THE EFFECTS OF CLEAT LOCATION ON MUSCLE RECRUITMENT STRATEGIES OF CYCLING

by

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(Under the Direction of KATHY SIMPSON)

ABSTRACT

Introduction: Foot placement may play an important role in muscle recruitment patterns that may affect cycling performance. **Objective:** The purpose of this study was to determine if muscular activity of the thigh and leg muscles shifted when a more posterior, compared to traditional, cleat location is used. **Methods**: Surface electromyography (sEMG) and kinematic data (1200 Hz and 120 fps, respectively) were collected from eleven (11) experienced cyclists performing at a constant pedal rate (80rpm) for two cleat location conditions: neutral (NTL) and posterior (POS). sEMG was obtained for eight (8) muscles. Ergometer positions were altered to maintain consistent kinematics between conditions. Root mean square (RMS) data for sEMG were analyzed using paired t-tests for each muscle. **Results**: RMS-EMG burst magnitude and RMS-EMG peak as a function of crank position were effected cleat conditions ($p < 0.05$).

INDEX WORDS: CYCLING, CLEAT, KINEMATICS, ELECTROMYOGRAPHY

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OF CYCLING

by

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DECLARATION

The work presented in this thesis is, to the best of my knowledge and belief, original, except as acknowledged in the text, and the material has not been submitted, either in whole or in part, for a degree at this or any other university.

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NOMENCLATURE

VO_2– rate of oxygen consumption

EMG – electromyography

sEMG – surface electromyography

RMS – root mean square

RMS-EMG – root mean square of electromyographic signal

RMS-EMG_{max}- maximum value for root mean square of electromyographic signal

RMS-EMG_{peak}- peak value, as a function of crank angle, for root mean square of electromyographic signal

SOL – soleus

MGA – medial gastrocnemius

LGA – lateral gastrocnemius

TA – tibialis anterior

VMO – vastus medialis oblique

VL – vastus lateralis

BF – biceps femoris

GM – gluteus maximus

CHAPTER 1

INTRODUCTION

Background

Cycling began as an outlet for entertainment with James Starley's invention of the mechanically driven 'safety' bicycle in 1885 (Wilson, 2004). Due to the hand-built nature and subsequent limitations of availability, only the most financially exclusive individuals had access. Subsequently, modern assembly technology has enabled mass-production bicycles to reach the entire spectrum of the socio-economic ladder. With eventual worldwide reach, the bicycle transitioned from a source of amusement for the elite to serving as a modern day resource to accommodate a multitude of daily duties, such as transportation and exercise.

The sport continues to be a popular mode of exercise for recreational, amateur and professional cyclists and subsequently creates substantial financial opportunity for many companies to capitalize on the possibility of performance gains through new products. The United States' cycling industry alone sold over 18 million adult bicycles in 2008 (Bicycle Retailer and Industry News, 2008). According to Sports Business Research Network approximately 38.1 million Americans over the age of seven rode a bicycle in 2009 with 4.3 million of them partaking in the activity greater than 110 times in the year (Sport Business Research Network, 2009).

This has resulted in a substantial growth of the retail bicycle industry over the past three decades. In 2008, via specialty retailers, American cyclists spent six billion dollars purchasing bicycles, related equipment and accessories (Bicycle Retailer and

Industry News, 2008). To maintain or improve sales growth, manufacturers must find methods to improve a cyclist's performance via their equipment.

Of great importance to any cyclist are equipment features aimed at performance enhancement. While cycling equipment and components that are lighter in weight and made with higher-quality materials are known to facilitate a more enjoyable experience for the rider, there are many unanswered questions about how these components affect the human/bike interface in regards to anatomical and biomechanical parameters associated with cycling.

As shown in Figure 1, integrated pedal systems, or clipless pedals, are one example of such a performance-driven accessory. In theory, the primary goal of integrated pedal systems was to improve the overall effectiveness of the pedaling motion. The introduction of the first commercially available pedal system established a widely accepted (three-hole) cleat-mounting standard. Clipless-pedal systems require the use of a cycling-specific shoe that requires a cleat to be fixed to the bottom of a rigid-soled shoe. When pressed together firmly, the cleat on the bottom side of the shoe mechanically locks atop the pedal. The fixed cleat allows for only minor adjustment in anterior/posterior, medio/lateral and rotational directions.

Figure 1. Clipless pedal design requires the use of cleats and cleat mounting hardware.

Today, several manufacturers produce integrated systems and shoes; consequently, modern cycling shoes have threaded holes in the sole in various arrangements, dependent on brand, or desired purpose. Despite minor variations, the necessary threading to accept cleat hardware is consistently located underneath an anatomical landmark that was thought to be the most appropriate, that is, in the region of the third metatarsophalangeal joint (MPJ).

However, whether the third MPJ is the most optimal location for performance effectiveness is not known. My investigation into the history of this standardized approach yielded very little data to support this as the most suitable location for cleat placement (Ericson, 1986; Litzenberger, et al., 2008; Mandroukas, 1990; Van Sickle & Hull, 2007). To date, no lay or scientific literature clearly identifies foundations supporting the current cleat location standards of integrated pedal systems as the most appropriate.

However, I believe that there may be a better cleat location for two reasons: the current cleat standard promotes a higher likelihood for injury (Gregor, et al., 1987; Gregor & Wheeler, 1994; Hockenbury, 1999; Pruitt & Matheny, 2006) and decreases performance (Cannon, et al., 2007; Ericson & Nisell, 1988; Gonzalez & Hull, 1989; Korff, et al., 2007; Litzenberger, et al., 2008; Mandroukas, 1990; Mornieux, et al., 2008; Too, 1990; Van Sickle & Hull, 2007). For this study, performance effectiveness is the main focus, although the results also have indirect implications for injury prevention/causation.

I believe performance effectiveness is not optimal for the current cleat standard because I predict that it requires greater force production by the posterior musculature of the lower leg than is necessary to pedal effectively. I predict a more posterior cleat location, compared to the current standard, decreases posterior musculature activity of the lower leg. A decrease in activity of the lower leg could come from a reduction in eccentric, isometric, or co-contractions necessary to facilitate pedaling. For traditional cleat placements, researchers found that during pedaling, the medial and lateral gastrocnemius act primarily to stabilize the ankle joint. Meanwhile, the prime mover, the soleus, contributes to generating positive angular momentum during the power phase (Raasch, et al., 1997; Zajac, et al., 2002). Sanderson and colleagues (Sanderson, et al., 2006), however, claimed that a posterior cleat placement reduces what they termed “moments of opposition” from muscles, increasing “muscular unison”. The researchers stated that, during the pedaling motion, the soleus was acting eccentrically, while the gastrocnemius acted concentrically, indicating a “moment of opposition” within the triceps surae complex. Ideally, muscles will work in “unison” throughout the cycle pedal motion, which was found to be the case with posterior cleat locations.

In terms of injury prevention/causation, long-term effects of a highly repetitious motion like cycling exacerbate the importance of synergistic behavior of muscles. Overuse injuries are believed to commonly result from long-term exposure to these moments of opposition (Gregor & Wheeler, 1994). Moreover, as the pedal/shoe interface is the point of contact that distributes work done by the legs to the bicycle, the aforementioned overuse injuries can often be attributed directly to maladjustment of the cleat. One remedy for many of these conditions is a more posterior cleat location (Pruitt & Matheny, 2006). Van Sickle (Van Sickle, 2007) observed that posterior cleat locations (55% total foot length), decreased muscle force demand on the triceps surae by approximately 65%. This was deduced to be due to reduced ankle extension moment (Van Sickle & Hull, 2007).

Compared to more posterior cleat locations, the current standard cleat location creates a longer lever arm about the ankle, as seen in Figure 2. This moment leads to significant stress in the Achilles' tendon and unnecessary muscular work from the triceps surae (Gregor et al., 1987; Mademli, et al., 2009).

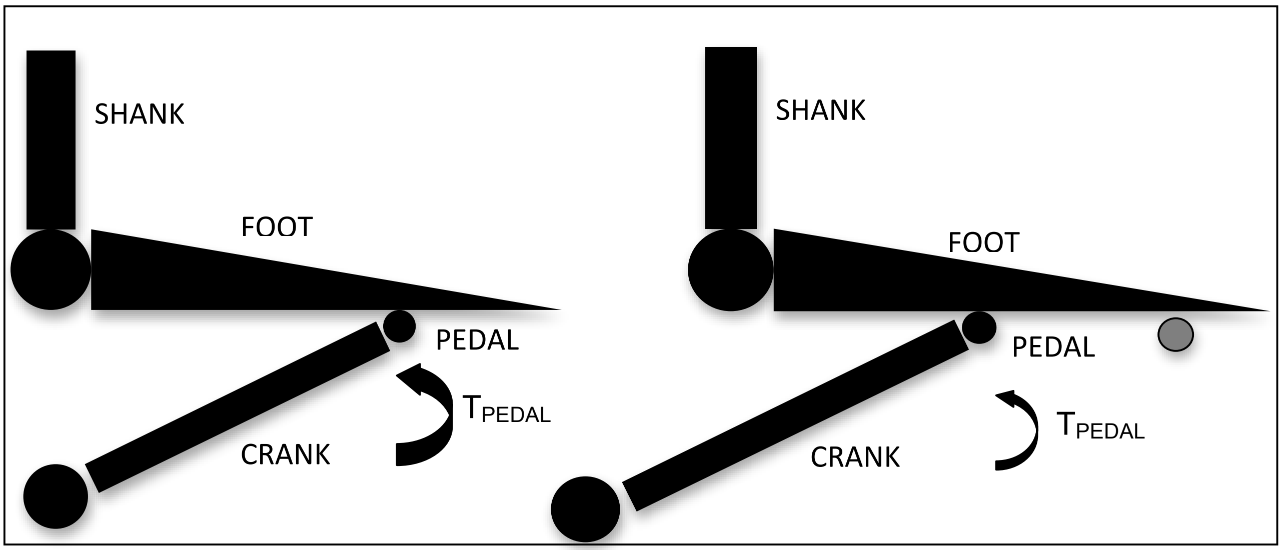

Figure 2.Physical description of neutral (NTL) (left) and posterior (POS) (right) cleat locations and resulting pedal torque.

Injury reduction and prevention are important to cyclists. Reports estimate that nearly 85% of all recreational cyclists will experience an overuse injury and 36% of those require medical attention (Wilber, et al., 1995). The most common of these injuries are: metatarsalgia, patellar tendonitis, and Achilles' tendonitis (Asplund & St Pierre, 2004; Hockenbury, 1999; Mellion, 1991). As cycling is a non-impact sport, most injuries sustained while participating are a result of faulty pedaling mechanics (Farrell, et al., 2003; Gregor & Wheeler, 1994; Mellion, 1991; Pruitt & Matheny, 2006; Sanner & O'Halloran, 2000).

Determination of optimal cleat location could eliminate the cleat/pedal interface from contributing to common cycling injuries and promote greater pedaling effectiveness. I believe that shoe manufacturers are, by locating the cleat hardware under the third MPJ, inadvertently increasing the force required by the posterior musculature of the lower leg during the pedaling stroke.

Purpose of the Study

Therefore, determining the role cleat location has on muscular recruitment patterns of the lower extremity is of great significance. Thus, the primary purpose of this study was to determine if moving the cleat location from the standard, neutral position (NTL) to a more posterior (POS) cleat position would change the electromyographic activity of muscles of the lower limbs demonstrated while cycling in a seated position. As explained previously, in general, I predicted that the POS compared to the NTL cleat location would decrease activity of the triceps surae and increase activity of the quadriceps, hamstring, and gluteal muscle groups.

Hypotheses

For the magnitude of peak muscular activation (root-mean square electromyogram; "RMS-EMG") ($RMS\text{-}EMG_{max}$) the POS compared to NTL cleat placement would display:

a. Decreased triceps surae (soleus [SOL], medial and lateral gastrocnemius [MGA and LGA, respectively]) RMS-EMG.
b. Increased RMS-EMG of the thigh muscles (gluteus maximus [GM], vastus lateralis [VL], vastus medialis [VMO], biceps femoris [BF]), and tibialis anterior (TA).

For the timing of peak muscular activation as a function of crank angle ($RMS\text{-}EMG_{peak}$) the POS, compared to NTL, cleat would display:

a. Later in the crank cycle for MGA and LGA

For the POS condition, compared to NTL, temporal activation patterns would exhibit:

a. Delayed activation for SOL, MGA and LGA
b. Delayed activation for BF, VMO, VL and GM

Significance of the Study

Little investigation has taken place regarding the isolated role cleat location has on muscular activity during cycling (Ericson, 1986; Litzenberger, et al., 2008; Mandroukas, 1990; Van Sickle & Hull, 2007). In a recent review of original research articles, investigators stated a lack of research involving the isolation of the shoe-pedal interface (Hug & Dorel, 2009). However, it is apparent that performance enhancement and the prevention of overuse injuries are both related to cleat locations (Ericson, 1986; Gregor & Wheeler, 1994; Litzenberger, et al., 2008; Pruitt & Matheny, 2006; Van Sickle

& Hull, 2007). The effect various cleat conditions have on the kinematics of the pedal stroke can aid in the understanding of how cleat location can be optimized, in terms of overuse injury manifestations and muscle recruitment patterns. By isolating cleat location, the biomechanical principles associated with the pedal stroke can be viewed exclusively as a result of various cleat conditions, in order to obtain the most optimal cleat location.

Aside from the injury prevention and rehabilitation implications this research has, there are also details that potentially affect the design of both cycling shoes and pedals for the bicycle industry. Previous research regarding equipment performance has been conducted to investigate other contact points of the bicycle, such as saddle and handlebar positions and shapes (Barratt, et al., 2011; Bressel, et al., 2009; Carpes, et al., 2009). These data drive the bicycle industry to spend millions of dollars to develop new, or improve upon, existing designs of cycling equipment while obtaining only marginal results. Meanwhile, rarely have these companies pursued the anatomical or biomechanical interface, and surely none have isolated exclusively the role of the cleat/pedal interface to investigate muscular recruitment patterns. This research will incorporate variables that are important to the positioning of the rider, and thus the resulting performance attributes. By investigating the relationship between cleat location and muscular activity, current industry standards regarding this equipment and its application may be altered.

