participants did not perceive this condition hazardous enough to affect their bicycle control. Although this kind of environment is rare, it is an important observation for future studies on cycle lane design. In order to get an insight into cyclists' behaviour, it has to be studied in realistic environments. The present study used road barriers on one side, physically preventing cyclists from laterally deviate from the lane. This would probably be perceived safer when compared to moving obstacles, such as cars or heavy goods vehicles. This could also be one of the reasons why Lee et al. (2015) recommended a minimum desired width for a cycle lane to be 2 meters. In their study, they recorded lateral motion while cycling in different cycle lane designs at various speeds in traffic and observed that the essential manoeuvring area for safe cycling should be at least 2 metres.

The effect of the physical restriction was also observed when the participants in the present study cycled through a narrow gap. Steering rate was found to be significantly increased just prior to the gap compared to after passing the gap. This could be explained by the circumstance that cyclists want to ensure smooth passage in the middle of the gap with a minimal lateral deviation of the bicycle, but once they are safely through, they use less active control which results in a smaller steering rate. This type of behaviour was also observed in other forms of locomotion (Warren & Whang, 1987; Wilmut & Barnett, 2010). For example, during walking, participants rotated their shoulders to leave a safety margin when passing though an aperture, but returned back to normal posture once they had passed through (Wilmut & Barnett, 2010).

Steering and roll rate are associated with the rider's expertise. Lower values are linked with smoother steering and roll motions, which could in turn indicate better control of a bicycle (Fonda et al., 2015). The present study used steering and roll rate as the two main parameters to quantify bicycle rider control.

Due to the limitations in the experimental design of the present study, clear conclusions on what the minimal width of a cycle lane cannot be made. However, it was clearly demonstrated that the cycle lane design (infrastructure and width) affects the cyclist's perception of safety, which in turn affects the control of the bicycle. One of the limitations of the present study was also the inability to monitor the lateral position of the bicycle, which could provide more insight into how close to the physical objects participants cycled and what is the desired safety margin. This could provide more information on the essential manoeuvring space, as discussed in other studies (Lee et al., 2015). With the addition of the information on lateral position and implementing other scenarios (e.g. turns, start/stop, overtaking cyclists and various speeds), a clearer picture could be established on how cycle lanes should be designed to maximise safety. However, the results of this study demonstrate the importance of including the aspect of rider control when designing cycling facilities, although this warrants further research.

5.06 Conclusion

The present study demonstrated that cycle lane design affects the cyclist's perception of safety, steering rate and roll rate. Therefore, it can be concluded that cycle lane design affects bicycle rider control. Physical restriction of the manoeuvring area affects both perception and control compared to a condition where a lane is just marked on the ground. Furthermore, it was observed that cyclists perform steering with a higher rate at narrower lanes compared to the wider ones. These results provide valuable information for designing safer cycle lanes. Further research on cycle lane design should implement realistic infrastructure (physical restriction), but should also include other scenarios that realistically describe a regular commute.

CHAPTER 6. GENERAL DISCUSSION

6.01 Overview

Bicycle rider control has not received a lot of attention in the scientific literature, leaving some of the most fundamental components of this skill unexplained. As cycling represents an important part of sustainable transport, and with more and more people using bicycles, to increase the safety of cyclists it is crucial to understand the skill of riding a bicycle and the constraints that affect this skill. The main goal of this thesis was to explore bicycle rider control skills and examine the effects of different constraints on the control of a bicycle. To address these issues, it was first necessary to develop a valid methodology that can be used in studying rider control. The first part of the experimental chapters in this thesis (CHAPTER 2) presented three separate experiments that aimed to validate different methods and data analysis often used in studying human motor control and motor behaviour, but not always validated in the specific context of cycling: motion capturing system, electromyography and angular rate sensors. The second part (CHAPTER 3, CHAPTER 4 and CHAPTER 5) focused on constraints that were hypothesised to affect bicycle rider control.

With this thesis, it has been demonstrated that 3-dimensional kinematics should be used to get a valid interpretation of the knee angle during cycling on an ergometer. Observations from the second methodological experiment showed that several cycles (70 or more) need to be averaged to form a valid ensembled average of an EMG pattern. Furthermore, to prevent invalid temporal EMG activity interpretation, at least 4 crank position points should be used. Lastly, steering and bicycle roll rate recorded with a single angular rate sensor during outdoor cycling exhibited high inter-session reliability and can be further used to study the effects of constraints. The second part of this thesis demonstrated that expertise, body position and environment affect bicycle rider control.

This chapter (CHAPTER 6) is split into four parts starting with a chapter-by-chapter discussion. Firstly, methodology used in the present thesis is addressed and presented in a broader view. This is followed by the discussion on constraints that affect bicycle rider control with a summary of the results and what these add to theory and practice. Discussion finishes with a section on limitations, further research and main conclusions.

6.02 Methodology

An important consideration in research methodology is that the method or tool is sufficiently reliable with an acceptable measurement error (Atkinson & Nevill, 1998). Only if the data collected is reliable, sensitive and valid can the true effects of a constraint be studied. This thesis examined reliability and validity of three different measurement approaches used for studying human movement: 1) kinematics inside a lab facility, 2) muscle activity during ergometer cycling and 3) kinematics of a bicycle outside a lab facility.

6.02.1 Muscle activity

One of the initial goals of this PhD work was to study inter-muscular coordination during cycling where a rider needs to maintain balance (not ergometer cycling). However, the results from the experiment presented in section 2.02 demonstrated that over 70 cycles need to be averaged to form a reliable ensembled average. This is substantially more compared to some of the existing literature (Hug, Turpin, Couturier, & Dorel, 2011; Jobson et al., 2013). One of the reasons for a high number of cycles required to form an ensembled average could lie in the physiological variability with there being more than one way to perform the movement (Dorel, Guilhem, Couturier, & Hug, 2012). Even though pedalling is a movement on a pre-defined trajectory (i.e. trajectory of the pedal), it can still be performed in many different ways. For example, a rider can produce more force from the hip extensor muscles or from the knee extensor muscles. Either way, the same power output will be achieved but just with a different muscle activation strategy (Fonda & Sarabon, 2010), leading to cycle-to-cycle variability. Inter-muscular coordination represents an important part of bicycle control and warrants further research. However, before conducting such study, one should examine if muscle activity exhibits higher cycle-to-cycle variability when cycling outdoors with a need to balance a bicycle. As shown by Moore et al. (2011), maintaining balance is, to some extent, achieved also by laterally moving the knees. That and other variability induced outdoors (wind, surface, etc.) could influence muscle activity patterns, as well as increase the within-session variability. Therefore, it is possible that reliable and valid EMG recordings outdoor could only be achieved by even more than 70 cycles to form an ensembled average.

Based on the data from the experiment presented in this thesis (Section 2.02), in order to obtain valid EMG recordings for one minute of cycling at an average cadence of 70 rpm (i.e. 70 cycles) and at the speed of approximately 18 km/h, more than 300 m of controlled steady state cycling should be recorded. With limited equipment in the laboratory where this PhD work was conducted, this was not possible and therefore was not further studied. Also, 300 m of controlled environment (wind, obstacles) is not easy to find.

