THE EFFECTS OF CLEAT PLACEMENT ON MUSCLE MECHANICS AND METABOLIC EFFICIENCY IN PROLONGED SUB-MAXIMAL CYCLING

A THESIS SUBMITTED TO THE GRADUATE SCHOOL IN PARTIAL

FULFILLMENT OF THE REQUIREMENTS FOR THE DEGREE

MASTER OF SCIENCE

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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Daniel J. Leib

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ABSTRACT

THESIS: The Effects of Cleat Placement on Muscle Mechanics and Metabolic Efficiency in Prolonged Sub-Maximal Cycling

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This study quantified the changes in pedaling mechanics and energy expenditure accompanying a posterior shift in cleat placement during prolonged cycling. Six male competitive cyclists participated. Each subject was asked to complete two separate hour long rides using traditional cleat placement and a novel heel placement, respectively. Expired gasses, kinematics, and EMG from 7 lower limb muscles were collected at three time intervals during each ride. No significant difference in O_2 utilization was seen ($p=0.905$). A significant difference was seen in sagittal plane knee angle ($p=0.008$) and angular velocity ($p=0.003$) in the heel condition, demonstrating a more extended knee and lower peaks in angular velocity. Musculo-tendon kinematic data showed no differences. Tibialis anterior (TA) iEMG was higher in the heel condition, and SOL and TA showed differences in timing between conditions. These results demonstrate changes in ankling patterns and knee joint kinematics as adaptations to heel pedaling.

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NOMENCLATURE

VO_2 - rate of oxygen consumption

EMG - electromyography

iEMG - integrated electromyography

RMS - root-mean-square

SOL - soleus

GAS - medial gastrocnemius

TA - tibialis anterior

BF - biceps femoris (long head)

RF - rectus femoris

VM - vastis medius

GLUT - gluteus maximus

CHAPTER 1

DEVELOPMENT OF THE PROBLEM

Introduction

Competitive cycling is a sport where victory and defeat are often decided by very small differences in ability and performance. Advances in cycling performance have come from a combination of athlete experimentation, commercial product development and testing, and formal research. Many advances in training protocols have come from metabolic expenditure research investigating changes in VO_2 and lactate threshold after riders undergo various training programs. Other improvements have come from advances in dietary research, such as caloric requirements, nutrient ratios, and supplementation. Still other advances have come from changes in factors extrinsic to the rider, such as cycle frame and gear design. A current issue rising in the cycling community is the placement of the cleat on the cycling shoe(Friel 2007; Friel 2007). Several highly competitive riders have are now using cycling shoes that situate the cleat at the middle of the arch of the foot rather than in the traditional location at the ball of the foot. Currently, most evidence concerning alterations in cleat placement is anecdotal, though research is starting to build in this area(Van Sickle and Hull 2007).

Most formal research on cycling has focused on the metabolic aspects of performance, working under the assumption that cyclists who are able to achieve higher oxygen utilization rates will be able to sustain faster riding speeds for longer periods of time. Many of these studies lack significant applicability to competitive cyclists, however, as the subjects used are often untrained and the studies are not of extended duration. While some information may be gleaned, competitive cyclists will probably not see the same level of metabolic adaptation to a particular stimulus as their untrained counterparts due to higher initial fitness levels(Holloszy and Coyle 1984).

Though metabolic factors are important in cycling performance, perhaps a better direct predictor of cycling velocity is the mechanical power output of the cycle-rider system, as it is this power output directly that propels the rider(Abbiss and Laursen 2005). It is useful, then, to measure energy expenditure at a particular power output or a particular velocity, as these measures have a more direct impact on cycling performance. For example, if rider A is able to maintain a given velocity at a certain VO_2 while rider B is able to maintain the same velocity at a lower VO_2, rider B will have a comparative advantage(McArdle 2007). This concept of a given O_2 utilization at a given workload is called gross efficiency (GE) and is calculated simply by dividing work performed by the energy cost(Sidossis, Horowitz et al. 1992).

Many factors influence a rider's mechanical output, most of which are related to the configuration of the bicycle. When the configuration of a bicycle is altered in some manner, be it changing the seat height, handle bar position, or crank length, the rider's posture is altered. These alterations in posture can have significant effects on factors related to muscle mechanics, such as muscle lengths and joint kinematics. One aspect of rider posture that is well understood is knee angle. It is recommended in the literature that a rider's knee be at between 25 and 35 degrees when the leg of interest is at dead drop(bottom position of the crank cycle)(Peveler 2007). Using this position is a good general guideline for high power output, pedaling efficiency, and low rates of pain and injury.

Since changing the rider's posture can affect muscle mechanics and riding kinematics, it is important to understand how it is that the rider uses his or her muscles to generates the force needed to propel the bicycle at a desired speed(Raasch and Zajac 1999). The roles of various muscles have been investigated through EMG, force transducer instrumented pedals, and computational modeling(Redfield and Hull 1986; Brown, Kautz et al. 1996; Neptune and van den Bogert 1997; Raasch and Zajac 1999; Neptune, Kautz et al. 2000; Chen, Kautz et al. 2001; Baum and Li 2003; Duc, Betik et al. 2005; Raymond 2005; Sanderson, Martin et al. 2006; Bieuzen, Lepers et al. 2007). Though there is some individual variability and some aspects that may not be fully understood, uni-articulate muscles serve to add energy to the limb of their insertion while bi-articulate muscles serve to both transfer energy between segments as well as generate energy (Raymond 2005). When muscles shorten more quickly or are closer or further to their optimal length, their ability to produce force is altered in accordance with the length-tension-velocity relationship(Herzog 2000). Alterations in cycling posture and technique may elicit a favorable change in these relationships, allowing for greater force to be produced by a given muscle without additional energy expenditure. Therefore, it is important to investigate how these muscle lengths and shortening velocities are altered when aspects of cycling are perturbed. Since *in vitro* muscle lengths are not easily measured without expensive and sometimes impractical medical imaging devices, computational modeling is a way to make a strong approximation to physiological muscle kinematics. There are a number of methods available for modeling muscular kinematics, including simple geometric relationships, more complicated models using muscle wrapping, and using relationships between joint angles and muscle lengths directly measured through cadaveric study(van der Krogt, Doorenbosch et al. 2008). Sanderson et al have applied the methods of Grieve et al in a study of triceps surae kinematics, describing changes in the kinematics of the muscles at different cadences such as acting out of unison and over difference ranges of length and velocity(Sanderson, Martin et al. 2006). Sanderson et al found that the gastrocnemius operated over a shorter range of excursion at high cadences but with higher velocity, whereas the soleus decreased its range of motion as well but increased in shortening velocity to a lesser extent. Given its success in previous studies, computational modeling can serve as a powerful tool in investigating changes in muscle mechanics in cycling.

PURPOSE

Altering the position in which a rider clips into the shoe-cycle interface may have a significant impact on muscular kinematics, energy expenditure, and neuromuscular dynamics. Currently, there is a paucity of evidence in the literature regarding what changes occur when cleat placement is altered, though potential benefits including increased pedal efficiency and a shift in force demand on individual muscles have been postulated(Frame 2007). It would be useful, then, to quantify these changes that occur with altered cleat placement through regimented protocol.

The purpose of this study is to examine the changes in energy expenditure through expired gasses, muscular kinematics through examination of muscular lengths and lengthening velocities, angular kinematics through joint angles and angular velocities, and neuromuscular excitation through timing changes and integrated EMG calculations associated with alterations in the antero-posterior position of cleat placement.

Expected Outcomes

Due to the same external load being placed on the subjects and the results of Van Sickle et al (Van Sickle and Hull 2007)we do not expect to see a difference in oxygen consumption between conditions. Since the mechanics of the task will necessarily change in the heel position, however, we would expect to see an increase in TA activity and a decrease in SOL activity due to changing ankling patterns. The GAS will not experience a significant change due to its multiple roles as a bi-articulate muscle. We also expect increases in integrated EMG of the gluteus musculature as well as the biceps femoris due to the posterior placement of the cleats and increased reliance on hip extension to produce force. Since aspects of the subjects' cycle configuration other than cleat position will be held constant, it is not expected that joint angles and velocities or muscular kinematic changes will occur about joints above the ankle.

SIGNIFICANCE

Altering the antero-posterior position of cleats on a cycling shoe may elicit differences in energy expenditure, muscle mechanics, and neuromuscular dynamics. While pedal efficiency and changes in energy expenditure over short duration rides have been assessed, there is currently no research available on posterior cleat placement over long-duration, sub-maximal cycling exercise. The data gained from this study may be used by competitive cyclists to better assess choices in shoe equipment design for optimal performance.

METHODS

Subjects

Seven competitive male cyclists aged 18-45 years were recruited for this study. To qualify, subjects must ride at least 100 miles per week in season and have competed in at least one race in the past calendar year.

Testing

All testing was done at the Ball State University Biomechanics Lab. Prior to the first testing session, subjects attended a one hour initial meeting session. In this meeting, subjects filled out an informed consent form approved by the Ball State IRB and asked any questions. The subject then performed a 30 minute familiarization ride using the heel clipped position on a Velotron cycle ergometer (Racermate Inc., Seattle, WA) fit to his normal cycle configuration.

A total of seven male competitive cyclists were recruited for testing. Each subject underwent two testing sessions, one clipped in a traditional toe position and one in the heel. Which condition was tested first was pseudo-randomized according to subject number in order to control for a learning effect.

To control for cleat position in the traditional toe position, cleats were positioned posteriorly from the toe end of the riding platforms by 25% of the subject's shod foot length. The same was done for the heel position, except that 25% of the subject's shod foot length was measured from the most posterior portion of the shoe. The subjects' feet were secured into the platforms using clips and a custom Velcro strap configuration.

At the beginning of each testing session, the subject performed a ten minute warm up protocol which included a 30-second torque-velocity (T-V) test as described by Rouffet et al(Rouffet and Hautier 2007) in order to obtain a maximal EMG signal representative of EMG activity during cycling. The load chosen for the test was 9Watts/kg, which is similar to peak values that may be obtained during a Wingate protocol(cite***). This test was done using the rider's own road cycle and cleats for both collection sessions on a CompuTrainer (Racermate Inc., Seattle, WA).During this T-V test and for all subsequent collections, subjects were fitted with 14 EMG electrodes bilaterally on the soleus, gastrocnemius, tibialis anterior, biceps femoris long head, rectus femoris, vastus medius, and gluteus maximus using a Delsys Bagnoli system (Delsys, Boston, MA). All EMG data were band-pass filtered at 6-600Hz and RMS processed using custom MATLAB scripts for later analysis.

After this warm-up period, subjects were fitted with retro-reflective markers according to the Vicon standard Plug-In-Gait marker set (Vicon, Oxford, UK). This simple marker set was deemed appropriate due to the primarily sagittal nature of movement in cycling and the likelihood that some markers may fall off during the hour long testing procedure; more complicated marker sets would present a greater likelihood of this occurring. Subjects then completed a 1 hour ride at 175 Watts in the designated position on the Velotron. At 10, 35, and 59 minutes, kinematic, EMG, and expired gasses were collected for a period of one minute. These times were chosen to represent even spacing during the ride after a 10 minute period designed to allow the subject to reach steady-state exercise. Expired gasses were collected using a Parvomedics metabolic cart (Parvomedics, Sandy, UT) and a standard one-way valve mouthpiece.After data collection, kinematics were filtered using a Woltring filtering routine with MSE of 20 and processed using Vicon's OLGA procedure (Vicon, Oxford, UK). Text files for use with OpenSim(Delp, Anderson et al. 2007) were also exported, as were EMG data. In order to obtain muscle lengths, models were generated for each subject in OpenSim(Delp, Anderson et al. 2007) using a modified BothLegs.osim file. These models were scaled and driven using collected kinematics, and muscle lengths for the same seven muscles EMG data was collected on generated using the plot function. Joint angular velocities and muscle lengthening velocities were obtained, angle data included, and all data normalized to crank cycles using custom written MATLAB scripts

(Mathworks, Natick, MA). Position data of a marker placed on the pedal was used to normalize data to percent crank cycle.

For each time point, the first four usable crank cycles were extracted for all kinematic and EMG data. Data from these four crank cycles were then averaged, resulting in three sets of data per cleat condition. Since preliminary analysis of the data showed no significant change in values over the three time points, the three time points were averaged, resulting in one set of data per condition for statistical analysis. From the data gathered on angular and muscular kinematics, a RMS value was calculated on each angle and angular velocity curve for each subject, resulting in 6 RMS values per angle and angular velocity measurement for each condition. Muscular data were analyzed on the dominant limb, resulting in 7 pairs of RMS values for muscular lengths and 7 pairs for lengthening velocity. These data were entered in to SPSS and analyzed using a dependent samples t-test for each variable. Though it would be more statistically correct to analyze the variables together in an ANOVA with repeated measures or, more broadly, a multiple ANOVA with repeated measures, it was felt that changes in one joint will de facto affect another, thus potentially masking any true differences between conditions for both muscular and joint kinematic data. EMG average curves were analyzed using cross-correlation in SPSS, with integrated EMG being analyzed using separate dependent samples t-tests per muscle.

