Intrinsic factors, performance and dynamic kinematics in optimisation of cycling biomechanics

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by

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I drafted the research proposal and successfully applied for ethical clearance. The data for this research project was captured with assistance of R Theo. I personally analysed all the data and drafted the current published manuscript. Apart from the normal guidance from my supervisors Dr J Swart and Dr J Fisher, I did not receive any other assistance.

Chapter 5

***W Holliday**, J Fisher, R Theo and J Swart. Cycling: Joint kinematics and muscle activity in steady state cycling. (In submission)*

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Chapter 1

A review of the literature

INTRODUCTION

Bike fitting has been described as “the detailed process of evaluating the cyclist’s physical and performance requirements, and systematically adjusting the bike to meet the goals and needs of the cyclist” (Medicine of Cycling, 2013). A properly configured bicycle is essential for optimal performance and injury prevention, and can have an influence on the cyclist’s perception of comfort (Silberman et al., 2005; Wishv-Roth, 2009). Cyclists have been searching for the optimal position to gain power whilst remaining injury free on their bicycles long before the science evolved.

The aspect of bicycle configuration that has been the focus of most studies to date regarding body position on the bicycle is the saddle height and related knee and ankle flexion angles (Bini, Hume, Croft, and Andrew Kilding, 2011). Currently there are three main methods used in clinical practice to set the saddle height: anthropometrics (inseam length and trochanteric leg length), static knee flexion angle methods and dynamic methods (during pedalling).

The saddle, however, only forms one of the contact points with the bicycle. A cyclist has three points of contact with the bicycle: the handlebars, pedals and saddle. The freely chosen bicycle configuration and subsequent cyclist kinematics, muscle recruitment patterns and physiological responses can be influenced by adjusting any of these contact points. The handlebars, pedal crank arm length, and both saddle height and fore-aft position can be adjusted to place the shoulder, hip and ankle joints in an optimal cycling position. Previous research on the correct positioning of these components is based on personal perspectives and comfort (De Vey Mestdagh, 1998; Silberman et al., 2005; Burt, 2014), whereas saddle height recommendations have been based on scientific static methods. With the advancement of technology we are now able to record the cyclist’s position in full three dimensional motion capture.

The ability to capture data during real-time cycling enables the bike fitter to assess the body position dynamically. It is known that the cycling position and muscle recruitment patterns adapt with fatigue and with maximal exertion (So, J. Ng, and G. Ng, 2005; Dingwell et al., 2008; Peveler, Shew, et al., 2012), however little is known on how training or racing at either prolonged steady state or at varying intensities affects the position and muscle activity of cyclists.

This review explores the current existing literature pertaining to the influence of intrinsic factors on individual bicycle configuration, the different methods of bicycle configuration, namely static and dynamic kinematics, as well as saddle pressure mapping, and the change in the cyclist's kinematics and muscle magnitudes during both steady state cycling and increasing workloads.

Peer-reviewed journals, books, theses and conference proceedings were searched using PUBMED and Google Scholar databases. The keywords used in the search of literature were: bicycle; biomechanics; kinetics; kinematics; EMG; static; dynamic; comfort; posture; pressure mapping. The literature was not restricted by a time period.

Optimal static saddle height configuration for performance and injury prevention

Over the years there have been numerous studies on optimal saddle height configuration (Bini, Hume, Croft, and Andrew Kilding, 2011). Some empirical methods of setting saddle height have been used by cyclists. One of the most basic methods of setting saddle height, which can be used by cyclists with limited technological knowledge or equipment, is known as the heel-toe method. The cyclist sits on the saddle and places their heel on the pedal at bottom dead centre (BDC). The saddle height is adjusted until the knee is locked in an extended position. Greg LeMond, a notable cyclist and later coach, set his saddle at a height of the leg inseam multiplied by 0.883 (Greg LeMond, 1990). Even though this method was tested in the wind tunnel and was a very popular method at the time, with new equipment such as clipless pedals, it has now become rather outdated (Burke, 2003).

Hamley and Thomas (1967) were perhaps the first to publish scientific saddle height recommendations. The test protocol involved the time required to complete a preset load of 500kg/m. To perform this test they used a progressive load increase device which provided a correlation between physiological and physical factors. The results demonstrated that the most effective saddle height for power output was at 109% of the leg inseam length. They thus proposed that for optimal cycling performance, a method of saddle height setting would be to adjust the distance from the pedal surface to the top of the saddle as measured through the seat tube to a value equal to 109% of the leg inseam length.

Perhaps the most commonly used method to date is the Holmes method (Holmes, Pruitt, and Whalen, 1994). This method recommends that the knee angle is set between 25 and 35 degrees of flexion to limit overuse knee injuries. Knee flexion angle (KFA) is measured with a goniometer with the cyclist in a stationary position, with the pedal horizontal and the crank arm in the lowest or 6 o'clock position and the pedal surface in a horizontal orientation. This is an easy and inexpensive measurement to perform (De Vey Mestdagh, 1998).

Peveler, Bishop, Smith, Richardson and Whitehorn (2005) compared these methods on setting saddle height. The inseam of the leg was measured and used to set the saddle according to both the Hamley method (inseam multiplied by 1.09), and the LeMond method (inseam multiplied by 0.883). Thirdly, the saddle was configured as per the heel-toe method described earlier. In each of these positions the KFA was measured to compare it to the Holmes method which recommends a KFA of 25-35°. The three methods fell into the Holmes recommended KFA 55-70% of the time. They concluded that for cyclists who were not prone to knee injury that the saddle height be set at 109% inseam length for optimal performance and those who may be susceptible to injury, to remain within the 25-35° range, compromising on economy yet reducing the risk of injury (Peveler, Bishop, et al., 2005).

Subsequent to Hamley's recommendations a number of studies have investigated changes in saddle height on various physiological outcomes. The effects of saddle height changes on oxygen consumption were observed during a continuous work protocol from 50 to 200 Watts, with five different saddle heights (Shennum and DeVries, 1976). This study concluded that 100% or 103% of inseam height was the most efficient, and a range of 103% to 104% of leg inseam length was therefore recommended as the optimal saddle height. A study that followed assessed economy at three different saddle heights; 95, 100 and 105% of trochanteric height (Nordeen-Snyder, 1977). It was determined that a saddle height set at 100% of trochanteric height was most efficient.

Since then studies have focused on various experimental methods of setting saddle height in order to optimise muscle activity (Jorge and Hull, 1986; Ericson, Nisell, and Nemeth, 1988) or compressive forces through the knee (McCoy and J. Gregor, 1989). Muscle activation at a saddle height set at 100% and 95% of trochanteric height was investigated, and there was an increased muscle activity in the quadriceps and hamstring muscles at the lower saddle height (Jorge and Hull, 1986). An increased muscle activation was associated with an increased muscle force, and thus the saddle height set at 95% trochanteric height was recommended. Further to this, the compressive forces through the tibiofemoral joint were assessed at three different saddle heights; 94, 100 and 106% of trochanteric height (McCoy and J. Gregor, 1989). No significant effects of saddle height on knee load during cycling were demonstrated.

Peveler, Pounders and Bishop (2007) investigated the effect of saddle height on anaerobic power production. The Hamley and Holmes methods were compared once again, with the 30 second Wingate protocol repeated at three saddle heights; 25° KFA, 35° KFA and at 109% of inseam length. There was no significant difference between trained cyclists and non-cyclists between saddle heights for peak power or mean power. Further analysis was done by dividing the subjects into those that fell within the recommended 25° to 35° KFA range and those that fell outside the recommended range when using the 109% inseam saddle height method. There was a general loss in power in those that fell outside of the recommended range, especially at saddle heights that elicited a KFA greater than 35°. There was no significant difference between saddle heights with less than 25° KFA. In a follow-on study, VO_2 was found to be significantly lower at 25° KFA (signaling greater economy) compared to the 35° KFA or 109% inseam (Peveler, 2008). It was further confirmed by another study, that a saddle set at 25° KFA was more economical in comparison to a saddle height set at 35° KFA or at 109% inseam (Peveler and Green, 2011). From this series of studies it was determined that setting a saddle height using a static method between 25-35° KFA is optimal for injury prevention and performance, and closer to 25° for a more performance focus (Peveler, Pounders, and Bishop, 2007; Peveler, 2008; Peveler and Green, 2011).

Bini, Hume and Croft (2011) have published a comprehensive review on the different methods of setting saddle height up to 2011, and despite the 25-35° KFA method considered as the golden standard for static methods of setting saddle height, further studies are being published. **Table 1.1** displays a summary of the prominent and more recently published studies.

TABLE 1.1: Studies investigating optimal saddle height.

Study	Method of setting saddle height	Outcome measures	Participants / paper type	Main results and notes	Recommendations for setting static saddle height
(Hamley and Thomas, 1967)	Percentage of inseam length	Time to exhaustion during constant load cycling	100	109% inseam length minimised time to exhaustion	109% inseam length
(Shennum and DeVries, 1976)	100, 103, 106, 109 and 112% of inseam length	VO_2, VCO_2, V_E, HR	5		Saddle set at 103-104% inseam length resulted in maximum power output
(Nordeen-Snyder, 1977)	95, 100 and 105% Trochanteric height	VO_2	10 women		100% Trochanteric saddle height most economical
(Holmes, Pruitt, and Whalen, 1994)	Knee flexion angle	Lower extremity overuse injuries	Review	To minimise knee joint load, aim for 25-35°	25-35° KFA
(Peveler, Bishop, et al., 2005)	109% inseam length (Hamley and Thomas) LeMond method Heel-toe method	To determine which method best fit into the recommended 25-35° KFA	14 male cyclists 5 female cyclists	No significant difference between Hamley and LeMond method. Significant difference between Hamley and heel-toe method. Hamley method fell into the 25-35° KFA 55% of the time	Holmes method, 25-35° KFA

(Peveler, Pounders, and Bishop, 2007)	25° KFA 35° KFA 109% inseam length (Hamley and Thomas)	Anaerobic power	9 male trained cyclists 3 non-trained male cyclists 15 female non-trained cyclists	a) Using 109% inseam to set saddle height, fell outside 25-35° KFA 63% of the time b) When outside recommended KFA, there was a loss in power, especially at lower saddle heights c) When within recommended KFA there was no difference in power	Holmes method, 25-35° KFA
(Peveler, 2008)	25° KFA 35° KFA 109% inseam length (Hamley and Thomas)	VO_2	5 male cyclists 2 male non-cyclists 9 female non-cyclists	A 25° KFA produced a significantly lower VO_2 compared to 35° KFA and 109% inseam	For increased economy, a KFA closer to 25°
(Peveler and Green, 2011)	25° KFA 35° KFA 109% inseam length (Hamley and Thomas)	VO_2 anaerobic power	11 well-trained males	Economy was better at 25° KFA compared to 35°and 109% inseam length. Power production was better at 25° compared to 109% inseam length.	For better economy and power production, a KFA closer to 25°

(Bini, Hume, and Croft, 2011)	Review of literature	a) Comparison of lower leg length measurements and knee angle methods b) Effects of saddle height on performance c) Effects of saddle height on knee injury risk	Review	a) The knee flexion angle method recommended b) Saddle height set to the Holmes method has better evidence for improved performance c) A knee flexed at 25-30° has been related to lowering the knee joint load and thus injuries	Holmes method, 25-35° KFA

KFA = knee flexion angle. HR = heart rate

Intrinsic factors related to bicycle configuration and performance

In conjunction with the Medicine of Cycling definition of bike fitting, “optimal range fit” is described as the ability to accomplish all the goals of the bike fit with the cyclist inside the optimal ranges for the type of riding they are performing (Medicine of Cycling, 2013). This is defined as the “ideal individualised position”. Similar to each cycling discipline having different demands and idealised positions, each individual cyclist will have specific goals and differing riding levels and skills, not to mention a range of body morphological characteristics. As already summarised, there is extensive research regarding the optimal saddle height and related KFA, however there is no evidence-based consensus nor defined ranges for the remaining anatomical joint ranges of motion or contact point positions on the bicycle.

There are guidelines for ankle and elbow ranges, although these are based on personal experience more than scientific data (Burke, 2003; Silberman et al., 2005; Burt, 2014). Complicated formulae to determine saddle setback, handlebar reach and handlebar drop have been investigated (De Vey Mestdagh, 1998; Iriberri, Muriel, and Larrazabal, 2008), however most bike fitting experts have suggested that the final position should be based on comfort and what appears acceptable visually (**Table 1.2**). The limitation of formulae is that they do not always take into consideration individual anthropometrics nor the pedalling characteristics of the cyclist (Peveler, Pounders, and Bishop, 2007). The range of 25° to 35° for KFA is supported by a large number of studies and can be considered the gold standard for setting saddle height scientifically for both performance (power output and economy) and injury prevention. Similar ranges should be developed using scientific methods for the remaining major joints of the body, taking into account the individual anthropometrics of the cyclists, their flexibility, riding style, comfort and performance goals.

Anthropometrics and bicycle configuration

There are several methods based on the anthropometric characteristics of the cyclist which have been described to configure the cyclist optimally on their bicycle (De Vey Mestdagh, 1998; Silberman et al., 2005). These recommendations are based on existing biomechanical research outside of the field of cycling as well as personal opinion based on experience.

Optimal saddle height, which is the most discussed aspect of bicycle configuration and is reviewed in greater detail above, is determined by leg length measurements and KFA. It has been recommended that the optimal saddle height should take into account individual femur and tibial leg length variations (Peveler, 2008). Similarly, the crank length is commonly determined by the individual leg length, with a ratio of approximately 20% of the inseam length being recommended (Gross and Bennett, 1976). It has been recommended that the length of an individual's foot and upper leg should also be taken into consideration when determining the cleat position and saddle setback respectively (De Vey Mestdagh, 1998). The handlebar reach and handlebar drop position have been described more subjectively, with full arm and upper body length described as being important considerations in optimal frame size, stem length and handlebar height (Silberman et al., 2005).

Flexibility and bicycle configuration

Assessment of the cyclist's lower back and hamstring flexibility plays an important role in bike fitting (Kotler, Babu, and Robidoux, 2016). Lumbosacral flexibility can determine how much handlebar reach and handlebar height a cyclist can tolerate, where hamstring flexibility may have an impact on saddle height.

A lower saddle height was selected by cyclists with reduced flexibility of the hamstring muscles (Burke, 2003), however this was contradicted by the results of the study by Hynd, Crowle and Stephenson (2014), who determined that hamstring flexibility did not have an effect on pre-selected saddle height. It was suggested that further studies be conducted to determine if low-level hamstring flexibility may have an influence on the cyclist's posture and bicycle configuration (Ferrer-Roca et al., 2012). A cyclist's spinal flexibility therefore may have an influence on handlebar reach and handlebar drop. This is also relevant to the cycling discipline in triathlon which is often performed as a time trial, and therefore may require a greater upper body flexion in order to reduce the frontal area for optimal aerodynamics (Burke, 2003).

Less than 50% of experienced cyclists met the recommended standards for flexibility and strength in a study conducted to assess intrinsic and extrinsic factors related to injury of club level cyclists (Dahlquist, Leisz, and Finkelstein, 2015). Only 22-25% of the participants met the required definition of 'Good-Okay' for the active knee extension test, indicating that less than a quarter of the participants had reasonable hamstring flexibility. Additionally, cyclists that had bike fits for optimal performance and aerodynamics, still had some degree of discomfort (Dahlquist, Leisz, and Finkelstein, 2015).

Comfort and individual riding style

Cyclists' perceptions of comfort should be considered as they are related to improvements in performance and injury prevention (Priego Quesada et al., 2017). An online survey identifying factors of bicycle comfort was conducted, and of the 244 respondents, 90% of the cyclists agreed that comfort is a concern when riding a bicycle, while 46% of enthusiastic cyclists agree that comfort is reached at the expense of performance (Ayachi, Dorey, and Guastavino, 2015). A recreational cyclist will generally prefer to sit more upright, whereas a competitive cyclist will want to be positioned in a more aerodynamic position, adopting a greater trunk flexion angle (Priego Quesada et al., 2017). There are further differences between cyclists, with novice cyclists demonstrating more variability in the pedalling technique than a more experienced cyclist (Peveler, Shew, et al., 2012), and these should be considered during bike fitting.

Performance goals

One of the main goals of bike fitting is to improve performance (Silberman et al., 2005; Medicine of Cycling, 2013; Burt, 2014). Optimal saddle height for performance has been well researched and discussed above, however, other variables related to bicycle configuration and power are largely based on empirical suggestions (De Vey Mestdagh, 1998; Burke, 2003; Silberman et al., 2005; Burt, 2014) and limited scientific studies. The saddle fore-aft position alters the effective seat tube angle and this may determine relative muscle contributions to the pedal force (Hayot et al., 2013). For example, a forward saddle position was associated with a greater peak of the quadriceps muscles during the first half of the crank rotation, whereas a greater peak of the plantarflexor and hamstring muscles was demonstrated at a further rearwards saddle position (Hayot et al., 2013). Further research demonstrated that a steeper effective seat tube angle resulted in a significant increase of the Rectus Femoris muscle activity, particularly during the downstroke, which is an important phase for power production (Duggan, Donne, and Fleming, 2017).

Another variable that is lacking scientific evidence is the relationship between flexibility and cycling performance. No significant differences were demonstrated between 'successful' and 'less successful' cyclists' hamstring, iliopsoas and quadriceps flexibility characteristics (Coetzee and Malan, 2018). A review of running-related literature concluded that there are mixed results with regards to flexibility and greater running economy, however, the overall opinion was that increased flexibility improved performance (Barnes and Kilding, 2014). This should be explored further with regards to cycling and optimal performance.

TABLE 1.2: Summary of guidelines for other variables of bicycle configuration.

Variable	Recommendation	Based upon	Study
Saddle setback	Formula related to upper leg length	Personal perspective	(De Vey Mestdagh, 1998)
	Plumbline and knee over pedal spindle (static)	Personal experience and recommendations	(Burke, 2003; Silberman et al., 2005; Burt, 2014)
Handlebar reach	Formula determined by arm length and torso length	Personal perspective	(De Vey Mestdagh, 1998)
	Plumbline from cyclist's nose dropped to centre of stem, hands in drops	Personal experience and recommendations	(Burke, 2003)
	Comfort in the drops, elbows flexed 60° to 70° With the knees at their maximal height and forward position, the distance between the elbows and knees should be small, 1 to 2 inches (2–5cm)	Personal experience and recommendations	(Silberman et al., 2005)
	Related to forearm length	Personal experience and recommendations	(Andy Pruitt and Matheny, 2006)
	Individual, comfort	Personal experience and recommendations	(Burt, 2014)
Handlebar height	Formula determined by arm length and torso length	Personal perspective	(De Vey Mestdagh, 1998)
	1-2 inches below saddle for small cyclists 4 inches below saddle for tall cyclists	Personal experience and recommendations	(Burke, 2003)

	Hands on the brake hoods, arms slightly flexed, the torso should flex to about 45° in relation to a non-sloping top tube	Personal experience and recommendations	(Silberman et al., 2005)
	Racer and competitive recreational cyclists' torso angle 30-45° Casual cyclist 50-60° torso angle	Personal experience and recommendations	(Andy Pruitt and Matheny, 2006)
	Individual, comfort	Personal experience and recommendations	(Burt, 2014)

Static versus dynamic methods for bike fitting

The kinematic comparison of alterations to knee and ankle angles from resting measures to active pedalling during a graded exercise protocol has been investigated, and it was established that knee and ankle angles changed significantly from a stationary position to a dynamic pedalling action measured using 2D video analysis (Peveler, Shew, et al., 2012).

Kinematics measured in 3D are however considered more accurate compared to 2D systems, as the 2D systems cannot measure movement in the transverse plane (Couto et al., 2008) and despite the small sample size, there were frontal plane differences in the ankle joint ranges between 2D and 3D (Umberger and Martin, 2001). This new dynamic method of bike fitting, which has been recommended, is now being used but is limited by the paucity of studies of the optimal ranges for dynamic KFA and other joints of the body. It is therefore important to establish new normative ranges for joint angles, as static measurements do not always agree with dynamic measurements (Ferrer-Roca et al., 2012). During the dynamic movement, cyclists generally adopt a degree of ankle plantarflexion during the late knee extension phase of the pedalling action (Peveler, Shew, et al., 2012). Other cyclists may use a combination of both ankle flexion and extension, known as ankling, when the cyclist lowers the heel with the knee extension and hip extension movement and pulls the heel up during the knee flexion phase of the pedal revolution. The chosen movement patterns will influence the kinematic chain and alter the associated KFA for any given saddle height with increased plantarflexion (pedal heel up action) also increasing the KFA.

Comparing static (Holmes method) to dynamic (photogrammetry of reflective markers for 3D analysis) measures of the lower limb joint angles during cycling, demonstrated significant changes for both the ankle and knee joint when transitioning from static to dynamic (Bini, Hume, and Croft, 2011). Ankle plantarflexion and KFA increased by approximately 8° from static to dynamic measures.

Using a 3D motion analysis system, the implications for Iliotibial Band Friction Syndrome (ITBFS) during force and repetition in cycling were investigated (Farrell, Reisinger, and Tillman, 2003). Saddle height for each subject was set with the knee at 25-30° flexion at BDC by using a goniometer before the start of the trial. During the cycling tests the KFA, as measured with a 3D motion analysis system, reached 30-35°, and upon further investigation it was determined that lateral pelvic rocking contributed 5-6° to this KFA increase. The authors thus recommended a dynamic KFA of 30-40° as the optimal range when measured dynamically.

In a more recent study, dynamic KFA's measured with an electrogoniometer and a high speed camera (2D kinematics) were significantly underestimated when compared to 3D kinematics (Fonda, Sarabon, and Li, 2014). The KFA as measured with a goniometer was also underestimated compared to 3D and 2D kinematics. The authors suggested that goniometer use should be discouraged for bike fitting, and that precise 2D video analysis can only be reached by adding a 2.2° correction factor to the knee angle assessment. Practically, they suggested that a high speed 2D video system with the correction factor would be suitable for commercial bike fitting centres, but that scientific studies should use 3D motion analysis for knee angle assessment during cycling, as it had the most valid results.

Based on these findings, Bini et al. (2011), Fonda et al. (2014) and Farrell et al. (2003) have all suggested that dynamic rather than static analysis should be used to adequately describe the lower limb cycling motion and to optimise bike fit.

There have been no studies thus far comparing the static and dynamic hip, shoulder and elbow flexion angles, and the relationship between these angles and knee and ankle flexion angles. It is therefore important to investigate the difference between static and dynamic angles measured during cycling, as this change in the kinematic chain may have an effect on performance, economy and injury risk.

TABLE 1.3: Studies comparing static and dynamic measure.

Study	Method of setting saddle height	Outcome measures	Participants	Main results and notes
(Umberger and Martin, 2001)	Matched to cyclist's own bicycle	To test that 2D and 3D models adequately represent sagittal and frontal plane lower extremity motion during cycling	4 experienced male cyclists	2D and 3D sagittal plane kinematics were similar, however 3D approach considered the more reasonable option Very few subjects
(Farrell, Reisinger, and Tillman, 2003)	KFA set at 25-30° with a goniometer	Implications for ITBFS in cycling	6 male recreational athletes 4 female recreational athletes	Sub-finding: The KFA recorded statically with a goniometer was 25-30°, however this reached 30-35° during 3D motion analysis The lateral pelvic tilt contributed approximately 5-6° to the KFA increase
(Bini, 2012)	Preferred saddle height High (-10° of preferred KFA) Low (+10° of preferred KFA)	Effects of saddle height on patellofemoral and tibiofemoral forces	24 competitive cyclists or triathletes	Sub-finding: Video analysis of the KFA in BDC position showed a greater KFA than static measures, however no values given

(Peveler, Shew, et al., 2012)	Saddle set to Holmes method at 25° KFA	Change from static to dynamic of knee and ankle, through a graded exercise: Stationary Level 1 (low intensity) RER of 1 Maximal exertion	34 recreational to highly-trained cyclists (28 males, 6 females)	Stationary KFA was lower than dynamic KFA in relation to Level 1 exertion, RER of 1 and at maximal exertion Stationary ankle angle was significantly lower in relation to level 1, RER of 1 and at maximal exertion
(Ferrer-Roca et al., 2012, postnote)	Static inseam saddle height on own bicycles, compared to dynamic KFA measured with 2D video	Compare static to 2D dynamic KFA Ride at 90-100rpm	23 high-level male cyclists	Inseam length to determine saddle height statically, did not coincide with the recommended 30-40°
(Fonda, Sarabon, and Li, 2014)	3 trials at different saddle heights: 25, 30 and 35° KFA using goniometer 2 trials at preferred saddle height	Compare 3 dynamic methods for knee angle measurement to each other and against static Goniometer, 2D, 3D and electrogoniometer taped to leg Intra-session reliability	6 elite 5 recreational cyclists	Electrogoniometer underestimated KFA compared to 2D and 3D Static underestimated the KFA compared to 2D and 3D, and overestimated compared to electrogoniometer All 3 dynamic achieved high intra-session reliability 3D best with a KFA correction factor of 2.2° Goniometer should be discontinued

KFA = knee flexion angle. ITBFS = Iliotibial band friction syndrome. BDC = bottom dead centre. RER = respiratory exchange ratio

Kinematics and muscle activity during cycling

It has been established that there are differences between static and dynamic measures of joint kinematics in cycling (**Table 1.3**). The general guidelines used for bicycle configuration focus on saddle height and the related KFA, however many do not take into account the goals and needs of the cyclists, nor the exercise intensity and workload.

Full body kinematics

Bicycle configuration guidelines should consider the body position adopted during events (Bini, Hume, and Croft, 2014). Bini et al. (2014) investigated the different body positions of cyclists and triathletes on the bicycle. A large difference in bicycle configuration and body position between competitive cyclists and competitive triathletes was demonstrated, specifically with regards to the frontal area and trunk and pelvic angles. Competitive triathletes demonstrated significantly lower frontal areas compared to competitive cyclists, with greater anterior knee projections and greater trunk and pelvic angles. This aggressive position adopted by triathletes is in order to reduce their frontal drag area (Burke, 2003). Only a moderate difference was revealed between competitive and recreational cyclists' trunk angles (Bini, Hume, and Croft, 2014). A limitation of this study was the use of static poses, which may differ during dynamic cycling analysis, and it was suggested that further research should compare joint kinematics during a dynamic assessment (Bini, Hume, and Croft, 2014). Similarly, dynamic bike fitting should be conducted at the intensity that a cyclist will perform the majority of his training or racing in (Peveler, Shew, et al., 2012).

Peveler et al. (2012), compared the alterations to ankle and knee angles when transitioning from a static position to active pedalling. This demonstrated differences in both ankle and knee angles, as well as demonstrating a difference in joint kinematics with increasing cycling intensities. The ankle plantarflexion and KFA were significantly lower at higher intensities. An increase in ankle dorsiflexion in response to increased intensity occurred in 71% of the subjects, and the authors proposed that this high percentage was due to the KFA being set according to the Holmes method, not at the cyclist's own freely chosen configuration.

Those results were similar to another study which did configure the test subjects' bicycles to match their own preferred setup (Kautz et al., 1991). Their results also indicated an increase in ankle dorsiflexion with an increase in workload. It was proposed that this was caused by the greater forces being applied on the pedal at higher intensities.

The ankle joint functions to transfer force from the legs to the crank (Bini and Diefenthaeler, 2010). The changes in ankle mean angles into dorsiflexion with increased intensity may be attributable to cyclists adapting their pedalling technique to overcome fatigue and the higher workload, as well as trying to maintain a controlled pedalling cadence (Bini, Diefenthaeler, and Mota, 2010; Bini, Senger, et al., 2012; Peveler, Shew, et al., 2012).

An increase into knee and hip extension was demonstrated at maximal workloads (Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010) and could be linked to a shift in forward position on the bicycle (Bini, Senger, et al., 2012). Cyclists, using their own bicycles at their preferred bicycle configuration, may intuitively move forward on the saddle in order to enhance the contribution of knee joint extensor muscles to deliver power on the pedal (Dingwell et al., 2008). However, when the cyclist's saddle height was set, this could have resulted in the cyclists changing their ankle and knee angle during the test in order to ride at a KFA they were accustomed to (Peveler, Shew, et al., 2012).

The studies reviewed above matched the bicycle configuration to the cyclist's own bicycle, however the hip angle was measured as an angle along the length of the femur parallel to the floor. This is not a clinical relevant hip angle, and spinal and upper body position have not been assessed for changes at different intensities.

The only study to date that has assessed the relationship between workload intensity and 3D kinematics demonstrated no workload effects on any of the variables (Bini, Dagnese, et al., 2016). However only hip adduction, thigh rotation, shank rotation, pelvis inclination, and spine inclination and rotation were analysed. The main findings were a small to moderate difference in lateral spine inclination and spine rotation between recreational and competitive cyclists. The subjects were instructed to ride for only 30 s at the required intensity, with only the last 15 s used for analysis. This measurement duration and overall dynamic duration may be inadequate to allow the subjects to attain the desired intensity and adjusted posture for the required cycling intensity. Likewise, the research discussed has only investigated the effects of exhaustion or maximal effort on lower limb joint kinematics.

TABLE 1.4: Workload effects on lower limb joint kinematics.

Study	Saddle height and hip measurements	Aims of study	Testing methods	Participants	Main results and notes
(Bini and Diefenthaeler, 2010)	Matched to own bicycle Vicon Hip angle measured relative to horizontal and not true hip angle	Compare joint kinetics and kinematics during an incremental cycling test to exhaustion	60, 75, 90 and 100% of PO_{max}	11 competitive male cyclists	Ankle DF increased at 100% PO_{max} Hip extension increased at 100% PO_{max}
(Bini, Diefenthaeler, and Mota, 2010)	Bicycle configuration not specified Vicon Hip angle measured relative to horizontal and not true hip angle	Analyse the joint forces and kinematics during cycling to exhaustion	Workload set at PO_{max} until exhaustion	10 well-trained male cyclists	Ankle DF increased with fatigue Hip and knee more extended with fatigue
(Sayers et al., 2012)	Own bicycles attached to a flywheel	To determine whether changes in 3D lower limb kinematics occur during the drive phase in sustained TT cycling	Six 10min work periods: 8min at 88% OBLA, 90 s effort phase at 140% of OBLA, followed by 30 s rest at 60% OBLA	10 experienced male cyclists	Increase into hip extension Increase into ankle DF

(Bini, Senger, et al., 2012)	Matched to cyclists bicycle Non-athletes set saddle height according to trochanteric height to floor Hip angle measured relative to horizontal and not true hip angle One high speed camera perpendicular to motion plane	a) Compare cyclists and non-cyclists lower limb kinematics b) Assess the effects of different workloads on joint kinematics in cyclists and non-cyclists	VT1 -25 W VT1 +25 W VT2 -25 W VT2 +25 W PO_{max}	15 athletic males 14 non-athletic males	Ankle DF increased at maximal workload Increased hip extension at PO_{max} Greater forward body position was observed at PO_{max} compared to submaximal stages
(Bini, Dagnese, et al., 2016)	Configured to elicit $\sim$30° of KFA with crank aligned with seat tube angle using a goniometer Vicon	Compare 3D joint and segment kinematics between competitive and recreational cyclists across different workloads	30 s at workloads of 65, 75, 85 and 95% PO_{peak}	12 competitive male cyclists 12 recreational male cyclists	No workload effects were observed in any of the assessed variables Recreational cyclists presented larger ranges of motion for lateral spine inclination and spinal rotation compared to competitive cyclists

DF = dorsiflexion. TT = time trial. OBLA = onset of blood lactate accumulation. VT = ventilatory threshold. KFA = knee flexion angle

Muscle activity

Cyclists will train and compete at different intensity zones, yet it is still unclear as to how the upper and lower body joint kinematics and lower limb muscle activity are affected by cycling at different intensities. It is clinically important that the position of the cyclist in their training and racing session should be further investigated as it is not known how the biomechanics of the cyclist adjust during moderately high workloads.

The typical muscle activation pattern displayed during cycling has been studied in more depth due to the recent advances in technology (Hug and Dorel, 2009). For simplification, the pedal revolution can be divided into four quadrants, starting with the crank at the top dead centre position (0°/TDC). Quadrant 1 corresponds to 0-90°, quadrant 2 corresponds to 90-180°, quadrant 3 corresponds to 180-270° and quadrant 4 corresponds to 270-360°. Quadrant 1 and 2 are variably described as the active, push phase or knee extension phase, with the foot pushing down on the pedal. Quadrant 3 and 4 are described as the passive, pull phase or knee flexion phase, where the pedal returns to the TDC position. Gluteus Maximus (GMax) is a powerful hip extender and is active during the push phase; from TDC to approximately 130° of the crank rotation cycle. Vastus Lateralis Oblique (VLO) and Vastus Medialis Oblique (VMO) extend the knee and are activated from just before the TDC position and terminate activation at just beyond 90°. The Rectus Femoris (RF) is a bi-articular muscle and works to flex the hip, as well as extend the knee. It is active from approximately 270° to 90°. The Tibialis Anterior (TA) functions to dorsiflex the ankle, activating from 270° to lift the foot over the TDC and terminates shortly afterward. The gastrocnemius muscles are also bi-articular muscles and plantarflex the foot and flex the knee. They start just after the termination of TA at ± 30° and work until 270° in plantarflexing the ankle and therefore forcing the foot down on the pedal, until the beginning of knee flexion. The hamstring muscles have shown greater variability in study results, with some studies demonstrating activation from just after TDC through to the BDC position of the crank, or variably until 270° (Jorge and Hull, 1986; Dorel, Couturier, and Hug, 2008). The hamstrings are bi-articular muscles, which are active in the transfer

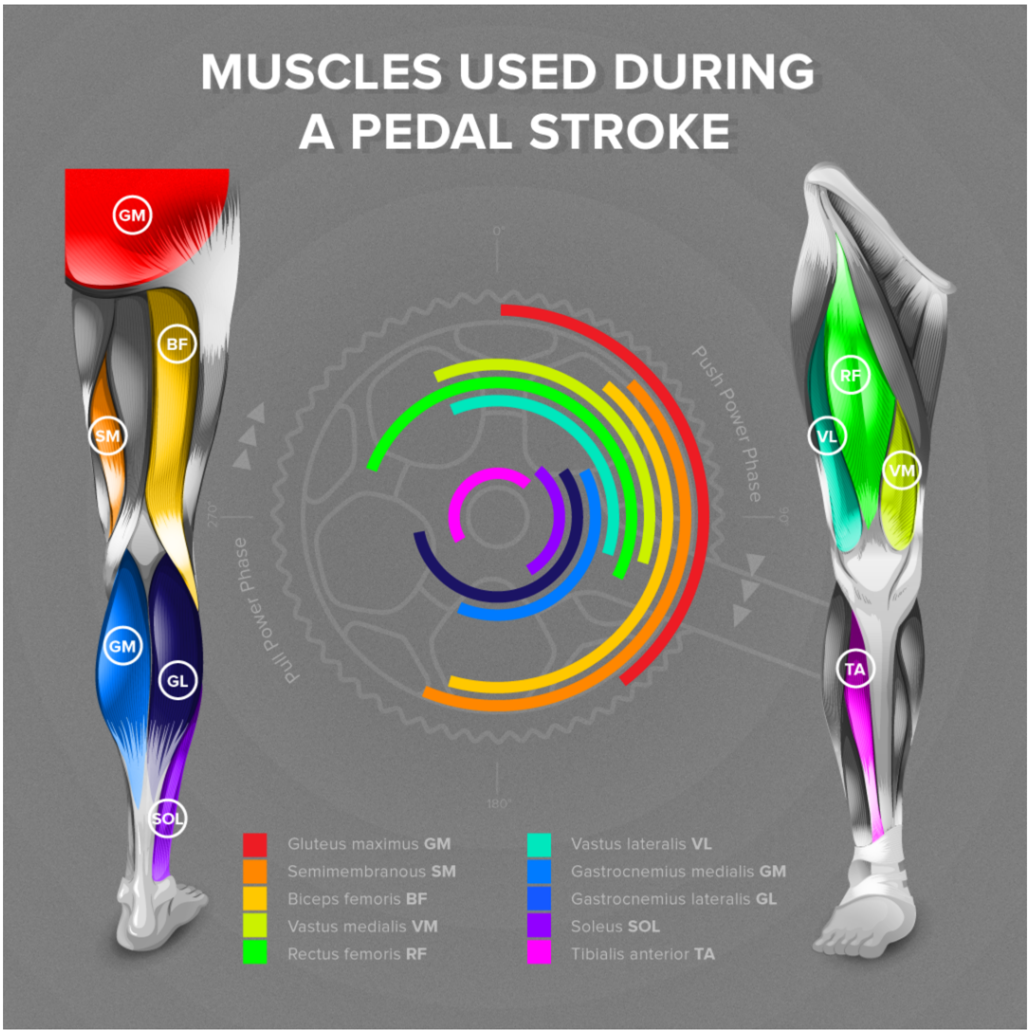

FIGURE 1.1: Muscles used during the pedal stroke. Photograph source unknown.

of energy at specific times in the pedalling cycle, and in the control of force direction on the pedal, whereas the uni-articular muscles have been linked to being the primary power producers (Hug and Dorel, 2009).