6.02.2 Kinematics

Cyclists' kinematics of the lower extremity during ergometer cycling does not change substantially (see Figure 6.1). In particular, knee angle during seated cycling can change only by lowering or raising the heel. The latter normally does not change substantially, which results in a constant knee angle (Bini & Diefenthaeler, 2009). Therefore, it is not surprising that the result of the experiment presented in section 2.01 demonstrated that, in order to obtain a reliable measure for the knee angle at the BDC, an average of five crank cycles is enough. This is substantially less compared to the minimum required number of cycles for a valid EMG ensembled average. However, with a smaller number of degrees of freedom to carry out a single pedal stroke (one complete crank cycle), it is not surprising that the movement is so reliable.

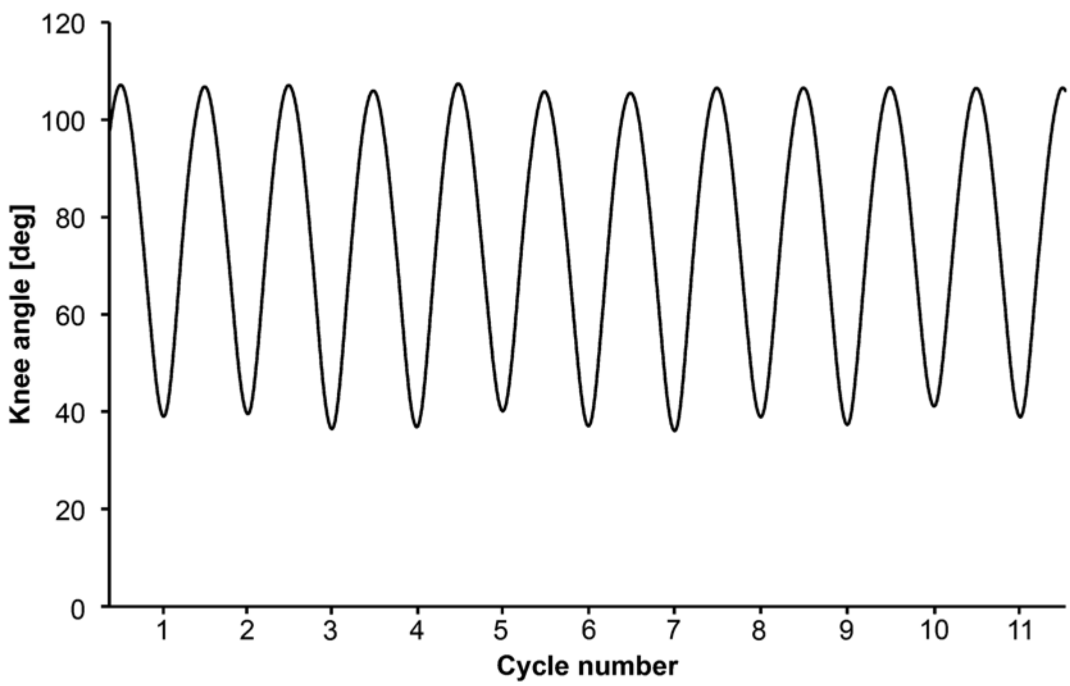


Figure 6.1: Knee angle during seated ergometer cycling recorded with a 3D motion capturing system.

Knee angle at the BDC represents one of the most common measures used in the bike fitting process (Burke, 1994; de Vey Mestdagh, 1998; Peveler & Green, 2011). However, it has never been clearly defined as to how one should obtain this measure. Section 2.01 demonstrates that a valid interpretation of the knee angle can only be obtained by using a 3D motion capturing system or a high-speed camera with a correction factor (adding 2.2° to the measured value). This experiment also demonstrated that a manual goniometer used for static measurement of the knee angle does not accurately reflect the knee angle in dynamic conditions (pedalling).

The results of this study were directly applied to the experiment examining the effect of body position on bicycle rider control (CHAPTER 4), where the knee angle at the BDC was assessed with a 3D motion capturing system. Further research on that topic should focus on standardisation procedures in the process of bike fitting with clear description of the methodology. Only that way a method can be accurately used among practitioners and clinicians.

6.02.3 Steering and roll rate

The last methodological experiment was chronologically conducted after the two experiments that had examined the effects of expertise and body position on bicycle rider control. The motivation to validate recordings outdoors with a single angular rate motion sensor came from the observations that the limited length of the cycle lane indoors, alongside the laboratory environment, presents a limitation when studying bicycle rider control. This is especially true when considering the environment as a constraint. Therefore, it had to be studied in an outdoor environment to ensure higher ecological validity. On the other hand, an outdoor environment affects perception differently than a laboratory environment, which could subsequently affect the control of a bicycle. The experiment presented in CHAPTER 3 showed high inter-session reliability of the lateral deviation, which indicates that cyclists perform the task of riding in straight line similarly on different occasions. However, the task in this experiment was in a controlled environment and over a limited length. Therefore, a mobile device with a single angular rate sensor was developed and mounted on the bicycle stem to directly measure steering rate and, to some degree, roll rate (see section 2.03) in outdoor conditions, to ensure higher ecological validity. A similar approach has been used in the previous literature (for example, see Cain (2013)), but has never been tested for reliability, especially when used in an outdoor setup. It needs to be noted that a single angular rate sensor cannot directly measure roll rate if it is mounted on the stem. As the stem rotates around two axes, roll can be influenced by the position of the stem in the transverse plane. Most of the protocols presented in this thesis involved riding in a straight line, which limits the error between the roll of the bicycle frame and roll of the stem. Despite the limitation in the experimental setup, in all of the experiments where the angular rate sensor was used, a repeated measures design has been employed. Therefore, the effects of a constraint on bicycle roll rate can be studied using a single angular rate sensor. In optimal conditions, an additional angular rate sensor mounted on the bicycle's frame should be used which would also allow direct comparisons with other studies.

From the experiment presented in CHAPTER 3 and the experiment presented in section 2.02, it can be concluded that motor behaviour during cycling exhibits high inter-session reliability. Therefore, changes in bicycle rider control (trajectory or steering/roll rate) observed in a repeated measures design reflects the effect of a constraint.

6.03 Constraints

The experiments presented in this thesis have demonstrated that the three tested constraints affect bicycle rider control. Constraints in general are considered as boundaries that can alter one's motor behaviour and can reduce or increase the degrees of freedom available to the system (Newell & van Emmerik, 1989). The three constraints examined in the present thesis were: 1) expertise, 2) body position and 3) environment.

6.03.1 Expertise

Participants from the experiment presented in CHAPTER 3 were either cyclists who had ridden a bicycle over 3000 km per year, or non-cyclists who had not ridden a bicycle for more than 100 km per year and had no history of regular cycling activity. Both groups were recruited on a basis of total kilometres they rode per year in the last 5 years. Although these may not be the best inclusion criteria, it still showed that more experienced cyclists performed significantly less lateral motion. That means they were able to control the bicycle on a predefined trajectory better than the group of less experienced cyclists.