Limitations

Although all riders must meet the stated inclusion and exclusion criteria, some riders may be better aerobically and/or anaerobically conditioned than others. Strength of the rider has been indicated as a possible reason for EMG timing changes and will not be explicitly controlled for. Also, while riders will be asked to log and duplicate food intake and testing will occur at the same time of day for each test, it is possible unforeseen and uncontrollable differences such as amount of sleep will affect outcome measures.

Delimitations

All subjects will be male competitive cyclists living and training in the Muncie, IN area and thus reflect that population.

Summary

As competitive cyclists are always looking to gain an edge over their opponents, much research has been done on aspects of cycling performance. One aspect that is beginning to garner more attention is cleat placement which is important because the shoe-pedal interface is a critical site for energy exchange from rider to cycle. One aspect of this interface that can be altered is the antero-posterior position of the cleat on the shoe. In order to investigate the effect of this placement, the two reasonable extremes in placement were tested; traditional toe placement and heel placement. Experimental kinematics, expired gasses, and EMG data were collected directly and muscle length data generated post hoc using OpenSim in an effort to understand what changes occur. These changes were analyzed using cross-correlations and t-tests that, while maybe not strictly appropriate statistically speaking, serve to shed more light on changes that are occurring than would be possible with combined tests that may mask important changes.

CHAPTER 2

REVIEW OF LITERATURE

INTRODUCTION

Competitive cycling is a sport where very small differences in training and equipment design can mean the difference between winning a race and coming up short. While much research has been done on metabolic, training, and equipment factors affecting cycling performance, very little research has focused in differences affected by cleat placement. Since the cleat-pedal interface is where the rider delivers force to the pedal to propel the cycle during every pedal stroke, it is important that this interface be optimal for best performance to be achieved.

This literature review will cover the current, relevant literature associated with the gross mechanics and neuro-mechanics of cycling and how it relates to performance outcomes. Also discussed will be computational modeling, an emerging tool in the measurement of muscle mechanical variables and how it can be applied to cycling research.

Factors Affecting Cycling Performance

There are many ways to measure and predict performance outcomes in competitive road cycling, from metabolic expenditure to mechanical power output. These variables can be altered in many ways, from using different training protocols to changing cycle configurations. While metabolic measures and training protocols are very important, these factors are very well represented in the literature(Abbiss and Laursen 2005). Once training is in place, racing speed can be improved through changes in biomechanical factors that affect mechanical power output and ultimately improve the velocity of the cycle-rider system.

Performance in competitive cycling is typically measured in four ways: time to completion of a given course, O_2 consumption, lactate threshold, , and mechanical power output. Time to complete a given course is the primary performance measure in competitive cycling as it determines who wins a race. There are a large number of smaller factors that contribute to this global measure, however, and there has been much investigation into which factors are most important. Oxygen consumption is one of those factors often investigated. Oxygen consumption has been used for decades as a performance measure in endurance sports due to oxygen's pivotal role in muscular energy production. This variable is typically expressed in relative terms of oxygen consumed per kilogram of bodyweight. Since road cycling is predominantly an aerobic sport, it makes sense to use this measure as a potential performance indicator as the ability to utilize more oxygen should result in a greater ability to maintain a given cadence(Abbiss and Laursen 2005). Often, measurements of oxygen consumption are expressed in relation to energy expenditure and useful work performed. In cycling, Sidossis et al and Ashe et al expressed this relationship as gross efficiency (GE)(Sidossis, Horowitz et al. 1992; Ashe 2003). GE is a useful way of capturing aspects of metabolic work compared to useful mechanical work in a single expression. This factor is calculated as reported in Figure 1. When work performed is kept constant, however, the simple measure of oxygen consumption represents the same relevant information. Despite the importance of O_2 consumption in muscular energy production this variable fails to address the entirety of metabolic indicators of performance. A rider's lactate threshold is another metabolic indicator of performance, as it has been shown that well-trained athletes are generally able to perform at an intensity slightly under their lactate threshold for extended periods. Though it has been

Figure 1

Gross Efficiency = work performed [kcal·min^{-1}] ÷ energy cost [kcal·min^{-1}] x 100

Work performed = power output [kgm·min^{-1}] x 1 kcal·426.6.4 kgm^{-1}

Energy cost = VO_2 [l·min^{-1}] x Y kcal·l^{-1} O_2

Energy cost in kcal·min^{-1} is calculated as a function of the caloric equivalent of 1 l of oxygen (Y) at various values of R.

Formula for calculating gross efficiency (Coast et al., 1986)

determined in recent research that it is not lactate itself that increases acidosis and decreases performance, this threshold does still correlate to performance outcomes(Faria, Parker et al. 2005). It stands to reason, then, that the greater an athlete's lactate threshold the better the performance(Abbiss and Laursen 2005).

While these metabolic factors are important in determining performance, a perhaps more important factor is mechanical output of the cycle-rider system. It is possible for a rider to have an impressive relative capacity to use oxygen and operate at a significant level of that threshold at length but not produce the mechanical power needed to propel the system fast enough to win a race(Abbiss and Laursen 2005). It is useful, then, to measure the rider's ability to achieve both a peak and sustained mechanical power output, as a greater power output would provide for a greater cycle-rider velocity(Korff, Romer et al. 2007).

There are many factors that influence a rider's mechanical power output. One aspect is the rider's posture on their bicycle. Alterations in posture influence muscle mechanics such as muscle lengths, lever arms, and shortening velocity. There are several principle ways in which a rider can alter posture. Riding posture is typically measured when a rider is seated on the saddle with the crank at dead drop. The primary measure used to assess posture is knee angle. This angle ties in closely to a primary aspect of the bicycle configuration; seat height. Typically, it is recommended in the literature that a cyclist should have a knee angle of between 25 degrees and 35 degrees, or the height from the seat to the pedal should be 109% of inseam(Peveler 2007). Once the seat height is set to achieve the correct knee angle, the next posture concern is the rider's hip angle. There are no values present in the literature that describe the “best” hip angles for a road cyclist, however. This angle is usually set by altering the forward positions of the seat and handle bars so that the rider maintains a neutral lordosis while maintaining an appropriate aerodynamic position. Shoulder and elbow angles are affected by seat and handle bar placement, but are typically fitted so that the rider is comfortable while maintaining the other joint angles. While it has been found that many riders produce more power in an upright posture(Ashe 2003), this posture is not appropriate for riding at high velocity due to wind resistance and is not representative of typical racing posture. The last aspect of posture usually controlled for is ankle angle. This angle is predominantly influenced by the rider's technique, bike configuration, and shoe cleat placement. Proper technique for a road cyclists is usually defined by "pushing and pulling" the pedal rather than "mashing"(Korff, Romer et al. 2007). The “pushing and pulling” technique is characterized by an emphasis of exerting force forward and backward rather than straight down, which characterizes “mashing.” Cleat placement is usually kept constant under the ball of the foot (at the base of the middle metatarsals), though there has recently been a trend in cycling towards moving the cleat position to under the arch of the foot(Friel 2007). The effects of this altered cleat placement on cycling kinetics, kinematics, and energetics are not currently well understood(Van Sickle and Hull 2007).

There are many factors that affect how a competitive road cyclist performs. Metabolic expenditure, oxygen usage, and an athlete’s lactate threshold are all improved through physical training. In modern cycling, however, physical training is only part of what goes in to winning a race. There are other biomechanical aspects that are improved with changes in posture, including kinematic variables, muscle mechanics and, ultimately, speed on the road.

Muscle Mechanics and Neurological Control

The ability of muscles to produce force is an important aspect of cycling as active muscular force production and subsequent energy transfer to the pedals is the primary method by which the rider propels the bicycle (Raasch and Zajac 1999). In order to understand how to best manipulate mechanical factors of a bicycle to produce force, it is important to understand how skeletal muscle tissue functions in this capacity.

It is currently widely accepted that the way skeletal muscle produces force actively is through cross-bridge cycling as initiated through neuromuscular excitation (Rassier, MacIntosh et al. 1999). The cross-bridge theory was originally postulated by Huxley in 1957 stating that skeletal muscle produces force by active means through interactions between actin and myosin proteins. Myosin joins to active sites on actin in the presence of calcium ions and then releases through chemical interactions with adenosine tri-phosphate (ATP). Using this model several aspects of muscle mechanics become apparent. First is that there is an optimal length at which muscles are best able to produce force to that the greatest number of available cross-bridge sites are available due to the physical structure of actin and myosin (Rassier, MacIntosh et al. 1999). While this property of muscle is more appropriately discussed at the level of the sarcomere from a theoretical sense(Rassier, MacIntosh et al. 1999), it is currently not practical to measure individual sarcomere length during active muscle contraction outside of an MRI device(Maganaris 2001). While it has been shown that sarcomeres within a muscle fiber stretch differentially based on their proximity to the end of the fiber while under load(Carlsen, Knappeis et al. 1961), the error introduced due to invalid sarcomere length estimation based on working within the context of a whole muscle framework is probably not great (Rassier, MacIntosh et al. 1999). Related to the force-length properties of muscle are the force-velocity properties(Rassier, MacIntosh et al. 1999). Generally speaking, muscle force production changes with contraction velocity; the more negative the contraction velocity, the greater the force skeletal muscle can produce at any given instant during maximal contractions. A similar relationship is seen during sub-maximal, dynamic contractions, though the relationship is altered slightly(Rassier, MacIntosh et al. 1999). The length-tension-velocity properties of whole skeletal muscle are dependent then on joint kinematics, muscular origin and insertion points, and the tension on the muscle. Tension on the muscle can be prohibitively expensive and invasive to measure as it would require placement of force transducers directly on a subject's musculature. Since origins and insertions cannot be easily measured without medical imaging equipment, changes in joint kinematics and loads will have the largest, easily measurable and manipulable effects on muscle mechanics(Rassier, MacIntosh et al. 1999). Joint kinematics also have the greatest effect on a given muscle's moment arm, an important factor in torque produced about the joint the muscle crosses. Since many aspects of muscle mechanics are largely dependent on joint kinematics, these length-tension-velocity characteristics of muscle become important in relatively closed-chain tasks such as cycling due to the role they play when kinematic variables are altered by alterations in cycle configuration or altered pedaling mechanics(Brown, Kautz et al. 1996; Ashe 2003; Sanderson, Martin et al. 2006).

Two other important and closely related aspects of force production in cycling are activation dynamics and motor control. Activation dynamics in cycling has been extensively investigated(Neptune, Kautz et al. 1997; Raasch, Zajac et al. 1997; MacIntosh, Neptune et al. 2000; Neptune and Herzog 2000; Neptune, Kautz et al. 2000; Raymond 2005; Rouffet and Hautier 2007). Muscles typically investigated in cycling studies include the vastus medialis(VM), vastus lateralis(VL), rectus femoris(RF), tibialis anterior(TA), biceps femoris(BF), gastrocnemius(GAS), soleus(SOL), and gluteus maximus(GLUT) as these muscles represent the primary force producing muscles throughout the crank cycle(Rouffet and Hautier 2007). While other muscles such as tibialis posterior and the peroneals do contribute, their measurement through surface EMG is difficult and fine-wire EMG is often impractical.

Determining a “normal” neuromuscular activation pattern during cycling is no small task, as activation patterns can vary greatly among different cadences, power outputs, and rider skill level(Raymond 2005). Even among experienced cyclists, varying patterns can arise due to differences in strength levels, limb lengths, and other biomechanical and cognitive factors(Coyle, Feltner et al. 1991). There are, however, some trends that can be observed. During cycling at a cadence within normal ranges for untrained cyclists (60-80RPM), VM,VL,RF,RA,BF, and GLUT most active for first half of propulsion; BF,GAS, and GLUT tend to maintain activity to bottom dead center. The RF experiences a gradual rise and fall of EMG amplitude, while the vasti undergo a relatively quick rise and fall. The medial and lateral hamstrings tend to peak in amplitude at 90 degrees from bottom dead center(Raymond 2005). In addition, one and two joint synergists often co-activated but differ in peak magnitude and timing(Prilutsky and

Gregory 2000). Activation patterns can change in elite cyclists, however. For this population, preferred cadence is approximately 90RPM(Coyle, Feltner et al. 1991), and EMG timings and amplitude have been shown to change with cadence(Raymond 2005) as well as skill level. Some have suggested that these changes could be training adaptations that optimize power output, minimize fatigue, and influence other factors of performance for the individual(Raymond 2005).

