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Effects of Lateral Shoe Wedges and Toe-in Foot Progression Angles on the Biomechanics of Knee Osteoarthritis during Stationary Cycling

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To the Graduate Council:

I am submitting herewith a dissertation written by Jacob Kale Gardner entitled "Effects of Lateral Shoe Wedges and Toe-in Foot Progression Angles on the Biomechanics of Knee Osteoarthritis during Stationary Cycling." I have examined the final electronic copy of this dissertation for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Doctor of Philosophy, with a major in Kinesiology and Sport Studies.

Songning Zhang, Major Professor

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Effects of Lateral Shoe Wedges and Toe-in Foot Progression Angles on the Biomechanics of Knee Osteoarthritis during Stationary Cycling

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DEDICATION

This dissertation is dedicated to my beloved wife, Keisha Gardner, for her unwavering love and support during our 10 years of marriage.

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ABSTRACT

Exercise is important for individuals with knee osteoarthritis (OA) but certain activities can be painful and discourage participation. Cycling is commonly prescribed for OA but practically no previous literature exists. Due to their altered knee kinematics, OA patients may be at greater risk of OA progression or other knee injuries during cycling. The purpose of Study One was to investigate the effects of lateral wedges on knee joint biomechanics and pain in patients with medial compartment knee OA. The purpose of Study Two was to investigate the effects of toe-in foot progression angles on the same variables. Thirteen OA subjects and 11 healthy subjects participated. A motion analysis system and custom instrumented pedal was used to collect 5 pedal cycles of kinematics and kinetics during 2 minutes of cycling in one neutral and two lateral wedge conditions (5° and 10°) for Study One and 2 toe-in conditions (5° and 10°) for Study Two. Subjects pedaled at 60 RPM and 80 watts and rated their knee pain on a visual analog scale.

Study One: There was a 22% decrease in the knee abduction moment with the 10° wedge. This finding was not accompanied by a decrease in knee adduction angle or pain. Additionally, there was an increase in vertical and horizontal PRF which may negate the advantages of the decreased KAM.

Study Two: For the OA subjects, there was a 61% (2.7°) and a 73% (3.2°) decrease in peak knee adduction angle compared to neutral. This finding was not accompanied by a decrease in pain or KAM because of high inter-subject variability. A simple linear regression showed a positive correlation between Kelgren-Lawrence (K/L) score and both peak knee adduction angle and KAM.

For OA patients, cycling with a 10° lateral wedge or a decreased foot progression angle may be beneficial in slowing the progression of OA or minimizing other knee injuries. Patients with a higher K/L score may have greater risk of injury. More research is needed to investigate the joint contact forces as well as long term effects of riding with wedges or toe-in foot angles.

TABLE OF CONTENTS

Page

LIST OF TABLES

LIST OF FIGURES

DEFINITION OF TERMS

1. *Crank cycle*: one revolution of the bicycle crank arm beginning and ending at top dead center.
2. *Top dead center (TDC)*: when the bicycle pedal is in the highest position.
3. *Bottom dead center (BDC)*: when the bicycle pedal is in the lowest position.
4. *Power phase*: phase of the crank cycle between TDC and BDC when the pedal is being pushed to propel the bicycle forward.
5. *Recovery phase*: phase of the crank cycle between BDC and TDC when the pedal is not being pushed to propel the bicycle forward.
6. *Peak medial PRF*: peak medial component of the resultant pedal reaction force during the power phase of the crank cycle
7. *Peak posterior PRF*: peak posterior component of the resultant pedal reaction force during the power phase of the crank cycle
8. *Peak vertical PRF*: peak vertical component of the resultant pedal reaction force during the power phase of the crank cycle
9. *Ankle inversion*: frontal plane angular deviation of the foot toward the midline of the body with respect to the tibia.
10. *Ankle eversion*: frontal plane angular deviation of the foot away from the midline of the body with respect to the tibia.
11. *Foot progression angle*: angular deviation of the long axis of the foot with respect to horizontal in the transverse plane.
12. *Peak ankle plantarflexion angle*: sagittal plane peak angular deviation of the foot away from the midline of the body with respect to the tibia.

13. *Peak ankle eversion angle*: frontal plane peak angular deviation of the foot away from the midline of the body with respect to the tibia.
14. *Peak ankle internal rotation angle*: transverse plane peak angular deviation of the foot toward the midline of the body with respect to the tibia.
15. *Peak knee flexion angle*: sagittal plane peak angular deviation of the tibia with respect to the femur.
16. *Peak knee adduction angle*: frontal plane peak angular deviation of the tibia towards the midline of the body relative to the femur around 90° of the power phase of the crank cycle.
17. *Peak knee external rotation angle*: transverse plane peak angular deviation of the tibia away from the midline of the body with respect to the femur.
18. *Peak internal knee extensor moment*: sagittal plane peak moment produced concentrically by the knee extensor muscles and ligaments to push the pedals around 90° of the power phase of the crank cycle.
19. *Peak internal knee abduction moment (KAM)*: frontal plane peak moment produced by the knee abductor muscles and lateral ligaments around 90° of the power phase of the crank cycle.
20. *Peak knee internal rotation moment*: transverse plane peak moment produced by the knee internal rotator muscles and ligaments around 90° of the power phase of the crank cycle.
21. *Abduction*: the frontal plane angular deviation of the tibia away from the midline of the body relative to the femur (same as knee valgus)

22. *Adduction*: the frontal plane angular deviation of the tibia toward the midline of the body relative to the femur (same as knee varus)

23. *Knee varus*: the frontal plane angular deviation of the tibia toward the midline of the body relative to the femur (same as knee adduction)

24. *Knee valgus*: the frontal plane angular deviation of the tibia away from the midline of the body relative to the femur (same as knee abduction)

CHAPTER I

INTRODUCTION

BACKGROUND

Osteoarthritis (OA) can have an incapacitating effect on people affected. The disease is prevalent in nearly 27 million people in USA alone (Lawrence et al., 2008) and joints that are commonly affected are the weight bearing joints of the lower extremities, namely the knees and hips (Lawrence et al., 2008). Exercise such as cycling is commonly prescribed by health professionals to reduce the loads placed on the joints and are effective for exercise in populations with knee injuries. However, it is not known if people with knee OA have different cycling patterns than healthy populations. Or perhaps, those with unilateral knee OA may experience asymmetrical patterns within their own limbs. If in fact persons with knee OA cycle differently, abnormal kinematics and kinetics may lead to further progression of the disease or increased pain experienced by the rider. If abnormal cycling kinematics and kinetics are present, it is possible that corrective conservative measures can be taken to encourage normal riding patterns and promote exercise in knee OA populations.

It is clear that cycling reduces loading on the knee joint by placing the majority of the rider's body weight on their seat during seated cycling (Burke, 2003). However, cycling produces a great demand on the muscles of the lower limbs, especially the knee extensors, as they are the driving force in propelling the bicycle forward. The increased muscle contraction in turn produces increased loading to the knee joint. Thus, knee injuries are still the leading complaint in cycling which has strong indication for an overuse injury mechanism (Dettori and

Norvell, 2006; Kennedy et al., 2007). For example, a common overuse injury during cycling, patellofemoral pain syndrome, is thought to occur because of an internal knee abduction moment during the downward pedal stroke (Boyd et al., 1997; Wolchok et al., 1998). Thus, proper alignment of the lower limbs that aid in reducing the internal knee abduction moment during cycling is an important factor for reducing overuse injuries experienced by the rider (Bailey et al., 2003; Gregersen et al., 2006a; Ruby and Hull, 1993). Additionally, compressive joint loads in cycling (about 1 to 2 body weights) have been estimated to be similar or slightly less than normal walking (about 2 to 2.5 body weights) (D'Lima et al., 2008; Ericson and Nisell, 1986a). However, due to the potentially large loads placed on the knee as a result of muscular contractions and the fact that there is a lack of information on knee osteoarthritis joint loads during cycling, it should be necessary to estimate the forces on the knee. This is important because a person with improper knee joint alignment during cycling has a greater potential for excessive knee joint loading which could have a negative impact on the knee over time.

During cycling of healthy population, frontal plane knee angles range from about 2 to 4 degrees of abduction to 1 to 6 degrees of adduction during the crank cycle (Bailey et al., 2003; Umberger and Martin, 2001). This small range of motion in the frontal plane indicates that there is not a large amount of abduction/adduction and the knee remains fairly neutral. A current ongoing study in our laboratory showed that participants with medial knee OA do not cycle with the normal frontal plane knee kinematics seen in healthy subjects in previous studies. Out of the 6 initial participants, 6 knees are continuously adducted throughout the crank cycle, 2 knees are continuously abducted, and 4 knees appear to have a normal range of motion. . With regard to the continuously adducted knees, the pattern seen is similar to that during gait in which patients with medial compartment OA walk with the knee in an adducted position (Cerejo et al., 2002). Bailey et al. (2003) found in their study that riders with a history of overuse knee pain had increased knee adduction/abduction angles when compared to the healthy controls. As discussed earlier, malalignment of the knee during cycling is a concern because it may exacerbate an existing condition such as knee OA or cause other problems such as overuse injuries with long term riding. Two possible solutions to knee malalignment during cycling could be borrowed from solutions used during gait, such as using lateral shoe wedges or a toe-in foot progression angle.

The internal knee abduction moment (KAM) during gait has been shown to be an important factor associated with knee OA (Baliunas et al., 2002; Cerejo et al., 2002). The KAM is a surrogate measure for loading to the medial compartment of the knee which is created as a response to an external adduction moment resulting from the ground reaction force (Schipplein and Andriacchi, 1991). The external adduction moment is defined as the product of the length of the moment arm from the knee joint center and the ground reaction force (GRF) vector in the frontal plane of the knee during walking (Hunt et al., 2006). This moment acts to adduct the knee during stance into a bow-legged or knee varus position (Cerejo et al., 2002); a condition that opens the lateral joint space while closing the medial joint space of the knee, resulting in increased stress on the medial compartment. Several studies have found a relationship between the magnitude of the adduction moment and the severity of knee OA (Cerejo et al., 2002; Mundermann et al., 2005; Sharma et al., 1998; Wada et al., 2001). Mundermann et al. (2005) found that people with more severe knee OA have a larger varus alignment (5.7°) than those with a less severe disease (0.3°). Additionally, malalignment in the knee has also been shown to be associated with the progression of knee OA (Cerejo et al., 2002).

Two possible mechanisms for reducing the KAM during walking, which have been verified by previous studies, is by placing a laterally posted orthotic in patients’ shoes (Butler et al., 2009; Butler et al., 2007; Hinman et al., 2009; Kerrigan et al., 2002) or by using variable stiffness walking shoes (Erhart et al., 2008, 2010b) which are shoes that have different stiffness on the medial side compared to the lateral side. The majority of people with knee OA have medial compartment knee OA. Thus a laterally posted orthotic would be used to place the ankle into a more everted position which pulls the knee more medial; effectively opening up the medial compartment. This causes a shift in the orientation of the ground reaction force vector so that the vector lies closer to the knee joint center, and thus decreases the GRF moment arm. Butler et al. (2009) showed that an average of 10 degree lateral wedge significantly reduced the peak internal knee abduction moment by 10% compared to a no wedge control condition. Erhart et al. (2010a) showed that by wearing variable stiffness walking shoes, participants were able to reduce the knee abduction moment by 6.6% and reduced pain compared to the subjects’ personal shoes. It is logical to assume that this method for reducing the KAM may be transferred to cycling. However, it is unknown if these modifications in cycling would produce similar results.

Other possible methods for reducing the KAM during walking are through simple gait modification strategies as demonstrated by several researchers (Fregly et al., 2007; Guo et al., 2007; Mundermann et al., 2008; Shull et al., 2012). Guo et al. (2007) attempted to reduce the KAM by requiring their participants to walk in an increased toe-out (foot progression) angle during walking and ascent/descent tasks. The results of the study showed that the participants were able to reduce their second peak KAM during the walking (40%) and stair ascent (11%) in a 15 degree increased foot progression angle compared to their self-selected foot progression angle. However, no beneficial changes were noted in the first peak KAM which is a measure that is more closely related to the severity and progression of medial knee OA. Thus, a toe-out method of gait change may not provide the desired load reductions in this population. Shull et al. (2012) attempted to reduce the KAM by having their participants walk in a toe-in foot progression angle (0.75 degree shank angle increase from baseline). They found that this method of walking reduced the first peak knee adduction moment by about 11% but the second peak KAM remained unchanged. This study provides promising results for a simple method to effectively reduce the KAM during walking, and may be a potential solution for realignining malaligned lower limb joints during cycling in the medial knee OA population.

STATEMENT OF THE PROBLEM

To our knowledge, no studies have explored the effects of limb alignment alterations on the internal knee abduction moment of knee OA patients during cycling. Changes in lower extremity alignment using lateral wedges and a toe-in foot progression angle could alter the frontal plane kinematics by placing the knee in a more medial position. This alignment change would decrease the length of the moment arm of the pedal reaction force to the knee joint center, thus, decreasing the KAM.

Therefore, the purpose of study one was to investigate the effects of lateral shoe wedges on peak knee adduction angle, peak internal knee abduction moment, and knee pain in healthy subjects, and subjects with medial compartment knee OA during moderate intensity stationary cycling.

The purpose of study two was to investigate the effects of a toe-in foot progression angle on peak knee adduction angle, peak internal knee abduction moment, and knee pain in healthy subjects and subjects with medial compartment knee OA during moderate intensity stationary cycling.

RESEARCH HYPOTHESES

Study One

1. It was hypothesized that lateral shoe wedges would reduce the peak adduction angle and the peak internal knee abduction moment in both healthy subjects and medial compartment knee OA patients during stationary cycling compared to a neutral control condition with no wedge.
2. In addition, for the OA subjects, due to the hypothesized decrease in knee adduction angles and adduction moments, it was expected that there would also be a decrease in knee pain with lateral wedges compared to a neutral foot position.

Study Two

1. It was hypothesized that toe-in foot progression angles would reduce the peak knee adduction angle and the peak internal knee abduction moment in both healthy subjects and medial compartment knee OA patients during stationary cycling compared to a neutral control condition.
2. In addition, for the OA subjects, due to the hypothesized decrease in knee adduction angles and adduction moments, it was expected that there would also be a decrease in knee pain with increased toe-in foot progression angles compared to a neutral foot position.

DELIMITATIONS

The exclusion criteria for the OA subjects included:

- Other osteoarthritis symptoms at the ankle or hip joint.
- Any lower extremity joint replacement.
- Any lower extremity joint arthroscopic surgery or intra-articular injection within past 3 months.
- Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis).
- BMI greater than 35.
- Inability to ride a stationary cycle ergometer without assistance.
- Neurologic disease (e.g. Parkinson's disease, stroke patients).
- Lower back pain referred to the lower limbs.
- Unable to see, hear, or follow instructions.
- Women who are pregnant or nursing.
- Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise.

The inclusion criteria for the OA subjects included:

- Men and women between the ages of 35 and 65.
- Radiographically diagnosed with medial compartment knee osteoarthritis in one or both knees, with or without patella-femoral knee osteoarthritis, by a rheumatologist with a grade 1 to 4 on the Kellgren-Lawrence scale.

The exclusion criteria for the healthy subjects included:

- Knee pain for at least 6 months during daily activities.
- Diagnosed with any type of lower extremity joint osteoarthritis.
- Any lower extremity joint replacement
- Any lower extremity joint arthroscopic surgery or intra-articular injection within past 3 months.
- Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis).
- BMI greater than 35.
- Inability to ride a stationary cycle ergometer without assistance.
- Neurologic disease (e.g. Parkinson's disease, stroke patients).
- Lower back pain referred to the lower limbs.
- Unable to see, hear, or follow instructions.
- Women who are pregnant or nursing.
- Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise.

The inclusion criteria for the healthy subjects included:

- Men and women between the ages of 35 and 65.

LIMITATIONS

1. All tests were conducted in a laboratory setting.
2. Skin marker placement in obese participants may not reflect accurate bony landmark locations.

3. Reflective markers used to track the feet during motion trials were placed on the shoe. Thus, foot motions within the shoe may not have been accurately captured.

4. No X-rays were performed on the healthy subjects. Thus, it was assumed that the healthy subjects did not have OA.

CHAPTER II

LITERATURE REVIEW

The purpose of study one was to investigate the effects of shoe wedges on knee biomechanics in healthy subjects and subjects with with medial compartment knee OA during moderate intensity stationary cycling. The purpose of study two was to investigate the effects of toe-in foot progression angles on knee biomechanics and pain in healthy subjects and subjects with medial compartment knee OA during moderate intensity stationary cycling. Additionally, each study will investigate the effects of the interventions on knee pain. Finally, each study will explore individual knee muscle forces and knee contact forces in patients with medial compartment knee OA during moderate intensity stationary cycling using a musculoskeletal modeling approach.

The focus of this chapter is to review existing literature on the biomechanics of cycling and knee osteoarthritis gait variables that may lead to the development and progression of OA. This work will attempt to make links between knee OA and cycling specifically for the management of knee osteoarthritis related symptoms.

BACKGROUND

Osteoarthritis (OA) can have an incapacitating effect on those afflicted by it. The disease is prevalent in nearly 27 million people in America alone (Lawrence et al., 2008) and joints that are commonly affected are the weight bearing joints of the lower extremities, namely the knees and hips (Lawrence et al., 2008). The cause of osteoarthritis is not completely understood but there are various known risk factors including age (Felson, 1990; Felson et al., 2000; Felson and

Zhang, 1998), female gender (Nevitt and Felson, 1996; Srikanth et al., 2005; Theis et al., 2007), muscle weakness (Baker et al., 2004; Slemenda et al., 1997; Slemenda et al., 1998), genetics (Evangelou et al., 2011; Felson et al., 1998), injury (Lohmander et al., 2004), overuse (Coggon et al., 2000; Felson et al., 1991), and obesity (Felson et al., 2000).

Obesity is the single most modifiable risk factor in the development and progression of OA and weight loss has been shown to reduce the debilitating symptoms that OA patients commonly encounter (Focht et al., 2005; Messier et al., 2005b; Messier et al., 2004). A study conducted by Messier et al. (2005b) demonstrated that for every 1 pound in body mass a person loses, the compressive load across the knee joints is reduced by 4 pounds. Other researchers have noted that weight loss by means of diet and exercise resulted in improvements in function and pain (Focht et al., 2005; Messier et al., 2004).

Additional support by the Osteoarthritis Research Society International (OARSI) has made 25 recommendations for the treatment of patients with OA of the hip or knee (Zhang et al., 2008). These recommendations are evidence based and expert advised treatments which have been shown to improve the symptoms of OA. Two of the highly recommended non-pharmacological treatments are regular aerobic and muscle strengthening exercise, and weight loss with maintenance of a healthy weight for overweight individuals (Zhang et al., 2008).

Based on existing evidence, it is important for people to maintain a healthy weight and if obese or overweight, it is crucial to their health that they lose weight. Additionally, the American College of Sports Medicine position stance recommends that most adults engage in moderate or vigorous intensity cardiorespiratory exercise (Garber et al., 2011). Current recommendations at the time of this writing are for greater than or equal to 30 min per day for greater than or equal to

5 days per week for moderate intensity, or greater than or equal to 20 min per day for greater than or equal to 3 days per week for vigorous intensity exercise. Alternatively, the exercise can be some combination of the two producing a total energy expenditure of greater than or equal to 500-1000 MET min per week (Garber et al., 2011).