The two groups differed in the variability, amplitude and velocity of steering and bicycle roll. Both the absolute amplitude and variability of the steering and bicycle roll are closely linked to the trajectory of riding (heading) as steering and roll work together (Schwab & Kooijman, 2010) to direct the bicycle's heading. Hence, the differences in lateral deviation and amplitude of bicycle control between the groups indicate that riders' motor behaviour across expertise does not significantly change in the way one can control the bicycle. However, the results also suggest that more experienced cyclists perform steering and roll motions in a smoother manner compared to the less experienced cyclists. Thus, steering and roll rate could represent a valid assessment of the level of expertise in cyclists. The experiment presented in CHAPTER 3 was not the first study examining cycling trajectory. Van den Ouden (2011) also examined the trajectory of riding but has not performed any further statistical analysis to quantify differences between more and less experienced cyclists. Therefore, no direct comparisons with the results from this thesis are possible.

Further subject-by-subject analysis from the experiment presented in CHAPTER 3 showed that not all cyclists from the experienced riders group performed with less lateral deviation as the cyclists from the inexperienced group. A follow up analysis revealed that cyclists from the inexperienced group who performed better had a training history from sports demanding high levels of coordination (e.g. gymnastics). It is therefore possible that the level of experience in cycling does not necessarily directly relates to the level of cycling-specific experience.

Observations from this thesis that steering and roll rate present a sensitive measure to distinguish between more and less experienced riders support some of the previous findings on the effect of expertise or training status in other tasks (Sarabon, Mlaker, & Markovic, 2010; Sarabon & Rosker, 2013). For example, frequency of oscillations during quiet standing is a common approach to assess body sway performance (Panjan & Sarabon, 2010). Furthermore, an increased steering rate is related to poorer driving performance and is therefore commonly used as a performance measure (Brookhuis, de Vries, & de Waard, 1991; Sturgis, 1982). Thus, it is not surprising that the rate of steering and bicycle roll change was found to be the most sensitive measure when examining the effects of constraints.

6.03.2 Seat height

Changing the seat height on a bicycle is probably the first modification a cyclist makes after purchasing a new bicycle. Although the search for the optimal seat height is a popular subject for improving physical performance and for preventing the occurrence of non-traumatic injuries (e.g. see Bini et al., (2011)), it has never been addressed to improve bicycle rider control. The results of the experiment presented in CHAPTER 4 show that alterations in body position, which were achieved by changing the seat height, have an effect on bicycle rider control. One of the observations from this experiment was that the trajectory of riding did not significantly change at different seat heights. On the other hand, there were significant changes in steering and roll rate and angle, with lower values at seat heights set 3% lower than previously recommended (Hamley & Thomas, 1967). That could indicate that cyclists control the trajectory of riding at less optimal body positions differently compared to the more optimal positions, which was shown in the variability of steering angle and steering roll rate. Based on these results, it was concluded that the seat height, set to 106% and 103% of the inner leg length for males and females respectively, improves control of the bicycle.

A decrease in steering and bicycle roll rate at a lower seat height are somehow expected as a lower centre of gravity presents better stability (Rosker et al., 2011). On the other hand, based on the observations from the experiment by Kooijman et al. (2011), it could also be hypothesised differently. Their example that a short stick balanced on end (an inverted pendulum) falls faster than a longer stick has been experimentally confirmed with a self-stable bicycle that contained counter-rotating wheels to eliminate the angular momentum and without a trail (see Figure 1.1). They have shown that this bicycle achieves self-stability due to the two frames, with a smaller mass at the front frame. A smaller frame falls faster than a bigger frame (rear frame), causing steering in the direction of the fall and maintaining stability. Based on these observations and examples, it could be hypothesised, theoretically, that by cycling with a seat height set higher than recommended, the centre of mass would be positioned higher with more self-stability of the bicycle and less active rider control (steering and bicycle roll). However, as shown in CHAPTER 4, this was not the case as smaller steering and bicycle roll rates were observed at the lower seat heights.

The reader should keep in mind that, regardless of these recommendations, the term “optimal” or “better” needs to be interpreted with caution, as only one aspect of bicycle rider control was examined (straight line cycling). For example, quick turns to avoid obstacles, cornering, etc. have not been examined but represent common and realistic movements that happen during cycling.

6.03.3 Cycle lane design

The last experiment of this PhD work aimed to examine the effects of the environmental constraints on the bicycle rider control. Regardless of the recommendations on the design, especially the width of a cycle lane (Winters et al., 2013), it can often be observed that cycle lanes are constructed poorly with limited space that substantially limits the manoeuvring area (Frings et al., 2014).

As demonstrated in the last empirical experiment of this PhD work, perception of the environment, especially when the infrastructure physically restricts the manoeuvring area, significantly affects bicycle rider control. Steering rate and roll rate were found to increase with an increased environmental demand (narrower cycle lane). The main conclusion from CHAPTER 5 was that perception of the environment affects the control of the bicycle. Even though, all conditions allowed cyclists to smoothly navigate along the cycle lane, an increase in steering and roll rate was observed for narrower cycle lanes. However, this phenomenon was mainly observed for the conditions where the infrastructure physically restricted the manoeuvring area, which means that conditions where a cycle lane has no physical restriction, just floor markings, does not represent hazardous conditions. Based on these observations, it can be concluded that the fear from a perceived hazardous situation affects motor behaviour during cycling. This was also previously shown during quiet standing on a balance beam higher from the ground, which provoked a fear of falling from a height (postural threat) and resulted in increased arousal and altered cognition (Adkin, Frank, Carpenter, & Peysar, 2002; Osler et al., 2013; Tersteeg, Marple-Horvat, & Loram, 2012). Moreover, Tersteeg et al., (2012) concluded that postural threat represents one of the most important mechanisms for disturbed locomotion and balance.

In the experiment presented in CHAPTER 5, the infrastructure with the gap presented a condition with a physical obstacle on both sides of a cyclist. It was shown that just prior to the gap, cyclists significantly increase the rate of steering and bicycle roll but once they pass the gap, both steering and roll rate decrease. It can be hypothesised that the fear of hitting either side of the gap (vertical poles) provokes changes in bicycle rider control. An increase in steering and roll rate is probably present to ensure as straight a trajectory as possible to pass through the gap, avoiding contact with any of the poles. This change in motor behaviour is similar to the one reported by Wilmut & Barnett (2010), who observed shoulder rotations during locomotion when navigating through a narrow gap even though the gap was wide enough for a safe passage without any postural alterations.

The results from the studies on locomotion (Wagman & Taylor, 2005; Warren, Kay, Zosh, Duchon, & Sahuc, 2001; Wilmut & Barnett, 2010) can not be directly related to the results found in CHAPTER 5 because any postural modification during cycling would not provide any means to make the bicycle-plus-rider system narrower. Therefore, the only way that a cyclist would pass a narrow gap as safely as possible is to ensure that the bicycle passes the gap as centrally as possible with minimum lateral movement. This is most likely achieved with an increased steering effort, which is reflected by an increase in steering rate just prior to passing the gap. Once the cyclists had passed the gap, steering and roll rate decreased back to, or even below, the baseline.