Changes in neuromuscular activation patterns are further altered with the pedaling task is perturbed. Some perturbations that a cyclist may experience are postural changes, fatigue, and equipment changes. While riding posture is relatively consistent for road cyclists, a rider may choose to stand, rather than sit, during certain portions of a race. Research by Brown et al showed that untrained cyclists attempted to change activation patterns to compensate to a more upright posture by preserving ankle kinematics(Brown, Kautz et al. 1996). Their work was further supported by the investigations of Chen et al, suggesting that preservation of cycling kinematics may be a primary goal of motor control strategies(Chen, Kautz et al. 2001). EMG response to fatigue in cycling is fairly similar to fatigue in most activities; EMG amplitude increases for a given level of force output and the percentage of signal at lower frequencies increases(Raymond 2005). Changes in equipment tend to elicit subtle changes in activation dynamics. Small changes in amplitude were observed in amplitude, but not timing, in a simulation study by Neptune and Herzog where the bicycle chain ring was altered in shape(Neptune and Herzog 2000). No difference has been observed with changes in shoe insole design(Jorge and Hull 1986; Raymond 2005), though it should be noted that cleat placement of the shoes did not change in that study. Changes in cadence will also influence a change in

EMG dynamics. Activation of the RF tends to be higher at lower cadences, such as 60RPM, and RF and BF burst occur earlier and exhibit double bursting patterns at a higher cadence(Raymond 2005). The strength of a cyclist will also have some influence of EMG activation patterns during a crank cycle. Weaker riders will tend to use a greater percentage of MVC across the crank cycle, while stronger riders tend to exhibit earlier bursting of BF and RF in the crank cycle(Bieuzen, Lepers et al. 2007).

These changes in neuromuscular excitation, along with numerous computer simulation studies, have been used to attempt to tease out the role of individual muscles used during cycling. Identifying the roles of individual muscles in a given motor task is not trivial; since many joints have a larger number of muscles than degrees of freedom, their roles can often be ambiguous and not easily solved for in inverse dynamical analysis(Anderson and Pandy 1999). Many computer simulations have been used to attempt to discern the roles of individual muscles by using moderately constrained dynamical optimization techniques(Raasch, Zajac et al. 1997; Neptune and Hull 1999; Chen, Kautz et al. 2001; Raymond 2005; Sanderson, Martin et al. 2006). The roles of uniarticular muscles have been fairly well established through these investigations. The role of the soleus, for example, appears to be primarily to transfer energy generated in more proximal musculature and segmental inertia to the pedal(Sanderson, Martin et al. 2006). The role of the vasti are to generate energy that is transferred through the shank to the soleus and then to the pedal during the down stroke(Raymond 2005). The gluteals primarily impart energy to the thigh segment, which is subsequently transferred distally by other structures(Raymond 2005). The roles of the bi-articulate muscles are more complex, serving both to regulate power transfer between segments and to generate energy themselves(Kautz and Neptune 2002). There are also is a primitive muscle activation pattern associated with cycling involving three antagonistic pairs muscle groups; a "Ext/Flex pair that accelerates the foot into extension or flexion with respect to the pelvis, an Ant/Post pair that accelerates the foot anteriorly or posteriorly with respect to the pelvis, and a Plant/Dorsi pair that accelerates the foot into plantarflexion or dorsiflexion(Neptune, Kautz et al. 2000)." While a two-pair method of excitation can be used to drive forward pedaling at a normal cadence, it fails when cadence and pedaling direction are altered(Neptune, Kautz et al. 2000).

Modeling as a Tool for Investigation

Many biomechanical aspects of cycling are difficult or currently not possible to directly measure, such as dynamic changes in fiber length or energy transfer between segments (expensive, specialized force measurement equipment) or dynamic changes in fiber length (requires medical imaging difficult to achieve on a bicycle). In order to investigate these difficult to measure values, mathematical modeling and simulation offer viable solutions for some aspects when applied appropriately. Since modeling has been applied so often in cycling research, it is useful to understand what sorts of modeling techniques have been used and the benefits as well as limitations of each.

Perhaps the first applications of computational modeling in the field of biomechanics dates back to the late 70's when Cavagna et al investigated various? aspects of gait. Their simple inverted pendulum model was fairly easy to design from a programming standpoint and computationally efficient. In this model, the center of mass of the body is considered to be the proximal end of the "leg" while the distal end was considered fixed to the ground and modeled with a simple hinge joint(Pandy 2003). However, it is only appropriate for investigating center of mass displacement due to gross oversimplifications of the thigh, shank, foot, and their joints(Pandy 2003). The next step in modeling was taken when the pendulum model was made to make contact with the ground at the point represented as the distal end of the leg and thus generate a simplified vertical component and antero-posterior component of the ground reaction force. In this model, gross kinetics could be investigated as a function of walking speed and ground contact angle. The model begins to exhibit unreasonable antero-posterior forces at higher moving velocities, however, such as with running(Pandy 2003).

At about the same time, investigations using a spring-mass model and multi-segment rigid bar linkage began to become more prevalent. Spring-mass models attempted to account for some of the assumptions made in the inverted pendulum model by adding a generic spring element in each limb designed to account for the gross compliance of the joints and soft tissues(Blickhan 1989). This model is better able to account for gross kinematics and kinetics for movements more ballistic than human gait and allows insight into how changes in gross stiffness of a limb may affect these variables. This type of model is also very computationally efficient. This type of model still lacks the complexity required to investigate changes and values at individual joints and muscles, however(Blickhan 1989). The multi-segment rigid-bar type of model was an attempt to correct this shortcoming. This type of model uses multiple rigid-bar segments linked with simple hinge joints to represent the human skeletal system. Initial versions of this type of model were strictly planar, two dimensional models actuated by joint moments(Redfield and Hull 1986). This type of model is appropriate for investigating movements that occur predominantly in one plane, such as gait and cycling, where the contributions of individual joints to gross kinetics are desired. This model variation can also be further expanded to include three dimensional movement and muscular actuators.

Currently, using a three dimensional, multi-segment model is standard practice for gait and motion analysis labs for most applications(Kirtley 2006). Typically, three dimensional kinematics and ground reaction forces are measured experimentally and then used to drive the model. Using this type of model, it is possible to investigate three-dimensional kinematics and kinetics across segments and joints using standard inverse dynamics techniques. While this type of model is considerably more computationally complex than planar models, current computer hardware technology is fast enough to make its use quite practical. Using simulated muscle tissues attached at locations on the model to mimic physiological muscle geometry, it is also possible to investigate muscular kinematics with this type of model(Zajac 1989).

When attempting to investigate the contributions of individual muscles to gross joint moments the issue of degrees-of-freedom (DOF) used in a model becomes painfully apparent. The appropriate number of DOF used to model each joint varies depending on the desired outcome variables of the model and the expected computational cost(Neptune 2000). Movements that happen predominantly in one plane, such as cycling, the knee, ankle, and hip joints can be built as hinge joints when flexion/extension are the main variables of interest. When investigating this set of information, the number of physiological muscles available across a joint exceeds the DOF of the joint. When this is the case, it becomes impossible to directly solve for the forces generated by each actuator(Anderson and Pandy 1999). A number of methods have been developed to address this problem. A simple method used in early musculoskeletal modeling was to combine muscles into "optimized" groups. This method can be useful when the actual contributions of individual muscles in a group, such as the muscles of the quadriceps, are not of concern, but rather their net contribution to the joint moment is of interest. Another simple method that can be used is to simply divide the forces required among all the muscles that can contribute to that movement. For example, the LifeMod modeling tool distributes forces required for knee extension evenly among the four modeled quadriceps muscles up to a set maximum force production value (BRG, San Clemente, CA). While this method is computationally efficient, it may not reflect physiological control strategies used for many activities and becomes problematic when investigating bi-articulate muscles due to two distinct sources of force demand. A method proposed by Herzog involves using a least-cubes algorithm to distribute force production unequally among muscles spanning a joint as this may represent a more physiological strategy, but this method is not well represented in the literature(Herzog 2000). A well-represented method for solving this problem is a technique known as Computed Muscle Control (CMC)(Thelen and Anderson 2006). CMC uses a static optimization framework xperimental kinematics and ground reaction forces (GRFs)are used as inputs to determine the accelerations of the model as well as the excitation history, and thus force produced, of each muscle. For each unit of time, accelerations of segments are derived that will produce a solution very similar to experimental kinematics. Muscle forces are then generated that will produce these

$$J = \sum_{i=1}^{m} V_i[a_i(t+T)]^2$$

Figure 2

segmental accelerations as well as match experimental GRFs. These forces are also set to minimize a cost function where V_i is the muscle volume of a given muscle and $a_i(t+T)$ is the activation of the given muscle at the desired muscle force; t represents the current time in the simulation and T represents the time interval of the kinematic sampling rate.(Thelen and Anderson 2006) This processes is then repeated for each successive frame of experimental kinematics. This method of muscle modeling accounts for the time delay between excitation and contraction and utilizes a Hill type muscle model that includes tendon properties, passive muscular properties, and active muscular properties(Thelen and Anderson 2006). A schematic of the CMC process is presented below, taken from Thelen and Anderson, 2006:

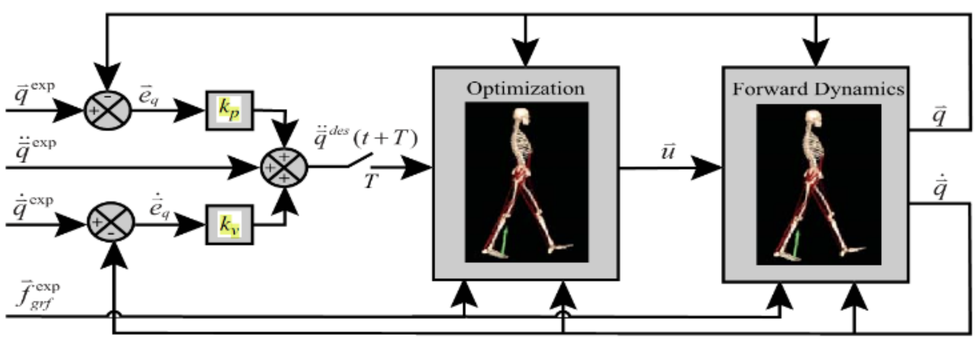

Figure 3

Muscular kinematics have been modeled in a number of different ways. Some researchers have used simple geometric rules to define the length of a muscle based upon the origin and insertion points and where in two or three dimensional space these points lie in space relative to each other(van der Krogt, Doorenbosch et al. 2008). The method of Grieve et al as implemented by Sanderson et al(Sanderson, Martin et al. 2006), involves direct estimation based upon values measured from cadavers directly when joint angles were manipulated. While this method is taken from physiological data, problems may arise when applied to individuals as muscular origins, insertions, and moment arms can vary considerably(Visser, Hoogkamer et al. 1990). More recent methods have used a combination of previous methods while incorporating techniques to enhance accuracy. The method of Delp et al incorporates origin and insertion location estimations derived from cadaveric data, using geometric methods to estimate Musculo-tendon unit length and moment arm length, and accounts for wrapping geometry using "via points" and "via cylinders." These points and cylinders account for error that would otherwise be introduced due to muscle geometry passing through bone segment geometry(Delp, Anderson et al. 2007; van der Krogt, Doorenbosch et al. 2008). This sort of complex geometry modeling tends to yield more consistent results across subjects and angular positions(Visser, Hoogkamer et al. 1990). Musculo-tendon unit shortening velocities can then be found by differentiating length data, and accelerations found by a further differentiation(Thelen and Anderson 2006).

Musculoskeletal modeling can be a time consuming task. Initially, all modeling had to be done using custom-written software. Development of the software required for a particular modeling goal could take months in itself depending on the developer's level of expertise. After development, depending on the complexity of the model, months of computational time could be required(Thelen and Anderson 2006). If a problem was uncovered in the software, the computational process might have to be done all over again. In an effort to make computational modeling more accessible for general use in biomechanics, a number of tools have been developed. One such tool is the AnyBody scripting language developed originally by Aalborg University Biomechanics and currently developed by AnyBody Technology (Aalborg, Denmark). This custom-developed scripting language is a combination of elements from several different programming languages and is designed for use among biomechanists with a programming background. Much like the MATLAB programming environment, it is also possible to graph and analyze outputs of the model within the same environment. Though this tool is simpler to use than, perhaps, C or COBOL, its use still requires strong knowledge of textual programming in addition to the mechanical principles being modeled. Another tool that has been developed is the LifeMod plug-in for the MSC.ADAMS software platform (Biomechanics Research Group, Inc, San Clemente, CA). This tool uses a graphical user interface (GUI) to guide a user through a number of steps in creating a musculoskeletal model rather than a traditional text editor. Since the tool is built upon a mature engineering modeling suite, it is possible to integrate features not traditionally used in biomechanical modeling such as object interaction and non-biological material properties at a lower computational cost than many custom-written programs. While this interface does allow for greater flexibility without requiring the same programming ability as a custom written model, problems can arise due to the particular methodologies of the software. Since it is not specifically written for human movement and other biomechanical applications, there can be times when a certain method for determining an outcome variable would be desirable, but is not easily achievable. Another very similar program is SIMM (Musculographics Inc., Santa Rosa, CA). This program is designed to interface well with Motion Analysis motion capture technology (Motion Analysis, Santa Rosa, CA), but can still suffer from a lack of efficient customizability, closed source code, and is tied closely to proprietary systems.