Diet and exercise continues to prove to be the healthiest and most effective form of weight loss. Unfortunately, weight loss can be a complex process that has difficulties of its own. One of those problems, specifically for overweight and osteoarthritic populations, is that excess load from the increased body mass increases the load on the knee joints (Messier et al., 2005b). This makes it very difficult, even painful for a person with arthritis to participate in load bearing exercises such as walking, jogging, or running, which are the most common and widely prescribed forms of exercise.

Other forms of exercise such as cycling are commonly prescribed by health professionals to reduce the loads placed on the joints and are effective for exercise in populations with knee injuries. Bicycling allows a person to get a good workout during exercise without inducing large impact loads to the lower extremity joints, as most of the body weight is carried by the seat, essentially relieving the lower extremities of bearing the load. Very few published studies exist for studying bicycling in osteoarthritic populations. However, it can be assumed that the non-weight bearing nature of cycling will reduce the knee joint loads of osteoarthritis sufferers.

What has not been answered in previous literature is the question of whether people with knee OA have different cycling patterns than healthy populations. Or perhaps, those with unilateral knee OA may experience asymmetrical patterns within their own limbs. If in fact persons with knee OA cycle differently, abnormal kinematics and kinetics may lead to further progression of the disease or increased pain experienced by the rider. This may discourage individuals from getting adequate exercise. If abnormal cycling kinematics and kinetics are present, it is possible that corrective conservative measures can be taken to encourage normal riding patterns and promote exercise in knee OA populations.

GAIT CHARACTERISTICS OF KNEE OSTEOARTHRITIS

It is important to understand the gait characteristics of knee OA because it is possible that the changes that patients experience with osteoarthritic gait may translate into cycling. Additionally, virtually no studies at the time of this writing have reported biomechanical variables of people with knee OA during cycling. Generally, individuals with knee OA have slower walking speed, less knee flexion angle at heel strike, less total knee range of motion (ROM), and increased knee internal abduction moment (KAM) compared to healthy subjects (Baliunas et al., 2002; Gok et al., 2002; Hurwitz et al., 2002; Kaufman et al., 2001; Mundermann et al., 2005).

Kinematics

During normal healthy walking, the sagittal plane knee angle typically ranges between 8 and 64 degrees of flexion throughout the gait cycle (Ounpuu, 1994). When only the stance phase is considered, the knee joint angle peaks at approximately 20 – 25 degrees around mid-stance (Hamill and Knutzen, 2009). In the frontal plane, typical knee joint angles in normal healthy walking range between approximately 2 degrees of abduction and 5 degrees of adduction (Boyer et al., 2012; Salsich and Long-Rossi, 2010).

The sagittal plane knee joint angle in people with osteoarthritis has been shown to be much less than healthy subjects. Zeni and Higginson (2009) showed that at self-selected and fast walking speeds, people with osteoarthritis experienced less knee flexion than their healthy counterparts. In fact, those with severe knee OA (KL score = 4) demonstrated about 10-12 degrees of knee joint excursion while those with moderate OA (KL score = 2 – 3) experienced excursions of approximately 13 to 17 degrees. The normal healthy controls experienced excursions of about 18 to 22 degrees depending on the walking speed. These results clearly indicate different characteristics of disease severity.

Astephen et al. (2008) performed a gait analysis comparing two knee OA groups (moderate and severe) with a control group. The subjects that were designated for joint replacement were placed in the severe group while those who were not were placed in the moderate group. These groups were verified with the Kellgren Lawrence Scale (KL). The patients in the severe group showed KL grades of 3-4 while those in the moderate group had KL grades of 1-4 (median of 2). The results of the study showed a relationship between OA severity and the peak knee flexion angle during stance. More specifically, the severe OA group saw the least amount of flexion (8 degrees), followed by the moderate OA group (14 degrees), and lastly by the asymptomatic group (~19 degrees). Additionally, the peak flexion angle during swing was decreased in the severe OA group compared to the other two groups. The angle for the severe group peaked at about 46 degrees while the moderate OA and asymptomatic groups experienced angles of about 6 degrees1 and 64 degrees respectively. Consequently, the severe OA group experienced a 16 - 18 degree reduction in knee angle range of motion compared to the other two groups.

While studying inter-limb differences in people with moderate unilateral knee OA, Briem and Snyder-Mackler (2009) found that when comparing the involved limb with the uninvolved limb the peak flexion angle of the involved knee was about 4.4 degrees less during the stance phase of walking. Interestingly, they also found a relationship between the sagittal plane knee flexion angle and the frontal plane varus angle. Specifically, those individuals with more knee varus were more likely to have less knee flexion. This relationship was evident in both involved and uninvolved knees but was only significant in the involved side.

It is clear from the presented studies that knee OA populations have less sagittal plane knee flexion during walking (i.e. they walk stiffer). It is not completely clear if these differences are a result of the OA or if they initially had a decrease in flexion angle which led to knee OA. Evidence from previous research would lead one to believe the stiff knee gait experienced by OA populations is a result of the progressive nature of the disease. It would seem counterintuitive that someone would walk more stiffly in order to decrease painful gait if indeed that is what they are trying to do. More than likely, the stiffer gait is a result of the stiffness within the knee due to joint damage, inflammation, and pain. More research is necessary in long term kinematic and kinetic changes in OA to understand the underlying mechanisms behind the disease progression.

Compressive Forces

Peak vertical impact ground reaction forces during healthy gait have been shown to range from about 1 to 1.2 times body weight (Hamill and Knutzen, 2009) during the stance phase of walking. However, the resultant knee joint compressive force during walking may be as high as 4 times body weight (Messier et al., 2005b).With this information in mind, it is clear that the knee joint has the potential to be severely impacted by high aberrant forces during walking. Knee joint forces for OA populations have been shown to be about 3 to 3.7 times body weight (Messier et al., 2005b; Schipplein and Andriacchi, 1991). Messier et al. (2005a) studied the knee joint loading of 10 older adults with varying degrees and locations of knee OA (i.e. severity ranges from mild to severe, and OA locations of medial, lateral, and patellofemoral compartments). The OA participants were matched with 10 healthy controls of similar age, sex, and body mass. All subjects walked at a self-selected walking pace while kinematic and kinetic data were recorded. The results of the study showed that the knee OA group experienced about a 25% decrease in compressive force across the knee, however, they also walked slower than the healthy controls. In fact, after statistically adjusting for walking speed by including walking speed as a covariate, the compressive knee joint load was nearly identical in both groups (3.67 BW in the OA group and 3.40 BW in the control group; $p = 0.49$). While the results of this study showed no differences in joint loads between the OA and control groups, they may have been influenced by the small study sample size and the lack of OA location and grade specificity. Another study reported that medial compartment OA patients encountered 4% higher knee joint reaction forces during walking compared to their healthy counterparts (Mundermann et al., 2005). While the differences reported appear to be small, the results suggest a relationship may exist between medial compartment OA and compressive knee joint loads.

Decisive evidence has not yet appeared in the literature about the relationship between knee OA and compressive forces. It is tempting to assume that an increase in compressive joint load has a detrimental effect on the knee joint structure. However, studies have shown that increased loading is actually beneficial to the health of the knee joint as indicated by an increase in knee cartilage volume with exercise (Kiviranta et al., 1988). Additionally, others studies have found that people with knee OA have been successful at lowering the loads on the affected limb in an attempt to reduce the pain during gait (Miyazaki et al., 2002; Mundermann et al., 2005).

Finally, a study by Chakravarty et al. (2008) showed that middle and older age long distance runners did not have an increase in OA progression compared to healthy non-runner controls studied over an 18 year period. The combination of these findings suggests that the response to joint loading may be dependent on the health of the knee cartilage. It can be argued that chronic compressive loads on the knee joint (such as in running) are not necessarily responsible for the development of knee OA, but rather have a large influence on the progression of OA once the disease has been acquired. This hypothesis is further supported as shown in a few studies where patients with knee OA who had higher loads at the knee during walking had an increased rate of cartilage breakdown compared to knee OA patients with lower knee loads (Miyazaki et al., 2002; Turner et al., 1985). However, true to the ambiguous nature of OA, other studies have shown that altered joint loading can lead to progressive degeneration of the articular surface in animal experiments (Buckwalter, 1995). Other research has shown that obesity is highly associated with incident OA of the hand, hip, and knee (Oliveria et al., 1999), suggesting that either the increased weight, or some other factor associated with obesity, has a degenerative influence on diarthrodial joints.

Internal Abduction Moment, Joint Laxity, and Malalignment

The internal knee abduction moment (KAM) during gait has been shown to be an important factor associated with knee OA (Cerejo et al., 2002; Mundermann et al., 2005; Sharma et al., 1998; Wada et al., 2001). The majority of the compressive load on the medial compartment of the knee is created by the external adduction moment which is countered by an internal KAM (Schipplein and Andriacchi, 1991). The external adduction moment is defined as the product of the length of the moment arm from the knee joint center and the ground reaction force (GRF) vector in the frontal plane of the knee during walking (Hunt et al., 2006). This moment acts to adduct the knee during stance into a bow-legged or knee varus position (Cerejo et al., 2002); a condition that opens the lateral joint space while closing the medial joint space of the knee, resulting in medial compartment joint space narrowing. Under a valgus stress, the medial joint space is increased under normal conditions and is described as medial joint laxity (Lewek et al., 2004). One study showed that in knee OA patients, a valgus stress at the knee resulted in a larger joint space opening when compared to healthy controls (Lewek et al., 2004). During gait, this mechanical abnormality redistributes the previously even load on the knee to a more medially directed load which may lead to increased joint space narrowing. In walking, there are typically two peaks present in the KAM. The first peak, which is associated with weight acceptance during stance, appears to have the largest influence on knee OA (Miyazaki et al., 2002).

An increase in KAM has been shown to affect the distribution of bone mineral content across the tibial plateaus (Hurwitz et al., 1998). This is important because osteophyte formation can occur with excessive or abnormal forces within a joint, which is a major component of osteoarthritis. In this study, the distribution of bone was examined between the medial and lateral sides of the tibia in 26 healthy males and females using dual energy X-ray absorptiometry. The subjects then participated in gait analysis in which kinematics and kinetics were recorded and an inverse dynamics analysis was used to calculate the internal KAM. The authors found significant differences in distribution of tibial plateau bone mineral density among the participants. They also found that the KAM was the single best predictor of the bone mineral distribution differences and established a linear relationship between the variables. Specifically, there was an increase in bone formation on the medial plateau when compared to the lateral plateau as the KAM was increased. The KAM accounted for 31% of the variation in bone mineral content redistribution and increased to 58% when body weight was accounted for (Hurwitz et al., 1998).

Other studies have found a relationship between the magnitude of KAM and the severity of knee OA (Chakravarty et al., 2008; Mundermann et al., 2005; Wada et al., 2001). For example, Mundermann et al. (2005) performed a gait analysis on 42 patients with bilateral medial compartment knee OA and 42 sex, age, height, and mass matched controls. The severity of the knee OA patients ranged from 1 to 4 on the Kellgren/Lawrence scale. The participants walked in their own shoes at a self-selected pace. The results of the study showed that the patients with more severe knee OA (KL grade of 3 or 4) demonstrated about 11% larger first peak KAM compared to their controls and about 28% greater than the less severe patients (KL grade 1 or 2). The authors also reported that even though the walking speeds were self-selected, the speeds were not different between groups, so the differences seen in the KAM cannot be attributed to walking speed.

As previously mentioned, the KAM has been associated with knee malalignment, particularly a knee varus position. According to Hurwitz et al. (2002), knee varus alignment is the best predictor of the first and second peak KAM in knee OA patients, accounting for 55% of first peak KAM variance and 56% of second peak variance. Additionally, the previously mentioned study by Mundermann et al. (2005) found that people with more severe knee OA have a larger varus alignment (5.7 degrees) than those with a less severe disease (0.3 degrees).

Malalignment in the knee has also been shown to be associated with the progression of knee OA. Cerejo et al. (2002) studied the knees of 230 OA patients who had varying degrees of

OA (KL grades 0 through 3). A KL grade of 4 was not included because it is the end stage of OA and disease progression is limited past this stage. OA progression odds ratios were determined from baseline to 18 month follow up. Knee varus or valgus alignments were measured statically using a full limb radiograph. The results of this study showed that in knees with mild OA (KL grade 2), the odds of medial compartment OA progression after 18 months was increased 4-fold by a varus alignment at baseline. In the same KL grade, the likelihood of lateral OA progression was increased 2-fold by a valgus alignment at baseline. For the moderate OA individuals (KL grade 3), the risk of progression of OA was increased 10-fold in both varus and valgus alignments.

One possible mechanism for reducing the KAM during walking is by placing a lateral wedge in patients' shoes. The majority of people with knee OA have medial compartment knee OA, and thus a laterally posted orthotic would be used. The orthotic causes a shift in the orientation of the leg (placing the knee closer to the bicycle frame) which ultimately alters the position of ground reaction force vector so that the vector is closer to the knee joint center. This decreases the distance between the GRF vector and the knee joint center (i.e. moment arm). Previous studies have in fact confirmed that wedges reduced the first peak KAM (Butler et al., 2009; Butler et al., 2007; Hinman et al., 2009; Kerrigan et al., 2002), by using wedges ranging from about 5 to 12 degrees. However, to date, the use of wedges has not been shown to be effective in slowing the progression of OA (Pham et al., 2004). For example, Pham et al. (2004) performed a 24 month intervention on 156 subjects (41 male, 115 female) to determine the long term effect of wedged insoles on knee pain and the rate of disease progression. They found that at the end of the 24 month period there was no difference in knee pain and function as measured by the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC), although they wedged insole group did have less NSAIDs intake over the two year period. They also found no difference in the rate of joint space narrowing between the two groups. While joint space narrowing is not the sole indicator of knee OA, the results may suggest there was no difference in disease progression.

Additionally, the reduction in KAM has not been strongly correlated with a reduction of pain in these studies. For example, Baker et al. (2007) performed a study with a cross-over design in which they had half of their participants wear a 5 degree lateral wedged insole for 6 weeks, and the other half wore a neutral insole. After a 4 week washout period, the groups switched. The Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) was used to assess the knee pain and function. At the end of the study, there were no statistical differences found between the two groups. However, this may be a result of a small amount of posting used in the intervention (Baker et al., 2007). Several researchers suggest that the location and angle of the wedge is subject specific, and not all patients have a positive response to the treatment (Butler et al., 2007).

Other possible methods for reducing the KAM during walking are through simple gait modification as demonstrated by several researchers (Fregly et al., 2007; Guo et al., 2007; Mundermann et al., 2008; Shull et al., 2012). Guo et al. (2007) attempted to reduce the KAM by requiring their participants to walk in an increased toe-out (foot progression) angle during walking and ascent/descent tasks. Ten participants with medial compartment OA (grade 1-4) walked in a self-selected and 15 degrees beyond self-selected foot progression angles. The results of the study showed that the participants were able to reduce their second peak KAM during the walking (40% decrease) and stair ascent (11% decrease) tasks in an increased foot progression angle compared to their self-selected foot progression angle. A potential downside to ascending stairs using a toe-out method is an increase in the first peak KAM. As mentioned previously, the 1st peak KAM appears to be more closely related to OA, and thus, a toe out method of gait change may not provide the desired load reductions in this population.

Fregly et al. (2007) also showed a promising gait modification for reducing the KAM during walking by training the participant to walk in a "medial thrusting" gait pattern. The modified gait pattern was prescribed for one patient with knee OA using an inverse dynamic optimization approach that reduced both first and second peak knee adduction moment peaks at the same time. Essentially, the cost function was to minimize the KAM with constraints that limited the how much the model could deviate from the patients specific gait pattern. The optimizer predicted a medial thrusting gait pattern. The optimization predicted a 32% reduction in the first peak KAM and a 56% reduction in the second peak KAM. Interestingly, after 9 months of gait retraining, the subject achieved a 37% and 55% reduction in first and second peak KAM, respectively. The biggest limitation of this study was the fact that only one participant was included. While these results seem very promising, the researchers also reported an increase in the external knee flexion moment during the medial thrusting gait which may actually cancel out the effects of the reduced KAM on the medial compartment force (Walter et al., 2010).

Building on the two previously mentioned studies, Shull et al. (Shull et al., 2012) attempted to reduce the KAM by having their participants walk in a toe-in foot progression angle. They had 12 subjects in their study (7 male and 5 female) with a mean age of 60 years who had radiographic evidence of medial compartment knee OA. This was an interesting gait retraining study in which they mounted a small vibrating device on the subject's shank. The subjects walked on an instrumented treadmill at a self selected pace in a normal walking pattern and then a toe-in pattern. During the toe-in trials, the patient must have maintained a shank angle of at least 0.75 degrees less than their baseline value otherwise the vibrating device would provide them with feeback and promt for increased toe-in angle. The authors found that this method of walking reduced the first peak knee adduction moment by about 11% but the second peak KAM and the knee flexion moment remained unchanged. This study provides promising results for a simple method to effectively reduce the KAM during walking, and may be a potential solution for alignining malaligned lower limb joints during future cycling studies.

In summary, existing literature supports the interconnected relationship of knee alignment with the internal knee abduction moment (Lewek et al., 2004; Wada et al., 2001). These factors have also been shown to be related to the severity and progression of knee OA (Cerejo et al., 2002; Chakravarty et al., 2008; Mundermann et al., 2005; Wada et al., 2001). However, the onset of knee OA is still unclear. On the one hand, it is possible that factors leading to joint space narrowing such as articular cartilage loss and meniscus degeneration are prerequisites for joint malalignment and laxity and lead to an increased KAM. On the other hand an increased KAM could lead to joint laxity and malalignment which may lead to increased loading and ultimately cartilage loss and meniscus degeneration. In other words, it is still unclear if joint laxity and malalignment lead to increased KAM or if the reverse is true. To make matters more confusing, we know that previous injuries such as a torn anterior cruciate ligament in the knee can lead to future development of OA. It is not clear if the OA develops because of a structural change in the knee joint, or by some other mechanism. More research is needed to clarify these questions, particularly longitudinal studies that may help to address the “chicken or the egg” conundrum. Additionally, promising gait retraining strategies have become apparent recently that may provide a reduction in the KAM. This could be very important considering the KAM seems to be so closely related to the severity and progression of knee OA.

Obesity and Associated Gait Changes

Obesity has been shown to be strongly associated with knee OA (Coggon et al., 2001; Felson et al., 1997; Leach et al., 1973; Manninen et al., 1996). Felson et al. (1997) performed analysis on data from a longitudinal study of the Framingham Cohort to determine possible risk factors associated with knee OA. Radiographs were originally taken on subjects between the years of 1983-1985, of whom, 979 subjects were free of OA. Approximately 10 years later (1992-1993), radiographs were taken in the same manner on 598 of the original subjects. Several risk factors were found for developing OA including being overweight at baseline. Additionally, it was found that for every 10 lbs. of weight gained during the time period studied, the risk of developing OA was increased by 40%. Coggon et al. (2001) performed a study in England in which 525 men and women were scheduled for surgery because of knee OA. The patients were matched by age, sex, and family practitioner with a control group. The results of the study showed that a BMI was significantly correlated to developing knee OA. For those individuals with a BMI less than 20 kg/m^2, the odds ratio was only 0.1 (95% CI: 0.0 – 0.5). However, for those who had a BMI greater than 30 kg/m^2, the odds ratio increased to 6.8 (95% CI: 4.4 – 10.5). Together these studies show strong evidence for a link between obesity and the development of knee OA.