Results from the experiment presented in CHAPTER 5 have practical implications, not least that highway engineers and policy makers need to consider bicycle rider control when designing cycling facilities. However, this topic warrants more research to examine other aspects important to consider when designing cycling facilities to promote greater perceived and actual safety.

6.04 Limitations and further research

The work presented in this thesis addressed several novel areas that have not yet been examined and this led to some of the methodological hurdles that had to first be explored. Each experiment presented in this thesis has certain limitations which were partially discussed at the end of each empirical chapter, and will not be repeated thenceforth.

The main limitation of this thesis is that primarily, only one aspect of cycling has been examined, i.e. riding in a straight line. Although this represents a major part of a regular commute (Kooijman et al., 2009), it still leaves out some very important secondary parts, such as corners, traffic light stops and overtaking cyclists to name but a few. Further research should therefore address other aspects of cycling such as these.

The greatest impact of safety would most likely be achieved by the implementation of more and safer cycling facilities, such as cycle lanes. Therefore, the author of this thesis sees this area as the most important for further research. That should include research on cycle lane design at various speeds, cornering, two-way lanes and position of the cycle lane (e.g. on a pavement or as part of the road). The present thesis demonstrated that cycle lane design affects bicycle rider control during straight line cycling. Similar equipment set-ups and protocols could be used for future research. However, research should focus on realistic scenarios that are easy to implement in urban environments and not just to find “optimal” designs.

6.05 Conclusion

The present thesis provides some insight into bicycle rider control skills. With a series of methodological questions, it has been shown what tools and protocols could be used in assessment of bicycle rider control. Moreover, it has been demonstrated that in order to increase cyclists’ safety; expertise, body position and cycle lane design need to be taken into account. Although none of the chapters completely close the story, it has been clearly demonstrated that constraints affect bicycle rider control and are indisputably important in attempts to increase cyclists’ safety.

CHAPTER 7. REFERENCES

Adkin, A. L., Frank, J. S., Carpenter, M. G., & Peysar, G. W. (2002). Fear of falling modifies anticipatory postural control. *Experimental Brain Research*, *143*(2), 160–70. doi:10.1007/s00221-001-0974-8

Andersen, L., Schnohr, P., Schroll, M., & Hein, H. (2000). ALl-cause mortality associated with physical activity during leisure time, work, sports, and cycling to work. *Archives of Internal Medicine*, *160*(11), 1621–1628. doi:10.1001/archinte.160.11.1621

Ashe, M. C., Scroop, G. C., Frisken, P. I., Amery, C. A., Wilkins, M. A., & Khan, K. M. (2003). Body position affects performance in untrained cyclists. *British Journal of Sports Medicine*, *37*(5), 441–444.

Atkinson, G., & Nevill, A. M. (1998). Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports Medicine*, *26*(4), 217–238.

Bartlett, R., Wheat, J., & Robins, M. (2007). Is movement variability important for sports biomechanists? *Sports Biomechanics*, *6*(2), 224–43. doi:10.1080/14763140701322994

Bauman, A. E., & Rissel, C. (2009). Cycling and health: an opportunity for positive change. *Medical Journal of Australia*, *190*(7), 347–348.

Bernstein, N. A. (1967). *The co-ordination and regulation of movements*. New York, USA: Pergamon Press.

Billaut, F., Basset, F. a, & Falgairette, G. (2005). Muscle coordination changes during intermittent cycling sprints. *Neuroscience Letters*, *380*(3), 265–269. doi:http://dx.doi.org/10.1016/j.neulet.2005.01.048

Bini, R., & Diefenthaeler, F. (2009). Mechanical Work And Coordinative Pattern Of Cycling: A Literature Review. *Kinesiology*, *41*(1), 25–39.

Bini, 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. doi:10.2165/11588740-000000000-00000

Bini, R., Hume, P., & Kilding, A. (2014). Saddle height effects on pedal forces, joint mechanical work and kinematics of cyclists and triathletes. *European Journal of Sport Science*, *14*(1), 44–52. doi:10.1080/17461391.2012.725105

Bini, R. R., Hume, P. A., Lanferdini, F. J., & Vaz, M. A. (2013). Effects of moving forward or backward on the saddle on knee joint forces during cycling. *Physical Therapy in Sport*, *14*(1), 23–27. doi:http://dx.doi.org/10.1016/j.ptsp.2012.02.003

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

Brookhuis, K. A., de Vries, G., & de Waard, D. (1991). The effects of mobile telephoning on driving performance. *Accident Analysis & Prevention*, *23*(4), 309–316. doi:10.1016/0001-4575(91)90008-S

Bruce, E. N., Goldman, M. D., & Mead, J. (1977). A digital computer technique for analyzing respiratory muscle EMG's. *Journal of Applied Physiology*, *43*(3), 551–556.

Burke, E. R. (1994). Proper fit of the bicycle. *Clinics in Sports Medicine*, *13*(1), 1–14.

Cain, S. M. (2013). *An Experimental Investigation of Human/Bicycle Dynamics and Rider Skill in Children and Adults*. University of Michigan.

Cain, S. M., Ashton-Miller, J. A., & Perkins, N. C. (2013). Measurement Of Human/Bicycle Balancing Dynamics And Rider Skill. In *37th annual meeting of the American Society of Biomechanics, Omaha, NE* (pp. 1–2).

Cain, S. M., & Perkins, N. C. (2012). Comparison of experimental data to a model for bicycle steady-state turning. *Vehicle System Dynamics*, *50*(8), 1341–1364. doi:10.1080/00423114.2011.650181

Cain, S., Ulrich, D., & Perkins, N. (2012). Quantifying Increased Skill Using Bicycle Kinematics as a Riders Learn to Ride Bicycles. In *ASME 2012 5th Annual Dynamic Systems and Control Conference joint with the JSME 2012 11th Motion and Vibration Conference* (pp. 195–199).

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.

Chapman, L. (2007). Transport and climate change: a review. *Journal of Transport Geography*, *15*(5), 354–367. doi:http://dx.doi.org/10.1016/j.jtrangeo.2006.11.008

Chataway, E. S., Kaplan, S., Nielsen, T. A. S., & Prato, C. G. (2014). Safety perceptions and reported behavior related to cycling in mixed traffic: A comparison between Brisbane and Copenhagen. *Transportation Research Part F: Traffic Psychology and Behaviour*, *23*(0), 32–43. doi:http://dx.doi.org/10.1016/j.trf.2013.12.021

Clark, J. E. (1995). On becoming skillful: patterns and constraints. *Research Quarterly for Exercise and Sport*, *66*(3), 173–183. doi:10.1080/02701367.1995.10608831

Connick, M. J., & Li, F.-X. (2013). The impact of altered task mechanics on timing and duration of eccentric bi-articular muscle contractions during cycling. *Journal of Electromyography and Kinesiology*, *23*(1), 223–229. doi:http://dx.doi.org/10.1016/j.jelekin.2012.08.012

Cross, K., & Fisher, G. (1977). A study of bicycle/motor-vehicle accidents: Identification of problem types and countermeasure approaches.

