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Bicycle Shoe Insoles and Their Effect on Lateral Knee Movement, Leg Muscle Activation Patterns, and Performance in Experienced Cyclists

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BICYCLE SHOE INSOLES AND THEIR EFFECT ON LATERAL KNEE MOVEMENT, LEG MUSCLE ACTIVATION PATTERNS, AND PERFORMANCE IN EXPERIENCED CYCLISTS

By

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BICYCLE SHOE INSOLES AND THEIR EFFECT ON LATERAL KNEE MOVEMENT, LEG MUSCLE ACTIVATION PATTERNS, AND PERFORMANCE IN EXPERIENCED CYCLISTS

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Orthotic insoles in cycling shoes are an intervention used to correct pedaling mechanics in riders, which has received little attention in the literature. This study was designed to test the hypothesis that the use of orthotic insoles in cycling shoes would alter the pedaling mechanics, muscle activity, and submaximal efficiency of healthy, experienced, male cyclists. Additionally, it was hypothesized that the insole that allowed the lowest level of lateral knee movement would produce the greatest improvements in these variables, and be related to the rider’s arch height. Nine cyclists were evaluated during four VO$_{2\text{max}}$ tests, using four different insole conditions (flat [no insole], low, medium, and high arch support) in a random order. High-definition video recordings were used to measure lateral knee movement, wireless electromyography to measure muscle activity, and telemetry-based gas analysis to determine cycling efficiency. VO$_{2\text{max}}$ tests were performed at least 48 hours apart to control for fatigue. The non-flat insole that resulted in the lowest level of lateral knee movement was identified for each leg: Spearman rank-order correlations showed no relationship between arch variables and this “best fit” insole. Because the best fit insole was not the same between feet for most participants, general linear mixed models were run 2 ways, with the best insole for the dominant leg and non-dominant leg identified as the overall “best fit” insoles. When the best fit for the dominant leg was identified as the overall “best fit” insole, it produced effects on
dominant knee lateral movement ($p=.001$) and heart rate at anaerobic threshold ($p=.014$). The non-dominant “best fit” insole had a significant effect on heart rate at anaerobic threshold ($p=.017$). There was also an interaction effect in the dominant leg hamstrings ratio between insoles and pedal float type ($p=.007$). The implication of these findings is that orthotic insoles may be an effective intervention to alter pedaling mechanics and upper leg muscle activation ratios about the knee, but have little effect on cycling performance.
For my parents
They always told me that a doctorate was a marathon, not a sprint. The past three-odd years have certainly proved that:

Thank you to DM&RA, for all your support, encouragement, and kicks in the pants when I needed them. I love you both dearly and wouldn’t be where I am without your help.

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To my committee: The brainstorming, the guidance, the encouragement, and the grounding… it has helped me develop as an academic, a person, and (in this case) an author. I remember this manuscript being a 500-word, 5 page list of bullet points. Your commitment to seeing me through this process has resulted in the manuscript here, and I’m eternally thankful to all of you for your help.

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Chapter 1

INTRODUCTION

Riders are fit to their bicycles at three points: the hands, the hips (ischial tuberosities), and the foot-shoe-cleat-pedal (FSCP) interface (Figure 1). There are three basic techniques that are used to effectively achieve this fit: experienced-based direct observation, technology-assisted indirect observation, and anthropometrically-based formula fitting (Hogg, 2012). None of these techniques take full advantage of existing knowledge concerning the FSCP interface and its potential effects on the mechanics of associated movements of the legs.

FSCP Interface

Researchers have examined the individual components of the FSCP interface, not considering how the structure can affect the mechanics of a rider’s entire lower extremity and performance when considered as an operational unit. Additionally, problems with the FSCP interface have the potential to cause discomfort or injury throughout the cyclist’s body (Asplund & St. Pierre, 2005).

Examination of the FSCP interface is confounded by the complex structure of the foot (Callaghan, 2005; Wozniak-Timmer, 1991). Foot inversion or eversion has been shown to alter knee moments during cycling (Ruby, Hull, Kirby, & Jenkins, 1992; Gregor, 1994; Johnston, 2007; Sanner & O’Halloran, 2000). Traditionally, the shoe-cleat interface has been examined for two purposes: equalizing leg length discrepancies and determining optimal fore-aft cleat placement (Asplund & St. Pierre, 2005; Callaghan, 2005). Spacers
have been used to correct leg length discrepancies, while fore-aft cleat placement has been examined for its influence on gastrocnemius activation (Ruby et al, 1992; Johnston, 2007; Sanner & O’Halloran, 2000; Van Sickle & Hull, 2007). In the literature, the cleat-pedal connection is not as well researched as the foot-shoe or shoe-cleat components. Research has also shown that increasing the freedom of transverse rotation of the foot relative to the pedal (float) will decrease the related joint moments and thus risk of injury (Asplund & St. Pierre, 2005; Callaghan, 2005; Ruby & Hull, 1993; Sanner & O’Halloran, 2000). Recently manufacturers have begun selling shoes packaged with modular insoles to compensate for each rider’s unique foot structure. Because of foot and ankle anatomy, changing any component within the FSCP interface should affect the mechanics of more proximal structures. These mechanics determine how wedges, spacers, or insoles can be used to facilitate the effective transfer of force from the legs to the pedals (Yang, 2013).