Limitations

Based on participation requirements, subjects were cyclists that were experienced with clipless pedal systems, and participate in the sport of cycling for a minimum of eight hours a month. As there are known differences between novice and expert riders as well as triathletes and cyclists, regarding muscular activity of pedaling, these results can only be applied to an expert category of cyclist (Candotti et al., 2007; Chapman, et al., 2007; Chapman, et al., 2008; Chapman, et al., 2009; Hug, et al., 2008; Korff et al., 2007). Also, participants will be exposed to a stationary bicycle void of any lateral sway in a laboratory environment, which has known biomechanical differences from overground, outdoor pedaling, limiting the generalizability of the findings to similar environments (Bertucci, et al., 2007). Thus, the ability to apply the findings to populations and environments that do not meet these standards is limited. Finally, crank arm length was the same for every participant and is known to affect muscle recruitment (Barratt et al., 2011; Hug & Dorel, 2009; Martin & Spirduso, 2001). However, crank arm length was the same for both cleat conditions.

Assumptions

Despite unfamiliarity with the POS cleat condition, I believed that participants were able to produce consistent muscular recruitment patterns, as demonstrated in earlier pilot testing. Furthermore, potentially new muscle recruitment strategies associated with POS cleat condition were not obstructed by a lack of neurological stimulation to muscles groups not typically targeted during the pedal motion.

Summary

Locating the cleats under the metatarsal heads has been the industry standard since the earliest clipless pedal and shoe designs were introduced to cyclists. Nearly four decades have passed without knowing if this truly is the most appropriate position for injury prevention and performance enhancement. This current standard location may be responsible, in part, for overuse injuries of the lower leg and increase the metabolic energy necessary to perform pedaling (Ericson, 1986; Gregor & Wheeler, 1994; Litzenberger, et al., 2008; Pruitt & Matheny, 2006; Van Sickle & Hull, 2007). Investigating the role cleat location has on the muscular recruitment patterns of the leg will aid in determining whether a more posterior cleat location is more optimal for improving performance and reducing injury potential. Based upon the findings of this research, new shoe and pedal designs may result.

CHAPTER 2

REVIEW OF LITERATURE

Introduction

The cleat/pedal interface serves as one of the three contact points between rider and bicycle and is solely responsible for distribution of energy between rider and bicycle. Understanding the parameters associated with this interface requires determining the biomechanical and muscular components involved with the pedal motion. Previous researchers have investigated extensively the role individual muscles play throughout the pedal stroke (Baum & Li, 2003; Bieuzen, et al., 2007; Brown, et al., 1996; Cannon et al., 2007; Chapman, et al., 2008; Chapman et al., 2009; Cruz & Bankoff, 2001; Dingwell, et al., 2008; Dorel, et al., 2009; Dorel, et al., 2009; Ericson, et al., 1985; Ericson, et al., 1986; Ericson, 1988; Gregor, et al., 1991; Hug, et al., 2003; Hug, et al., 2006; Hug, et al., 2004; Hug et al., 2008; Hug & Dorel, 2009; Jorge & Hull, 1986; L. Li & Caldwell, 1998; L. Li & Baum, 2004; L. Li, 2004; Litzenberger, et al., 2008; MacIntosh, et al., 2000; Prilutsky & Gregory, 2000; Raasch et al., 1997; Ryan & Gregor, 1992; Sanderson et al., 2006; Suzuki, et al., 1982; Wakeling & Horn, 2009). However, knowledge regarding the specific effects cleat location has on pedaling is much less developed (Litzenberger, et al., 2008; Mandroukas, 1990; Pruitt & Matheny, 2006; Van Sickle & Hull, 2007). The following review of literature will explore the biomechanical features associated with bicycle pedaling, activation and roles of specific leg musculature; and cleat location and muscle recruitment.

Cycling Biomechanics

Despite being constrained in a circular trajectory within a mostly sagittal plane, the cycling pedal motion is quite complex (Burke & Newsom, 1988). The most common biomechanical model for cycling treats the lower extremities for the cycling motion as a five-bar linkage (Hug & Dorel, 2009; Hull & Gonzalez, 1988). Components of this system include the skeletal structures of the pelvis, femur, tibia/fibula complex, and the many bones of the foot. The mechanical junction of the rider/bicycle interface, the pedal/crank combination, serves as the final mechanical joint associated with this linkage model (Hull & Gonzalez, 1988; Redfield & Hull, 1986). The primary objective of this lower extremity model during cycling is to generate and transfer power from the rider to the mechanical equipment of the bicycle, resulting in forward motion of the cycle (Candotti et al., 2007; Gregor & Wheeler, 1994).

From a mechanical standpoint, a crank rotating about a fixed, central axis dictates the bicycle pedal motion. The resulting trajectory of the foot/pedal interface is commonly observed by dividing the movement into two phases, based on muscular activity levels of thigh and lower leg muscles. This two-phase model identifies power- and recovery-phases and denotes the transitional period between the two with top-dead-center (TDC) and bottom-dead-center (BDC) identifiers (Burke & Newsom, 1988; Ericson, 1986; Faria, 1984; Gonzalez & Hull, 1989; Wilson, 2004). As the rider's limbs are in anti-phase with one another, the TDC and BDC identifiers are most commonly assigned to the side of the bicycle that contains the drivetrain components and are associated with 0° and 180° positions during the cycle, respectively (Burke & Newsom, 1988; Gonzalez & Hull, 1989). As the nomenclature denotes, the power-phase is the period of time where the leg generates and distributes power to the pedals, while the purpose of the recovery-phase serves to recuperate from this energy expenditure and to return the crank to the top position (Brown et al., 1996; Childers, et al., 2009; Coyle et al., 1991).

The power phase is the period of extensor force production where the rider must overcome the various sources of resistance to promote forward momentum of the bicycle (Childers et al., 2009; Wilson, 2004). Force generation associated with the power phase begins just after TDC and ends at BDC with peak force production occurring at approximately 110° (Sanderson, et al., 2000). Despite the rapid decay of force production just after the peak, the limbs continue to generate small amounts of positive impulse until reaching DBC (Mornieux et al., 2008; Sanderson et al., 2000).

As the power phase is occurring in one limb, the recovery phase of the pedal stroke is occurring in the other. The recovery phase is characterized by the relaxation of previously contracting extensors and subsequent contraction of hip and knee flexors (Ericson, 1988; Gregor et al., 1991; Hug et al., 2008; Jorge & Hull, 1986; Mornieux et al., 2008; Prilutsky & Gregory, 2000; Sanderson et al., 2000). Because the legs are in anti-phase with one another, the cyclist must not only overcome environmental factors such as wind resistance, gradient, and mass associated with the bicycle, but also must overcome the resistance associated with the motion of the opposing limb (Cruz & Bankoff, 2001; Mornieux et al., 2008; Wilson, 2004). Thus, the pedal cycle is comprised of overcoming both external and internal resistance factors to generate positive angular work (Wilson, 2004).

To accomplish the task of developing positive angular impulse at the crank, cyclists' must produce force in a meaningful capacity. Pedal force, and subsequently crank force, can be measured by force-sensing equipment capable of identifying the radial and tangential components during the power and recovery phases (Caldwell & Li, 2000; Kautz & Hull, 1993; Koninckx, et al., 2008; D. J. Sanderson & Black, 2003; Stapelfeldt, et al., 2007). Under ideal circumstances the cyclic trajectory of the pedaling motion has force development perpendicular, or tangential, to the crank arm at all positions throughout the pedal motion (Korff et al., 2007; Mornieux et al., 2008; Schmidt, et al., 2003; Stapelfeldt et al., 2007). Thus, the overall mechanical effectiveness of the pedal motion is determined by the ability of the rider to minimize the radial force while maximizing tangential force. Moreover, a simultaneous decrease in the negative angular impulse associated with the recovery phase is also beneficial to the development of mechanically efficient pedaling (Korff et al., 2007; Mornieux et al., 2008; Schmidt et al., 2003).

A commonly used technique to examine basic parameters of cycling is the index of force effectiveness. The index of force effectiveness is a measure of the ratio of positive to negative angular impulse by evaluating the 3-dimensional force applied to the pedal (Boyd, et al., 1997; Cannon et al., 2007; Korff et al., 2007; Stapelfeldt et al., 2007). Increasing the mechanical effectiveness of pedaling originates from equipment selection, pedal rate, fatigue and conscious attention to the pedal task (Cruz & Bankoff, 2001; Jorge & Hull, 1986; Korff et al., 2007; Mornieux et al., 2008; Sanderson & Black, 2003). During separate investigations Mornieux (2008), Cruz (2001) and Jorge (1985), et al., found that the selection of clipless versus toe-clip pedal influenced the mechanical effectiveness of the pedaling motion by increasing the development of tangential force, and subsequently the index of force effectiveness, with the use of clipless pedals. Tangential pedal force development is also influenced by the pedal rate, with the greatest index of mechanical effectiveness coming from pedal rates ranging from 60-80rpm (Sanderson et al., 2000). Finally, researchers utilizing force-measuring pedals equipped with digitized visual feedback observed a decrease in negative angular impulse when cyclists were consciously aware of the tangential application of force, resulting in a 57% increase in effectiveness (Mornieux et al., 2008). Sanderson et al. (2002) claim that during prolonged steady-state bouts of cycling, fatigue induced an increased peak positive angular impulse during the power phase as a result of an increase in the negative angular impulse during the recovery phase, resulting in a decrease in overall mechanical effectiveness.

Although the pedal motion is limited to the sagittal plane, the techniques required by the cyclist to generate and apply force to the bicycle in a meaningful manner are quite complex. However, the constraints of movement within a single plane, and subsequent minimization of degrees of freedom promote the development of very predictable motor programs (Bernstein, 1967). An outcome associated with the development of stable motor programs is the ability to measure the repetitious nature of muscle activity associated with the pedal motion.

Muscle Activity of the Pedal Cycle

The musculature involved during the bicycle pedal motion relies on very precise coordination because no one muscle can execute all required biomechanical functions (Brown et al., 1996; Ericson, 1988; Sanderson et al., 2000; Sanderson et al., 2006; Ting & McKay, 2007). In order for cyclists to develop and optimally apply force to the mechanical features of the bicycle proper organization of muscle recruitment is necessary (Kautz & Hull, 1993; Korff et al., 2007; Laplaud, et al., 2006). Moreover, orientation of the direction of force is required to increase the index of effectiveness (Laplaud et al., 2006). Appropriate timing of muscle activation to direct loads on each joint, the transfer of energy between joints, and finally the distribution of energy upon the mechanical apparatus of the bicycle are all required to accommodate the complex nature of the pedal motion (Brown et al., 1996; Burke, 2002; Hasson, et al., 2008).

Sagittal plane motion, dictated by the structural components of the bicycle, requires at least eight lower limb muscles to synchronize activation for both timing and intensity. Individual muscles associated with the five-bar linkage model lower limb include: hip extensors gluteus maximus (GM) and biceps femoris (BF); extensors of the knee, vastus lateralis (VL), rectus femoris (RF), and vastus medialis oblique (VMO); plantarflexors of the foot, soleus (SOL), medial and lateral gastrocnemius (MGA and LGA, respectively); and dorsiflexor of the foot, tibialis anterior (TA) (Burke & Newsom, 1988; Cruz & Bankoff, 2001; Ericson, et al., 1985; Ericson, 1988; Hug & Dorel, 2009; Laplaud et al., 2006).

The functional role of each muscle is of great importance to accomplish common pedal strategy (Childers et al., 2009; Coyle et al., 1991; Dorel et al., 2009). GM is a single-joint hip extensor that serves as a major power producer during the power phase of the pedal cycle (Childers et al., 2009; Dorel, et al., 2008; Hug & Dorel, 2009; Jorge & Hull, 1986). As a two-joint hip extensor and knee flexor BF has extensor activation patterns during the power phase and flexor activity at DBC to facilitate transition to the recovery phase (Childers et al., 2009; Hug & Dorel, 2009; Jorge & Hull, 1986). The quadriceps group, consistent of VM and VL are single-joint knee extensors responsible for the majority of the power production during the power phase between TDC and BDC (Childers et al., 2009; Hug & Dorel, 2009; Jorge & Hull, 1986). Contributing to knee extension during the power phase, the two-joint nature of RF also aids in hip flexion during the recovery phase (Childers et al., 2009; Hug & Dorel, 2009; Jorge & Hull, 1986). MGA and LGA are responsible for both flexing the knee just before DBC to promote transition from power to recovery phases as well as eccentric plantarflexion and stabilization of the ankle complex during the power phase via co-contraction (Childers et al., 2009; Hug & Dorel, 2009; Jorge & Hull, 1986). To accompany MGA and LGA, SOL is a single-joint muscle that is largely responsible for the force generation in plantarflexion of the foot (Childers et al., 2009; Hug & Dorel, 2009; Jorge & Hull, 1986). TA is a single-joint muscle that facilitates dorsiflexion of the foot during the recovery phase and also aids SOL, MGA and LGA in ankle stabilization during the power phase through eccentric, isometric and co-contractions (Childers et al., 2009; Hug & Dorel, 2009; Jorge & Hull, 1986).

Other important anatomical considerations of muscle are the pennation angle and the location of muscle insertion relative to a joint-center (Neptune & Kautz, 2001; Wakeling & Horn, 2009). Wakeling, et al. (2009), state that as the pennation angle of a muscle increases, a predisposition towards greater force development exists. Research also indicates that if a muscle inserts further away from a joint it is capable of generating greater force, but moves the joint through a smaller range of motion for a given muscle contraction (Wakeling & Horn, 2009). Therefore individual muscles from within a group may possess architectural differences that predispose them to different contributions during various motions. As stated previously, the complexity of the pedal motion requires not only suitable force development but also force direction, which is commonly associated with the differences of articulation of muscles (Elmer, Ba et al., 2011; Gregor et al., 1987; Hasson et al., 2008; Hull & Gonzalez, 1988).