Even though a link has been established between obesity and OA, it is not completely clear why obesity increases the risk of OA. Past studies have revealed that obese people clearly have increased vertical GRF and plantar pressures compared to normal weight populations (Browning and Kram, 2007; Hills et al., 2001; Messier et al., 1996; Mickle et al., 2006; Wearing et al., 2006a, b). However, running studies have shown high peak ground reaction forces too (actually higher forces than seen in obesity), and no relationship has been established between running and OA (Paty, 1994). In fact, a study in the early 80's has shown that compressive forces improve the health of the knee joint through an increase in cartilage synthesis (Palmoski et al., 1980). While a direct link between increased ground reaction forces and OA has not been made, it has been shown that increased GRF (by means of an increase in weight) may have an indirect effect on the progression of OA because of altered gait mechanics (Browning and Kram, 2007; Messier, 1994; Messier et al., 2005b; Schipplein and Andriacchi, 1991). Browning and Kram (2007) compared the gait of 10 obese individuals with 10 normal weight controls and found that obese individuals had greater absolute vertical GRF, sagittal plane knee moments, and step width. They also found that the obese individuals were able to reduce the GRF by simply walking slower than their healthy counterparts. Additionally, Messier (1994) found increased eversion rearfoot motion and an increased forefoot abduction (in relation to the rearfoot) in obese individuals. A change in alignment at the ankle may propagate up through the kinetic chain of the lower limb to have a negative effect on the knee. For example, if one were to have a more everted foot (similar to a collapsed arch), the knee would be forced into a more abducted position (knee valgus), which may compromise the ability of the knee joint to evenly distribute loads.

A factor likely more important to link OA and obesity is the compressive force across the knee joint. Messier et al. (2005b) studied the effects of weight loss on the knee joint compressive force in 116 patients with knee OA. After a 6 – 18 month follow-up, it was found that weight loss was associated with a reduction in compressive knee joint force. Specifically, for every 1 kg of weight loss, the average knee joint compressive force was decreased by about 40 Newtons (N). They also found a significant decrease in the knee internal abduction moment - which has been found to be associated with the progression of knee OA (Baliunas et al., 2002; Chakravarty et al., 2008) as previously mentioned.

While the link between obesity and OA is not completely understood, evidence clearly suggests a meaningful association. A reasonable explanation is that secondary gait changes are likely responsible for OA development, while the increased GRF (leading to increased compressive force across the knee joint) may play a more important role in the disease progression. As shown previously, when the knee experiences increased internal abduction moments, there is potential for a redistribution of bone across the tibial plateau. This may lead to break down of the cartilage and meniscus, followed by osteophyte formation.

CYCLING IN HEALTHY SUBJECTS AND IMPLICATIONS FOR OSTEOARTHRITIS

In cycling, terminology and bicycle part names can be overwhelming. To help make it less overwhelming, refer to the bicycle diagram in figure 1 that identifies the bicycle parts relevant to this review. Also, refer to figure 2 which depicts the pedaling/crank cycle typically seen in cycling. During pedaling, the top most position of the crank and pedal is referred as top dead center, while the bottom most position is bottom dead center. These positions correspond to 0 and 360 degrees for top dead center and 180 degrees for bottom dead center.

Typical Kinematics, Kinetics, and Muscle Activation

Cycling kinematics, kinetics, and muscle activation can be influenced by many bicycle and rider manipulations, as will be shown in the following literature review. However, before discussing the influences of these manipulations, it is important to provide the reader with some background information about typical kinematics, kinetics, and muscle activity during cycling.

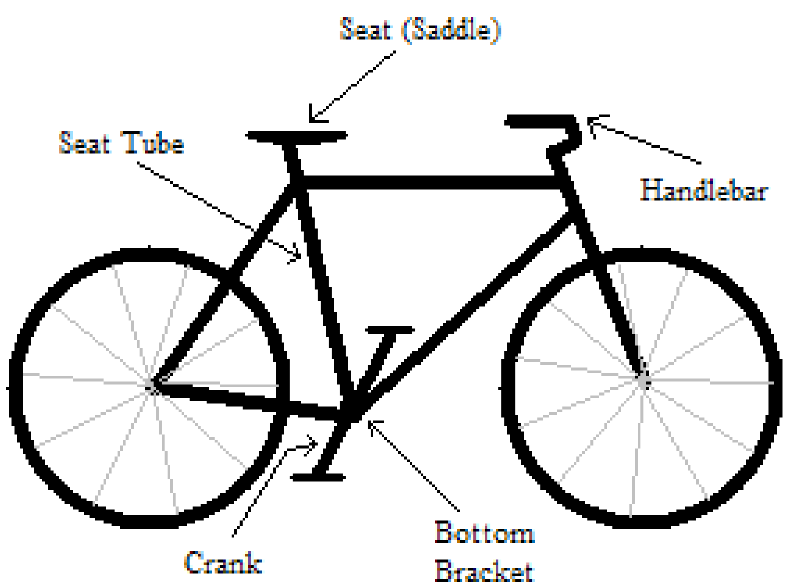

Figure 1: Diagram with labels of key components of the bicycle

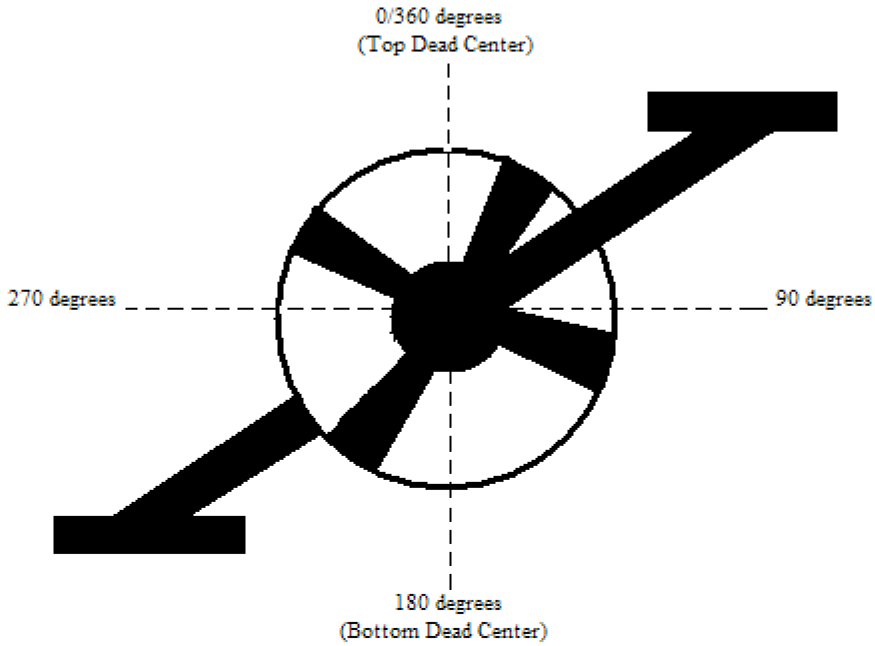

Figure 2: Diagram of the crank/pedaling cycle

Kinematics

Many authors have reported on the lower extremity joint kinematics during cycling, (Bailey et al., 2003; Bini et al., 2010; Carpes et al., 2009; Damiano et al., 2011; Edeline et al., 2001; Edeline et al., 2004; Ercison et al., 1988; Ericson et al., 1986a; Ericson et al., 1985a; Faghri and Trumbower, 2005; Gregersen and Hull, 2003; Hamley and Thomas, 1967; Heil et al., 1997; Heil et al., 1995; Mileva and Turner, 2003; Nordeensnyder, 1977; Peveler and Green, 2011; Price and Donne, 1997; Reiser et al., 2002; Reiser et al., 2001; Sanderson et al., 2006; Savelberg et al., 2003; Tamborindeguy and Rico Bini, 2011; Too and Landwer, 2000; Umberger and Martin, 2001) while general joint motions are similar due to the cyclical nature of the bicycle, differences exist depending on the seating arrangement and specific manipulation in each paper.

To give the reader an idea of typical joint motions seen in cycling, Ercison et al. (1988) showed that during normal cycling, defined as 120 Watt workload, 60 rpm pedal cadence, a seat height of 113% of the distance between the ischial tuberosity and the medial malleolus, and an anterior foot position, the mean hip, knee, and ankle ranges of motion were 38 degrees (32-70 degrees), 66 degrees (46-112 degrees), and 24 degrees (2 degrees plantarflexion to 22 degrees dorsiflexion) respectively. They also showed that the peak hip extension occurred right at bottom dead center during the pedal cycle, while the knee flexion peaked just before, and the ankle dorsiflexion peaked just after. Nearly 22 years later Bini et al. (2010) discussed kinematic changes with several different variable manipulations. In their reference position (seat height at 100% of greater trochanter height), typical joint ranges of motion were about 55, 65, and 25 degrees for the hip, knee, and ankle respectively. The differences in the hip angle between the two studies may be a result of trunk lean or joint convention, but it can be seen that joint range of motion during cycling remains similar even over a large span of time.

Most of the articles on joint kinetics have reported two dimensional joint angle calculations in the sagittal plane. However, it has been suggested that movements important to joint safety occur in all three planes of motion as discussed by Francis from (Burke, 1986). Thus, it is necessary to analyze joint motions in the frontal and transverse planes, not just in the sagittal plane. Umberger and Martin showed that using a 2D analysis of the cycling motions, resulted in large deviations from the 3D analysis (hip angle difference of 34 degrees) and therefore, care should be taken when interpreting 2D analysis of motions and 3D analysis should be used when possible (Umberger and Martin, 2001). Previous research showed that during normal bicycling with healthy subjects, frontal plane knee angles ranged from about 2 to 4 degrees of abduction to 1 to 6 degrees of adduction during the crank cycle (Bailey et al., 2003; Umberger and Martin, 2001). The maximum abduction angle occurred at about 90 to 200 degrees during the crank cycle, and the maximum adduction angle occurred at about 300 to 360 degrees during the crank cycle (Bailey et al., 2003). The small range of motion in the frontal plane indicates that during normal healthy cycling, there is not a large amount of abduction/adduction and the knee remains fairly neutral.

Establishing a proper seating arrangement for people with osteoarthritis to improve pain and function in the joint is essential for successful treatment of the OA symptoms. Analysis of joint kinematics in different seating arrangements may provide a good starting point to discover the proper configuration for pain reduction and functional improvement.

Kinetics

Like kinematics, kinetics in cycling has been studied by many researchers (Bini et al., 2010; Boyd et al., 1996; Boyd et al., 1997; Ericson, 1986; Ericson, 1988a; Ericson et al., 1986a; Ericson et al., 1986b; Ericson et al., 1985a; Gregersen and Hull, 2003; Gregersen et al., 2006a, b; Gregor et al., 1985; Kautz et al., 1994; Kautz and Neptune, 2002; Marsh et al., 2000; Neptune and Herzog, 1999; Neptune and Hull, 1995, 1998; Neptune and Kautz, 2000; Prilutsky and Gregory, 2000; Redfield and Hull, 1986; Reiser et al., 2002; Reiser et al., 2004; Ruby and Hull, 1993; Too and Landwer, 2000; Wolchok et al., 1998). Using Broker from (Burke, 2003) as a representative example, it is apparent that ankle, knee, and hip joints mainly demonstrate extensor moments during the down stroke phase of the pedaling cycle (0 to 180 degrees), while the knee extensor moment peaks at approximately 90 degrees. The hip, knee, and ankle joints reached maximum extension moments of 30, 40, and 40 Nm respectively (workload of 250 W and pedal cadence of 90 rpm). Around 125 degrees into the pedaling cycle, the knee switches from an extensor moment to a flexor moment even before the leg is fully extended. During the upstroke of the pedaling cycle (180 to 360 degrees), hip moments are initially extensor but become flexor near top dead center. The knee initially has a flexor moment but switches to extensor near the 270 degree mark during the pedal cycle. The ankle joint exhibits mostly plantarflexion moments throughout the cycle but experiences brief dorsiflexion moments around the upper most 30 degrees (approximately).

A study by Gregersen and Hull (2003) is one of the very few studies that analyzed the kinetics of the frontal and transverse planes in cycling (although only the knee joint was considered). They found large variations in joint moments between subjects when pedaling at a 90 rpm cadence and a 225 W workload. Overall, subjects mainly experienced muscular resistance of varus and internal rotation moments during the power phase (defined here as the phase when the knee experienced an extension moment) and valgus and external rotation moments during the recovery phase (when the knee experienced flexor moments). On average, subjects experienced peak adduction moments of 6 Nm, peak abduction moments of 7 Nm, peak internal rotation moments of 1 Nm, and external rotation moments of 2 Nm. The moments across the joints during cycling can be influenced by several factors. Knowledge of joint moments may help to establish proper riding configurations for reducing stress in the joints. On the whole, individuals with osteoarthritis may largely benefit from cycling if an optimized position and workload can be established which elicits small joint moments, particularly at the knee joint.

Muscle Activation

The measurement of muscular activity through the use of electromyography (EMG) during cycling has been studied by several researchers (Baum and Li, 2003; Chapman et al., 2008a; Chapman et al., 2008b; Chapman et al., 2006; Cruz and Bankoff, 2001; Dagnese et al., 2011; Dorel et al., 2008; Duc et al., 2008; Eisner et al., 1999; Ericson, 1988b; Ericson et al., 1985b; Gregersen et al., 2006b; Jorge and Hull, 1986; MacIntosh et al., 2000; Marsh and Martin, 1995; Neptune et al., 1997; Neptune et al., 2000; Prilutsky and Gregory, 2000; Raasch et al., 1997; Rouffet and Hautier, 2008; Ryan and Gregor, 1992; Sanderson et al., 2006; Sarre et al., 2003; Savelberg et al., 2003). Ericson et al. (1985b) did an EMG study on the activity of 11 different muscles in the lower extremities while manipulating several different factors including work load, pedal rate, seat height, foot position, and use/disuse of toe clips. The muscles analyzed were gluteus maximus and medius, rectus femoris, vastus lateralis and medialis, biceps femoris, medial hamstring, medial and lateral gastrocnemius, soleus, and tibialis anterior. They found that the three most active muscles during cycling were the vastus medialis, vastus lateralis, and soleus, accounting for 54, 50, and 37% of their isometric maximum voluntary contraction (MVC) respectively using an integrated EMG value (iEMG). There was low muscular activity found in the gluteus maximus (7% MVC), gluteus medius (11% MVC), rectus femoris (12% MVC), biceps femoris (12% MVC), medial hamstrings (10% MVC), and tibialis anterior (9% MVC). It was also found that the medial and lateral gastrocnemius were moderately active with values of 19 and 32% MVC respectively. These values indicate that during cycling, monoarticular muscles (vastus medialis, vastus lateralis, and soleus) are much more active than the biarticular muscles. Most muscle activity during this study had maximum values between 90 and 180 degrees during the crank cycle.

Ryan and Gregor (1992) showed the activity of 8 lower extremity muscles throughout the crank cycle when cycling at 250 W and 90 rpms. The gluteus maximus was active from top dead center until about 130 degrees. The biceps femoris and semitendinosis followed similar patterns and were active for much of the cycle likely due to the biarticular nature of the muscles. There was a small region of inactivity for the two muscles from approximately 270 degrees until top dead center during the upstroke. The rectus femoris muscle was active from about 90 degrees before top dead center until approximately 90 degrees after top dead center; likely active during hip flexion in the upstroke and knee extension during the down stroke. The vastus lateralis, a powerful knee extensor, was active from about 45 degrees before top dead center until about 90 degrees after top dead center. The gastrocnemius was active from top dead center until about 270 degrees; however, the bulk of its activity was from 45 to 180 degrees. The majority of the soleus activity was during the first 135 degrees of the down stroke, while the tibialis anterior was active during the last 90 degrees of the upstroke. As expected, the majority of the muscles were highly active during the propulsion phase of the crank cycle. This may be an indication that peak loading to the knee occurs during this phase of cycling.

When compared to walking, Ericson et al. (1985b) showed that muscular activity in cycling is about the same or much less in most muscles of the lower extremity. The two muscles that had much greater activity in cycling were the vastus medialis and lateralis. The authors state that the large amount of activity in these two muscles supports the general opinion that cycling is a good mode of exercise for strengthening the quadriceps muscles. However, it is possible that the high activity of these two muscles may also play a role in the common knee injuries seen in cycling.

Positioning and Workload

Researchers have shown that manipulations such as frame geometry (namely the seat tube angle), seat (also referred as “saddle”) height, crank arm length, foot position, pedaling cadence, and workload can all impact the cyclist. How these manipulations relate to variables such as kinematics, kinetics, muscle activation, metabolic efficiency, and power output during cycling will be discussed in the proceeding review.

Seat Tube Angle

The seat tube angle is the angle formed between the rear of the seat tube and level ground. The angle of the seat tube dictates how far forward or rearward relative to the bottom bracket a rider will sit (assuming no changes in fore and aft position of the seat). The typical range of seat tube angles on road bicycles is between 70 and 76 degrees (Ricard et al., 2006). Changes in seat tube angle have been shown to have an effect on power output (Price and Donne, 1997; Rankin and Neptune, 2010; Ricard et al., 2006; Umberger et al., 1998), cardiorespiratory response (Heil et al., 1997; Heil et al., 1995; Price and Donne, 1997), kinematics (Heil et al., 1997; Heil et al., 1995; Price and Donne, 1997; Umberger et al., 1998), and muscle activity (Rankin and Neptune, 2010; Ricard et al., 2006).

Heil et al. (1995) investigated the effect of seat tube angle on cardio respiratory responses in 25 trained triathletes and cyclists. They analyzed 4 seat tube angles (69, 76, 83, and 90 degrees) and their effects on oxygen consumption, heart rate, ventilation, and rating of perceived exertion during a 10 minute submaximal cycling test. They found that oxygen consumption, heart rate, and ratings of perceived exertion were all lower in the in the steeper seat tube angles (83 and 90 degrees) when compared to the 69 degree seat tube angle. They also reported on kinematic variables and noted that there was greater hip extension and ankle plantar flexion when the seat tube angle was increased. The authors concluded that the 69 degree seat tube angle was the only condition that was detrimental to performance based on cardiorespiratory responses. In a later article, Heil et al. (1997) found that cyclists optimized their oxygen consumption at frame geometries similar to the setup of their personal bicycles. This finding suggests a training effect may exist for experienced cyclists.

Price and Donne (1997) studied the effects of changing the seat tube angle as well as the seat height on the cardiorespiratory response and the lower extremity kinematics. They analyzed 14 experienced male cyclists riding at a constant workload of 200 watts at three seat tube angles (68, 74, and 80 degrees) and three seat heights (96, 100, and 104% of greater trochanter height). The investigators found that at all seat heights, VO2 was significantly lower and power and efficiency significantly higher at a seat tube angle of 80 degrees compared to the other two seat tube angles, and at 74 degrees compared to 68 degrees. In terms of kinematics, no changes were observed in ankle or knee angles with respect to changing seat tube angles. However, for the hip, both the minimum and maximum angles increased significantly at the 68 degree seat tube angle compared to the 80 degree angle at all seat heights. There was approximately a 10 degree difference for maximum angle and an 8 degree difference for minimum angle. Additionally, the maximum and minimum hip angles in the 68 degree condition were significantly larger than the 74 degree seat tube angle at 96 and 104% trochanter height conditions only.

Umberger et al. (1998) studied seat tube angle on peak power, mean power, and fatigue during short term anaerobic performance (15 seconds of maximal effort). The seat tube angles analyzed were 69, 76, 83, and 90 degrees. They also reported kinematics of the trunk, hip, knee, and ankle during their test of 12 healthy active participants. They found that peak power was significantly higher in the 69 degree seat tube angle when compared to the 90 degree angle, and the mean power was significantly higher in the 69, 76, and 83 degree angles when compared to the 90 degree seat tube angle. In terms of kinematics, mean hip angles increased as seat tube angle increased (range between 88 and 107 degrees) with no differences found in range of motion. The mean knee angle was greater at the 90 degree seat tube angle compared to all other seat tube positions (119 degrees compared to approximately 115 degrees) but the range of motion remained unchanged. The ankle range of motion was greater in the 83 and 90 degree conditions compared to the 69 and 76 degree conditions (41 compared to 35 degrees). The authors conclude that maximum short term power is greater in shallower seat tube angles which are accompanied by decreased mean hip angle which could affect the muscle lengths and moment arms of the hip extensors.