Davids, K., Glazier, P., Ara??jo, D., & Bartlett, R. (2003). Movement Systems as Dynamical Systems. *Sports Medicine*, *33*(4), 245–260. doi:10.2165/00007256-200333040-00001

Davis, R. M. (2001). BMJ bans “accidents.” *BMJ*, *322*(7298), 1320–1321. doi:10.1136/bmj.322.7298.1320

De Geus, B., Joncheere, J., & Meeusen, R. (2009). Commuter cycling: effect on physical performance in untrained men and women in Flanders: minimum dose to improve indexes of fitness. *Scand J Med Sci Sports*, *19*(2), 179–187. doi:10.1111/j.1600-0838.2008.00776.x

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

DfT. (2008). *Cycle infrastructure desing. Local Transport Note 02/08*. London, UK. Retrieved from www.dft.gov.uk

Dorel, S., Couturier, A., & Hug, F. (2008). Intra-session repeatability of lower limb muscles activation pattern during pedaling. *Journal of Electromyography and Kinesiology*, *18*(5), 857–865.

Dorel, S., Guilhem, G., Couturier, A., & Hug, F. (2012). Adjustment of muscle coordination during an all-out sprint cycling task. *Medicine and Science in Sports and Exercise*, *44*(11), 2154–64. doi:10.1249/MSS.0b013e3182625423

Ducheyne, F., De Bourdeaudhuij, I., Lenoir, M., & Cardon, G. (2014). Effects of a cycle training course on children’s cycling skills and levels of cycling to school. *Accident Analysis & Prevention*, *67*(0), 49–60. doi:http://dx.doi.org/10.1016/j.aap.2014.01.023

Ducheyne, F., De Bourdeaudhuij, I., Lenoir, M., Cardon, G., & Spittaels, H. (2013). Children’s cycling skills: Development of a test and determination of individual and environmental correlates. *Accident Analysis & Prevention*, *59*(0), 38–45. doi:http://dx.doi.org/10.1016/j.aap.2012.06.021

Ericsson, K. A. (2014). Why expert performance is special and cannot be extrapolated from studies of performance in the general population: A response to criticisms. *Intelligence*, *45*, 81–103. doi:10.1016/j.intell.2013.12.001

Fernández-Peña, E., Lucertini, F., & Ditroilo, M. (2009). A maximal isokinetic pedalling exercise for EMG normalization in cycling. *Journal of Electromyography and Kinesiology*, *19*(3), e162–170. doi:10.1016/j.jelekin.2007.11.013

Ferrer-Roca, V., Bescós, R., Roig, A., Galilea, P., Valero, O., García-López, J., … Garcia-Lopez, J. (2014). Acute effects of small changes in bicycle saddle height on gross efficiency and lower limb kinematics. *Journal of Strength and Conditioning Research*, *28*(3), 784–791. doi:10.1519/JSC.0b013e3182a1f1a9

Ferrer-Roca, V., Roig, A., Galilea, P., & García-López, J. (2012). Influence of Saddle Height on Lower Limb Kinematics in Well-Trained Cyclists: Static Vs. Dynamic Evaluation in Bike Fitting. *Journal of Strength & Conditioning Research*, *26*(11), 3025–3029.

Fleiss, J. L. (1999). Reliability of Measurement. In *The Design and Analysis of Clinical Experiments* (pp. 1–32). John Wiley & Sons, Inc. doi:10.1002/9781118032923.ch1

Fonda, B., Panjan, A., Markovic, G., & Sarabon, N. (2011). Adjusted saddle position counteracts the modified muscle activation patterns during uphill cycling. *Journal of Electromyography and Kinesiology*, *21*(5), 854–860.

Fonda, B., & Sarabon, N. (2010). Biomechanics of Cycling. *Sport Science Review*, *19*(1/2), 187–210. doi:10.2478/v10237-011-0012-0

Fonda, B., Sarabon, N., Blacklock, R., & Li, F. (2014). Effects of changing seat height on bike handling. In *Journal of Science and Cycling* (Vol. 3, p. 19).

Fonda, B., Sarabon, N., & Li, F.-X. (2014). Validity and reliability of different kinematics methods used for bike fitting. *Journal of Sport Sciences*, *32*(10), 940–946. doi:10.1080/02640414.2013.868919

Fonda, B., Sarabon, N., & Li, F.-X. (2015). Bicycle Rider Control Skills: expertise and assessment. *Journal of Sport Sciences*, *in press*. doi:10.1080/02640414.2015.1039049

Frings, D., Parkin, J., & Ridley, A. M. (2014). The effects of cycle lanes, vehicle to kerb distance and vehicle type on cyclists' attention allocation during junction negotiation. *Accident; Analysis and Prevention*, *72*, 411–21. doi:10.1016/j.aap.2014.07.034

Gámez, J., Zarzoso, M., Raventós, A., Valero, M., Alcántara, E., López, A., … Vera, P. (2008). Determination of the Optimal Saddle Height for Leisure Cycling (P188). In *The Engineering of Sport 7* (pp. 255–260).

Gibson, J. J. (1979). *The ecological approach to visual perception*. Boston: Houghton Mifflin Company.

Godley, S. T., Triggs, T. J., & Fildes, B. N. (2004). Perceptual lane width, wide perceptual road centre markings and driving speeds. *Ergonomics*, *47*(3), 237–256. doi:10.1080/00140130310001629711

Gregor, R. J., Broker, J. P., & Ryan, M. M. (1991). The biomechanics of cycling. *Exercise and Sport Sciences Reviews*, *19*, 127–169.

Hamer, M., & Chida, Y. (2008). Active commuting and cardiovascular risk: a meta-analytic review. *Prev Med*, *46*(1), 9–13. doi:10.1016/j.ypmed.2007.03.006

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

Harris, M. A., Reynolds, C. C. O., Winters, M., Chipman, M., Cripton, P. A., Cusimano, M. D., & Teschke, K. (2011). The Bicyclists' Injuries and the Cycling Environment study: a protocol to tackle methodological issues facing studies of bicycling safety. *Injury Prevention : Journal of the International Society for Child and Adolescent Injury Prevention*, *17*(5), e6. doi:10.1136/injuryprev-2011-040071

Hendriksen, I. J. M., Simons, M., Garre, F. G., & Hildebrandt, V. H. (2010). The association between commuter cycling and sickness absence. *Preventive Medicine*, *51*(2), 132–135. doi:http://dx.doi.org/10.1016/j.ypmed.2010.05.007

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology*, *10*(5), 361–374.

Hillman, M. (1993). Cycling and the promotion of health. *Policy Studies*, *14*(2), 49–58. doi:10.1080/01442879308423639

Hillman, M., & Morgan, D. (1992). *Cycling: towards health and safety*. Oxford University Press.

Holmes, J. C., Pruitt, A. L., & Whalen, N. J. (1994). Lower extremity overuse in bicycling. *Clinics in Sports Medicine*, *13*(1), 187–205.