**Proposed Mechanisms**

Although classified as a hinge joint, the structure of the tibiotalar joint allows an element of rotation. Orthotic insoles are designed to control foot pronation or supination, thereby reducing rotation of the tibia. Bringing the foot to a neutral position may help align the leg, increasing comfort, performance, and safety (Yang, 2013). There is considerable research establishing the use of insoles to support or correct motion in the tibiotalar joint of runners (McMillan & Payne, 2008; Eng & Pierrynowski, 1994); but little research exists examining their effectiveness in producing similar corrections in cyclists during pedaling. To our knowledge, only one study has examined insole use in cyclists, and a limited number of variables (ankle frontal plane inversion/eversion, knee joint ROM, VL
EMG, muscle power) were assessed. Of these, only ankle motion and muscle power demonstrated significant changes due to insole condition (Yang, 2013). These findings support the hypothesis that insole use can affect foot position, and potentially, performance in cyclists.

**Objectives of the Proposed Study**

The purpose of the current study examined the impact that four levels of insole support can have on kinematics, muscle activation levels (EMG), and performance in moderately active male cyclists. All participants were tested under all insole conditions, so it was possible to identify which insole provides the “best fit”; operationally defined as the insole that causes the least amount of lateral knee movement while pedaling. Additionally, the strength of the relationships between individuals’ arch heights and their best-fit insoles were evaluated.

Because the impact of insoles during cycling is not known, this study tested the hypothesis that there would be at least one insole for each subject that would minimize lateral knee movement while pedaling. The effect of this reduction in movement was expected to alter or reduce EMG activation levels of the quadriceps and hamstrings and improve performance during a VO$_{2\text{max}}$ test performed on a cycle ergometer adjusted to simulate the cyclist’s road bicycle.
Chapter 2

METHODOLOGY

Participants
This study was approved by the Medical Science Institutional Review Board of the Human Subjects Research Office at the University of Miami. All participants read and signed a written informed consent approved by the review board. Participants were recruited from local cycling groups. Potential participants were required to have more than one year of cycling-specific training, to have used their current pedals for the past 3 months, and to meet the requirements of the PAR-Q and American Council on Exercise (ACE) health history questionnaires.

Experimental Procedures
A CONSORT flow chart detailing the number of participants recruited, screened, enrolled, tested, and analyzed is shown in Figure 2. After providing written consent, qualified participants lay on an athletic training table, and measurements were taken of their left and right legs (anterior superior iliac spine [ASIS] to medial malleolus), femurs (greater trochanter to lateral knee joint line), and tibias (lateral joint line to lateral malleolus). Standing and sitting arch heights, foot lengths (from medio-lateral axes at the front of the great toe to the rearmost projection of the heel), and heel-1st metatarsal lengths were measured. All measurements were made by the same certified athletic trainer (ECC), and are shown in Table 1. Participants were also required to bring their normal road bicycle to their screening visit, and the laboratory cycle ergometer (Velotron
Pro, Racermate, Seattle, WA) was configured as closely as possible to the participant’s road bike. The participant then mounted the ergometer and further adjustments were made using the following parameters: (1) bottom bracket-to-saddle height, (2) front of saddle-to-handlebars distance, (3) handlebars-to-bottom bracket distance, (4) bottom bracket-saddle setback, and (5) crank length. These values were recorded and used during all subsequent trials, and participants were asked to not alter any dimensions of their normal road bicycle nor adjust shoe cleat position for the duration of their participation in the study.

Next, each subject’s foot measurements were taken with the JAK-Tool Arch Height Index Measurement System (AHIMS; JAK Tool and Model, Cranbury, NJ). The AHIMS was used to measure subject foot length, heel-1st metatarsal length, and arch height while sitting. This was their “unloaded” arch height. Foot length, heel-1st metatarsal and arch heights were also taken while each subject was standing. To control the amount of load they were placing on their longitudinal arches, participants were asked to put most of their weight on the foot being measured; the contralateral foot could remain on the ground for balance purposes only. The difference between sitting (unloaded) and standing (loaded) arch height was called “arch compliance”, or informally, “collapse”. Following these procedures, participants were given an opportunity to familiarize themselves with the Velotron ergometer and other equipment and appointments were made for all testing sessions.

For every laboratory visit, participants were instructed to arrive well rested (8-hrs of sleep), well hydrated (500ml of water within an hour of experimental trial), not having
consumed caffeine, and fasted over the preceding 8-hour period. They were required to bring their normal cycling shoes and pedals. Their pedals were affixed to the Velotron. They were also given the option of bringing their own road bicycle saddle.