Uni-articular muscles are classified as muscles that have their origin and insertion only crossing one joint, for example, the TA muscle which originates from the tibial condyle and crosses the ankle joint to insert into the medial cuneiform and first metatarsal. Bi-articular muscles will thus cross two joints between origin and insertion, such as the RF which originates from the anterior superior iliac spine (ASIS), crosses the hip and knee joint to insert into the patella via the quadriceps tendon. Bi-articular muscles are complex to understand, however, are thought to transfer force between the joints and control the direction of the movement (Von Tscharner, 2000). Prilutsky and Gregor (2000), suggested that fatigue and Rating of Perceived Exertion (RPE) may be reduced by preferential activation of bi-articular muscles during certain phases of the cycling movement, as well as co-activation of uni-articular muscles with their bi-articular antagonists. More recently the muscle coordination during an all-out sprint cycling task was investigated (Dorel, Guilhem, et al., 2012). Fifteen well-trained cyclists performed two submaximal exercises, followed by an all-out seated sprint at 80% of their optimal pedalling rate. The relative contribution of all of the lower limb muscles tested displayed a significant change between the submaximal and maximal cycling exercises. The increase in the duration of all muscle activity during the sprint is suggestive of a strategy to enhance the work generated by each of the muscle groups. During the all-out sprint, there was a large increase in hip flexor activity, a lesser extent to the knee flexor activity, whereas the plantarflexors and knee extensors displayed an even smaller increase. The large increase in activity of the RF muscle during an all-out sprint is possibly explained by its bi-articular function, and the authors suggested that it functions largely as a hip flexor during the sprint (Dorel, Guilhem, et al., 2012). During a 60 minute self-paced cycling time trial, participants were asked to perform a one minute all-out sprint every ten minutes (Kay, Marino, et al., 2001). There was a decrease in RF electromyography (EMG) activity during the cycling sprints, which may be indicative of an alteration in the coordination pattern of the cycling movement with the development of fatigue and it is possible that alternative muscles are recruited as fatigue accumulates in working muscles.

All of the above-mentioned studies were investigated at maximal power or to exhaustion and it is difficult to distinguish the change in EMG activity levels with regards to the effects of power output increases or the onset of muscle fatigue. Racing at a workload of 55-60% VO_{2max} has been suggested as a strategic way to maximise power output while minimising the risk of early fatigue (Blake, Champoux, and Wakeling, 2012).

Fatigue should be viewed as an ongoing process that changes the neuromuscular functional state (Kay and Marino, 2000). It has been demonstrated that a five hour cycling exercise progressively reduces the maximal voluntary force-generating capability in the quadriceps muscle (Lepers et al., 2002). The contractile properties of the VLO and VMO were significantly altered after the first hour, whereas central drive and excitability were affected towards the end of the trial.

From the research reviewed it is clear that there is a change in the coordinative pattern of muscles which occurs with the onset of fatigue induced by sprinting and prolonged cycling. Change in the EMG mean frequency signal preceded change in movement kinematics (Dingwell et al., 2008). Transient fatigue was demonstrated in every muscle in each subject during this trial, and this EMG change preceded changes in kinematics. Early into the fatigue protocol, most subjects shifted towards a greater trunk lean angle and all subjects displayed an increase into dorsiflexion of the ankle angle. In conclusion, they established that as fatigue occurs, cyclists may change their muscle activation patterns to maintain performance. This subsequently may lead to maladaptive joint loading caused by changes in kinematics with fatigue.

TABLE 1.5: Workload effects on lower limb muscle activity.

Study	Muscles used for EMG analysis	Aims of study	Testing methods	Participants	Main results and notes
(Ericson, Nisell, Arborelius, et al., 1985)	GMax, VLO, RF, VMO, BF, ST, MG	EMG activity during different workloads	Power output increased from 120 W to 240 W	11 healthy subjects	An increase in workload significantly increased the mean maximum activity in all the muscles investigated
(Hautier et al., 2000)	GMax, RF, VLO, LG, BF	EMG changes during fatigue produced by repeated maximal sprints	15 repeated 5 s sprints, with a 25 s rest period between each sprint	8 male subjects 2 female subjects All trained for 9 weeks and detrained for 7 weeks	GMax and VLO remained unchanged after maximal cycling sprints, however the force and power required were reduced After fatigue, BF and LG were less activated There is an adaptation of the muscular coordination pattern to transfer force and power to the pedal

(Laplaud, Hug, and Grélot, 2006)	VLO, RF, VMO, SM, BF, LG, MG, TA	Investigate the reproducibility of 8 lower limb muscle activity levels during a pedalling exercise performed to exhaustion	2 x incremental tests until exhaustion	8 healthy male subjects	Good reproducibility of activity level of muscles during a progressive pedalling exercise performed until exhaustion Subnote: All muscles demonstrated an increase in RMS values throughout the incremental exercise
(Dingwell et al., 2008)	VLO, BF, LG, TA	Changes in movement kinematics and muscle activity as fatigue progresses	Cycled at 100% VO_{2max} until exhaustion	7 highly-trained male cyclists	Significant muscle fatigue in BF and LG across all subjects Muscle fatigue preceded changes in trunk angle

(Blake, Champoux, and Wakeling, 2012)	TA, MG, LG, Sol, VMO, RF, VLO, ST, BF, GMax	To determine muscle timing and coordination, pedal force application and total muscle activity that maximises cycling efficiency	3min intervals at 25, 40, 55, 60, 75 and 90% VO_{2max}	9 experienced competitive male cyclists	Muscle coordination patterns vary with workload GMax increased activity from low to high resistance RF and TA demonstrated increased intensity across the TDC with increasing resistance In order to maximise the muscle coordination patterns used in competition, it was suggested that cyclists train in similar conditions

Gluteus Maximus (GMax), Vastus Medialis Oblique (VMO), Vastus Lateralis Oblique (VLO), Tibialis Anterior (TA), Rectus Femoris (RF), Biceps Femoris (BF), Semitendinosus (ST), Semimembranosus (SM), Medial Gastrocnemius (MG), Lateral Gastrocnemius (LG) and Soleus (Sol) muscles. RMS = root mean square. TDC = top dead centre

Saddle Pressure Mapping

With advances in technology, we are now able to measure the pressure at the interface between the cyclist and the saddle. With saddle-related discomfort reported at a prevalence of 50-91% (Leibovitch and Mor, 2005), and despite large commercial companies conducting trials to try and improve saddle comfort and reduce saddle-related pathologies, a better understanding of saddle pressure and optimisation of saddle positioning is required. Saddle pressure measurements may also provide valuable information about the orientation of the pelvis, stability and movement patterns.

The reliability and validity of bicycle seat interface pressure measurements has been investigated (Bressel and Cronin, 2005). It was concluded that the within-trial reliability was excellent for both mean and peak pressure values. The between-trial saddle pressures demonstrated fair to excellent reliability for all areas of the saddle, except anterior saddle pressure, which displayed poor reliability. They stated that errors may arise from a lack of conformity between the saddle contours and the pressure mat.

Differences induced by varying workloads was investigated on a standard saddle (Potter et al., 2008). Their results are similar to the results by Bressel and Cronin (2005) where there was a 39% greater pressure on the saddle at 118 W compared to 300 W, and the more recent study investigating the effects of workload on seat pressure between two different saddles (Carpes et al., 2009). The saddle design had little effect on the seat pressure, however there was a statistically significant change for the different workloads. The authors all suggested that the reduction in force and pressure were as a result of the increased force applied on the pedals with the increase in power, the opposing force which would act through the hip and knee to lift the pelvis off the saddle. A limitation of the most recent study (Carpes et al., 2009) was the saddle pressure system, as it did not give specific pressures for the anterior, posterior, left or right zones of the saddle, and they recommended that further research be done to investigate a range of workloads and compare pressure load and distribution in the various saddle zones, as this may provide a better understanding to genital discomfort and optimisation of saddle positioning.

SUMMARY

Numerous studies have investigated the optimal static saddle height, and general consensus agrees that a KFA of 25-35° is optimal for performance and injury prevention (Peveler, Bishop, et al., 2005; Bini, Hume, and Croft, 2011). With the increased use of technology used in bike fitting, and a trend towards more dynamic fits, the science behind the bike fitting methods needs to be updated on a regular basis. Firstly, the reliability and validity of each measuring system should be investigated.

Static bike fitting is advantageous as it is a simple, cheaper method and more repeatable (Visentini and Clarsen, 2016), however a dynamic method is a more accurate representation of the cyclist's position, especially when load is added (Peveler, Shew, et al., 2012). Dynamic kinematic measurements may differ depending on the cyclist's relative power output and these should therefore be assessed at a specific percentage of maximal heart rate or power output. These same principles apply to choosing the correct saddle for a cyclist. Previous research has demonstrated gender-related differences in both the saddle load (Potter et al., 2008) and joint angles on the bike (De Vey Mestdagh, 1998; Sauer et al., 2007) and therefore gender should be taken into account. The training conditions and pedalling style of cyclists should also be individually based and taken into account. Likewise, individual anthropometrics and flexibility should be assessed to determine optimal bicycle configuration for each cyclist, and configuration and power correlations should be investigated further.

RESEARCH QUESTIONS

Question 1

Methods for optimising bicycle configuration have advanced in recent years. Previous research is based on static methods, yet new technology allows us to measure full body kinematics dynamically in either two or three dimensions. These measuring tools have yet to be compared to determine reliability.

1) Are the measurement tools used in bike fitting reliable?

We aim to compare three methods of measuring the ankle, knee, hip, shoulder and elbow joint angles of cyclists. A comparison of the goniometer, digital inclinometer and 3D motion capture will be conducted to assess the difference between static and dynamic measures. A secondary aim of the study is to assess the with-in subject reliability of these three differing measurement techniques for each of the respective joints.

Question 2

Bike fitting ranges for optimal positioning should take into account individual anthropometrics, flexibility, cycling style, comfort and performance goals. Previous research has based optimal recommendations on complicated formulae or personal experiences and empirical knowledge.

2) Is there a relationship between freely chosen bicycle configuration and individual anthropometrics, flexibility and training?

We aim to determine the static flexion angle ranges chosen by cyclists of the ankle, knee, hip, shoulder and elbow. Basic anthropometrics, flexibility, and training history and volume factors will be analysed to determine a correlation with freely chosen bicycle configuration.

Question 3

The smallest of performance margins can determine success or failure in sport. Recommendations for bicycle configuration have previously been determined based on personal experiences, comfort and injury prevention. However there have been no studies to date that have determined the relationship between cycling performance and the various components of the bicycle configuration such as saddle setback, handlebar height and handlebar reach. Likewise, the relationship between flexibility and power production in cycling has not been investigated.

3) Is there a relationship between freely chosen bicycle configuration, flexibility and performance in cycling?

The aim of this study is to determine if a relationship exists between individual bicycle configuration, and flexibility, and cycling performance. Guidelines for clinical application will also be determined.

Question 4

Body joint kinematics and muscle recruitment patterns may be affected by training intensity and duration. It is known that knee and ankle joint ranges, as well as lower limb muscles, are altered with fatigue and maximal effort. However, it is important to study these variables during a submaximal steady state cycle, before the onset of fatigue, as cyclists will train long steady endurance sessions.

4) How do the full body kinematics and EMG muscle magnitudes alter during a steady state submaximal cycle?

We aim to assess the change in full body kinematics and EMG muscle magnitudes of seven lower limb muscles during an hour-long steady state cycle at 60% of maximal VO_2 power.

Question 5

A method for training based on different heart rate intensities has previously been described. During a race, a cyclist will spend the majority of time at 60-80% of heart rate intensity, with only a fraction of the total time at 90% or higher. Previous research has investigated how the body position and muscle recruitment patterns are altered with absolute fatigue or maximal power, however it is important to assess the changes at differing intensities in order to train in similar conditions.

5) How do the full body kinematics and EMG muscle magnitudes alter during differing cycling intensities?

We aim to assess the change in full body kinematics and EMG magnitudes of seven lower limb muscles during three different intensities, namely 60, 80 and 90% of maximal heart rate.

Question 6

With new technology we are now able to assess the pressure at the interface between the saddle and the cyclist. In order to optimise the saddle choice and the cyclist's position on the saddle, the pressure at the saddle interface should be investigated at the various intensities a cyclist will encounter during training and racing.

6) How do the saddle pressure mapping variables alter during differing cycling intensities?

We therefore aim to assess the change in saddle pressure indexes during three maximal heart rate intensities, 60, 80 and 90%, and provide guidelines on how best to test for optimal saddle choice and positioning.

These six studies were designed to answer different research questions outlined above. These studies are summarised in **Table 1.6**.
The most commonly used abbreviations are listed in **Table 1.7**.
To meet the stylistic requirements of a thesis, the format of the published papers have been adjusted accordingly and abbreviations of units and terms standardised throughout. Chapter 2, 6 and 7 have been peer reviewed and published. Chapter 5 is in journal review. In Chapter 8, the outcomes of the six different studies are summarised, synthesised and interpreted as a whole. Recommendations are made for future research. Additionally, a clinical guideline is presented.

TABLE 1.6: Descriptive overview of the six initial studies presented in this thesis.

Study	Title and research aim
Study 1 (Chapter 2)	**Static versus dynamic kinematics in cyclists: A comparison of goniometer, inclinometer and 3D motion capture** Research aim: The aim of the study was to assess the difference between static and dynamic measures of the ankle, knee, hip, shoulder and elbow.
Study 2 (Chapter 3)	**Freely chosen bicycle configuration relative to individual anthropometrics and flexibility** Research aim: The aim of this study was to assess the individual cyclist's anthropometry, flexibility, and training history and volume relative to their own freely chosen bicycle configuration.
Study 3 (Chapter 4)	**Performance variables relative to freely chosen bicycle configuration and flexibility** Research aim: The aim of this study was to determine if a relationship between power production, bicycle configuration, and flexibility exists, and how best to apply this clinically to the cyclist and the bike fitting process.
Study 4 (Chapter 5)	**Cycling: Joint kinematics and muscle activity in steady state cycling** Research aim: The aim of this study was to assess EMG magnitudes of seven lower limb muscles as well as 3D kinematics of the full body during a steady state cycle at 60% of VO_{2max} power.
Study 5 (Chapter 6)	**Cycling: Joint kinematics and muscle activity during differing intensities** Research aim: The aim of this study was to assess the change in lower limb EMG magnitudes and 3D kinematics of the full body during three different heart rate intensities, namely 60, 80 and 90% of maximum heart rate.

Study 6 (Chapter 7)	**The effects of relative cycling intensity on saddle pressure indexes** Research aims: The aim of this study was to assess the change in saddle pressure indexes during three different intensities, namely 60, 80 and 90% of maximum heart rate.

TABLE 1.7: List of the most commonly used abbreviations.

Abbreviation	Description
3DMC	3D motion capture
°C	Degrees Celsius
ANOVA	Analysis of variance
ASIS	Anterior Superior Iliac Spine
BDC	Bottom dead centre
BF	Biceps Femoris
bpm	beats per minute
C7	Cervical vertebra 7
CoP	Centre of Pressure
DF	Dorsiflexion
EMG	Electromyography
GMax	Gluteus Maximus
GM	Goniometer
HR	Heart rate
Hz	Hertz
ICC	Intra-class correlation coefficient
IM	Inclinometer
KFA	Knee flexion angle
kg	Kilogram
km	Kilometre
LG	Lateral Gastrocnemius
MG	Medial Gastrocnemius
MHR	Maximum heart rate
ml	Millilitre
min	Minutes
PF	Plantarflexion
PO	Power output

PPO	Peak power output
PSIS	Posterior Superior Iliac Spine
RF	Rectus Femoris
RMS	Root mean square
RPE	Rate of perceived exertion
rpm	Revolutions per minute
s	Seconds
SD	Standard Deviation
SENIAM	Surface EMG for Non-invasive Assessment of Muscles
Sol	Soleus
T5	Thoracic vertebra 5
T10	Thoracic vertebra 10
TA	Tibialis Anterior
TDC	Top dead centre
TEM	Typical Error of Measurement
VLO	Vastus Lateralis Oblique
VMO	Vastus Medialis Oblique
VO_2	Volume Oxygen
W	Watts

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Chapter 2

Static versus dynamic kinematics in cyclists: A comparison of goniometer, inclinometer and 3D motion capture

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ABSTRACT

Kinematic measurements conducted during bike set-ups utilise either static or dynamic measures. There is currently limited data on reliability of static and dynamic measures nor consensus on which is the optimal method. The aim of the study was to assess the difference between static and dynamic measures of the ankle, knee, hip, shoulder and elbow. Nineteen subjects performed three separate trials of a 10min duration at a fixed workload (70% of peak power output). Static measures were taken with a standard goniometer (GM), an inclinometer (IM) and dynamic three dimensional motion capture (3DMC) using an eight camera motion capture system. Static and dynamic joint angles were compared over the three trials to assess repeatability of the measurements and differences between static and dynamic values. There was a positive correlation between GM and IM measures for all joints. Only the knee, shoulder and elbow were positively correlated between GM and 3DMC, and IM and 3DMC. Although all three instruments were reliable, 3D motion analysis utilised different landmarks for most joints and produced different means. Changes in knee flexion angle from static to dynamic are attributable to changes in the positioning of the foot. Controlling for this factor, the differences are negated. It was demonstrated that 3DMC is not interchangeable with GM and IM, and it is recommended that 3DMC develop independent reference values for bicycle configuration.

Keywords: *Bicycle; bike fitting; static; dynamic; 3D analysis; kinematics*

Highlights

- The TEM for all of the five joints as measured with all three instruments was low, and consistent with previous research.
- Although all three instruments were reliable, 3DMC analysis utilised different landmarks and therefore produced different outcomes.
- The alterations in knee flexion angle previously reported when changing from static to dynamic measurements are attributable to the positioning of the foot during the static measurements. When this is controlled for the measurement differences between the methods was negated.
- This study shows that 3D motion capture is not interchangeable with goniometer and inclinometer measurements, and it is recommended that 3D motion capture develop its own independent reference values for clinic-based bicycle configuration.

INTRODUCTION

Maximising performance and comfort, and minimising injury in cycling, requires understanding both the bicycle configuration and the rider position, necessitating research into the optimal bicycle configuration (Wishv-Roth, 2009; Bini, Hume, Croft, and Kilding, 2011; Priego Quesada et al., 2017). The aspect of bicycle configuration that has been the focus of most studies to date regarding body position on the bicycle is the saddle height and related knee and ankle flexion angles (Bini, Hume, and Croft, 2011).

To date there are three main methods used in clinical practice to set the saddle height: anthropometrics (inseam length and trochanteric height), a static knee flexion angle (KFA) method and dynamic methods (during pedalling). Hamley and Thomas (1967) proposed that for optimal cycling performance, a method of optimising saddle setting would be to adjust the height to 109% of the inseam. The Holmes method (Holmes, Pruitt, and Whalen, 1994), which is indicated for injury prevention (Peveler, Pounders, and Bishop, 2007), recommends that the KFA be set between 25° and 35°. This is commonly performed statically using a goniometer (GM) and is inexpensive and easy to perform (De Vey Mestdagh, 1998). Peveler et al. (2007) recommended that using an adjusted saddle position, a 25-35° KFA was optimal for both injury prevention and increased performance. Peveler and Green (2011) also found that setting saddle height using the 25° KFA method was more economical, with lower oxygen consumption and produced a lower rating of perceived exertion, than that produced when setting the saddle height using inseam length. In a series of studies performed, it was found that the use of 109% of inseam resulted in subjects falling outside the recommended 25-35° KFA up to 74% of the time (Peveler, Bishop, et al., 2005; Peveler, Pounders, and Bishop, 2007; Peveler, 2008). Static methods, based on the anthropometric measurement of 106-109% of leg inseam length, did not agree with dynamic 2D video camera methods (Ferrer-Roca et al., 2012).

Kinematics of the knee and ankle flexion angles changed significantly from a stationary position to a pedalling action (Peveler, Shew, et al., 2012). Stationary ankle angle was significantly lower in relation to the three active levels, as was stationary KFA. It was apparent that alterations to knee and ankle angles occur during active pedalling and that these angles alter as resistance to pedalling increases and thus a dynamic KFA of 30-40° has been recommended (Farrell, Reisinger, and Tillman, 2003).

As part of his PhD thesis, Bini (Bini, Hume, Croft, and Kilding, 2011) studied the comparison of static to dynamic measures of the lower limb joint angles in cycling using photogrammetry of reflective markers for 3D analysis and the Holmes method for static measurements. He found that both ankle plantarflexion and KFA's increased by 8° from static to dynamic measures. At bottom dead centre (BDC), a static measure using the manual GM underestimated the KFA by up to 38.2% compared to both 2D and 3D kinematics (Fonda, Sarabon, and Li, 2014). Based on these findings, Bini et al. (2011), Ferrer-Roca et al (2012), Peveler et al (2012) and Fonda et al (2014) all suggested that kinematic rather than static analysis should be used to optimise bike fit.

No significant differences were found between 2D sagittal plane kinematics for the respective angles measured in 3D during cycling (Umberger and Martin, 2001). However, there were frontal plane deviations between 2D and 3D ankle angles and it was suggested that more testing would be required to significantly prove or disprove a relationship between 2D and 3D measures, as the study was underpowered in that it only had four subjects. In a more recent study, dynamic KFA's measured with an electrogoniometer and a high speed camera (2D kinematics), were significantly underestimated when compared to 3D kinematics (Fonda, Sarabon, and Li, 2014) and a correction factor of an additional 2.2° to the KFA when using a high speed camera for 2D kinematics was recommended. However, both studies recommend analysis in 3D as the preferred method to adequately describe the lower limb cycling motion.

Studies have reported KFA's using static measures. Although some researchers have recommended the use of dynamic measures (Farrell, Reisinger, and Tillman, 2003) there have been no studies thus far comparing the static and dynamic ankle, hip, shoulder and elbow flexion angles, and the relationship between changes in these angles and their influence on KFA's. It is important to study the difference between static and dynamic angles measured during cycling, as changes in the kinematic chain may have an effect on performance, economy and injury risk.

Accordingly, the aim of this study was to assess the difference between static and dynamic measures of the ankle, knee, hip, shoulder and elbow, using a standard GM, an inclinometer (IM) and 3D motion capture (3DMC). To date, only the knee and ankle measures have been compared. The second aim of this study was to assess the within-subject reliability of these three differing measurement techniques for each of the respective joints (Zlowodski and Bhandari, 2009; Hopkins, 2013).

METHODS

Participants

Nineteen male road cyclists (32 $\pm$ 9 years, 75.7 $\pm$ 7.7kg, 178.7 $\pm$ 4.7cm) conforming to Level 2 or greater (De Pauw et al., 2013) were recruited for this study. The general characteristics and performance parameters of the 19 cyclists are shown in **Table 2.1**. The third motion capture trial from three subjects could not be included due to injuries and technical difficulties and their data average from only two data sets instead of three were used.

Prior to testing, each participant was informed of the risks and stresses associated with participation in the research trial, were personally interviewed about their training history, completed a Physical Activity Readiness Questionnaire (PAR-Q) (Whaley, Brubaker, and Otto, 2007) and signed an informed consent form. The study was approved by the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Cape Town, and conformed to the principles of the World Medical Association Declaration of Helsinki (World Medical Association, 2013).

Testing procedure

The participants reported to the laboratory on four separate occasions (1 week apart, over 4 weeks). Participants used their own cycling shoes and pedals. A CycleOps 400 Indoor Pro Cycle (Power Tap: Saris Cycling Group®. Madison, WI. USA) was used for all trials.

TABLE 2.1: General characteristics of cyclists (n=19)

Variable	**Mean ± SD**
Age (years)	32 ± 9
Body mass (kg)	75.7 ± 7.7
Stature (cm)	178.7 ± 4.7
Trochanteric leg length (cm)	94.3 ± 2.8
Percentage body fat (%)	8.6 ± 2.8
Sum of seven skinfolds (mm)	58.5 ± 14.9
Cape Town Cycle Tour race time (minutes)*	199 ± 28
PPO (W)	352 ± 34.9
PPO (W/kg)	4.7 ± 0.4
VO_{2max} (ml/kg/min) Relative	54.5 ± 6.3

**Cape Town Cycle Tour. 109km road cycle race*

On the first visit to the laboratory, participant's anthropometric measurements were taken and their own bicycle was measured to determine saddle height, saddle setback, handlebar reach and handlebar drop configurations (**Appendix 2.A**). The participant was then seated on the CycleOps ergometer which was set up to match their freely chosen position and static joint angles were measured using the GM and IM as described below. The riding position was standardised with the cyclist's hands on the brake hoods in order to avoid changes on metabolic cost due to modification of the trunk angle (Heil, Derrick, and Whittlesey, 1997).

The VirtualTraining app (VirtualTraining, version 1.7.3, Czech Republic) was set according to the participants age, weight and height. Following a standardised warm-up (**Appendix 2.B**) (Lamberts et al., 2009) and a 3min rest they completed a peak power output (PPO) and peak oxygen consumption test to determine the required workload for the experimental trials. Gas analysis was monitored over 15 s intervals using an on-line breath-by-breath gas analyser and pneumotach (Oxycon, Viasis, Hoechberg, Germany). Participants started pedalling at 100 W and resistance was increased by continuous ramp protocol at a rate of 20 W every 60 s until the participant was exhausted and could not sustain a cadence of at least 60 revolutions per minute (rpm). PPO was calculated by averaging the power output for the final minute of the VO_{2peak} test. VO_{2peak} was recorded as the highest VO_2 reading recorded for 30 s during the test.

On the subsequent visits to the laboratory, the participant was placed on the CycleOps ergometer set up according to their own bicycle measurements as previously recorded. The static joint angle measurements were repeated using a GM and an IM. The researcher then attached the set of reflective markers to the participant. A static calibration of the motion capture system was performed before the rider was seated on the CycleOps ergometer.

After the warm-up and a 1min rest they commenced a 10min cycle at 70% of their PPO. Participants were instructed to remain seated during the trial and not to alter their riding position, e.g. no standing whilst pedalling or changing the handgrip position. 3DMC was recorded for 10 s at the start of every minute of the 10min cycle. Participants were not informed when the 3D kinematic data was to be recorded, so as to prevent them changing their pedalling action. The participants repeated this procedure on a third and fourth visit to the laboratory one week apart.

Instruments

Static joint flexion angle determination of the ankle, knee, hip, shoulder and elbow were taken using a GM with 25cm arm length (Whitehall, G300 model). The measurements were taken by the primary investigator and repeated until three consistent measures (max difference of 1°) were recorded. A second static measurement was performed using a digital IM (Digi-Pas® DWL-80E model). The IM was calibrated using the manufacturer-provided instructions and standardised using both a horizontal and vertical calibrated surface as reference points. The participants were asked to stop at BDC during a revolution, following which the KFA was measured with the rider in their natural foot and lower leg riding position. This differs from the previously described Holmes method (1994), where the pedal was placed in a horizontal position to take the KFA measurement. A full description of the techniques used during static angle measurement using the GM and IM are described in **Appendix 2.C**.

For 3D kinematic measurements the participants were fitted with reflective markers as per the recommended manufacturers full-body plug-in gait model (*Plug-in Gait model details* 2008) prior to the experimental trials. Plug-in gait is a biomechanical model based on the Newington-Helen Hayes gait model that calculates joint kinematics and kinetics from the XYZ marker positions and specific subject anthropometric measurements. The Vicon full body plug-in gait marker set allows for the measurement of all joint locations and angles of rotation as well as the calculation of joint moments. The standard full marker set was modified by placing the tenth thoracic vertebra (T10) marker over the fifth thoracic vertebra (T5) instead. This was done to more closely approximate the static methods used to measure shoulder flexion angle. Reflective markers were also placed on the pedal spindle and crank axis.

An eight camera motion capture system (Oxford Metric Vicon) was used to capture kinematic data and was recorded at a sampling rate of 250 Hz.

The 3DMC markers were placed by the primary investigator by measuring and recording the placement of each marker to increase the accuracy of placement in subsequent trials. Prior training was conducted, under the supervision of experienced researchers, to ensure correct positioning of markers.

Data analysis

The static measures were captured during the first, second and third testing session. Averages from the 3DMC were recorded during the second, third and fourth testing sessions. 3DMC was recorded for 10 s at the start of every minute of the 10min cycle. The first two quality data sets from the 10min recording period were used for 3D analysis of the ankle, knee, hip, shoulder and elbow joint angles. The angles were then averaged over the 10 s. Three dimensional kinematic data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12 Hz and analysis was then performed using MATLAB® (The Mathwork Inc.). Ankle and knee angles for each trial were obtained by dividing the 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 hip angle for each trial was obtained by dividing the data into individual crank cycles using the top dead centre (TDC) pedal position determined as the point at which the pedal reflective marker reached its maximal vertical position, i.e. 360°. Shoulder and elbow angles were taken as an average over the 360° cycle. Shoulder flexion angle was analysed by assessing the angle subtended by a line passing through the T5 and seventh cervical vertebra (C7) markers on the spine and a line intersecting the shoulder, upper arm and elbow markers.

Statistical methods

All data are expressed as mean and standard deviation (mean $\pm$ SD). Reliability for each variable was assessed by calculating intra-class correlation coefficient (ICC) and the 90% confidence intervals (CIs). Typical error of the measurements (TEM) were calculated with 90% CIs, using a spreadsheet downloaded from `http://www.newstats.org` (Hopkins, 2013).

The agreement between measures was evaluated using the ICC (3,1) for the values of each instrument. Scatter and Bland-Altman plots were performed using GraphPad Prism v7.0a (GraphPad Software, San Diego, CA, USA) to assess agreement visually. The Bland-Altman plot displays a scatter plot of the average IM, GM and 3DMC measurements versus their differences, in degrees. Pearson's correlation coefficient was calculated for the averages and differences of the Bland-Altman plot. A level of significance of 0.05 was assumed.

ICC values of less than 0.5 indicate poor reliability, values between 0.5 and 0.75 indicate moderate reliability, 0.75–0.9 indicate good reliability, and value greater than 0.9 indicate excellent reliability (Portney and Watkins, 2000).

RESULTS

Reliability

The descriptive data (mean ± SD), ICC and TEM for each joint and method of measurement can be found in **Table 2.2**.

The TEM was low for all three methods across all joints. IM TEM ranged from 2.6° (hip) to 4.5° (shoulder). GM TEM ranged from 2.7° (ankle) to 4.1° (shoulder). 3DMC TEM ranged from 3.1° (shoulder) to 4.8° (hip).

There was a moderate to good inter-session reliability of all three measurement tools.

TABLE 2.2: Joint mean ± standard deviation, 90% ICC and TEM

		Mean ± SD	ICC(90%)	TEM
Ankle	GM	110.1 ± 5.6	0.80 (0.66 - 0.90)	2.7 (2.3 - 3.3)
	IM	110.3 ± 6.2	0.78 (0.64 - 0.89)	3.0 (2.6 - 3.8)
	3DMC	95.7 ± 5.9	0.68 (0.48 - 0.84)	3.5 (2.9 - 4.4)
Knee	GM	35.0 ± 7.0	0.81 (0.69 - 0.91)	3.2 (2.7 - 4.0)
	IM	38.7 ± 7.5	0.85 (0.74 - 0.93)	3.1 (2.6 - 3.8)
	3DMC	35.5 ± 6.6	0.72 (0.53 - 0.86)	3.7 (3.0 - 4.6)
Hip	GM	70.3 ± 5.5	0.58 (0.38 - 0.77)	3.7 (3.1 - 4.6)
	IM	77.3 ± 5.6	0.81 (0.68 - 0.90)	2.6 (2.2 - 3.2)
	3DMC	87.5 ± 8.5	0.705 (0.48 - 0.85)	4.8 (3.9 - 6.4)
Shoulder	GM	115.0 ± 6.4	0.62 (0.42 - 0.79)	4.1 (3.5 - 5.1)
	IM	111.6 ± 8.0	0.71 (0.54 - 0.85)	4.5 (3.8 - 5.6)
	3DMC	104.7 ± 8.0	0.87 (0.77 - 0.94)	3.1 (2.6 - 3.9)
Elbow	GM	22.7 ± 7.4	0.8 (0.66 - 0.90)	3.5 (3.0 - 4.3)
	IM	22.7 ± 9.1	0.85 (0.75 - 0.93)	3.7 (3.2 - 4.6)
	3DMC	35.4 ± 7.5	0.74 (0.56 - 0.87)	4.0 (3.4 - 5.1)

Ankle range >90° indicates plantarflexion. Goniometer=GM. Inclinometer = IM. 3D motion capture = 3DMC.

Instrument correlations

There was a moderate to excellent correlation between GM and IM measurements, for all of the measured joints. This suggests that the GM and IM measurements are relatively interchangeable. The static to dynamic methods did not correlate as well, with a range from poor correlation (ankle IM/MC ICC = 0.25) to a moderate correlation (shoulder IM/MC ICC = 0.72, knee GM/MC ICC = 0.73). There were statistically significant correlations between the two static methods and 3DMC for only the knee, shoulder and elbow joints These values, together with ICC, mean bias and 95% CIs are presented in **Figure 2.1**.