More recently, a concerted effort has been made by a group led by Scott Delp to produce an open-source biomechanical modeling tool called OpenSim(Delp, Anderson et al. 2007). This software was started as a response to the closed-source nature of other modeling tools such as AnyBody, LifeMod, and SIMM to provide a framework where biomechanics researchers could collaborate and share work without being tied to any particular software system. Since OpenSim's source code is publicly available, researchers are free to use the tools already provided in the software or write their own module if certain functionality is missing. Since this tool is built by and for the biomechanical community, most basic calculations that are useful to biomechanics are already available, as are more advanced tools like muscle force modeling. These tools are also based off of algorithms published in academic literature, such as Thelen's CMC algorithm and Schutte's muscle model design(Schutte 1993). Since these previously published algorithms are often what current research and specialized models are based off of, incorporating them into the software design saves new researchers the time of recoding them for every study, while the open nature of the software allows new algorithms to be added as they are developed and validated.

In summary, there are many useful tools available for modeling muscle mechanics. While each have their own strengths and weaknesses, OpenSim is emerging as a preferred method of modeling human muscle mechanics due to the fact that it is open and based upon research previously validated in the biomechanics literature. While other tools may be useful in certain circumstances, for general human movement analysis, OpenSim is a powerful and validated tool that is efficient to use.

Summary

Many factors affect cycling performance outcomes. Metabolic, mechanical, and biomechanical aspects can all play a role. Metabolic factors, including oxygen consumption and lactate threshold, are important and perhaps some of the most studied variables in all of sport. Despite their importance, however, it is the eventual generation and output of mechanical power that propels a rider over a course. A rider's posture is an important determinant of many mechanical variables, so seat height, knee angle, and cleat placement can have a profound influence on how a rider is able to generate power.

Since muscular contraction is the primary means by which a rider propels a bicycle, understanding neurological control and muscular dynamics is an important part of understanding cycling performance. One way to infer properties of how a muscle is behaving in a system is to investigate the kinemtaics of the Musculo-tendon system in conjunction with joint kinematics and EMG. Even in the absence of force data, these variables can provide significant insight into how a system is operating.

Many variables relating to muscular dynamics are not feasible to measure *in vivo* during a cycling task. A good solution to this problem is the use of computational modeling, where muscle behavior is estimated given a series of input parameters. While comparatively new to biomechanics, recent work has validated many of the methods commonly used in the literature and it is reasonable to use these methods with the appropriate understanding of their strengths and limitations.

CHAPTER 3

METHODS

As competitive cyclists are always looking to gain an edge over their opponents, much research has been done on aspects of cycling performance. One aspect that is beginning to see more attention is cleat placement, important because the shoe-pedal interface is a critical site for energy exchange from rider to cycle. One part of this interface that can be altered is the antero-posterior position of the cleat on the shoe. In order to investigate the effect of this placement, the two reasonable extremes in placement were tested; traditional toe placement and heel placement.

Experimental Procedures

Seven experienced male cyclists were recruited from the Ball State cycling club and from personal and professional contacts of Jeff Frame, competitive cyclist and professional staff of the Ball State Biomechanics Lab. After initial recruitment, participants were stepped through procedures, potential risks, and benefits associated with study participation. Once the recruit agreed to participate, he signed an informed consent form and scheduled a time for familiarization procedures. The subject was also assigned a generic subject number at this time.

In order to familiarize the rider with the novel cleat placement and custom cleat placement device being used, subjects were instructed to ride for ten minutes at 175Watts on the Velotron cycle ergometer (RacerMate Inc., Seattle, WA) to be used in testing. The ergometer was adjusted to the subject's own road cycle dimensions for the toe position. In order to best isolate cleat position as a variable, these aspects of cycle configuration were adjusted to best duplicate riding posture: seat height, seat antero-posterior position (fore-aft), handle bar drop height, reach, and knee angle. These measurements were performed and/or confirmed by a certified Serrota bike fit specialist and recorded for use in testing sessions.

After arriving at the Biomechanics Laboratory, subjects first changed into cycling apparel consisting of spandex (or similar) shorts and a form-fitting shirt. Once properly attired, the investigators adjusted the Velotron Dynafit Pro (RacerMate Inc., Seattle, WA) cycle ergometer to their preferred position. Once adjusted, it was verified that the subject's knee angle measures between 25 and 35 degrees as measured by a hand-held goniometer when the ipsilateral leg is at the bottom dead center position. If the subject's preferred adjustments did not cause the knee angle to fall within this range, small adjustments were made to seat height so that this condition was met. Once the ergometer was suitably adjusted, subjects were affixed with 14 surface EMG electrodes using a 16 channel Bagnoli system (Delsys Inc., Boston, MA). The muscles monitored were the SOL, GAS, TA, BF, RF, VM, and GLUT on each leg (Bieuzen, Lepers et al. 2007; Rouffet and Hautier 2007). Placement was done according to the guidelines in *Introduction to Surface Electromyography* (Cram 1998). The site for each electrode was first palpated, shaved to remove body hair, lightly abraded using fine sandpaper to remove dead skin cells, and swabbed with alcohol. After allowing the alcohol to dry, electrodes were affixed using double-sided medical tape and wrapped firmly using pre-wrap. The electrodes were then connected to two relay boxes affixed to a neoprene vest. The relay boxes were then wired to a main switch box, then connected to a Vicon MX controller box to be synchronized together with kinematic data using Vicon Workstation software (Vicon Corporation, Oxford, UK). EMG placement was then tested using a series of manual muscle tests designed to illicit response from an intended muscle. A walking trial was used first to detect appropriate bursting. If clear bursting was not observed, the electrode was replaced, and another manual test performed. Once EMG placement was completed, subjects performed a 5 minute warm up on the CompuTrainer (RacerMate Inc., Seattle, WA) in order to warm up both the rider and the ergometer machinery. After this warm-up, the CompuTrainer (RacerMate Inc., Seattle, WA) was calibrated according to manual instructions. After the rider and machinery were sufficiently warmed up, the rider began a protocol to provide a continued warm-up leading in to a torque-velocity (T-V) test as described by Rouffet et al(Rouffet and

Hautier 2007). The load chosen for the T-V test was 9Watts/kg, a load similar to that experienced during a Wingate power output test(Bar-Or 1987). The protocol began with five minutes of riding at a light load (75Watts), thirty seconds of the T-V test at full load, then five more minutes at a light load to allow for a cool down period. After this protocol was complete, the rider was affixed with retro-reflective markers according to the Vicon standard Plug-in-Gait (PiG) (Vicon Corporation, Oxford, UK) marker set. In total, 39 markers were placed on each subject on selected anatomical landmarks. After the markers were placed, a static collection trial was collected for subject calibration. A 10 camera Vicon MX collection system collecting at 100Hz was used for all kinematic data collection; EMG was collected at 2400Hz.

After this static trial was completed, the rider either simply clipped in to the pre-configured Velotron (RacerMate Inc., Seattle, WA) for the toe condition or the Velotron was adjusted for changes in initial position for the heel condition. Once the rider was configured and clipped in, the one hour ride began. For this portion of data collection, the load supplied by the ergometer was 175Watts. Kinematic and EMG data were collected at 10 minutes, 35 minutes, and 59 minutes during the protocol for thirty seconds. Expired gasses were collected for four minutes prior to/encompassing each kinematic/EMG collection to allow for human error inherent in putting on the mask and maintaining a normal cadence. Data from each collection were verified for completeness directly after collection.

After the first data collection, locations of EMG electrodes and markers placed on the skin were marked using a permanent marker for greater repeatability between sessions. This method proved to be problematic, however, as several of the riders went swimming between sessions and cleared the marks accidentally. EMG and marker placement was duplicated as best as possible for these subjects during the second session.

Modeling Methods

Since Vicon software does not include an algorithm well-accepted in the biomechanics community for computing dynamic muscle lengths, kinematic data were exported in *trial*.trc format for use in OpenSim(Delp, Anderson et al. 2007). OpenSim is capable of producing a set of estimated muscle lengths during dynamic movement using the combined methodologies of the program contributors(Delp, Anderson et al. 2007). This method uses estimates of muscle origins and insertions based upon work performed by Dr. Scott Delp and others. The muscles are represented geometrically using line segments and via points to represent muscle wrapping about joints, condyles, and other structures(Anderson and Pandy 1999; Delp, Anderson et al. 2007).

A modified version of the BothLegs.osim example lower body model was used for this investigation. This model includes 86 soft tissue actuators and 24 degrees of freedom. The model was modified to ignore gravitational effects as no forces were available to input into the system. A custom *markerset*.xml file was created to allow the model to be driven using the PiG marker set.

In order to generate subject-specific muscle lengths, the model is first scaled using the toe condition static trial collected kinematics. Scaling factors in OpenSim are calculated as a function of marker placement. Markers used to scale each body are presented in Figure 4.

Measurements	Marker Pairs			
pelvis	RPSI	LPSI	LASI	RASI
thigh	RASI	RKNE	LASI	LKNE
shank	RKNE	RANK	LKNE	LANK
foot	RHEE	RTOE	LHEE	LTOE

Figure 4.

Once the model has been scaled, OpenSim performs internal calculations to move "expected" marker locations to better fit collected data and adjusts model posture to fit the static pose.

Once scaling is complete, the model can be driven more accurately using experimental kinematics. This is accomplished by attempting to synchronize the "expected" locations of the markers on each body determined in the scaling procedures with the location of the markers in the *trial*.trc file frame-by-frame using weighted static optimization techniques. For this analysis, greater weights were used for pelvic, knee, ankle, and foot markers, as their placement was deemed more reliable due to the presence of a distinct bony landmark. The RMS error between experimental kinematics and model kinematics is reported for each frame and output to a file. These errors are not included in this paper due to extreme length, but are available for later analysis if required. Once this procedure is complete, a file containing the model kinematics for the trial is generated and saved as *trial*.mot. In the next step, the last modeling computational step required for this study, the muscular actuators attempt to duplicate kinematics stored in the *trial*.mot file using Thelen's CMC algorithm(Thelen, Anderson et al. 2003). This algorithm, in the absence of applied forces, primarily attempts to reduce the average residual error between motion generated by the model actuators and the kinematics contained in the *trial*.mot file. This portion of analysis is referred to as a Reduced

Residual Analysis (RRA) in OpenSim Of particular interest in this study is that actuator length is held to be such that there is no slack between origin, insertion, and via points; that is, the distance between these points is minimized when calculating Musculo-tendon length. Once this RRA has been performed, Musculo-tendon length can then be plotted over the time course of the movement. In this study, the same muscles used for EMG analysis were also exported in this fashion.

Data Reduction

Kinematic data were processed in Workstation (Vicon Corporation, Oxford, UK). Point data and EMG data were collected using a 10 camera MX system with F-series cameras at 100Hz and a 16 channel Delsys Bagnoli system (Delsys Inc., Boston, MA). Point data were reconstructed, labeled, and filtered using a Woltring filtering routine with predicted MSE of 20 due to the data's predicted smooth motion. Gaps in the data are also filled using this implementation of the Woltring routine; trajectories are predicted using a quintinc spline algorithm. Trials were truncated to four crank cycles in which all markers were present and all data exported to text files for further analysis.

Joint positional data were generated using Vicon's OLGA processing scheme. Angles are calculated according to the PiG model(Vicon 2008) with the addition of calculations performed by the OLGA; this algorithm essentially filters data by changing coordinate data to best represent rigid placement, as well as corrects for potential error and movement in the thigh and shank markers by adjusting their position to reduce cross-talk between knee flexion and rotation(Roren 2005).

EMG data were band-pass filtered between 6 and 600Hz using a custom MATLAB (Mathworks, Boston, MA) script.

Since the integrator used in OpenSim outputs 1000 data points per second of data, muscle length data were resampled to 100Hz using a custom-written MATLAB (MathWorks, Boston, MA) script. This script averages ten data point clusters and writes a new matrix containing these averages.