**Maximum Oxygen Uptake**

Respiratory gases were collected continuously during testing using a portable ergospirometry device (Oxycon Mobile, Jaeger, Hoechberg, Germany). The Oxycon Mobile system has been shown to accurately measure VO$_2$, V$_E$, and RER at submaximal workloads (Rosdahl, Gullstrand, Salier-Eriksson, Johansson, & Schantz, 2010). Prior to data collection the participant was fitted with a mask that covered both the nose and mouth. Breath-by-breath data included respiratory rate (RR; breaths·min$^{-1}$), pulmonary ventilation (V$_E$; L·min$^{-1}$), oxygen uptake relative to body weight (VO$_2$; ml·kg$^{-1}$·min$^{-1}$), expired carbon dioxide relative to body weight (VCO$_2$; ml·kg$^{-1}$·min$^{-1}$), and respiratory exchange ratio (RER). By default, the Oxycon Mobile system takes averages over five-second intervals of breath-by-breath. Heart rate (HR) was continuously measured using a Polar T31 Coded Transmitter (Polar Electro Inc., Lake Success, New York), and transmitted via short-range telemetry to the Oxycon Mobile receiver unit. Prior to each testing session, air flow calibration was performed using the automatic flow calibrator and the gas analyzer was calibrated against a certified gas mixture of 16% O$_2$ and 4% CO$_2$. Participants preceded all VO$_{2\text{max}}$ tests with a 5-minute warm-up on the Velotron at a self-selected, comfortable resistance. Following the warm-up, the subject began the test at 100W, and the Velotron was programmed to increase resistance by 50W every two minutes, with the goal of reaching the subject’s VO$_{2\text{max}}$ in approximately 10 minutes.
(Buchfuhrer et al, 1983). The resistance levels were identical for every subject for each insole condition. If participants did not voluntarily terminate a VO2max test, the research team terminated the test based on two of the four following criteria: 1) a plateau in VO2 despite an increase in workload; 2) a heart rate within 10-15 bpm of age-predicted maximum; 3) an RER greater than 1.10; or, 4) a decrease in cadence below 80rpm.

**Electromyography**

Participants were prepared for electromyographical (EMG) data collection by initially locating the motor points for the right and left biceps femoris (BF), semitendinosus (ST), vastus lateralis (VL), and vastus medialis (VM). To assure accurate electrode placements, the areas of highest probability for each motor point location were initially determined utilizing previously established anatomical landmarks (McHugh, Tyler, Greenberg, & Gleim, 2002). The skin surfaces at each placement site were shaved, abraded, and cleansed with alcohol to remove dead surface tissues and oils, and reduce skin impedance. Disposable Ag/AgCl dual electrodes (Noraxon Dual Electrodes, Noraxon USA, Inc. Scottsdale, Arizona) were positioned over the areas of skin preparation parallel with the underlying muscle fibers as determined by a line drawn from the muscle’s origin to its insertion. To reduce movement artifacts and prevent electrodes from losing contact with the skin due to perspiration during the testing, the electrodes were held against the skin using PowerFlex self-adherent wrap (Andover Healthcare Inc., Salisbury, Massachusetts).
Raw EMG signals were recorded during the last ten seconds of each 2-minute stage using a wireless EMG telemetry system (Noraxon USA, Scottsdale, AZ) with an input impedance of 2 MV and a common mode rejection ratio (CMRR) of 100 dB. The gain was set at 2,000, with band pass filtering between of 1 and 500 Hz. Signals were sampled at 1,500Hz, digitized using a 16-bit A/D converter (Noraxon USA, Scottsdale, AZ) and stored using a microcomputer. EMG for each subject was normalized to the peak value seen for maximum voluntary isometric contractions (MVC) of each muscle during seated knee flexion and extension performed on a Biodex System 2 dynamometer (Biodex Medical Systems, Shirley, NY). This was computed by dividing the root mean square (rms) of a given EMG signal by the average rmsEMG value seen for the middle 3s of a 5s MVC. Twelve MVCs were done prior to each trial (3 extensions and 3 flexions for left and right legs) in order to allow comparisons between insole testing conditions. Magnets were placed on the end of the crank arms of the ergometer. During testing, these magnets tripped a reed switch on the bicycle frame, producing a spike recorded on a separate channel that allowed us to divide the EMG signal into segments encompassing rotations of a single-pedal.

EMG ratios were calculated using the following procedures. Using the pedal revolution defined by the reed switch pulses, the 5 “middle” pedal cycles of the 10s data collection period were identified. EMG for each muscle for this period was isolated, the root mean square value was calculated, and the NrmsEMG value calculated as described above. NrmsEMG values were then used to calculate a ratio of medial-to-lateral muscle activity.
for the quadriceps and hamstrings muscle groups. These normalized values were averaged within trials to give one measure to compare across trials (insole conditions).