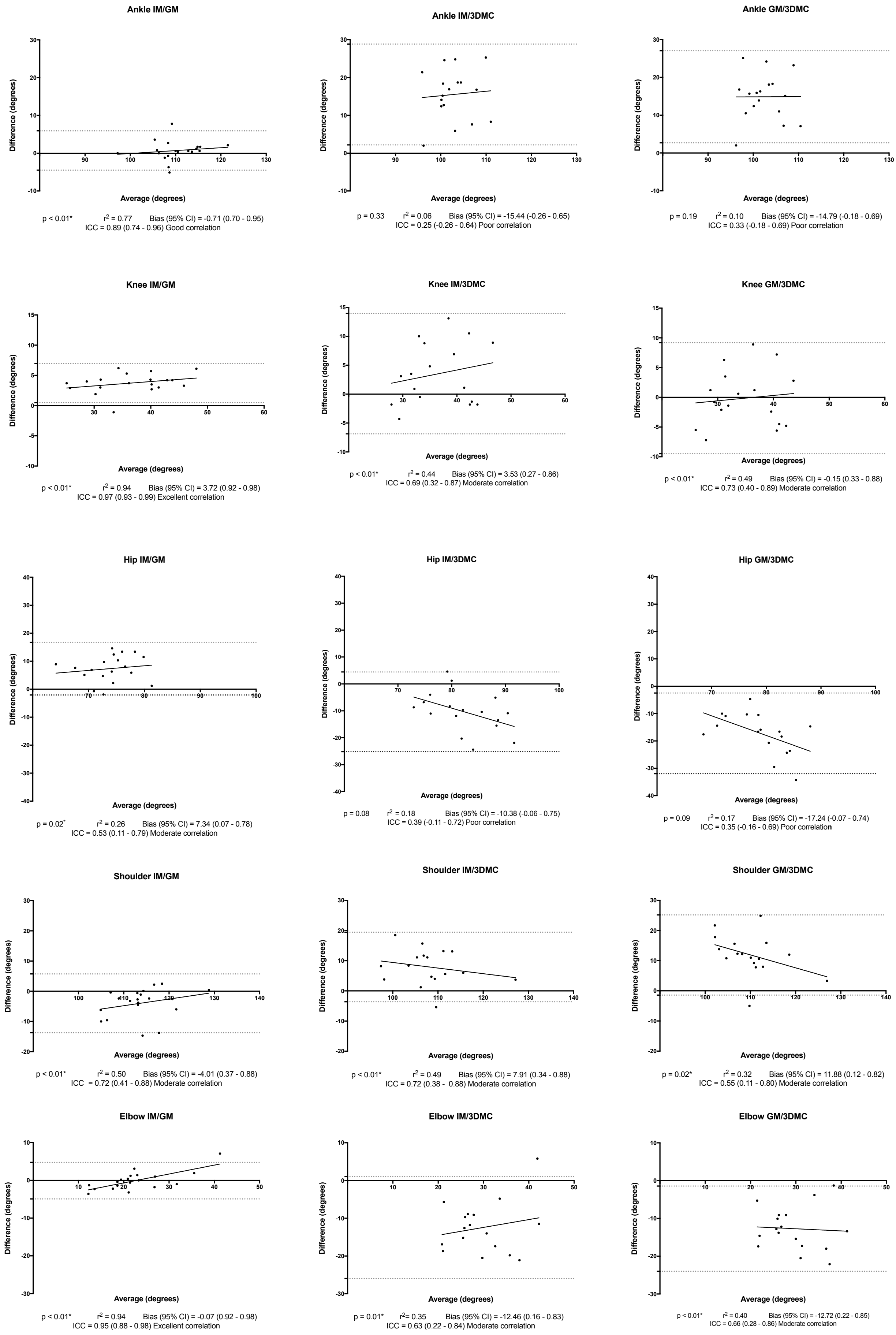

FIGURE 2.1: Scatter and Bland-Altman graphs for each joint, *indicates positive correlation. IM=Inclinometer GM=Goniometer 3DMC=3D motion capture

DISCUSSION

The purpose of this study was to assess the difference between static and dynamic measures of the ankle, knee, hip, shoulder and elbow, using a standard GM, an IM and 3DMC, and to assess the reliability of these three differing measurement techniques for each of the respective joints.

The first important finding is that the TEM for all of the five joints as measured with all three instruments was low and consistent with previous research (McGinley et al., 2009; Kolber, Vega, et al., 2011; Milanese et al., 2014). Contrary to the general perception (Braghin et al., 2016; Schurr et al., 2017), 3DMC was not more reliable than either goniometer or inclinometer measurement. Across all three methods, the range of TEM was relatively low (from 2.6° to 4.8°) and this agrees with the recommendation that 3DMC errors between 2° and 5° are to be regarded as reasonable (McGinley et al., 2009). All three methods can therefore be considered reliable.

There was a general trend towards higher inter-session reliability for the IM measurements when compared to either GM or 3DMC. There was a similar trend for higher reliability with the IM measures when compared to the GM in a study assessing reliability of shoulder mobility (Kolber and Hanney, 2012).

The static measurements for the ankle joint demonstrated a moderate to excellent degree of correlation; however, there was a poor to moderate correlation between either static method and 3D kinematic measurements for the ankle joint. This may partially be explained by the static and dynamic measurements using different anatomical landmarks. The IM and GM measurements were taken parallel to the long axis of the fibula (pointing towards the fibular head) and parallel to the long axis of the fifth metatarsal, with the lateral malleolus as the centre of measurement, whereas the 3DMC system measures the ankle angle from the marker on the posterior calcaneus to the marker placed on the dorsum at the base of the first metatarsal, intersecting the line from the marker on lateral malleolus to the lateral knee joint. This is shown in **Figure 2.2C** and **Figure 2.2D**. Shoe length and height may also contribute to the difference in means.

The knee was moderately to excellently correlated between all three instruments (ICC = 0.69 – 0.97). This is almost certainly due to the methods used whereby the ankle joint was positioned in the natural riding position when taking the static measures, as opposed to the Holmes method, where the pedal is placed horizontal to the floor when measuring the KFA (Holmes, Pruitt, and Whalen, 1994). By taking into account the natural position of the foot whilst riding, this eliminates the kinematic changes that occur when transitioning from static to dynamic measurement. It is our hypothesis that ankle plantarflexion drives the KFA, and by measuring the KFA in a static position that resembles the natural riding position, the measurements correlate well. The notion that the ankle and knee joint movement are closely related was previously highlighted by Peveler et al (2012)(page 3008):

> “The distance from the pedal axle in the 6-o’clock position to the top of the saddle for a given subject does not alter. When plantarflexion occurs at the ankle, knee angle must also alter, because of the pelvis maintaining a stable position with little movement on the saddle and the cycling shoe being locked in the clipless pedal”.

Peveler et al. (2012) also continues to explain that the alteration to knee and ankle angles from static to dynamic pedalling may be a result of the cyclist attempting to obtain an ankle and KFA similar to those in which they are accustomed. This may also explain the reason why Fonda et al. (2014) found that GM measures significantly underestimated KFA’s compared to 3D kinematics. Both these studies were done with pre-determined saddle heights, compared to the current study where participants cycled at their preferred configuration. Once again the two static measurement (IM and GM) correlations demonstrated a smaller difference compared to static and dynamic measures, which demonstrated broad 95% Limits of Agreement (knee IM/3DMC: -6.86 – 13.94; GM/3DMC: -9.49 – 9.18) as well as lower r^2 values (knee IM/3DMC: 0.44; GM/3DMC: 0.49).

There was a large degree of variability in the hip measurements. Once again only the static measurements had a moderate correlation (ICC = 0.53), and there was poor correlation between static and dynamic measures with poor to good 95% CI's. There was a trend for higher reliability with the IM measurements (TEM 2.6°) when compared to goniometry (TEM 3.7°) and 3DMC (TEM 4.8°). Although the 3DMC system was reliable, this is not higher than when using the IM. The 3DMC system has a higher degree of technical difficulty, in terms of application in a clinical setting, compared to GM and IM. Marker placement inaccuracy and inconsistency may lower the reliability and negatively affect the clinical analysis of the data (Sinclair, Hebron, and Taylor, 2014). In addition, the 3DMC measures a different hip angle to that measured with the IM and GM. The 3DMC system uses a perpendicular line bisecting the anterior superior iliac spine and posterior superior iliac spine, and a line bisecting the knee joint centre and greater trochanter to determine hip flexion angle. Both the GM and IM measure the hip joint as the angle subtended by the area expanding below the iliac crest from the third lumbar vertebra to the sacrum, to the line bisecting the greater trochanter and lateral femoral condyle, with the pedal at the TDC position. Other researchers have highlighted technical difficulties with the 3D markers used in measuring the hip angle (Neptune and Hull, 1996), and similar difficulties were experienced in this study. Pelvic rotation and lateral movement needs to be taken into account when measuring the hip angle when using 3DMC analysis.

All three measurement methods for the shoulder joint demonstrated a moderate degree of correlation. However, there was a bias for 3D measurements in comparison with static measurements with mean 3D kinematic measurements for shoulder flexion 10.3° lower than mean values obtained using the GM and 6.3° lower than the IM. Displacement of the reflective marker during protraction of the scapula may account for this difference in the means. The 3DMC shoulder marker is placed when the arm is in the anatomical position. This marker's axis subsequently moves from a coronal plane towards a sagittal plane during scapular protraction and results in anterior displacement of the marker ball in relation to the head of the humerus. As a result, the axis of the humerus measured by 3D kinematics may be distorted towards measurement of lower shoulder flexion angle when the rider places his hands on the hoods of the bicycle. This is shown in **Figure 2.2A** and **Figure 2.2B**. Similarly, a mean difference of approximately 10° was found between static and dynamic measures of scapulothoracic tilt, and this became increasing unreliable at elevation angles greater than 60° (Maclean et al., 2013).

The static measurements for the elbow demonstrated an excellent degree of correlation, and a moderate correlation between the static and dynamic measurements. Although all three measurement methods were significantly correlated, there are wide limits of agreement when either of the static measurements were compared to the dynamic measurements. There was a bias for the 3DMC system measurements in comparison with static measurements, with mean 3D kinematic measurements for elbow flexion 12.7° higher than mean values obtained using both the GM and IM. This may once again be due to the displacement of the shoulder marker as described above.

In addition to the proposed rationale for differences in the kinematic values between static and dynamic methods, it is also possible that rotational movements in the axial plane and between planes which are incorporated into the analysis by the 3DMC software may account for some of the differences between static and dynamic 3D measurements. A limitation to this study was not taking a static recording with the 3DMC system before the rider began pedalling. It has been recommended by Peveler et al. (2012) to standardise static angles for comparison with angles during 3DMC.

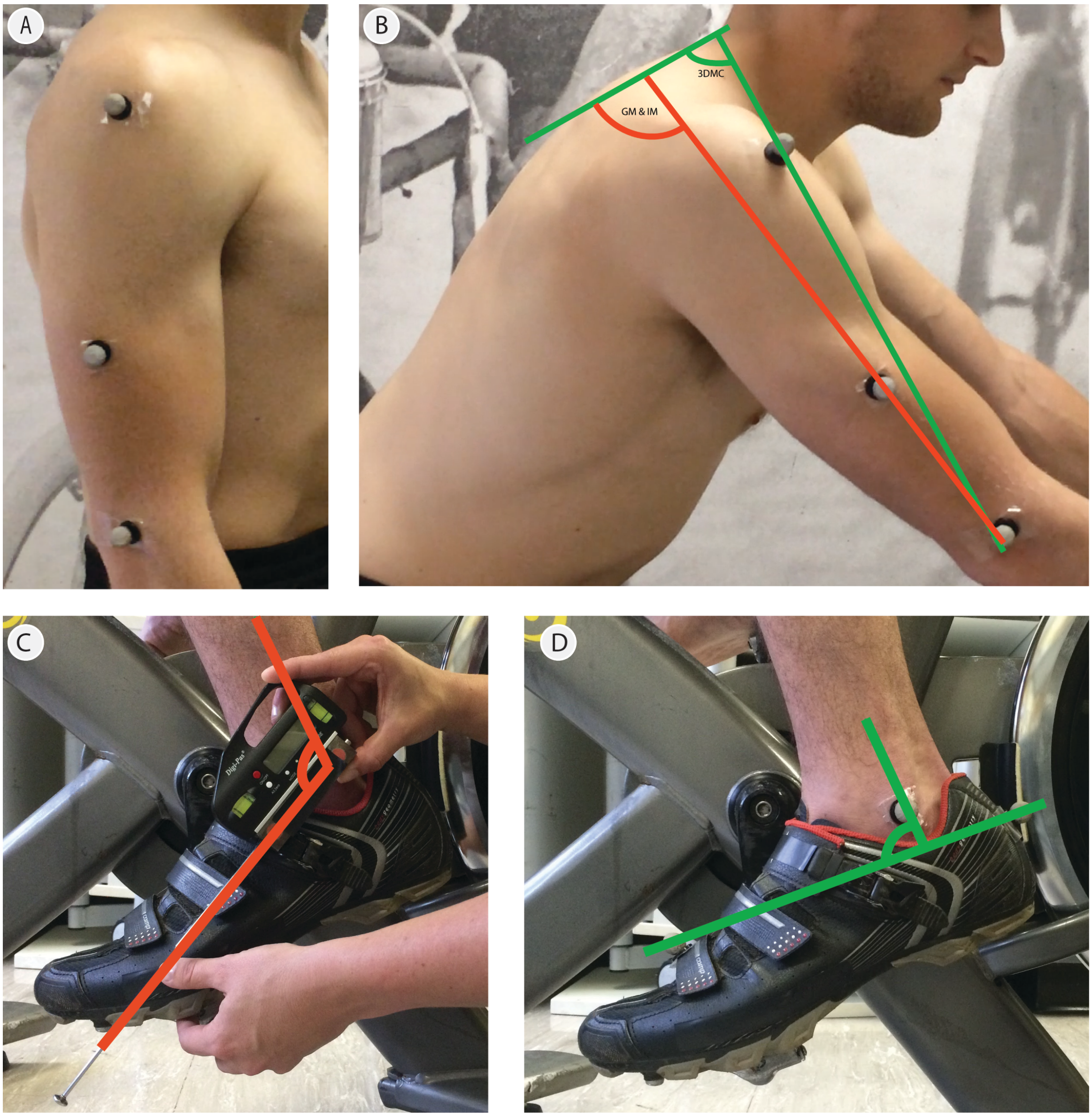

FIGURE 2.2: Displacement of shoulder marker: arm in neutral position when markers placed on body (A), hands on the hoods showing anterior displacement of marker (B). Different methods of measuring the ankle angle: IM and GM measurements (C), 3D motion capture measurement (D).

CONCLUSION

Firstly, despite the perception that 3DMC analysis is the gold standard for kinematics, when applied in a clinical setting, the TEM between sessions is higher for this method than for the two static methods. Overall the static measurements demonstrated moderate to excellent reliability with smaller Limits of Agreement, than the dynamic measurements which only demonstrated poor to good reliability. Although all three instruments were reliable, 3DMC analysis utilised different landmarks for most joints and therefore produced different outcomes. In addition, it is also possible that rotational movements in the axial plane and between planes which are incorporated into the analysis by the 3DMC software may account for some of the differences between static and dynamic 3D measurements.

Secondly, the alterations in KFA previously reported when changing from static to dynamic measurements are attributable to the positioning of the foot during the static measurements. Foot positioning should be standardised, either according to the Holmes method, with the pedal horizontal to the ground, or as was done in this study, with the cyclist stopping and maintaining their natural foot position at BDC. When this is controlled for the measurement differences between the methods was negated. Joint angle measurements using a GM and IM are both reliable and valid, are easy to use and inexpensive. GM and IM joint measurements have been shown to be relatively interchangeable. 3DMC is expensive, timely and aversive, and although reliable, great care is required to place the markers in the exact location from one trial to the next. This study shows that the 3DMC model used in this study is not interchangeable with GM and IM measurements, and it is recommended that 3DMC develop its own independent reference values for clinic-based bicycle configuration.

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Appendix

2.A Bicycle configuration measurements

Saddle height:

The saddle height was measured from the centre of the crank axle to the top of the saddle, passing through the centre of the bicycle seat tube and seat post.

Saddle setback:

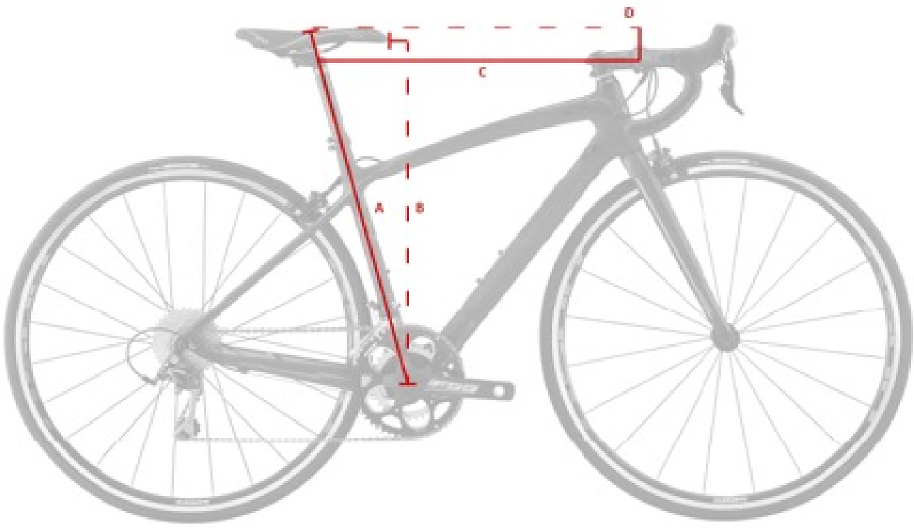
Saddle setback was measured as the horizontal distance from the front of the saddle to the centre of the crank axle.

Handlebar reach:

The handlebar reach was measured horizontally, from the centre of the handlebar clamping point to the centre of the seat post or seat tube. This measurement is for bicycle frames with a 74° seat tube angle.

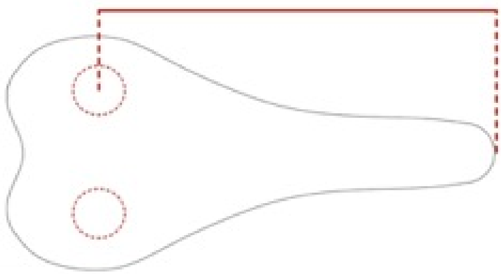
Handlebar drop:

Handlebar drop values were measured as the vertical distance from the top of the saddle surface to the centre of the handlebar clamping joint.



Saddle length:

The saddle was measured (a standard saddle is 22.5cm in length) from the centre of the ischial tuberosity padding to the front of the saddle.



2.B Lamberts Submaximal Cycle Test

This is a 17 minute warm-up protocol, simulated on a flat course with three different exercise intensities defined by different target heart rates. During each of these stages, the cyclists had to elicit and maintain their heart rate ($\pm$1 bpm) corresponding to 60%, 80% and 90% of their maximum heart rate (MHR). The cyclists needed to either cycle faster or slower to achieve their heart rate when it deviated by at least 2 bpm. The MHR of each participant was measured during the peak power output test, and this was used to calculate the target heart rates for the warm-up. The target heart rates for the initial warm-up before the peak power output test were based on a predicted MHR (220-age). The first stage of the test involved cycling for 6 minutes at 60% MHR, the second stage involved another 6 minute cycle at 80% of MHR. The third stage was a 3 minute cycle at 90% of MHR. After completing the warm-up test, participants were asked to stop cycling, sit up straight and to recover for 90 seconds.

2.C Static angle measurements

Maximum hip flexion angle was determined by requesting the participant to stop pedalling with the pedal at the top dead centre position. Measurements were taken along the long axis of the femur (as determined by the position of the greater trochanter and the lateral knee joint space). The position of the pelvis was determined by the angle of the posterior surface of the lowest lumbar vertebrae and the proximal three sacral vertebrae. This was determined as the area in the midline, directly below a line joining the iliac crests. The first inclinometer (IM) measurement was taken by positioning the bottom 'red dot' (**Figure 2.3**) of the IM over the top of the greater trochanter with the extension arm positioned at the lateral condyle of femur. The second measurement was taken with the positioning of the IM on the posterior surface of the lower lumbar vertebrae with the top 'red dot' perpendicular to the iliac crests. The maximum hip flexion angle was calculated from these two measurements by applying plane geometry formulas.

Knee flexion angle was assessed by requesting the participant to stop pedalling with the pedal at the bottom of the pedal stroke in the 6 o'clock position, without altering their natural ankle angle. The tester was subjectively assessing that the heel was not dropped as the cyclist adopted a resting position. The measurements were taken parallel to the long axis of the femur and parallel to the long axis of the fibula (as determined by the position of the greater trochanter and the lateral malleolus). The goniometer (GM) axis was positioned at the lateral epicondyle of the femur. The IM was positioned at the top of the greater trochanter with the extension arm positioned at the lateral condyle of femur for the first measurement. The second measurement was taken with the positioning of the IM on the lateral malleolus with the extension arm positioned at the head of fibula. The knee flexion angle was calculated from these two measurements by applying plane geometry formulas.

Ankle flexion angle was assessed by requesting the participant to stop pedalling with the pedal at the bottom of the pedal stroke in the 6 o'clock position. They were asked not to alter their ankle angle as they stopped. The tester was subjectively assessing that the heel was not dropped as the cyclist adopted a resting position. The measurements were taken parallel to the long axis of the fibula (pointing towards the fibular head) and parallel to the long axis of the 5th metatarsal. The GM axis was positioned at the lateral calcaneus at the bisection of fibula and 5th metatarsal. The IM was positioned at the centre of the lateral malleolus with the extension arm positioned to the head of fibula for the first measurement. The second measurement was taken with the positioning of the IM on the lateral malleolus with the extension arm positioned parallel to the 5th metatarsal. The ankle flexion angle was calculated from these two measurements by applying plane geometry formulas.

Shoulder flexion angle was assessed with the participant's hands on the brake hoods and with the participant sitting on the saddle. The participant was instructed to pedal a few revolutions before stopping and was instructed not to change their position. The measurements were taken along the longitudinal axis of the humerus pointing towards the lateral epicondyle and with the other arm of the GM parallel to the long axis of the first four thoracic vertebrae, determined as the area in the midline, directly below the vertebrae prominens. The GM axis was positioned at the lateral aspect of the centre of the humeral head approximately 1cm below the acromion process. The IM was positioned at the lateral aspect of the centre of the humeral head approximately 2.5cm below the acromion process with the extension arm positioned in line to the lateral epicondyle for the first measurement. The second measurement was taken with the positioning of the IM on the first 4 thoracic vertebrae with the 'red dot' perpendicular to the centre of the humeral head. The shoulder flexion angle was calculated from these two measurements by applying plane geometry formulas.

Elbow flexion angle was taken with the participant's hands on the brake hoods and with the participant sitting on the saddle. The participant was instructed to pedal a few revolutions before stopping in a comfortable position, and was instructed not to change their position. The measurements were taken along the longitudinal axis of the humerus and with the other arm of the GM parallel to longitudinal axis of the radius (as determined by the position of the acromion process and the styloid process of the radius). The GM axis was positioned over the lateral epicondyle of the humerus. The IM was positioned over the lateral epicondyle of the humerus with the extension arm positioned to the centre of the humeral head to determine the first measurement. The second measurement was taken with the positioning of the IM over the lateral epicondyle of the humerus with the extension arm positioned to the longitudinal axis of the radius (as determined by the position of the acromion process and the styloid process of the radius). The elbow flexion angle was calculated from these two measurements by applying plane geometry formulas.

The methods used for static goniometry measurement are valid and reliable (Norkin and White, 2003) and have been used in previous studies (Peveler, Bishop, et al., 2005; Peveler, Shew, et al., 2012). The interrater reliability of the goniometry measurement of shoulder flexion has been shown to be extremely high, regardless of the sitting or supine testing position (Sabari et al., 1998). It is recommended that the standard protocols for goniometric assessment of shoulder flexion are consistent with regard to the GM alignment, and that the participant be consistently placed in the same testing position. Only a few studies have researched the reliability of the GM and IM in measuring joint angles. To date the reliability of both methods is high in the ankle and shoulder joint angle measurements (Kolber and Hanney, 2012; Konor et al., 2012).

FIGURE 2.3: Digital inclinometer. The 'red dot' on the IM helps to standardise positioning on anatomical landmarks during static joint angles measurement.

Chapter 3

Freely chosen bicycle configuration relative to individual anthropometrics and flexibility

ABSTRACT

Introduction: Intrinsic factors such as leg length, flexibility and training history are factors that should be considered in the individual bicycle configuration process. Bike fitting methods do not always take these variables into account, and as yet there have been limited studies examining how these variables can affect the cyclist's position on the bicycle. The main aims of this study were to establish how individual anthropometrics, training history and flexibility factors may influence cyclists' freely chosen bicycle configuration, and to determine the full body static flexion angles chosen by cyclists on the bicycle.

Methods: Fifty male cyclists were recruited for the study. Individual bicycle configuration, static joint angles, anthropometrics, flexibility and training history were recorded. A Pearson correlation coefficient was computed to assess the relationship between preferred bicycle configuration and anthropometrics, flexibility and training history.

Results: Stature, leg length and arm length were moderately correlated with saddle setback. Leg length and hamstring flexibility demonstrated moderate correlations with handlebar drop. Hamstring flexibility was also correlated with stature. There was no significant relationship between hamstring flexibility and total saddle height. There were no significant relationships between training history and bicycle configuration. Average joint kinematic ranges were similar to previous recommendations, additional recommendations for the hip and shoulder angle were suggested.

Conclusion: The results from this study are recommendations for how individual variables may influence static bicycle fitting. Anthropometrics and flexibility should be taken into account for optimal bicycle configuration.

INTRODUCTION

Bike fit is defined as the detailed process of evaluating the cyclist's physical and performance requirements and systematically adjusting the bike to meet the cyclist's goals and needs (Medicine of Cycling, 2013). Bicycle configuration can have an influence on the cyclist's performance and perception of comfort (Silberman et al., 2005). While most studies to date have discussed the configuration normative values that are recommended for power and injury prevention (Peveler, Pounders, and Bishop, 2007; Bini, Hume, and Croft, 2011; Peveler and Green, 2011), it is important to address the adjustable components of the bicycle and both the cyclist's perceptive and anthropometric measures in order to optimise their comfort and cycling position. An online survey identifying factors of bicycle comfort was conducted, and of the 244 respondents, 90% of the cyclists agreed that comfort is a concern when riding a bicycle, while 46% of enthusiastic cyclists agree that comfort is reached at the expense of performance (Ayachi, Dorey, and Guastavino, 2015).

For increased performance and injury prevention, saddle height has been recommended to be set statically at a knee flexion angle (KFA) of 25-35° (Peveler, Pounders, and Bishop, 2007). To date there have been no other researched ranges of optimal angles for the other joints of the body on the bicycle. There are guidelines for ankle and elbow ranges, although these are based on personal experience more than scientific data. De Vey Mestdagh (1998) has suggested complicated formulae to determine saddle setback, handlebar reach and handlebar drop, however most bike fitting experts have suggested that the final position be based on comfort and what visually appears acceptable (**Table 3.1**). The drawback to formulae is that they do not always take into consideration individual anthropometrics or pedalling styles (Peveler, Pounders, and Bishop, 2007). The range of 25° to 35° for KFA has been used extensively and proves to be the gold standard for setting saddle height statically as it takes into consideration all individual factors of the cyclists (Peveler, Bishop, et al., 2005). Adjustments are made within the range to accommodate individual anthropometric needs and different types of pedalling such as with the heel up or down (Peveler and Green, 2011). Other joints of the body and configuration variables should have similar recommended ranges for optimal bicycle configuration.

Saddle setback has commonly and anecdotally been determined by the ‘knee over pedal spindle’ or ‘KOPS’ method. This involves dropping a plumbline from the anterior knee with the pedal in the forward or 3 o’clock position. The plumbline should fall in line with the pedal axis or just posterior to this. This method has not been proven scientifically and although suggested to prevent injury, there is no data to support this. Another formula is to take the upper leg length into account in determining optimal position (De Vey Mestdagh, 1998). This too is based on personal experience and a review of existing ergonomics, although these have not been published.

There are two popular methods for setting handlebar reach and handlebar drop, however neither have any scientific support. The first determines the final handlebar position as a measure of the arm and torso length, although this formula does not take into account the bicycle frame geometry nor the type of handlebar (Burt, 2014). The other most common method of setting handlebar reach and handlebar drop is related to torso angle, with a recommendation ranging from 30° to 60° (Silberman et al., 2005; Andy Pruitt and Matheny, 2006; Burt, 2014). The torso angle in this approach is measured as an angle from a line parallel to the floor bisecting a line from the hip joint centre to the glenohumeral joint centre. This angle negates the natural curves of the lumbar and thoracic spine. However, the authors have further suggested that handlebar height depends on training status, strength, individual comfort, and spinal and hamstring flexibility.

TABLE 3.1: Summary of guidelines for other variables of bicycle configuration.

Variable	Recommendation	Based upon	Study
Saddle height	25-35° knee flexion angle	Scientifically based	(Holmes, Pruitt, and Whalen, 1994; Peveler, Bishop, et al., 2005; Peveler, Pounders, and Bishop, 2007; Peveler, 2008; Bini, Hume, and Croft, 2011; Peveler and Green, 2011)
Saddle setback	Formula related to upper leg length	Personal perspective	(De Vey Mestdagh, 1998)
	Plumbline and knee over pedal spindle (static)	Personal experience and recommendations	(Burke, 2003; Silberman et al., 2005; Burt, 2014)
Handlebar reach	Formula determined by arm length and torso length	Personal perspective	(De Vey Mestdagh, 1998)
	Plumbline from cyclist's nose dropped to centre of stem, hands in drops	Personal experience and recommendations	(Burke, 2003)

	Comfort in the drops, elbows flexed 60° to 70° With the knees at their maximal height and forward position, the distance between the elbows and knees should be small, 1 to 2 inches (2–5 cm)	Personal experience and recommendations	(Silberman et al., 2005)
	Related to forearm length	Personal experience and recommendations	(Andy Pruitt and Matheny, 2006)
	Individual, comfort	Personal experience and recommendations	(Burt, 2014)
Handlebar height	Formula determined by arm length and torso length	Personal perspective	(De Vey Mestdagh, 1998)
	1-2 inches below saddle for small cyclists 4 inches below saddle for tall cyclists	Personal experience and recommendations	(Burke, 2003)
	Hands on the brake hoods, arms slightly flexed, the torso should flex to about 45° in relation to a non-sloping top tube	Personal experience and recommendations	(Silberman et al., 2005)
	Racer and competitive recreational cyclists' torso angle 30-45° Casual cyclist 50-60° torso angle	Personal experience and recommendations	(Andy Pruitt and Matheny, 2006)
	Individual, comfort	Personal experience and recommendations	(Burt, 2014)

TABLE 3.2: Previously recommended ranges for optimal positioning.

Joint	Recommendations	Based upon	Study
Ankle	13° plantarflexion at bottom dead centre	Personal perspective	(De Vey Mestdagh, 1998)
	5-15° plantarflexion at bottom dead centre	Personal experience and recommendations	(Burt, 2014)
Knee	25-35° flexion		(Holmes, Pruitt, and Whalen, 1994; Peveler, Bishop, et al., 2005; Peveler, Pounders, and Bishop, 2007; Peveler, 2008; Bini, Hume, and Croft, 2011; Peveler and Green, 2011)
Hip	55-65° on road bike (measured as an angle along the femur to the greater trochanter to the shoulder)	Personal experience and recommendations	(Burt, 2014)
Shoulder	None to date		None to date
Elbow	20-30°	Personal experience and recommendations	(Burt, 2014)
Torso angle	45-55° recreational 45-30° fast road cyclists	Personal experience and recommendations	(Burt, 2014)
	45° to non-sloping top tube	Personal experience and recommendations	(Silberman et al., 2005)
	30-45° racing or competitive recreational 40-50° fitness cyclists 50-60° casual cyclists	Personal experience and recommendations	(Andy Pruitt and Matheny, 2006)

It has previously been reported that cyclists with reduced flexibility of the hamstrings tended to select lower saddle heights (Ferrer-Roca et al., 2012). However, Hynd, Crowle, and Stephenson (2014) determined that hamstring flexibility did not have an effect on pre-selected saddle height, and this was in agreement with another study demonstrating that a single variable could not predetermine optimal saddle height (Muyor, Alacid, and López-Miñarro, 2011). Muyor et al. (2011) concluded that hamstring flexibility does not have an influence on the thoracic and pelvic postures adopted whilst sitting on a bicycle. Ferrer-Roca et al. (2012) suggested that further studies should be conducted to determine if low-level hamstring flexibility may have an influence on the cyclist's posture and bicycle configuration. One could also hypothesise that handlebar reach and handlebar drop can be influenced by the cyclist's spinal flexibility. De Vey Mestdagh (1998) in his personal search for an optimum cycling position, determines handlebar reach and handlebar drop values by measuring arm and torso length, with recommended heights determined by averages as well as comfort levels.

Dahlquist, Leisz, and Finkelstein (2015) investigated the performance of 63 recreational road cyclists compared with established norms regarding strength and flexibility measures. Hamstring and lumbar flexibility were tested, as well as static goniometer measurements of the torso, elbow, hip and knee angles on their own bicycles. Despite 59% of the participants having had a professional bike fit, less than 50% of the participants met the recommended flexibility, strength and bike fit norms. The professional fitments conducted varied from visual inspection to computerised systems, and some cyclists were fitted for optimal performance and aerodynamics, resulting in a degree of discomfort. The study concluded that further studies should be conducted as there is a need for better definitions of normative values for intrinsic factors related to cycling. Our aims were therefore:

- To determine the static flexion angle ranges of the ankle, knee, hip, shoulder and elbow adopted by cyclists.

- Basic anthropometrics were analysed to determine the relationship on freely chosen bicycle configuration.
- Training history and training volume were analysed to determine a correlation with freely chosen bicycle configuration.
- To determine if measures of flexibility (Sit and reach test, Knee Extension Angle test, Fingertip to floor and modified Schober test) were associated with the position adopted by cyclists.

We hypothesised that stature, leg length and arm length would have a significant correlation with saddle height, saddle setback and handlebar reach. Spinal flexibility was predicted to have a positive correlation with handlebar reach and handlebar drop. Lastly, increased training history or training load was expected to correlate with handlebar drop and saddle height.

METHODS

Participants

Fifty well-trained male road cyclists (30 $\pm$ 9 years, 76.5 $\pm$ 7.9kg, 180.7 $\pm$ 5.6cm) conforming to Level 2 or greater (De Pauw et al., 2013) were recruited for this study. The general characteristics and performance parameters of the 50 cyclists are shown in **Table 3.3**. Participants were excluded if they had made any changes to their bicycle configuration in the past three months, or if they experienced any pain or discomfort on their current bicycle configuration. Prior to testing, each participant was informed of the risks and stresses associated with participation in the research trial, were personally interviewed about their training history, completed a Physical Activity Readiness Questionnaire (PAR-Q) (Whaley, Brubaker, and Otto, 2007) and signed an informed consent form. The study was approved

TABLE 3.3: General characteristics of cyclists (n=50)

Variable	**Mean $\pm$ SD**
Age (years)	30 $\pm$ 9
Body mass (kg)	76.5 $\pm$ 7.9
Stature (cm)	180.7 $\pm$ 5.6
Trochanteric leg length (cm)	97.5 $\pm$ 4.4
Arm length (cm)	80.8 $\pm$ 3.6
Percentage body fat (%)	11.9 $\pm$ 4.7
Sum of seven skinfolds (mm)	61.5 $\pm$ 20.2
PPO (W)	387.7 $\pm$ 53.1
PPO (W/kg)	5.1 $\pm$ 0.7
VO_{2max} (ml/kg/min) Relative	58.8 $\pm$ 7.7

by the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Cape Town, and conformed to the principles of the World Medical Association Declaration of Helsinki (World Medical Association, 2013).

Testing procedure

The participants reported to the laboratory with their own bicycle, cycling shoes and pedals. During the visit to the laboratory, the participant's anthropometry measurements including height, weight, arm length, trochanteric leg length and sum of seven skinfolds (triceps, biceps, supra-iliac, sub-scapular, calf, thigh and abdomen) were taken, and body fat percentage was determined (Ross and Marfell-Jones, 1991).

The participants then underwent the following flexibility testing:

1. Sit and reach test.
2. Knee Extension Angle (KEA) to assess hamstring length.
3. Fingertip to floor test
4. Modified Schober test.

See **Appendix 3.A** for detailed descriptions of testing.

The participant's own bicycle configuration was measured according to a set of objectively reproducible measurements:

- Seat height

  The seat height was measured from the centre of the crank axle to the top of the saddle, passing through a reference line set at 74 degrees to the horizontal to standardise the seat tube angle.

- Saddle setback

  Saddle setback was measured as the horizontal distance from the front of the saddle to the centre of the crank axle. The front of the saddle was determined based on a standardised distance of 22.5cm from the contact point of the ischia to the front of the saddle. For saddles which did not conform to these measurements, a correction value was applied to the measured setback.