According to researchers, the number of joints a muscle crosses is also an important consideration accountable for the unique performance of specific muscles (Hasson et al., 2008; Kautz & Hull, 1993; O'Brien, 1991; Sanderson et al., 2006). The muscles of the five-bar linkage model are of either a mono- or bi-articular nature, meaning they cross either one or two joints, respectively (Redfield & Hull, 1986). The mono-articular muscles associated with cycling are: TA, SOL, VL, VM and GM; and bi-articular muscles: MG, LG, BF, and RF. Studies have shown that with complex motions, such as cycling, mono-articular muscles primarily generate energy and perform mechanical work, whereas the bi-articulate muscles function to transfer and aid in force direction at critical times throughout the pedal cycle (Hautier et al., 2000; Mileva & Turner, 2003; Sanderson et al., 2006). Mileva, et al. (2003), claim that the lines of action of mono-articular muscles are directed more or less lengthwise, while bi-articular muscles often have considerable transverse component, causing a differentiation in anatomical function. Therefore, to support necessary adaptations in muscle activity during obligatory alterations in movement mechanics, control of muscle groups based upon their biomechanical role of mono- or bi-articular nature is essential (Mileva & Turner, 2003). Architectural roles of muscles facilitate synergistic activity within the five-bar linkage model to maximize positive angular impulse during the power phase while minimizing negative angular impulse during the recovery phase (Ting & McKay, 2007; Zajac et al., 2002).

The muscle synergies of cycling express predictable patterns of activation, across multiple muscles, to execute a common function (Ting & McKay, 2007). Without synergistic behavior, the leg would fail to form proper movement patterns matching those required by the mechanical apparatus of the bicycle (Hug et al., 2004; Ryan & Gregor, 1992). During the propulsive phase of pedaling, several agonist/antagonist muscle pairs activate together. Often these patterns arise due to requirements of both torque generation about a joint and torque necessary to establish the direction of force on a pedal. According to van Ingen Schenau, et al. (van Ingen Schenau, et al., 1992), co-activation of mono-articular agonists and their bi-articular antagonists appears to provide a unique solution to the conflicting requirements often present during the pedal cycle. Moreover, co-contraction of agonist/antagonist groupings may also provide joint stability by reducing bone displacement and subsequent rotations.

Muscle activity of the lower limb shows very high day-to-day reproducibility regarding intensity of activation (Chapman, et al., 2008; Dorel et al., 2008; Hug & Dorel, 2009; Ryan & Gregor, 1992). Dorel, et al. (2008) also demonstrated good intra-session repeatability of ten lower limb muscle activation patterns during cycling, both in terms of intensity and temporal activation. However, several factors can alter the sequential order and activation levels of muscle activation (Ting & McKay, 2007). Seat height, pedal rate, power output, shoe-pedal interface and fatigue are all known to alter the predictability of recruitment patterns (Ericson et al., 1985; Hug et al., 2004; Hug & Dorel, 2009; Koninckx et al., 2008; Martin & Spirduso, 2001; Neptune, et al., 1997; Sanderson et al., 2000; Sanderson & Black, 2003; Sarre, et al., 2005; Ting & McKay, 2007). Alterations in these variables change joint angles, muscle moments, and thus length-tension and force-velocity relationships of muscles, resulting in differences in force production (Too, 1990). The number of joints a muscle crosses also contributes to the predictability of pedaling patterns, with the variability of bi-articulate muscles significantly higher than mono-articular (Johnston, 2007; Neptune & Kautz, 2001; Ryan & Gregor, 1992).

Saddle height is defined as the distance between the center of rotation of the crank and the top of the saddle. Disagreement between researchers regarding the most economical saddle height has encouraged significant exploration. Unfortunately, agreement on muscle activation intensity and timing has failed to reach consistent conclusions due to inconsistency regarding power output, pedal rate and methods for determining saddle height (Bini, et al., 2010; Carpes et al., 2009; Ericson et al., 1986; Hug & Dorel, 2009; Sanderson et al., 2006; Sanderson & Amoroso, 2009). However, modifications in activation patterns due to saddle height manipulation are observed in all muscles associated with the five-bar linkage model for pedaling (Ericson et al., 1985; Hull & Gonzalez, 1988; Jorge & Hull, 1986).

Of the factors known to contribute to alterations in muscle activation patterns, pedal rate, or cadence, is the most sensitive (Bieuzen et al., 2007; Gonzalez & Hull, 1989; Marsh & Martin, 1995; Prilutsky & Gregory, 2000; Prilutsky, 2000; Samozino, et al., 2007; Sarre & Lepers, 2007). The most predictable patterns arise from approximately 60rpm pedal rates, with a linear shift in peak power to earlier points in the pedal cycle observed up to 120rpm (Marsh & Martin, 1995; Martin & Spirduso, 2001; Sanderson et al., 2006; Sarre & Lepers, 2007). It is hypothesized that the earlier activation is necessary to compensate for the electro-mechanical delay (EMD) associated with rapid muscle activation demands. In order for the active muscle to produce power at optimal crank angles, physiological adaptations in EMD promote earlier activations in linear fashion up to 120rpm (Neptune et al., 1997). Sarre, et al. also claimed muscle activation patterns in bi-articular muscles show double bursts in EMG output at pedal rate greater than 100rpm (Sarre & Lepers, 2007). Despite various reports of inconsistent behavior of individual muscle intensity and timing, reoccurring results regarding angular impulse are present. Typically, a decrease in positive angular impulse during the power phase, and increased negative impulse during the recovery phase is associated with increasing pedal rates (Sanderson et al., 2006). Thus, extremely high pedal rates seem to negatively affect the pedaling index of effectiveness. As a final artifact of alterations in pedal rate, Sanderson (2006) found the ankle joint to become significantly more plantarflexed and exhibit a decreased range of motion, with the knee becoming significantly less extended, during bouts of pedaling exceeding 100rpm.

Power output has shown to influence muscle activation intensity (Bigland-Ritchie & Woods, 1984; Hug et al., 2003; Laplaud et al., 2006; Lucia, et al., 2003) but not significantly impact muscle activation timing (Jorge & Hull, 1986). Because power output is an artifact of both instantaneous force production and pedal rate, the following will focus exclusively on research where pedal rate was constant during exposure to varying workload conditions. Progressive pedaling tests performed have shown linear relationships between the RMS and workload levels (Bigland-Ritchie & Woods, 1984). Other research has shown a non-linear increase of RMS after a certain workload was reached (Hug et al., 2003; Lucia, et al., 2004). Despite the disagreement on RMS response to increased workloads, investigations failed to dissociate the influence fatigue and power output have on EMG activity level, as fatigue is known to positively influence EMG activity (Bigland-Ritchie & Woods, 1984; Hug et al., 2003; Hug et al., 2006; Lucia et al., 2004). Sanderson, et al. (Sanderson et al., 2000; Sanderson & Black, 2003) found increases in power output demands elicited a decrease in peak negative force. Thus, as power demands increased, cyclists improve the effective application of force by decreasing the amount of force demanded of the propulsive leg to overcome the recovery leg. Despite substantial consistency in the literature, a clear increase in EMG activation with constant-load bouts performed at different intensities is witnessed in all eight muscles associated with the five-bar linkage model (Ericson et al., 1986; Sarre et al., 2005).

Traditional pedals only allow the application of an effective force during the power phase, while clipless style pedal systems theoretically promote positive angular impulse during the recovery phase (Burke & Newsom, 1988; Cruz & Bankoff, 2001; Koninckx et al., 2008; Mornieux et al., 2008). While the various studies that have investigated this interface have found variations in EMG intensity, they draw inconsistent conclusions and fail to report alterations in timing. Crucial to the link between EMG activity and the shoe/pedal interface is the timing of activation, especially in bi-articular muscles. Also critical to muscular responses of the lower limb is the anterior and posterior positions of foot placement over the pedal. Van Sickle, et al. (2007) found that with foot positions posterior to those traditionally used, a 27% and 54% decrease in plantarflexor activity was found with mid- and rear-foot positions, respectively. Although it is often difficult to draw definitive conclusions about the effects associated with the shoe/pedal interface has on EMG activity, there is clear evidence that there is a correlation and any future research need take this into consideration (Mornieux et al., 2008).

Fatigue can be defined as an exercise-induced decrease in the ability to generate muscle force, regardless of whether or not the task can be sustained (Bigland-Ritchie & Woods, 1984; Dingwell et al., 2008). In cycling studies where athletes participate in fatigue-inducing bouts, EMG activity progressively increases until exhaustion (Dingwell et al., 2008; Sarre et al., 2005). For example, Hettinga, et al. (2000) measured EMG activity in BF and VL during a 4000m cycling time-trial and found decreased mechanical output during the latter stages, accompanied by and increase in RMS-EMG. This increase in EMG amplitude is associated with an increase in motor unit recruitment, as compensation for the decrease in force of contraction occurring in fatigued muscle fibers (Dingwell et al., 2008; Hug et al., 2006). An increase in firing rate and/or synchronization of muscle activity has also been attributed to adaptations in EMG activity (Housh et al., 2000). Researchers have stated that fatigue-induced EMD also influences activation timing (Cavanagh & Komi, 1979; Dingwell et al., 2008; Knaflitz & Molinari, 2003; Laplaud et al., 2006; Sarre& Lepers, 2007). Thus, changes in EMG activity during fatigue can be attributed to increased motor unit recruitment or variations in coordination strategies within muscle groups. However, due to limitations of EMG it is difficult to dissociate the effects of neuromuscular fatigue and the supposed changes in coordination patterns of the leg. Further research is needed to clarify the influence of fatigue on timing and coordination of the lower limbs during cycling.

Exploration of Cleat Location

The shoe/pedal interface is one of three contact points between rider and bicycle and is solely responsible for the transmission of energy between the rider and bicycle. Despite the importance of the shoe/pedal interface, to date there are very little data from which consistent correlation can be observed, between the location of cleats and the resulting muscle recruitment strategies (Hug & Dorel, 2009). Of the investigations that have taken place, a lack of consistency among researchers throughout much of the methodology leaves little room to form valuable conclusions. These inconsistencies result in a lack of understanding of the role cleat location has on temporal and intensity features of muscle contractions. Despite a lack of consistent findings among researchers, each investigation holds merit in the aid of future research design.

Ericson, et al. (1986), Mandroukas, et al. (1990), Litzenberger, et al. (2008), Leib (2008) and Van Sickle (2007) each investigated the effects of cleat location on muscular activity. However, the level of detail in reporting of methods, regarding the alterations in rider position between various cleat locations, is inconsistent and in some cases lacking altogether. In each of these studies, where claims are made regarding the affects of kinematics on muscle recruitment strategies, researchers failed to clearly document how the rider's kinematics were controlled for when cleat location was altered. These inconsistent methods for rider positioning are clearly an obstacle, as the effects of kinematics on muscular activity are well documented (Bertucci et al., 2007; Bini, et al., 2011; Brown et al., 1996; Chapman et al., 2008; Dorel et al., 2009; Hug & Dorel, 2009; Litzenberger, et al., 2008; Martin & Spirduso, 2001).

Van Sickle, et al. (2007) and Leib (2008) both investigated metabolic responses to posterior cleat locations, of which Van Sickle's findings are often cited as evidence that posterior cleat locations elicit no beneficial physiological response. However, Van Sickle et al., used a workload assignment of 90% LT that will often elicit excess post-exercise oxygen consumption, decreasing the likelihood for differences in VO_2 between conditions in repeated testing. Furthermore, Van Sickle utilized a cross-sectional design, where a longitudinal study may be more beneficial in understanding these effects, due to potential adaptation in muscular recruitment and subsequent physiological conditioning. Leib (2008) also studied metabolic responses of posterior and traditional cleat locations but defined posterior cleat location as the center of the calcaneus, which is an extreme and unrealistic cleat position.

Finally, Mandroukas, et al. (1990) and Litzenberger, et al. (2008) both investigated the physiological effects of cleat location, yet neither clearly identified the methods used to determine posterior cleat location. Mandroukas, et al. stated that traditional cleat location was defined by the second metatarsus joint, with posterior locations approximately 10cm aft. Furthermore, the researcher did not state how this was adapted for variations in rider foot length and only reported rate of perceived exertion (RPE). Litzenberger, et al. utilized the tarsometatarsal joints to identify the posterior position, but did not clearly state how this location was identified for the rider, as these joints are not clearly identifiable without radiographic technology. In either case, while researchers claim utilizing posterior cleat locations to elicit responses, the methodology and results among investigations is too disparate to deduce meaningful relationships.

CHAPTER 3

METHODS

Design

Within-subject, repeated measures with cleat position as the independent variable.

Participants

Eleven experienced male cyclists were recruited from the University of Georgia Cycling Team and the local cycling community. All participants satisfied the following inclusionary criteria: was a healthy male or female; age was between 18-44; familiar with clipless pedal systems; and cycled a minimum of 8 hours per month. A potential participant was excluded from the study if the participant had any of the following exclusionary criteria: current or chronic injury potentially affecting the participant's performance or safety; had previous lower extremity injury requiring medical attention which limits the range of motion for any lower limb joints; had any problems with balance not remedied; had any illness or medical condition that could negatively affect performance or safety; self-reported any symptoms that would potentially affect the participant's performance or safety, including: discomfort, pain, light-headedness, dizziness, fainting and/or nausea, or recent asthma episode; or physical activity level rated as "low" based on score on the Physical Activity Questionnaire. After initial recruitment, participants were informed of procedures, potential risks, and benefits associated with study participation. Once the recruit agreed to participate, informed consent forms were signed, in accordance with the University of Georgia Institutional Review Board.

Cleat positions

Two cleat locations were tested. For the neutral (NTL) cleat location, the cleat was placed at a midpoint of the longitudinal difference between the first and fifth metatarsophalangeal joints (MPJ), such that the NTL cleat location lay beneath the third MPJ. The cleat for the posterior (POS) cleat condition was located ½ the distance between the NTL position and the posterior edge of the calcaneus. Van Sickle and Hull (Van Sickle & Hull, 2007) also utilized this alternative cleat location. To accommodate the POS cleat location, research-specific shoes (2010 Specialized Comp™, Specialized Bicycle Components - Morgan Hill, CA) were modified by drilling holes in locations necessary to obtain proper cleat placement, based on anatomical landmarks of each participant's unshod foot. Details of participant measurements and corresponding cleat locations can be seen in Figure 3 and Figure 4.