**Movement Analysis**

Following EMG preparation, retroreflective markers were placed on each rider’s left and right tibial tuberosities. These were held in place with Leukotape (Beiersdorf AG, Hamburg, DE), and further secured with black PowerFlex tape (Andover Healthcare, Salisbury, MA) to create a high-contrast background. A picture of the subject prepared for an exercise trial is presented in (Figure 3). These markers were video recorded while the participants rode each trial, and 10-second clips from the end of each 2 minute resistance stage were cut from the whole video. Within these 10-second clips, the middle 5 pedal cycles beginning with the right leg at the bottom dead center (BDC) of the pedal stroke were identified, and the right and left markers were tracked for these 5 cycles. Raw data were analyzed in the Kinovea open-source software suite to quantify lateral knee movement (maximum lateral position and minimum internal position) as estimated by the movement of markers on the knee. The Kinovea video analysis software was calibrated to a known dimension on the bicycle frame and the maximum lateral distances the right and left markers travelled were measured. These values were averaged within trials to provide a singular measure representative of lateral knee movement for each leg per insole.
Test Conditions

Four insole conditions were evaluated: baseline, low, medium/neutral, and high support. The low, medium, and high arch support levels were created using a widely-sold two-layer cycling shoe-specific insole with a built-in “low” level of arch support (1:1 Insole system, Pearl Izumi, Louisville, CO). “High” and “medium” support levels were attained by use of high-density foam inserts that fit between the two layers at the appropriate position. The baseline insole was each subject’s current insole. Participants passed a “no current arch support” inclusion criteria, essentially making the baseline insole the ‘flat’ arch support condition. Every participant performed their baseline test first, and then was randomized to an insole testing order using the low, medium, and high conditions. “Best Fit” insoles were defined as those non-baseline insoles which caused the least amount of lateral knee movement for the dominant and non-dominant legs.

Five measures were analyzed to ascertain performance changes. Oxygen consumption (VO₂) was measured via gas analysis that took 5-second averages continuously throughout each trial. To negate the effect of any potential spikes in the data, 20-second averages were used. Oxygen consumption at anaerobic threshold (VO₂AT: Defined by 20-second average VO₂ prior to the point VE/VO₂ broke from a linear increase) was used as a way to compare submaximal VO₂ across participants and insole conditions. Heart rates and Velotron resistances (stage, in W) at the same point (anaerobic threshold) were also compared. Additionally, the highest VO₂ value seen during each trial (VO₂Peak) and VO₂AT as a percentage of VO₂Peak were also compared across participants. Because most trials did not go to voluntary exhaustion (participants were stopped
according to cutoff criteria), the common “Time to Exhaustion” measure was not used in this study.

**Statistical Design & Analysis**

Results were analyzed using SPSS Statistics for Windows, Version 20.0 (IBM Corp., Armonk, NY). Descriptive statistics are reported as Means ± SE, and frequencies are provided for categorical data. A 5 x 2 general linear mixed model (GLMM) was used to test if there was a significant difference among insole conditions (baseline, low, medium, high, and “best fit”) and normally used pedals (centering or non-centering float). GLMM was chosen for two primary reasons; firstly, due to the definition of “best fit” the model calculates estimates while accounting for missing data, which the classic ANOVA cannot. Secondly, the GLMM procedure allows the direct specification of the residual covariance matrix, which negates need to satisfy the restricting assumptions of homogeneity of covariance matrices and sphericity. Covariance structures were chosen based on the method that resulted in minimization of Akaike and Bayesian Information Criterions. The GLMMs require that the residuals for each group by insole within the model are normally distributed (McCulloch, 2006). This was assessed using the Shapiro-Wilk test. If violated, the Box-Cox test was performed to find a transformation that would satisfy the assumption. If no transformations uniformly correct for normality of residuals, across groups and insoles, the violation was noted and care taken interpreting the inferential test.
The exercise testing caused participants to sweat. This would occasionally cause EMG electrodes to become unstuck, especially towards the end of each trial, which would result in extreme values to be recorded by the software. To prevent equipment failure such as this from affecting results, data were screened to remove extreme or unrealistic readings using Tukey’s outlier labeling rule (Hoaglin & Iglewicz, 1987) with $k = 2.1$. GLMM was also used to detect the presence of any carry-over effect. The same dependent variables were examined, this time using “Order” and “Pedal System” as the independent variables.

The secondary research question, examining the relationship between arch height and “best fit” insole, was examined using a Spearman Rank-Order Correlation. Spearman’s rho was chosen because arch support level was an ordinate variable based on manufacturer design of the insoles used. Additionally, because we expected arch heights to be similar by “best fit” group, rho was a more appropriate method of identifying a monotonic relationship. The assumptions of linearity, monotonicity, and no outliers were examined using a scatterplot. Alpha was set at .05 for all tests, excluding Shapiro-Wilk, which used .01 due to its sensitivity in small samples.
Chapter 3

RESULTS

Nine participants (32 ± 6.3 years old, 178.6 ± 6.1cm, 83.6 ± 11.9kg; Table 2) completed all 4 VO\textsubscript{2}max tests and were included in the analysis. The inclusion criteria, especially the experience requirements, limited subject recruitment. Most of the competitive cyclists who qualified for the study were in the middle of the training and racing season, which dampened the enthusiasm to participate in the testing this study required. Of the twenty participants who were screened, there were three non-completers (did not finish all four tests), four were not able to participate due to scheduling conflicts, two participants who ceased communication, one elite athlete whose training schedule would not accommodate the volume of testing, and one dropout due to an injury unrelated to the study.