- Handlebar reach

  The handlebar reach was measured horizontally, from the centre of the handlebar clamping point to the centre of the 74° seat tube reference line.

- Handlebar drop

  Handlebar drop values were measured as the vertical distance from the top of the saddle surface to the centre of the handlebar clamping point.

- Crank length

A CycleOps 400 Indoor Pro Cycle (Power Tap: Saris Cycling Group®. Madison, WI. USA) was used for the purpose of a follow-on study. Saddle height, saddle setback, handlebar reach and handlebar height were set to match the configuration of the participant's own bicycle. Static joint angles of the ankle, knee, hip, shoulder and elbow were taken on the bicycle, using a digital inclinometer (Digi-Pas® DWL-90E model) as previously described (Holliday et al., 2017).

This was followed by an incremental exercise test to volitional exhaustion to determine eligibility in the study. The participants performed a standard warm-up and after a three minute rest period completed a Peak Power Output (PPO) and Peak Oxygen Consumption test. The CycleOps VirtualTraining app (VirtualTraining, version 1.7.3, Czech Republic) was used to control the ergometer and was set according to the participant's individual characteristics of age, mass and stature. Heart rate was captured by a Suunto® T6C heart rate monitor (Suunto Oy, Vanata, Finland). Gas analysis was monitored over 15 s intervals using an on-line breath-by-breath gas analyser and pneumotach (Oxycon, Viasis, Hoechberg, Germany). Participants started exercising at a workload of 100 W and resistance was increased by continuous ramp protocol at a rate of 20 W every 60 s until the participant was exhausted and could not sustain a cadence of at least 60 revolutions per minute (rpm). PPO was calculated by averaging the power output for the final minute of the VO_{2peak} test. VO_{2peak} was recorded as the highest VO_2 reading recorded for 30 s during the test. Maximum heart rate was recorded as the highest heart rate achieved during the incremental exercise test.

Data analysis

Stature, leg, and arm length were analysed for correlations with seat height and absolute handlebar reach and drop values. Total saddle height was calculated as the sum of the measured seat height and the crank length. The individual configuration of each participant's bicycle was analysed as a relative value as follows:

- Total saddle height was calculated as a percentage of trochanteric leg length.
- Saddle setback was calculated as a percentage of seat height.
- Handlebar drop was calculated as a percentage of seat height.
- Handlebar reach was calculated as a percentage of stature.

Statistical analysis

All bicycle configuration measurements, joint kinematics, anthropometrics, flexibility and training data are expressed as means and standard deviation (mean $\pm$ SD). A Pearson product-moment correlation coefficient was computed to assess the relationship between preferred bicycle configuration and anthropometrics, flexibility and training history. The statistical analyses were performed using GraphPad Prism v7.0c (GraphPad Software, San Diego, CA, USA). Correlation coefficient descriptors are defined as trivial (r = 0.0 - 0.1), small (0.1 - 0.3), moderate (0.3 - 0.5), large (0.5 – 0.7), very large (0.7 – 0.9) and almost perfect (0.9 – 1) (Hopkins, 2013).

RESULTS

The minimum, maximum and mean $\pm$ SD values of the participants for bicycle configuration, joint angles, flexibility results and training history are shown in **Table 3.4**.

There was a large correlation between stature and absolute handlebar reach (r = 0.51, p <0.001), however, arm length did not correlate with absolute handlebar reach (r = 0.24, p = 0.10). There was a very large correlation between leg length and seat height (r = 0.77, p <0.001). There was a moderate correlation between stature and saddle setback (r = 0.47, p <0.001), as well as a moderate correlation between leg length and saddle setback (r = 0.46, p = 0.003) and between arm length and saddle setback (r = 0.42, p = 0.003). There was a moderate correlation between leg length and handlebar drop (r = 0.40, p = 0.021), as well as a moderate correlation between KEA and handlebar drop (r = 0.47, p <0.001). There was a moderate correlation between KEA and stature (r = 0.31, p = 0.029). Individual scatter plots summarise the results (**Figure 3.1**).

There was no significant relationship between hamstring flexibility and total saddle height (r = 0.005 and p = 0.978). There were also no significant relationships between training history, training load and bicycle configuration.

TABLE 3.4: Mean $\pm$ standard deviation of bicycle configurations, joint angles, flexibility results and training history of participants.

Variable	**Minimum**	**Maximum**	**Mean $\pm$ SD**
Bicycle configuration			
Saddle height (seat height + crank length)	870	1040	942.79 $\pm$ 37.42mm
Saddle height as a % of leg length	93.93	103.32	97.14 $\pm$ 2.18%
Setback as a % of seat height	5.75	15.11	10.32 $\pm$ 2.25%
Drop as a % of seat height	1.66	21.45	13.03 $\pm$ 3.59%
Reach as a % of stature	33.78	38.77	35.92 $\pm$ 1.15%
Joint angles			
Ankle (BDC)	97°	133°	116 $\pm$ 7°
Knee (BDC)	20°	51°	36 $\pm$ 7°
Hip (TDC)	67°	86°	77 $\pm$ 5°
Shoulder	99°	129°	112 $\pm$ 7°
Elbow	3°	45°	19 $\pm$ 8°
Flexibility			
Sit and reach (cm)	16.00	54.00	38.99 $\pm$ 8.58cm
Knee Extension Angle	8°	80°	47 $\pm$ 16°
Fingertip to floor (cm)	-14.50	29.00	-0.18 $\pm$ 9.61cm
Modified Schober (cm)	20.00	25.00	21.88 $\pm$ 0.91cm
Training history			
Consecutive years of training	1.50	24.00	5.97 $\pm$ 4.21yrs
Average hours of training per week in last 3 months	4.00	20.00	11.04 $\pm$ 3.79hrs

BDC = bottom dead centre. TDC = top dead centre.

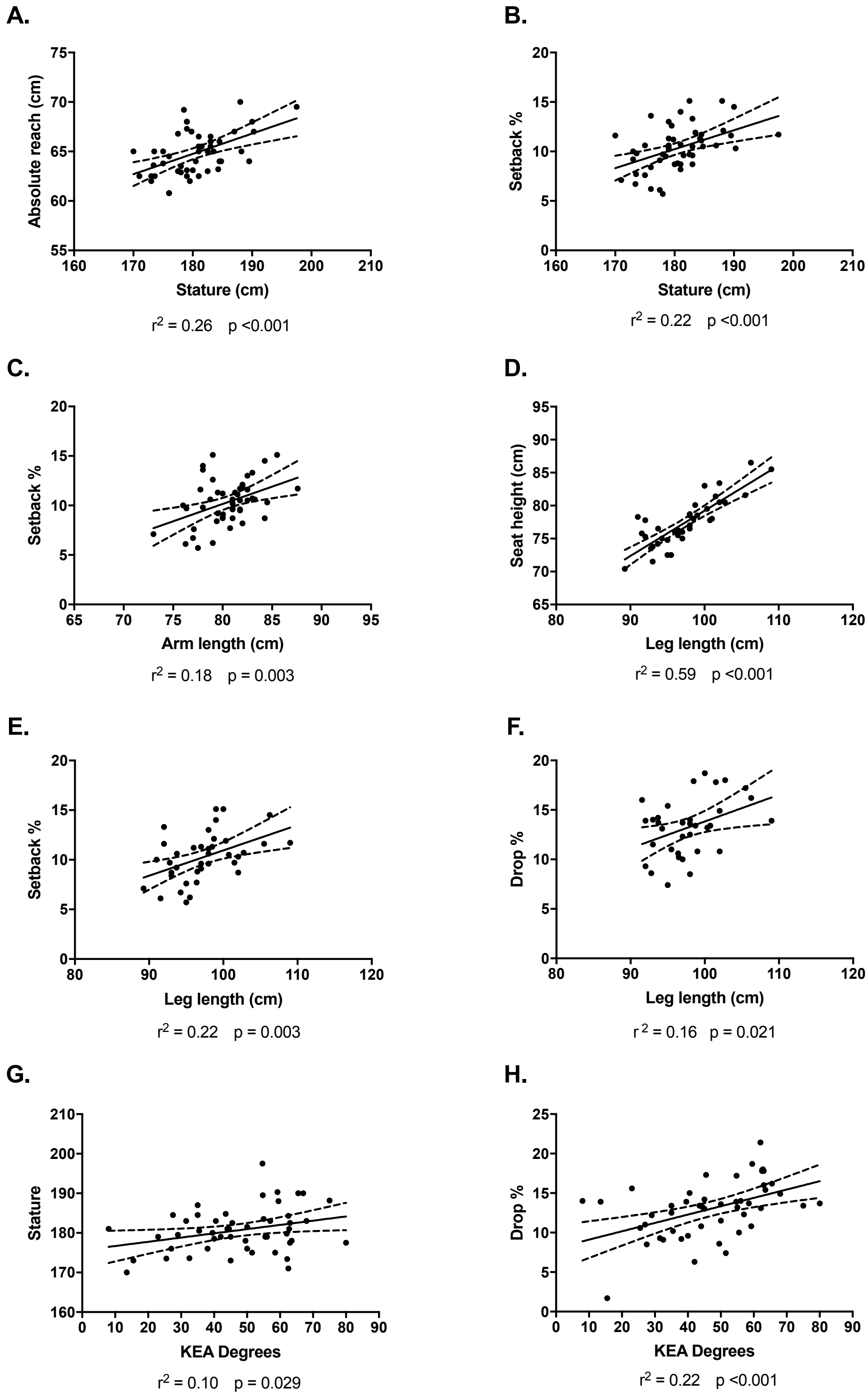


FIGURE 3.1: Scatter plots of significant correlations. A. Stature correlated to absolute handlebar reach. B. Stature correlated to saddle setback. C. Arm length correlated to saddle setback. D. Leg length correlated to seat height. E. Leg length correlated to saddle setback. F. Leg length correlated to handlebar drop. G. Knee Extension Angle (KEA) correlated to stature. H. KEA correlated to handlebar drop. A fully extended leg is measured as 90°, indicating good KEA flexibility.

DISCUSSION

Freely chosen bicycle configuration and joint angles

The optimal bicycle configuration studies to date have established a set of norms that aim to maximise power and minimise injury, yet often neglect to take cyclists' individual anthropometrics and comfort into consideration. The purpose of this study was to determine the association between intrinsic factors and freely chosen bicycle configuration, and to establish ranges of optimal angles for all the joints of the body as a starting point for static bicycle configuration.

Freely chosen bicycle configuration resulted in a mean KFA range of approximately 29° to 43°, with a mean KFA of 36°. This is similar to the findings of Dahlquist et al. (2015) who demonstrated a mean KFA of 34° despite more than 50% of their participants having had a professional bike fit. Cyclists tend to opt for a range of KFA similar to the recommended range of 25° to 35° for optimal power (Peveler, Pounders, and Bishop, 2007) with some cyclists selecting a lower saddle height than recommended. It should be taken into consideration that the KFA was measured in a natural riding position, not with the pedal horizontal as recommended by Holmes, Pruitt and Whalen (1994). These values may therefore conform more closely to those measured using dynamic methods, as it has been demonstrated that a change from static (using the Holmes method) to dynamic measurement of KFA differs by approximately 8° (Bini, Hume, and Croft, 2011). Our mean KFA of 36° may therefore correlate to approximately 28° using the Holme's method and falls close to the original recommendations of 25° to 35° for optimal performance and injury prevention (Holmes, Pruitt, and Whalen, 1994). The ankle and elbow joints also demonstrated similar joint ranges to previous recommendations.

There are limited recommendations for an optimal or comfortable hip flexion angle on the bicycle. Previous studies have determined hip flexion angle as a line bisecting the length of the femur and a line horizontal to the floor (Sanderson and Black, 2003; Bini and Diefenthaeler, 2010; Bini, Senger, et al., 2012), or as an angle bisecting the length of the femur and a line from the hip joint centre to the glenohumeral joint centre (Dingwell et al., 2008). These measures exclude the spinal segments and do not measure the hip joint independently (long axis of femur and lumbar spine-sacrum). From our results we recommend a static hip flexion angle range from 72-82° with the measured leg at top centre pedal position, and as an angle from the length of the femur bisecting a line parallel to the lower lumbar spine and sacrum.

Similar to the hip, the shoulder angle is often simplistically determined as an angle between the elbow, acromion and hip joint centre. A clinical shoulder angle will take the thoracic spine into account, as was done in this study. There has been no recommendation for optimal shoulder angle whilst riding with the hands on the hoods position. Our results recommend positioning the static shoulder angle in a range from 105° to 119° to set handlebar drop and handlebar reach.

Anthropometrics, flexibility and bicycle configuration

As expected, we demonstrated moderate to strong correlations between leg length and seat height, and saddle setback. This is in keeping with previous recommendations of measuring saddle height as a percentage of leg length (Hamley and Thomas, 1967; Nordeen-Snyder, 1977; De Vey Mestdagh, 1998) In addition there was a moderate to strong correlation between stature and absolute handlebar reach, and stature and saddle setback. A surprising finding was that arm length did not correlate with handlebar drop or reach.

Although not scientifically validated in his book, Andy Pruitt takes into consideration hamstring and lower back flexibility to determine handlebar reach (Andy Pruitt and Matheny, 2006). Despite our hypothesis that the sit and reach test would have a significant correlation with handlebar reach and handlebar drop being disproven, it is our suggestion that handlebar reach still take into account individual flexibility and comfort in this position. There was, however, a significant correlation between handlebar drop and hamstring flexibility. A greater saddle setback was also correlated with an increased hamstring flexibility.

From our results we can also determine that the taller a person and the longer their legs or arms, the more saddle setback they can achieve. This is similar to the method described by De Vey Mestdagh (1998) who suggests a formula related to the length of the cyclist's upper leg. The increased leg length and ability to stretch the hamstrings appear to enable the cyclist to adopt a more aggressive handlebar drop position. As the cyclist leans forwards to place his hands on the handlebars, the lumbar spine needs to flex and the pelvis should rotate anteriorly. It has previously been demonstrated that elite cyclists have a greater anterior pelvic tilt compared to non-cyclists (McEvoy, Wilkie, and Williams, 2007) and that greater hamstring flexibility allows for more lumbar flexion range and pelvic tilt (Congdon, Bohannon, and Tiberio, 2005). Even though no correlation between hamstring extensibility and spinal curvatures and pelvic tilt on the bicycle was determined (Muyor, López-Miñarro, and Alacid, 2011), an outdated hamstring test to assess extensibility was used, as well as testing with the hands in the drops handlebar position. Similarly, should the cyclist wish to adopt a more aerodynamic position with an increased handlebar drop, they should aim to improve their hamstring flexibility in order to comfortably maintain this position.

Previous reports have suggested that the more experienced cyclist will adopt a lower handlebar drop position (Burt, 2014), and that a fairly new cyclist to the sport will cycle in a more upright position (I. Priego Quesada et al., 2017). We demonstrated no relationship between a history of cycling and bicycle configuration, although none of our cyclists were new to the sport, having cycled consistently for more than 18 months. Likewise there were no significant correlations between training history and freely chosen bicycle configuration.

Despite small to moderate correlations, which indicate association between variables, the regression analysis yielded low r^2 values. This indicates that further research with a higher sample size is required to effectively establish if one or multiple variables are the cause of another (i.e. higher saddle height directly results in an increased VO_{2max} in an individual).

CONCLUSION

The average static joint ranges demonstrated in this study are similar to previous recommendations, but additionally hip and shoulder angle ranges are now suggested. We recommend that one starts the bicycle fitting by systematically guiding the cyclist into these recommended ranges, and then further optimising the fit using more elaborate methods, starting with the saddle. Most bike fitters have agreed that the saddle is a good starting point (Burt, 2014). Perception, comfort and anthropometric measures of the cyclist should also be taken into account (J. Priego Quesada et al., 2018). A taller cyclist with longer arms and legs will guide the bicycle configuration into a greater saddle setback and increased handlebar drop position. Likewise handlebar drop can also be guided by hamstring flexibility.

The results from this study provide further recommendations for static bicycle fitting. Future research should explore the effect of steady state or differing cycling intensities on these joint angle ranges.

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Appendix

3.A Flexibility testing

1. Sit and reach test

A standard sit-and-reach box (30.5cm in height) with a sliding ruler on top centre was used. The ruler was placed so that the 35cm mark was in line with the participant's toes in a neutral ankle position. In this way, there was always a positive reading. 0cm indicated reduced flexibility and 50cm high flexibility. The participants were asked to sit on the plinth with legs together and knees straight. The participant was not wearing shoes and the soles of their feet were placed against the edge of the sit-and-reach box. Verbal instructions were given as follows "Place one hand on top of the other, with palms down and keeping your knees, arms and fingers straight, slowly reach forwards towards your toes. Reach as far as possible. Hold that maximal stretch for 6 seconds." The test administrator ensured that the heels remained at the 35cm mark and that knees were fully extended throughout the test. Scores were recorded in centimetres to the nearest 0.5cm using the ruler on the sit-and-reach box. This test has moderate validity, acceptable reproducibility and is a simple procedure to administer (Ayala et al., 2012).

2. Knee Extension Angle to determine hamstring length.

Hamstring length was determined by using the Knee Extension Angle test (Scott Davis et al., 2008). Participants were lying supine with the spine and pelvis in a neutral position, hips and knees fully extended. The first inclinometer was placed on the distal end of the tested leg, immediately superior to the patella. A second inclinometer was placed on the distal anterior tibia, along the distal edge aligned with the superior aspect of the medial malleolus. The tested leg was raised passively by the tester to 90° of hip flexion as recorded by the inclinometer placed on the distal thigh. The participant's knee was then passively straightened until a strong stretch was reportedly felt in the posterior thigh by the participant. The contralateral leg was fixed in an extended position on the plinth by a strap over the distal thigh. The Knee Extension Angle was measured in degrees using the inclinometer placed on the tibia. The benchmark was the ability to extend to within 10° of full extension. A fully extended leg is measured as 90°, indicating good flexibility.

3. Fingertip to floor test

The fingertip to floor test is a quick and easy test to measure trunk flexion. The participant stood on a 20cm high box with shoes removed and feet together. He was asked to bend forward as far as possible, whilst maintaining knees, arms and fingers fully extended. The vertical distance between the top of the box and the tip of the participant's middle finger was measured in centimetres. The distance is positive if the participant did not touch the box and expressed in a negative figure if able to go beyond the top of the box.

4. Modified Schober test

The modified Schober test was done with the participant standing upright and barefoot. The tester palpated the lumbosacral joint, and placed the first marker 5cm below this, and the second marker 10cm above the lumbosacral joint (total of 15cm between markers). The participant was asked again to bend forward as far as possible. The tester measures the distance between the markers. The increase in the distance between the marks was then calculated as the measure of lumbar flexion range of motion.

Both the fingertip to floor and modified Schober tests were used together to evaluate the participant's lumbar flexion range of motion and functional ability to bend forward. The fingertip to floor test has excellent reliability and good sensitivity (Perret et al., 2001) and both the modified Schober test and fingertip to floor test have excellent intertester reliability (Robinson and Mengshoel, 2014). Both tests are best used in a test-retest situation to assess treatment effect on lumbar range of movement and flexibility.

Chapter 4

Performance variables relative to freely chosen bicycle configuration and flexibility

ABSTRACT

Introduction: Cycling races are often won by the smallest of margins. Research has focused on optimal saddle height for performance, however the relationship between freely chosen bicycle configuration and individual factors such as anthropometrics and flexibility have not yet been investigated adequately. The aim of this study was to determine if a relationship between power production, bicycle configuration and flexibility exists, and how best to apply this clinically to the cyclist during the bike fitting process.

Methods: Fifty male cyclists were recruited for the study. Individual anthropometrics, flexibility and individual bicycle configuration were recorded before the participants performed a peak power output and peak oxygen consumption test to determine their VO_{2max}. A Pearson correlation coefficient was computed to assess the relationship between preferred bicycle configuration, anthropometrics and flexibility.

Results: There was a significant correlation between performance and hamstring flexibility, handlebar drop, saddle setback and ankle plantarflexion. An increased lumbar flexibility demonstrated an inverse relationship with relative VO_{2max}. A more anteriorly rotated pelvis correlated with improved hamstring flexibility, hip flexion angle and an increased handlebar drop.

Conclusion: The results from this study have clinical implications for bike fitters and cyclists. Greater saddle setback and lower handlebar height may increase peak power output. Improving a cyclist's flexibility and ability to adopt an anteriorly rotated pelvis and lower handlebar height may increase the force generated in the push phase of the pedal stroke and thus improve cycling performance.

INTRODUCTION

Sporting performance at an elite level can be determined by the smallest of margins. Differences of less than 1% can determine the outcome between winning a race or finishing out of points. For example, the 1989 Tour de France was won by a difference of only two-thousandths of a percent. As a result, research studies and practical investigations into various interventions such as nutritional supplementation, training methods (E. Faria, Parker, and I. Faria, 2005), aerodynamics (Oggiano et al., 2008) and biomechanical factors (Peveler, Pounders, and Bishop, 2007; Peveler and Green, 2011) relating to performance are popular and frequently undertaken in both the field and laboratory setting.

There are limited scientific studies which have investigated the relationship between cycling performance and the various components of bicycle configuration such as saddle setback, handlebar reach and handlebar drop. Recommendations for the optimal setting of these variables have previously been based on personal perspective, comfort and injury prevention (De Vey Mestdagh, 1998; Burke, 2003; Silberman et al., 2005; Burt, 2014).

Similarly, there are limited studies to date which have evaluated the flexibility of the lower limb and torso and how this relates to the production of power output in cycling. A study investigated the flexibility characteristics of ‘successful’ and ‘less successful’ road cyclists as predictors for performance capability (Coetzee and Malan, 2018). The passive straight leg raise test, modified Thomas Iliopsoas test and modified Thomas Quadriceps test were conducted, with no significant differences found between the two participant groups. A direct relationship between performance and flexibility was however not investigated. In running related research, there are mixed results regarding flexibility and improved running economy. In a review of the literature (Barnes and Kilding, 2014), equivocal results with regards to the effect of stretching and flexibility on running economy were demonstrated. However, the general consensus was that better flexibility resulted in improved running performance.

The aim of this study was to determine if a relationship between power production and bicycle configuration and flexibility exists, and how best to apply this clinically to the cyclist and the bike fitting process.

METHODS

Participants

Fifty well-trained male road cyclists (30 $\pm$ 9 years, 76.5 $\pm$ 7.9kg, 180.7 $\pm$ 5.6cm) conforming to Level 2 or greater (De Pauw et al., 2013) were recruited for this study. The general characteristics and performance parameters of the 50 cyclists are shown in **Table 4.1**. Participants were excluded if they had made any changes to their bicycle configuration in the past three months, or if they experienced any pain or discomfort on their current bicycle configuration. Prior to testing, each participant was informed of the risks and stresses associated with participation in the research trial, were personally interviewed about their training history, completed a Physical Activity Readiness Questionnaire (PAR-Q) (Whaley, Brubaker, and Otto, 2007) and signed an informed consent form. The study was approved by the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Cape Town, and conformed to the principles of the World Medical Association Declaration of Helsinki (World Medical Association, 2013).

TABLE 4.1: General characteristics of cyclists (n=50)

Variable	**Mean $\pm$ SD**
Age (years)	30 $\pm$ 9
Body mass (kg)	76.5 $\pm$ 7.9
Stature (cm)	180.7 $\pm$ 5.6
Trochanteric leg length (cm)	97.5 $\pm$ 4.4
Percentage body fat (%)	11.9 $\pm$ 4.7
Sum of seven skinfolds (mm)	61.5 $\pm$ 20.2
PPO (W)	387.7 $\pm$ 53.1
PPO (W/kg)	5.1 $\pm$ 0.7
VO_{2max} (ml/kg/min) Relative	58.8 $\pm$ 7.7

Testing procedure

The participants reported to the laboratory with their own bicycle, cycling shoes and pedals. On the visit to the laboratory, the participant's anthropometric measurements including stature, body mass, arm length, trochanteric leg length and sum of seven skinfolds (triceps, biceps, supra-iliac, sub-scapular, calf, thigh and abdomen) were taken, and body fat percentage was determined (Ross and Marfell-Jones, 1991).

The participants then underwent a series of flexibility tests:

1. Sit and reach test.
2. Knee Extension Angle (KEA) to assess hamstring length.
3. Fingertip to floor test.
4. Modified Schober test.

See **Appendix 4.A** for detailed descriptions of testing.

The participant's own bicycle configuration was measured according to a set of objectively reproducible measurements:

- Seat height

  The seat height was measured from the centre of the crank axle to the top of the saddle, passing through a reference line set at 74 degrees to the horizontal to standardise the seat tube angle.

- Saddle setback

  Saddle setback was measured as the horizontal distance from the front of the saddle to the centre of the crank axle. The front of the saddle was determined based on a standardised distance of 22.5cm from the contact point of the ischia to the front of the saddle. For saddles which did not conform to these measurements, a correction value was applied to the measured setback.

- Handlebar reach

  The handlebar reach was measured horizontally, from the centre of the handlebar clamping point to the centre of the 74° seat tube reference line.

- Handlebar drop

  Handlebar drop values were measured as the vertical distance from the top of the saddle surface to the centre of the handlebar clamping point.

- Crank length

A CycleOps 400 Indoor Pro Cycle (Power Tap: Saris Cycling Group®. Madison, WI. USA) ergometer was used during all performance testing. Saddle height, saddle setback, handlebar reach, and handlebar height were set to match the configuration of the participant's own bicycle. Static joint angles of the ankle, knee, hip, shoulder and elbow were taken on the bicycle, using a digital inclinometer (Digi-Pas® DWL-90E model) as previously described (Holliday et al., 2017). The position of the pelvis was determined by the angle of the posterior surface of the lowest lumbar vertebrae and the proximal three sacral vertebrae to the horizontal. This was determined as the area in the midline, directly below a line joining the iliac crests. A lower angle indicates more of an anterior pelvic tilt.

This was followed by an incremental exercise test to volitional exhaustion. The participants performed a standard warm-up and after a three minute rest period completed a Peak Power Output (PPO) and Peak Oxygen Consumption test. The CycleOps VirtualTraining app (VirtualTraining, version 1.7.3, Czech Republic) was used to control the ergometer and was set according to the participant's individual characteristics of age, mass and stature. Heart rate was captured by a Suunto® T6C heart rate monitor (Suunto Oy, Vanata, Finland). Expiratory gas analysis was monitored over 15 s intervals using an on-line breath-by-breath gas analyser and pneumotach (Oxycon, Viasis, Hoechberg, Germany). Participants started exercising at a workload of 100 W and workload was increased by means of a continuous ramp protocol at a rate of 20 W every 60 s until the participant was exhausted and could not sustain a cadence of at least 60 revolutions per minute (rpm). PPO was calculated as the average of the power output for the final minute of the VO_{2peak} test. VO_{2peak} was recorded as the highest VO_2 reading recorded for 30 s during the test. Maximum heart rate was recorded as the highest heart rate achieved during the incremental exercise test.

Data analysis

Total saddle height was calculated as the sum of the measured seat height and the crank length. The individual configuration of each participant's bicycle was analysed as a relative value as follows:

- Total saddle height was calculated as a percentage of trochanteric leg length.
- Saddle setback was calculated as a percentage of seat height.
- Handlebar drop was calculated as a percentage of seat height.
- Handlebar reach was calculated as a percentage of stature.

Statistical analysis

All bicycle configuration measurements, joint kinematics, flexibility and performance variables are expressed as means and standard deviation (mean $\pm$ SD). A Pearson product-moment correlation coefficient was computed to assess the relationship between peak power output and VO_{2max} with flexibility, preferred bicycle configuration and joint kinematics. The statistical analyses were performed using GraphPad Prism v7.0c (GraphPad Software, San Diego, CA, USA). Correlation coefficient descriptors are defined as trivial (r = 0.0 - 0.1), small (0.1 - 0.3), moderate (0.3 - 0.5), large (0.5 – 0.7), very large (0.7 – 0.9) and almost perfect (0.9 – 1) (Hopkins, 2013).

RESULTS

The minimum, maximum and mean $\pm$ SD values of the participants for bicycle configuration, joint angles, flexibility results and training history are shown in **Table 4.2**.

Flexibility

There was a small correlation between hamstring flexibility (as measured using the KEA) and a moderate correlation with relative VO_{2max} (r = 0.29, p = 0.046) and relative PPO (r = 0.33, p = 0.022) (**Figure 4.1**). There was also a moderate relationship between the KEA and absolute PPO (r = 0.46, p = 0.001) and absolute VO_{2max} (r = 0.43, p = 0.002) (**Figure 4.2**). An increased lumbar flexibility (as measured with the modified Schober test) demonstrated an inverse relationship with relative VO_{2max} (r = -0.41, p = 0.004), indicating that the more flexible the lumbar spine the lower the relative VO_{2max} (**Figure 4.1**).

Bicycle configuration

A greater handlebar drop position moderately correlated with a greater relative PPO (r = 0.44, p = 0.002) (**Figure 4.1**), as well as a largely significant absolute PPO (r = 0.51, p <0.001) and a moderate correlation with absolute VO_{2max} (r = 0.31, p = 0.033). An increased saddle setback had a moderate correlation with an increased absolute PPO (r = 0.35, p = 0.017) (**Figure 4.2**).

TABLE 4.2: Mean ± standard deviation of bicycle configurations, joint angles, flexibility results and training history of participants.

Variable	Minimum	Maximum	Mean ± SD
Bicycle configuration			
Saddle height (seat height + crank length)	870	1040	942.79 ± 37.42mm
Saddle height as a % of leg length	93.93	103.32	97.14 ± 2.18%
Setback as a % of seat height	5.75	15.11	10.32 ± 2.25%
Drop as a % of seat height	1.66	21.45	13.03 ± 3.59%
Reach as a % of stature	33.78	38.77	35.92 ± 1.15%
Joint angles			
Ankle (BDC)	97°	133°	116 ± 7°
Knee (BDC)	20°	51°	36 ± 7°
Hip (TDC)	67°	86°	77 ± 5°
Pelvis	49°	70°	59 ± 5°
Shoulder	99°	129°	112 ± 7°
Elbow	3°	45°	19 ± 8°
Flexibility			
Sit and reach (cm)	16.00	54.00	38.99 ± 8.58cm
Knee Extension Angle	8°	80°	47 ± 16°
Fingertip to floor (cm)	-14.50	29.00	-0.18 ± 9.61cm
Modified Schober (cm)	20.00	25.00	21.88 ± 0.91cm
Training history			
Consecutive years of training	1.50	24.00	5.97 ± 4.21yrs
Average hours of training per week in last 3 months	4.00	20.00	11.04 ± 3.79hrs

BDC = bottom dead centre. TDC = top dead centre.

Joint kinematics

An increase in ankle plantarflexion resulted in a higher absolute PPO and absolute and relative VO_{2max} (**Figure 4.2**). There were no statistically significant correlations between any of the other body joints and power production. Although there was no significant correlation between pelvic angle and any of the performance variables, a more anteriorly rotated pelvis moderately correlated to KEA ($r = -0.43$, $p = 0.002$), had a very large correlation with hip flexion angle at TDC ($r = 0.76$, $p < 0.001$) and moderately correlated with an increased handlebar drop ($r = -0.34$, $p = 0.017$) (**Figure 4.3**).

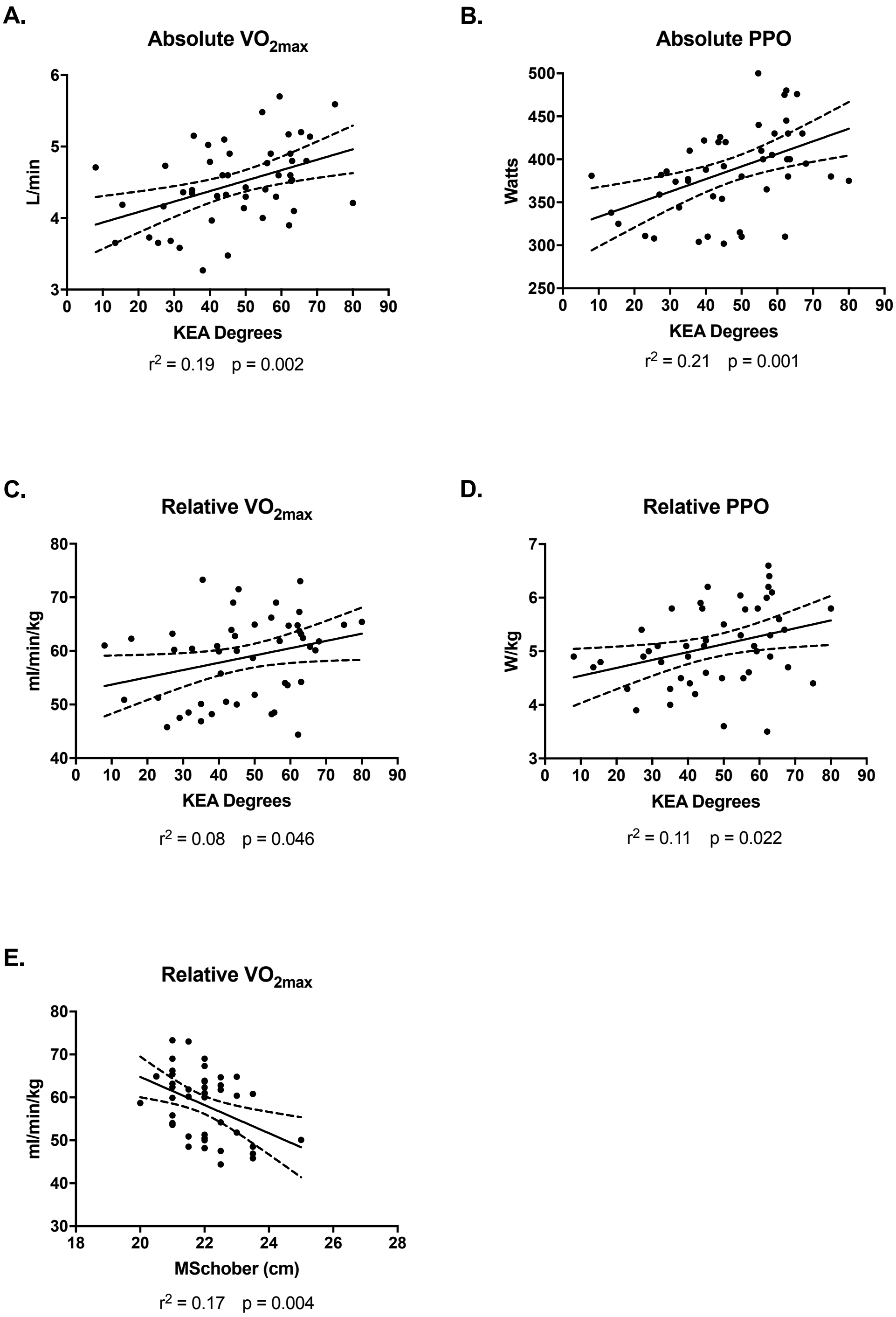

FIGURE 4.1: Scatter plots of significant flexibility correlations. A. Knee Extension Angle (KEA) correlated to Absolute VO_{2max} B. KEA correlated to Absolute PPO. C. KEA correlated to Relative VO_{2max}. D. KEA correlated to Relative PPO. E. Modified Schober correlated to Relative VO_{2max}.