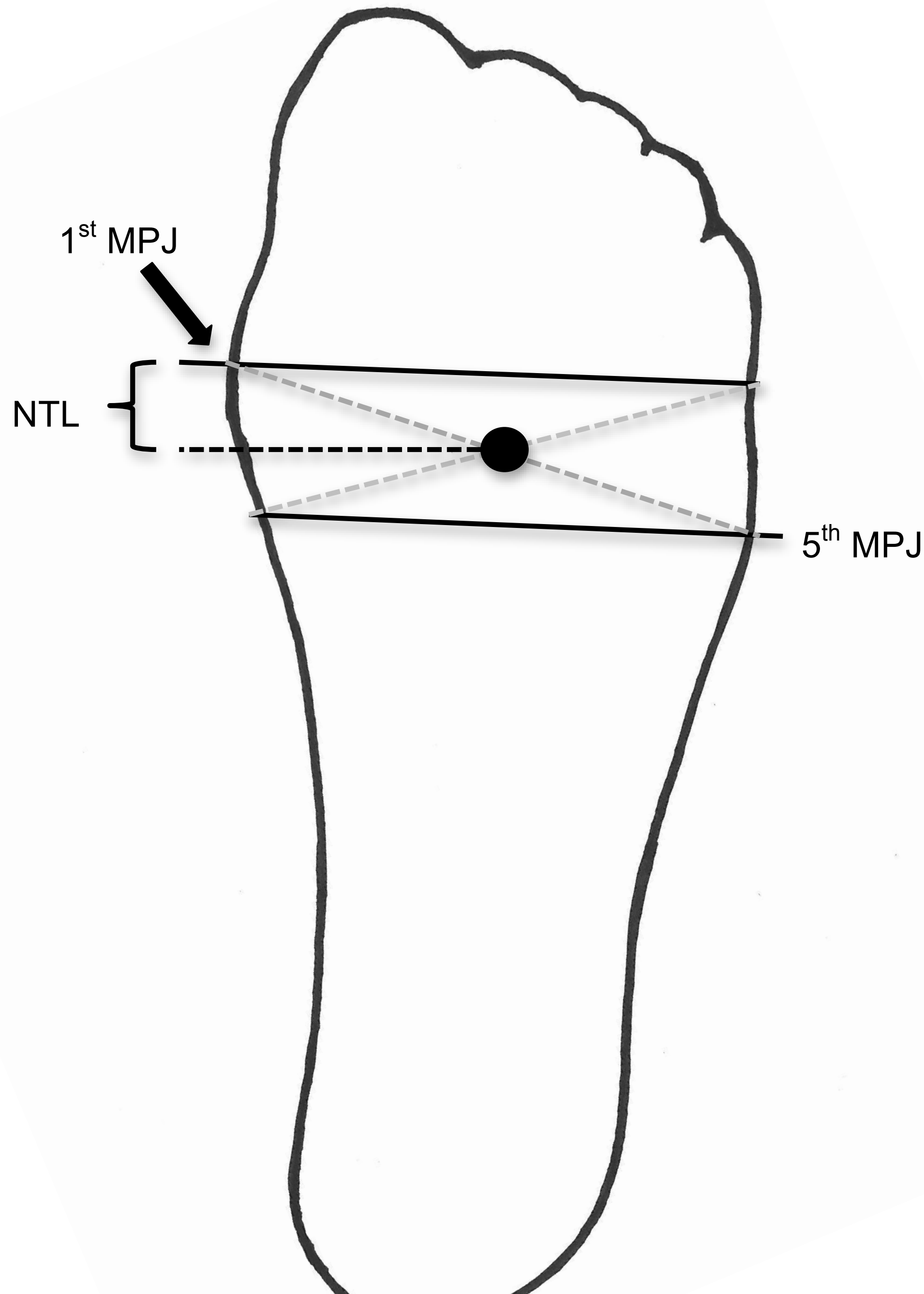

Figure 3. Identification of NTL cleat location – 1st and 5th MPJ landmarks are used to determine approximate location for 3rd MPJ

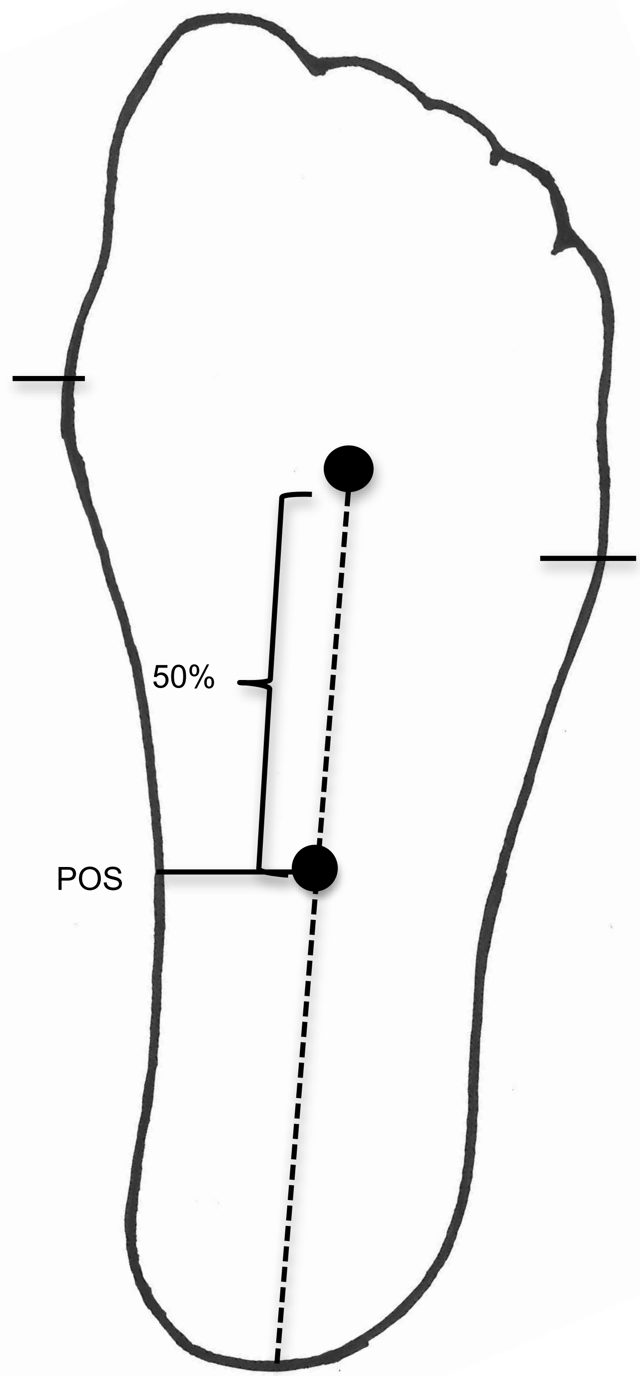

Figure 4. Identification of POS cleat location – 50% distance of NTL to posterior edge of calcaneus.

Experimental Procedures

The participant came to two test sessions. The first session consisted of fitting the shoe and cycle to the participant. The participant performed cycling tasks during the second session while EMG and kinematic data were obtained.

Preparatory and VO_2 max test session: After appropriate documentation was completed, anatomical features of each foot were recorded in order to properly position the cleats. To do this, a tracing of the border of each foot, and locations of the first and fifth metatarsophalangeal joints (MPJ) were identified while the subject was standing in typical bilateral weight-bearing stance. Subjects tried on modified shoes to determine proper sizing. Once the correct shoe was chosen, cleats for NTL and posterior POS conditions were placed on the bottom of each shoe at the locations previously described.

In order to maintain similar cycling positions and motions between cleat conditions, and to reproduce the cyclist's natural body positioning as best as possible, various measurements were taken. These aspects of cycle configuration were adjusted to best duplicate riding posture: seat height, seat antero-posterior position (fore-aft), handlebar vertical drop and horizontal reach, and knee joint angle. Each subject, for both cleat conditions, was positioned with the following measurements: 30° knee flexion at point of terminal extension; anterior aspect of patella located vertically over the 3rd MPJ with cranks in horizontal position; absolute trunk angle 40° from the horizontal; and relative shoulder joint angle of 90°. All measurements were recorded for both cleat positions, as seen in Figure 5.

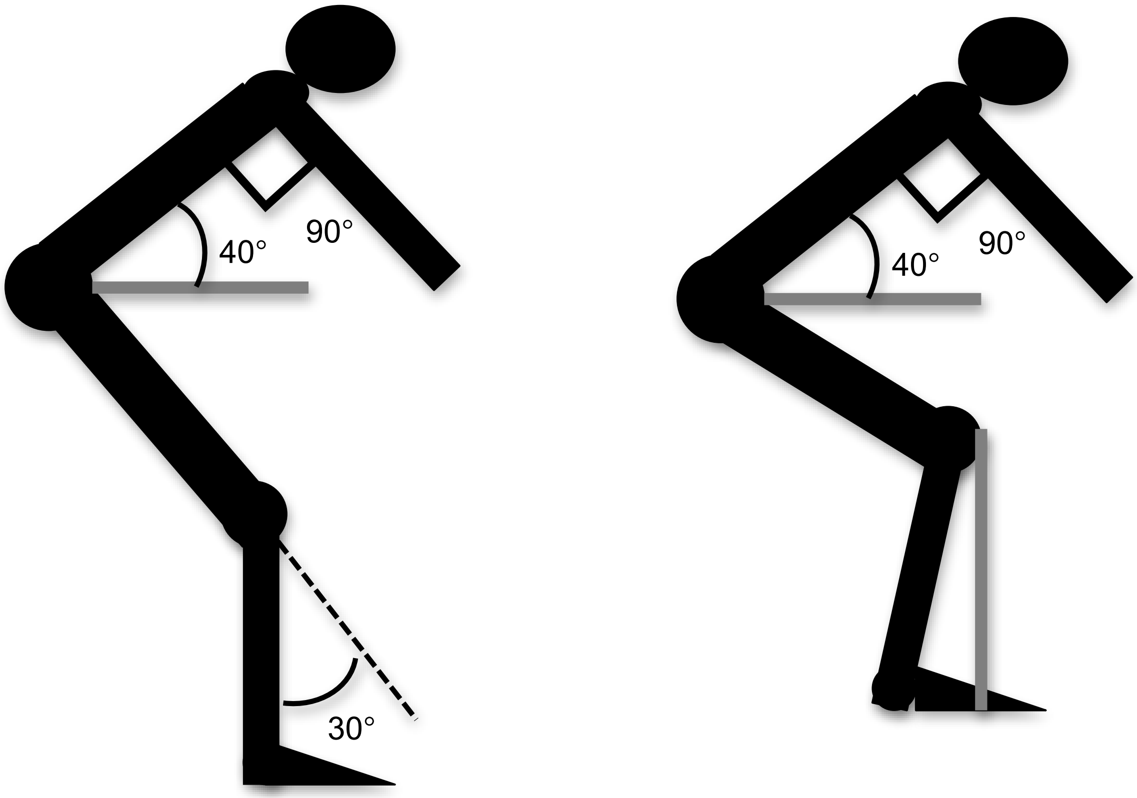

Figure 5. Rider position for NTL and POS. For saddle height, crank positioned to elicit maximal knee joint extension of 30° (left). Saddle fore/aft is set with anterior aspect of patella directly over 3rd MPJ with crank in forward-horizontal position (right).

Next, at the UGA Biomechanics Laboratory, participants underwent a maximally-graded exercise test (VO_{2max}) during cycling. First, subjects were informed of the protocol design of the maximally graded exercise test (VO_{2max}) and allowed to ask any questions or voice any concerns regarding the maximal effort. Participants then changed into cycling apparel consisting of spandex (or similar) shorts and a form-fitting shirt and had body mass and height measurements recorded. Next, once properly attired, the cycle ergometer (Lode Excalibur; Lode, Groningen, Nederland) was adjusted to obtain the body positioning described above with the participant's own footwear and cleat configuration. This was done for the participant's comfort during exhaustive exercise. Once the ergometer was properly adjusted, subjects were fitted with the necessary headgear and heart-rate monitor to complete a VO_{2max} test. Utilizing a TrueMax 2400™ metabolic cart (ParvoMedics, Salt Lake City, UT) participants warmed up with a self-selected pedal rate for five minutes at 100W while gas exchange was recorded at 15-second intervals.

For the test, following the warm-up, a ramp protocol of 1W every 2 seconds was used to elicit a voluntary, maximal effort. Completion of the test was determined by either voluntary cessation by the cyclist, the pedal rate fell below 50rpm, or a clear decrease or plateau in oxygen uptake was displayed despite an increase in workload. After termination of the test, participants were allowed to cool-down at a preferred workload.

The test was considered successful if there was an increase in VE/VO_2 with a non-concomitant increase in VE/VCO_2 (Wasserman, et al., 1994), a respiratory exchange ratio (RER) of ≥1.1 was achieved, and the subject demonstrated a heart rate equal to or greater than their age-predicted maximum (ACSM, 2010). The TrueMax 2400 proprietary algorithm to detect the ventilatory threshold elicited during each test was utilized for each VO_{2max} test. Finally, the second data collection session was scheduled for 3-8 days post-metabolic test.

Kinematic and EMG data collection session: Participant preparation: Prior to engaging in any procedures, participants were asked to verify their current medical- and health-status to ensure that they had remained healthy and injury-free. Similar to the initial testing session, participants wore cycling-specific clothing. Anthropometric measures of the ankle and knee width, leg length, and body mass and height were recorded.