Hubbard, M., Hess, R., Moore, J. K., & Peterson, D. L. (2011). Human Control of Bicycle Dynamics with Experimental Validation and Implications for Bike Handling and Design. In *NSF Engineering Research and Innovation Conference*. Atlanta, Georgia.

Hug, F. (2011). Can muscle coordination be precisely studied by surface electromyography? *Journal of Electromyography and Kinesiology*, *21*(1), 1–12. doi:10.1016/j.jelekin.2010.08.009

Hug, F., & Dorel, S. (2009). Electromyographic analysis of pedaling: a review. *Journal of Electromyography and Kinesiology*, *19*(2), 182–198.

Hug, F., Turpin, N. a, Couturier, A., & Dorel, S. (2011). Consistency of muscle synergies during pedaling across different mechanical constraints. *Journal of Neurophysiology*, *106*(1), 91–103. doi:10.1152/jn.01096.2010

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.

Hull, T., Wootten, D., Boyd, T., & Hull, M. L. (1996). An improved accuracy six-load component pedal dynamometer for cycling. *Journal of Biomechanics*, *9290*(95), 1105–1110.

Illingworth, C. M., Noble, D., Bell, D., Kemn, I., Roche, C., & Pascoe, J. (1981). 150 bicycle injuries in children: a comparison with accidents due to other causes. *Injury*, *13*(1), 7–9. doi:http://dx.doi.org/10.1016/0020-1383(81)90083-8

Jacobs, R., & Macpherson, J. M. (1996). Two functional muscle groupings during postural equilibrium tasks in standing cats. *Journal of Neurophysiology*, *76*(4), 2402–11.

Jacobsen, P. L. (2003). Safety in numbers: more walkers and bicyclists, safer walking and bicycling. *Injury Prevention*, *9*(3), 205–209. doi:10.1136/ip.9.3.205

Janssen, W. G. M., Bussmann, H. B. J., & Stam, H. J. (2002). Determinants of the Sit-to-Stand Movement: A Review. *Physical Therapy*, *82*(9), 866–879.

Jobson, S. A., Hopker, J., Arkesteijn, M., & Passfield, L. (2013). Inter- and intra-session reliability of muscle activity patterns during cycling. *Journal of Electromyography and Kinesiology:*, *23*(1), 230–237.

Knowls, J., Adams, S., Cuerden, R., Savill, S., Tight, M., & Limited, T. R. L. (2009). *Collisions Involving Cyclists on Britain's Roads: Establishing the Causes* (Vol. PPR 445). Department for Transport.

Kooijman, J. D. G., Meijaard, J. P., Papadopoulos, J. M., Ruina, A., & Schwab, a L. (2011). A bicycle can be self-stable without gyroscopic or caster effects. *Science (New York, N.Y.)*, *332*(6027), 339–342. doi:10.1126/science.1201959

Kooijman, J. D. G., & Schwab, A. L. (2013). A review on bicycle and motorcycle rider control with a perspective on handling qualities. *Vehicle System Dynamics*, *51*(11), 1722–1764. doi:10.1080/00423114.2013.824990

Kooijman, J. D. G., Schwab, A. L., & Moore, J. K. (2009). Some observations on human control of a bicycle. *ASME 2009 International Design and Engineering Technical Conferences & Computers and Information in Engineering Conference*. San Diego, California, USA.

Laplaud, D., Hug, F., & Grélot, L. (2006). Reproducibility of eight lower limb muscles activity level in the course of an incremental pedaling exercise. *Journal of Electromyography and Kinesiology*, *16*(2), 158–166.

Latash, M. L., Scholz, J. P., & Schöner, G. (2002). Motor Control Strategies Revealed in the Structure of Motor Variability. *Exercise and Sport Sciences Reviews*, *30*(1), 26–31.

Lee, C., Shin, H. C., Kang, S., & Lee, J.-B. (2015). Measurement of desirable minimum one-way bike lane width. *KSCE Journal of Civil Engineering*. doi:10.1007/s12205-015-0467-0

Lee, D. N., & Lishman, J. R. (1975). Visual proprioceptive control of stance. *Journal of Human Movement Studies*.

Legg, S. J., Laurs, E., & Hedderley, D. I. (2003). How safe is cycling with a schoolbag? *Ergonomics*, *46*(8), 859–869. doi:10.1080/0014013031000107586

Li, L. (2004). Neuromuscular control and coordination in cycling. *Research Quarterly for Exercise and Sport*, *75*(1), 16–22.

Lindström, M. (2008). Means of transportation to work and overweight and obesity: A population-based study in southern Sweden. *Preventive Medicine*, *46*(1), 22–28. doi:http://dx.doi.org/10.1016/j.ypmed.2007.07.012

Lloyd, D. (2013). *Reported road casualties in Great Britan: annual report 2012*. (D. Lloyd, Ed.)*Statistical Release*. Department for Transport.

Macarthur, C., Parkin, P. C., Sidky, M., & Wallace, W. (1998). Evaluation of a bicycle skills training program for young children: a randomized controlled trial. *Injury Prevention*, *4*(2), 116–121.

Maimaris, C., Summer, C. L., Browning, C., & Palmer, C. R. (1994). Injury patterns in cyclists attending an accident and emergency department: a comparison of helmet wearers and non-wearers. *British Medical Journal*, *308*(6943), 1537–1540. doi:10.1136/bmj.308.6943.1537

Mathias, J. P., Barsi, G. I., van de Ruit, M., & Grey, M. J. (2014). Rapid acquisition of the transcranial magnetic stimulation stimulus response curve. *Brain Stimulation*, *7*(1), 59–65. doi:10.1016/j.brs.2013.08.003

McLean, J. R., & Hoffmann, E. R. (1972). The effects of lane width on driver steering control and performance. In *Proceedings of the Sixth Australian Road Research Board Conference* (pp. 418–440).

Meijaard, J. P. P. P., Papadopoulos, J. M., Ruina, A., & Schwab, a. L. L. (2007). Linearized dynamics equations for the balance and steer of a bicycle: a benchmark and review. *Proceedings of the Royal Society A: Mathematical, Physical and Engineering Science*, *463*(2084), 1955–1982. doi:10.1098/rspa.2007.1857

Moore, J. K. (2009). *Human Control of a Bicycle (unpublished doctoral dissertation)*. *Mechanical and Aerospace Engineering*. University of California, Davis.

Moore, J. K., Kooijman, J. D., Schwab, A. L., & Hubbard, M. (2011). Rider motion identification during normal bicycling by means of principal component analysis. *Multibody System Dynamics*, *25*, 1–20.

Newell, A., Rosenbloom, P. S., & Anderson, J. R. (1981). Mechanisms of skill acquisition and the law of practice. In *Cognitive skills and their acquisition* (pp. 1–55). New York: Erlbaum.