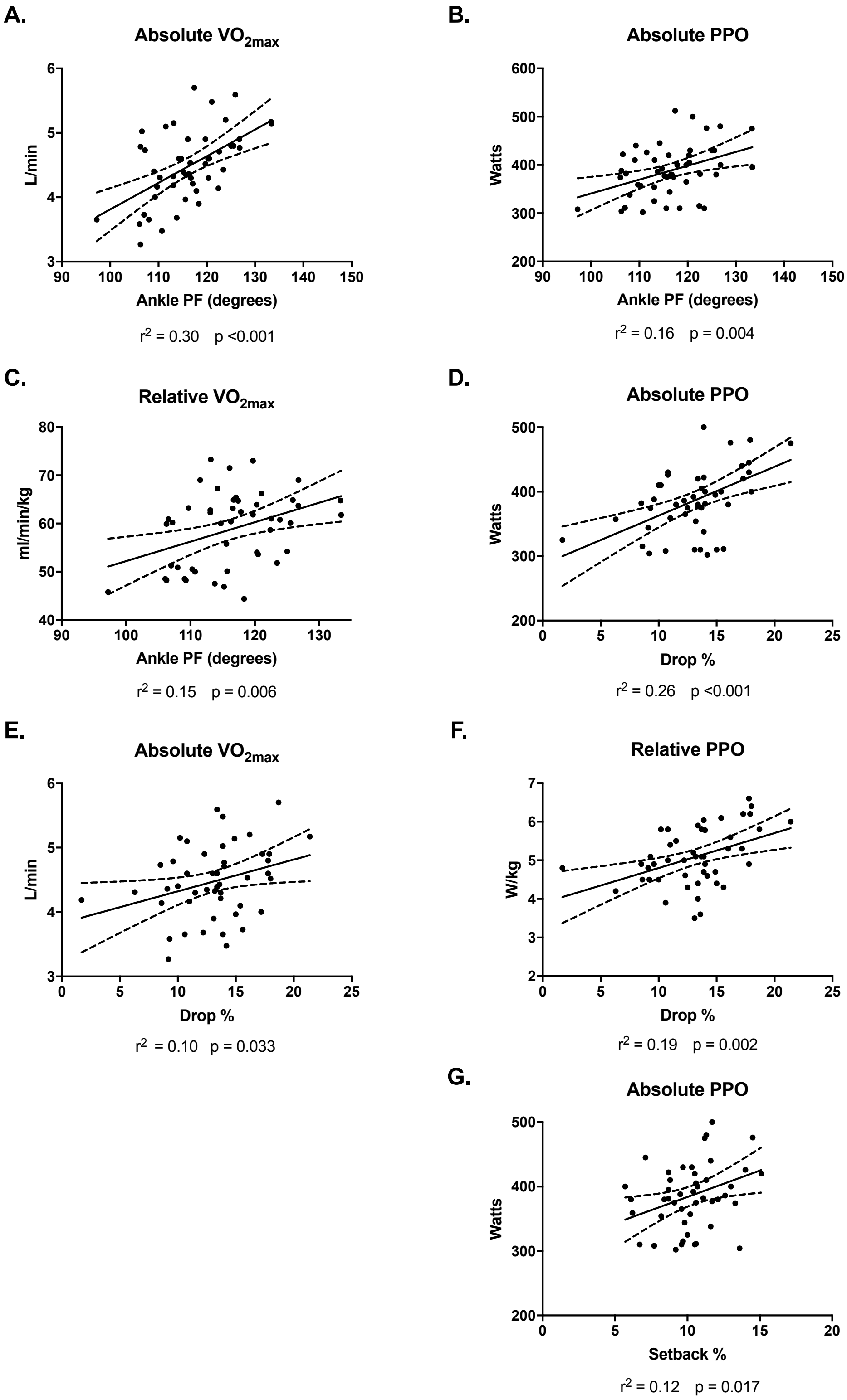

FIGURE 4.2: Scatter plots of significant joint and bicycle configuration correlations. A. Ankle plantarflexion (PF) correlated to Absolute VO_{2max}. B. Ankle PF correlated to Absolute PPO. C. Ankle PF correlated to Relative VO_{2max}. D. Handlebar drop correlated to Absolute PPO. E. Handlebar drop correlated to Absolute VO_{2max}. F. Handlebar drop correlated to Relative PPO. G. Saddle setback correlated to Absolute PPO.

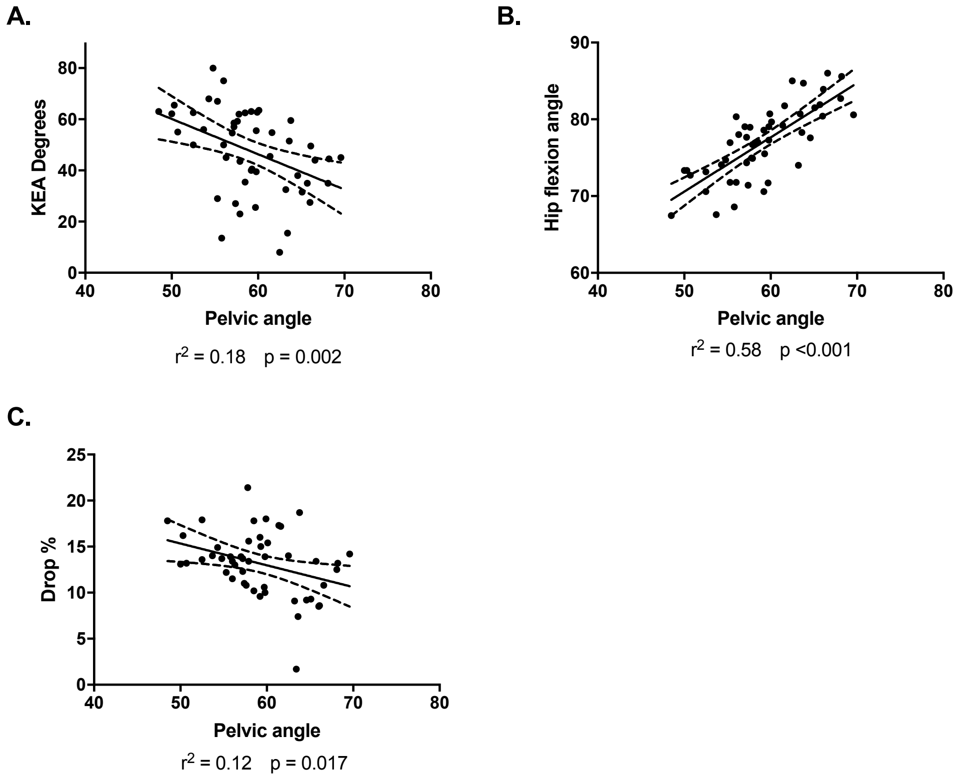

FIGURE 4.3: Significant pelvic angle correlations. A. Pelvic angle correlated to Knee Extension Angle (KEA). B. Pelvic angle correlated to hip flexion angle. C. Pelvic angle correlated to handlebar drop.

DISCUSSION

The aim of this study was to determine whether a relationship between power production, bicycle configuration and flexibility exists, and how best to apply this clinically to the cyclist and the bike fitting process.

Our data demonstrate that increased hamstring flexibility and a lower handlebar position correlated significantly with relative VO_{2max} and peak power output. The hamstring muscles originate from the ischial tuberosity of the pelvis and insert below the knee onto the medial tibial tuberosity and fibular head. These bi-articular muscles work to extend the hip and flex the knee and are stretched when the hip is flexed and the knee is extended. A limitation in hamstring flexibility may therefore cause the pelvis to rotate posteriorly while a greater hamstring flexibility may allow the cyclist to selectively rotate the pelvis into a more anteriorly rotated position. This may be facilitated by a lower handlebar position.

A secondary gain of adjusting the handlebars into a relatively lower position may be a more favourable position for aerodynamic drag reduction (Oggiano et al., 2008), however, aerodynamic positioning was not a factor which was measured in this study.

Lumbar flexibility, as measured with the modified Schober test which assesses the lumbar vertebrae flexibility independent of other joints, demonstrated a significant inverse correlation with an increased relative VO_{2max}. There is no existing data to indicate that an increased or decreased spinal mobility can alter performance in cycling, however it has been previously demonstrated that elite cyclists have a greater anterior pelvic tilt angle compared to non-cyclists (McEvoy, Wilkie, and Williams, 2007). A similar inverse relationship between spinal flexibility and anterior pelvic tilt at varying intensities of cycling has been described (Sauer et al., 2007). An increased lumbar flexibility correlated to less of an anterior pelvic tilt. These studies align with our own findings. Although pelvic tilt was not correlated with an increased VO_{2max} or PPO, there was a correlation between a more anteriorly rotated pelvic tilt and hamstring flexibility, hip flexion angle and handlebar drop. We suggest that a reduction in lumbar flexibility and an increase in hamstring flexibility promotes the adoption of a more anteriorly rotated pelvis and that this is facilitated by a lower handlebar position.

Our data confirm that an anteriorly rotated pelvic position, with a lower handlebar height, is correlated with the greatest hip flexion at top dead centre (TDC). Previous studies have demonstrated that a greater hip flexion angle with a flexed knee (ie TDC position) places the Gluteus Maximus and possibly the hamstring muscles in an optimal length-tension position, which leads to a greater hip extension torque (Bazett-Jones et al., 2017), as well as a greater knee extension torque (Ema, Wakahara, and Kawakami, 2017). The increased hip and knee extension torque would imply a greater force applied to the pedal in the push phase, thus producing a more powerful downstroke.

The plantarflexor muscles are predominately recruited with the downward propulsive phase (So, J. Ng, and G. Ng, 2005), and it has previously been demonstrated that the plantarflexor peak force is increased with a greater saddle setback position (Hayot et al., 2013). The results from our study confirm this as there was a significant relation between greater saddle setback, ankle plantarflexion angle and absolute PPO. Ankle plantarflexion may be explained as secondary to the anteriorly rotated pelvis position. When adopting the anteriorly rotated pelvis position, the increased stretch of the hamstring muscles may result in a reflexive increase in knee flexion to reduce dynamic tensioning of the hamstring muscle during the knee extension phase. Ankle plantarflexion facilitates increased knee flexion angle (Peveler, Shew, et al., 2012). Saddle fore-aft position alters the effective seat tube angle which may affect muscle activation patterns of the lower limbs in cycling. To date there has been conflicting evidence in the literature with regards to the alteration of muscle activity primarily due to inadequate kinematic controls in these studies. However, more recent research has demonstrated that there is a significant increase in muscle activity in the Bicep Femoris, Gluteus Maximus and Medial Gastrocnemius with reductions in effective seat tube angle (McDonald et al., 2018). Increased saddle setback may therefore selectively recruit a larger muscle mass and facilitate higher power outputs.

Despite previous research demonstrating a relationship between knee flexion angle and power production (Peveler, Pounders, and Bishop, 2007; Peveler and Green, 2011), there was no significant relationship between these variables in our study. It is not immediately evident why this is the case.

CONCLUSION

The findings of this study demonstrate that there is a small to moderate correlation between flexibility and power, as well as some bicycle configuration variables. An increased hamstring flexibility allows for a greater anterior pelvic tilt which, combined with a lower handlebar height, positions the lower limb muscles optimally for force generation.

There is limited research investigating the optimal fore-aft position of the saddle and handlebar height and most of the published recommendations are based on personal experience (De Vey Mestdagh, 1998; Silberman et al., 2005). Greater saddle setback and lower handlebar height may increase peak power output while a lower handlebar position may be an indication of increased hamstring flexibility and facilitate a more anteriorly rotated, powerful position. However, further research should focus on optimal guidelines for these other variables of bicycle configuration, not limited to the well-researched optimal saddle height.

The results from this study have clinical implications for bike fitters and cyclists. Improving a cyclist's flexibility and ability to adopt an anteriorly rotated pelvis and lower handlebar height, may increase the force generated in the push phase of the pedal stroke and thus improve their cycling performance.

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Appendix

4.A Flexibility testing

1. Sit and reach test

A standard sit-and-reach box (30.5cm in height) with a sliding ruler on top centre was used. The ruler was placed so that the 35cm mark was in line with the participant's toes in a neutral ankle position. In this way, there was always a positive reading. 0cm indicated reduced flexibility and 50cm high flexibility. The participants were asked to sit on the plinth with legs together and knees straight. The participant was not wearing shoes and the soles of their feet were placed against the edge of the sit-and-reach box. Verbal instructions were given as follows "Place one hand on top of the other, with palms down and keeping your knees, arms and fingers straight, slowly reach forwards towards your toes. Reach as far as possible. Hold that maximal stretch for 6 seconds." The test administrator ensured that the heels remained at the 35cm mark and that knees were fully extended throughout the test. Scores were recorded in centimetres to the nearest 0.5cm using the ruler on the sit-and-reach box. This test has moderate validity, acceptable reproducibility and is a simple procedure to administer (Ayala et al., 2012).

2. Knee Extension Angle to determine hamstring length.

Hamstring length was determined by using the Knee Extension Angle test (Scott Davis et al., 2008). Participants were lying supine with the spine and pelvis in a neutral position, hips and knees fully extended. The first inclinometer was placed on the distal end of the tested leg, immediately superior to the patella. A second inclinometer was placed on the distal anterior tibia, along the distal edge aligned with the superior aspect of the medial malleolus. The tested leg was raised passively by the tester to 90° of hip flexion as recorded by the inclinometer placed on the distal thigh. The participant's knee was then passively straightened until a strong stretch was reportedly felt in the posterior thigh by the participant. The contralateral leg was fixed in an extended position on the plinth by a strap over the distal thigh. The Knee Extension Angle was measured in degrees using the inclinometer placed on the tibia. The benchmark was the ability to extend to within 10° of full extension. A fully extended leg is measured as 90°, indicating good flexibility.

3. Fingertip to floor test

The fingertip to floor test is a quick and easy test to measure trunk flexion. The participant stood on a 20cm high box with shoes removed and feet together. He was asked to bend forward as far as possible, whilst maintaining knees, arms and fingers fully extended. The vertical distance between the top of the box and the tip of the participant's middle finger was measured in centimetres. The distance is positive if the participant did not touch the box and expressed in a negative figure if able to go beyond the top of the box.

4. Modified Schober test

The modified Schober test was done with the participant standing upright and barefoot. The tester palpated the lumbosacral joint and placed the first marker 5cm below this, and the second marker 10cm above the lumbosacral joint (total of 15cm between markers). The participant was asked again to bend forward as far as possible. The tester measures the distance between the markers. The increase in the distance between the marks was then calculated as the measure of lumbar flexion range of motion.

Both the fingertip to floor and modified Schober tests were used together to evaluate the participant's lumbar flexion range of motion and functional ability to bend forward. The fingertip to floor test has excellent reliability and good sensitivity (Perret et al., 2001) and both the modified Schober test and fingertip to floor test have excellent intertester reliability (Robinson and Mengshoel, 2014). Both tests are best used in a test-retest situation to assess treatment effect on lumbar range of movement and flexibility.

Chapter 5

Cycling: Joint kinematics and muscle activity in steady state cycling

W Holliday, *R Theo, J Fisher and J Swart*

ABSTRACT

It is important for bike fitters to consider the effects of training intensity and duration on kinematics and muscle activity. Cycling kinematics and electromyographic (EMG) patterns may change with fatigue and with increased workload intensity, however limited data has been published on the effects of a prolonged steady state cycle on cyclists' three dimensional whole-body kinematics and EMG magnitudes. We therefore aimed to assess the changes in lower limb EMG magnitudes and 3D kinematics of the ankle, knee, hip, lumbar, thoracic, shoulder and elbow angles of cyclists during a 60 minute steady state cycle at 60% of their VO_{2max} power.

Seventeen well-trained male cyclists were enrolled for the study. Participants performed a VO_{2max} test, followed by two trials of an hour duration each, where kinematic and EMG magnitudes were analysed for each third of the one hour steady state cycle.

Steady state training at a moderately high intensity does not appear to have any meaningful effect on cyclists' kinematic variables or muscle magnitudes. Bike fitting for steady state endurance rides can be performed as per previous recommendations, since no positional nor muscle magnitude changes were encountered.

Keywords: *Cycling; biomechanics; kinematics; EMG; bike fitting*

INTRODUCTION

Many a cyclist will seek out a bike fit with the purpose of enhancing their cycling experience. In order to optimise the cyclist's performance and minimise their risk of injury, the bike fitter should fully understand the interaction between the bicycle configuration and the cyclist's position as well as the type of training the cyclist will undertake. With the aim to maximise the use of muscle coordination patterns learned during training, it has been suggested that cyclists could benefit by training in the same conditions that they would race in (Blake, Champoux, and Wakeling, 2012). Multi-stage professional road cycling races are characterised by high intensity spurts interspersed with lower intensity recovery periods. Cyclists will spend approximately 70-90% of the time at a low exercise intensity (depending on the terrain), and 10-30% at a moderate to high exercise intensity (Padilla et al., 2001).

Cyclists will prepare for races by variably combining long steady endurance sessions with high intensity maximal workload interval sessions (Laursen and Jenkins, 2002; Laursen, 2010). It is important to analyse the muscle magnitudes experienced during these rides in order to strengthen the muscles in the correct range and intensity. Likewise, analysing the position of the cyclist on their bicycle during these training rides and races is important to provide insight into preventing overuse injuries. There is very little data to date on the effects of a steady state cycle on kinematics and EMG magnitudes, however, it is known that kinematics of the lower limb change with fatigue and changes in intensity (So, J. Ng, and G. Ng, 2005; Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010).

Change in the coordinative pattern of muscles of the lower limb occurs with the onset of fatigue induced by maximal intensity cycling (Dingwell et al., 2008). The same study demonstrated that a change in the EMG median frequency signal preceded a change in movement kinematics, and early into the protocol most subjects shifted towards a greater trunk lean angle while all subjects displayed an increase into dorsiflexion of the ankle joint. In conclusion, they found that as fatigue occurs, cyclists may change their muscle activation patterns to maintain performance. This subsequently may lead to maladaptive joint loading caused by changes in kinematics with fatigue at maximal effort. An increase of knee and hip extension has previously been demonstrated at maximal workloads (Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010) and could be linked to a shift in forward position on the bicycle (Bini, Senger, et al., 2012).

Only one study has investigated neuromuscular changes during steady state cycling at a lower intensity (55% maximal aerobic power) (Lepers et al., 2002). They only measured the Vastus Medialis Oblique (VMO) and Vastus Lateralis Oblique (VLO) muscles and these demonstrated a progressively reduced maximal voluntary force-generating capability. The contractile properties of the muscle were significantly altered after the first hour, whereas central drive and excitability were more affected towards the end of the five hour trial. No study to date has measured EMG across multiple muscle groups in the lower limb during prolonged steady state cycling. In addition, no studies to date have assessed how prolonged cycling at a moderately high intensity affects full body kinematics. How the existing findings translate to kinematics and EMG magnitudes during steady state cycling therefore needs to be investigated further. Nor are there guidelines for clinicians and bike fitters for optimising full body kinematics in 3D for prolonged exercise. Proper knowledge of how the muscle magnitudes and the position of the cyclists alter during steady state training is essential to minimise injury and optimise power output.

The aim of this study was therefore to assess EMG magnitudes of seven lower limb muscles as well as 3D kinematics of the ankle, knee, hip, lumbar, thoracic, shoulder and elbow angles of cyclists during a steady state cycle at 60% of their VO_{2max} power.

It was hypothesised that there may be small but significant changes in full body kinematics and muscle magnitudes in the last third of the testing protocol.

METHODS

Participants

Seventeen well-trained male road cyclists (31 $\pm$ 9 years, 75.5 $\pm$ 7.5kg, 178.4 $\pm$ 4.4cm) conforming to Level 2 or greater (De Pauw et al., 2013) were recruited for this study. The general characteristics and performance parameters of the 17 cyclists are shown in **Table 5.1**. Prior to testing, each participant was informed of the risks and stresses associated with participation in the research trial, were personally interviewed about their training history, completed a Physical Activity Readiness Questionnaire (PAR-Q) (Whaley, Brubaker, and Otto, 2007) and signed an informed consent form. The study was approved by the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Cape Town, and conformed to the principles of the World Medical Association Declaration of Helsinki (World Medical Association, 2013).

Testing procedure

The participants reported to the laboratory on three separate occasions (one week apart, over three weeks) with their own cycling shoes and pedals.

A CycleOps 400 Indoor Pro Cycle (Power Tap: Saris Cycling Group®. Madison, WI. USA) was used for all trials. Saddle height, saddle setback, handlebar reach and handlebar height were set to match the configuration of the participant's own bicycle as previously described (see **Appendix 5.A**) (Holliday et al., 2017).

TABLE 5.1: General characteristics of cyclists (n=17)

Variable	**Mean ± SD**
Age (years)	31 ± 9
Body mass (kg)	75.5 ± 7.5
Stature (cm)	178.4 ± 4.4
Trochanteric leg length (cm)	95.0 ± 2.7
Percentage body fat (%)	8.4 ± 2.8
Sum of seven skinfolds (mm)	57.6 ± 15.4
PPO (W)	354.0 ± 34.5
PPO (W/kg)	4.7 ± 0.4
VO_{2max} (ml/kg/min) Relative	55.2 ± 6.4

On the first visit to the laboratory the participant's anthropometrics were measured, followed by an incremental exercise test to volitional exhaustion. The CycleOps VirtualTraining app (VirtualTraining, version 1.7.3, Czech Republic) was used to control the ergometer and was set according to the participant's individual characteristics of age, mass and stature. Heart rate for all sessions was captured by a Suunto® T6C heart rate monitor (Suunto Oy, Vanata, Finland). The participant completed a Peak Power Output (PPO) and Peak Oxygen Consumption test to determine the required workload for the experimental trials. Gas analysis was monitored over 15 s intervals using an on-line breath-by-breath gas analyser and pneumotach (Oxycon, Viasis, Hoechberg, Germany). Participants started exercising at a workload of 100 W and resistance was increased by continuous ramp protocol at a rate of 20 W every 60 s until the participant was exhausted and could not sustain a cadence of at least 60 revolutions per minute (rpm). PPO was calculated by averaging the power output for the final minute of the VO_{2peak} test. VO_{2peak} was recorded as the highest VO_2 reading recorded for 30 s during the test.

On the second and third visit to the laboratory the researcher attached the EMG and 3D motion capture markers to the participant (Hermens et al., 1999; Vicon Motion Systems Limited, 2008). This was followed by a static calibration of the motion capture system before the participant was seated on the CycleOps ergometer.

Each participant performed a one hour steady state cycle at 60% of the peak power as measured during the PPO test. Rating of Perceived Exertion (RPE) and oxygen consumption were recorded at the end of each 20min interval. RPE was recorded using the Borg 6-20 RPE scale (Borg, 1982). The mask was fitted during measurements of oxygen consumption and removed during the remaining time intervals. Power output, heart rate, speed, cadence and distance were recorded continuously for later analysis. Participants were requested to maintain a cadence as close to 90rpm as possible throughout the trial. Participants were instructed to remain seated and not to alter their riding position during the trial, i.e. no standing whilst pedalling or changing the handgrip position. The riding position was standardised with the cyclist's hands on the brake hoods in order to avoid changes in metabolic cost due to modification of the trunk angle (Heil, Derrick, and Whittlesey, 1997). Participants were not informed when the 3D kinematic data was to be recorded, so as to prevent them from changing their pedalling action.

The participants repeated this procedure on a third visit to the laboratory one week later. All kinematic and EMG data was averaged over the testing sessions. By doing a repeat session, the reliability of the study's data was increased, suggesting that the hypothesised changes in kinematics may be reported with confidence and would help to reduce the risk of errors in the data interpretation (Lamberts et al., 2009; Hopkins, 2013).

Instrumentation and analysis

The hour-long steady state cycle was divided into thirds for analysis over time. The EMG and 3D motion capture were synchronised at the start of each data recording interval. Data were recorded simultaneously for 15 s three times during every third of the full hour. Specifically, during the first third at 10, 12 and 14min, during the second third at 30, 32 and 34min, and in the last third at 50, 52 and 54min. The 3D motion capture markers and EMG electrodes were placed by the primary investigator by measuring and recording the placement of each marker and electrode to increase the accuracy of placement in subsequent trials (Tsushima, Morris, and McGinley, 2003).

3D kinematics

An eight camera motion capture system (Oxford Metric Vicon, Oxford, UK) was used to capture kinematic data, and was recorded at a sampling rate of 250 Hz. The Vicon full body plug-in gait marker set allows for the measurement of all joint locations and angles of rotation as well as the calculation of joint moments. Plug-in gait is a biomechanical model based on the Newington-Helen Hayes gait model that calculates joint kinematics and kinetics from the XYZ marker positions and specific subject anthropometric measurements. The standard full marker set was modified by placing the tenth thoracic (T10) vertebra marker over the fifth thoracic (T5) vertebra instead. This was done to more closely approximate static methods used to measure shoulder flexion angle. All other joint angles and segments were defined as per the manual (*Plug-in Gait model details* 2008). Reflective markers were also placed on the pedal spindle and crank axis, to define a global coordinate system.

Analysis of the 3D kinematic data was performed using MATLAB (The Mathwork®, USA). The 3D kinematic data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12 Hz. Analysis of ten revolutions from each intensity stage was performed on the range of the ankle, knee, hip, shoulder and elbow joint angles, as well as the lumbar and thoracic spine. Ankle and knee angles for each trial were reported at bottom dead centre (BDC) pedal position, determined as the point at which the pedal reflective marker reached its minimal vertical position, i.e. 180°. Full knee extension is equal to 0°. Ankle neutral is equal to 90° (dorsiflexion $\leq$90°, plantarflexion >90°). The hip flexion angle for each trial was reported at top dead centre (TDC) pedal position, determined as the point at which the pedal reflective marker reached its maximal vertical position, i.e. 360°. Thoracic flexion was calculated relative to the global coordinate system, indicating a forward thoracic tilt or lean on the bicycle. Shoulder, elbow and spinal angles were taken as an average over the 360° cycle.

Electromyography

The EMG activity during the hour-long steady state cycle was recorded using an 8-channel EMG system (Telemyo 2400 G2, Noraxon, USA, Inc., Arizona, USA). Two electrodes (Blue Sensor, Medicotest, Denmark) were placed on the belly of the right Gluteus Maximus (GMax), VMO, VLO, Tibialis Anterior (TA), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Biceps Femoris (BF) muscles. Prior to placing the electrodes on the skin, the skin over the muscle was shaved and cleaned with ethanol. The placement and location of the electrodes were according to the recommendations by SENIAM (Surface EMG for Non-invasive Assessment of Muscles) (Hermens et al., 1999).

All EMG activity was sampled at 1984 Hz, thus providing raw data at a high enough frequency for reliable data collection and quantitative analysis. A 50 Hz notch filter was applied to filter out the power line noise. The signal was filtered using a 15-500 Hz band pass filter to allow movement artefact below 15 Hz and non-physiological signals above 500 Hz to be removed. The data were smoothed using root mean squared analysis (RMS), which was calculated for a 50ms window.

Ten revolutions from each data set were used for EMG analysis, which was performed using MATLAB (The Mathwork®, USA). The processed EMG data were further analysed into each quadrant of the cycle revolution, where quadrant 1 represents 0-90°, quadrant 2: 90-180°, quadrant 3: 180-270° and quadrant 4: 270-360°. The average of each quadrant, from each third of the hour-long cycle, was expressed as a percentage of the average obtained during ten full revolutions from the first third, i.e. data from each third of the trial were normalised against data from the first third and then expressed as quadrant activity which was a percentage of the average value for a full revolution.

For example, the average magnitude during the second third, in quadrant 3 was calculated as follows:

$$magnitude_{avg} = \frac{average\ of\ 10\ revolutions\ during\ \{2/3\}\ stage\ in\ quadrant\ \{3\}}{average\ of\ 10\ revolutions\ during\ 1/3\ stage\ over\ a\ full\ revolution}\%$$

Statistical methods

All joint kinematic and EMG magnitude data are expressed as means and standard deviation (mean $\pm$ SD). The data were statistically tested using a one-way ANOVA with repeated measures. When significant main effects were found, a Tukey test was used for post-hoc analysis. Significance was accepted when p-value <0.05. The statistical analyses were performed using GraphPad Prism v7.0a (GraphPad Software, San Diego, CA, USA).

RESULTS

There were no significant changes in kinematic variables during the hour-long steady state cycle (**Figure 5.1**). Mean values, SD's and p-values are displayed in **Table 5.2**.

There were no significant changes for muscle EMG signal amplitude for any of the seven measured muscle groups between any of the three time points in any of the quadrants (**Figure 5.2**). The p-values ranged from 0.07 (GMax, 0-90° quadrant) to 0.97 (TA, 0-90° quadrant). The mean values and standard deviations for each quadrant are shown in **Table 5.3**.

There was a significant increase in RPE from the first to the second and last third of the test, however no significant change from the second to the last third (**Table 5.2**). There were no significant changes in the VO_2 and V_E measurements for the hour-long steady state cycle. Although statistically significant, there were minor clinically insignificant changes in RER from the first to the second third, and from the first to the last third of the trial (**Table 5.2**).

TABLE 5.2: Mean ± standard deviation of each third of the hour-long steady state cycle and p-values for joint angles, rate of perceived exertion and metabolic gases.

	1/3	**2/3**	**3/3**	**p-value**
Ankle (BDC)	$97 \pm 5°$	$96 \pm 6°$	$96 \pm 6°$	0.23
Knee (BDC)	$37 \pm 7°$	$36 \pm 7°$	$36 \pm 7°$	0.16
Hip (TDC)	$122 \pm 5°$	$122 \pm 5°$	$122 \pm 5°$	0.75
Lumbar flexion	$48 \pm 12°$	$48 \pm 11°$	$48 \pm 12°$	0.86
Thoracic lean	$63 \pm 7°$	$63 \pm 6°$	$63 \pm 6°$	0.88
Shoulder	$104 \pm 7°$	$106 \pm 7°$	$106 \pm 7°$	0.10
Elbow	$34 \pm 8°$	$34 \pm 8°$	$34 \pm 8°$	0.29
RPE	$13 \pm 1^{*}$	$14 \pm 1^{*}$	14 ± 1	$<0.01^{*}$
VO_2 (ml/min)	3065.63 ± 340.81	3004.87 ± 355.36	3069.55 ± 337.13	0.28
RER	$0.90 \pm 0.04^{*\dagger}$	$0.89 \pm 0.04^{*}$	$0.88 \pm 0.04^{\dagger}$	$0.01^{*\dagger}$
V_E	73.84 ± 10.64	72.49 ± 9.26	74.38 ± 10.94	0.30

* significant change between first and second third of the hour-long steady state cycle
† significant change between first and last third of the hour-long steady state cycle
RPE = rate of perceived exertion. BDC = bottom dead centre. TDC = top dead centre.

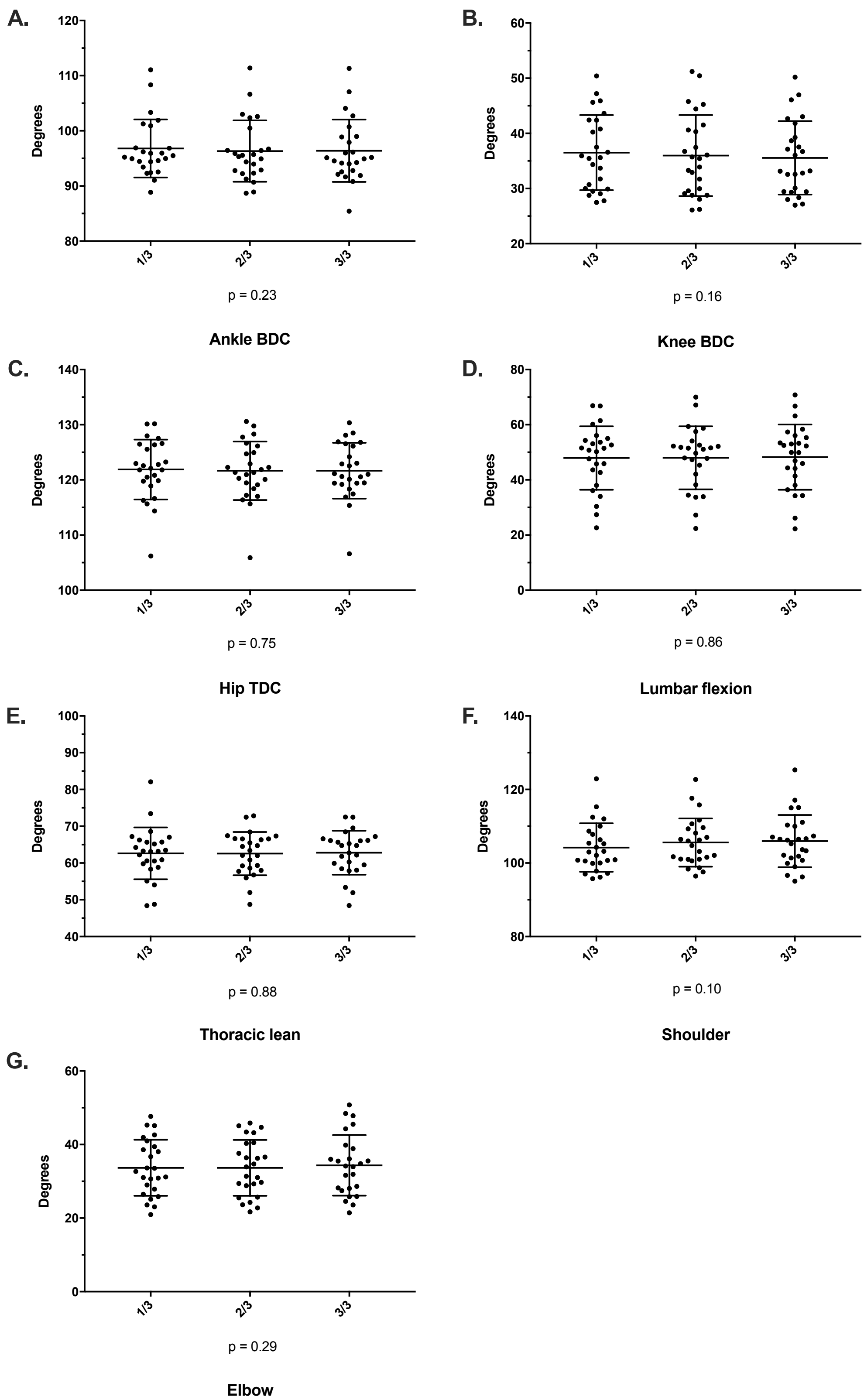

FIGURE 5.1: Joint angles over the hour-long steady state cycle with p-values.

TABLE 5.3: Mean ± standard deviation in percentages for each muscle in each quadrant, and p-value during each third of the steady state cycle.

Muscle	**Quadrant**	**1/3**	**2/3**	**3/3**	**p-value**
GMax	0-90°	227.64 ± 44.61	264.15 ± 68.28	266.39 ± 97.45	0.07
	90-180°	84.14 ± 18.86	101.18 ± 38.40	86.33 ± 48.60	0.12
	180-270°	44.74 ± 26.33	44.68 ± 29.21	42.06 ± 31.67	0.85
	270-360°	44.11 ± 20.64	50.29 ± 28.03	56.99 ± 30.34	0.12
VMO	0-90°	277.45 ± 22.87	283.77 ± 40.34	287.70 ± 54.83	0.37
	90-180°	44.82 ± 22.31	48.53 ± 21.65	47.68 ± 20.02	0.47
	180-270°	11.75 ± 5.32	12.34 ± 5.60	13.40 ± 7.48	0.19
	270-360°	65.97 ± 24.90	68.85 ± 26.45	71.77 ± 28.46	0.25
VLO	0-90°	273.22 ± 23.45	282.66 ± 39.40	284.88 ± 51.43	0.24
	90-180°	44.24 ± 29.35	47.95 ± 27.07	45.73 ± 19.63	0.55
	180-270°	9.88 ± 5.52	10.75 ± 6.62	12.30 ± 13.57	0.41
	270-360°	72.66 ± 22.54	75.67 ± 29.68	82.07 ± 34.06	0.11
TA	0-90°	79.44 ± 43.71	79.41 ± 444.17	80.09 ± 44.05	0.97
	90-180°	65.97 ± 35.46	68.18 ± 40.76	71.07 ± 39.91	0.78
	180-270°	60.98 ± 29.84	57.67 ± 30.89	67.64 ± 37.72	0.40
	270-360°	185.06 ± 62.54	184.37 ± 61.07	189.81 ± 62.89	0.80
RF	0-90°	162.23 ± 51.63	183.67 ± 54.81	177.90 ± 61.94	0.07
	90-180°	46.39 ± 28.11	49.38 ± 25.05	50.52 ± 27.55	0.49
	180-270°	54.08 ± 23.34	47.47 ± 21.47	52.74 ± 25.77	0.15
	270-360°	131.70 ± 48.95	130.91 ± 51.97	136.04 ± 63.51	0.79
MG	0-90°	66.72 ± 23.06	72.35 ± 29.39	70.43 ± 30.72	0.25
	90-180°	254.77 ± 26.45	242.03 ± 29.93	240.52 ± 33.88	0.06
	180-270°	64.50 ± 30.21	71.05 ± 37.06	71.27 ± 40.35	0.40
	270-360°	14.15 ± 9.63	13.54 ± 8.32	14.78 ± 9.38	0.61
BF	0-90°	122.23 ± 36.28	130.27 ± 45.91	123.05 ± 50.76	0.51
	90-180°	198.81 ± 36.04	206 ± 59.23	192.05 ± 68.78	0.47
	180-270°	48.39 ± 29.00	53.45 ± 32.71	48.25 ± 37.40	0.46
	270-360°	27.13 ± 14.12	29.54 ± 17.79	33.63 ± 19.73	0.11

Gluteus Maximus (GMax), Vastus Medialis Oblique (VMO), Vastus Lateralis Oblique (VLO), Tibialis Anterior (TA), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Biceps Femoris (BF) muscles.