Average left and right lateral knee displacements were computed for each insole condition. “Best Fit”, previously defined as the non-baseline insole that showed the least lateral knee displacement, varied between sides in 6 of 9 cases. Because left and right best fit insoles were not consistent across participants, it was decided to perform analyses by functional “dominant” and “non-dominant” legs. Leg dominance was ascertained by asking which foot each subject would put forward when performing a standing start on a bicycle. Three participants identified their left leg as their dominant leg. One of two covariance structures, either Compound Symmetry or First-Order Autoregressive, was chosen for all GLMMs (Appendix B, Table 3).
Best fit insole had no relationship to standing (loaded) arch height for either the dominant or non-dominant sides, $r_s=.055$ (p=.89) and $r_s=.114$ (p=.77), respectively. Additionally, there was no relationship between sitting (unloaded) arch height and best fit insole on the dominant ($r_s=-.054$, p=.89) or non-dominant ($r_s=-.057$, p=.885) side. Lastly, the difference between sitting and standing arch heights (arch compliance) was not related to best fit insole for the dominant, $r_s=-.103$, p=.791 or non-dominant sides, $r_s=-.601$, p=.087. Scatterplots of the results are shown in Figure 4.

**Order**

A 5 x 2 restricted GLMM was run to examine effect of trial order on dependent variables. The model was restricted to just the main effects of insole and trial order. Two significant effects were identified: Carry-over on Dominant Knee Lateral Movement (Figure 5A), and carry-over on the Non-Dominant Hamstrings Ratio (Figure 5B). Dominant Knee Lateral Movement during the third and fourth trial were significantly higher than the first test ($1.052\text{cm}$, p=.005; $1.173\text{cm}$, p=.002, respectively). The other carry-over effect was for the Non-Dominant Hamstrings Ratio. From the second to third trial, there was a decrease of $0.67$ (p=.003), and from the second to fourth trial, there was a decrease of $0.441$ (p=.04), indicating a relative shift towards lateral hamstring activation during the third trial and fourth trials.

**Knee Displacement**

GLMMs were run to examine knee displacement for the dominant leg and non-dominant legs. For dominant leg, the pattern of differences across insole conditions between pedal
float types was not significant, $F(4,19)=.591$, $p=.673$. There were no differences in pedal float types, $F(1,8)=.022$, $p=.886$. Insoles had a significant effect on lateral knee movement, $F(4,19)=7.837$, $p=.001$ (Figure 6A). Pairwise comparisons revealed that best fit (M=5.228cm, SE=.343cm) insoles caused significantly less lateral knee movement than high (M=6.038cm, SE=.363cm) and medium (M=6.535cm, SE=.387cm) levels of arch support, $p=.015$ and $p=.001$, respectively.

For non-dominant leg, the pattern of differences across insole conditions between pedal float types was not significant, $F(4,17)=.321$, $p=.86$. There were no differences due to pedal float types, $F(1,7)=1.714$, $p=.232$. Insoles had no effect on lateral knee movement, $F(4,17)=1.826$, $p=.169$.

**Quadriceps Activation Ratios**

For the dominant leg, VMO/VL activation ratio differences across insole conditions between pedal float types were not significant, $F(4,18)=.5$, $p=.736$. Pedal float type had no significant effect on VMO/VL ratio, $F(1,7)=.026$, $p=.878$. Insoles caused no changes in the quadriceps activation ratios, $F(4,18)=.649$, $p=.635$.

For the non-dominant leg, VMO/VL activation ratio differences across insole conditions between pedal float types were also not significant, $F(4,17)=.531$, $p=.715$. Pedal float type had no significant effect on VMO/VL ratio, $F(1,8)=.016$, $p=.902$. Insoles caused no changes in the quadriceps activation ratios, $F(4,17)=1.493$, $p=.248$. 
Hamstrings Activation Ratios

The effect of arch support on ST/BF ratios in the dominant leg differed between pedal types, $F(4,18)=4.94, p=.007$ (Figure 7A). With non-centering float pedals, the best fit insole condition ($M=0.738, SE=0.275$) produced a significantly lower ST/BF ratio than the medium condition ($M=1.513, SE=0.275$), $p=.005$, shifting the balance of muscle activation to favor the lateral side in the best fit insole condition. With centering float pedals, the best fit insoles ($M=1.06, SE=0.246$) showed a significantly higher ratio than the medium insole ($M=0.302, SE=0.323$), $p=.021$, implying a shift towards relative medial hamstring activation in the best fit insole. Pedal float type had no significant effect on ST/BF ratio, $F(1,7)=.005, p=.947$; and, ST/BF ratios were not significantly different by insole condition, $F(4,19)=2.206, p=.108$.