Ricard et al. (2006) studied the effect of seat tube angle (72 and 82 degrees) on muscle activation and anaerobic power during a Wingate test. The investigators analyzed 4 muscles of the lower extremity including vastus lateralis and medialis, semimembranosus, and the biceps femoris of 12 experienced cyclists. They found that variation in seat tube angle had no effect on the power output during the Wingate test. However, they did find that in all muscles analyzed, muscle activation was lower in the 82 degree seat tube angle. Although, only the biceps femoris was found significant and showed an approximately 32% decrease in muscle activation compared to the 72 degree condition. An increased seat tube angle places the rider in a more forward position, which allows the rider to produce greater hip extension torque with lower levels of biceps femoris activation. The authors conclude that an increased seat tube angle may reduce muscular fatigue without affecting maximal power production.

Rankin and Neptune (2010) performed a muscle actuated forward dynamics simulation of pedaling to determine the optimal seat height, pelvic orientation, and seat tube angle for maximum power output. They studied a range of seat tube angles from 65 to 110 degrees and found that a seat height at 102% of greater trochanter height, and a seat tube angle of 85.1 degrees, produced the most power. They attributed their findings to the lower extremity kinematics in which their optimal position placed the major power producing muscles in the most favorable region of the force-length-velocity curves. The authors did note however, that power differences in changes in seat tube angles varied at most by 1%, which the authors believe to be the result of similar lower extremity joint kinematics for each position studied.

To summarize the findings from manipulating seat tube angle, previous literature indicates that seat tube angles greater than 76 degrees improve cardiorespiratory efficiency (Heil et al., 1995) and decrease muscle activation (Ricard et al., 2006), but has an ambiguous effect on overall power. Some research suggests increased power at a seat angle less than 70 degrees (Umberger et al., 1998), others claim increased power at angles greater than 80 degrees (Price and Donne, 1997; Rankin and Neptune, 2010), and still others claim no difference in power output with changing seat tube angle (Ricard et al., 2006). An increase in seat tube angle places the rider in a more forward position relative to the crank, which increases hip extension but not hip ROM, and has relatively little effect on the knee and increases the ankle ROM (Heil et al., 1995; Price and Donne, 1997; Umberger et al., 1998).

The effect of changing seat tube angles on knee osteoarthritis may not be easily deducted from the existing literature. There may be good reason to suggest that increasing the seat tube angle would alleviate symptoms of knee OA based on muscle activation findings from (Ricard et al., 2006) but no research exists to accept or refute this claim. Additionally, to our knowledge, there are no studies that investigate the effect of seat tube angle on joint kinetics, which would be an important research topic for finding the optimal configuration for reduced joint loading.

Crank Arm Length

The length of the crank arm has been studied by a few investigators as it relates to cycling performance (Hull and Gonzalez, 1988; Inbar et al., 1983; Macdermid and Edwards, 2010; Martin and Spirduso, 2001; Morris and Londeree, 1997; Too and Landwer, 2000; Yoshihuku and Herzog, 1996) as well as lower extremity kinematics (Barratt et al., 2011; Too and Landwer, 2000) and kinetics (Barratt et al., 2011). The crank arm is the part of the bicycle that the pedal attaches to. Its distance is measured from the center of the axis of rotation of the pedal to the center of the axis of rotation of the crank (bottom bracket). The crank arm is not adjustable on a bicycle but can be replaced with other cranks of varying lengths. Generally speaking, longer crank arms allow the rider to sustain a given work rate with less force production because of the extended moment arm of the lever. However, a longer crank arm will also increase the range of motion (ROM) that the lower leg moves through. Thus the crank arm length may be a variable of interest when fitting bicycles for diseased populations such as in knee osteoarthritis.

Inbar et al. (1983) studied the effect of crank arm length on power output in the Wingate anaerobic cycle test. They analyzed crank lengths of 125 – 225 mm and found that there was an 8% difference in cycling power between the two most extreme crank lengths and that crank length was highly correlated to leg length. This would indicate that crank length selection has a large impact on power output. However, since this publication, other researchers have noted that the Wingate test does not elicit maximum short term cycling power (Dotan and Baror, 1983; Martin and Spirduso, 2001), and that Inbar et al. (1983) did not account for pedaling rate (Martin and Spirduso, 2001), which would likely diminish the large correlation between crank length and power output.

In 1990, Yoshihuku and Herzog (1996) performed a maximum muscular power output simulation, comparing two different optimal muscle length assumptions to identify the optimal riding configuration. They attempted to optimize crank length, pedaling rate, pelvic inclination, and seat height, and found that large changes in any one parameter elicits relatively small changes in the total power output. Additionally, they noted that the optimization of a single variable is simultaneously dependent on all other variables involved. They concluded that maximum power output varies 0-10% for crank lengths between 130 and 210 mm depending on how the optimal muscle length is defined.

Morris and Londeree (1997) tested the long term effects of VO_2 with 3 different crank arm lengths (165, 170, and 175 mm). They allowed their subjects to get used to each crank arm length prior to testing by requiring 2 weeks of riding at 225 km per week before each test session. They had each subject ride for 105 minutes and found that oxygen consumption changed between crank arm lengths, but when correlated with leg lengths, no significant correlations were found. They also noted that taking 2 weeks to adapt to crank lengths was unnecessary which was in agreement with a later article by Neptune and Herzog (2000) who showed that muscular coordination adaptations occur within the first 10-20 cycles of an unfamiliar task.

Martin and Spirduso (2001) tested the maximal power output at 5 different crank lengths (120, 145, 170, 195, and 220 mm) for 16 cyclists while accounting for pedal rate and pedal speed. They found that the 145 and 170 mm cranks produced a significantly larger amount of power than the 120 and 220 mm cranks, but only by 4%. They also found that the optimal pedaling rate decreased with crank length, but the optimal pedaling speed increased with crank length. Additionally, they found that optimum crank length was significantly correlated to the leg and tibia length (20% of leg length or 41% of tibia length) however, these values accounted for 20.5 and 21.5% of variability in maximum power output respectively when pedal speed and pedal rate were controlled.

A more recent article by Macdermid and Edwards (2010) studied the effect of three different commonly used crank lengths (170, 172.5, 175 mm) on supramaximal and isokinetic power output, and maximal aerobic capacity in 7 female cross country cyclists of similar stature.

Contrary to previous literature (Inbar et al., 1983; Martin and Spirduso, 2001), the authors found that a small change in crank arm length (170 mm to 175 mm) elicited a significant response to performance. They showed that the amount of time it takes to achieve peak power output during supramaximal exertion is reduced with the 170 mm crank length when compared to the 175 mm crank length. No other differences were found which may indicate that shorter crank arms are beneficial for short duration bursts of power and are not detrimental to long term aerobic capacity suggesting an advantage in race situations.

Barratt et al. (2011) studied the effects of crank length on joint specific power using inverse dynamics in 15 experienced cyclists. They wanted to know how crank length changes would affect the relative lower extremity joint powers during maximum cycling effort when 1: pedal rate was optimized for maximum power for each specific crank length and 2: when pedal rate was constant at 120 rpm. They analyzed 5 different crank lengths (150, 165, 170, 175, 190 mm) and found that joint specific powers did not differ across the crank lengths when pedal rate was optimized for maximum power. However, they did find that when pedal rate was held constant at 120 rpm, the 150 mm cranks resulted in a greater knee power and smaller hip power compared to the 190 mm cranks. Additionally, the authors found that increasing crank length resulted in an increase in angular velocities of the lower extremity joints due to the larger joint excursions. This may suggest that the joint specific powers in the lower extremity are directed by the shortening velocities of the muscles spanning the joints. The results of this study indicate that joint specific power is reliant on pedaling rate. When pedaling rate is optimal for maximum power, changes in crank length do not affect joint specific powers or lower extremity joint powers.

To summarize findings from crank length changes, the literature suggests that optimal crank length for peak power to be anywhere between 145 and 180 mm. It was also noted that when pedal rate and pedal speed are optimized for maximum power output, minimal changes in overall power and joint specific power are identified with crank length changes. However, if pedaling rate is held constant, smaller crank lengths are likely to result in less knee flexion power and more hip extension power. Shorter crank arm lengths may provide an advantage in short duration bursts of power as in a race scenario. It can also be seen that changes in crank arm length have relatively little effect on long duration aerobic capacity.

The effect of crank length changes on knee osteoarthritis is still unclear. It can be theorized from Barratt et al. (2011) that longer crank lengths may benefit the knee in terms of reducing muscular power. However, Barratt et al. (2011) also showed that increasing crank arm length increases knee and hip joint excursion, and in turn, angular velocities. It is unclear if a person with osteoarthritis would benefit from larger or smaller joint ranges of motion while cycling.

Seat Height

Probably the most influential factor in cycling performance is the seat height. A study by Hamley and Thomas (1967) was conducted to determine the optimal seat height for maximum anaerobic power output in a cycle ergometer test. Subjects were to achieve a predetermined power output of 500 kg*m/min as quickly as they could at various seat heights. The trials that achieved the shortest time to the predetermined power output resulted in the optimal seat height. The results suggested that the optimal seat position was located at 109% of the pubic symphysis height (the distance from the floor to the pubic symphysis usually measured with cycling shoes on).

Over the years, other methods of determining proper seat height have been developed. Some authors have determined the optimal seat height by measuring from the floor to the bony prominence of the greater trochanter. Nordeensnyder (1977) found that the optimal seat position for the most efficient oxygen consumption was at 100% of trochanter height when compared with oxygen consumption of 95% and 105% of trochanter height. Price and Donne (1997) performed a similar test of different seat heights at different seat tube angles. They tested seat heights of 96%, 100%, and 104% of trochanter height and found that the 104% height resulted in the largest oxygen consumption and increased participants' heart rate. They did not find differences between the 100% and 96% heights, suggesting a range of optimal seat heights between 96 and 100% of trochanteric height.

Greg Lemond, a three time Tour de France winner, has also recommended a method to determine optimal seat height. His method uses the pubic symphysis height multiplied by 88.3% to determine the seat height from the center of the bottom bracket to the top of the saddle (Burke, 2003). While this method is similar to the method developed by Hamley and Thomas, it does not account for the length of the crank arm (the arm the pedal is attached to). The crank arm length on a bicycle is not one set length and therefore this seat height method may result in different seat heights for different bikes and riders.

Another method to determine seat height relates to the protection of the knee joint during cycling. This method, proposed by Holmes et al. (1994) recommends positioning the seat height so that the angle of the knee, when the foot is in the lowest crank position (bottom dead center), is at 25 – 35 degrees. The authors note that if the seat height is too low, there will be pain associated with the posterior knee, and if too high the pain will likely be in the anterior knee. The authors attribute the reasoning behind this specific range of knee angle to several potential knee injuries during cycling: chondromalatia, patellar tendinitis, quadriceps tendinitis, medial patella femoral ligament irritation/medial patella femoral plica, iliotibial band syndrome, and biceps tendonitis. The first four mentioned conditions are related to the anterior knee and are likely to occur if the saddle is too low or too far forward. The authors recommend a knee angle of 25 degrees for each of these conditions. Iliotibial band syndrome is commonly seen when the band actively crosses the lateral femoral condyle at a knee angle of approximately 30 degrees and is usually accompanied by internal tibial rotation. The authors recommend a knee angle between 30 and 35 degrees for this condition. Biceps tendonitis may occur if the saddle height is too high. This position necessitates increased knee extension which may stress the posterior knee. For this condition, the authors recommend a knee angle between 30 and 35 degrees (Holmes et al., 1994).

Peveler and Green (2011) compared the economy and anaerobic power between the Holmes's method and the Hamley and Thomas method to determine how closely each relates to one another. They showed that the use of the Hamley and Thomas method of a saddle height set at 109% of pubic symphysis height resulted in different saddle heights than the Holmes method and fell out of the 25-35 degree knee angle range 73% of the time. Furthermore, the 25 degree knee angle position resulted in significantly better economy than both the 35 degree condition and the 109% pubic symphysis height condition (44.77 ml kg min vs. 45.22 and 45.98 ml kg min respectively). The 25 degree knee angle condition also resulted in a greater mean power output than the 35 degree condition (672.37 W vs. 654.71 W), and the mean power of the 109% pubic symphysis height condition (662.86 W) was also greater than the 35 degree condition. Peveler and Green (Peveler and Green, 2011) concluded that the use of the 25 degree knee angle method resulted in the best performance while keeping the knee within the desired range for injury prevention.

Vrints et al. (2011) studied the saddle position on the moment generating capacity of the lower extremity joints during maximum effort cycling for 5 seconds. The seat heights chosen were 109% of inner leg length plus or minus 2 cm. Their results show that a decrease in seat height results in a decrease in maximum power output accompanied with a decrease in moment generating capacity of the rectus femoris, biceps femoris, and vastus intermedius at the knee. No changes were found in the hip and ankle joints suggesting lower saddle positions mainly affect the knee joint kinematics which in-turn affect the moment generating capacities of the muscles surround the joint.

Perhaps more relevant to the osteoarthritic population is the effect of seat height changes on joint loading. Ericson and Nisell (1987) studied the effect of three different seat heights (among other manipulations) on 2-dimensional, patellofemoral joint forces in the sagittal plane. They used a kinetic model in combination with previously collected joint moments to calculate the estimated joint forces. The three seat heights were 102, 113, and 120% of the distance between the ischial tuberosity and medial malleolus. They found an inverse relationship between saddle height and patellofemoral joint forces. They showed that as the seat height was decreased, the knee load moment and knee joint angle increased, which caused an increase in patellofemoral joint forces.

Ericson and Nisell (1986b) also reported on the tibiofemoral joint forces during cycling over the same seat heights as mentioned in the preceding paragraph. Similar to the patellofemoral joint results, they found that as the seat height was decreased from the high to low position, the tibiofemoral compressive forces increased. Force magnitudes were reported at approximately 1.3 times body weight in the low seat position and 0.8 times body weight in the high seat position. However, no significant changes in joint force occurred in the anterior direction (average forces of about 0.05 times body weight).

A more recent article (Tamborindeguy and Rico Bini, 2011) discussed the sagittal plane compression and shear forces across the tibiofemoral joint with small variations in seat heights (97, 100, and 103% of greater trochanter height). They tested nine cyclists at a low workload and found no differences in either shear or compressive forces across the knee joint. The authors were in agreement with many other studies that suggest knee loading is more affected by work load than other cycling parameters.

Cadence and Workload

A major topic studied by many researchers is the effect of cycling cadence and workload on performance (Ericson and Nisell, 1988; Martin and Spirduso, 2001; Sanderson et al., 2000) as well as kinematics (Bini et al., 2010; Edeline et al., 2004; Hull and Gonzalez, 1988), kinetics (Bini et al., 2010; Ericson et al., 1986b; Ericson et al., 1985a; Ericson and Nisell, 1986b, 1988; Neptune and Herzog, 1999; Neptune et al., 1999; Redfield and Hull, 1986; Sanderson et al., 2000), and muscle activation (Baum and Li, 2003; Ericson, 1988a; Ericson et al., 1985b; Jorge and Hull, 1986; MacIntosh et al., 2000; Marsh and Martin, 1995; Neptune et al., 1997; Sanderson et al., 2006; Sarre et al., 2003).

A study by Bini et al. (2010) analyzed the lower extremity joint kinematics and kinetics of 9 untrained male cyclists. They found that increases in pedal cadence from 40 to 70 rpm and increases in workload from 0 to 10 N had no effect on hip or knee joint kinematics but mean ankle angle did increase and ROM decreased with an increase in pedal rate. They also found that while pedal cadence had no effect on the mechanical work of the joints, an increase in workload did result in an increase in mechanical work for each of the lower extremity joints.

In other studies, Ericson et al. (1985b) found that increasing the cycle workload increased the activation of all muscles studied, and increasing pedal rate increased all muscular activity except for rectus femoris and biceps femoris. Neptune et al. (1999) showed that as pedal rate increased in their study, bilateral asymmetries typically seen in cycling tended to diminish. Redfield and Hull found that increasing pedal rate increased ankle, knee, and hip moments, but decreased the torque at the crank (likely due to the force velocity relationship). However, when compared to Ericson et al. (1986) notable differences were seen at the knee joint. For example, Ericson et al. (1986b) discovered that hip moments were significantly impacted by both pedal rate and workload, however, the knee moment was only influenced by the workload. The differences seen may be due to differences in pedal rates tested.

To summarize, even though cadence and workload have little effect on the lower extremity joint kinematics, noteworthy differences are seen in the joint kinetics and muscle activation. In most studies reviewed, as workload increased, the ankle, knee, and hip joint moments increased as well. Typically, an increase in joint moments were seen with increasing pedal cadence, but wasn’t always the case. Muscle activation was most influenced by workload even though pedal cadence did impact most muscles. These results indicates that workload has a much larger impact on joint kinetics and muscle activation and should be highly considered when planning cycling interventions for individuals with knee injuries such as osteoarthritis. It is important if the patient is using a bicycle with gears that they do not use too hard of a gear, for fear of increasing joint loads and perhaps increasing joint pain.

Lower Limb Alignment and the Effects of Shoe Wedges and Foot Progression Angles

Alignment

To this point in the review it is clear that cycling reduces loading on the knee joint by placing the majority of the rider's body weight on their seat during seated cycling. However, cycling produces a great demand on the muscles and joints of the lower limbs, as they are the driving force in propelling the bicycle forward. Thus, knee injuries are still the leading complaint in cycling which has strong indication for an overuse injury mechanism. For example, a common overuse injury during cycling, patellofemoral pain syndrome, is thought to occur because of an adduction (varus) knee moment during the downward pedal stroke (Boyd et al., 1997; Wolchok et al., 1998). Thus, proper alignment of the lower limbs during cycling is an important factor for reducing overuse injuries experienced by the rider (Bailey et al., 2003; Gregersen et al., 2006a; Ruby and Hull, 1993).

Bailey et al. (2003) studied the frontal plane motions of 24 experienced male cyclists. The cyclists were either classified as injury free or had a history of knee overuse injuries. The researchers found that the cyclists with a history of injury exhibited 1.9 degrees greater peak shank adduction angle and 4.9 degrees greater ankle dorsiflexion angle compared to the injury free group. Additionally, the shank angle of the healthy subjects hovered around a neutral position (range of -2.5 degrees abduction to 1 degrees adduction) while the shank angle of the injured group remained in an abducted position throughout the crank cycle (range of -4 to -2 degrees). The average differences between the injured and injury free group was about 2.5 degrees throughout the crank cycle. The authors concluded that the results support the potential for a possible mechanism for overuse knee injuries during cycling.

While proper lower limb alignment is important, Ruby and Hull (1993) also showed that too much cleat restriction in the cycling shoe may be unfavorable to knee loading. They studied the adduction/abduction (varus/valgus) and inversion/eversion knee moments of cyclists while using four different pedal platforms. One pedal did not allow any movements between the cycling shoe and the pedal, while the other three allowed for either medial/lateral translation, adduction/abduction rotation, or inversion/eversion rotation. The authors found no differences in knee loading between the fixed platform and the pedal that allowed for medial/lateral translation. However, they did find that the pedal that allowed for inversion/eversion movements significantly reduced the varus/valgus knee moments, while the pedals that allowed for abduction/adduction movements significantly reduced both the internal/external and abduction/adduction knee moments.