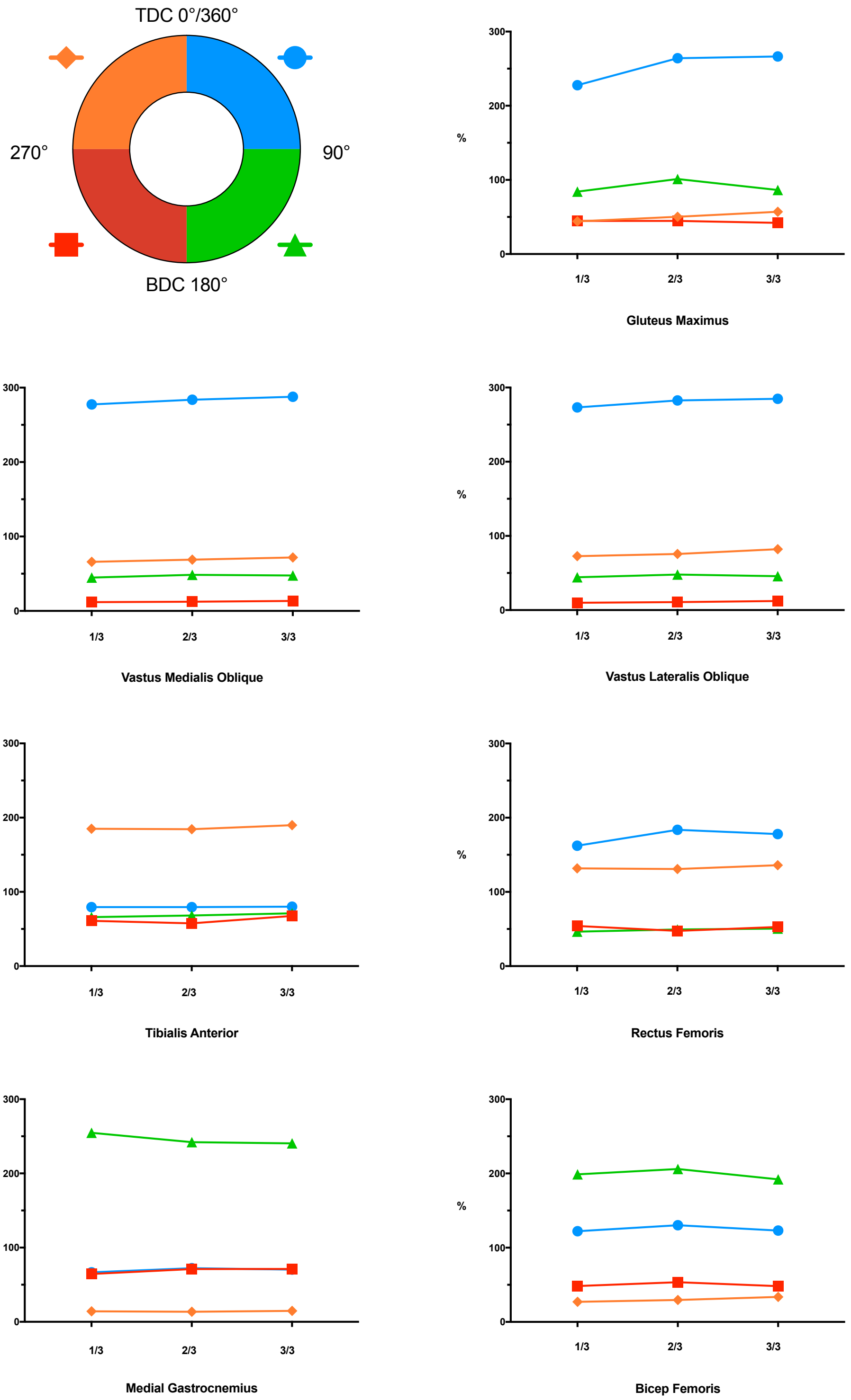

FIGURE 5.2: EMG magnitudes over the hour-long steady state cycle, in each quadrant.

DISCUSSION

Despite performing at a moderately high intensity (60% of VO_{2max}) for a full hour duration we did not find any significant changes in the kinematics of any of the seven joints we measured. Nor did we demonstrate any significant changes in the muscle magnitudes during this hour-long cycle. These are similar to the findings by Sayers, Tweddle, Every, and Wiegand (2012), who investigated the changes in the drive phase of the lower limb during a 60 minute stochastic time trial. In their study only the kinematics of the lower limb were recorded during the steady state phases of the 60min trial and are therefore somewhat comparable. There was a high level of consistency in the sagittal plane movements, with only the ankle and hip demonstrating significant changes. These changes occurred in ankle dorsiflexion and hip extension and, although significant, were less than 3°. This is within the normal typical error of measurement values found for 3D motion capture (Holliday et al., 2017). It has been suggested that the ankle moves into dorsiflexion to increase stability around the ankle joint in order to transfer force effectively to the pedals to maintain the power output at higher workloads (Bini and Diefenthaeler, 2010). Changes to ankle mean angles into dorsiflexion have been demonstrated with increased intensity (Bini and Diefenthaeler, 2010; Bini, Senger, et al., 2012; Peveler, Shew, et al., 2012). As Sayers et al. (2012) utilised a stochastic protocol with a higher mean intensity and 90 s intervals at 140% of onset of blood lactate accumulation (OBLA) power they may have induced greater fatigue than our protocol and thus induced these changes into ankle dorsiflexion. Prolonged cycling at higher workloads and stochastic high intensity cycling may therefore induce compensatory changes which were not demonstrated in our study.

Previous research has suggested that the optimal static configuration of the knee flexion angle should be within a range of 25-35° (Peveler, Bishop, et al., 2005). Our findings confirm those of other studies that the knee joint kinematics do not alter during the steady state ride and suggests that bike fitters can use static kinematics to configure the cyclist's bicycles as previously recommended (Peveler, Pounders, and Bishop, 2007), or by dynamic measures, knowing that the knee will remain within this optimal range during steady state training rides. Similarly, recommendations that may be developed in time for hip, shoulder or other joints can also be reliably applied without concern that these would change during steady state exercise bouts.

Our second finding was that there was no statistically significant change in EMG magnitudes over the hour-long steady state ride for all seven muscle groups of the lower limb measured. Ours is the first study to measure multiple muscle groups during prolonged cycling. One previous study (Lepers et al., 2002), concluded that the quadriceps muscle only demonstrated signs of altered contractile properties beyond the first hour of steady state riding. However this study only measured activity in the VMO and VLO muscles and demonstrated a reduced EMG signal in these two muscle groups as well as changes in M-wave properties and contractility. Another study (Kay et al., 2001), has also demonstrated reduced EMG activity in the RF muscle with the onset of fatigue. Both of these studies suggest that there is reduced activity in the quadriceps with prolonged fatiguing cycling exercise and that, in a compensatory manner, other muscle groups such as GMax may contribute to knee extensor force during the drive phase of pedalling (Kay et al., 2001; Lepers et al., 2002). However, this has not been demonstrated to date. We did see an increase of 17% in GMax activity in the first quadrant of the drive phase between the first and third periods of our trial. However, these changes did not reach significance (p-value = 0.07) (**Table 5.3**). Perhaps if the cyclists in our study had continued to exercise for a longer duration, they may have begun to demonstrate changes in the activation patterns as a result of accumulation of muscle fatigue.

Changes in the RPE increased from the first to the second third of the trial which is consistent with an increase in afferent activity mediated by peripheral fatigue processes and consistent with previous studies of prolonged exercise at fixed workloads (Swart et al., 2009; Eston, 2012). During the final third the RPE did not increase further. This is in keeping with known duration trials where the afferent feedback and hence RPE ratio is attenuated as the endpoint approaches (Swart et al., 2009; St Clair Gibson, Swart, and

Tucker, 2018).

CONCLUSION

During configuration of a cyclist's position on the bicycle it is important for bike fitters to consider the effects of training intensity and duration on kinematics and muscle activity. From a practical perspective, steady state training conducted at a moderately high intensity does not appear to have any meaningful effect on cyclists' kinematic variables or muscle magnitudes. Higher intensities and more prolonged exercise durations may well have an effect (Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010; Bini, Senger, et al., 2012) and these should be investigated with further studies using similar methods to our study to assess all important joints and muscle groups and not simply limited to lower limb kinematics or a few muscle groups.

Bike fitting for steady state endurance rides can be performed as per previous research recommendations to prevent injury and to maximise performance (Peveler, Pounders, and Bishop, 2007).

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Appendix

5.A Bicycle configuration measurements

Saddle height:

The saddle height was measured from the centre of the crank axle to the top of the saddle, passing through the centre of the bicycle seat tube and seat post.

Saddle setback:

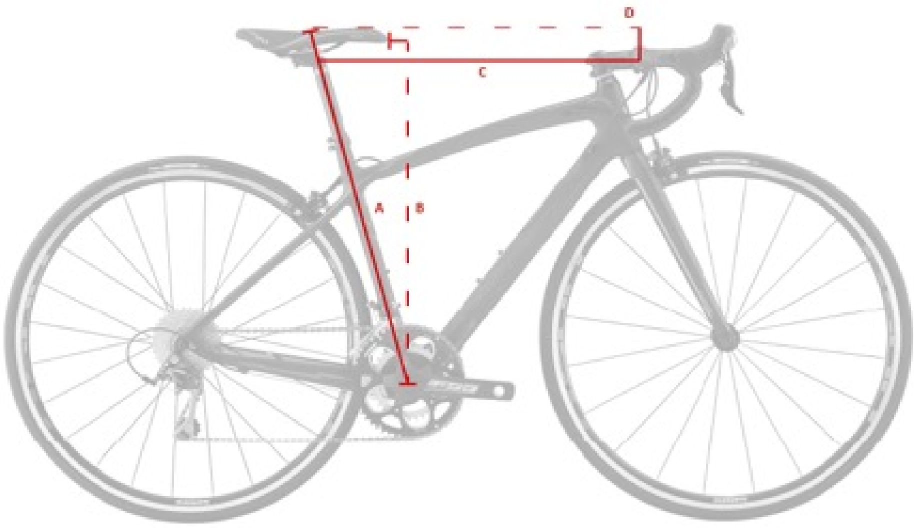
Saddle setback was measured as the horizontal distance from the front of the saddle to the centre of the crank axle.

Handlebar reach:

The handlebar reach was measured horizontally, from the centre of the handlebar clamping point to the centre of the seat post or seat tube. This measurement is for bicycle frames with a 74° seat tube angle.

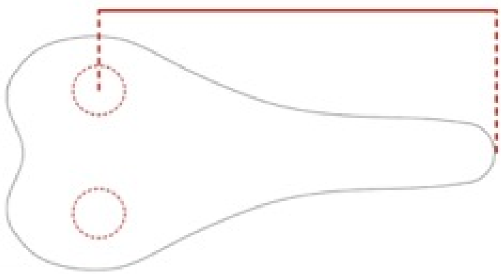
Handlebar drop:

Handlebar drop values were measured as the vertical distance from the top of the saddle surface to the centre of the handlebar clamping joint.



Saddle length:

The saddle was measured (a standard saddle is 22.5cm in length) from the centre of the ischial tuberosity padding to the front of the saddle.



Chapter 6

Cycling: Joint kinematics and muscle activity during differing intensities

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ABSTRACT

Full body kinematics and electromyographic (EMG) patterns may alter based on the workloads that are encountered during cycling. Understanding the effect of differing intensities on the cyclist can guide clinicians and bike fitters in improving specific muscle recruitment and cycling posture to optimise training and racing. We aimed to assess the changes in lower limb EMG magnitudes and full body 3D kinematics of seventeen well-trained cyclists at three different exercise intensities: 60, 80 and 90% of maximum heart rate. Significant results were demonstrated for all the joints except the hip and shoulder. Cyclists' ankle dorsiflexion and knee extension increased between 6-9% with higher intensities. The elbow adopted a significantly more flexed position, increasing flexion by 39% from 60% to 90% intensity, whilst the lumbar and thoracic flexion increased by 7% at the higher intensity. There were significant increases in EMG signal amplitude at higher intensities for all muscle groups measured. These results will guide clinicians in strengthening specific muscles at specific ranges of the cycling pedal revolution. Guidelines for optimal bicycle configuration should take into account the full body position of the cyclist as well as the training and racing intensity when assessing kinematics.

Keywords: *Biomechanics, electromyography, bike fitting*

INTRODUCTION

Optimal static bicycle configuration has been the topic of numerous studies (Peveler, Bishop, et al., 2005; Peveler, Pounders, and Bishop, 2007; Peveler, 2008; Bini, Hume, and Croft, 2011). The freely chosen bicycle configuration and subsequent cyclist kinematics, muscle activity and physiological responses can be influenced by adjusting any of the contact points on the bicycle (Burt, 2014). Previous research on the correct positioning of the handlebars, pedal crank arm length and saddle fore-aft position is based on personal perspectives and comfort (De Vey Mestdagh, 1998; Silberman et al., 2005; Burt, 2014), whereas static saddle height recommendations have been based on scientific methods (Bini, Hume, and Croft, 2011). Currently there are three main methods used in clinical practice to set the saddle height: anthropometrics (inseam length and trochanteric leg length), static knee flexion angle methods and dynamic methods (during pedalling). The recommended static method is the Holmes method (Peveler, Bishop, et al., 2005). The cyclist is in a stationary seated position with the crank arm in the lowest or 6 o'clock position and the pedal surface in a horizontal orientation. Knee flexion angle (KFA) is measured with a goniometer and recommended to be in a range between 25° and 35° (where full knee extension is equal to 0°). It has been demonstrated that setting the saddle at this KFA range statically is optimal for injury prevention and performance (Peveler, 2008).

More recently it has been recommended that bike fitting be conducted in a dynamic functional manner, as kinematics can be influenced by cycling workload (Ferrer-Roca et al., 2012; Peveler, Shew, et al., 2012). With the advancement of technology we are now able to record the cyclist's position in full three dimensional motion capture, however, as yet there are limited scientific recommendations for optimal joint ranges for dynamic bicycle configuration. Static recommendations for optimal bicycle configuration cannot be transferred to dynamic methods as the difference between static and dynamic lower limb angles has been highlighted (Peveler, Shew, et al., 2012; Fonda, Sarabon, and F. Li, 2014; Bini, Dagnese, et al., 2016; Holliday et al., 2017). The range of knee flexion recommended during static assessment using the Holmes method (25-35°) increases by ∼5-8° (to approximately 30-40° KFA) depending on the study and the relative workload intensity (Farrell, Reisinger, and Tillman, 2003; Peveler, Shew, et al., 2012; Fonda, Sarabon, and F. Li, 2014). Increased knee and hip extension were demonstrated at maximal workloads (Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010) and could be linked to a shift in forward position on the bicycle (Bini, Senger, et al., 2012). This forward position on the bicycle was demonstrated during sustained high intensity cycling (Sayers and Tweddle, 2012). Increased sagittal plane thoracic angle (i.e. the thoracic segment moving anteriorly relative to the crank arm) occurred towards the end of the protocol and was suggested to be linked to cyclists shifting their body forwards as they fatigued, enabling more weight to be exerted onto the pedals. This was further confirmed where a greater trunk lean angle was demonstrated during a fatiguing protocol, with all participants also displaying an increase into dorsiflexion at the ankle joint (Dingwell et al., 2008). There was a positive association, such that an increase in EMG median frequency signal preceded an increase in movement kinematics. These suboptimal positional changes with fatigue may lead to maladaptive joint loading and thus may result in an increased risk of repetitive strain or long term injuries. It was concluded that as fatigue occurs, cyclists changed their body position and muscle activation patterns to maintain performance (Dingwell et al., 2008).

The typical muscle activation pattern displayed during cycling has been studied in more depth due to the recent advances in technology (Hug and Dorel, 2009). Likewise, numerous studies have investigated muscle recruitment patterns in the final stages of exhaustion during cycling, and it is known that EMG patterns change with the onset of fatigue (So, J. Ng, and G. Ng, 2005). Lower limb muscle coordination during an all-out sprint cycling task displayed a significant change between the submaximal and maximal cycling exercises (Dorel et al., 2012). The increase in duration of all muscle activity during the sprint is suggestive of a strategy to enhance the work generated by each of the muscle groups. During the all-out sprint, there was a large increase in hip flexor activity, a lesser extent to the knee flexor activity, whereas the plantar flexors and knee extensors displayed an even smaller increase. It is possible that alternative muscles are recruited as fatigue accumulates in working muscles, as demonstrated by a decrease in Rectus Femoris EMG activity during all-out cycling sprints (Kay et al., 2001).

These studies were investigated at maximal power or to exhaustion and it is known that the body position on the bicycle and the muscle recruitment patterns are altered compared to riding at low intensities. Knowledge of how the muscles adapt to differing intensities, in conjunction with the position the cyclist is in, would help clinicians and bike fitters to strengthen those muscles in that range, at that cycling intensity. Racing at a workload of 55-60% VO_{2max} has been suggested as a strategic way to maximise power output while minimising the risk of early fatigue (Blake, Champoux, and Wakeling, 2012).

In order to maximise the use of muscle coordination patterns learned during training, it has been recommended that the cyclist train in similar conditions that they race in (Blake, Champoux, and Wakeling, 2012). The research published to date explores the adaptations of the lower limb kinematics and muscle activity with maximal effort or fatigue. The cyclist will however spend only a fraction of the race or training at absolute fatigue and/or maximal effort greater than 90% heart rate intensity, with the majority of the ride shifting between 60-80% heart rate intensity (Palmer et al., 1994; Padilla et al., 2001).

It is beneficial for clinicians and bike fitters to understand how the full body kinematics and lower limb muscles are affected by differing intensities encountered in cycling training and racing, not only with fatigue or maximal efforts. The only study to date that has assessed the relationship between workload intensity and 3D kinematics, demonstrated a small to moderate difference in lateral spine inclination and spine rotation between recreational and competitive cyclists (Bini, Dagnese, et al., 2016). Currently there are no studies investigating full body 3D kinematics simultaneously as well as lower limb muscle activity at differing intensities. The aim of this study was therefore to assess how the full body kinematics and specific muscle magnitude is affected by different intensities that are encountered in cycling. Furthermore, we aim to guide clinicians and bike fitters with recommendations for which joints or body segments to focus on during dynamic bike fitting and how cycling intensity during a bike fit may be of importance.

It was hypothesised that the upper body would adopt a more flexed position with intensity, whilst the ankle would move into a more dorsiflexed position and the knee into a more extended position, and that individual muscle activity would also increase proportionally. Furthermore, we hypothesise that the spinal segments and upper limb joints will demonstrate significant changes and that cycling intensity will have an impact on the bike fitting process.

METHODS

Participants

Seventeen well-trained male road cyclists (31 ± 9 years, 75.5 ± 7.5kg, 178.4 ± 4.4cm) conforming to Level 2 or greater (De Pauw et al., 2013) were recruited for this study. Level 2 is described as having a relative VO_{2max} between 45 - 54.9ml/kg/min, and a relative Peak Power Output (PPO) between 3.6 and 4.5W/kg. The general characteristics and performance parameters of the 17 cyclists are shown in **Table 6.1**. Prior to testing, each participant was informed of the risks and stresses associated with participation in the research trial, were personally interviewed about their training history, completed a Physical Activity Readiness Questionnaire (PAR-Q) (Whaley, Brubaker, and Otto, 2007) and signed an informed consent form. The study was approved by the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Cape Town, and conformed to the principles of the World Medical Association Declaration of Helsinki (World Medical Association, 2013).

Testing procedure

The participants reported to the laboratory on three separate occasions (one week apart, over three weeks) with their own cycling shoes and pedals. A CycleOps 400 Indoor Pro

TABLE 6.1: General characteristics of cyclists (n=17)

Variable	**Mean $\pm$ SD**
Age (years)	31 $\pm$ 9
Body mass (kg)	75.5 $\pm$ 7.5
Stature (cm)	178.4 $\pm$ 4.4
Trochanteric leg length (cm)	95.0 $\pm$ 2.7
Percentage body fat (%)	8.4 $\pm$ 2.8
Sum of seven skinfolds (mm)	57.6 $\pm$ 15.4
PPO (W)	354.0 $\pm$ 34.5
PPO (W/kg)	4.7 $\pm$ 0.4
VO_{2max} (ml/kg/min) Relative	55.2 $\pm$ 6.4

Cycle (Power Tap: Saris Cycling Group®. Madison, WI. USA) was used for all trials. Saddle height, saddle setback, handlebar reach and handlebar height were set to match the configuration of the participant's own bicycle as previously described (Holliday et al., 2017).

On the first visit to the laboratory the participant's anthropometric measurements were taken, followed by an incremental exercise test to volitional exhaustion. The CycleOps VirtualTraining app (VirtualTraining, version 1.7.3, Czech Republic) was used to control the ergometer and was set according to the participant's individual characteristics of age, mass and stature. The participant completed a PPO and Peak Oxygen Consumption test to determine the required workload for the experimental trials. Gas analysis was monitored over 15 s intervals using an on-line breath-by-breath gas analyser and pneumotach (Oxycon, Viasis, Hoechberg, Germany). Participants started exercising at a workload of 100 W and resistance was increased by continuous ramp protocol at a rate of 20 W every 60 s until the participant was exhausted and could not sustain a cadence of at least 60 revolutions per minute (rpm). PPO was calculated by averaging the power output for the final minute of the VO_{2peak} test. VO_{2peak} was recorded as the highest VO_2 reading recorded for 30 s during the test. The maximum heart rate (MHR) of each participant was calculated during the PPO test, and was used to calculate the target heart rates for the intensity protocol.

On the second and third visit to the laboratory the researcher attached the EMG and 3D motion capture markers to the participant (Hermens et al., 1999; *Plug-in Gait model details* 2008). This was followed by a static calibration of the motion capture system before the participant was seated on the CycleOps ergometer.

Each participant performed a fifteen minute exercise protocol at three different workload intensities based on the Lamberts Submaximal Cycle Test (Lamberts et al., 2009), which was previously demonstrated to be highly reliable, with an ICC of R = 0.96 and typical error of measurement (TEM) less than 2 beats per minute (bpm). The first stage of the protocol involved cycling for six minutes at 60% MHR, followed immediately by six minutes at 80% MHR and a further three minutes at 90% MHR. Cyclists have to elicit and maintain their heart rate with resistance increased or decreased to avoid their heart rate deviating by more than 2bmp. Participants were requested to maintain a cadence as close to 90rpm as possible throughout the trial. Participants were instructed to remain seated and not to alter their riding position during the trial, i.e. no standing whilst pedalling or changing the handgrip position. The riding position was standardised with the cyclist's hands on the brake hoods in order to avoid changes in metabolic cost due to modification of the trunk angle (Heil, Derrick, and Whittlesey, 1997).

Rating of Perceived Exertion (RPE) was recorded at the end of each intensity period using the Borg 6-20 RPE scale (Borg, 1982). Power output, heart rate, speed, cadence and distance were recorded continuously for later analysis.

The participants repeated this procedure on a third visit to the laboratory one week later. By doing a repeat session, the reliability of the study's data was increased, suggesting that the hypothesised changes in kinematics may be reported with confidence and would also help to reduce the risk of errors in the data interpretation (Lamberts et al., 2009; Hopkins, 2013).

Instrumentation and analysis

Three dimensional motion capture and EMG were recorded simultaneously during the second and third testing sessions. Data were recorded for 15 s during the second minute of each intensity interval, specifically at 2, 8 and 13min. The 3D motion capture markers and EMG electrodes were placed by the primary investigator by measuring and recording the placement of each marker and electrode to increase the accuracy of placement in subsequent trials (Tsushima, Morris, and McGinley, 2003).

3D kinematics

An eight camera motion capture system (Oxford Metric Vicon, Oxford, UK) was used to capture kinematic data, and was recorded at a sampling rate of 250 Hz. The Vicon full body plug-in gait marker set allows for the measurement of all joint locations and angles of rotation as well as the calculation of joint moments. Plug-in gait is a biomechanical model based on the Newington-Helen Hayes gait model that calculates joint kinematics and kinetics from the XYZ marker positions and specific subject anthropometric measurements. The standard full marker set was modified by placing the tenth thoracic (T10) vertebra marker over the fifth thoracic (T5) vertebra instead. This was done to more closely approximate static methods used to measure shoulder flexion angle. All other joint angles and segments were defined as per the manual (*Plug-in Gait model details* 2008). Reflective markers were also placed on the pedal spindle and crank axis, to define a global coordinate system.

Analysis of the 3D kinematic data was performed using MATLAB (The Mathwork®, USA). The 3D kinematic data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12 Hz. Analysis of ten revolutions from each intensity stage was performed on the range of the ankle, knee, hip, shoulder and elbow joint angles, as well as the lumbar and thoracic spine. Ankle and knee angles for each trial were reported at bottom dead centre (BDC) pedal position, determined as the point at which the pedal reflective marker reached its minimal vertical position, i.e. 180°. Full knee extension is equal to 0°. Ankle neutral is equal to 90° (dorsiflexion $\leq$90°, plantarflexion >90°). The hip flexion angle for each trial was reported at top dead centre (TDC) pedal position, determined as the point at which the pedal reflective marker reached its maximal vertical position, i.e. 360°. Thoracic flexion was calculated relative to the global coordinate system, indicating a forward thoracic tilt or lean on the bicycle. Shoulder, elbow and spinal angles were taken as an average over the 360° cycle.

Electromyography

The EMG activity during the testing sessions was recorded using an 8-channel EMG system (Telemyo 2400 G2, Noraxon, USA, Inc., Arizona, USA). Two electrodes (Blue Sensor, Medicotest, Denmark) were placed on the belly of the right Gluteus Maximus (GMax), Vastus Medialis Oblique (VMO), Vastus Lateralis Oblique (VLO), Tibialis Anterior (TA), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Biceps Femoris (BF) muscles. Prior to placing the electrodes on the skin, the skin over the muscle was shaved and cleaned with ethanol. The placement and location of the electrodes were according to the recommendations by SENIAM (Surface EMG for Non-invasive Assessment of Muscles)(Hermens et al., 1999).

All EMG activity was sampled at 1984 Hz, thus providing raw data at a high enough frequency for reliable data collection and quantitative data analyses. A 50 Hz notch filter was applied to filter out the power line noise. The signal was filtered using a 15-500 Hz band pass filter to allow movement artefact below 15 Hz and non-physiological signals above 500 Hz to be removed. The data were smoothed using root mean squared analysis (RMS), which was calculated for a 50ms window. Ten revolutions from each data set were used for EMG analysis, which was performed using MATLAB (The Mathwork®, USA). The processed EMG data were further analysed into each quadrant of the cycle revolution, where quadrant 1 represents 0-90°, quadrant 2: 90-180°, quadrant 3: 180-270° and quadrant 4: 270-360°. The average magnitude from each intensity level, from each quadrant, was expressed as a percentage of the average magnitude obtained during ten full revolutions from the first intensity level.

For example, the average magnitude during the 80% intensity stage, in quadrant 3 was calculated as follows:

$$magnitude_{avg} = \frac{average\ of\ 10\ revolutions\ during\ \{80\%\}\ stage\ in\ quadrant\ \{3\}}{average\ of\ 10\ revolutions\ during\ 60\%\ stage\ over\ a\ full\ revolution}\%$$

Statistical methods

All joint kinematic and EMG magnitude data are expressed as means and standard deviation (mean $\pm$ SD). The data were statistically tested using a one-way ANOVA with repeated measures. When significant main effects were found, a Tukey test was used for post-hoc analysis. Significance was accepted when p-value <0.05. The statistical analyses were performed using GraphPad Prism v7.0a (GraphPad Software, San Diego, CA, USA).

RESULTS

The mean $\pm$ SD and p-values for all the joint kinematics can be found in **Table 6.2** and **Figure 6.1**. There was a significant change in all joints across all intensities, except for the hip and the shoulder joint. The ankle joint progressively moved into dorsiflexion with the increased intensity with a decrease in mean from 100 $\pm$ 5° at 60%, 97 $\pm$ 5° at 80% and 94 $\pm$ 6° at 90%, (F(1.215, 27.95) = 26.79). The knee flexion decreased progressively with an increase in intensity, with a decrease in mean from 37 $\pm$ 7° at 60%, 35 $\pm$ 6° at 80% and 34 $\pm$ 6° at 90%, (F(1.75, 40.19) = 17.45). The spinal flexion increased with an increase in intensity, with an increase from 45 $\pm$ 9° at 60%, 47 $\pm$ 11° at 80% and 48 $\pm$ 11° at 90%, (F(1.68, 36.94) = 17.80). The thoracic angle increased with an increase in intensity, with an increase in mean from 60 $\pm$ 5° at 60%, 62 $\pm$ 5° at 80% and 64 $\pm$ 5° at 90%, (F(1.37, 30.16) = 21.59). Elbow flexion increased progressively with increased intensity with an increase in mean from 31 $\pm$ 5° at 60%, 36 $\pm$ 5° at 80% and 43 $\pm$ 10° at 90%, ($F(1.23, 29.45) = 35.50$).

The mean $\pm$ SD and p-values for all muscle EMG magnitudes can be found in **Table 6.3** and **Figure 6.2**. There were significant changes in all muscle groups with increasing intensity. The change was most visible in the quadrant that the muscle has been shown to be most active in, and between 60% to 80% and 60% to 90% intensity. E.g. VLO activity increased mostly in quadrant 1 which is the period during which knee extension is used to generate pedalling power.

The RPE increased progressively and linearly in keeping with the increased intensity from a score of 9 $\pm$ 1 for 60%, 13 $\pm$ 2 for 80% and 16 $\pm$ 2 for 90% intensity (**Table 6.2**).

TABLE 6.2: Mean ± standard deviation and p-values for joint kinematics at different intensities, rate of perceived exertion, and average heart rate, cadence, speed and power.

	60%	**80%**	**90%**	**p-value**
Ankle (BDC)	100 ± 5°	97 ± 5°	94 ± 6°	<0.001$^{*\dagger\pm}$
Knee (BDC)	37 ± 7°	35 ± 6°	34 ± 6°	<0.001*†
Hip (TDC)	122 ± 6°	122 ± 6°	122 ±6°	0.86
Lumbar flexion	45 ± 9°	47 ± 11°	48 ± 11°	<0.001$^{*\dagger\pm}$
Thoracic lean	60 ± 5°	62 ± 5°	64 ± 5°	<0.001$^{*\dagger\pm}$
Shoulder	103 ± 9°	104 ± 10°	104 ± 8°	0.82
Elbow	31 ± 5°	36 ± 5°	43 ± 10°	<0.001$^{*\dagger\pm}$
RPE	9 ± 1	13 ± 2	16 ± 2	<0.001$^{*\dagger\pm}$
Average HR (bpm)	109 ± 6	144 ± 6	164 ± 7	<0.001$^{*\dagger\pm}$
Average cadence (rpm)	88 ± 4	92 ± 2	92 ± 4	0.40
Average speed (km/hr)	36 ± 3	38 ± 4	35 ± 4	0.002$^{*\pm}$
Average power (W)	133 ± 17	240 ± 35	303 ± 45	<0.001$^{*\dagger\pm}$

* significant change between 60% and 80% maximum heart rate (MHR)
† significant change between 60% and 90% MHR
$^{\pm}$ significant change between 80% and 90% MHR
RPE = rate of perceived exertion. BDC = bottom dead centre. TDC = top dead centre.

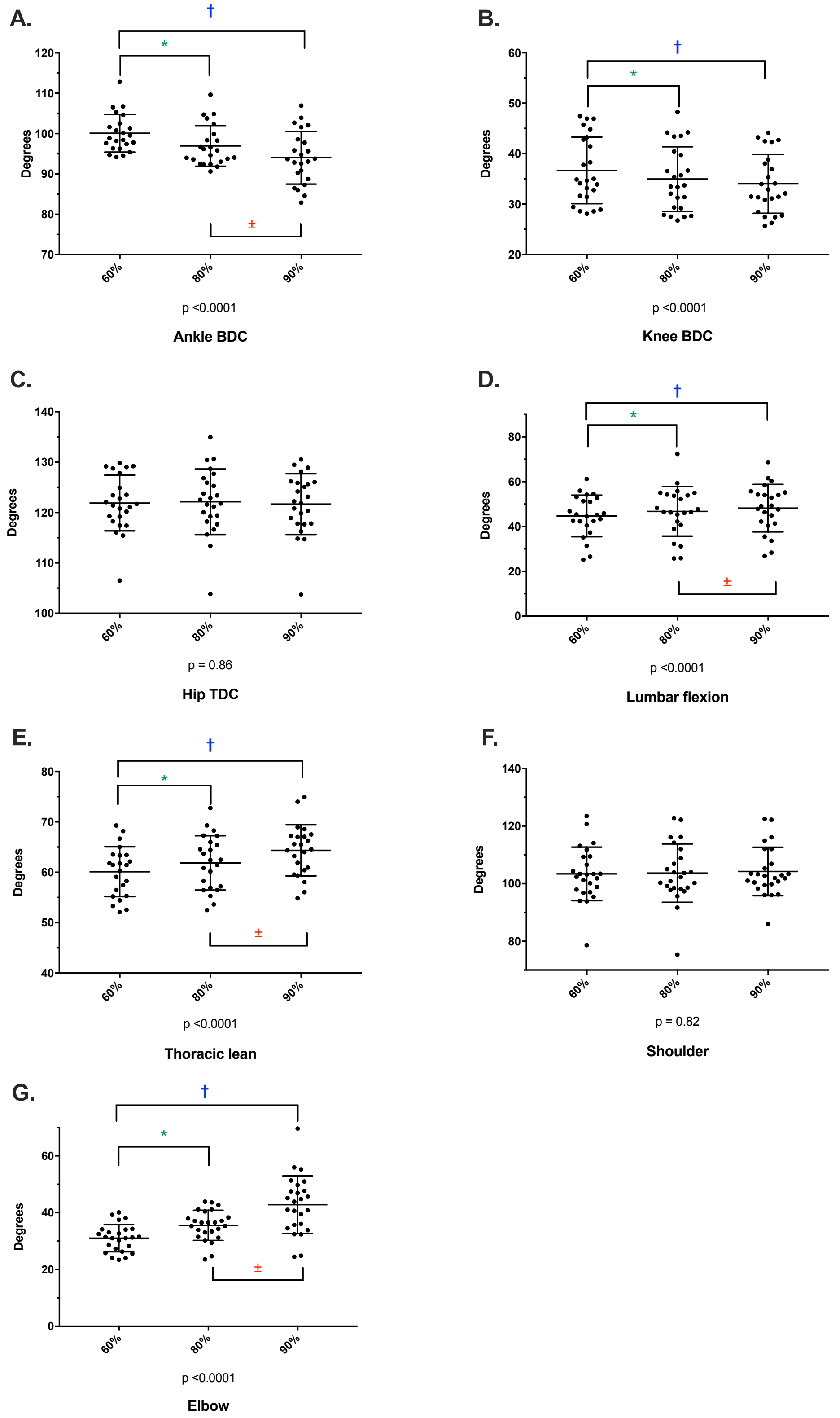

FIGURE 6.1: Joint angles over the different intensities with p-values.
*significant difference between 60% and 80% maximum heart rate (MHR)
±significant difference between 80% and 90% MHR
†significant difference between 60% and 90% MHR.