All kinematic data were then combined into text files via Microsoft Excel (Microsoft Inc, Redmond, VA) for use in MATLAB. This file contains coordinate data for the markers placed on the left and right pedal (RPED and LPED), joint angle data, and muscle length data bilaterally. Muscle length data during quiet standing was also included in a separate text file for normalization purposes. A separate text file containing EMG data was also included for each collection, as well as a separate text file containing EMG data from the T-V test for normalization purposes. In all, 15 text files were included for each subject, representing three collection points for each condition.

To begin processing, muscle lengths were first normalized to appropriate limb lengths(Sanderson, Martin et al. 2006) as generated through PiG processing. Muscle lengthening velocities and joint angular velocities were then calculated by differentiating length and joint position data with respect to time. After these values were found, data were split into crank cycles, averaged over all four crank cycles per collection point. EMG were then RMS processed using a 25ms moving window(Rouffet and Hautier 2007), normalized to the maximum value for each muscle found in the T-V test, and integrated, then all data were averaged over all three collection points. This resulted in a single set of data for each variable being analyzed for each condition per subject. Resulting from this set of processing/analyses were a single text file for each of these variables containing the appropriate 101 point vector or scalar: muscle length, muscle lengthening velocity, saggital joint angle, sagittal joint angular velocity, normalized EMG linear envelopes, and iEMG values. From these average vectors, RMS values were calculated for each condition for muscle length (compared to quiet standing length), muscle lengthening velocity (compared to zero), joint angles (also compared to zero, or angles in anatomical position for the reference frame used), and joint angular velocities (also compared to zero). These RMS values were then written to a text file for later statistical analysis.

Analysis

Significance for all t-tests and ANOVA comparisons was set at p=0.05. Joint kinematic data were imported into SPSS for analysis. The RMS values for each subject for each variable were calculated by comparing each normalized curve to a 101 point vector containing all zeroes, representing both ideal joint angle during quiet standing and joint angular velocities during quiet standing. The MATLAB script rmsd.m was used for this analysis. The RMS values were then compared using separate multiple ANOVAs for joint position and angular velocity data, resulting in two multiple ANOVA comparisons including sagittal plane data for the hip, knee, and ankle.

Muscle kinematic data were exported to Microsoft Excel for analysis. In order to investigate differences between muscle lengths and velocities without detecting concurrent changes (such as the TA necessarily shortening when the SOL lengthens as would be detected in an ANOVA) and to avoid accumulating large family-wise error through a series of individual t-tests, this data was plotted for each muscle and each variable per condition. This resulted in 14 plots, representing 7 muscles of the dominant leg (both conditions on each plot) and 2 dependent variables per plot. To show potential differences between curves, the 95% confidence interval (CI) for each curve was also plotted at 10% crank cycle intervals.

EMG curves were analyzed using cross-correlation according to the method of Li and Caldwell(Li and Caldwell 1999) to detect time domain changes and dependent samples t-tests used to detect differences in iEMG. Only the dominant leg for each subject was used for analysis to improve statistical power. Data averaged per muscle over all subjects were used for the cross-correlation analyses, resulting in 7 tests. iEMG were analyzed per muscle using averaged values as well, resulting in 7 dependent-samples t-tests.

CHAPTER 4

RESEARCH ARTICLE

Title:	The Effects of Cleat Placement on Muscle Mechanics and Metabolic Efficiency in Prolonged Sub-Maximal Cycling
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Abstract: The goal of this study was to quantify the changes in pedaling mechanics and energy expenditure accompanying a posterior shift in cleat placement during prolonged sub-maximal cycling. Seven male competitive cyclists from central Indiana were recruited to participate. Each subject was asked to complete two separate hour long rides, once using a traditional cleat placement, once using a novel heel placement. Expired gasses, lower limb kinematics, and EMG from 7 bilateral lower limb muscles (soleus (SOL), gastrocnemius, tibialis anterior (TA), bicep femoris long head, rectus femoris, vastus medialis, and gluteus maximus) were collected at three time intervals during each ride. No significant difference in O_2 utilization was seen between conditions ($p=0.905$). A significant difference was seen in sagittal plane knee angle ($p=0.008$) and angular velocity ($p=0.003$) RMS values. Muscle kinematic data analyzed using 95% confidence intervals comparing normalized curves showed no distinct differences. TA integrated EMG (iEMG) was higher in the heel condition, and SOL and TA showed differences in timing between conditions. These results demonstrate potential changes in ankling patterns and knee joint kinematics as adaptations to heel pedaling.

Introduction

Competitive cycling is a sport where victory is often decided by very small differences in ability and performance. Advances in cycling performance have come from a combination of athlete experimentation, commercial product development and testing, and formal research. Many advances in training protocols have come from metabolic expenditure research investigating changes in oxygen consumption and lactate threshold after riders undergo various training programs. Other improvements have come from advances in dietary research, such as caloric requirements, nutrient ratios, and supplementation. Still other advances have come from changes in factors extrinsic to the rider, such as cycle frame and gear design.

Though metabolic factors are important in cycling performance, perhaps a better direct predictor of cycling velocity is the mechanical power output of the cycle-rider system, as it is this power output directly that propels the rider(Abbiss and Laursen 2005). It is useful, then, to measure oxygen consumption at a particular power output or a particular velocity, as these measures have a more direct impact on cycling performance. If rider A is able to maintain a given velocity at a certain VO_2 while rider B is able to maintain the same velocity at a lower VO_2, rider B will have a comparative advantage(McArdle 2007). This concept of a given O_2 consumption at a given workload is called gross efficiency (GE) and is calculated simply by dividing work performed by the energy cost(Sidossis, Horowitz et al. 1992).

Many factors influence a rider's mechanical output, most of which are related to the configuration of the bicycle. When the configuration of a bicycle is altered in some manner, be it changing the seat height, handle bar position, or crank length, the rider's posture is altered. These alterations in posture can significantly affect factors related to muscle mechanics, such as muscle lengths and joint kinematics. One aspect of rider posture that is well understood is knee angle. It is recommended in the literature that a rider's knee be at between 25 and 35 degrees when the leg of interest is at dead drop(bottom position of the crank cycle) with the foot parallel to the ground(Peveler 2007). Using this position is a good general guideline for high power output, pedaling efficiency, and low rates of pain and injury.

Since changing the rider's posture can affect muscle mechanics and riding kinematics, it is important to understand how it is that the rider uses his or her muscles to generate the force needed to propel the bicycle at a desired speed(Raasch and Zajac 1999). The roles of various muscles have been investigated through EMG, force transducer instrumented pedals, and computational modeling(Redfield and Hull 1986; Brown, Kautz et al. 1996; Neptune and van den Bogert 1997; Raasch and Zajac 1999; Neptune, Kautz et al. 2000; Chen, Kautz et al. 2001; Baum and Li 2003; Duc, Betik et al. 2005; Raymond 2005; Sanderson, Martin et al. 2006; Bieuzen, Lepers et al. 2007). Though there is some individual variability and some aspects that may not be fully understood, uni-articulate muscles serve to add energy to the limb of their insertion while bi-articulate muscles serve to both transfer energy between segments as well as generate energy (Raymond 2005). When muscles shorten more quickly or are closer or further to their optimal length, their ability to produce force is altered in accordance with the length-tension-velocity relationship(Herzog 2000). Alterations in cycling posture and technique may elicit a favorable change in these relationships, allowing for greater force to be produced by a given muscle without additional energy expenditure. Therefore, it is important to investigate how these muscle lengths and shortening velocities are altered when aspects of cycling are perturbed. Since *in vitro* muscle lengths are not easily measured without expensive and sometimes impractical medical imaging devices, computational modeling is a way to make a strong approximation to physiological muscle kinematics. There are a number of methods available for modeling muscular kinematics, including simple geometric relationships, more complicated models using muscle wrapping, and using relationships between joint angles and muscle lengths directly measured through cadaveric study(van der Krogt, Doorenbosch et al. 2008). Sanderson et al have applied the methods of Grieve et al in a study of triceps surae kinematics, describing changes in the kinematics of the muscles at different cadences such as acting out of unison and over difference ranges of length and velocity(Sanderson, Martin et al. 2006). Sanderson et al found that the gastrocnemius operated over a shorter range of excursion at high cadences but with higher velocity, whereas the soleus decreased its range of motion as well but increased in shortening velocity to a lesser extent.

Altering the position in which a rider clips into the shoe-cycle interface may have a significant impact on muscular kinematics, energy expenditure, and neuromuscular dynamics. Currently, there is a paucity of evidence in the literature regarding what changes occur when cleat placement is altered, though potential benefits including increased pedal efficiency and a shift in force demand on individual muscles have been postulated(Frame 2007). It would be useful, then, to quantify these changes that occur with altered cleat placement through regimented protocol.

The purpose of this study was to examine the changes in energy expenditure through expired gasses, muscular kinematics through examination of muscular lengths and lengthening velocities, angular kinematics through joint angles and angular velocities, and neuromuscular excitation through timing changes and integrated EMG calculations associated with alterations in the antero-posterior position of cleat placement.

Due to the same external load being placed on the subjects and the results of Van Sickle et al (Van Sickle and Hull 2007)we do not expect to see a difference in oxygen consumption between conditions. Since the mechanics of the task will necessarily change in the heel position, however, we would expect to see an increase in TA activity and a decrease in SOL activity due to changing ankling patterns. The GAS will not experience a significant change due to its multiple roles as a bi-articulate muscle. We also expect increases in integrated EMG of the gluteus musculature as well as the biceps femoris due to the posterior placement of the cleats and increased reliance on hip extension to produce force. Since aspects of the subjects' cycle configuration other than cleat position will be held constant, it is not expected that joint angles and velocities or muscular kinematic changes will occur about joints above the ankle.

Methods

Subjects

Seven competitive male cyclists between the ages of 18 and 45 were recruited (mean=24.5+/-4.3) from the central Indiana area. After the procedure had been explained, each subject gave written, informed consent to participate. The procedures were approved by the Ball State University Institutional Review Board. To participate, subjects must ride at least 100 miles per week in-season, have competed at least once in the past calendar year, and be free from orthopedic injury.

Experimental Protocol

Subjects were brought in for an initial familiarization visit during which each rider rode for 10 minutes at 175 Watts in each cleat position on a Velotron cycle ergometer (CE) (RacerMate Inc., Seattle, WA). During this ride, any questions the rider had were answered, the next two collection sessions scheduled, and instructions for the following collection sessions were outlined. The traditional and heel cleat placement parameters were defined as 25% of the shod foot length posterior to the most anterior portion of the shoe and anterior to the most posterior portion of the shoe, respectively. Measurements from the subject's own cycle were also collected during this session and later used to fit the Velotron to the same dimensions.

Subjects completed their first collection session no more than one week after the familiarization session. The order of cleat placement for each subject was pseudo-randomized to maintain group equality. For each session, subjects were first fitted with 7 electrodes bilaterally on the soleus, gastrocnemius (GAS), tibialis anterior, bicep femoris long head (BF), rectus femoris (RF), vastus medialis (VM), and gluteus maximus (GM) after sites were palpated, shaved, abraded, and swabbed with alcohol. EMG data were collected using a Delsys Bagnoli (Delsys Inc., Boston, MA) system sampled at 2400Hz.

The subject's own road bike was then attached to a CompuTrainer (RacerMate Inc., Seattle, WA) ergometer and the ergometer was calibrated. Subjects next performed a 10 minute warm up period that included a thirty second torque-velocity (T-V) maximal EMG test(Rouffet and Hautier 2007). Subjects rode for 5 minutes at 75 Watts, then experienced a load of 9 Watts/kg for thirty seconds, a load similar to that experienced during a Wingate test(Inbar 1996). Subjects were instructed to maintain a normal riding posture during this test. After this 30 second T-V test, the load was reduced to 75 Watts for an additional 4.5 minutes. This procedure was performed in the traditional position for both collection sessions.

Subjects were next affixed with 39 retro-reflective markers in the Vicon Plug-In-Gait protocol (Vicon, Lake Forest, CA) and performed a 1 hour ride on the CE. Kinematic data were collected using a 10 F-series camera system at 100Hz at 10 minutes, 35 minutes, and 59 minutes, as were metabolic data using a Parvomedics metabolic cart

(Parvomedics, Sandy, UT) and EMG data. A marker was also placed on the pedal axis for use in time synchronization in MATLAB (Mathworks, Boston, MA). This procedure was followed for both collection conditions.

Data Reduction

Point data were filtered and interpolated using a Woltring filtering routine with a predicted MSE of 20. EMG data were band-pass filtered at 6-600Hz using a custom MATLAB script. O_2 data were averaged over two data points surrounding the collection time (30 seconds and 1 minute). Joint angles and lower body limb lengths were generated using Vicon's OLGA procedure. Musculo-tendon length data were generated using subject-specific models in OpenSim driven by collected kinematics. Data from the dominant leg were used for analysis.