Participants were then affixed with 16 pairs of surface EMG bi-polar electrodes using a 16-channel Myopac system (RUN Technologies MPRD101-Receiver/Decoder, Laguna Hills, CA; sampling rate = 1200; CMRR = 90db min. @ 60 Hz). The muscles monitored were the soleus (SOL), medial gastrocnemius (MGA), lateral gastrocnemius (LGA), tibialis anterior (TA), biceps femoris (BF), vastus lateralis (VL), vastus medialis oblique (VMO), and gluteus maximus (GM) on each leg, with the tibial tuberosity serving as the location for the common ground electrode (Bieuzen et al., 2007; Hug et al., 2004). Electrode placement was done according to the guidelines of Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) (Hermens, et al., 2000). Although both legs were monitored, only the right limb will be reported on for this study. The site for each electrode was first palpated, shaved to remove body hair, lightly abraded using course gauze to remove dead skin cells, and swabbed with alcohol to remove oils and lotions. After allowing the alcohol to dry, pairs of bipolar Ag-AgCl electrodes (Vinyl 1-3/8" Biopac Systems EL503 Electrodes - Goleta, CA) were placed on previously-specified muscle locations using a 2cm inter-electrode distance (Beck et al., 2009; Cram & Kasman, 1998; Farina, 2006; Hermens et al., 2000; Hug & Dorel, 2009; Mademli et al., 2004; Malek, et al., 2006; Mercer, et al., 2006; Petrofsky, 1979; U.S. Department of Health and Human Services, 1992). Electrode leads were attached to the amplifier/data conditioning unit belt pack that was secured to the participant's thoracic area using a neoprene vest. Signals were sent, via fiber optic cable, to a patch panel, then the Vicon MX™net controller box. The MX net was used to synchronize the kinematic data using Vicon Workstation™ software (Vicon Corporation, Oxford, UK). EMG electrode placement and amplification was tested using a series of maximal voluntary isometric contractions. Gain settings were adjusted as necessary, per subject, to elicit optimal signal-to-noise ratios. Once EMG placement signal testing was completed, pre-wrap and athletic tape was used to secure electrode placement and lead wires to minimize signal artifact, with attention being paid to the future location of reflective markers and pedal motion. To obtain kinematic data, and subsequent temporal muscular activation pattern data, 17 reflective markers were placed as recommended by Vaughan and as used in the Vicon Plug-in-Gait™software module for the lower extremity. Locations included: toe (second metatarsal head); heel (calcaneus at same height as toe marker); ankle (lateral malleolus passing through transmalleolar axis); tibia (lower lateral 1/3 of shank, following plane of transmalleolar axis); knee joint center (lateral aspect of knee center-of-rotation); thigh (lower lateral 1/3 of thigh); anterior superior iliac spine (ASIS) (directly over anterior superior iliac spine); posterior superior iliac spine (PSIS)(directly over posterior superior iliac pine). To identify cycle period and cleat condition, two makers were placed on the lateral aspect of the right and left shoe, directly above the cleat location. For each cleat condition, a static calibration was completed with the participant standing with feet shoulder width apart and with the feet facing forward. This was done to later reconstruct the participant's natural standing position.

Instrumentation

High-speed digital video motion measurement system: An MX-40VICON™camera system (Vicon, Ltd., Oxford, UK), comprised of 7 high-speed (240 Hz), semi-permanently mounted digital-video cameras (visible-red light sensitive C-MOS photodiodes, 4.1 megapixel; exposure time = 1/1000 s) and Workstation™software, were used to capture the locations of the reflective markers on the participant's lower extremity. The rider-ergometer system was positioned centrally to the motion capture cameras. In conjunction with the Vicon™ motion capture system, a single digital video recorder was set in a medial-lateral position and was time-sequenced with the Vicon™ system to obtain video-capture information at 60Hz sampling rate.

Surface electromyography: A Myopac Surface Electromyography Unit (MYOPAC, Laguna Hills, CA) equipped with MPRD101-Reciever/Decoder and 27 Channel Amplifier/Encoder with Fiber Optic Transmitter was used to capture myoelectric activity from the participant's lower extremity. 32 Vinyl 1-3/8" Ag-AgCl Biopac Systems EL503 Electrodes (Biopac, Goleta, CA) were used in a bi-polar arrangement for 8 muscles per leg. SEMG sampling rate of 1200Hz was set consistently. EMG gain settings (2000 or 5000) were altered to accommodate proper signal-to-noise ratios and documented for each participant and specific location.

Metabolic measurement: TrueMax 2400™ metabolic cart (ParvoMedics, Salt Lake City, UT) was used to monitor VO_2 by gas exchange measurement.

Anthropometric equipment: Each lower extremity body segment length and circumference were measured using standard equipment (digital weight scale, sliding calipers and tape measure)

Protocol

Data Collection: Prior to the cycling testing, the participant was informed of the tasks and instructed to maintain the workload as shown by the visual display on the ergometer. The participant put on the cycling shoe selected during the first session, and the ergometer and cleat position were adjusted using the documentation from the first session.

For the testing, a total of four 10-minute trials were completed at 80 rpm for each workload-cleat combination. For each cleat condition, two workloads were performed: 50%VT followed by 85%VT. Only the 85% workload will be presented in the data reduction and results sections of this study, as variability associated with 50% VT was excessive. During each trial, at minutes 7,8,9 and 10 a 10-second sample of Vicon, EMG and digital video were all simultaneously obtained. Participants were not made aware of the intervals when data were being collected. Upon completion of each trial, participants were asked to dismount the bicycle while researchers made adjustments to the equipment, to either accurately assign a new position, or to blind the participant's anticipation of adjustments. To complete ergometer position changes and recalibrate the motion capture system, participants were allowed 8 minutes of rest between trials. The order of the two cleat positions were counterbalanced among the participants.

Data Reduction and Analysis

From each trial, 10 complete cycles of the right limb were selected for analysis. Although left and right limbs were monitored for EMG and VICON data, only the right limb data has been used for reporting. One cycle began at top-dead-center (TDC) and ended after one complete revolution. Data for all four trials were used; therefore, 40 cycles were analyzed. MATLAB™ (The MathWorks, Natick, MA) programs were written to obtain all quantities and all variables, with exceptions noted below for kinematic data reduction.

Kinematic Data Reduction: The raw marker locations were reconstructed into 3D coordinate data using the proprietary method in the Vicon Workstation® software (v. 4.3.1, Oxford-Metrics, Ltd), and smoothed using Woltring's (1985) generalized, cross-validation spline (GVSPL). Joint coordinate systems (Cardan's method) for the ankle, knee, and hip joints for the right limb were used to determine clinical joint angles. Maximum flexion/extension angles of the right lower extremity will be used to verify angular kinematic conditions are similar across all conditions.

EMG Data Reduction: Raw sEMG data from each muscle for each cleat condition was full-wave rectified and then filtered using 4th order bandpass Butterworth digital filter (30 to 200 Hz). Bandpass filter parameters were assigned as a result of pilot data frequency analysis. Root mean square (RMS) EMG (T = 50 ms, equivalent to 3.18 Hz low-pass filter) was generated, as that has been shown to be a data form most correlated to muscle force (Cram &Kasman, 1998). To determine the RMS threshold values needed to detect burst onset and offset, an ensemble of 40 pedal cycles was calculated. Threshold values were set at 20% of maximal RMS as seen in the literature (Hug & Dorel, 2009). Maximal RMS was identified as the peak value of muscle contraction. An RMS-EMG burst was considered to occur during an interval of time during which the RMS-EMG magnitude was equal to or greater than the threshold value. For each burst of a given muscle and pedal cycle, the magnitude of the activation was defined as the mean of the RMS-EMG displayed during the burst. Onset and offset times were expressed as a function of crank angle, identified by degree of rotation. To understand the role cleat location has on muscle recruitment of the leg, several dependent variables of muscular activity were explored. The magnitude of peak MG-RMS values were obtained for each muscle, as an indicator of increased or decreased muscle activity associated with a cleat condition.

Statistical analysis: Paired t-tests were used to test differences among the cleat conditions ($p < .05$) using mean scores between cleat conditions for the following parameters: maximal RMS-EMG; peak RMS-EMG as a function of crank angle; and RMS-EMG burst onset/offset times as a function of crank angle for all muscles. Confidence interval (CI) was 95% confidence was used to ascertain measurement precision. Effect size will be determined utilizing Cohen's *d.*

CHAPTER 4

RESULTS

The characteristics for the 11 participants are shown below (Table 1). As seen, age, body mass, height and VO2max ranges for participants were broad. Of particular interest is the range of experience, as this may have direct affects on the adopted pedaling technique during trials (Chapman, et al., 2008). Experience was determined using a questionnaire (Appendix B) where participants self-identified the number of years cycling.

Table 1. Participant characteristics

	Mean ± SD	Range
Age (yr)	28 ± 7	21 – 43
Body Mass (kg)	73 ± 11	60 – 96.5
Height (cm)	175 ± 6	165 – 186
Experience (yr)	9 ± 5	4 – 22
VO_{2max} (ml/kg/min)	55.5 ± 5.1	49.5 – 64.3

Magnitude of Maximal Muscular Activation

The descriptive data of the maximal muscular activity (RMS-EMG_{max}) for the neutral (NTL) and posterior (POS) conditions are presented in Figure 6. Table 2 shows the descriptives, 95%CI and statistical outcomes. T-test statistics demonstrated that, for the POS versus NTL cleat conditions, muscular activity was decreased for the triceps surae (SOL, LGA, MGA) and increased for GM. However, TA, VMO, VL, and BF did not display significantly different outcomes between NTL and POS conditions.

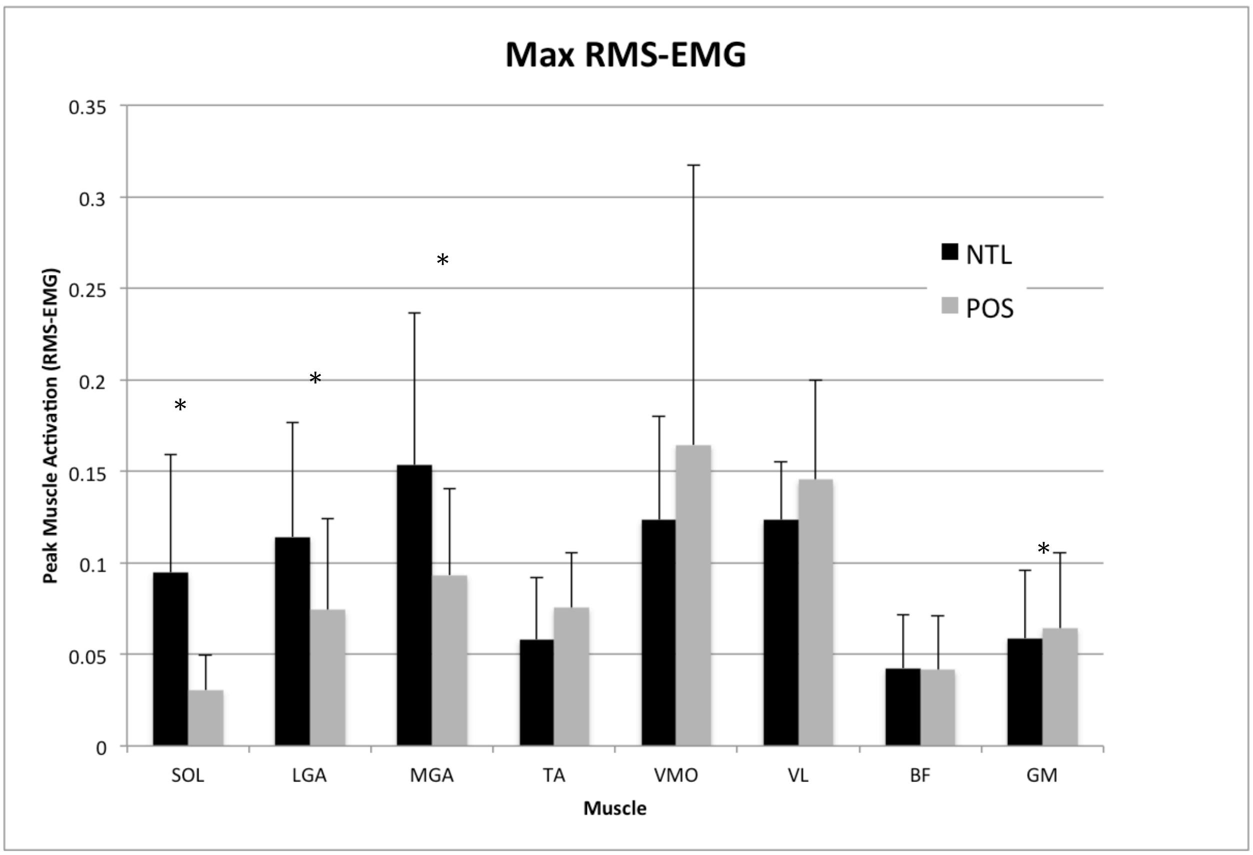

Figure 6. Group means for RMS-EMG_{max} for neutral (NTL) and posterior (POS) conditions for monitored muscles of the right limb. Asterisks represent statistical significance.

Table 2. Means (mV) and SD, and the 95% confidence intervals of maximal EMG-RMSmax values.

		Mean ± SD	95% CI	*p* value
SOL	NTL	0.095 ± 0.019	0.08 - 0.11	**0.009**
	POS	0.031 ± 0.006	0.03 - 0.03	
LGA	NTL	0.114 ± 0.063	0.08 - 0.15	**0.002**
	POS	0.074 ± 0.050	0.04 - 0.1	
MGA	NTL	0.153 ± 0.083	0.1 - 0.2	**0.013**
	POS	0.093 ± 0.047	0.07 - 0.12	
TA	NTL	0.058 ± 0.034	0.01 - 0.06	0.052
	POS	0.076 ± 0.030	0.06 - 0.09	
VMO	NTL	0.124 ± 0.056	0.09 - 0.16	0.221
	POS	0.164 ± 0.153	0 - 0.18	
VL	NTL	0.123 ± 0.032	0.1 - 0.14	0.063
	POS	0.145 ± 0.055	0.11 - 0.18	
BF	NTL	0.043 ± 0.029	0.08 - 0.12	0.817
	POS	0.042 ± 0.029	0.09 - 0.12	
GM	NTL	0.059 ± 0.037	0.04 - 0.08	**0.018**
	POS	0.064 ± 0.041	0.04 - 0.09	

Note: **Bold** *p* value = a significant t-test comparison ($p < .05$)

Peak Muscular Activation as a Function of Crank Angle

RMS values for peak muscle activation (RMS-EMG$_{peak}$) as a function of crank position during the power phase of the pedal cycle are presented in Figure 7. Statistically-significant outcomes are also shown. Group means, standard deviations, 95% confidence intervals and statistical outcomes are listed in Table 3. T-test statistics demonstrated that for the POS cleat condition, only vastus lateralis (VL) displayed a difference in timing of peak muscular activity, as a function of crank position. For VL, peak muscle contraction occurred 32.3° later in the crank cycle for the POS condition compared to NTL.