TABLE 6.3: Mean ± standard deviation in percentages for each muscle in each quadrant, and p-value during each different intensity.

Muscle	Quadrant	60%	80%	90%	p-value
GMax	0-90°	232.60 ± 54.51	486.44 ± 165.64	624.37 ± 289.36	<0.001$^{*\dagger\pm}$
	90-180°	60.60 ± 24.34	138.42 ± 67.38	205.08 ± 93.85	<0.001$^{*\dagger\pm}$
	180-270°	40.31 ± 28.79	50.23 ± 36.69	59.17 ± 32.91	<0.001†
	270-360°	57.63 ± 22.93	68.45 ± 35.53	82.22 ± 52.82	0.01$^{\dagger\pm}$
VMO	0-90°	281.55 ± 26.97	387.97 ± 84.33	430.42 ± 114.39	<0.001$^{*\dagger\pm}$
	90-180°	37.16 ± 18.74	47.12 ± 23.55	54.00 ± 24.50	0.001*†
	180-270°	9.92 ± 5.62	15.48 ± 9.52	18.99 ± 12.07	<0.001$^{*\dagger\pm}$
	270-360°	74.30 ± 25.20	104.06 ± 41.75	120.46 ± 69.23	0.01*†
VLO	0-90°	280.88 ± 26.29	396.08 ± 90.17	471.79 ± 130.03	<0.001$^{*\dagger\pm}$
	90-180°	32.92 ± 16.27	43.86 ± 22.36	61.27 ± 47.42	0.01*†
	180-270°	8.61 ± 3.06	11.86 ± 4.81	13.14 ± 5.02	<0.001*†
	270-360°	79.80 ± 21.61	112.33 ± 47.44	140.76 ± 71.31	0.001*†
TA	0-90°	61.85 ± 30.47	92.68 ± 59.60	106.75 ± 66.44	<0.001$^{*\dagger\pm}$
	90-180°	68.07 ± 39.44	87.41 ± 51.46	92.95 ± 54.12	0.05
	180-270°	62.95 ± 32.32	78.30 ± 35.82	103.16 ± 63.33	0.03
	270-360°	199.37 ± 52.00	267.89 ± 94.84	272.06 ± 132.52	0.02*
RF	0-90°	146.17 ± 54.65	228.44 ± 142.76	296.10 ± 181.83	<0.001$^{*\dagger\pm}$
	90-180°	31.78 ± 16.95	37.76 ± 19.14	52.86 ± 26.60	<0.001$^{*\dagger\pm}$
	180-270°	64.42 ± 25.55	63.37 ± 29.74	68.19 ± 35.12	0.71
	270-360°	156.95 ± 53.29	214.05 ± 89.06	244.66 ± 118.12	0.003*†
MG	0-90°	77.12 ± 34.67	81.49 ± 34.07	85.20 ± 41.59	0.44
	90-180°	259.81 ± 27.19	269.59 ± 43.28	259.77 ± 45.78	0.31
	180-270°	52.93 ± 28.90	66.54 ± 33.32	75.24 ± 40.37	0.01†
	270-360°	10.73 ± 4.44	12.71 ± 5.64	13.07 ± 5.45	0.02†
BF	0-90°	121.07 ± 48.40	210.98 ± 73.27	293.15 ± 129.19	<0.001$^{*\dagger\pm}$
	90-180°	206.17 ± 57.91	325.20 v83.63	440.58 ± 145.11	<0.001$^{*\dagger\pm}$
	180-270°	45.14 ± 29.46	87.53 ± 62.95	129.06 ± 75.81	<0.001$^{*\dagger\pm}$
	270-360°	27.18 ± 7.40	43.93 ± 19.77	48.13 ± 23.06	<0.001*†

* significant change between 60% and 80% maximum heart rate (MHR)
† significant change between 60% and 90% MHR
$^{\pm}$ significant change between 80% and 90% MHR
Gluteus Maximus (GMax), Vastus Medialis Oblique (VMO), Vastus Lateralis Oblique (VLO), Tibialis Anterior (TA), Rectus Femoris (RF), Medial Gastrocnemius (MG) and Biceps Femoris (BF) muscles.

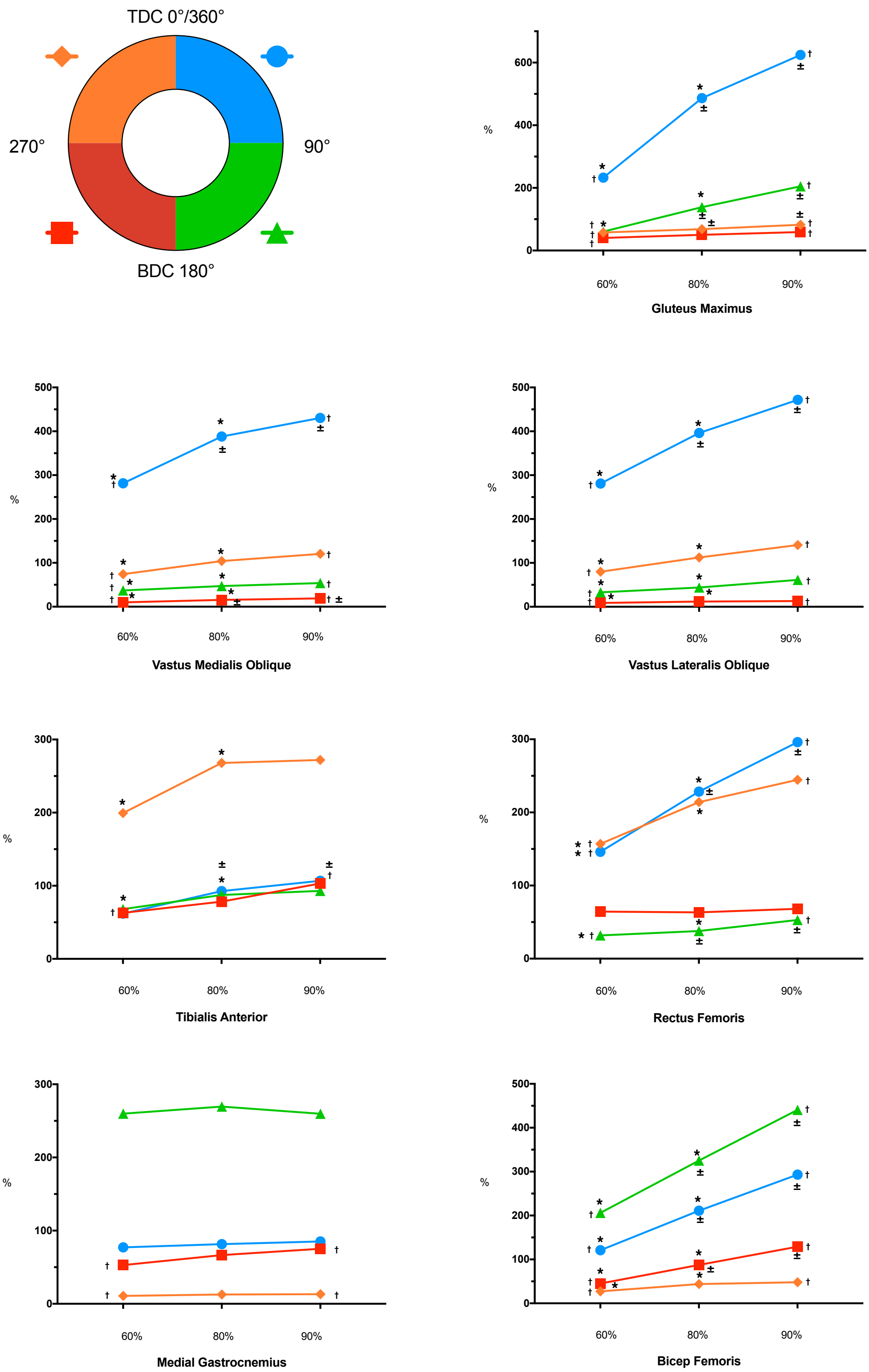

FIGURE 6.2: EMG magnitudes over different intensities, in each quadrant.
*significant difference between 60% and 80% maximum heart rate (MHR)
±significant difference between 80% and 90% MHR
†significant difference between 60% and 90% MHR.

DISCUSSION AND IMPLICATIONS

The ankle and knee normative values for static bike fitting have been well researched (Hamley and Thomas, 1967; Holmes, Pruitt, and Whalen, 1994; Peveler, Bishop, et al., 2005; Peveler, Pounders, and Bishop, 2007; Peveler, 2008; Bini, Hume, and Croft, 2011; Bini, Hume, Croft, and Kilding, 2011) and more recently the difference between static and dynamic angles have been highlighted (Peveler, Shew, et al., 2012; Fonda, Sarabon, and F. Li, 2014; Bini, Dagnese, et al., 2016; Holliday et al., 2017). As yet, there are no normative data to describe the dynamic values for full body joint angles recommended for cycling at differing intensities.

The muscle recruitment pattern during cycling has also been well researched (So, J. Ng, and G. Ng, 2005), including intramuscular EMG (Chapman et al., 2010; Silva et al., 2016). The purpose of this study was not to report the activation patterns, but to assess the changes in the lower limb EMG magnitudes during differing intensities as it is not clear how these may change with the natural adoption of different body positions as intensities increase, nor whether the joint kinematics change significantly.

This study produced similar results to that of Blake et al. (2012), showing that there was a general increase in muscle activation across muscle groups as the intensity increased. Blake et al. (2012) analysed nine male cyclists at a low (25-55% VO_{2max}) and a high intensity (60-90% VO_{2max}). In keeping with this previous study, our study also demonstrated an increase in EMG activity of TA and RF across the top and early part of the pedal cycle (0-90°) with increasing intensity. TA was shown to work predominately in the fourth quadrant, thus suggesting an increase into ankle dorsiflexion near the TDC. There was a significant change in RF from 60-90% MHR in the first, second and fourth quadrants which correspond to hip flexion, driving the knee over the TDC of the pedal revolution and knee extension in the push phase. The significant changes in TA and RF may indicate that these are the muscles responsible for driving the pedal across the TDC, an area where the major muscle groups are unable to exert effective force to drive crank rotation.

The role of GMax is to extend the hip joint and there were significant increases in EMG signal through all three intensities in the pushing phase of the pedal revolution (from 0-180°). Similarly, the VMO and VLO extend the knee joint in the same push phase of the pedal revolution, and there were significant increases in EMG signal through all three intensities in the first quadrant. The MG worked predominantly in the second quadrant, corresponding to the second half of the push phase of the pedal revolution, however the magnitude remained constant with only minor significant changes in the third and fourth quadrants between 60% and 90%. Even though Soleus was not examined in this study, it has been demonstrated that Soleus and MG work together from 340° through to 270° in the pedal revolution to stabilise the ankle and to transfer force to the pedal exerted by the relatively large GMax and quadriceps muscles (Jorge and Hull, 1986; Fonda and Sarabon, 2010). As such, even at lower workloads the force applied by MG in order to stabilise the ankle may be relatively higher. Similar results have been reported by Blake et al. (2012) where GMax had the largest increase in activity from a low to a high intensity, VMO and VLO were both highly active in the push phase with increasing intensity and the MG showed very little change with increasing workloads.

Numerous studies have shown an increase in hip and knee extension, as well as ankle dorsiflexion, with incremental cycling (Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010). The previous knee and ankle findings are consistent with our study, suggesting a movement into dorsiflexion to increase stability around the ankle joint in order to transfer force effectively to the pedals to maintain the power output. The movement into dorsiflexion may increase efficiency of MG or increase passive tension in the muscle tendon unit to assist with force transfer. The ankle increased into dorsiflexion by 6° between 60% and 90% intensity, which is a greater difference than the reported TEM of 3.5° (Holliday et al., 2017). As the bicycle contact points are fixed, this increase in ankle dorsiflexion requires an increase in knee extension (Peveler, Shew, et al., 2012). Dynamic bike fitting systems recommend a dynamic KFA of 30-40°, however there is no research validating this specific range. Previous research has demonstrated a difference in KFA of between 5° and 8° in static relative to dynamic measures (Farrell, Reisinger, and Tillman, 2003; Fonda, Sarabon, and F. Li, 2014; Holliday et al., 2017). The results from this study also demonstrated a difference in KFA at low and high intensities, and it is therefore possible to infer that optimal KFA at BDC position using dynamic measurements should range from 33-43° at low intensity and 30-40° at high intensity. Although statistically significant, from a clinical and practical perspective, it is recommended that the use of dynamic 2D and 3D kinematic data should interpret knee flexion in relation to the relative intensity during data capture.

There were no significant hip joint angle changes in any of the quadrants, at any of the intensities. This differs from previous studies that have shown hip extension increases with incremental cycling (Sanderson and Black, 2003; Bini and Diefenthaeler, 2010; Bini, Diefenthaeler, and Mota, 2010). The hip angles in previous studies were measured as an angle bisecting the length of the femur and a line parallel to the floor or as an angle bisecting the length of the femur and a line from the hip joint centre to the shoulder centre. These measures exclude the spinal segments and do not measure the independent hip joint angle (long axis of femur and lumbar spine-sacrum), as was done in this study.

Similar to the hip, the shoulder angle is often determined as an angle between the elbow, acromion and hip joint centre. A clinical shoulder angle will take the thoracic spine into account, as was done in this study. There were no significant changes in the shoulder angle, at any of the different intensities, yet the elbow and thoracic lean angle changed significantly between all three intensities. This is consistent with research where there was a significant change in forward body position on the bicycle at maximal power output (Bini, Senger, et al., 2012; Sayers and Tweddle, 2012). It was suggested that cyclists increased their trunk lean angle in response to muscular fatigue, and that changes in EMG preceded changes in mean trunk lean angle (Dingwell et al., 2008). It was hypothesised that the increase in trunk lean angle was in order to focus on increasing hip extensor muscle length and reducing knee flexor moment (Dingwell et al., 2008; Bini, Senger, et al., 2012).

Our findings that the hip joint position remained unchanged while significant lumbar flexion did occur, indicate that the previous basic methods of measuring the angles of the body, without taking into consideration the spine, should be discarded. The spine consists of 33 bones and each joint has varying degrees of movement. It is clear from this study that movement occurs in the lumbar, thoracic and elbow joints with increased intensity, not at the hip or shoulder. A possible rationale for this change in position may relate to the transfer of force across the hip joint. GMax demonstrated the largest change in EMG signal from low to high workloads. Increased GMax activity may aid the transfer of the increased force across the hip joint by stabilising the pelvis (L. Li and Caldwell, 1998). The increase in lumbar flexion and elbow flexion may therefore be a compensatory mechanism to stabilise the pelvis through the contact points at the hands as the forces across the hip joint increase (Grant, Watson, and Baker, 2015). Future research on more detailed spinal segment kinematics as well as spinal and upper limb EMG analysis with increasing intensity should be considered.

CONCLUSION

It is clear from this study that the magnitudes of muscles used during cycling increase with increasing intensity. The ankle adopts a more dorsiflexed position and the knee moves into a more extended position with an increase in cycling intensity. The elbow and lumbar and thoracic spinal segments also adopt a more flexed position as intensity increases. Previous recommendations for optimal cycling position have been suggested for the lower limb, however from these results it is essential that lumbar and thoracic spinal segments are also taken into account. Guidelines for optimal bicycle configuration should therefore consider the full body kinematics as well conducting the bike fit at an intensity applicable to a cyclist's individual training and racing goals.

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Chapter 7

The effects of relative cycling intensity on saddle pressure indexes

***W Holliday**, J Fisher and J Swart*

ABSTRACT

Objectives: To compare pressure load and distribution in various saddle zones through a range of workloads in order to provide clinicians and bike fitters with a better understanding of how to optimise saddle positioning.

Design: Experimental, quantitative study.

Methods: Saddle pressure of seventeen well-trained male cyclists was recorded at 60, 80 and 90% of maximal heart rate, based on data collected during a peak power output test.

Results: Loaded area increased significantly and progressively with increased workload while mean pressure did not change significantly. Point of load indexes in longitudinal and transverse planes both increased significantly and progressively with increases in workload. Distribution of load did not change with intensity.

Conclusions: Saddle pressure mapping should ideally be performed at an intensity similar to that which the cyclist will encounter during the majority of their training and racing. Comparative measurements of saddle pressures should also standardise workload intensity to ensure reliability of these measurements.

Keywords: *Bicycling; sport medicine; sports performance; ergonomics; bike fitting*

INTRODUCTION

The saddle is one of three contact points between the cyclist and the bicycle. A cyclist will make contact with the saddle at the ischial tuberosities and the pubic rami, which differ in shape and width from person to person. Due to this individual variation, large commercial companies have conducted studies to try and improve saddle comfort and reduce saddle-related pathologies (for example ©Specialized Bicycle Components). However, the extent of the body weight and thus pressure being transferred through each anatomical point depends on the cyclist's individual riding position and the anterior-posterior rotation angle of the pelvis on the saddle (Bressel and Larson, 2003).

With advances in technology we are now able to measure the pressure at the interface between the cyclist and the saddle. The reliability and validity of bicycle seat interface pressure measurements have previously been studied (Bressel and Cronin, 2005). The saddle was divided into three sections to differentiate anterior, posterior left and posterior right. It was concluded that the within-trial reliability is excellent for both mean and peak pressure values. The between-trial saddle pressures demonstrate moderate to excellent reliability for all areas of the saddle, except the anterior saddle pressure, which demonstrates poor reliability. However, errors may arise from lack of conformity between the saddle contours and the pressure mat, yet these are minimal.

Differences in saddle pressure indexes at varying workloads were also investigated in 22 cyclists who rode at 100 W and then 200 W on one standard saddle (Potter et al., 2008). The saddle was divided into five regions for analysis; total saddle, anterior, posterior, posterior left and posterior right regions. The vertical force and medio-lateral forces were both significantly reduced at 200 W compared to 100 W. Maximum pressure in the posterior region also significantly decreased with increasing workload. The forward position of the anterior center of pressure and the mediolateral width of the posterior center of pressures were both greater at the higher workload. The reduction in pressure at the higher workload was suggested to be as a result of the increased force applied to the pedals and transmitted through to the pelvis with the increase in power, which would reduce load on the saddle. This is similar to the findings by Bressel and Cronin (2005) who demonstrated a 39% reduction in pressure on the saddle at 300 W compared to 118 W. However, both these studies were conducted with the cyclist in a position they were not accustomed to (i.e. the bicycle configuration was not set at their freely chosen position). This may have had an impact on the results, as they may have shifted forwards or backwards on the saddle to obtain a more comfortable riding position.

More recently the effects of workload on saddle pressure between two different saddles has been investigated (Carpes et al., 2009). Eleven male and eleven female recreational cyclists volunteered for the study and rode at two different workloads, 150 W and 300 W, on two different saddle designs; one standard flat-surfaced saddle and one cutout saddle with a full-centre recess and a hole through the nose. The saddle design had little effect on the seat pressure, however there was a statistically significant increase in mean saddle pressure with the increase in pedalling workload. They postulated that the increased saddle pressure in different workloads may be due to the applied force to the pedals to gain forward propulsion. However, a limitation of this study was the saddle pressure system, as it did not give specific pressures for the anterior, posterior, left or right zones of the saddle, and they recommended that further research be done investigating a range of workloads to compare pressure load and distribution in the various saddle zones. The American College of Sports Medicine (2013) has described using heart rate ranges as a method for training intensity. During a race, a cyclist's heart rate intensity will vary according to the topographical profile of the race and overall distance (E. Faria, Parker, and I. Faria, 2005). The cyclist will spend only a fraction of the race or training at absolute fatigue and/or maximal effort greater than 90% heart rate intensity, with the majority of the ride shifting between 60-80% heart rate intensity (Palmer et al., 1994; Padilla et al., 2001). Investigating this range of workloads may provide clinicians and bike fitters with a better understanding on how best to optimise saddle positioning for cyclist's individual training or racing intensity.

The aim of this study was therefore to assess the change in saddle pressure indexes during three different intensities, namely 60, 80 and 90% of maximum heart rate. It was hypothesised that the mean saddle pressure as well as the area of loading would increase with the increase in intensity.

METHODS

Seventeen well-trained male road cyclists (28 $\pm$ 7 years, 75.5 $\pm$ 8.3kg, 181.9 $\pm$ 4.5cm, PPO 5.3 $\pm$ 0.8 W/kg) conforming to Level 2 or greater (De Pauw et al., 2013) were recruited for this study. Prior to testing each participant was informed of the risks and stresses associated with participation in the research trial, were personally interviewed about their training history, completed a Physical Activity Readiness Questionnaire (PAR-Q) (Whaley, Brubaker, and Otto, 2007) and signed an informed consent form. The study was approved by the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Cape Town, and conformed to the principles of the World Medical Association Declaration of Helsinki (World Medical Association, 2013).

The participants reported to the laboratory on two separate occasions with their own bicycle and cycling shoes. The participant's bicycle was loaded onto a Wahoo Kickr Smart Trainer (Wahoo Fitness®, 2018) and they rode in their own freely chosen bicycle configuration. On the first visit to the laboratory the participant's anthropometrics were taken, followed by an incremental exercise test to volitional exhaustion. The participants performed a standard warm-up and after a three minute rest period completed a Peak Power Output (PPO) and Peak Oxygen Consumption test to determine the required heart rate for the experimental trials. Gas analysis was monitored over 15 s intervals using an on-line breath-by-breath gas analyser and pneumotach (Oxycon, Viasis, Hoechberg, Germany). Participants started exercising at a workload of 100 W and resistance was increased by continuous ramp protocol at a rate of 20 W every 60 s until the participant was exhausted and could not sustain a cadence of at least 60 revolutions per minute (rpm). PPO was calculated by averaging the power output for the final minute of the VO_{2peak} test. VO_{2peak} was recorded as the highest VO_2 reading recorded for 30 s during the test. Maximum heart rate (MHR) was recorded as the highest heart rate achieved during the incremental exercise test.

On the second visit to the laboratory, each participant's bicycle was loaded onto a Wahoo Kickr Smart Trainer (Wahoo Fitness©, 2018) and a standard saddle was fitted to their bicycle, ensuring that saddle height and setback remained the same. A standard saddle (Fabric® Scoop Elite Shallow, 142mm) was used for all participants to reliably compare the data as previously recommended (Bressel and Cronin, 2005). The saddle pressure mapping mat (Gebiomized®) was placed on the saddle by the same investigator to ensure repeatability of the positioning throughout the trial.

Each participant performed a standardised warm-up followed by a fifteen minute exercise trial at three different workload intensities based on the Lamberts Submaximal Cycle Test (Lamberts et al., 2009), which proved to be highly reliable. The intensity was set at 60% of their individual MHR, recorded during the VO_{2max} test, for the first six minutes, immediately followed by six minutes at 80% MHR and a further three minutes at 90% MHR. Resistance was increased via the Wahoo Fitness app (v5.13.3) until the desired heart rate was achieved. The saddle pressure mapping was recorded during the last minute of each stage, with the participant's hands on the hoods and seated. Participants were requested to maintain a cadence as close to 90rpm as possible throughout the trial. Participants were not informed when the saddle pressure mapping data was to be recorded, so as to prevent them changing their pedalling action.

A saddle pressure mapping system (GebioMized®) was used for all testing. The system comprises a thin flexible mat containing 64 sensors, which was fitted over the saddle. The pressure mat was calibrated as per manufacturer instructions. The data collected was transmitted wirelessly to the manufacture software which was installed on a standard Windows computer. The GebioMized system generates a report on the following:

- Mean pressure; defined as the average instantaneous peak of the maximum pressure recorded at each sensor in each area
- Loaded area for the anterior pubic bone, rear left sit bone and rear right sit bone zones (**Figure 7.1**)
- Absolute maximum of force; defined as the maximum instantaneous peak force
- Mean of total force; sum of forces recorded by each sensor divided by the number of sensors
- Longitudinal (front to back) and transverse (left to right) mean movement of the centre of pressure (CoP), also known as the point of load incidence (**Figure 7.1**)

The system classifies the cyclists sitting position as either Front or Rear and determines a regression line angle, indicating pelvis orientation.

Dynamic saddle pressure mapping data was recorded for the final ten seconds during the last minute of each interval. Specifically at 5, 11 and 14min. The data were analysed using the manufacturer software. All mean pressure measurements were normalised to body weight.

All data are expressed as means $\pm$ standard deviation (mean $\pm$SD). Analysed variables were statistically tested using a one-way ANOVA with repeated measures. When significant main effects were found, a Tukey test was used for post-hoc analysis. Significance was accepted when p-value <0.05. The statistical analyses were performed using GraphPad Prism v7.0a (GraphPad Software, San Diego, CA, USA).

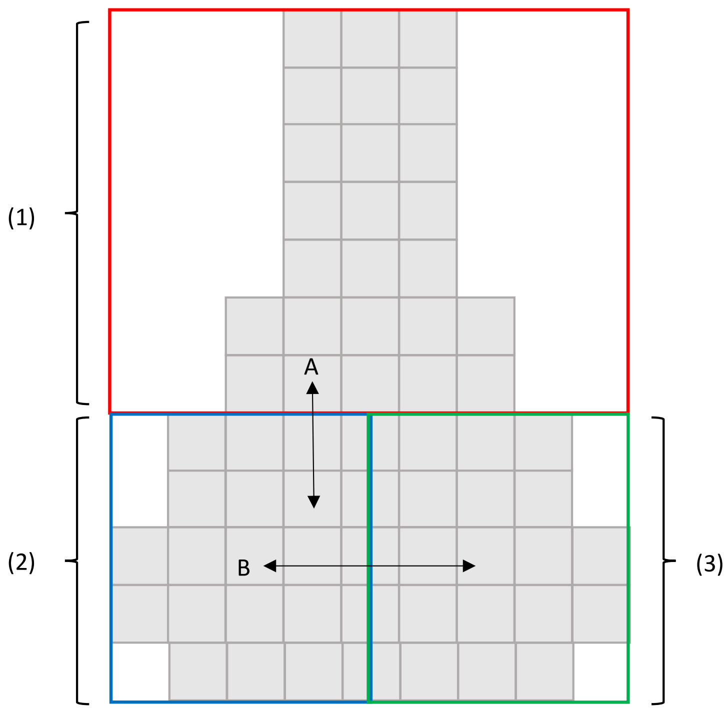

FIGURE 7.1: Pressure mat showing anterior pubic bone (1), rear left sit bone (2) and rear right sit bone (3) zones. Longitudinal (A) and Transverse (B) movement of the Centre of Pressure. Each square depicts a pressure sensor.

RESULTS

There were significant changes in the loaded area of the pubic bone zone with an increase in intensity, with changes in means from 5058.82 ± 1323.49mm^2 at 60%, 5247.06 ± 1278.03 at 80% and 5445.59 ± 1233.97 at 90%, $F(1.56, 21.8) = 6.62$, p-value = 0.01. There were significant changes in the loaded area of the left sit bone zone, $F(1.85, 29.65) = 20.80$, p-value <0.01, and right sit bone zone, $F(1.33, 21.24) = 6.18$, p-value = 0.01. The left sit bone zone demonstrated significant changes between all three intensities, with an increase in mean from 4905.88 ± 994.38mm^2 at 60%, 5325.00 ± 902.04 at 80% and 5630.88 ± 764.93 at 90%. The right sit bone zone demonstrated significant changes between 60% (5195.59 ± 1013.40) and 90% (5604.41 ± 1092.81) only (**Table 7.1** and **Figure 7.2**). The total loaded area as a percentage of the total area demonstrated a significant increase between all intensities, from 42.71 ± 9.47% at 60%, 46.06 ± 4.99 at 80% and 47.76 ± 4.25 at 90%, $F(1.59, 23.88) = 22.64$ and p-value <0.01 (**Table 7.1** and **Figure 7.2**).

The movement of the CoP, along the longitudinal and transverse axes, increased significantly. The longitudinal axis demonstrated a significant change between 60% (21.82 ± 7.41mm) and 80% (30.18 ± 9.46mm) and between 60% and 90% (36.00 ± 13.96mm) intensity, $F(1.43, 22.94) = 18.78$, p-value <0.0001. The transverse axis demonstrated a significant change between all three intensities, increasing from 19.18 ± 7.41mm at 60%, 33.65 ± 12.84 at 80% and 43.06 ± 18.19 at 90%, $F(1.51, 24.16) = 39.17$, p-value <0.0001 (**Table 7.1** and **Figure 7.2**).

There were no significant changes in mean pressure for the pubic bone zone nor the rear left and right sit bone zones. Similarly there were no significant changes in absolute maximum of force and front or rear pressure distribution. The mean of total force demonstrated a significant decrease between all three intensities, from 463.47 ± 97.20N at 60%, 407.94 ± 106.29 at 80% and 380.00 ± 107.56 at 90%, $F(1.66, 26.02) = 26.02$, p-value <0.01 (**Table 7.1** and **Figure 7.2**).

TABLE 7.1: Mean ± standard deviation of pressure mapping variables.

Variable		60%	80%	90%	p-value
Mean pressure (Normalised to body weight) (mbar)	Pubic bone	251.71 ± 114.75 (3.32 ± 1.43)	257.41 ± 121.85 (3.38 ± 1.47)	253.76 ± 123.14 (3.34 ± 1.48)	0.93
	Sit bone (L)	237.94 ± 48.42 (3.18 ± 0.67)	241.24 ± 58.78 (3.22 ± 0.77)	241.71 ± 59.27 (3.21 ± 0.74)	0.91
	Sit bone (R)	257.76 ± 69.10 (3.46 ± 1.04)	244.35 ± 69.08 (3.27 ± 1.00)	252.41 ± 56.96 (3.37 ± 0.79)	0.36
Loaded Area (mm^2)	Pubic bone	5058.82 ± 1323.49	5247.06 ± 1278.03	5445.59 ± 1233.97	0.01†
	Sit bone (L)	4905.88 ± 994.38	5325.00 ± 902.04	5630.88 ± 764.93	<0.001*†±
	Sit bone (R)	5195.59 ± 1013.40	5476.47 ± 1081.32	5604.41 ± 1092.81	0.01†
Absolute Maximum of Force (N)		595.53 ± 125.61	574.00 ± 129.27	563.29 ± 128.69	0.30
Mean of Total Force (N)		463.47 ± 97.20	407.94 ± 106.29	380.00 ± 107.56	0.001*†±
Loaded area/ total area (%)		42.71 ± 9.47	46.06 ± 4.99	47.76 ± 4.25	0.001*†±
Area of pressure (%)	Front	47.94 ± 15.47	49.06 ± 15.94	48.65 ± 15.00	0.74
	Rear	52.06 ± 15.47	50.94 ± 15.94	51.35 ± 15.00	0.74
Point of load incidence (mm)	Longitudinal axis	21.82 ± 7.41	30.18 ± 9.46	36.00 ± 13.96	0.001*†
	Transverse axis	19.18 ± 7.41	33.65 ± 12.84	43.06 ± 18.19	0.001*†±

* significant change between 60 and 80% maximum heart rate (MHR)
† significant change between 60% and 90% MHR
± significant change between 80 and 90% MHR

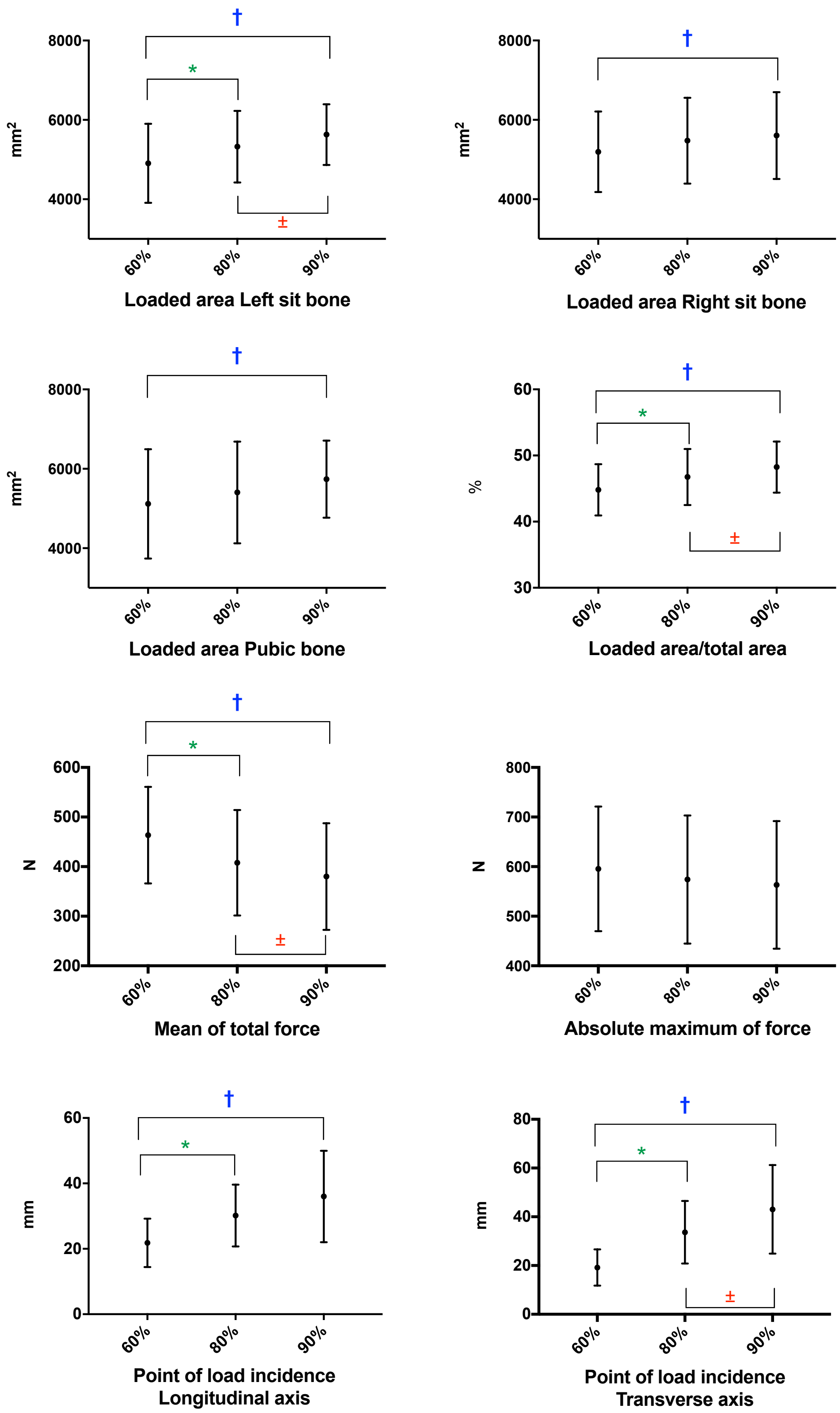

FIGURE 7.2: Loaded area Left and Right sit bone, Pubic bone and Point of load incidence, Longitudinal and Transverse axis, Mean of total force and Absolute maximum of force. *significant difference between 60% and 80%, †significant difference between 60% to 90%, and ±significant difference between 80% and 90% maximum heart rate.