Shoe Wedging

There are situations in healthy cycling in which the rider may have excessive knee abduction (i.e. they have a more medially placed knee position throughout the crank cycle). This has been associated with potential for development of overuse injury (Burke, 1986). To counter this misalignment, it is common for cyclists to use a wedge between their shoe and the pedal, or a medially wedged (posted) in-shoe orthotic. Only few scientific publications are available addressing the effectiveness of these types of devices. One such study by Sanderson et al. (1994)

filmed the frontal plane movements of 28 experienced cyclists using 16 mm film. The researchers placed markers on the tibial tuberosity of each leg and measured the distance from the marker to the frame of the bicycle. They studied the effect of a 10 degree varus and 10 degree valgus wedge placed between the shoe and pedal and compared this to a neutral foot alignment. They found that the distance between the marker and the frame in the neutral position ranged from 7.5 to 10.5 cm. For the valgus wedge the distance ranged from 7 to 12 cm, and for the varus wedge the distance ranged from 7.5 to 10 cm. There was not a statistical difference between the neutral condition and either of the wedge conditions, but there was a significant difference between the two wedge conditions.

A more recent study by Gregersen et al. (2006a) analyzed the frontal and transverse plane knee moments in 15 competitive cyclists without history of knee overuse injury. Additionally, they studied the muscle activation from the vastus lateralis (VL), vastus medialis (VM), and tensor fascia latae (TFL) muscles during the pedal cycle. In this study, the cyclists pedaled in 5 inversion/eversion angles (5 and 10 degrees of inversion and eversion, and a neutral position). The main overall findings showed that both the peak KAM and the average value of the KAM was significantly decreased from neutral when the foot was everted and significantly increased when the foot was inverted ($p < 0.0001$). More specifically, the 10 degree everted condition reduced the peak moment by 4.29 Nm (55% decrease from neutral), and the 10 degree inverted condition increased the peak moment by 3.69 Nm (47% increase from neutral). They also found that when the foot was everted, the peak VM activation increased relative to that of the VL and the TFL muscle activity was decreased. The authors concluded that everting the foot during cycling could reduce the potential for overuse knee injuries such as patellofemoral pain syndrome. This reduced injury potential was made possible by potentially reducing lateral patellar tracking by reducing the KAM, increasing the VM activation with respect to the VL, and by decreasing the TFL activation. The patterns found in this study may be a good implication for diseased populations such as in patients with medial knee osteoarthritis.

Foot Progression Angle

Very little literature exists on the effects of foot progression angles on cycling biomechanics. Similar studies by Ruby and Hull (1993) and Boyd et al. (1997) analyzed the three dimensional intersegmental knee loads in healthy subjects (11 subjects in Ruby and Hull, and 10 subjects in Boyd et al.). The subjects in the Ruby and Hull study pedaled with a pedal platform that allowed freedom of foot movements in medial/lateral translation, adduction/abduction, and in inversion/eversion rotation separately. The subjects in Boyd et al.'s study pedaled in similar conditions but also in a condition that allowed freedom of movement in adduction/abduction and inversion/eversion simultaneously. The pedals had the option of being fixed in a neutral position as well, which was used as the control condition. In the study by Ruby and Hull (1993) the freedom of rotational movement in the adduction/abduction directions significantly reduced the adduction and abduction knee moments when compared to a fixed cleat condition. This finding was not consistent in the study by Boyd et al. (1997) which found no difference in adduction/abduction moments compared to a fixed cleat condition. The dissimilarity between the two studies can possibly be explained by the inherent differences in pedaling mechanics between the subjects in the two studies. Boyd et al. (1997) noted that in the transverse plane, the knee moment exhibited a pattern where an internal axial moment occurred in the down stroke of the pedal cycle in 5 of the 10 subjects. The rest of the subjects exhibited no consistent pattern. In the Ruby and Hull study, only 3 of the 11 subjects exhibited an internal axial moment during the down stroke of the pedal cycle.

Alignment in Preliminary Research

A preliminary study in our laboratory showed that participants with knee OA do not cycle with the normal frontal plane knee kinematics seen by previous studies. Out of the 6 initial participants, 6 knees are continuously adducted throughout the crank cycle, 2 knees are continuously abducted (valgus), and 4 knees appear to have a normal range of motion that begin the crank cycle in knee adduction and end in abduction. With regard to the continuously adducted knees, the pattern seen is similar to that during gait in which patients with medial compartment OA walk with the knee in an adducted (varus) position. Some possible solutions to the malalignment seen during cycling could possibly be borrowed from those seen in gait, such as using lateral shoe wedges or increasing the toe-in foot progression angle during the cycling bouts. More research is needed to determine the optimal riding patterns for people with knee OA.

Cycling Summary

While research suggests that cycling may be beneficial for knee injuries, it is also important to note that knee injuries are the most common injuries in cycling (Asplund and St Pierre, 2004; Dettori and Norvell, 2006; Kennedy et al., 2007). Some studies have pursued the cycling benefits for injuries, but most of the research has focused on optimizing performance. Thus, more research is needed for optimizing cycling for chronic injuries such as osteoarthritis, and determining ways to reduce joint loading and pain while improving overall joint function.

While cycling can be a very taxing aerobic and muscular exercise, it appears that clinicians may be justified in their decision to prescribe it as exercise for patients with knee injuries or OA because cycling induces small pedal reaction forces which may result in small joint reaction forces at the knee. This is thought to be beneficial for individuals with osteoarthritis, since compressive loading has been shown to influence the progression of OA. This is almost entirely due to the fact that the majority of the rider's body weight is supported by the bicycle seat and not their legs as in gait (that is unless they stand to pedal). Additionally, due to the closed chain nature of the exercise, cycling also results in small frontal plane knee joint moments in healthy populations. However, it is not clear if similar moments would be found in persons with knee OA who may have large potential for poor knee alignment while cycling. If in fact people with knee OA have poor alignment while cycling, there is potential for increased loading on the individual compartments of the knee (depending on how the knee is aligned). Furthermore, cycling induces a high demand of the lower extremity musculature. While the GRF's due to body weight are reduced in cycling, the muscles surrounding the knee joint are highly active because they are the major contributors to propelling the bicycle forward. Thus, increased knee joint forces due to muscle contractions may occur.

Questions arise in the use of cycling exercise related to joint moments and muscular activity. Cycling has been shown to increase joint moments and in most cases resulted in increased muscle activation (especially around the knee). However, increased ranges of motion during cycling may impact joint pain or function in osteoarthritic populations. Is it better or worse to increase the joint range of motion during exercise for osteoarthritic populations? While previous research has shown that the knee OA population walks in a stiffer gait, a question that arises is 'should we focus on recuperating the lost range of motion or should we accommodate it?' In cycling, many manipulations can be made to the bicycle that can increase or decrease joint ranges of motion and an ideal position has not been identified for osteoarthritis. This is likely a much more complex issue than it seems and will more than likely result in seating configurations that are different for each person.

Proper frontal plane joint alignment in cycling appears to be an important factor in preventing injuries in healthy populations. Frontal plane alignment is also important during gait for both healthy and knee OA populations. Therefore, it can be hypothesized that proper frontal plane alignment would be important for OA populations during cycling. As of now, it is unclear if correcting a frontal plane malalignment would produce clinically meaningful results during cycling and thus is a suggested topic for future research. One possible option is to test the effect of wedged shoe insoles on the knee alignment during cycling in an attempt to decrease joint loading.

LITERATURE REVIEW SUMMARY

This chapter reviewed relevant previous literature on knee osteoarthritis and related variables associated with gait. Additionally, it discussed the comprehensive effects of cycling on the benefits and potential pitfalls associated with osteoarthritis. Much research has been done on knee OA, yet it still remains a global issue which currently is incurable. Despite the fact that knee OA does not have a cure, there are well known risk factors associated with the disease development and progression. While some of the factors associated with knee OA are out of the control of the individual (age, sex, genetics, and previous injury), it is worth noting that many other important factors are still modifiable such as obesity, nutrition, and muscle weakness. Thus, while OA is not completely understood, and to date there is no cure for OA, there is hope that progression and symptoms of the disease can be lessened through various methods including exercise.

Exercise in OA is somewhat of a "double edged sword." While exercise is beneficial for decreasing the rate of progression, large impacts during exercise may exacerbate the disease. Physicians commonly prescribe low impact exercises for people with OA including treadmill walking and cycling. These activities are beneficial in providing individuals with OA good modes of exercise while reducing the vertical compressive load to the knee joint. Cycling appears to be a good choice for knee OA, as it produces small amounts of compressive loads on the knee joints. However, as this review has suggested, the bicycle still has the potential to exacerbate knee problems for the rider. Thus, simply riding a bicycle will likely not be the best solution for people with knee OA. More often than not, knee OA patients will require a customized cycling intervention for their specific needs. Not only is it important to configure a bicycle specifically for the rider, it is also important to ensure that there is no malalignment of the lower extremity joints while riding. This is especially true of patients with OA since they typically have poor alignment associated with their disease. More research is needed on the effects of cycling for exercise in people with knee osteoarthritis.

CHAPTER III

METHODS

PARTICIPANTS

Healthy Subjects

Eleven healthy male and female participants (age: 50.0 ± 9.7 yrs., height: 1.75 ± 0.12 m, weight: 80.17 ± 23.13 kg, BMI: 25.9 ± 5.4 kg/m^2) between the ages of 35 and 65 volunteered for participation in this study (age, height, weight). The subjects were pain free in their lower extremities for at least 6 months prior to the study. They were not diagnosed with any type of lower extremity osteoarthritis, never had a joint replacement, and did not have arthroscopic surgery or intra-articular injection within three months prior to the study. Additionally, the subjects must not have a neurological disease, low back pain referred to the legs, women who were pregnant or nursing, or cardiovascular risk factors that would preclude them from participation in aerobic exercise. The participants must have had a BMI of no more than 35 kg/m^2, and they must have been able to walk and ride a stationary cycle without aid. Prior to testing, each subject read and signed the informed consent that was approved by the University of Tennessee Institutional Review Board.

OA Subjects

Thirteen participants (age: 56.8 ± 5.2 yrs., height: 1.80 ± 0.14 m, weight: 83.2 ± 22.3 kg, BMI: 26.6 ± 3.6 kg/m^2) with knee osteoarthritis between the ages of 35 and 65 volunteered for participation in this study (age, height, weight). Each participant with OA had medial compartment tibiofemoral osteoarthritis in either one or both of their knees. To be included in the study, the OA participants must have had at least a grade 1 on the Kellgren-Lawrence score which was verified with radiographs and a diagnosis by a rheumatologist. While the requirement for medial compartment OA was strictly enforced, the participants were not excluded from the study if they had additional OA in the lateral compartment of their knee(s). Additionally, they were not excluded if their tibiofemoral OA was accompanied by patellofemoral OA. In addition, subjects were excluded from the study if they had OA in the hip or ankle joints, had previously had a lower extremity joint replacement, had knee joint arthroscopic surgery or intra-articular injections within 3 months prior to testing, had systemic inflammatory arthritis such as rheumatoid or psoriatic arthritis, had lower back pain that referred to the lower limbs, had a BMI greater than or equal to 35 kg/m^2, women who were pregnant or nursing, or those who had cardiovascular disease or other risk factor which precluded participation in aerobic exercise. All OA subjects must have been experiencing pain the majority of the days of the week, for at least the previous 6 months. If subjects were taking any type of medication for their pain, they were asked to cease its use 2 days prior to the study.

Before taking the x rays, each subject read and signed the study informed consent which was approved by the University of Tennessee and Medical Center's Institutional Review Boards. For the X-rays, the subjects performed bilateral standing while anterior/posterior radiographs were taken of both knees in the frontal plane. Additionally, a sagittal plane radiograph of each knee was collected while the subject was in a bent knee stance.

INSTRUMENTATION

3D High-Speed Motion Capture: A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to acquire three-dimensional (3D) kinematics of the trunk, pelvis, and bilateral thighs, shanks, and feet of the subjects. The subjects wore tight fitting workout clothing such as that used for cycling (i.e. spandex). If the subjects did not own this type of clothing, spandex laboratory shorts were supplied. Reflective anatomical markers were used to identify segment joint centers and were placed bilaterally on the subject's 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, greater trochanter, left and right iliac crests, and left and right acromion processes. Non-collinear tracking markers were attached to rigid thermoplastic shells and then attached to the trunk, pelvis, thighs, and shanks using hook and loop wraps. For the feet, three markers were placed on the outer surface of the shoe at the superior, inferior, and lateral heel.

Cycle Ergometer: A Lode Excalibur Sport cycle ergometer (Lode, Groningen, Netherlands) was used for the cycle testing. The ergometer was electro-mechanically braked which allowed for a precise workload setting that was independent of the pedal cadence. Additionally, the ergometer had removable pedals, and had the capability of adjusting the seat and handlebar to fit each rider.

Customized Instrumented Pedals: A customized instrumented bike pedal was used on the Lode cycle ergometer, which allows recordings of three dimensional forces and moments (Figure 3). The assembly contained two 3D force sensors (Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A, Kistler, Switzerland). The coordinate system for the pedal is shown in Figure 4. The charge amplifiers were necessary to convert the charge measured by the force sensors to a voltage value used by the Nexus software. The sensors could be placed in either the left or right pedal depending on the desired limb to be analyzed. A dummy pedal of the same mass and design was used on the opposite limb to minimize asymmetries during the testing. The kinetics from the instrumented pedals and 3D kinematics were recorded through the Vicon Nexus system simultaneously. Prior to using the pedal assembly in the current research project, extensive calibration testing was done to ensure the pedal measurements were accurate.

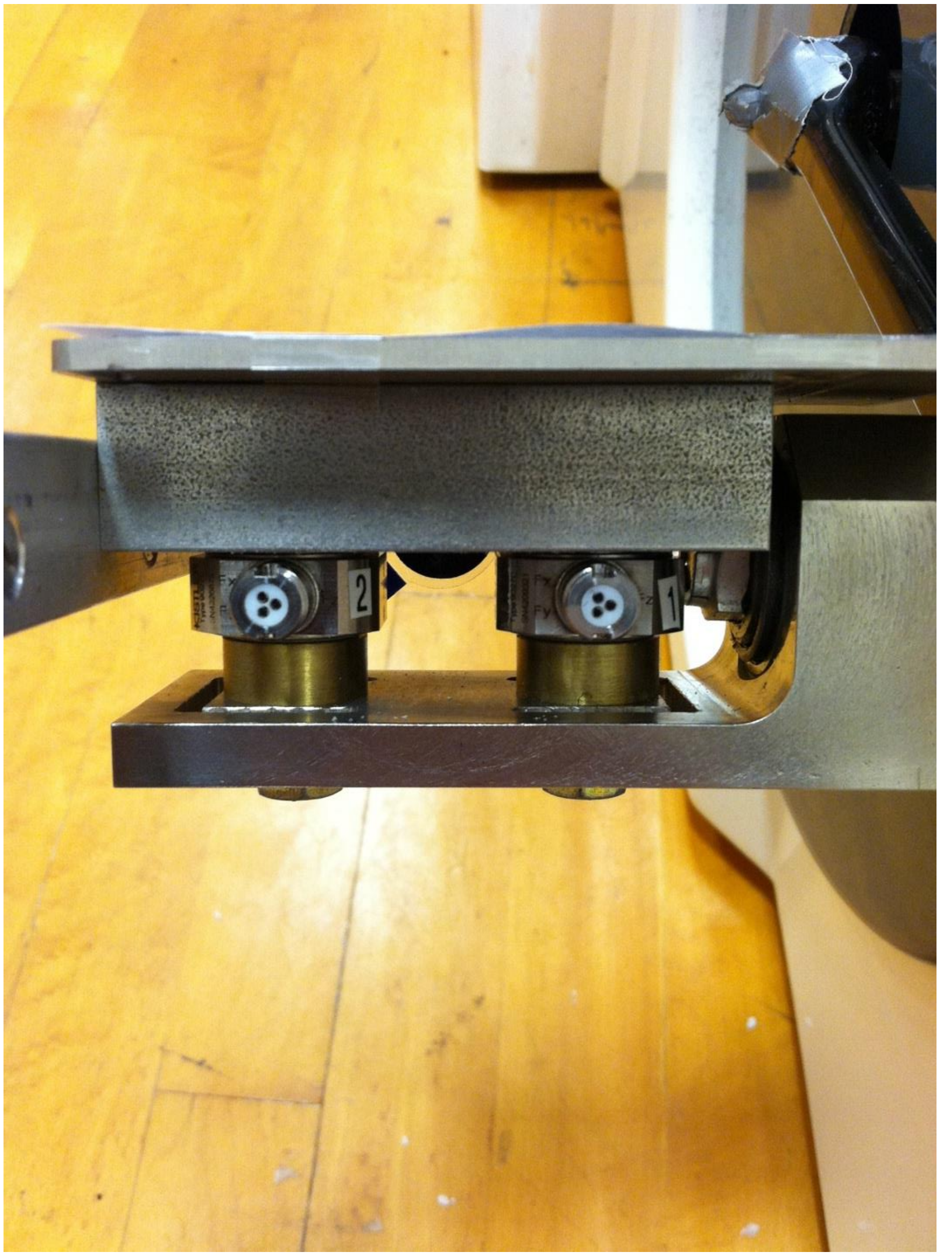

Figure 3: Photo of the instrumented pedal assembly

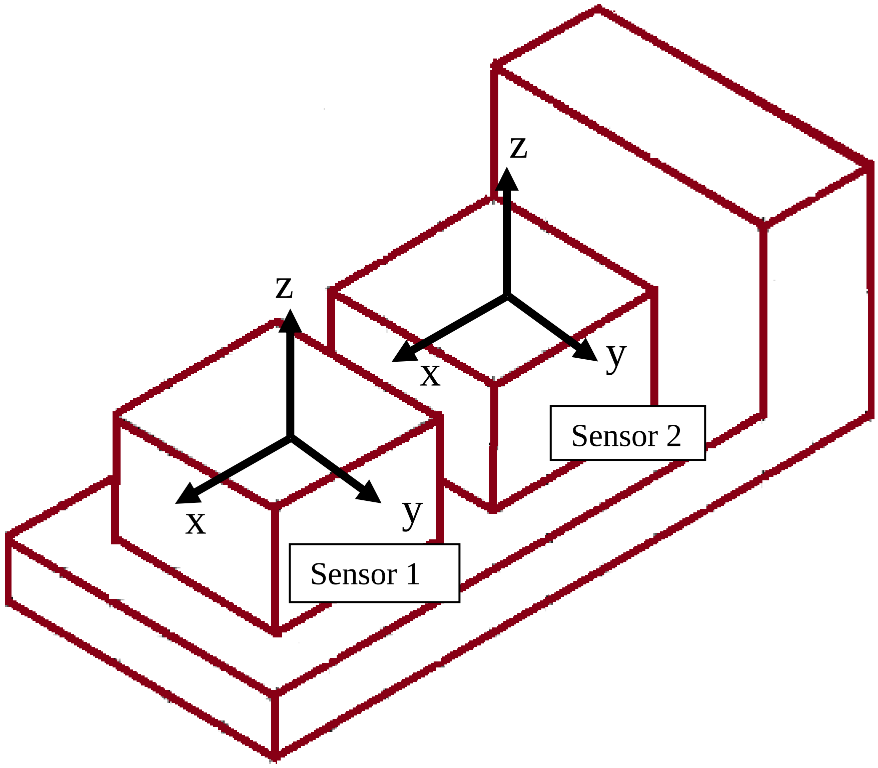

Figure 4: Coordinate system for the right pedal assembly. Note: the top plate has been removed to show the force sensors.