Kinematic (joint, point, and muscle) data and EMG data were input in to a custom MATLAB script for further processing. Data were synchronized to crank cycles using the position of the marker on each pedal axis. Four crank cycles per collection period were normalized then averaged for each collection period, then all three collection periods per condition averaged due to lack of observed changes over time. Joint angular velocities and Musculo-tendon lengthening velocities were found by differentiating position and length data respectively. Average EMG envelopes were integrated as well.

Data Analysis

Root-mean-square (RMS) values were generated for joint kinematic data by comparing data to zero and were analyzed using a multiple ANOVA with repeated measures design. Muscle kinematic data were analyzed using 95% confidence intervals (CI) applied to the average curve for each measure. If the CI between conditions did not overlap, differences were considered significant. EMG timing differences were analyzed using cross-correlation according to the method of Li and Caldwell(Li and Caldwell 1999). iEMG was compared using one dependent samples t-test per muscle to avoid masking of potential differences that may occur due to co-contraction. O_2 data were compared using a dependent samples t-test.

Results

Statistical significance for all applicable tests was set at p=.05. Six of the seven initial subjects were used in analysis; one was excluded because of failure to complete the hour long ride in the heel condition due to fatigue.

No differences were seen between conditions in oxygen utilization using a dependent samples T-test (p=.905). The knee was more extended in the heel position and knee angular velocity experienced less extreme positive and negative peaks while no differences were seen at other joints (see Tables 1 and 2).

--Table 1--

--Table 2--

A summary of results for EMG data is presented in Table 3.

--Table 3--

Differences in EMG timing were detected at the SOL, GAS, TA, and GM. SOL exhibited low level activity for most of the crank cycle in the heel position instead of a clear bursting pattern. GAS exhibited a shorter burst at a similar time in the crank cycle in heel position as well as a clear second burst near 90% of the crank cycle. The TA showed a longer burst early in the cycle and a longer, more distinct burst near the middle of the cycle in the heel position. GLUT activity showed slightly delayed burst and earlier termination in the heel position. These results are displayed in Figure 4.

--Figure 1---

Only TA showed a significant increase in iEMG in the heel position (p=.029). No other differences in iEMG were observed.

Musculo-tendon kinematics were compared using a confidence interval (CI) comparison approach. 95% CI were calculated at each 10% of the crank cycle for each kinematic variable per condition; if the CI overlapped, curves were not considered statistically different. Differences were not seen in Musculo-tendon kinematics CI measurements with the exception of GAS lengthening velocity slightly increasing in the heel condition at approximately 50% of the crank cycle.

--Figure 2---

Discussion

The purpose of this study was to examine the changes in energy expenditure through expired gasses, muscular kinematics through examination of muscular lengths and lengthening velocities, angular kinematics through joint angles and angular velocities, and neuromuscular excitation through timing changes and integrated EMG associated with alterations in the antero-posterior position of cleat placement. The results of O_2 consumption in this study agree with the findings of Van Sickle and Hull (Van Sickle and Hull 2007). Their results showed no change in O_2 consumption at three different cleat positions for a short duration ride and this was confirmed to hold true over longer duration rides as well in this study. No change in O_2 consumption in this protocol also shows no change in gross efficiency as mechanical work was maintained at the same level between conditions.

Changes in ankle kinematics, though strongly expected, were not observed in this study. This may be explained due to exceptionally high variability between subjects in this measurement. Figures 2 and 3 display ankle joint kinematics for the traditional and heel positions with variability expressed as 95% CI. The wide variability seen in the heel condition is likely the reason significant differences were not seen between conditions at this joint. The change in knee angle kinematics seen between conditions may be explained by the inability of the rider to modulate knee angle through ankle kinematics in the heel position. Using traditional cleat placement, riders may effectively change the length of their leg by plantarflexing; this pushes against the pedal axis and forces greater knee flexion if contact with the seat is maintained. Changes in muscular kinematics about the knee were not seen despite these changes in joint kinematics. While the lack of change in bi-articulate muscles may be attributed to kinematics of the other spanned joint, one would expect uniarticulate such as the VM to show a difference if knee joint kinematics showed a change from traditional to heel position. It is possible, then, that aspects of the modeling methods used to obtain muscle length and lengthening velocity data were not sensitive enough to detect these changes if they should exist. Some major sources of error that may affect outcomes are the simplicity of the marker set used, methods by which the model is scaled in OpenSim, and a generic placement of muscular origins and insertions based upon averaged cadaver and MRI data(Delp, Anderson et al. 2007). Changes in hip kinematics were not expected nor seen potentially due to the ability of the cyclist to modulate this angle with slight trunk position changes.

While several changes were noted in EMG timing, they were different from what was expected. The near-constant low levels of SOL activity in the heel condition may occur as a co-contraction to antagonize TA activity, though due to lack of SOL bursting does not directly corroborate this idea. It is also possible that subjects were attempting to use SOL as a means of controlling pedal kinematics which, having a much shorter foot segment lever, may have required near constant activation. Greater TA activity overall and a distinct burst midway through the crank cycle may reflect an increased reliance on the TA to maintain pedal kinematics and thus pedal efficiency(Chapman, Vicenzino et al. 2008). A more distinct second burst of activity in GAS may indicate a greater role as a knee flexor in the heel condition, though this is difficult to determine as it may also play a role in controlling pedal kinematics during the end of the pedal stroke as it crosses the ankle joint as well. No change in iEMG for either BF or GLUT was noted, indicating that posterior cleat placement may not change the average level at which hip musculature is recruited.

Changes in muscular kinematics at the ankle, though expected, were not seen. Again, this could be partially due to variability in ankling patterns employed by the subjects. While the root cause of this variability cannot be explicitly extracted from this data set, it is possible that subjects were not able to develop an optimized control pattern due to the severely shortened lever arm of the foot. It may also be that the criteria used for subject selection was not stringent enough to recruit sufficiently experienced cyclists; Chapman et al have reported levels of EMG timing variability in inexperienced cyclists that greatly exceeds that of elite cyclists(Chapman, Vicenzino et al. 2008) which would be compounded by the novelty of the heel placement task(Chapman, Vicenzino et al.

2007). Another source of variability in different variables could be the cadence chosen. While 90 RPM is typically cited as a normal cadence for experienced cyclists (Neptune, Kautz et al. 1997; Neptune and Hull 1999; MacIntosh, Neptune et al. 2000; Raymond 2005), several of the subjects noted verbally that they would have felt more comfortable at a higher cadence. It may be possible that variability would be reduced with a longer time to adapt to the task, potentially with extended training in this position, or if the cleat position is moved slightly more anteriorly to allow for greater use of the foot/ankle complex as a control mechanism for pedal kinematics.

In summary, joint kinematic changes reached statistical significance only at the knee, though this is predominantly due to wide variability in ankle kinemtaics. Musculo-tendon kinematic changes were not seen except for a brief period in the GAS. These lack of statistical changes may be accounted for, again, due to wide variability in the sample. EMG differences were seen between conditions and are summarized in Table 3.

Future research should approach the question of cleat placement, at least in part, as a motor learning task. The wide variability seen at the ankle in the heel condition suggests that riders may not have adapted optimally to the novel demands associated with posterior cleat placement, and thus a prolonged training study may help elucidate what biomechanical changes can occur once a rider is familiar with the task. Allowing riders to pedal at a freely chosen cadence to eliminate further novelty in the task may be advisable as well, though the metabolic, kinematic, and neurodynamical consequences of subjects potentially riding at different cadences would have to be considered.

Angle RMS t-test Value Summary Table				
	RMS Mean Traditional (+/-SD)	RMS Mean Heel (+/-SD)	p value	Change in Heel Condition
Ankle	0.45 +/- (.136)	0.33 +/- (.172)	0.145	none
Knee	1.84 +/- (.105)	1.72 +/- (.089)	*0.008	More Extended
Hip	1.2 +/- (.106)	1.2 +/- (.106)	0.84	none

Table 1
*significant at the 0.05 level
RMS values for angular position data and the biomechanical outcome. Angular position curves were compared to zero (anatomical position).

Angular Velocity RMS t-test Value Summary Table				
	RMS Mean Traditional (+/-SD)	RMS Mean Heel (+/-SD)	p value	Change in Heel Condition
Ankle	0.69 +/- (.2)	0.5 +/- (.27)	0.134	None
Knee	2.76 +/- (.18)	2.56 +/- (.15)	*0.003	lower peaks
Hip	1.78 +/- (.14)	1.76 +/- (.09)	0.623	None

Table 2
*significant at the 0.05 level
RMS values for angular velocity data and the biomechanical outcome. Angular velocity curves were compared to zero (quiet standing).

EMG Summary Table						
	iEMG Traditional Mean +/-SD	iEMG Heel Mean +/-SD	iEMG p value	C-C Bounds	Significant Shift	Difference in Heel Condition
SOL	315.23 +/- (224.24)	462.93 +/- (693.85)	.676	.258-.579	Yes	Very low, near constant activity rather than bursting
GAS	697.61 +/- (213.25)	338.97 +/- (213.1)	.051	.784-1.18	Yes	Later in cycle with clear second burst
TA	488.63 +/- (125.46)	1192.9 +/- (513.91)	*.029	.492-.733	Yes	Longer duration with distinct second burst at mid crank cycle
BF	488.09 +/- (224.05)	587.11 +/- (104.7)	.24	.965-.984	No	n/a
RF	259.94 +/- (126.11)	321.19 +/- (124.96)	.382	.932-.969	No	n/a
VM	639.7 +/- (152.21)	593.47 +/- (109.2)	.499	.959-.981	No	n/a
GM	175.37 +/- (110.03)	184.27 +/- (41.67)	.877	.987-.994	Yes	Later on and earlier off

Table 3
*significant at 0.05 level
EMG results for both conditions. Note that units for iEMG are percent activation*percent crank cycle.

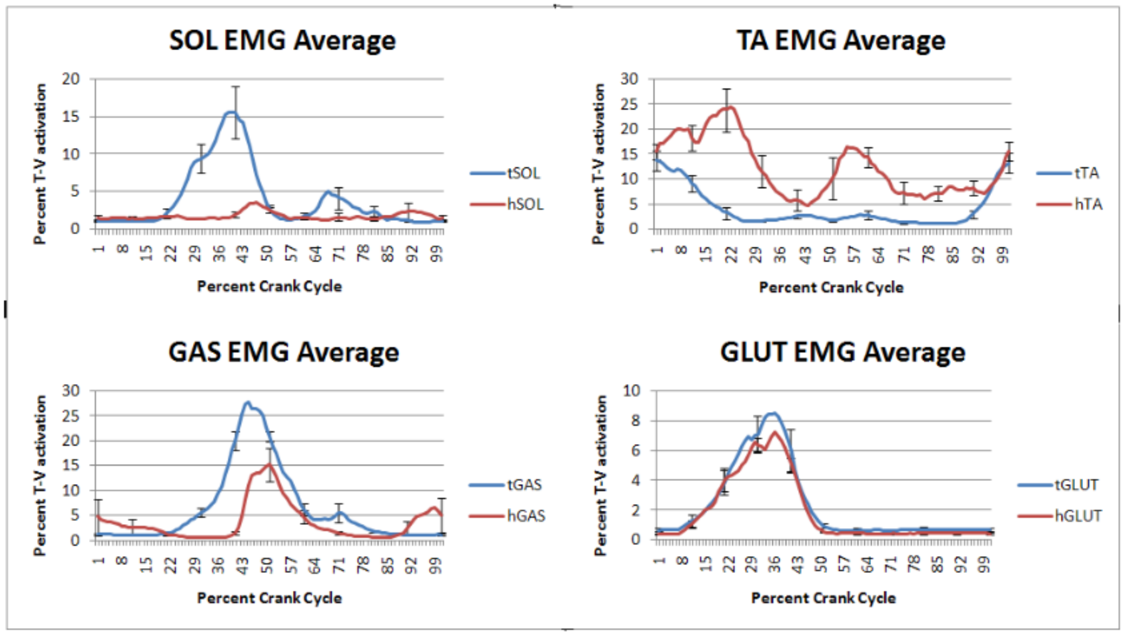

Figure 1
EMG graphs with significant shifts according to cross-correlation with 95% CI displayed

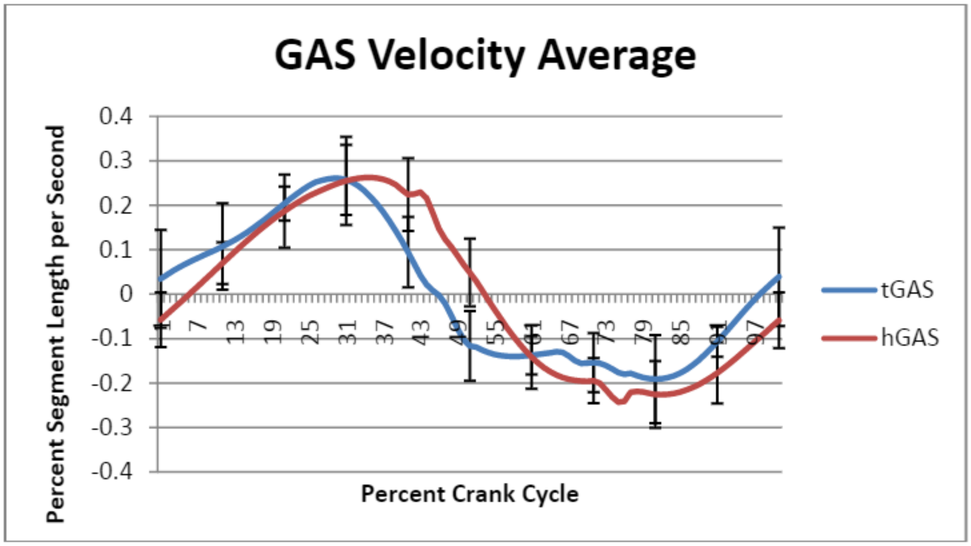

Figure 2
Musculo-tendon lengthening velocity of the gastrocnemius with 95% confidence interval displayed. Overlap of CI indicates no difference between curves.