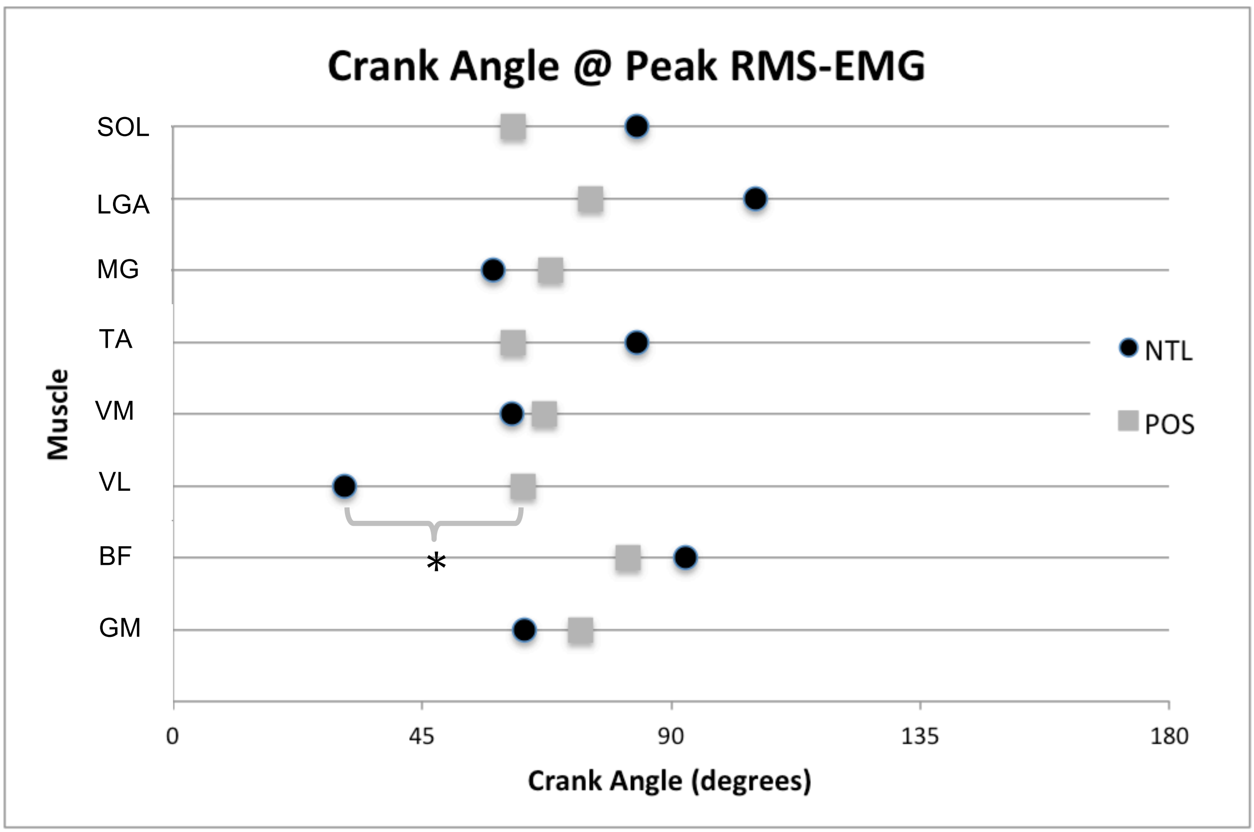

Figure 7. Peak muscular activity (RMS-EMG$_{peak}$) as a function of crank position during the power phase. An asterisk indicates statistical significance.

Table 3. Means and SD, and the 95% confidence intervals for neutral (NTL) and posterior (POS) peak EMG-RMS values as a function of crank angle.

		Mean ± SD	95% CI	*p* value
SOL	NTL	84.0 ± 9.4	78.6 - 89.5	0.23
	POS	61.4 ± 65.4	22.7 - 100.0	
LGA	NTL	105.2 ± 69.6	64.1 - 146.3	0.35
	POS	75.4 ± 63.9	37.6 - 113.2	
MGA	NTL	57.8 ± 24.0	43.6 - 72.0	0.43
	POS	68.3 ± 43.7	42.5 - 94.1	
TA	NTL	72.4 ± 123.4	0.0 - 145.3	0.59
	POS	44.3 ± 94.8	0.0 - 100.3	
VMO	NTL	61.2 ± 61.2	25.0 - 97.4	0.62
	POS	67.1 ± 33.6	47.3 - 86.9	
VL	NTL	31.0 ± 22.6	17.7 - 44.4	**0.03**
	POS	63.3 ± 41.8	38.6 - 87.9	
BF	NTL	92.6 ± 38.3	70.0 - 115.3	0.49
	POS	82.3 ± 57.1	48.6 - 115.9	
GM	NTL	63.6 ± 23.4	49.7 - 77.4	0.06
	POS	73.6 ± 24.2	59.3 - 87.9	

Note: **Bold** *p* value = a significant t-test comparison ($p < .05$).

RMS-EMG Burst On/Off Timing as a Function of Crank Angle

RMS-EMG burst on/off times, as a function of crank angle, during the pedal cycle are presented in Figure 8. Group means, standard deviations, 95% confidence intervals and statistical outcomes are listed in Table 4. For the posterior (POS) cleat condition, no muscles displayed a significant difference in muscular burst timing compared to neutral (NTL).

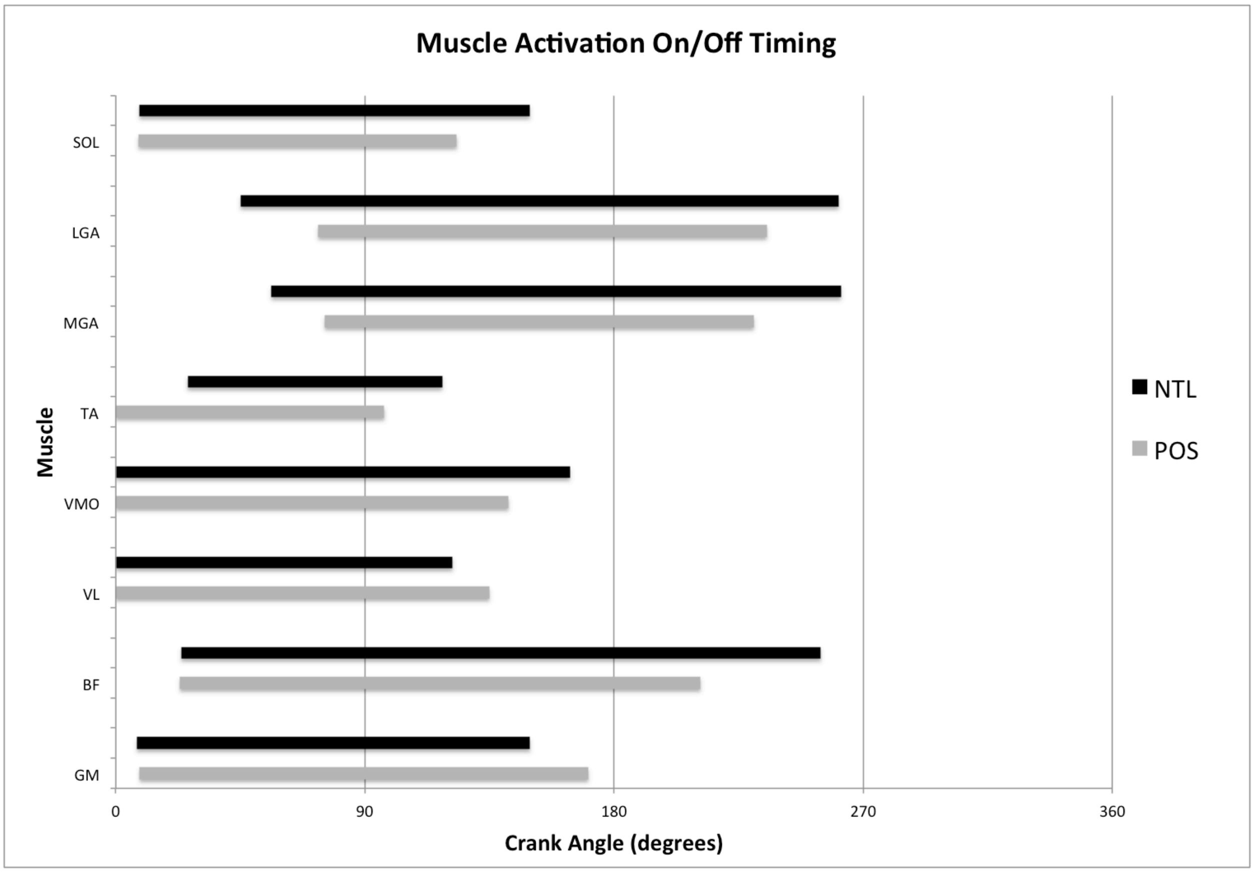

Figure 8. RMS-EMG burst on/off times as a function of crank angle for neutral (NTL) and posterior (POS) conditions.

Table 4. Means and SD, and the 95% confidence intervals of scores for neutral (NTL) and posterior (POS) RMS-EMG burst on/off values as a function of crank angle.

			Mean ± SD	95% CI	*p* value
SOL	ON	NTL	8.6 ± 9.1	3.2 - 13.9	0.96
		POS	8.2 ± 18.3	0.0 - 19.0	
	OFF	NTL	150.0 ± 19.1	138.7 - 161.3	0.49
		POS	122.9 ± 121.5	51.1 - 194.7	
LGA	ON	NTL	45.1 ± 23.4	31.3 - 58.9	0.11
		POS	73.1 ± 45.2	46.4 - 99.8	
	OFF	NTL	260.9 ± 39.1	237.8 - 284.1	0.49
		POS	234.7 ± 111.8	168.7 - 300.8	
MGA	ON	NTL	56.2 ± 22.6	42.8 - 69.6	0.23
		POS	75.5 ± 47.5	47.4 - 103.5	
	OFF	NTL	205.7 ± 70.2	164.3 - 247.2	0.49
		POS	230.2 ± 87.9	178.2 - 282.1	
TA	ON	NTL	25.9 ± 82.0	0.0 - 74.4	0.32
		POS	0.00	-	
	OFF	NTL	118.3 ± 129.5	41.8 - 194.8	0.68
		POS	97.2 ± 91.1	43.3 - 151.0	
VMO	ON	NTL	0.00	-	-
		POS	0.00	-	
	OFF	NTL	145.6 ± 49.1	116.7 - 174.6	0.77
		POS	141.8 ± 22.9	128.3 - 155.4	
VL	ON	NTL	0.00	-	-
		POS	0.00	-	
	OFF	NTL	121.6 ± 9.5	116 - 127.3	0.12
		POS	134.9 ± 26.6	119.2 - 150.6	
BF	ON	NTL	23.6 ± 32.4	4.5 - 42.8	0.95
		POS	23.2 ± 34.9	2.5 - 43.9	
	OFF	NTL	255.0 ± 83.4	205.7 - 304.3	0.11
		POS	210.6 ± 104.8	148.7 - 272.6	
GM	ON	NTL	7.6 ± 13.2	0.0 - 15.5	0.77
		POS	8.6 ± 11.7	1.7 - 15.6	
	OFF	NTL	150.5 ± 25.0	135.7 - 165.2	0.33
		POS	170.9 ± 63.6	133.4 - 208.5	

Note: **Bold** *p* value = a significant t-test comparison ($p < .05$)

Kinematic Data for Neutral and Posterior Cleat Conditions

Kinematic data are displayed in Table 5, including group means, standard deviations, and statistical outcomes. There were no significant differences in lower leg kinematics between cleat conditions.

Table 5. Means and SD, and the 95% confidence intervals of difference scores for neutral (NTL) and posterior (POS) kinematic flexion/extension rang of motion.

		Flexion/Extension Angle		Difference		95% CI		
		MIN	MAX	MIN	MAX	MIN	MAX	*p* Value
HIP	NTL	40.8	86.7	-2.0	-3.7	36.1 - 45.5	80.9 - 92.6	0.21
	POS	38.8	83.0			34.1 - 43.5	75.5 - 90.5	
KNEE	NTL	30.9	110.6	-2.0	-8.3	27.9 - 33.9	105.6 - 115.6	0.083
	POS	28.9	102.3			25.6 - 32.2	95.9 - 108.7	
ANKLE	NTL	-5.5	17.1	2.1	-1.5	9.25 - 1.75	14.7 to 19.5	0.068
	POS	-3.4	15.6			7.28 - 0.5	12.1 to 19.2	

Note: **Bold** *p* value = a significant t-test comparison ($p < .05$)

CHAPTER 5

DISCUSSION

The primary purpose of this study was to determine if moving the cleat location from the standard, neutral position (NTL) to a more posterior (POS) cleat position would change the electromyographic activity of muscles of the lower limbs demonstrated while cycling in a seated position. In contrast to the NTL cleat location, I predicted that the POS cleat location would decrease activity of the triceps surae (SOL, LGA, MGA) due to a shortened lever arm about the ankle joint. Subsequently, an increased activity of the thigh musculature (VMO, VL, BF, GM) was anticipated for POS in order to maintain consistent work output with decreased assistance of the triceps surae. In addition, I expected peak EMG-RMS values to occur later in the crank cycle for MGA and LGA with POS. The rationale behind this is anticipation that these muscles would change their primary role of action. As stated previously, the primary action for MGA and LGA during the power phase is ankle joint stabilization, by eccentric or isometric contraction. During the recovery phase the primary role of these muscles is to act as knee flexors, by concentric contraction. However, it is well documented that the large eccentric and isometric contractions during the power phase prevents properly timed concentric contraction at DBC, the transition from power phase to recovery phase. As a result, there is a loss of tangential force applied to the pedal. Therefore, I surmised that by minimizing the eccentric contraction during the power phase, a more properly timed concentric contraction would facilitate knee flexion, and thus more likely to induce positive angular work through this transition. Finally, for the POS versus the NTL condition, it was expected that temporal activation patterns (as a function of crank angle) of the triceps surae and thigh muscles would be delayed. The results of this study support some of these hypotheses and, therefore, some of the predicted justifications.