DISCUSSION

The purpose of this study was to assess the interaction between the cyclist and the saddle during three different cycling intensities. We observed a progressive increase of the loaded area with increasing intensity and this occurred in all three of the saddle zones. In addition, there was a decrease in mean total force with increasing intensity and an increase in CoP movement with increasing intensity.

Mean pressure remained unchanged. This variable is the mean of the instantaneous peak pressures recorded for sensors in each area. This may be confused with the mean pressure over time. Researchers and practitioners should be aware of this definition to avoid incorrect interpretation of the outcome values.

In contrast to our findings, Carpes et al, (2009) demonstrated an increase in saddle pressure between two different workloads, 150 W and 300 W. However these participants were instructed to maintain their trunk angle at 60°, with the researchers ensuring this position was kept constant during the trials. It is possible that this intervention may have altered the relative distribution of pressure on the saddle. It has previously been demonstrated that as intensity increases, riders naturally flex the thoracic and lumbar spine and flex the elbow joint while maintaining the kinematics of the hip and shoulder joint (Holliday et al., 2019). In contrast, in our study the participants were allowed to adopt their freely chosen posture. Previously, no change in pelvic anterior-posterior rotation with an increase in intensity was demonstrated (Bini et al., 2016), and our data confirms this as the relative distribution of pressure (front and rear) on the saddle did not alter significantly. However as intensity increases there is a progressively larger loaded area of the perineal area in contact with the saddle.

Two studies have paradoxically demonstrated a reduction in pressure with an increase in workload. Mean and peak pressures were greater at 118 W compared to 300 W when riding with the hands holding on the tops of the handlebars (Bressel and Cronin, 2005).

Likewise, total, anterior and posterior maximum pressure all decreased at 200 W compared to 100 W, with an increase in anterior and medio-lateral CoP movement (Potter et al., 2008). As the power increased to 200 W, the vertical force on the saddle decreased. The authors postulated that a greater power output would necessitate a greater force application at the pedals. As this force is mediated primarily through hip and knee extension it would act to reduce the load on the saddle. This is confirmed by our findings that there was a significant decrease in mean force with an increase in intensity.

The movement of the CoP (also known as the point of load incidence), in both the longitudinal and transverse planes, increased progressively with intensity. This is in keeping with the previous findings by Potter et al (2008). This is a measure of stability and is used in clinical practice to assess the stability of the rider position when adapting the bike fit parameters. Natural pelvic roll from side to side has been demonstrated to occur in cycling (Farrell, Reisinger, and Tillman, 2003; Sauer et al., 2007). This pelvic rocking can be exaggerated at higher speeds (Farrell, Reisinger, and Tillman, 2003) and our results indicate that this pelvic rocking increases at higher workloads independent of cadence, which remained unaltered. As both transverse and longitudinal movements increase with increasing workload, any comparative measurements when adapting contact point position should be compared at the same relative workload intensity.

CONCLUSION

The contact area between the cyclist and the saddle and the CoP movement increases with intensity while total saddle force decreases. Although the hand position should be standardised, the cyclist should be allowed to adopt a natural riding position when comparing saddle pressure measurements after altering contact point position or when measuring pressure using different saddle designs.

ACKNOWLEDGEMENTS

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Practical applications

- Guidelines for optimal saddle choice should take into account the training discipline and the intended riding intensity of the cyclist, such as for recovery, endurance, tempo, threshold and superthreshold.
- Cyclists should adopt a natural riding position for saddle pressure mapping during bike fitting.
- For cyclists who are interested in training or racing at high intensities, we recommend that assessments be conducted at 80% of MHR or a similar standardised intensity to better replicate the forces produced during high intensity cycling and racing.

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Chapter 8

Summary of research findings

SUMMARY OF RESEARCH FINDINGS

A cyclist usually seeks a bike fit for two main reasons: to improve their performance or to do so comfortably (Ayachi, Dorey, and Guastavino, 2015; I. Priego Quesada et al., 2017; J. Priego Quesada et al., 2018). Bike fitting can be grouped into two categories: the optimal range fit and the accommodated fit (Medicine of Cycling, 2013). The optimal range fit is defined as a set of variables that the fitter is able to accomplish whereby the cyclist is positioned within, what are considered, optimal ranges for the type of cycling they are partaking in, i.e. the ideal individualised position that meets the goals of the cyclist is met. Whereas the accommodated fit is considered when there are limitations, either due to the cyclist's own limitations (flexibility for example) or to components of the bicycle that are not ideally sized (the frame or integrated stem and handlebars for example). An unsurpassed position is when the bicycle is optimally configured to the cyclist and the cyclist is considered to be in an ideal, comfortable and sustainable position on the bicycle.

In order to achieve this unsurpassed position, the individual anthropometrics, flexibility, training history, type of cycling and comfort of the cyclist needs to be taken into account. To date there are many static methods used to configure the bicycle optimally, however with the advancement of technology, 2D and 3D dynamic methods are being used but with limited regard to the paucity of scientific reference points recommended for the optimal fit. Previous methods used for configuring the bicycle have not always considered comfort, and thus cyclists may have experienced a degree of discomfort in order to perform optimally (Ayachi, Dorey, and Guastavino, 2015). Nor did the previous methods take into consideration individual characteristics such as leg length and pedalling styles (Peveler and Green, 2011). These should be considered during customising the configuration for comfort. The other common goal of most sports participants is to improve their performance. Similarly, the recommended ranges for performance likewise do not always consider individual variations of riding style or type of training and racing.

Once ranges for configuration, which consider individual intrinsic factors, have been established, the bicycle configuration can be adjusted according to the type of training the cyclist undertakes. Cyclists will prepare for races by variably combining long steady endurance sessions with high intensity maximal or near-maximal workload interval sessions (Laursen and Jenkins, 2002; Laursen, 2010). The cyclist will spend only a fraction of the race or training at absolute limits of fatigue and/or maximal effort greater than 90% heart rate intensity, with the majority of the ride shifting between 60-80% heart rate intensity (Palmer et al., 1994; Padilla et al., 2001). The cycling position and muscle activity may therefore change according to the types of training conducted.

Therefore, to further optimise bicycle configuration, it is important to determine how the cycling modality or discipline and the training intensity can have an effect on the saddle interface pressure. The saddle interface is one of the most common contributing factors to pain or discomfort whilst cycling (Dettori and Norvell, 2006).

The overall aims of this thesis were therefore to determine the reliability and validity of different measurements systems currently used in bicycle configuration, how intrinsic factors may affect bicycle configuration and performance, as well as determining how cycling intensity may influence full body cycling kinematics, muscle electromyography signal amplitude and saddle pressure mapping indexes.

An outline of the main findings and questions presented in Chapter 1 of the study are summarised briefly:

1) Are the measurement tools used in bike fitting reliable?

In chapter two we aimed to assess the difference between static and dynamic measures of the ankle, knee, hip, shoulder and elbow, as well as the within-subject reliability of these three measurement techniques. It is beneficial to be aware of the difference between static and dynamic angles measured during cycling, as changes in the kinematic chain may have an effect on performance, economy and injury risk.

Measures were taken statically with a standard goniometer (GM) and an inclinometer

(IM), and dynamically with three dimensional motion capture (3DMC) using an 8 camera motion capture system. All three instruments were valid and reliable with a low typical error of measurement (TEM) across all three techniques, for all of the measured joints. There was a general trend towards higher inter-session reliability for the IM measurements when compared to either GM or 3DMC. The study demonstrated a positive correlation between GM and IM measures for all joints. Only the knee, shoulder and elbow were positively correlated between GM and 3DMC, and IM and 3DMC. The 3D motion analysis utilises different landmarks for most joints and produces different means. The 3DMC therefore measures a different hip angle to that measured with the IM and GM. Both the GM and IM measure the hip joint as the angle subtended by the area expanding below the iliac crest from the third lumbar vertebra to the sacrum, to the line bisecting the greater trochanter and lateral femoral condyle. To determine the hip flexion angle the 3DMC system uses a perpendicular line bisecting the anterior superior iliac spine (ASIS) and posterior superior iliac spine (PSIS), and a line bisecting the knee joint centre and greater trochanter. In addition to this, it is also possible that the 3D analysis software, which incorporates rotational movements in the axial plane (such as pelvic rotation), may account for some of the differences between static and dynamic 3D measurements. The ankle was also measured in 3DMC using different landmarks to the static methods and in addition, the shoe length and height may have contributed to the difference in means.

Joint measurements using a GM or IM were demonstrated to be relatively interchangeable, and are inexpensive and easy to use (De Vey Mestdagh, 1998). Motion capture is expensive, timely and aversive, and although reliable, great care is required to place the markers in the exact location from one trial to the next.

It was demonstrated that the 3DMC model used in this study is thus not interchangeable with GM and IM, and it is recommended that 3DMC develop its own independent reference values for clinician-based bicycle configuration.

2) Is there a relationship between freely chosen bicycle configuration and individual anthropometrics, flexibility and training?

The main aim of the second study (Ch 3) was to establish how individual anthropometrics, training history and flexibility factors may be associated with cyclists' freely chosen bicycle configuration, and to determine the full body static flexion angles chosen by cyclists on their bicycles. In order to fully optimise cyclist's position on their bicycle, individual anthropometrics and comfort should be taken into consideration. Individual bicycle configuration, static joint angles on the bicycle, anthropometrics, flexibility and training history were recorded to determine if there was a relationship between any of these variables.

There was a strong correlation between leg length and seat height and a moderate correlation between leg length and saddle setback. There was a strong correlation between stature and handlebar reach. However, there was no statistically significant correlation between arm length and handlebar reach. There were also moderate correlations between stature and saddle setback; and arm length and saddle setback. There was a moderate correlation between leg length and handlebar drop, and hamstring flexibility and stature. It appears that a cyclist with longer legs and a greater hamstring flexibility adopts a greater saddle setback and handlebar drop. The increased hamstring flexibility may allow the pelvis to rotate anteriorly, which in turn allows the cyclist to reach a lower handlebar height more comfortably.

Our results demonstrated similar average kinematic ranges of the ankle, knee and elbow joints to previous recommendations (Holmes, Pruitt, and Whalen, 1994; De Vey Mestdagh, 1998; Burt, 2014). Suggestions for hip and shoulder joint ranges were recommended as there is an absence of existing research with respect to recommendations for these joint angles.

The results from this study demonstrate that individual variables may influence static bicycle fitting, and thus anthropometrics and flexibility should be taken into account for optimal bicycle configuration.

3) Is there a relationship between freely chosen bicycle configuration, performance, and flexibility?

In Chapter 4 we aimed to determine if relationships between power production, bicycle configuration and flexibility exist, and how best to apply these clinically to the cyclist during the bike fitting process. Individual anthropometrics, flexibility and individual bicycle configuration were recorded and correlated to individual power and performance.

There was a small to moderate correlation between performance and hamstring flexibility, handlebar drop, saddle setback and ankle plantarflexion. A greater hamstring flexibility may allow the cyclist to adopt a more anteriorly rotated pelvic position, which may also be facilitated by a lower handlebar position. The greater saddle setback and handlebar drop may place the cyclist in a more powerful position from which to maximise gluteal and plantarflexor muscle force in the push phase of the pedal revolution.

The results from this study have clinical implications for bike fitters and cyclists. Greater saddle setback and lower handlebar height may increase peak power output. Improving a cyclist's flexibility and ability to adopt an anteriorly rotated pelvis and lower handlebar height may increase the force generated in the push phase of the pedal stroke and thus improve cycling performance.

4) How do the full body kinematics and EMG muscle magnitudes alter during a steady state submaximal cycle?

We aimed to assess the changes in lower limb EMG magnitudes and 3D kinematics of the ankle, knee, hip, lumbar, thoracic, shoulder and elbow angles of cyclists during a 60 minute steady state cycle at 60% of their VO_{2max} power in Chapter 5.

Participants performed two trials of an hour duration each, where kinematic and EMG magnitudes were analysed for each third of the one hour steady state cycle. There were no statistically significant changes in muscle EMG magnitudes nor were there any changes in joint kinematics over the hour-long steady state cycle at a moderately high intensity. This has a positive implication for bicycle configuration as bike fitting for steady state endurance rides can be performed as per existing recommendations as no body position nor muscle EMG changes were encountered.

5) How do the full body kinematics and EMG muscle magnitudes alter during differing cycling intensities?

Full body kinematics and muscle EMG magnitudes may alter based on the workloads that are encountered during cycling. Understanding the effect of differing intensities on the cyclist can guide clinicians and bike fitters in improving specific muscle strength and cycling posture to optimise training and racing. In Chapter 6 we aimed to assess changes in lower limb EMG magnitudes and full body 3D kinematics of well-trained cyclists at three different exercise intensities: 60, 80 and 90% of maximum heart rate.

Cyclists' ankle dorsiflexion and knee extension increased with higher intensities, whilst the elbow and lumbar and thoracic segments adopted a more flexed position. There were no changes in the clinical hip and shoulder angles. Previous studies have demonstrated an increase in ankle dorsiflexion and knee extension, however hip and shoulder angles were previously measured with simplistic methods, not taking the spinal flexion into account. Our results demonstrate that altered kinematics occur in the lumbar and thoracic spine with increasing intensity and it is recommended that these changes should be considered when optimising the bike fitting process for high intensity cycling.

There were significant increases in EMG signal amplitude at higher intensities for all muscle groups measured. These results can guide clinicians in strengthening specific muscles to optimise force production at specific points in the pedal revolution.

Guidelines for optimal bicycle configuration should take into account the full body position of the cyclist as well as the cyclist's training and racing intensity when assessing kinematics. Comparisons of the joint angles and muscles between steady state cycling and cycling at differing intensities are shown in **Figure 8.1** to **Figure 8.4**.

6) How do the saddle pressure mapping variables alter during differing cycling intensities?

Finally, in Chapter 7 we compared pressure load and distribution in various saddle zones through a range of workloads. This was to provide clinicians and bike fitters with a better understanding of how to optimise saddle positioning.

Saddle pressure indexes of a group of well-trained cyclists were recorded at 60, 80 and 90% of maximal heart rate. Loaded area increased significantly and progressively with increased workload while mean pressure did not change significantly. It has previously been suggested that at greater power outputs the greater applied force on the pedals is transmitted through the kinetic chain and results in a force elevating the pelvis from the saddle (Potter et al., 2008), and thus our mean pressure data remained constant despite greater forces and movement during each pedal stroke.

Point of load indexes in longitudinal and transverse planes both increased significantly and progressively with increases in workload. Point of load indexes are used as a measure of stability and can be used in clinical bike fitting to assess the stability of the cyclist's position when adapting the bike fit parameters. Natural pelvic roll from side to side has been demonstrated to occur in cycling (Farrell, Reisinger, and Tillman, 2003; Sauer et al., 2007), and can be exaggerated at higher speeds (Farrell et al., 2003). Our results indicate that this pelvic rocking increases at higher workloads independent of cadence, which remained unaltered.

Saddle pressure mapping should ideally be performed at an intensity similar to that which the cyclist will encounter during the majority of their training and racing. Comparative measurements of saddle pressures should also standardise workload intensity to ensure reliability of these measurements.

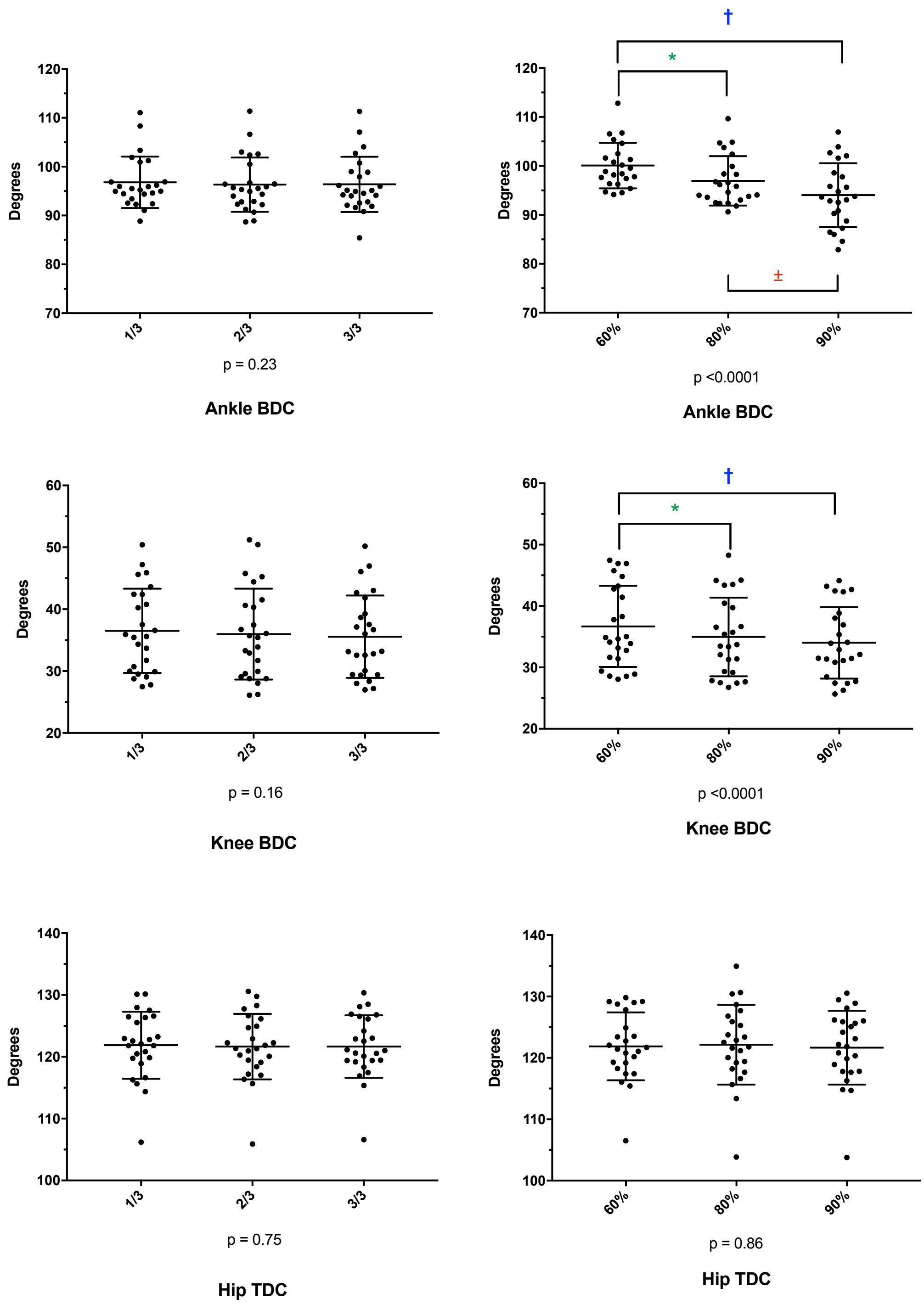

FIGURE 8.1: Comparison of steady state cycling and differing intensities for the ankle, knee and hip joints.

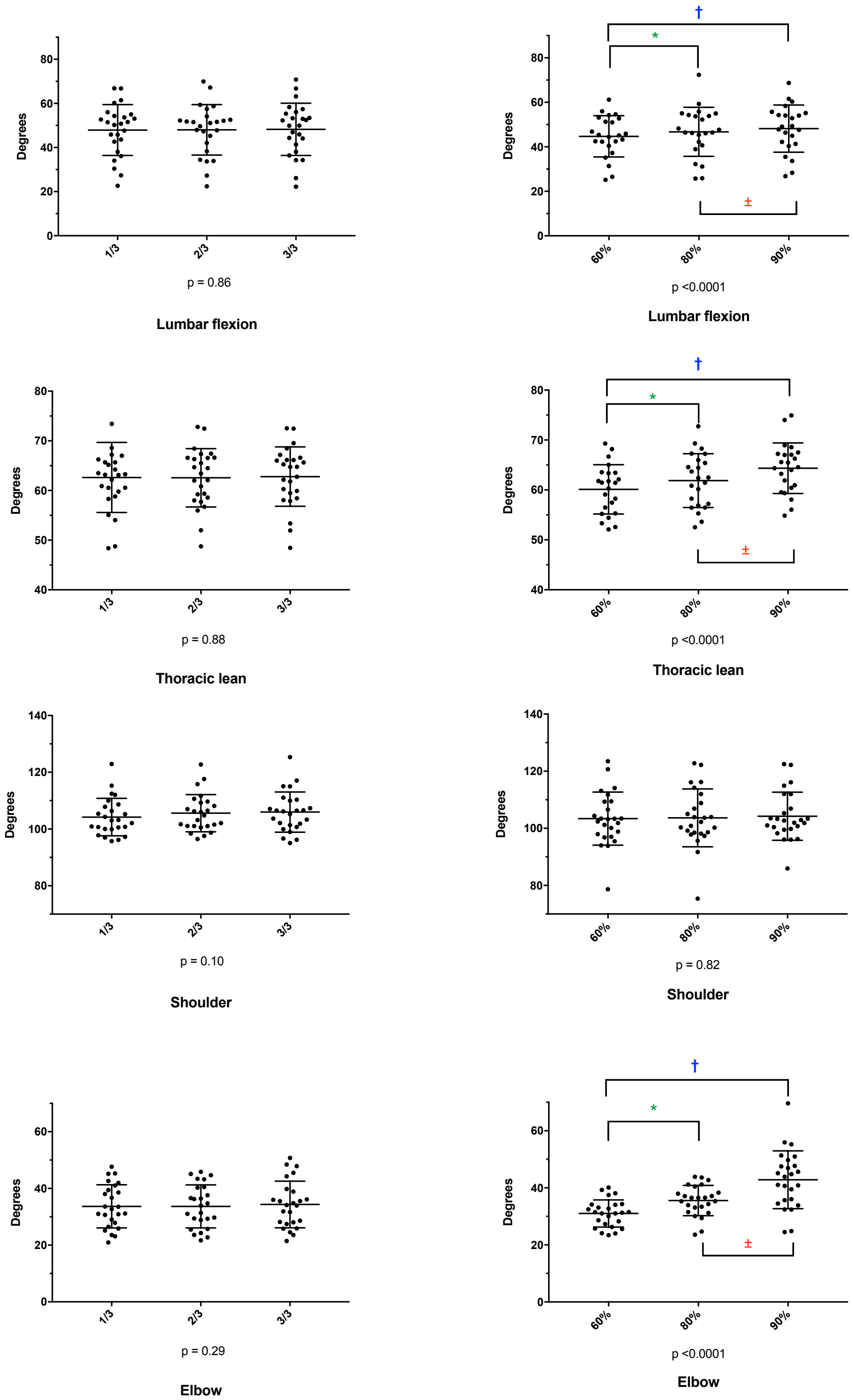

FIGURE 8.2: Comparison of steady state cycling and differing intensities for the lumbar, thoracic, shoulder and elbow joints.

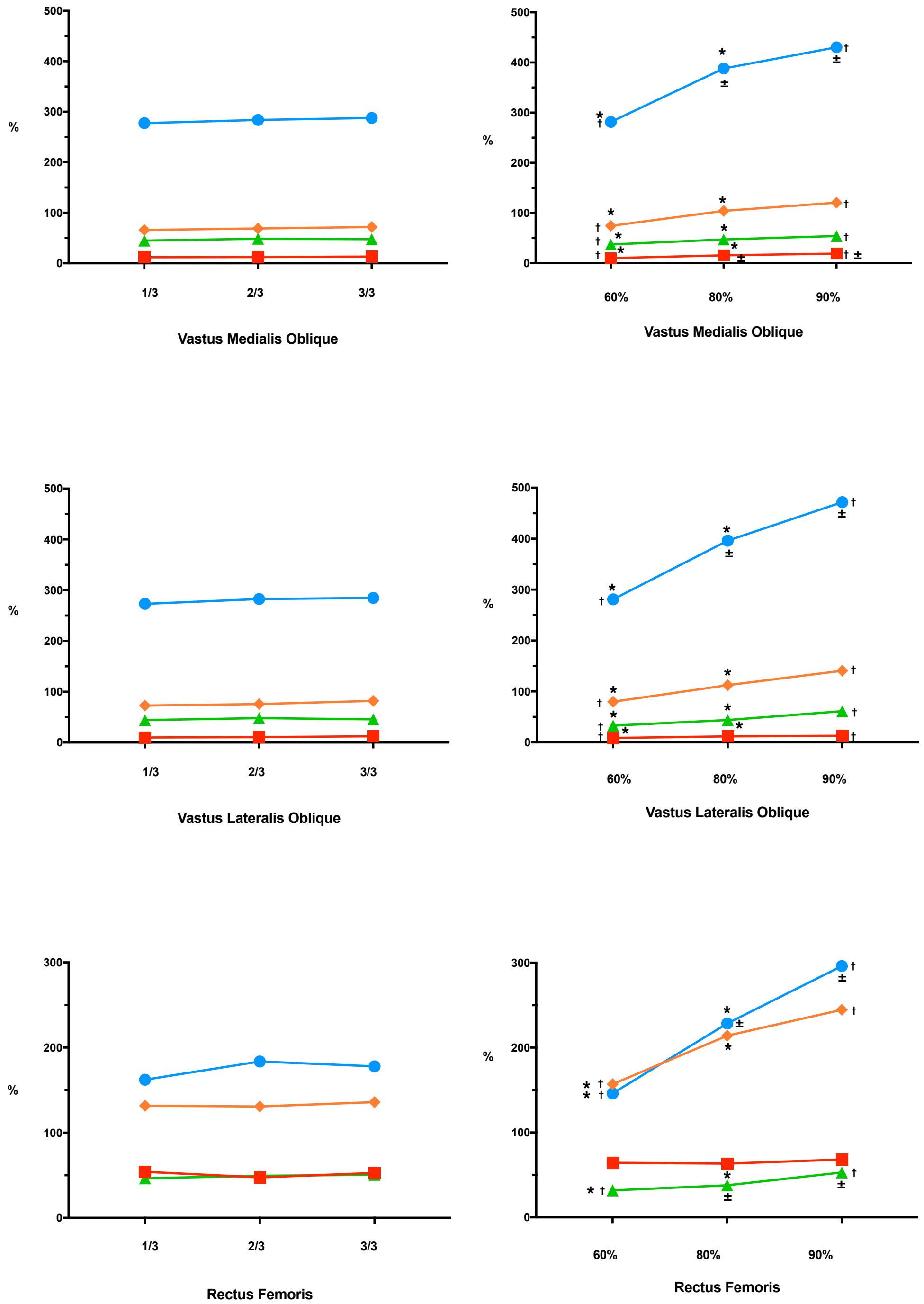

FIGURE 8.3: Comparison of steady state cycling and differing intensities for the Vastus Medialis Oblique, Vastus Lateralis Oblique and Rectus Femoris.

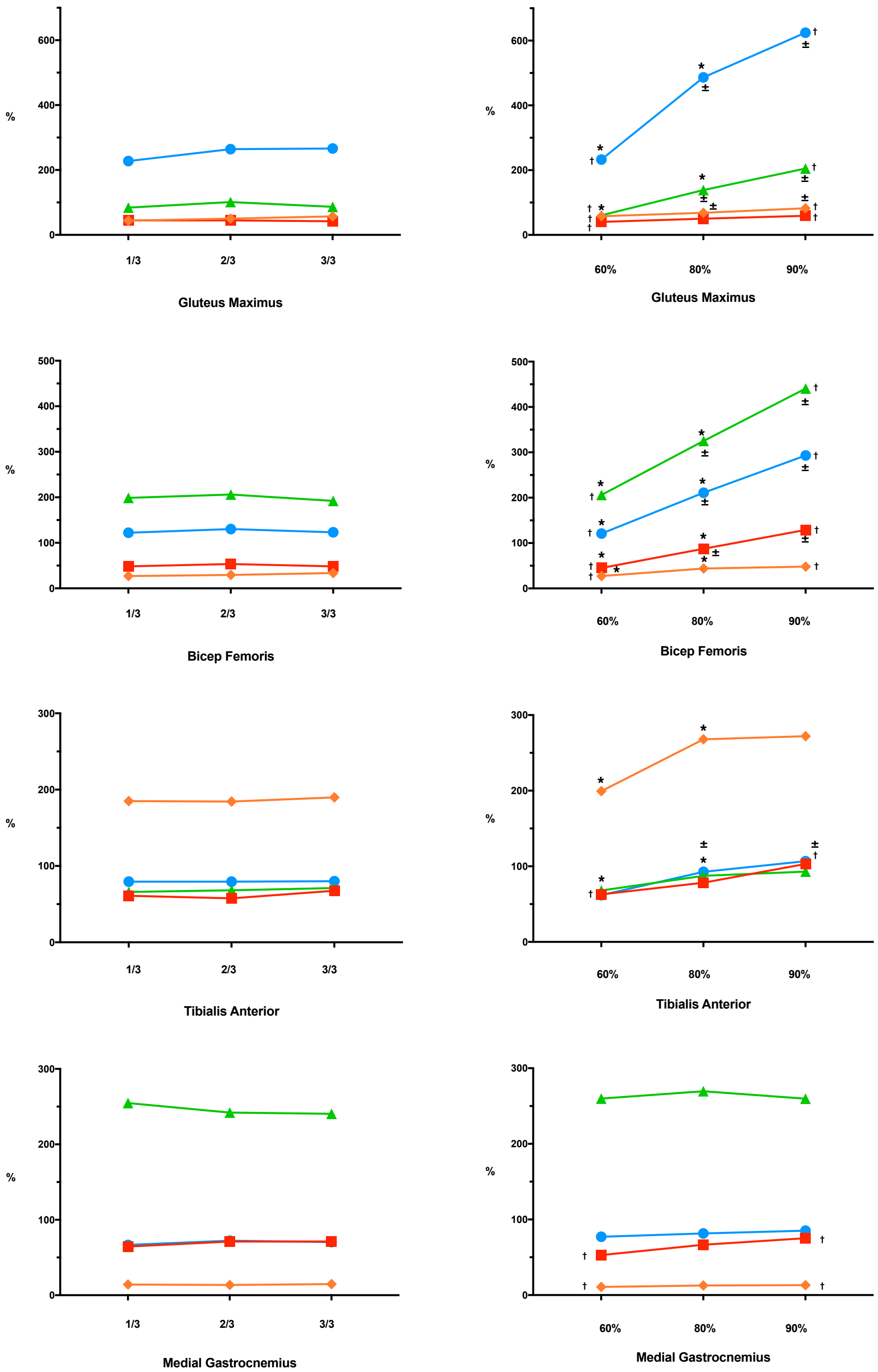

FIGURE 8.4: Comparison of steady state cycling and differing intensities for the Gluteus Maximus, Bicep Femoris, Tibialis Anterior and Medial Gastrocnemius.

FUTURE RESEARCH QUESTIONS

Although this thesis has added further insight to guide clinicians during both static and dynamic methods of bike fitting, additional research is needed to establish clinical bike fitting guidelines which address individual anthropometrics, flexibility, as well as training and racing goals.

We have shown that the static guidelines for bicycle configuration cannot be transferred to dynamic methods, and future research should focus on developing a clinically based reference guide for 3D motion capture to be used during bike fits.

We demonstrated that there are significant changes in the ankle, knee, lumbar, thoracic and elbow angles, as well as muscle magnitudes during differing cycling intensities. Future research should focus on how best to position a cyclist for these intensities and which range of motion to best strengthen the muscles in order to work optimally. Similarly, the pelvic tilt during cycling has an impact on both power performance and saddle comfort and positioning. Future research should focus on measuring the pelvic tilt and the effect this has on performance and saddle comfort.

CONCLUSION

The bike fitting industry has grown exponentially, with the title ‘Bike Fitter’ becoming an established career option. New methods in which to optimise bicycle configuration have developed considerably in the last 5-10 years. Bicycle configuration has become a science rather than the art form it was back in the formative years of competitive cycling.

With the advancement of new technology and measuring systems, this thesis aimed to determine the reliability and validity of different measurement systems currently used in bicycle configuration. Furthermore, intrinsic factors which may affect bicycle configuration and performance were determined, as well as determining how cycling intensity may influence full body cycling kinematics, muscle magnitudes and saddle pressure mapping.

The previous methods of using static measurements for bike fitting cannot be transferred to dynamic methods. Static bike fitting may be advantageous as it is a simple, less costly method and is highly reliable (Visentini and Clarsen, 2016), however a dynamic method is a more accurate representation of the cyclist's position and movement, especially during increases in workload (Peveler, Shew, et al., 2012). Dynamic kinematic measurements differ depending on the cyclist's relative power output and should be assessed at a specific percentage of maximal heart rate or power output.

Due to the time-consuming process of measuring and analysing dynamic methods, it is recommended that one starts with a static bike fit. The use of static methods to measure joint angles can be used to guide the initial position for saddle height, saddle setback, handlebar reach and handlebar drop, before moving onto dynamic assessment techniques to fine tune the fit at the intended racing or training intensity. Bike fitting should also be viewed as an ongoing process; as cyclists' performance parameters, strength and flexibility may change as they train for specific races. The individual variables such as riding style, stature, arm and leg length, flexibility and performance of cyclists should be considered when configuring their bicycle. Individual anthropometrics, flexibility and power ouptut will guide bike fitters into configuring the cyclist's position on the bicycle optimally. The bike fit should be adjusted to match the intrinsic factors as well as individual goals and strength of the cyclist.

Muscle EMG amplitudes will increase with increased intensity. The results from this thesis can provide insights to guide clinicians and cyclists in strengthening specific muscles at specific ranges of the cycling pedal revolution. Similarly, for cyclists wanting to adopt and sustain a lower handlebar drop position more comfortably, they may need to increase their hamstring flexibility.

It has become clear throughout this thesis that previous methods of measuring the hip and shoulder angles were overly simplistic and did not take into account the spinal flexion angles nor the rotation of the pelvis. From our results it has been demonstrated that the pelvis angle has a large role in the power produced and that the lumbar and thoracic spine kinematics change with increasing cycling intensities. The access to motion capture systems, and the overall advantages of assessing dynamically, makes it imperative that future bike fitting systems include the pelvis, lumbar and thoracic spine kinematics in their measurements and assessment.

In summary, static kinematics are a valid and reliable method that provide a useful tool in the process of bike fitting. They provide an easy, rapid and cost-effective means of assessing the cyclist that can be used in the early phases of a more complex fitting or as a stand-alone fitting process for the more cost-effective and shorter bike fit. Individual intrinsic factors and flexibility can be used as a guide for the initial bicycle configuration. Dynamic methods, including pressure mapping, aerodynamics and force pedals, can then be used to fine tune the bicycle configuration according to the specific needs and riding intensities of the cyclist.

SUMMARY OF CLINICAL GUIDELINES

- Static measurements demonstrated moderate to excellent reliability with smaller Limits of Agreement than the dynamic measurements which only demonstrated poor to good reliability
- Dynamic 3D motion capture is not interchangeable with static (goniometer and inclinometer) measurements
- There is a need for dynamic reference values for all joints of the body
- Formulae used for bike fitting are not optimal as they do not take into account individual anthropometrics nor riding style
- Comfort and individual anthropometrics of cyclists need to be taken into account when performing bike fitting
- Greater hamstring flexibility allows for a better position on the bicycle for increased performance
- Increased hamstring flexibility is associated with an increased handlebar drop
- Increased handlebar drop is associated with greater performance
- For endurance steady state cycling, bike fitting can be performed as per previous recommendations as the cycling position and muscle activity remain constant before fatigue is reached
- With increasing cycling intensity, joint kinematics and EMG magnitudes alter
- It is important to consider the training or racing intensity when performing a bike fit in order to maximise cyclist's full body position and muscle activity
- Pelvis, lumbar and thoracic kinematics should be assessed during bike fitting

- Guidelines for saddle choice should take cycling intensity into account

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