Newell, K. M., & van Emmerik, R. E. A. (1989). The acquisition of coordination: Preliminary analysis of learning to write. *Human Movement Science*, *8*(1), 17–32. doi:10.1016/0167-9457(89)90021-3

Noakes, T. (1995). Fatal Cycling Injuries. *Sports Medicine*, *20*(5), 348–362. doi:10.2165/00007256-199520050-00006

Nordeen-Snyder, K. S. (1977). The effect of bicycle seat height variation upon oxygen consumption and lower limb kinematics. *Medicine and Science in Sports*, *9*(2), 113–117.

Oja, P., Titze, S., Bauman, A., de Geus, B., Krenn, P., Reger-Nash, B., & Kohlberger, T. (2011). Health benefits of cycling: a systematic review. *Scandinavian Journal of Medicine & Science in Sports*, *21*(4), 496–509. doi:10.1111/j.1600-0838.2011.01299.x

Osler, C. J., Tersteeg, M. C. A., Reynolds, R. F., & Loram, I. D. (2013). Postural threat differentially affects the feedforward and feedback components of the vestibular-evoked balance response. *The European Journal of Neuroscience*, *38*(8), 3239–47. doi:10.1111/ejn.12336

Panjan, A., & Sarabon, N. (2010). Review of methods for the evaluation of human body balance. *Sport Science Review*, *19*(5/6), 131–163.

Parkin, J., & Meyers, C. (2010). The effect of cycle lanes on the proximity between motor traffic and cycle traffic. *Accident; Analysis and Prevention*, *42*(1), 159–65. doi:10.1016/j.aap.2009.07.018

Peveler, W., Pounders, J., & Bishop, P. (2007). Effects of saddle height on anaerobic power production in cycling. *Journal of Strength & Conditioning Research*, *21*(4), 1023–1027. doi:10.1519/r-20316.1

Peveler, W. W. (2008). Effects of Saddle Height on Economy in Cycling. *The Journal of Strength & Conditioning Research*, *22*(4), 1355–1359 10.1519/JSC.0b013e318173dac6.

Peveler, W. W., Bishop, P., Smith, J., Richardson, M., Whitehorn, E., Astorino, T., … Dalleck, L. (2005). Comparing methods for setting saddle height in trained cyclists. *JEP 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 & Conditioning Research*, *25*(3), 629–633.

Potter, J. J., Sauer, J. L., Weisshaar, C. L., Thelen, D. G., & Ploeg, H.-L. (2008). Gender differences in bicycle saddle pressure distribution during seated cycling. *Medicine and Science in Sports and Exercise*, *40*(6), 1126–1134. doi:10.1249/MSS.0b013e3181666eea

Prokop, T., Schubert, M., & Berger, W. (1997). Visual influence on human locomotion Modulation to changes in optic flow. *Experimental Brain Research*, *114*(1), 63–70. doi:10.1007/PL00005624

Prothero, R. (2014). *2011 Census Analysis - Cycling to Work*. Office for National Statistics.

Pruitt, A. L. (2006). *Andy Pruitt's Complete Medical Guide for Cyclists*. VeloPress.

Pucher, J., Dill, J., & Handy, S. (2010). Infrastructure, programs, and policies to increase bicycling: An international review. *Preventive Medicine*, *50, Supple*(0), S106–S125. doi:http://dx.doi.org/10.1016/j.ypmed.2009.07.028

Raasch, C. C., & Zajac, F. E. (1999). Locomotor strategy for pedaling: muscle groups and biomechanical functions. *Journal of Neurophysiology*, *82*(2), 515–525.

Rankine, W. J. M. (1869). On the dynamical principles of the motion of velocipedes. *The Engineer*, *28*(79), 129.

Raymond, C. H., Joseph, K. F., Gabriel, Y. F., So, R. C. H. H., Ng, J. K.-F., & Ng, G. Y. F. F. (2005). Muscle Recruitment Pattern in Cycling. *Physical Therapy in Sport*, *6*(2), 89–96. doi:10.1016/j.ptsp.2005.02.004

Reynolds, C. C. O., Harris, M. A., Teschke, K., Cripton, P. A., & Winters, M. (2009). The impact of transportation infrastructure on bicycling injuries and crashes: a review of the literature. *Environmental Health : A Global Access Science Source*, *8*(1), 47. doi:10.1186/1476-069X-8-47

Rosker, J., Markovic, G., & Sarabon, N. (2011). Effects of vertical center of mass redistribution on body sway parameters during quiet standing. *Gait & Posture*, *33*(3), 452–456. doi:http://dx.doi.org/10.1016/j.gaitpost.2010.12.023

Ryan, M. M., & Gregor, R. (1992). EMG profiles of lower extremity muscles during cycling at constant workload and cadence. *Journal of Electromyography and Kinesiology*, *2*(2), 69–80.

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), e465–471.

Sarabon, N., Fonda, B., & Markovic, G. (2012). Change of muscle activation patterns in uphill cycling of varying slope. *European Journal of Applied Physiology*, *112*(7), 2615–2623. doi:10.1007/s00421-011-2236-1

Sarabon, N., Mlaker, B. B., & Markovic, G. (2010). A novel tool for the assessment of dynamic balance in healthy individuals. *Gait & Posture*, *31*(2), 261–264. doi:10.1016/j.gaitpost.2009.11.001

Sarabon, N., & Rosker, J. (2013). Effect of 14 days of bed rest in older adults on parameters of the body sway and on the local ankle function. *Journal of Electromyography and Kinesiology*, *23*(6), 1505–1511.

Sarabon, N., Rosker, J., Loefler, S., & Kern, H. (2013). The effect of vision elimination during quiet stance tasks with different feet positions. *Gait & Posture*, *38*(4), 708–711.

Sauer, J. L., Potter, J. J., Weisshaar, C. L., Ploeg, H., & Thelen, D. G. (2007). Influence of gender, power, and hand position on pelvic motion during seated cycling. *Medicine and Science in Sports and Exercise*, *39*(12), 2204.

Schepers, P., & den Brinker, B. (2011). What do cyclists need to see to avoid single-bicycle crashes? *Ergonomics*, *54*(4), 315–327. doi:10.1080/00140139.2011.558633

Schmidt, R. A., & Lee, T. (1988). *Motor Control and Learning, 5E*. Campaign, IL: Human kinetics.

Schmidt, R. A., & Wrisberg, C. A. (2004). *Motor Learning and Performance*. Campaign, IL: Human Kinetics.

Schwab, A., & Kooijman, J. (2010). Controllability of a Bicycle. In *5th Asian Conference on Multibody Dynamics*. Kyoto, Japan.

Schwab, A. L. (2012). Bicycling safety and the lateral stability of the bicycle. In *Proceedings, International Cycling Safety Conference*. Helmond, The Netherlands.

Schwab, A. L., de Lange, P. D. L., & Moore, J. K. (2012). Rider optimal control identification in bicycling. *ASME 2012 5th Annual Dynamic Systems and Control Conference Joint with the JSME 2012 11th Motion and Vibration Conference*.