The effect of arch support on ST/BF ratios in the non-dominant leg did not differ between centering and non-centering float pedals, $F(3,18)=.023, p=.995$ (Figure 7B). Pedal float type had no significant effect on ST/BF ratio, $F(1,8)=.12, p=.737$. Insoles produced no changes in the hamstring activation ratios, $F(4,17)=1.471, p=.254$.

Performance at Anaerobic Threshold

For heart rate data at anaerobic threshold (HR_{AT}), the assumption of normality was violated. A Box-Cox test was run to determine a transformation to remedy this. No reasonable transformation resulted in the assumption of normality being met. Accordingly, the raw data for heart rate at anaerobic threshold were analyzed, and should be interpreted in that context.
For the dominant leg, the effect of arch support on HR\textsubscript{AT} did not differ between pedal float types, \(F(4,18)=1.796, p=.173\). Pedal float type had no significant effect on HR\textsubscript{AT}, \(F(1,7)=.000, p=.987\). HR\textsubscript{AT} was significantly different by insole, \(F(4,18)=4.172, p=.014\). Pairwise comparisons showed that the effect was caused by differences between the baseline, low, medium, and high insole groups, and that the best fit insole (M=152.8, SE=4.4 bpm) was not significantly different from any other insole conditions.

For the non-dominant leg, the effect of arch support on HR\textsubscript{AT} did not differ between pedal float types, \(F(4,18)=2.776, p=.059\). Pedal float type had no significant effect on HR\textsubscript{AT}, \(F(1,7)=.013, p=.911\). HR\textsubscript{AT} was significantly different by insole, \(F(4,18)=4.018, p=.017\) (Figure 9C). Pairwise comparisons showed the best fit insole (M=153 bpm, SE=4.633) displayed a lower heart rate than the baseline insole condition (M=157.4 bpm, SE=4.6), \(p=.044\).

Other measures of submaximal efficiency (cycling economy at anaerobic threshold) showed no significant main effects or interactions. There were no significant effects on oxygen consumption at anaerobic threshold (VO\textsubscript{2AT}). Power output at anaerobic threshold was not changed by insole condition, pedal float type, or their interaction for dominant or non-dominant best fit insoles. Peak oxygen consumption during each trial (VO\textsubscript{2Peak}) and VO\textsubscript{2AT} as a percentage of VO\textsubscript{2Peak} also remained unchanged.
Chapter 4

DISCUSSION

The results of this study indicate that insoles can affect lateral knee movement, supporting their use by clinicians and bicycle fitters. As we hypothesized, the “best fit” insole resulted in a lower level of lateral knee movement compared to the other insole conditions. However, contrary to what was hypothesized, “best fit” was not related to an individual’s arch height, and in two-thirds of participants, differed by side. Our results indicate that despite having similar arch height and compliance levels for dominant and non-dominant legs, different levels of arch support demonstrate the “best” mechanical results for each leg. These findings indicate that when fitting a rider with orthotic insoles, care should be taken not to generalize one level of arch support as “best” or most effective application for both legs. Nigg, Nurse, and Stefanyshyn (1999) assert that it “should be possible to match subject characteristics with insert and orthotic characteristics” indicating their assertion that improved alignment, strength, and compliance can be used, among others, as possible diagnostic markers of a subject’s “optimal” insole. However, they performed no physical analyses to confirm this assertion; and ultimately proposed that comfort may be the key variable related to fit, which could explain why “best fit” insoles were often different between legs.

In addition to comfort, imbalances in strength, or leg dominance, could explain the dissimilar optimal insole heights between legs. Perry and Bekey (1981) showed the relationship between EMG and force could differ due joint positioning as seen in the current study; while Öunpuu and Winter (1989) noted that symmetry should not be
assumed in gait. These statement, combined with findings in controlled studies literature establishing differences between dominant and non-dominant legs (Chhibber & Singh, 1970; Jacobs, Uhl, Seeley, Sterling, & Goodrich, 2005), support our conclusion that anthropometric similarities between dominant and non-dominant legs, specifically represented in the current study as differences in longitudinal arch characteristics, are not a strong enough basis by which to assign overall “best fit” insoles.

We found no significant changes in mediolateral pull ratios of the quadriceps muscles across any insole condition. In theory, alteration of ankle position by insoles while pedaling would affect tibial version, and the position of the tibial tuberosity with respect to the quadriceps muscles. This change in position would then be manifested as changes in EMG activity of the quadriceps muscles. Research supports the assertion that orthotic insoles affect tibial version (Gross & Foxworth, 2013); however, no changes in EMG activity were seen in this study. This may be because the position of the tibial tuberosity (the insertion for the patellar tendon) has a less-than-expected effect on the action of the quadriceps muscles because it is not the direct insertion of the muscles evaluated, which first act on the patella (Lee, Morris, & Csintalan, 2003). Alternatively, the magnitude of alterations in tibial version due to arch support may not have been enough to elicit changes in EMG values of the quadriceps muscles.