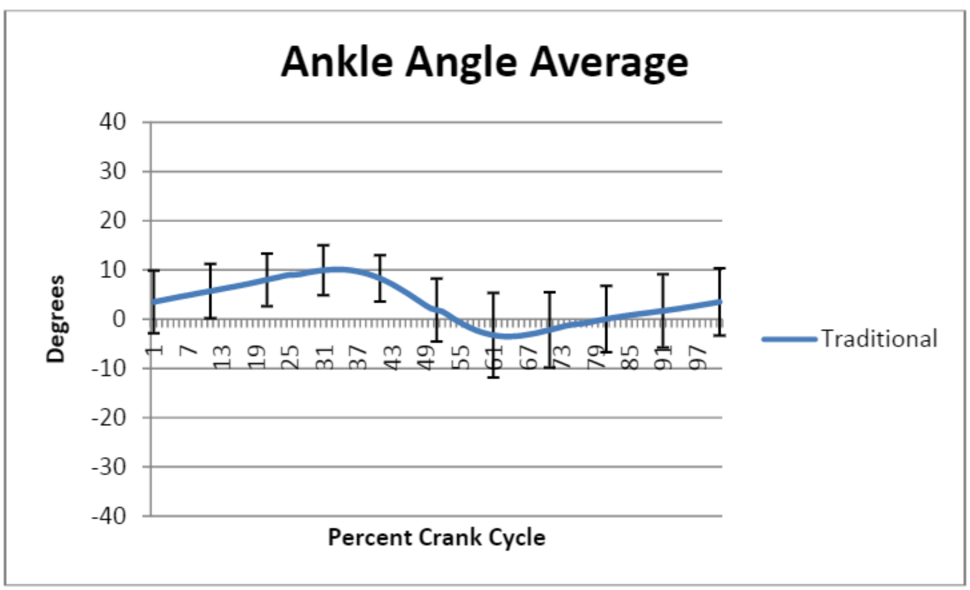

Figure 3
The traditional condition sagittal plane ankle angle displayed as a function of percent crank cycle with 95% CI displayed.

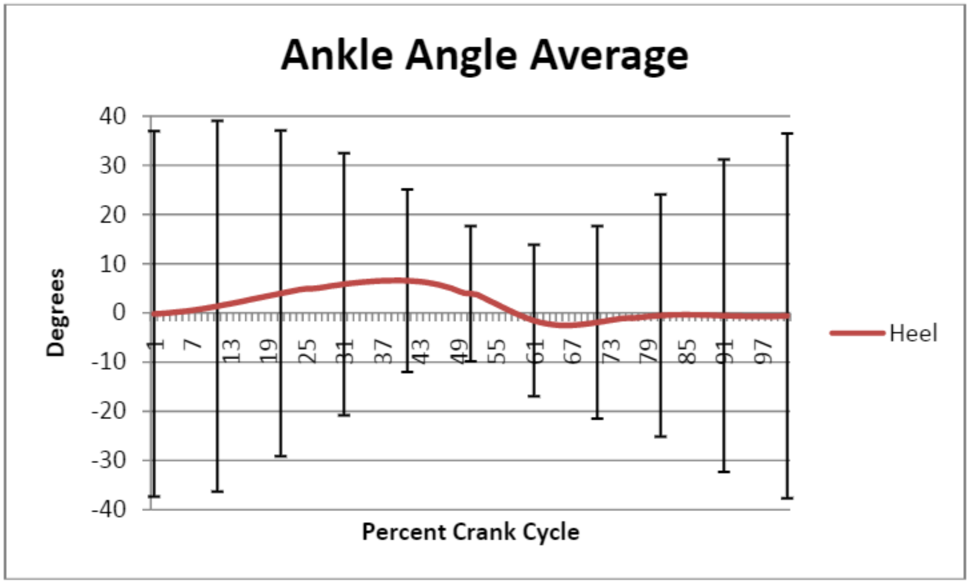

Figure 4
The heel condition sagittal plane ankle angle displayed as a function of percent crank cycle with 95% CI displayed.

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CHAPTER 5

SUMMARY AND CONCLUSIONS

Summary

Aspects of cycling performance have been a focus of investigation in biomechanics both as a means to itself, and also as a framework for investigating human movement within a constrained system. Many aspects of cycling have been well investigated, but the effect of cleat placement on metabolic, kinematic, and electromyographic outcomes in competitive cyclists remains unknown. The purpose of this project was to determine what, if any, differences existed in these factors when cycling in a traditional cleat placement v heel cleat placement. Six male competitive cyclists between the ages of 18 and 45 completed two 1 hour long rides on separate in two cleat positions while metabolic, kinematic, and EMG data were collected.

Oxygen consumption was not different between conditions (p=0.905), which is consistent with the findings of Van Sickle et al(Van Sickle and Hull 2007). Respective means for traditional and heel condition were 35.00ml/kg*min and 35.09ml/kg*min. Few joint kinematic changes were seen between conditions; only the knee joint angle (p=.008, more extended) and knee joint velocity (p=.003, smaller peaks) RMS values changed between conditions. No differences were noted in comparing the 95% confidence intervals for muscle length and velocity curves with the exception of a very brief difference in GAS velocity at around 50% of the crank cycle where the GAS was lengthening more slowly in the heel condition. Some differences were found in EMG timing through the use of cross-correlation(Li and Caldwell 1999) and iEMG values. A summary of those differences is included in Appendix A.

Conclusions

That no difference was found in oxygen consumption indicates that heel pedaling is at least no less efficient than traditional pedaling when completing an hour long ride at 175 Watts. This is important for competitive cycling as increasing or decreasing oxygen consumption at a given workload can have a large impact on overall performance(Abbiss and Laursen 2005). This also further corroborates the results of Van Sickle et al who showed that there is no difference in oxygen consumption over a shorter ride at a higher intensity in three difference cleat positions(Van Sickle and Hull 2007).

The lack of differences and large variability seen in joint and Musculo-tendon kinematic data indicate that, although changes may be occurring, the subjects that participated in this protocol did not adapt in a uniform manner. The exception to this, the knee joint, may have been forced into greater flexion in the heel position due to an inability of the subject to effectively shorten the leg by plantarflexing against the axis of the pedal that the rider may take advantage of in the traditional position. The data thus suggest that variability in ankling patterns is of primary concern when analyzing cleat placement and a rider's kinematics in the traditional position may have an impact on their kinematics and, by extension, EMG patterns, in other positions. This idea is further supported by an overall increase in TA activity and differences in several muscles regarding EMG timing despite limited change seen in muscular kinematics.

Recommendations for Future Research

Given the high level of variability seen in many measures in this study, it is recommended that further research in cycling related to cleat placement consist of large samples. With a small sample size, changes that may be occurring can easily be washed out by variability between subjects when the cleat is placed at the heel. In order to control for this variability, it is also advisable to screen riders for similar kinematics in the traditional cleat position. It may also be helpful to control for the fit of the bike to the subjects. In this study, the ergometer was adjusted to mimic the subject's own bike, but the configuration of the subjects' bikes was not controlled.

Given such high variability in kinematics and the increase in TA activity that was observed, it may also be more appropriate to focus investigation on cleat placement nearer the middle of the foot to allow the plantarflexors to play a larger role in controlling pedal dynamics. Alternatively, it may be advisable to investigate changes in musculoskeletal variables over time in a training study to determine if heel pedaling is simply a complicated task that requires a longer period to adapt than was provided in this protocol.

CHAPTER 6

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Appendix A

IRB APPLICATION

1. **TITLE, PURPOSE, RATIONALE**

1.1 Title. Effects of cleat placement on muscle mechanics and metabolic efficiency in steady-state cycling

1.2 Purpose of the study. The purpose of this present study is to investigate the differences in metabolic expenditure, muscle mechanics, and joint angles when pedaling for a prolonged ride using a traditional cleat placement or heel cleat placement.

1.3 Rationale. Competitive road cyclists, like most competitive athletes, are always looking for a way to improve performance. Cycling has received much attention in biomechanics and physiological research, but one factor affecting performance that has not received much attention is cleat placement with respect to foot position. Altering cleat placement has the potential to advantageously change cycling kinematics, muscle mechanics, and metabolic efficiency. The results of this study will provide information useful in determining if using a posterior placement of the cleat provides any performance advantage over traditional cleat placement for competitive cyclists.

Definition of Terms:

Modeling/simulation: Using sophisticated computer software to recreate human movement and physical characteristics.

Cleat: A part of the shoe that hooks in to the pedal on a competitive road cycle.

2. DESCRIPTION OF THE SUBJECT POPULATION

2.1 Number of subjects. Twenty participants will be enrolled in the study.

2.2 Describe the subject population. Participants will be apparently healthy men between the ages of 18 and 35 years.

2.3 Describe specified inclusion/exclusion criteria. Participants must be non-smokers, free from metabolic disease, ride in excess of 100 miles per week during the normal cycling season, and have competed in at least one road race in the past year. Any known disease or medications used that affect neuromuscular and/or cardiovascular health will result in exclusion from the study.

3. SUBJECT RECRUITMENT

3.1 Describe method of participant recruitment. Prospective participants will be recruited from the Ball State Cycling Club and the Delaware County Cycling Club by word of mouth and email solicitation.

4. METHODS AND PROCEDURES

4.1 Describe the methods and procedures used. Data collection for this study will take place at the Ball State Biomechanics Laboratory. Participants will be asked to visit this facility on two occasions, each less than a week apart, for approximately 3 hours per visit. Another visit to the campus biomechanics laboratory will be required to familiarize the participants with a heel-clipped riding posture. This visit will take approximately half an hour and the participant will be able to read and sign the informed consent form and ask any questions as well. Participants will also fill out a short questionnaire concerning inclusion/exclusion criteria at this visit.

Each data collection session will begin with participants changing into cycling-appropriate clothing including cycling shorts and athletic shoes. Participants will wear a specialized vest as well that is designed to minimize cable clutter in the to-be-described electromyography (EMG) collection system. Once the participant is properly attired, surface EMG electrodes will be placed on eight locations on both legs for a total of sixteen electrodes. These sites include the soleus, medial gastrocnemius, biceps femoris, lateral hamstring, rectus femoris, vastus medius, tibialis anterior, and gluteus maximus. The sites for electrodes will be shorn, lightly abraded, and cleansed with alcohol to improve signal fidelity. Once placed, electrode signal clarity will be tested using functional tests such as a bodyweight squat, and toe raise, in order to ensure signal clarity. Once placement has been verified, electrodes will be secured using athletic training pre-wrap and derma-tape, with cords and collection equipment properly secured on the vest. Next, 18 retro-reflective markers will be placed on the lower body in a standard Vicon Plug-in-Gait marker set to define the foot, lower leg, and thigh in space. The participant will then begin a fifteen minute warm up on a Velotron cycle ergometer. Feet will be affixed to the pedals using a custom-made device that allows for configuration in any position along the antero-posterior axis of the foot. Which position, toe or heel, each participant is tested on for their first data collection will be pseudo-randomized to ensure an even number of each case. The warm up will consist of 7 minutes of riding at 75 Watts, a 30 second effort at a load of 9 Watts/kg of body mass, and an additional 7 minutes and 30 seconds of riding at 75 Watts. The high load used in the middle of the warm up will be used for EMG maximal effort normative data and collected at 2kHz. After the warm-up period is finished, riders will ride for one hour at a load of 175 Watts. Three data collections will be done during this ride at 10 minutes, 35 minutes, and during the last minute of riding. Kinematics will be collected using a 10-camera Vicon system collecting at 60Hz. EMG will be collected at 2kHz. Metabolic data will be collected using a ParvoMedic metabolic cart. For this part of the data collection, subjects will wear a headgear attached to a 1-way valve and mouthpiece that directs expired air to the metabolic cart. The headgear will only be worn during the collection period. All data will be collected for one minute at each collection time. After the final collection, participants will be given a five minute cool down at 75 Watts and all collection equipment will be removed. The participant will then come for an identical testing session using the cleat placement not yet investigated within one week.