Visual 3D: Visual 3D (C-Motion Inc.), a 3D biomechanical analysis software suite was used for signal processing and computing 3D kinematics and kinetics.

Knee Pain and Function Assessment: The Knee Injury and Osteoarthritis Outcome Score (KOOS) was used to assess each subject's knee pain and function during the week prior to the testing session.

Visual Analog Scale for Pain: A 0 to 10 cm VAS numeric pain intensity scale was used to assess each participant's knee pain during the cycling protocol with 0 being no pain and 10 being worst pain possible. Subjects did could choose any real number between 0 and 10 (Figure 5).

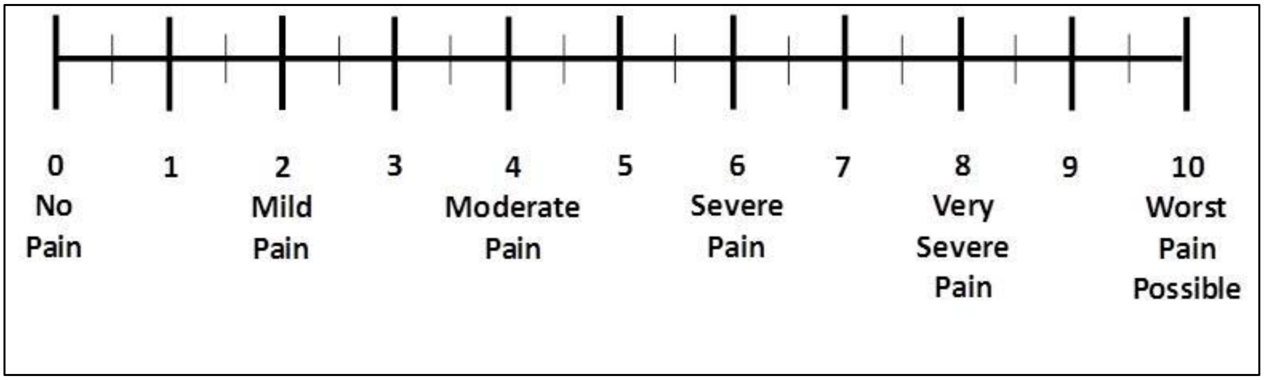

Figure 5: 0 to 10 cm numeric pain intensity scale

Customized Computer Programs: Customized computer programs (VB_V3D and VB_Table) were used to determine critical events of the 3D kinematic and kinetic variables of interest from the output of Visual3D, and were also used to compute additional parameters and organize the variables for statistical analyses.

EXPERIMENTAL PROTOCOL

Upon arrival to the biomechanics laboratory each subject filled out the KOOS survey for each of their knees (OA subjects only). Additionally, height and weight were recorded at this time. Subjects then did a walking warm up on a treadmill for 3 minutes to get a baseline pain measurement (using the VAS) in their knees. Reflective markers were then placed on the individual's body segments as described in the instrumentation section above. A static calibration trial was recorded and then the anatomical markers were removed. The subjects were asked to warm up on the cycle ergometer for 3 minutes.

The seat height on the cycle ergometer was set so that the angle of the subject's knee was at 30 degrees when the crank was at bottom dead center (BDC). This seat height was determined from preliminary work that revealed the least amount of knee pain within this range. This method has also been shown to minimize the risk of outside sources of knee pain such as patellofemoral pain syndrome (PFPS) and iliotibial band syndrome (ITBS) (Holmes et al., 1994). The horizontal seat depth was set so that the knee was in line with the pedal spindle when the crank was in the forward horizontal position (90°) (Burke, 2003). Each subjects' trunk angle was also controlled by placing the handlebars in a position that created a 90° angle between the trunk and the thigh.

The subjects pedaled in 5 cycling conditions for this study. There were two conditions in which a wedge was placed between the shoe and the pedal (5 and 10 degrees) on the lateral side of the foot (Figure 6). The wedge was simply a block of wood that was cut at the specified angle and then attached securely to the pedal. In the third and fourth conditions, subjects cycled with an increased (toe-in) foot progression angle. The foot progression angles were increased to 5 and 10 degrees relative to the antero-posterior axis of the pedal. The toe-in effect was created with a wedge that was placed between the anterior surface of the pedal body and the pedal toe-cage (Figure 7). This effectively increased the angle of the toe-cage which restricted the subject's foot to the desired toe-in angle. The fifth cycling condition was the control condition. The subjects pedaled in a neutral foot position which was established with a neutrally oriented pedal toe-cage.

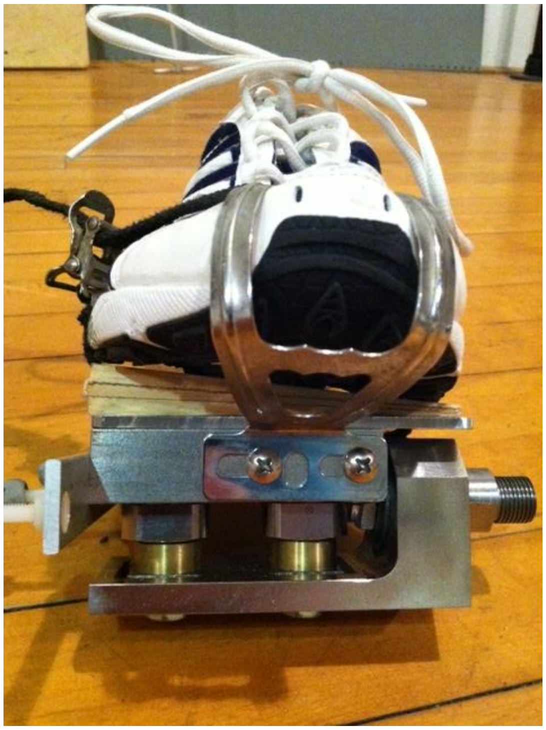

Figure 6: Photo of how the wedge condition was created on the pedal

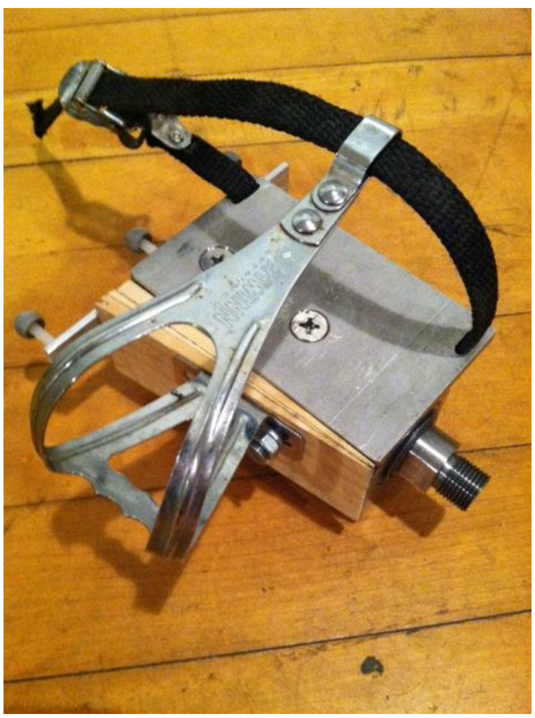

Figure 7: Photo of how the toe-in condition was created on the pedal

The subjects cycled for 2 minutes in each of the conditions at a pedal cadence of 60 RPM and a workload of 80 Watts. This workload was used to meet exercise recommendations by the American College of Sports Medicine (ACSM) (Garber et al., 2011) if the subject were to continue riding at this level for about 30 minutes. For example, subjects weighing between 70 and 100 kg who cycle at an 80 W workload would be working at an equivalent of about 4.5 to 5.5 metabolic equivalents (METS). The following equations were used to calculate the MET levels.

$$VO_2 = (10.8 \times Watts \div Body\ Mass) + 7$$

$$1\ MET = 3.5\ ml/kg/min$$

Where VO2 is the energy expenditure in ml/kg/min, Watts is the workload set on the cycle ergometer, Body Mass is in kilograms, and MET is metabolic equivalents. This criterion was chosen in an effort to approximate the level of exercise a person might engage in if they were cycling on their own accord. If the subjects continued to work at this level for 30 minutes, they would burn about 200 to 235 calories. Continuing this level of exercise for 5 days each week would result in a weekly energy expenditure of about 1015 to 1170 calories/week, which would meet current ACSM weekly energy expenditure recommendations (Garber et al., 2011).

The cycling conditions were randomized ahead of time to minimize any order and learning effect. Simultaneous recordings of kinematic (240 Hz) and kinetic (1200 HZ) data were performed on 5 consecutive pedaling cycles for each condition which began during the last 30 seconds of each trial. For the OA subjects, an enlarged numeric pain intensity scale was presented to the subjects during this time, and they rated the pain they felt, if any, in both of their knees. Subjects were given at least 2 minutes of rest between conditions.

DATA ANALYSIS AND STATISTICAL PROCEDURES

Data Analysis

The signals from the two pedal sensors in the x direction were summed to form a single Fx signal before entering the amplifier. The y and z signals were not combined across the two sensors before entering the amplifier so that the moments about the y (My) and z (Mz) axes could be calculated. Thus, the center of pressure (COP) was free to move in the medial lateral direction (x) along the pedal spindle, but was constrained by the pedal spindle. Forces, moments of force, and COP were calculated as follows:

Right Pedal

$FX = fx1 + fx2$	Medio-lateral force
$FY = fy1 + fy2$	Anterior-posterior force
$FZ = fz1 + fz2$	Vertical Force
$Mx' = FY * Az0$	Moment of X-axis about the top of the pedal
$My' = (a * fz1 - a * fz2) - FX * Az0$	Moment of Y-axis about the top of the pedal
$Mz' = (-a * Fy1 + a * Fy2)$	Moment of Z-axis about the top of the pedal
$Ax = -My'/FZ$	COP in the X-direction
$Ay = Mx'/FZ$	COP in the Y-direction

Left Pedal

$FX = (fx1 + fx2) * -1$	Medio-lateral force
$FY = fy1 + fy2$	Anterior-posterior force
$FZ = fz1 + fz2$	Vertical Force

$Mx' = FY * Az0$	Moment of X-axis about the top of the pedal
$My' = (-a * fz1 + a * fz2) - FX * Az0$	Moment of Y-axis about the top of the pedal
$Mz' = (a * Fy1 - a * Fy2)$	Moment of Z-axis about the top of the pedal
$Ax = -My'/FZ$	COP in the X-direction
$Ay = Mx'/FZ$	COP in the Y-direction

Where *fx1*, *fy1*, and *fz1* are the forces measured by sensor 1 in the x, y, and z direction respectively, *fx2*, *fy2*, and *fz2* are the forces measured by sensor 2 in the x, y, and z direction respectively, *a* is the distance between the two sensors, and *Az0* is the distance from the sensors to the top of the pedal.

Within Vicon Nexus, the 5 consecutive pedal cycles were separated to obtain 5 individual trials from top dead center (TDC, 0°) to TDC (360°). Visual 3D was used to obtain the 3D kinematic and kinetic computations for the lower extremity joints. A right hand rule was used to determine the polarity of the joint angles and moments and an X-Y-Z Cardan rotation sequence was used to compute joint angles. Marker and pedal reaction force data were each filtered using a zero lag, 4th order, digital Butterworth filter at 6 Hz (Gregersen et al., 2006a). Peak ankle, knee, and hip joint angle and moment data were identified and extracted using custom written programs (VB_V3D, VB_Table). For the kinematic and kinetic data, peaks were chosen at approximately 90° during the power phase of the crank cycle. This is the approximate time in the crank cycle when the rider is able to produce the most effort, which would have the greatest muscular impact on their joints. It should be noted that the moment variables were not normalized to any anthropometric feature (i.e. weight or height). In cycling we believe it is important not to normalize as the majority of the subject's weight is carried by the cycle ergometer seat and handlebars. Thus, by not normalizing, we are able to get a better understanding of the actual moment value across the knee joint.

Statistical Procedures

Independent samples t-tests were used to determine if KOOS scores for each subcategory were different between the two groups. A 2 x 3 (group x condition) mixed design analysis of variance (ANOVA) was used to detect differences between the cycling conditions and participant groups for pain and other selected variables (IBM SPSS Statistics 20, Chicago, IL). When an interaction was present, a pairwise t-test was performed in the post hoc analysis with Bonferroni adjustments to determine the location of the statistical differences. An alpha level of 0.05 was set a priori. Additionally, for study 2, a simple linear regression was performed for the OA patients to analyze the relationship between K/L score and the peak knee adduction angle and peak KAM. An alpha level of 0.05 was set a priori.

CHAPTER IV

THE EFFECTS OF LATERAL SHOE WEDGES ON JOINT BIOMECHANICS OF PATIENTS WITH MEDIAL COMPARTMENT KNEE OSTEOARTHRITIS DURING STATIONARY CYCLING

ABSTRACT

Cycling is commonly prescribed for individuals with knee osteoarthritis (OA) but practically no biomechanical literature exists on the topic. Individuals with OA may be at greater risk of OA progression or other knee injuries because of their altered knee kinematics. This study investigated the effects of lateral wedges on knee joint biomechanics and pain in patients with medial compartment knee OA. Thirteen OA subjects and 11 healthy subjects participated in this study. A motion analysis system was used to collect 5 pedal cycles of kinematics during 2 minutes of cycling in 1 neutral and 2 wedge (5° and 10°) conditions. Subjects pedaled at 60 RPM and 80 watts while a custom instrumented pedal was used to collect pedal reaction forces. Participants rated their knee pain on a visual analog scale each minute of each condition. There was a 22% decrease in the knee abduction moment with the 10° wedge. However, this finding was not accompanied by a decrease in knee adduction angle or subjective pain. Additionally, there was an increase in vertical and horizontal PRF which may negate the advantages of the decreased KAM. For medial knee OA patients, cycling with 10° lateral wedges may be a possible method to slow OA progression or minimize other knee injuries. More research is needed to investigate the joint contact forces as well as long term effects of lateral wedges.

Key Words: knee osteoarthritis, lateral wedges, knee moment, cycling, kinetics, kinematics

Running Title: Effects of wedges on knee OA during cycling

INTRODUCTION

Osteoarthritis (OA) can have an incapacitating effect on people affected. The disease is prevalent in nearly 27 million people in USA alone (Lawrence et al., 2008) and the knee joint is one of the most commonly affected joints (Lawrence et al., 2008). Exercise such as cycling is commonly prescribed by health professionals to reduce the body weight loads placed on the knees. However, knee injuries are still the leading complaint in cycling (Dettori and Norvell, 2006; Kennedy et al., 2007) and very little research has been done justify cycling for knee OA beyond the fact that the body weight load is reduced. It is unclear if people with knee OA have different cycling patterns than healthy populations. If in fact persons with knee OA cycle differently, abnormal kinematics and kinetics may lead to increased knee load and/or pain at the very least, and possibly further the development or progression of the disease. It is possible that corrective conservative measures can be borrowed from gait interventions to encourage normal riding patterns and promote exercise in knee OA populations.

During gait, the internal knee abduction moment (KAM) has been shown to be an important factor associated with knee OA (Andriacchi et al., 2000; Baliunas et al., 2002; Cerejo et al., 2002; Mundermann et al., 2005). The KAM is a surrogate measure for loading to the medial compartment of the knee which is created as a response to an external adduction moment resulting from the ground reaction force (Schipplein and Andriacchi, 1991). This moment acts to adduct the knee during stance into a bow-legged or knee varus position (Cerejo et al., 2002); a condition that opens the lateral joint space while closing the medial joint space of the knee, resulting in increased stress on the medial compartment. Several studies have found a relationship between the magnitude of the adduction moment and the severity of knee OA

(Cerejo et al., 2002; Mundermann et al., 2005; Sharma et al., 1998; Wada et al., 2001). Mundermann et al. (2005) found that people with more severe knee OA have a larger knee varus alignment (5.7°) than those with a less severe disease (0.3°). This malalignment in the knee has been shown to be associated with the progression of knee OA (Cerejo et al., 2002).

Possible mechanisms for reducing the KAM during walking, which have been verified by previous studies, is by placing a laterally posted orthotic in patients' shoes (Bennell et al., 2013; Butler et al., 2009; Butler et al., 2007; Hinman et al., 2009; Hinman et al., 2012; Kerrigan et al., 2002) or by using variable stiffness walking shoes (Erhart et al., 2008, 2010b). The application of a laterally posted orthotics or variable stiffness walking shoes are used to place the ankle into a more everted position which pulls the knee medially; effectively opening up the medial compartment joint space. It is logical to assume that this method for reducing the KAM may be transferred to cycling. Gregersen et al. (2006a) showed that increasing the ankle eversion angle in healthy, experienced cyclists, decreased the KAM by 55% and concluded that everting the foot may be beneficial towards preventing or decreasing patellofemoral pain syndrome in cycling. However, it is unknown if these modifications in OA patients during cycling would produce similar results.

Cycling reduces loading on the knee joint by placing the majority of the rider's body weight on their seat during seated cycling (Burke, 2003). However, cycling produces a great demand on the muscles of the lower limbs, especially the knee extensors, as they are the driving force in pushing the pedals to propel the bicycle forward. The increased muscle contraction in turn produces increased loading to the knee joint. Thus, knee injuries are still the leading complaint in cycling which has strong indication for an overuse injury mechanism (Dettori and

Norvell, 2006; Kennedy et al., 2007). For example, a common overuse injury during cycling, patellofemoral pain syndrome, is thought to occur because of abnormal non-driving knee moments (frontal and transverse planes) during the downward pedal stroke (Boyd et al., 1997; Wolchok et al., 1998). Thus, proper alignment of the lower limbs that aid in reducing the internal knee abduction moment during cycling is an important factor for reducing overuse injuries experienced by the rider (Bailey et al., 2003; Gregersen et al., 2006a; Ruby and Hull, 1993).

During cycling of healthy populations, frontal plane knee angles range from about 2 to 4 degrees of abduction to 1 to 6 degrees of adduction during the crank cycle (Bailey et al., 2003; Umberger and Martin, 2001). This small range of motion in the frontal plane indicates that the knee remains in a fairly neutral position throughout the crank cycle. However, results of a preliminary study in our laboratory showed that participants with medial knee OA do not cycle with the normal frontal plane knee kinematics that are typically seen in healthy participants of previous studies. Out of the 6 initial participants, 6 knees were continuously adducted throughout the crank cycle. The pattern seen is similar to that during gait in which patients with medial compartment OA walk with the knee in an adducted position (Cerejo et al., 2002). Bailey et al. (2003) found that riders with a history of overuse knee pain had increased knee abduction angles when compared to the healthy controls. As discussed earlier, malalignment of the knee during cycling is a concern because it may exacerbate an existing condition such as knee OA or cause other problems such as overuse injuries with long term riding.

To our knowledge, no studies have explored the effects of limb alignment alterations on the internal knee abduction moment and angle of knee OA patients during cycling. Changes in lower extremity alignment using lateral wedges could alter the frontal plane kinematics by placing the knee in a more medial position and decrease the length of the moment arm of the pedal reaction force to the knee joint center in the frontal-plane, thus, decreasing the KAM. Previous literature has also suggested that the sagittal plane (Walter et al., 2010) and transverse plane (Boyd et al., 1997; Ruby and Hull, 1993) knee moments may be important variables for discussing knee injuries. Therefore, kinematic and kinetic knee variables in all three planes of motion were analyzed in this study. Additionally, since we were directly manipulating the ankle joint by use of wedges, ankle kinematics in all three planes of motion were analyzed. Finally, PRF data were also analyzed in this study due to their direct influence on joint moments. Therefore, the primary purpose of this study was to investigate the effects of lateral shoe wedges on peak knee adduction angle and peak internal KAM in participants with medial compartment knee OA during stationary cycling. It was hypothesized that lateral shoe wedges would reduce the peak knee adduction angle and the peak KAM in participants with medial compartment knee OA during stationary cycling compared to a neutral control condition.