Schwab, A. L., & Meijaard, J. P. (2013). A review on bicycle dynamics and rider control. *Vehicle System Dynamics: International Journal of Vehicle Mechanics and Mobility*, *51*(7), 1059–1090. doi:10.1080/00423114.2013.793365

Scott, M. A., Li, F.-X., & Davids, K. (1997). Expertise and the regulation of gait in the approach phase of the long jump. *Journal of Sports Sciences*, *15*(6), 597–605.

Shackel, S. C., & Parkin, J. (2014). Influence of road markings, lane widths and driver behaviour on proximity and speed of vehicles overtaking cyclists. *Accident; Analysis and Prevention*, *73*, 100–8. doi:10.1016/j.aap.2014.08.015

Shennum, P. L., & DeVries, H. A. (1976). The effect of saddle height on oxygen consumption during bicycle ergometer work. *Medicine and Science in Sports*, *8*(2), 119–121.

Shiavi, R., Frigo, C., & Pedotti, A. (1998). Electromyographic signals during gait: criteria for envelope filtering and number of strides. *Medical & Biological Engineering & Computing*, *36*(2), 171–178.

Sturgis, S. P. (1982). Spectral Analysis of Driver Steering Behavior. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, *26*(3), 228–232. doi:10.1177/154193128202600309

Tersteeg, M. C. A., Marple-Horvat, D. E., & Loram, I. D. (2012). Cautious gait in relation to knowledge and vision of height: is altered visual information the dominant influence? *Journal of Neurophysiology*, *107*(10), 2686–91. doi:10.1152/jn.00875.2011

Teschke, K., Frendo, T., Shen, H., Harris, M. A., Reynolds, C. C., Cripton, P. A., … Winters, M. (2014a). Bicycling crash circumstances vary by route type: a cross-sectional analysis. *BMC Public Health*, *14*(1), 1205. doi:10.1186/1471-2458-14-1205

Teschke, K., Frendo, T., Shen, H., Harris, M. A., Reynolds, C. C., Cripton, P. A., … Winters, M. (2014b). Bicycling crash circumstances vary by route type: a cross-sectional analysis. *BMC Public Health*, *14*(1), 1205. doi:10.1186/1471-2458-14-1205

Teschke, K., Harris, M. A., Reynolds, C. C. O., Winters, M., Babul, S., Chipman, M., … Cripton., P. A. (2012). Route infrastructure and the risk of injuries to bicyclists: A case-crossover study. *American Journal of Public Health*, *102*(12), 2336–2343.

Thomas, B., & DeRobertis, M. (2013). The safety of urban cycle tracks: a review of the literature. *Accident; Analysis and Prevention*, *52*, 219–27. doi:10.1016/j.aap.2012.12.017

Turvey, M. T. (2010). Affordances and Prospective Control: An Outline of the Ontology.

Van den Ouden, J. H., Ouden, J. Van den, & den Ouden, J. H. Van. (2011). *Inventory of bicycle motion for the design of a bicycle simulator*. *Delft University of Technology, Delft (Unpublished master thesis)*. Delft University of Technology.

Vandenbulcke, G., Thomas, I., de Geus, B., Degraeuwe, B., Torfs, R., Meeusen, R., & Int Panis, L. (2009). Mapping bicycle use and the risk of accidents for commuters who cycle to work in Belgium. *Transport Policy*, *16*(2), 77–87. doi:http://dx.doi.org/10.1016/j.tranpol.2009.03.004

Vansteenkiste, P., Cardon, G., D'Hondt, E., Philippaerts, R., & Lenoir, M. (2013). The visual control of bicycle steering: The effects of speed and path width. *Accident Analysis & Prevention*, *51*(0), 222–227. doi:http://dx.doi.org/10.1016/j.aap.2012.11.025

Vansteenkiste, P., Cardon, G., & Lenoir, M. (2015). Visual guidance during bicycle steering through narrow lanes: A study in children. *Accident; Analysis and Prevention*, *78C*, 8–13. doi:10.1016/j.aap.2015.02.010

Vansteenkiste, P., Zeuwts, L., Cardon, G., Philippaerts, R., & Lenoir, M. (2014). The implications of low quality bicycle paths on gaze behavior of cyclists: A field test. *Transportation Research Part F: Traffic Psychology and Behaviour*, *23*(0), 81–87. doi:http://dx.doi.org/10.1016/j.trf.2013.12.019

Wagman, J. B., & Taylor, K. R. (2005). Perceiving affordances for aperture crossing for the person-plus-object system. *Ecological Psychology*, *17*(2), 105–130.

Walker, I. (2007). Drivers overtaking bicyclists: objective data on the effects of riding position, helmet use, vehicle type and apparent gender. *Accident; Analysis and Prevention*, *39*(2), 417–25. doi:10.1016/j.aap.2006.08.010

Warren Jr, W. H., & Whang, S. (1987). Visual guidance of walking through apertures: body-scaled information for affordances. *Journal of Experimental Psychology: Human Perception and Performance*, *13*(3), 371.

Warren, W. H., Kay, B. A., Zosh, W. D., Duchon, A. P., & Sahuc, S. (2001). Optic flow is used to control human walking. *Nature Neuroscience*, *4*(2), 213–6. doi:10.1038/84054

Whipple, F. J. W. (1899). The stability of the motion of a bicycle. *Quarterly Journal of Pure and Applied Mathematics*, *30*(120), 312–321.

Wierda, M., & Brookhuis, K. A. (1991). Analysis of cycling skill: A cognitive approach. *Applied Cognitive Psychology*, *5*(2), 113–122. doi:10.1002/acp.2350050205

Wilmut, K., & Barnett, A. L. (2010). Locomotor adjustments when navigating through apertures. *Human Movement Science*, *29*(2), 289–298.

Wilson, C., Simpson, S. E., Van Emmerik, R. E. A., & Hamill, J. (2008). Coordination variability and skill development in expert triple jumpers. *Sports Biomechanics*, *7*(1), 2–9. doi:10.1080/14763140701682983

Winter, D. a, Patla, a E., Prince, F., Ishac, M., & Gielo-Perczak, K. (1998). Stiffness control of balance in quiet standing. *Journal of Neurophysiology*, *80*(3), 1211–1221.

Winter, D. A. (1995). Human balance and posture control during standing and walking. *Gait & Posture*, *3*(4), 193–214. doi:http://dx.doi.org/10.1016/0966-6362(96)82849-9

Winters, M., Davidson, G., Kao, D., & Teschke, K. (2010). Motivators and deterrents of bicycling: comparing influences on decisions to ride. *Transportation*, *38*(1), 153–168. doi:10.1007/s11116-010-9284-y

Winters, M., & Teschke, K. (2010). Route preferences among adults in the near market for bicycling: findings of the cycling in cities study. *American Journal of Health Promotion*, *25*(1), 40–7. doi:10.4278/ajhp.081006-QUAN-236

Winters, M., Teschke, K., Babul, S., Becker, H. J. E. H. J., Brubacher, J. R., Chipman, M., … Rouphail, N. (2013). Breaking Barriers to Bicycling: Bicycle Lanes Best Practices and Pilot Treatments. *Accident; Analysis and Prevention*, *39*(2), 1–46. doi:http://dx.doi.org/10.1016/j.trf.2013.12.021