Magnitude of Maximal Muscular Activation

In regards to the magnitude of peak muscular activation (EMG-RMS_{max}), it was anticipated that POS would show decreased activity for the triceps surae (SOL, LGA, MGA) and increased activation for the thigh muscles tested (VMO, VL, BF, GM) during the power phase. Support of these predictions for EMG-RMS_{max} was mixed. During the power phase, the triceps surae demonstrated results consistent with predictions and with previous research (Ericson, 1986; Litzenberger, et al., 2008; Van Sickle & Hull, 2007). EMG-RMS_{max} was significantly decreased for SOL, MGA and LGA. However, EMG-RMS_{max} was not increased for all muscles of the thigh. While VMO and VL displayed a non-significant tendency of increased EMG-RMS_{max} activity, only GM showed increase of statistical significance. Meanwhile, BF did not increase EMG-RMS_{max} for POS, but rather decreased.

To understand the following potential explanations of the reduced triceps surae EMG-RMS_{max} during POS cleat cycling, assume that the foot is the system of interest, and the axis of rotation is the flexion/extension axis located within the talus. A 'moment' is defined as force x moment arm, whereby the moment arm is the perpendicular distance between the axis of rotation and the line of action of the force. For the moments acting on the foot about the ankle joint, there are the net ankle muscle moments, the resistance moment created by the resistance force of the pedal pushing against the cleat of the foot, and the moment created by the weight of the foot. All else equal, the moment created by the weight of the foot is similar between cleat conditions, and thus will not be considered in the following explanations.

Therefore, there are three possible explanations and/or interpretations presented for the triceps surae outcomes. First, it is feasible (although not provable with these data), that the resistance moment in the POS versus the NTL cleat position could be less. This is surmised to occur, as the POS cleat condition likely reduced the moment arm of the resistance force power phase. This explanation is consistent with previous research (Ericson, 1986; Litzenberger, et al., 2008; Van Sickle & Hull, 2007).

Second, if the resistance moment was less, then consequently, the net ankle muscle moment required could also be less, thereby requiring less triceps surae force. Less RMS-EMG may support (albeit indirectly) that less triceps muscle force was generated during the POS condition. This is suggested because RMS-EMG has been reported to be correlated with muscle force (assuming other factors affecting muscle force are held constant, e.g., contraction velocity and muscle length) (Burden & Bartlett, 1999; Farina, 2006).

Moreover, it is more likely that if reduced net ankle moments for the POS condition did occur, it was due to decreased muscle force and not muscle moment arms. This is suggested, as the ankle and knee joint kinematics were similar between cleat conditions.

The third possible explanation for reduced triceps activation for POS compared to NTL is that increased GM RMS-EMG activity compensated for any reduction in positive angular work done by the triceps surae. This finding is supported by outcomes of a previous research study that was similar to this study, in that a consistent workload was maintained for multiple cleat locations (Ericson et al., 1985). What did they find or say that supports this explanation beyond the fact that they also observed reduced triceps surae muscle force.

Some thigh extensor muscles, however, did not increase their activation as predicted to help counter decreased triceps surae activity. One explanation is that the increased GM RMS-EMG activity was sufficient to compensation of reduced triceps surae angular work. Another explanation is that other hip and knee extensor muscles increased their activation, but the EMG of these muscles were not obtained. As the vastus intermedius is known to produce substantial knee extensor moments (Zhang, 2003), this muscle could have contributed necessary work required to accomplish. However, as the location of this muscle is inferior to other muscles, sEMG could not be used.

Results from this study are similar to some of the equivocal findings of previous literature. When compared to traditional cleat locations, Ericson (1986), Litzenberger (2008) and Van Sickle (2007) determined that posterior cleat conditions elicit increased RMS-EMG activity from the thigh musculature and decreased activity from triceps surae. Litzenberger, et al. reported very little on magnitude of RMS-EMG. However, conclusions drawn by Litzenberger are consistent with these findings, in regard to triceps surae RMS-EMG magnitude decreasing with POS cleat conditions.

Other investigators observed some findings consistent with ours, but some differences, too. Van Sickle, et al. (2007), found no statistical support for changes in magnitude of RMS-EMG of thigh musculature with various cleat positions. However, their findings in regards to reduced triceps surae RMS-EMG with more posterior cleat positions are consistent with our results. Findings of Ericson, et al. (1986), also are consistent with our findings in regards to triceps surae. However, the exception was that of the MG magnitudes, where RMS-EMG did not decrease with more posterior cleat conditions.

Despite a lack of absolute agreement with previous research, I feel the results of this study follow similar patterns of muscle activation for both the upper and lower extremity musculature in regards to RMS-EMG$_{max}$ reported by investigators of previous research (Ericson, 1986; Litzenberger, Illes, Hren, Reichel, & and Sabo, 2008; Van Sickle & Hull, 2007).

At present, only three studies have been conducted where posterior cleat locations were utilized, while EMG was monitored (Ericson, 1986; Litzenberger, et al., 2008; Van Sickle & Hull, 2007). However, these studies each have their own limitations. Ericson (1986) reported intensity of sEMG activity, but did not include any temporal parameters. Furthermore, Ericson (1986) did not utilize participant-specific posterior cleat location. Litzenberger (2008) also did not utilize participant-specific cleat location or workloads, and only reported on the length of burst duration, but not on/off burst as a function of crank position. Finally, Van Sickle (2007) reported sEMG activity secondarily to metabolic responses, and did not include any temporal parameters of muscle activation. Therefore, this research serves to combine sEMG parameters valuable to a competitive cyclist and documentation necessary to replicate cleat and rider position.

Timing of Peak Muscular Activation

It was anticipated that RMS-EMG$_{peak}$ would occur later in the crank cycle for MGA and LGA during cycling with a POS versus a NTL cleat position. I predicted this because I surmised that the involved muscles would shift their activation levels to accomplish different mechanical goals for moving/stabilizing the foot-ankle complex during the transition from power phase to recovery phase. During NTL, the primary role of the predominant muscle group (triceps surae) is to stabilize the ankle-foot complex via eccentric/isometric actions (Dorel, et al., 2009; Elmer, et al., 2011; Ericson, et al., 1985; Gregor, et al., 1987). I hypothesize that during POS cycling, the gastrocnemius would serve as a secondary agonist at the knee joint, producing knee flexor torque to produce knee joint flexion motion.

However, the results were not consistent with these predicted outcomes. RMS-EMG_{peak} of LGA and MGA were not significantly different between NTL and POS. Furthermore, LGA RMS-EMG_{peak} occurred earlier in the crank cycle. Finally, not predicted, VL was the only muscle that did exhibit significant RMS-EMG_{peak} differences. The POS condition caused a shift of VL RMS-EMG_{peak} activity to (approximately 30°) later in the crank cycle.

Explanation for the lack of significant LGA and MGA finding like comes from sizeable intraparticipant variability, as shown by the large standard deviations for LGA and MGA were recruited. One likely reason for this variation is the burst patterns. For most participants, LGA and MGA show a single burst of activation that occurs between approximately 45° and 90° of the crank cycle. However, visual inspection of individual-participant ensemble graphs of RMS-EMG revealed that some participants displayed a second burst for these muscles between 160° and 200°.

Qualitatively, two interesting outcomes were observed for the participants who demonstrated a second burst. First, for those who displayed two bursts, often the first burst occurred approximately 20° later in the crank cycle. Second, these participants also appeared to display synergistic behavior with the BF, in that MGA and LGA work in conjunction with BF through the recovery phase to flex the knee, and these events typically occurred through the transition from the power phase to the recovery phase.

While not conclusive, and the proposed interpretations of these findings cannot be proven with these data, the data indicate that some participants did express tendencies of our predicted RMS-EMG$_{peak}$ behavior. This may be advantageous for producing greater moments to the crank arm-pedal-chainring. For participants who tended to shift the peak activation of the VL to later in the crank cycle with a more posterior cleat position, this likely indicates that the cyclist is being able to apply more of the resultant force in a perpendicular direction relative to the crank arm. Force produced in a perpendicular direction (compared to any other direction) relative to the crank arm produces the greatest moment (Koninckx, et al., 2008). Thus, the resulting force generation is more likely to contribute to increased positive angular work. Application of this finding is important to cyclists attempting to improve the amount of positive angular work throughout the pedal stroke.

RMS-EMG Burst On/Off Timing as a Function of Crank Angle

For the POS condition, it was expected that activation patterns relative to the crank angle of the triceps surae and thigh muscles would be delayed as compared to the NTL condition. For the triceps surae, reasoning behind these predictions is similar to those of RMS-EMG$_{peak}$. With the POS condition, a decrease in eccentric muscle activation of MGA and LGA during the power phase about the ankle should allow these muscles to act as concentric knee flexors during the transition from power to recovery phases (Sanderson et al., 2006). Furthermore, if the LGA and MGA become more active during the recovery phase, this can potentially lead to positive angular work being accomplished, increasing the mechanical energy of the pedal-crank-drivetrain system.

Thus, the contralateral leg would not have to produce as much force during this same time interval. As the contralateral limb is engaged in the power phase during this time, the relevant musculature of the contralateral limb could delay its activation, as less angular work would have to be generated to transition from recovery to power phases.

Results from this research did not support these hypotheses or corresponding rationales. Regarding the RMS-EMG burst on/off timing, a lack of statistical significance for any observed muscle prevents definitive conclusions. However, it is likely that the lack of significance is due, in part, to sizeable intraparticipant and/or interparticpant variation. As stated in a recent review, high variability for EMG has been commonly reported during cycling studies (Hug et al., 2008).

Differences of scales (time versus position) used among the studies to report activation timing, it is difficult to make such comparisons. Due to the inability to compare results, it is difficult to draw definitive conclusions regarding the muscle burst onset-offset or duration for this study. Commonly reported for RMS-EMG burst on/off timing, researchers report the percentage of the crank cycle a muscle is activated (Litzenberger, et al., 2008). While this can be advantageous to understand from a physiological perspective, what is of greater importance is a clear understanding of when muscles are active during the crank cycle, rather than the length of time. In this manner, researchers are better able to understand the role each muscle plays in generating positive angular work to the pedal/crank system, and investigating muscle mechanics that are sensitive to joint positioning.

Kinematic Data for Neutral and Posterior Cleat Conditions

Maintaining similar cycling kinematics between cleat conditions was important in order to reduce potential confounding factors related to muscle force generation, such as muscle length and velocity. If cycling kinematics were similar between cleat conditions, then the interpretation of the EMG activation magnitudes presented above may serve as a rough, indirect measure related to muscle force (Van Sickle & Hull, 2007). As stated in a recent review (Hug & Dorel, 2009), kinematic reporting is commonly overlooked in cycling literature. Because of this, it can be difficult to compare our results to such studies.

There were no significant differences in hip, knee or ankle range of motion between NTL and POS conditions. Lack of significance could have been due to a slight drop in statistical power for the kinematic results, as data for a single participant were missing. Moreover, lack of statistical significance does not equal similarity. However, based on the CI of each cleat condition, quantitative evidence supports a lack of statistical significance. Hence, it can be stated that preventative measures taken to re-create kinematics between cleat conditions was successful.

The results of this study are consistent to those of previous research. Currently, three studies have been conducted where posterior cleat locations were utilized, while EMG was monitored (Ericson, 1986; Litzenberger, et al., 2008; Van Sickle & Hull, 2007). Ericson (1986) reported intensity of sEMG activity, but did not include any temporal parameters. Furthermore, Ericson (1986) did not utilize participant-specific posterior cleat location. Litzenberger (2008) also did not utilize participant-specific cleat location or workloads, and only reported on the length of burst duration, but not on/off burst as a function of crank position. Finally, Van Sickle (2007) reported sEMG activity secondarily to metabolic responses, and did not include any temporal parameters of muscle activation.

Limitations

These results can only be applied to an expert category of male cyclists. Participants were cyclists that were experienced with clipless pedal systems, and they participate in the sport of cycling for a minimum of eight hours a month. As there are known differences between novice and expert riders, as well as between triathletes and cyclists, these outcomes are relevant only to this skill level of cyclists (Candotti et al., 2007; Chapman et al., 2007; Chapman, et al., 2008; Chapman et al., 2009; Hug et al., 2008; Korff et al., 2007). To minimize confounding effects of skill level and cycling technique, a homogenous group was desired. Moreover, novice riders don't frequently utilize clipless pedal systems. Additionally, these data are representative of competitive cyclists, who would most benefit from any performance enhancement stemming from these findings. Finally, there are no known data regarding gender differences during pedal tasks. As a result, it is assumed that all findings can be applied to the same category of female cyclist.

Also, participants performed the tasks on a stationary bicycle that may have produced outcomes different from those obtained for overground, outdoor pedaling. It is known that there are biomechanical differences between stationary and cycling in actual cycling conditions. Therefore, the cycling environment limits the generalizability of the findings to similar environments (Bertucci et al., 2007). Use of a stationary cycle, however, did facilitate the control of many confounding parameters known to contribute to alterations in muscle recruitment.

Additionally, crank arm length was the same for every participant and that may have affected muscle recruitment (Barratt et al., 2011; Hug & Dorel, 2009; Martin & Spirduso, 2001). Typically, each cyclist uses a crank arm length that is best for him/her, usually based on height. However, the biomechanical and physiological effects of a set crank arm length should have been similar across cleat conditions, as the crank arm length was the same for both cleat conditions.

Also, POS cleat conditions are considered novel and therefore may have contributed to the high interindividual and intraindividual variability of some muscle activation patterns. A pilot investigation was executed to determine adaptation period for pedaling in the novel cleat location and participants were instructed to pedal for seven minutes prior to data collection.

Last, the interpretations of this study regarding muscle force and moments are only speculations, as these quantities were not measured or estimated. Therefore, these explanations should be considered with great caution.

CHAPTER 6

SUMMARY AND CONCLUSIONS

Summary

The primary purpose of this study was to determine if moving the cleat location from the standard, neutral position (NTL) to a more posterior (POS) cleat position would change the electromyographic activity of muscles of the lower limbs demonstrated while cycling in a seated position. Of specific interest was determination of the maximal RMS-EMG magnitude, crank angle at peak RMS-EMG and the crank angles when muscle activation began and ended.

Eleven male participants volunteered for this study. Participants came to two test sessions. The first session consisted of fitting the shoe and cycle to the participant, a warmup and familiarization period and a VO_{2max} test. The participant performed cycling tasks during the second session while sEMG and kinematic data were obtained. The order of cleat conditions were counterbalanced among participants.