METHODS

Participants

Eleven healthy male and female participants (age: 50.0 ± 9.7 yrs., height: 1.75 ± 0.12 m, weight: 80.17 ± 23.13 kg, BMI: 25.9 ± 5.4 kg/m^2) and thirteen participants with knee OA (age: 56.8 ± 5.2 yrs., height: 1.80 ± 0.14 m, weight: 83.2 ± 22.3 kg, BMI: 26.6 ± 3.6 kg/m^2) between the ages of 35 and 65 volunteered for participation in this study. Each participant with OA had medial compartment tibiofemoral OA in either one or both of their knees. To be included in the study, the OA participants must have had at least a grade 1 on the Kellgren-Lawrence score (Kellgren and Lawrence, 1957) (Grade 1: N=5, Grade 2: N=3, Grade 3: N=3, Grade 4: N=2) which was diagnosed and verified with radiographs by a rheumatologist. The participants were not excluded from the study if they had additional patellofemoral OA or OA in the lateral compartment of their knee(s). In addition, participants were excluded from the study if they had OA in the hip or ankle joints, had previously had a lower extremity joint replacement, had knee joint arthroscopic surgery or intra-articular injections within 3 months prior to testing, had systemic inflammatory arthritis such as rheumatoid or psoriatic arthritis, had lower back pain that referred to the lower limbs. All OA subjects must have been experiencing pain the majority of the days of the week, for at least the previous 6 months. If subjects were taking any type of medication for their pain, they were asked to cease its use 2 days prior to the study. The healthy participants were pain free in their lower extremities for at least 6 months prior to the study and were not diagnosed with any type of lower extremity OA. All participants must have had a BMI of no more than 35 kg/m^2, and must have been able to walk and ride a stationary bike without aid. Each participant read and signed the informed consent that was approved by the Institutional Review Board.

For the X-rays, the OA participants performed bilateral standing in a semi flexed knee while anterior/posterior radiographs were taken of both knees in the frontal plane (Buckland-Wright et al., 2004). Additionally, a sagittal plane radiograph of each knee was collected while the participant stood in a semi flexed knee to determine the presence of patellofemoral OA.

Instrumentation

A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to acquire three-dimensional (3D) kinematics during the cycling test. The participants wore tight fitting spandex shorts and a t-shirt. To identify joint centers, anatomical markers were placed bilaterally on the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, left and right greater trochanters, left and right iliac crests, and left and right acromion processes. Four non-collinear tracking markers affixed to a semi-rigid thermoplastic shell was attached to the trunk, pelvis, thighs, and shanks using hook and loop wraps. For the feet, three markers were placed on the posterior and lateral side of heel counter of standard lab shoes (Noveto, Addidas).

A cycle ergometer (Excalibur Sport, Lode, Groningen, Netherlands) was used during testing. The ergometer was electro-mechanically braked which allowed for a precise workload setting that was independent of the pedal cadence. Additionally, the ergometer had removable pedals, and had the capability of adjusting the seat and handlebar to fit each rider.

A customized instrumented bike pedal was used on the Lode cycle ergometer, which allowed recordings of three dimensional forces and moments (Figure 1). The assembly contained two 3D force sensors (Type 9027C, Kistler, Switzerland) connected with two charge amplifiers (Type 5073A and 5072A, Kistler, Switzerland). The sensors could be placed in either the left or right pedal depending on the desired limb to be analyzed. A dummy pedal with the same mass and design was used on the opposite side. The pedal reaction forces and 3D kinematics were recorded through the Vicon Nexus system simultaneously.

Experimental Protocol

Upon arrival to the laboratory each participant filled out a KOOS (Knee Osteoarthritis Outcome Score) survey for each of their knees to assess knee pain and function during the week prior to the testing session. Participants then performed 3 minutes of treadmill walking at a self-selected pace which served as a warm-up and as a way to get a baseline VAS pain in their knees (one measurement before and one after the warm-up). Reflective markers were then placed on the individual's body segments for testing.

The seat height on the cycle ergometer was set so that the angle of the participant's knee was 30 degrees when the crank was at bottom dead center (Holmes et al., 1994). The horizontal seat depth was set so that the knee was in line with the pedal spindle when the crank was in the forward horizontal position (90°) (Burke, 2003). Each participant's trunk angle was also controlled by placing the handlebars in a position that created a 90° angle between the trunk and the thigh when the crank angle was at 90°. The participants were asked to warm up on the cycle ergometer for 3 minutes where knee pain levels were again recorded, one before and one after the warm-up.

The participants pedaled in 3 cycling conditions. The two wedge conditions included 5 and 10 degree wedges placed between the shoe and the pedal on the lateral side of the foot. The third was the control condition in which the participants pedaled with a neutral foot position established with a neutrally oriented pedal toe-cage. The testing conditions were randomized.

The cycling was performed for 2 minutes in each of the three conditions at a pedal cadence of 60 RPM and a workload of 80 Watts. Data were collected on 5 consecutive pedaling cycles from top dead center (TDC, 0°) to TDC (360°) for each condition, which began during the last 30 seconds of each trial. For the OA participants, an enlarged 0 to 10 numeric pain intensity scale was presented to the participants during this time, and they rated the pain in both of their knees (0 being no pain and 10 being worst pain possible). Participants could choose any real number between 0 and 10. Pain measurements for each knee were recorded at minutes 0, 1, and 2 during the cycling. Participants were given at least 2 minutes of rest between conditions.

Data and Statistical Analyses

Visual 3D (C-Motion Inc.), a 3D biomechanical analysis software suite, was used for signal processing and to obtain the 3D kinematic and kinetic computations for the lower extremity joints. A right hand rule was used to determine the polarity of the joint angles and moments and an X-Y-Z Cardan rotation sequence was used to compute joint angles. Both marker and pedal reaction force data were filtered using a zero lag, 4^{th} order, digital Butterworth filter at 6 Hz. Customized computer programs (VB_V3D and VB_Table) were used to determine critical events of the 3D kinematic and kinetic variables of interest from the output of Visual3D. For the kinematic and kinetic data, peaks were chosen at approximately 90° during the power phase of the crank cycle. This is the approximate time in the crank cycle when the rider is able to produce the most effort, which would have the greatest muscular impact on their joints. It should be noted that the moment variables were not normalized to any anthropometric feature (i.e. weight or height). In cycling we believe it is important not to normalize as the majority of the subject's weight is carried by the cycle ergometer seat and handlebars. Thus, by not normalizing, we are able to get a better understanding of the actual moment value across the knee joint.

Independent samples t-tests were used to determine if KOOS scores for each subcategory were different between the two groups. A 2 x 3 (group x condition) mixed design analysis of variance (ANOVA) was used to detect differences between the cycling conditions and participant groups for pain and other selected variables (IBM SPSS Statistics 20, Chicago, IL). When an interaction was present, a pairwise t-test was performed in the post hoc analysis with

Bonferroni adjustments to determine the location of the statistical differences. An alpha level of 0.05 was set a priori.

RESULTS

KOOS and VAS Knee Pain

All subscales of KOOS were (pain, symptoms, activities of daily living, sports, and quality of life) lower in the OA group when compared to the healthy group (all p-values <0.001, Table 1). During treadmill walking, the VAS pain scores were 0.00 ± 0.00 for the healthy group and 1.19 ± 1.48 for the OA group. The VAS pain scores during cycling were 0.03 ± 0.08 cm, 0.00 ± 0.00 cm, and 0.00 ± 0.00 cm for the healthy group and 1.15 ± 1.10 cm, 1.05 ± 0.89 cm, and 1.12 ± 0.79 cm for the OA group for the neutral, 5° wedge, and 10° wedge conditions respectively. The ANOVA revealed no interaction ($p=0.743$) or condition ($p=0.425$) effects. There was a group main effect found with the OA group experiencing more pain than the healthy group ($p<0.001$).

Pedal Reaction Forces

None of the PRF variables revealed a significant group or interaction effect. The peak vertical PRF was significantly greater in both 5° wedge ($p=0.006$) and 10° wedge ($p=0.039$) conditions compared to neutral (Table 2). Additionally, the peak medial PRF for the 10° wedge approached significance with the ANOVA revealing a statistically significant condition main effect ($p=0.043$) but the post hoc test revealing a borderline, but insignificant result ($p=0.050$). For this particular variable, a post hoc analysis without a Bonferroni adjustment was also performed (using LSD Procedure) and the results revealed that the peak medial PRF in the 10° wedge condition was significantly greater than the neutral condition ($p=0.017$).

Ankle Joint Angles

None of the ankle joint variables revealed a significant group or interaction effect. The post hoc comparisons confirmed that the peak eversion angle was increased in the 5° wedge ($p<0.001$) and the 10° wedge ($p<0.001$) when compared to the neutral condition (Figure 8). The eversion angle was also significantly different between the two wedged conditions ($p=0.002$). Additionally, the peak internal rotation angle was significantly decreased in both the 5° wedge ($p=0.005$) and 10° wedge ($p<0.001$) conditions when compared to neutral (Table 3).

Knee Joint Angles

Figure 9 shows representative knee adduction angles across conditions for one subject. None of the knee angle variables revealed a significant group or interaction effect. The peak knee flexion angle was significantly greater in the 5° ($p<0.001$) and 10° ($p<0.001$) wedge conditions compared to neutral (Table 4). There was also a significant difference found in the peak flexion angle between the two wedge conditions ($p<0.001$).

Knee Joint Moments

Figure 10 shows representative knee abduction moments across conditions for one subject. None of the knee moment variables revealed a significant group or interaction effect. The peak abduction moment was significantly decreased in the 10° wedge condition compared to neutral ($p=0.033$, Table 5).

DISCUSSION

The purpose of this study was to examine the effects of lateral wedges on knee joint biomechanics and pain in patients with medial compartment knee OA during stationary cycling. The primary hypothesis of this study was that increasing the eversion ankle angle by way of wedges would decrease the knee adduction angle, internal knee abduction moment, and knee joint pain during cycling. Our hypothesis was partially supported with the 10° wedge condition resulting in a 22% reduction of peak KAM for the OA subjects compared to neutral. However, this finding was not accompanied by a reduction in knee adduction angle or knee pain when compared to neutral.

The pain values in this study were on the low end of the VAS pain scale which may be a good indicator that cycling is an effective mode of exercise to help reduce pain. It may also mean that the OA subjects used in this study did not have a significant amount of pain to begin with. There was very little difference in pain in the OA subjects during cycling compared to treadmill walking. In all conditions, pain was lower in cycling compared to walking. However, these differences were very small (largest difference of 0.14 cm in the 5° wedge), so it is unclear if the cycling was actually effective in reducing pain compared to walking, or if these particular subjects simply did not experience much pain on the day of testing. Five of the thirteen OA subjects had K/L grades of 1 which is considered mild OA. This is further supported by the KOOS scores which out of the 5 subscales the OA subjects scored highest on Activities of Daily Living, and Pain. This suggests that the OA subjects cope well with their disease and were less affected compared to the other KOOS subscales. However, our subjects scored similar (in some cases worse) than the comparison group of the OA data for which the KOOS was partially developed. Interestingly, the participants in that study were required to have a KL score ≥ 2. Nevertheless, it is possible that OA patients with more severe knee pain would show greater reductions in pain when the wedges were introduced during cycling.

For the ankle, both the 5° and 10° wedges were effective in increasing the eversion angle, however, not to the extent of the actual slope of the wedge. On average across both groups, the 5° wedge increased the ankle eversion angle by about 3.2°, while the 10° wedge increased the angle by 5°. There are several reasons this could be the case, including shoe sole flexibility and movement of the foot inside the shoe. However, the most obvious reason is that when the foot was everted, the shank angle did not remain vertical, but leaned more medially. This of course would mean the knees were pulled closer to a more desirable, neutral position. Nonetheless, the interventions did not induce a statistically significant reduction in peak knee adduction angle as we expected.

For the knee, we did see a marginal main effect ($p = 0.054$) and a decreasing trend in the adduction angle across wedge angles. But it is clear from Table 4 that between subject variability was very high. Even though there was a lot of variability in the knee adduction angle, the largest mean difference from the neutral condition in the OA group was in the 5° wedge and was only 0.6°. Though the main effect was nearly significant, this small difference in adduction angle is more than likely not a clinically meaningful result. It is unclear why the wedges did not decrease the peak adduction angle as hypothesized. It is possible that some subjects did not like how the wedges felt, so they pulled their knees back laterally in an attempt to emulate their typical riding style (i.e. more like the neutral condition). However, this is purely speculation and we did not collect subject perception data to substantiate this theory. Furthermore, the standard deviations for the neutral condition, not just the wedge conditions, were also very high. Additionally, the sample size of this study was relatively small considering the high variability of OA studies. The more obvious explanation is that since the knee adduction angle is less restricted in cycling than other variables (such as hip and ankle angles), there is more room left for greater inter-subject variability.

An interesting observation we made in collecting preliminary data was that there was a clear difference in knee adduction angle in OA subjects when compared to healthy subjects of previous cycling literature. Specifically, healthy subjects of previous studies exhibited frontal plane knee angles that hovered around zero (Bailey et al., 2003; Umberger and Martin, 2001). For example, Bailey et al. (2003) showed frontal plane knee angles in healthy subjects ranging from -2.5° abduction to 1° adduction. However, our preliminary data showed that most OA subjects' frontal plane knee angles remained adducted for the entire crank cycle. This suggests that there is a clear malalignment issue in OA subjects during cycling. This malalignment may contribute to the progression of knee OA or may even contribute to overuse knee injuries which are already a big concern in cycling. Qualitatively, the OA group in our study did show a larger adduction angle during cycling when compared to the healthy group (Table 4). Quantitatively, there was about a 41% difference between the two groups (2.4°), however, a statistical difference was not found because of the high variability. Surprisingly, the healthy group did not have a neutral frontal plane knee angle (12.5° to 1.7° adduction) as seen in previous studies. It is possible that the difference between our study and previous studies is due to several factors. First, the kinematic collection procedure was different between our study and Bailey's. For the frontal plane, Bailey et al. used a 2 dimensional technique by placing two markers on the anterior shank and measuring the angle between these markers and the right horizontal. Secondly, there was a clear difference in the cycling experience of the subjects. Bailey's subjects had an average of 7.6 years of regular cycling experience, while our subjects had no prior cycling requirements. Lastly, it is possible that the age of the subjects could play a role since the average age of subjects in Bailey's study was 28 years. While clear methodological differences exist between the studies, it is unknown if these differences contribute to the study findings.

For knee osteoarthritis patients, the KAM has been adopted as the surrogate measure for medial compartment knee joint loading (Schipplein and Andriacchi, 1991). In gait studies, importance has been placed on reducing the KAM with the overarching goal of lessening the severity and/or progression of OA. As mentioned previously, our hypothesis on this primary variable was partially supported by the results that the 10° wedge condition introduced a significant reduction of the peak KAM by 1.73 Nm (22.4%) for the OA group and by 0.87 Nm (9.7%) for the healthy group. This result certainly appears promising for reducing medial compartment knee joint loading during cycling. However, the question remains how these loads relate to those seen during daily walking. Butler et al. (2007) showed a 10° lateral wedge reduced the KAM by 10% compared to neutral. Hinman et al. (2012) showed that a 5° lateral wedge reduced the KAM by 5.8%. A more recent study by Bennell et al. (2013) showed that by wearing a modified shoe with a 4 - 6° lateral wedge insole, OA patients reduced their KAM by 7.2%. Our OA subject cycling results with the 10° wedge appear to reduce the KAM by a greater percentage (22%) than during walking. We did see a 28% reduction in the KAM for the OA group in the 5° wedge condition, however, the result was not significant. It is difficult to conjecture the implications of the larger percent decrease in KAM for OA subjects in cycling compared to walking. The magnitude of the KAM and the loading to the knee joint is greater during gait. During walking, however, the KAM is typically normalized (usually by body weight and height). The normalized KAM for OA subjects during normal over-ground walking is about 3.8 % Body Weight * Height (Baliunas et al., 2002; Bennell et al., 2013; Hinman et al., 2012). This is equivalent to 17.2 Nm un-normalized for a 1.8 m tall adult weighing 83.2 kg; the averages of the OA subjects for our study. With a peak 7.7 Nm KAM in the neutral cycling condition in the current study, OA subjects can expect about 2 times less peak absolute KAM than when compared to walking, and even more when cycling with wedges.. So, while there is clearly a greater percent reduction of KAM in cycling compared to gait when using wedges, it is not completely clear how clinically meaningful this percent change is when the absolute KAM is initially much less than gait. It is clear from a multitude of studies that much emphasis is placed on reducing the KAM for individuals with OA. Based on the results of this study, clinicians are justified in prescribing cycling for OA as it is clear that cycling reduces the KAM by as much as half during normal cycling. For those requiring an even greater reduction in KAM, a 10° wedge may be a good option. Prospective studies would be beneficial to help determine the clinical importance. We believe it is important not to normalize the moments for cycling because the majority of the body weight is supported by the bicycle seat and not by the legs during seated cycling. Thus, the KAM experienced during cycling is due to the muscular effort required to push the pedals, rather than due to the loading of body weight during in gait.

We could only find one published study that tested the effects of wedges during cycling (Gregersen et al., 2006a). This group studied the effects of 5 and 10 degree wedges in 15 competitive cyclists between the ages of 18 and 30 years. The KAM reached by the OA subjects in our study are nearly identical to the subjects in Gregersen's study in the neutral condition (difference of 0.12 Nm) but their percentage change in the 10° wedge was much greater (55%). It is not completely clear why our subjects did not realize a similar size of change in the KAM, but it is likely that the difference is a result of the age, cycling experience level, riding style, and injury status of the subjects. Gregersen et al. (2006a) concluded that everting the foot may be beneficial in either preventing or lessening patellofemoral pain syndrome. This conclusion may also hold true for the knee OA patients in our study who experienced similar, albeit less drastic changes. An interesting result of this study is that even though the KAM was decreased in the 10° wedge, there was an increase in the peak medial and vertical PRF compared to neutral (Table 2). Since the KAM is a result of a combination of the frontal plane PRF and the frontal plane moment arm from the PRF to the knee joint center, the reduction in the KAM must have come from a relatively greater reduction of the moment arm. Additional work to calculate the length of the PRF moment arm would be needed to verify this deduction.

One concern with the wedges is that it is possible that attempting to reduce the KAM with wedges may inadvertently increase the knee extensor moment. Previous gait literature suggests that even if a reduction is seen in the KAM, an increase in the knee extensor moment may negate any beneficial effects (Walter et al., 2010). While we did not find a difference in the peak knee extensor moment among conditions, we did find an increasing trend that approached significance ($p=0.056$, Table 5). Additionally, there was a significant increase in the vertical PRF which may indicate that the subjects had to push harder when using the wedges in order to keep the pedals moving at a consistent 60 RPM. Considering these two variables together, it appears that using the wedges may require the subjects to put forth greater muscular effort. This may in fact result in an increase in compression across the knee joint. Future examination on knee joint load and muscle forces using a musculoskeletal modeling approach is warranted to verify this finding.

One unexpected finding from this study was that the peak knee flexion angle increased with the wedged conditions compared to neutral. This was initially surprising to us because the manipulation in the testing conditions was to increase the ankle eversion angle. However, further consideration suggested that the knee flexion angle would be increased at the bottom of the crank cycle because the wedges essentially lifted the foot slightly off of the pedal. This would be similar to riding with a shortened crank arm without adjusting the seat height to compensate. We did not change the height of the seat with different wedge conditions, thus, it is unclear if a small seat height adjustment would produce different results in either of the groups. We suggest that in future studies, it may be necessary to adjust the seat height to accommodate any changes to the pedal height.