Musculoskeletal Modeling

Once 3-D kinematic data has been collected, this data will be used to produce a subject-specific model using commercially available musculoskeletal modeling software. This process will require no extra commitment from participants. From these subject-specific models information about muscle lengths and shortening velocities, two important aspects of muscle mechanics and force production, will be generated.

Statistical Analysis

Metabolic data will be compared across the groups between the two conditions at each data point using a two-way (cleat position x time point) multiple ANOVA. EMG data will be normalized to the collection during warm-up per visit per individual. Integrated EMG per quadrant will be analyzed across the entire group, between conditions, using a quadrant approach. The values at 0% (top position), 10%, 20%, to 90% of a crank cycle will be compared using a 2-way multiple ANOVA. Significance for all ANOVA analysis will be set at 0.05. Muscle lengths will be generated directly using the modeling package, and shortening velocities will be determined by differentiating length values. These values will be compared using confidence intervals at the same percentages of a crank cycle.

If the confidence intervals for the splines across conditions overlap, the values will not be considered to be statistically different. If the confidence intervals do not overlap, values will be considered to be statistically different.

5. ANONYMITY/CONFIDENTIALITY OF DATA

5.1 Describe how data will be collected and stored. Subject confidentiality will be protected to the maximal extent allowable by law. Digital data will be kept on a password protected, encrypted external hard drive. Paper records will be kept in a locked filing cabinet in the Biomechanics Laboratory. Files will be maintained for a minimum of three years following study completion. Participants will be assigned a generic number for all communications, forms, and publications.

6. POTENTIAL RISKS AND BENEFITS

6.1 Describe potential risks and discomforts. As with any cycling activity, there is a small risk of injury associated with cycling according to this protocol. Participants may also feel slightly uncomfortable during the placement of the gluteal EMG electrode. There is also a slight risk infection associated with shaving the skin for electrode placement.

6.2 Describe how risks will be minimized. The population used for this study minimizes the risk of injury as they are relatively young, healthy, and free from orthopedic problems. All parts of the procedure will be conducted in a professional manner, and sterile procedures will be used when shaving and abrading the skin. Single-use razors will be used and disposed of in a hazardous waste container.

6.3 Describe potential benefits. Results of this study will provide insight into optimal cleat placement for competitive road cyclists.

7. SUBJECT INCENTIVES/INDUCEMENTS TO PARTICIPATE

7.1 Describe any incentives/inducements to participate that will be offered to the subject. Participants will be given a Motion Analysis and Therapy Complex branded water bottle at the completion of the second data collection. Participants will also receive performance information regarding their oxygen use and cycling form.

8. OTHER FINANCIAL CONSIDERATIONS

8.1 Describe any financial expense to the subject. This study requires no financial responsibility of the participants.

8.2 Describe any provisions for compensation for research-related injury. Emergency medical treatment is available in the event of injury. The subject will assume all costs of medical care that is provided. In the unlikely event of injury or illness of any kind as a result of participation, Ball State University, its agents and employees, will assume whatever responsibility is required by law.

9. INFORMED CONSENT

9.1 Prior to participation in the study, participants will be given a verbal and written explanation of study procedures, after which they will be asked to provide informed consent. Even after agreeing to participate and providing signed consent, participants are free to withdraw from the study at any time at no penalty of any kind.

Appendix B

INFORMED CONSENT DOCUMENT

Consent to Participate in a Research Study

EFFECTS OF CLEAT PLACEMENT ON MUSCLE MECHANICS AND METABOLIC EFFICIENCY IN STEADY-STATE CYCLING

WHY AM I BEING INVITED TO TAKE PART IN THIS RESEARCH?

You are being asked to participate in research investigating advantages and disadvantages to clipping in to a road bike at the heel of the shoe instead of the traditional cleat placement. You are being invited because you are a competitive cyclist and are a member of the community this research may benefit.

WHO IS DOING THE STUDY?

The person in charge of this study is Daniel Leib, HFI, who is a graduate student in biomechanics at Ball State University. Dr. Eric Dugan, director of the Ball State Biomechanics program, will be supervising the research. Jeffery Frame, MS, lab manager, will be assisting during data collections as well.

WHAT IS THE PURPOSE OF THIS STUDY?

Since some high level riders are making the choice to use cycling shoes with a non-standard cleat placement, it is important to understand how this affects the way a person rides. The purpose of this study is to determine advantages and disadvantages associated with using a non-traditional cleat placement during a prolonged ride under typical road cycling conditions. Different factors investigated will be energy used by the rider, how moving the cleat changes posture, and how muscle involvement is changed.

WHERE IS THE STUDY GOING TO TAKE PLACE AND HOW LONG WILL IT LAST?

The research procedures will be conducted at the Ball State University Biomechanics Laboratory, Muncie, IN.

You will need to attend the Biomechanics Laboratory on 3 occasions. The first time will be to sign the informed consent form, ask any questions, and ride a stationary cycle for 30 minutes to get used to the new cleat placement. The second and third visits to the Biomechanics Laboratory will be for testing. You will be at the lab for these visits for approximately two and a half hours. These visits will be scheduled within the same week, and at the same time of day each time.

WHAT WILL I BE ASKED TO DO?

You will be asked to ride on a Velotron cycle ergometer, a stationary bicycle, for one hour and fifteen minutes per collection session. The first fifteen minutes of each ride will be a warm up at 75 Watts that includes a short maximal effort burst to calibrate our equipment. You will then ride for an hour at 175 Watts, a typical load for an amateur competitive cyclist. During this ride, we will collect motion capture data using small markers taped to your skin, muscle activity data using small electrodes also taped to the skin, and energy usage using a metabolic cart. Energy usage measurements will require wearing a headgear and mouthpiece for the duration of three short collections during the rides only. The headgear will not be worn when this measurement is not being collected.

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

There are no expected serious risks or discomforts expected from this study. As this study does require riding a bicycle for an hour at a time, some fatigue normal for exercise may be experienced. Electrode placement does require shaving small areas of hair from the legs if you do not already shave normally and light abrasion to remove dead skin cells. There will be a short maximal effort bout early in the ride that may pose a risk of injury, but this risk is no more than you might experience in normal training.

WILL I BENEFIT FROM TAKING PART IN THIS STUDY?

Participation in this study may not directly benefit you, though energy expenditure data and feedback on riding posture will be provided to you if desired. The data from this study has the potential to provide information valuable to cleat placement advantages and disadvantages for competitive cyclists.

DO I HAVE TO TAKE PART IN THE STUDY?

Taking part in this study is completely voluntary, and, should you chose to participate, you may withdraw at any time and for any reason without penalty.

IF I DON’T WANT TO TAKE PART IN THE STUDY, ARE THERE OTHER CHOICES?

If you choose not to participate in the study, or do not meet the study inclusion criteria, there are currently no other studies being performed by the primary investigator.

WHAT WILL IT COST ME TO PARTICIPATE?

There will be no cost to participate in the study. You will have to pay for transportation costs to and from the Ball State Biomechanics Laboratory if applicable.

WHO WILL SEE THE INFORMATION THAT I GIVE?

We will keep confidential all research records that identify you to the extent allowed by law. When data is reported, you will be identified with an assigned number related to initial study date and the date you joined the study rather than your real name. The data from this study may be published either in scholarly journals or through presentations at conferences, but you will not be individually identified in any way.

All records will be stored in a locked file cabinet in a locked room or on a password protected computer using standard 128-bit encryption protocols. This electronic data will be stored at the Ball State Biomechanics Laboratory for at least three years. Only the primary investigators and members of the research team will have access to these records. By signing this form, however, you allow the research investigators to make my records available to the Institutional Review Board (IRB) Offices at Ball State University and regulatory agencies as required by law.

CAN MY TAKING PART IN THE STUDY END EARLY?

You are free to withdraw from the study at any time and for any reason without penalty if you so choose. It may also become necessary for you to be excluded from the study by the investigators; if this becomes necessary, there will be no other penalty to you. Any information already gathered on your cycling performance will still be made available to you if you desire.

WHAT HAPPENS IF I GET HURT OR SICK DURING THE STUDY?

If you believe you are hurt or if you get sick because of something that is done during the study, you should call Daniel Leib at 765-285-5178 or at 314-910-8057 immediately. If an accidental injury occurs during your participation in the study, appropriate emergency measures will be taken. The primary investigator is fully certified in Red Cross CPR and emergency First Aid. It is important for you to understand that if you suffer from an injury as a direct result of participation in this research, you should obtain medical care in the same manner as you would ordinarily obtain treatment. No compensation (such as lost wages, medical cost reimbursement, lost time or discomfort) or additional treatment will be made available to you except as otherwise specified in this consent form. You are not waiving any of your legal rights by signing this consent form.

WILL I RECEIVE ANY REWARDS FOR TAKING PART IN THIS STUDY?

You will receive a university-branded water bottle for participating in this study. Your cycling performance data collected in this study will be made available to you as well if desired.

WHAT IF I HAVE QUESTIONS?

Please ask any questions you might have before signing this consent form and agreeing to participate; if you have any questions at a later date, you may contact the primary investigator Daniel Leib at 765-285-5178 or Eric Dugan at (765) 285-5139. If you have any questions about your rights as a volunteer in this research, contact Melanie Morris, Coordinator of Research Compliance, Office of Academic Research and Sponsored Programs, Ball State University, Muncie, IN. 47306 (765) 285-5070, irb@bsu.edu. We will give you a copy of this consent form to take with you.

WHAT ELSE DO I NEED TO KNOW?

You will be informed if any new information is learned which may influence your willingness to continue taking part in this study.

CONSENT

I have read the above information about "Effects of cleat placement on muscle mechanics and metabolic efficiency in steady-state cycling" and have been given an opportunity to ask questions. I agree to participate in this study and I have been given a copy of this consent document for my own records.

____

Participant Signature Date

____

Address

____

Participant Name Printed

____

Name of person providing information to the participant Date

____

Signature of Investigator

Principal Student Investigator	Thesis Advisor
Daniel Leib, HFI	Eric L. Dugan, Ph.D.
Ball State University	Ball State University
Biomechanics Laboratory	Biomechanics Laboratory
McKinley Avenue, PL 202	McKinley Avenue, PL 204
Muncie, IN 47306	Muncie, IN 47306
(765) 285-5178	(765) 285-5139
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Appendix C

PAR-Q

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO	
☐	☐	**1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	**2. Do you feel pain in your chest when you do physical activity?**
☐	☐	**3. In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	**4. Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	**5. Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	**6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	**7. Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ________________________________

SIGNATURE ____________________________ DATE ____________________

SIGNATURE OF PARENT ____________________ WITNESS ____________________
or GUARDIAN (for participants under the age of majority)

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

 Supported by: Health Canada Santé Canada

continued on other side...

Appendix D

IRB APPROVAL

BALL STATE UNIVERSITY

ACADEMIC AFFAIRS
OFFICE OF ACADEMIC RESEARCH AND SPONSORED PROGRAMS

Muncie, Indiana 47306-0155
Phone: 765-285-1600
Fax: 765-285-1624

INSTITUTIONAL REVIEW BOARD

DATE:	May 14, 2008
TO:	Daniel Leib
FROM:	Institutional Review Board Leonard Kaminsky, Chair
RE:	IRB protocol # 82928-4
TITLE:	The Effects of Cleat Placement on Muscle Mechanics and Metabolic Efficiency in Prolonged Sub-Maximal Cycling
SUBMISSION TYPE:	Revision
ACTION:	APPROVED
APPROVAL DATE:	05/14/2008
EXPIRATION DATE:	03/18/2009
REVIEW TYPE:	Expedited

The Institutional Review Board has approved your modification for the above protocol, effective May 14, 2008 through March 19, 2009. All research under this protocol must be conducted in accordance with the approved submission.

As a reminder, it is the responsibility of the P.I. and/or faculty sponsor to inform the IRB in a timely manner:

- when the project is completed,
- if the project is to be continued beyond the approved end date,
- if the project is to be modified,
- if the project encounters problems, or
- if the project is discontinued.

Any of the above notifications should be addressed in writing and submitted electronically to the IRB (http://www.bsu.edu/irb). Please reference the IRB protocol number given above in any communication to the IRB regarding this project. Be sure to allow sufficient time for review and approval of requests for modification or continuation. If you have questions, please contact Research Compliance at (765) 285-5070 or irb@bsu.edu.

cc: Eric Dugan