During the cycling task, the spatial locations of reflective markers located on the cyclist's shoe and shorts were captured for the pelvis and legs using digital video. sEMG signals also were obtained for eight muscles of each leg. Paired t-tests were used to test differences among the cleat conditions ($p < .05$) the following variables: maximal RMS-EMG magnitude; crank angles at which the RMS-EMG bursts began and ended; and the angle at which the peak RMS-EMG magnitude occurred.

The results of this study demonstrated that triceps surae muscles (SOL, LGA, MGA) RMS-EMG_{max} were decreased for the POS compared to the NTL condition as predicted. However, for the thigh muscles, only the RMS-EMG_{max} of GM increased for POS as predicted. Among the thigh, only VL changed the crank angle at which peak RMS-EMG occurred. This was not anticipated. Peak RMS-EMG occurred at an angle approximately 30° further in the crank cycle. Lack of differences in other variables was most likely due to high interindividual variability. Of interest, was the finding of some participants who displayed two bursts for LGA and MGA for POS condition.

Conclusions

Based on the results of this study, posterior cleat locations alter the magnitude and temporal muscular recruitment strategies of seated cycling when compared to neutral cleat placement. Outcomes from this research provide indications that there may be a benefit to a more posterior cleat condition. Therefore, there is a need to further investigate the physiological and biomechanical effects of cleat location before we know whether a more posterior position will be beneficial for long-term performance enhancement and injury prevention.

Recommendations

There are 4 recommendations:

1. Riders frequently had very large intra-individual variability for sEMG intensity of recruitment patterns. Moreover, there was evidence of two burst patterns for some. Therefore, future studies should utilize larger participant sample sizes, as well as collect more trials/participant.
2. Potentially contributing to the inter-individual variability of these results is the large range of participant experience. It would be beneficial for future research designs to utilize participants with more similar cycling experience.
3. Because posterior cleat locations are considered novel, it would be of great interest to understand how long-term adaptation affects the ability to produce sEMG activity with less variability. Longitudinal studies are needed to better understand the implications of posterior cleat locations.
4. Of great interest to competitive cyclist is mechanical power generation and the effectiveness of pedaling. The inclusion of force-monitoring pedals would provide additional information regarding the efficacy of alternative cleat locations. Therefore, future studies should incorporate pedals equipped with multi-axis force measurement systems.

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APPENDIX A

INFORMED CONSENT

INFORMED CONSENT FOR RESEARCH PARTICIPATION

I, __________________________ agree to participate in the research study entitled, "***The effects cleat positions on muscle recruitment patterns during seated cycling***", that is being conducted by Dr. Kathy Simpson (706-542-4385) and Mr. Thomas McDaniel (graduate student, 706-542-4132), Department of Kinesiology at the University of Georgia, Athens, GA. I understand that this participation is entirely voluntary; I can refuse to participate or withdraw my consent at anytime without penalty or loss of benefits to which I am otherwise entitled. I can have the results of my participation, to the extent that can be identified as mine returned to me, removed from the research records, or destroyed. Additionally, if I am ineligible or become ineligible during my participation in the study and am released from further involvement in the study by the researchers, there also are no penalties or financial charges. My decision to participate/not participate or to withdraw my consent at any time if I so choose, will in no way affect any current or future participation in research conducted at the University of Georgia Biomechanics Laboratory.

The purpose of this study is to investigate how muscles perform differently when the cleat location is moved in the forwards or backwards direction and how this affects pedal forces and cycling technique. We hope to determine how cleat position affects muscle torques, muscle activation patterns and joint kinematics (movements). Knowledge gained from this research will be the initial step towards determining an optimal cleat position to maximize performance and prevent injury.

To be eligible, I must be healthy, free from any leg, foot, back or neck pain, or injury having required major medical attention or surgery; and free from any medical condition that is a) not being treated successfully or b) not monitored/treated by a medical physician and/or c) may adversely affect my safety or performance.
<u>Note</u>. The researchersalso reserve the right, now or at any time during testing, to ask for additional medical clearance from my physician before further testing can occur if they believe that I may have a health condition/injury/impairment that could affect my ability to safely complete the tasks, or if I am unsure that any medical conditions, impairments or illnesses I have may affect my safety or performance.

My part in this study will last for approximately 1.5 – 2 hr.If I am eligible to continue participation, if feasible, all study procedures will be accomplished in one day. However, it also is possible or may be necessary to complete the tasks over multiple days.

The procedures are as follows: I will come to the University of Georgia Biomechanics Laboratory, and I will sign this informed consent form after having the procedures explained to me and any questions I have answered. Next, we will determine my eligibility to participate in the study. First, the researcher will review with me my answers on the confidential, current health status questionnaire that I completed before the test session began. This questionnaire contains my history of injury/disease and any known balance difficulties. The second questionnaire I will complete prior to testing is for informational purposes only to help us understand the physical activities I typically engage in. I will then fill out a physical activity questionnaire.

Certain measurements of my body dimensions, e.g., height, weight, will be made. Similar to how animations are made for movies and video games, I will have reflective markers placed on various locations on my skin and/or clothing. The locations of these markers will be captured during testing by digital motion cameras used in animation. These marker locations are used later to reconstruct the movements of my body and limbs. One regular video camera will be used to capture my movements to later help the researchers track the marker movements only if necessary. I will also select a pair of laboratory cycling shoes which best represent the fit of my own cycling specific shoes to wear during testing. Finally, I will have surface electromyography electrodes placed at various locations on my legs to monitor muscular activity while pedaling a stationary cycle. The electrodes are like a 'listening' device, as they pick up electrical signals that tell the muscle fibers within a muscle to contract; they do not emit any signals or electricity of their own. When relevant, I will have a researcher of the same gender apply electrodes to specific muscle areas.

For the cycling testing, I first will undergo a warm-up consisting of 10 minutes of pedaling at a workload of 125W for males and 75W for female (similar to a very light training ride). Next, I will pedal at a workload of 175W for males and 125W for females (similar to a moderate training ride) for 15 minutes at each of three different cleat positions. Before pedaling at a new cleat position, I will have a rest of approx. 5 min (or longer, if I wish) and the height of my saddle will be adjusted as needed. This part of the testing will take approximately 45 minutes. The markers will be removed, and if I wish, I can see an initial look at my movements, digital video files and/or some of my data.

Performing any physical activity has some inherent risk of injury. However, the potential risk of injury is minimal, as the cycling task involved is of moderate intensity and lasting much less time than a typical training ride. As I am healthy, with good physical functional capacity, and used to riding with cycling cleats, I am very unlikely to experience fatigue or a fall off of a stationary bike. However, to further reduce the risk of these problems, the researchers will have the following in place: a) to avoid fatigue, I am required to only pedal for 15 minutes at a time before resting; b)I will have a researcher be ready to steady me while getting off/on the bike and during cycling; c) I will have a designated researcher take care of me throughout the testing, including monitoring how I feel and watching me for any signs of discomfort or other problems; d) I will tell the researchers immediately if I feel any signs of discomfort, pain, dizziness or other physical symptoms that could influence my health and safety; and e) the researchers will stop testing immediately if any researcher believes that I not be able to perform the tasks safely or that I may be exhibiting symptoms of a physical problem.

Thus, I am informed that I am to tell the researchers immediately if I begin to experience any discomfort, pain, nausea, dizziness or other atypical symptoms. Testing will be stopped immediately, and the researchers and I will discuss whether the problem can be resolved immediately and testing can continue; if I should postpone testing until a later date; or testing will be terminated. Although unlikely, discomfort or muscle soreness in the legs may occur for a few days after participation. This muscle soreness is what is felt sometimes when starting new physical activity or increasing the intensity/amount of existing training. The researchers will exercise all reasonable care to protect me from harm as a result of my participation. In the event of an injury as an immediate and direct result of my participation, the researchers' sole responsibility is to transport me to an appropriate facility if additional care is needed. The researchers will not provide any compensation or payment for medical care. As a participant, I do not give up or waive any of my legal rights.

The only people who will know that I am a research participant are members of the research team; and, if medical clearance is required for participation, the doctor I choose to provide my clearance.. No identifying information about me or provided by me during the research will be shared with others, except if necessary to protect my rights or welfare (for example, if I am injured and need emergency care); or if required by law. Only research team members who assist with data collection will see me. All of my data will be coded using a participant ID number that is known only to the researchers. As only the reflective markers are visible to the special motion capture cameras, my recorded performances will be confidential and identifiable only by my participant number. The digital video files of my cycling performance will only be used by the researchers to help them track the marker locations from the other cameras if needed (which is rarely the case). If possible, my face will not be visible on the digital video clips. All the rest of the data are non-identifiable. All data, including the electronic video files, will remain in a secured area. Personal health information will not be disclosed and used for any analysis. Only the primary and co-investigators will have access to the master list that that identifies me with my participant ID number, as it will be kept secure and separate from other data files. The master list and digital video files will be destroyed when analysis is finished or 3 years from the completion date of this study, whichever comes first.

For any further questions about the research please contact: Co-Investigator, Thomas McDaniel at 706/542-4132 (tmcdan78@uga.edu)or Principal-Investigator, Dr. Kathy Simpson, at 706/542-4385 (kjsimpsonuga@gmail.com).

I understand that, by my signature on this form, I am agreeing to take part in this research project and understand that I will receive a signed copy of this consent form for my records.

My signature ____ Date __________
Please sign both copies, keep one and return one to the researcher.

________________________________ ____________________________________ ___________

Name of Researcher(s) **Researcher Signature** **Date**

Dr. Kathy J. Simpson 706/542-4385 kjsimpsonuga@gmail.com
Thomas M McDaniel 706/542-4132 tmcdan78@uga.edu
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Additional questions or problems regarding your rights as a research participant should be addressed to The Chairperson, Institutional Review Board, University of Georgia, 612 Boyd Graduate Studies Research Center, Athens, Georgia 30602-7411; Telephone (706) 542-3199; E-Mail Address IRB@uga.edu.

APPENDIX B

HEALTH STATUS AND PHYSICAL ACTIVITY QUESTIONNAIRE

For researcher's use only
PP# ________
Date ________

The purpose of this questionnaire is to help us assess your past medical history and current health status to ensure your safety and that you have no current or past conditions that would affect your performance today. Second, we are gathering information about your prior and current participation in cycling related activities.

Please ask the researcher if you have any questions or need assistance. Your participation is greatly appreciated!

Age: _____yr

Gender: (Place an X in appropriate blank) ____ Female ____ Male

MEDICAL HISTORY AND CURRENT HEALTH STATUS

Medical History

- Please circle the "Y" (yes) or "N" (no).
- If more room is needed to answer a question, continue answer on back of page.)

General Health

- Have you been diagnosed with diabetes? **Y / N**
- Have you ever had an oral glucose tolerance test? **Y / N**
- Have you ever been told by a physician that you have Osteoporosis / Osteopenia? **Y / N**
- Have you ever been told by a physician that you have a heart condition? **Y / N**
- Have you or anyone in your immediate family ever had a heart attack, stroke, or cardiovascular disease before the age of 50? **Y / N**

Y / N

- Have you ever been told by a physician that you have high blood pressure? **Y / N**

- Have you ever been told by a physician that you have high cholesterol? **Y / N**

- Have you ever been told by a physician that you have a thyroid problem? **Y / N**

- Have you ever been told by a physician that you have kidney disease?

- Do you feel angina-like symptoms (pain or pressure in your chest, neck, shoulders, or arms)? **Y / N**

- Do you ever lose your balance because of dizziness? **Y / N**

- Do you ever lose consciousness? **Y / N**

- Do you consider most of your days stressful? **Y / N**

- Do you consider your eating habits healthy Overall? (Lower in fats and fried foods, higher in fruits, veggies and grains)? **Y / N**

- Have you had any major surgeries? **Y / N**

- Do you consider yourself generally healthy? **Y / N**

- Do you currently smoke cigarettes? **Y / N**

- Are you a former smoker? **Y / N**

- Have you ever been told by a physician that you have asthma? **Y / N**

- Any known drug allergies? Y / N
  - If yes, please explain:
    ____

- Are there any health related issues not listed in the previous series of questions you feel we should know about?
  ____
  ____
  ____

- Do you have any current injuries that effect your ability to perform cycling-related exercise, if so please explain (chronic or acute):
  ____
  ____

Please indicate any injuries occurring in the past 15, 30, or 60 day period which may limit range of motion or performance in general (ie. sprains, strains, ligament or muscle tears, undiagnosed pain lasting more than one week, etc.)

	Ankle	**Knee**	**Hip**
15 Days			
30 Days			
60 Days			

<u>Current Health Status</u>

1. If you have any of the following symptoms, please place a check in the blank provided.

   ___tired ___ dizzy ___ trouble with balance ___ muscle soreness ___ unusual clumsiness

2. Are you currently experiencing any physical discomfort or pain? Y/N

   - If yes, please explain.____
     ____

3. Are you currently ill? Y/N

- If yes, please explain.

____

____

4. How much sleep did you get the night before last? _______ hr last night? ________ hr

5. Are you currently taking any prescription or over-the-counter medications? Y/N

If yes to above, if you have possibly experienced side effects:

a. List the medicine(s):

____

____

____

____

b. List side effects that you attribute to medicines (including but not limited to: *pain, discomfort, dizziness, trouble with balance, coordination difficulties, vision or hearing-related problems, muscle aches, trouble understanding directions, inability to concentrate):*

____

____

____

____

PHYSICAL ACTIVITY and CYCLING EXPERIENCE:

This study is to learn more about the biomechanics of cycling. The following questions are to help researchers understand more about the organization of muscle recruitment while cycling in a seated position.
Please let the researcher know if you have questions.

1. Are you experienced with "clipless" style pedal systems? Y / N
2. Approximately how long have you been cycling? ______ yrs.
3. Approximately how many hours a week are you currently cycling? ______hrs.
4. Is your current time investment to cycling what you would consider to be normal for you? Y / N
5. Do you use orthotics or custom insoles in your cycling shoes? Y / N
   - If yes, please explain: ____
6. Do you currently partake in "spin" classes as an alternative to outdoor cycling? Y / N
   - If yes, how often? ____
7. Do you ever partake in maximal efforts while cycling? Y / N
   - If yes, how often?____
   - Do you ever experience discomfort outside of what you would consider 'normal' during these maximal efforts? Please explain.

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