Hamstring activation ratios remained unchanged for the non-dominant leg across insole conditions. When the dominant leg’s best fit insole was used, hamstring EMG ratios changed across insole conditions, between pedal float types. In both significant pairwise
comparisons, the medium insole was seen to alter the ST/BF ratios compared to the best fit insole. Changes in hamstrings EMG may be affected to a greater extent than the quadriceps because alterations in ankle position by arch support affects tibial torsion (Lynn & Costigan, 2009; Gross & Foxworth, 2013), and therefore, generate a higher tension level through the direct insertion points of the hamstrings than the quadriceps muscles. The change across pedal systems suggests an external torque in the transverse plane of the foot which affects the action of muscles up the leg. The pull to midline seen in centering-float style pedals may be responsible. If future research corroborates this assertion, this force is something clinicians should keep in mind when applying orthotic insoles to correct pedaling mechanics.

Although this study found few differences in EMG ratios, it does contribute to the literature by helping establish “normal” EMG ratios during cycling. Similar to Cerny (1995) and Lynn (2009), this information will assist future studies by allowing researchers and clinicians to identify abnormal ratios.

We measured few changes in physiological variables in this study. In the case of heart rate at anaerobic threshold, the decrease in beats per minute would be detrimental to performance. A training effect would result in lower heart rates and higher stroke volumes at a given work rate, and while this may cause a decrease in heart rate for a given work effort, the trials in this study were not spaced far enough apart that a training effect would be a plausible explanation for this effect. Under the circumstances, the
long-established effects of hydration status on heart rate response (Saltin, 1964a; Saltin, 1964b; Candas, Libert, Brandenberger, Sagot, & Kahn, 1988), or existing fatigue prior to starting a trial, are likely responsible for the changes seen.

Koch, Frölich, Emrich, and Urhausen (2013) demonstrated that power output slightly decreased during a Wingate test when subjects used a stiff carbon fiber insole. While the current study used soft insoles, the results of the two studies were similar. Power output at anaerobic threshold did not change by insole condition. Measures of sub-maximal oxygen consumption were also unchanged. The results of this study further establish that insoles do not appear to provide a physiological improvement in cycling performance.

**Limitations**

This study used a carefully selected, mostly homogeneous sample. It should be noted that a homogeneous sample limits the generalizability of this study. Excluding female riders, while removing gender as a confounding variable, also means that these results do not take into account large anatomical differences between riders of different genders (increased Q-angle; Horton & Hall, 1989: altered patellofemoral joint biomechanics; Csintalan, Schulz, Woo, McMahon, & Lee, 2002). In addition to being homogeneous, the sample size (n=9) was also smaller than optimal for this study.

Also, this study recruited riders who used both centering-float and non-centering-float pedal systems. This is important to note because “non-centering” float allows the cleat (and shoe and foot) to rotate freely above the pedal body. Centering-float-style pedal
systems allow rotation before the cleat disengages from the pedal, but are built in such a way that the spring tension of the pedal is constantly trying to pull the cleat (and thus, shoe) back into line with the pedal body. Because pedals showed effect on the dominant leg hamstring ratio, we cannot completely rule out a float-type effect on leg muscle activation patterns.

Lastly, the results of this study represent acute changes due to insole use. Previous research examining changes in kinematics and kinetics with long-term insole use has concentrated on treatment of osteoarthritis in the knees, but suggests that insoles may cause chronic changes. The current study design used one trial per insole, with no acclimatization period, and thus would not have shown such changes. Though the literature concentrates on walking gait, Alsancak (2012) showed that long-term insole use significantly reduced knee varus moment. Toda and Tsukimura (2004) also demonstrated subjective improvement in gait function via visual analogue scale (VAS) and Lequesne Index scores in women with osteoarthritis. Schwellnus, Jordaan, and Noakes (1990) showed that insole use attenuated injury rates over a 9-week training period. Stöggl, Haudum, Birklbauer, Murrer, and Müller (2010) showed that 10 weeks of training with an experimental shoe resulted in gait characteristics similar to walking in control shoes. These results support the assertion that insoles in cycling shoes may show more effects on mechanical, electromyographical, and performance variables after prolonged use and adaptation. Future studies on insoles in cycling shoes should be longitudinal in nature, to allow the evaluation of any effects insoles might have after adaptation phases or prolonged use.
Chapter 5

CONCLUSIONS

The effects orthotic insoles have on skeletal alignment are still debatable (Nigg et al, 1999; (Gross & Foxworth, 2013). Orthotic support in running shoes is an area of active research (McMillan & Payne, 2008); however, arch support in cycling shoes has little support in the literature, though bicycle fitters often use them as a corrective intervention. Unfortunately, we cannot support the use of orthotic insoles in cycling shoes as a mechanical intervention. While the identified “best fit” insole did produce significantly lower lateral knee movement than the other experimental insole conditions, it was not significantly lower than the baseline insoles the subjects already used. Although the “best fit” insole was not mechanically better than baseline, it was also no worse. It may be that using the rider-identified more comfortable of the two may result in safer and/or more mechanically efficient pedaling motion (Callaghan, 2005).

Since arch height was not related to the “Best Fit” insole in this study, an important goal for future research is to identify alternative predictors of individuals’ “Best Fit” insoles. As outlined above, Nigg et al. (1999) offered some guidance to identify characteristics that would make predicting a subject’s ideal insole a simpler task; however, the findings of the current study indicate that neither arch height, nor arch compliance are suitable predictors.

Though arch supports are a widely-known intervention, there is little research explaining their effects. Future research should use larger and more varied samples to further
examine the effects of float type, gender differences, or forefoot correction (which can be
applied at the insole or the cleat) on motion of the leg. Study designs should also explore
the long term effects of training with foot position correction. Longitudinal studies, or
trial designs using longer, steady-state rides or time trials, may identify power output or
oxygen consumption effects that were not seen in this study. Additionally, since “best
fit” insoles were often different from one side to the other, a similar study should be
carried out with each leg’s “best fit” insole used on that respective leg, to examine if the
combination of the “best” for each leg produces greater impacts of electromyographic
and cardiorespiratory variables than those recorded in this study. In conclusion, arch
support insoles in cycling shoes may be an effective intervention for altering pedaling
kinematics, but not necessarily for increasing performance. Continued study of the
distinct effects of alterations in the FSCP interface can eventually be synthesized to build
a complete model to explain the behavior of the pedaling leg.
REFERENCES


APPENDIX A

FIGURES

Figure 1: Shimano FSCP interface, patent #US 6925908 B2. The image shows the mechanics of shoe attachment to the pedal system, via cleats attached to the bottom of the shoe.
Figure 2: Consort Flow Chart

**Enrollment**

Assessed for eligibility (n = 20)

Exclusion (n = 7)
- Do not meet inclusion criteria
- Decline to participate
- Other reasons

Randomized (n = 13)

**Allocation**

Allocated to insole order cohort (n = 13)
- Cohort orders (n = 6; LNH, LHN, NLH, NHL, HNL, HLN)
- Participants per cohort (n = 2)

**Follow-Up**

Lost to follow-up (n = 4)

**Analysis**

Analyzed: (n = 9)
Centering: (n = 5); Non-centering (n = 4)
Excluded data points (equipment failure) (n = 9)
Figure 3: Subject prepared for exercise trial
A)

- Standing Arch Height
- Sitting Arch Height
- Arch Compliance (Right Y Axis)
- Linear (Standing Arch Height)
- Linear (Sitting Arch Height)
- Linear (Arch Compliance (Right Y Axis))

Dominant Foot Best Fit Insole
Figure 4: Scatterplots of arch variables v “Best Fit” insoles for A) Dominant and B) Non-Dominant (d) legs.
Figure 5: The effects of carry-over on A) dominant knee lateral movement and B) non-dominant hamstrings ratio. * Significantly different from 1st Trial, p<.05; **significantly different from 3rd and 4th trials, p<.05.
Figure 6: The effects of insole condition on lateral knee movement in the A) dominant and B) non-dominant legs. * Significantly higher than “Best Fit”, p<.05.
Figure 7: The interactive effects of insole type and non-centering □ and centering □ pedals on A) dominant and B) non-dominant hamstrings ratios. * Significantly higher than “Best Fit” insole, p<.05; † Significantly lower than “Best Fit” insole, p<.05. The “Low” insole condition was classified as the non-dominant “Best Fit” for all riders using centering float pedals, causing a missing column in 7B.
Figure 8: The effects of insole on heart rate at anaerobic threshold (HR_AT) for A) dominant and B) non-dominant legs. †Significantly higher than "Low" and "Medium" insoles, p<.05. * Significantly higher than "Best Fit", "Low", and "Medium" insoles, p<.05.
### Table 1: Subject Demographics

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Table 2: Arch Measurements
A) Insole Analysis

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<th>HR</th>
<th>HR_{AT}</th>
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B) Order Analysis

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Table 3: Covariance Structures for GLMM models. “CS” = Compound Symmetry; “AR1” = 1st Order Autoregressive, “KM” = Knee Movement, “QR” = Quadriceps Ratio, “HR” = Hamstrings Ratio, “HR_{AT}” = Heart Rate at Anaerobic Threshold, “VO_{2AT}” = VO$_2$ at Anaerobic Threshold, “VO_{2P}” = Peak VO$_2$, “AT_{P}” = VO$_2$ at Anaerobic Threshold as a percentage of Peak VO$_2$, “ATW” = Watts (resistance) at Anaerobic Threshold.