It is worth noting that while there was decrease in the KAM with the 10° wedge condition, we did not see a decrease in KAM in the 5° wedge condition compared to neutral. This presents the question about how many degrees of wedging are sufficient to produce the desired result. The current cycling market has in-shoe wedges as well as wedges for cycling cleats that are available for correcting excessive knee abduction (these wedges would be placed on the medial side of the shoe or cleat). It is interesting to note that these wedges come in very small degree increments (typically 1.5°). It is unclear if the individuals requiring these types of wedges only need a very small correction, or if it takes a less severely angled wedge to correct an abducted knee angle compared to an adducted knee angle. It is possible that we did not see the desired changes because of the stiffness of our laboratory shoes. Those who cycle with cleats typically have very stiff soled cycling shoes which may increase the chances of achieving the desired result. It is unclear if the changes we made to our pedals would have a different result on the knee joint biomechanics if a stiff soled shoe was used instead of our lab shoes.

The observed changes in this study appear to be minimal, however, it is important to note that this study only reported on the acute effects of the use of wedges during cycling as the cycling was only performed for 2 minutes. It is unknown if the acute effects would diminish over a longer period of time or if they would persist. Due to the repetitive nature of cycling, and considering that knee injuries are repetitive overuse injuries, we suggest that these small changes may compound and have a large impact over time. This is potentially beneficial for the OA sufferer who finds himself struggling with getting adequate exercise while minimizing harmful effects to his knees. The workload (80 Watts) and RPM (60 RPM) were fairly mild for the majority of our subjects. This workload was used to meet exercise recommendations by the American College of Sports Medicine (Garber et al., 2011) for exercising at this level for 30 minutes per day for 5 days each week. For example, subjects weighing between 70 and 100 kg who cycle at an 80 W workload would be working at an equivalent of about 4.5 to 5.5 metabolic equivalents (METs). Many people tend to exercise above this MET level, thus, different results in KAM and other critical variables may exist with differing workloads. However, a study by D'Lima et al. (2008) showed that the peak tibial contact force in an instrumented knee replacement of 3 older adults did not change with an increase in cadence from 60 to 90 RPM. They also showed that the peak tibial contact force was not different when resistance levels were increased from level 2 to 3 on their cycle ergometer. While it is not clear how resistance levels of

2 and 3 compare to wattage values in our study, it appears at least in older, knee replacement individuals, that compressive forces across the knee may not increase with small increases in workload. Additionally, we included OA subjects with a K/L score of 1 which is technically considered a diseased joint, but generally a K/L of 1 represents a mild stage of OA. While these participants still met our criteria for experiencing pain on a regular basis, it is unknown if our results would have been different if only patients with a minimum K/L of 2 were included.

CONCLUSION

The findings of this study indicate that cycling can reduce the KAM by as much as half when compared to previous walking studies. The use of a 10° lateral wedge during seated cycling was effective in reducing the KAM in healthy and knee OA subjects when compared to a neutral condition. However, this finding was not accompanied by a decrease in knee adduction angle, or subjective knee pain. Furthermore, even though a decrease in KAM was observed, there was a notable increase in the vertical PRF which may be an indicator of increased knee joint loading. It is important to remember that these results were due to an acute bout of cycling, and it is possible that the relatively small findings may add up and prove to be significant over time. This is the first study to report changes in knee joint biomechanical variables with the use of wedges in knee OA patients during stationary cycling. Further work is warranted to examine the knee joint loading and muscle forces using musculoskeletal modeling, as well as the long term effects of using wedges during cycling.

CHAPTER IV APPENDIX: TABLES AND FIGURES

Tables

Table 1: KOOS subscale normalized scores

Subscale	Group	Mean	SD	p value
Pain	Healthy	95.73	8.27	<0.001
	OA	70.38	11.93	
Symptom	Healthy	93.09	10.00	<0.001
	OA	56.85	11.86	
ADL	Healthy	96.55	7.16	<0.001
	OA	72.92	15.71	
Sports	Healthy	91.36	15.51	<0.001
	OA	53.69	24.03	
QOL	Healthy	91.00	16.58	<0.001
	OA	51.62	13.06	

ADL: Activities of daily living; QOL: Quality of life

Table 2: Peak PRF during the downward phase of the crank cycle (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Wedge	10° Wedge	Neutral	5° Wedge	10° Wedge	Grp	Cond	Int
Peak Medial PRF (N)	-30.70±11.83	-34.98±10.99	-35.64±9.40	-28.59±8.87	-27.42±14.16	-30.95±8.57	0.242	**0.043**	0.155
Peak Posterior PRF (N)	-64.69±19.98	-74.11±24.18	-68.54±25.82	-80.46±17.67	-78.59±24.71	-75.10±28.49	0.333	0.306	0.088
Peak Vertical PRF (N)	236.35±46.60	251.33±46.30[#]	250.21±51.44[#]	234.49±34.24	244.96±30.64[#]	244.17±37.00[#]	0.772	**0.007**	0.82

[#]: Significantly different than neutral; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 3: Peak Ankle Joint Angles (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Wedge	10° Wedge	Neutral	5° Wedge	10° Wedge	Grp	Cond	Int
Peak Plantarflexion Angle (°)	-8.9±10.7	-7.8±11.6	-8.0±11.3	-6.0±8.5	-6.8±9.8	-8.1±12.8	0.767	0.806	0.595
Peak Internal Rotation Angle (°)	9.2±7.6	7.1±8.5#	5.8±8.0#	8.1±7.1	6.8±6.9#	5.9±7.7#	0.892	**<0.001**	0.431

#: Significantly different than neutral; &: Significantly different than 5° wedge. Plantarflexion and external rotation are negative; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 4: Peak knee joint angles during the downward phase of the crank cycle (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Wedge	10° Wedge	Neutral	5° Wedge	10° Wedge	Grp	Cond	Int
Peak Flexion Angle (°)	-44.9±7.8	-47.2±7.1#	-48.8±7.7#&	-39.8±8.1	-45.0±7.3#	-48.1±7.5#&	0.383	**<0.001**	0.095
1st Peak Adduction Angle (°)	2.2±5.3	1.6±5.8	1.2±5.5	4.4±5.6	3.8±6.1	4.1±7.1	0.32	0.054	0.707
Peak External Rotation Angle (°)	-5.0±5.2	-4.3±4.3	-4.6±5.0	-2.9±5.4	-2.9±4.8	-3.1±5.0	0.41	0.469	0.528

#: Significantly different than neutral; &: Significantly different than 5° wedge. Flexion, abduction, and external rotation are negative; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 5: Peak knee joint moments during the downward phase of the crank cycle (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Wedge	10° Wedge	Neutral	5° Wedge	10° Wedge	Grp	Cond	Int
Peak Extensor Moment (Nm)	26.27±9.60	30.35±9.97	29.07±8.94	27.97±7.42	28.88±10.30	28.46±10.62	0.972	0.056	0.271
Peak Abduction Moment (Nm)	-9.00±4.74	-8.98±5.31	-8.13±4.19#	-7.72±4.76	-5.53±3.34	-5.99±3.66#	0.191	**0.034**	0.131
Peak Internal Rotation Moment (Nm)	7.98±4.29	8.08±4.22	8.47±3.46	6.58±3.32	5.64±4.05	6.02±3.97	0.169	0.705	0.446

#: Significantly different than neutral. Extensor, abduction, and external rotation moments are negative; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Figures

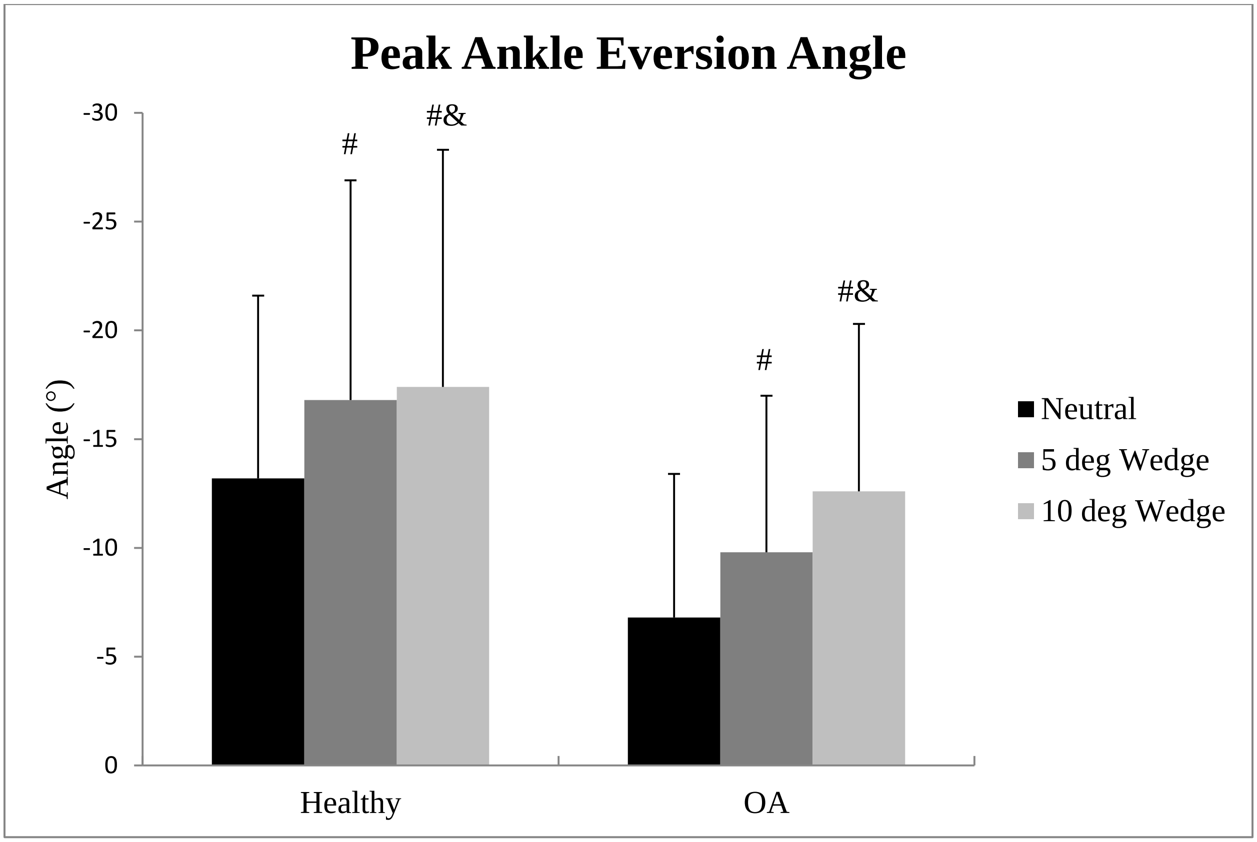

Figure 8: Peak ankle eversion angles. #: Significantly different than neutral; &: Significantly different than 5° Wedge.

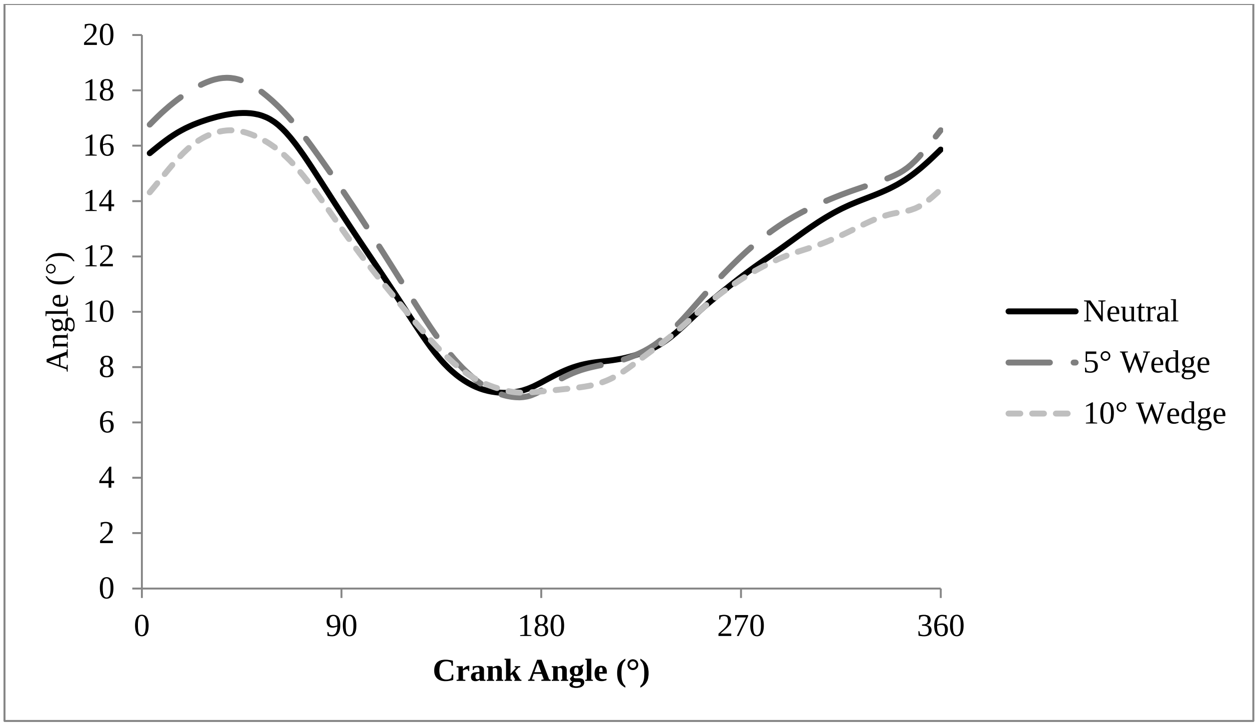

Figure 9: Representative knee adduction angle from one subject.

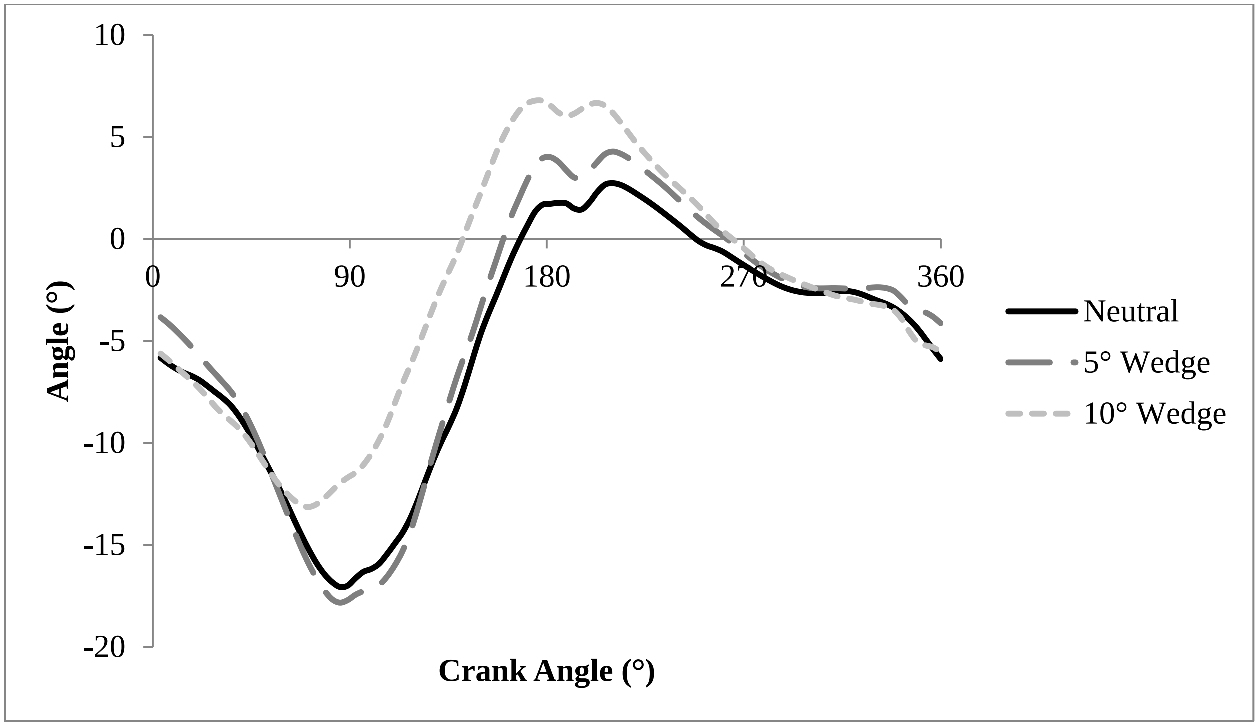

Figure 10: Representative knee abduction moment from one subject.

CHAPTER V

THE EFFECTS OF TOE-IN FOOT PROGRESSION ANGLES ON JOINT BIOMECHANICS OF PATIENTS WITH MEDIAL COMPARTMENT KNEE OSTEOARTHRITIS DURING STATIONARY CYCLING

ABSTRACT

Exercise is important for individuals with knee osteoarthritis (OA) but certain activities can be painful and discourage participation. Cycling is commonly prescribed for knee OA but practically no previous literature exists on the topic. Due to their altered knee kinematics, OA patients may be at greater risk of OA progression or other knee injuries during cycling. This study investigated the effects of reduced foot progression angles (i.e. toe-in) on knee joint biomechanics and pain in subjects with medial compartment knee OA. Thirteen OA subjects and 11 healthy subjects participated in this study. A motion analysis system and custom instrumented pedal was used to collect 5 pedal cycles of kinematics and kinetics during 2 minutes of cycling in 1 neutral and 2 toe-in conditions (5° and 10°). Subjects pedaled at 60 RPM and 80 watts and rated their knee pain on a visual analog scale for each condition. For the OA subjects, there was a 61% (2.7°) and a 73% (3.2°) decrease in peak knee adduction angle compared to neutral. This finding was not accompanied by a decrease in pain or peak knee abduction moment (KAM) because of high inter-subject variability. A simple linear regression showed a positive correlation between Kelgren-Lawrence (K/L) score and both peak knee adduction angle and KAM. For medial knee OA patients, cycling with a decreased foot progression angle may be beneficial in slowing the progression of OA or minimizing other knee injuries. Patients with a higher K/L score may have greater potential for beneficial results. More research is needed to investigate the joint contact forces as well as long term effects of riding with toe-in foot angles.

Key Words: knee osteoarthritis, toe-in, foot progression angle, knee moment, cycling, kinetics, kinematics

Running Title: Effects of toe-in on knee OA during cycling

INTRODUCTION

It has been estimated that nearly 27 million people in the USA suffered from osteoarthritis (OA) (Lawrence et al., 2008). While several treatments have been suggested to cope with the disease, there is, unfortunately, no cure for OA at this time. The medial compartment of the knee is the most common joint affected by OA (Lawrence et al., 2008) and has garnished much attention for treatment and symptom alleviation. While seemingly counterintuitive, exercise is one of the best treatment options for OA (Hochberg et al., 2012; Jordan et al., 2003; Zhang et al., 2008). The problem with exercise in OA sufferers is that certain activities can increase joint loading, be painful, and ultimately discourage continued exercise. Thus, to remedy the situation, exercises that reduce the joint loading such as cycling are commonly recommended by health professionals (Mangione et al., 1999). Unfortunately, beyond the fact that cycling reduces the body weight loading on the knees, there is practically no previous literature to suggest that cycling is beneficial for knee OA sufferers. In fact, knee injuries are the leading complaint in cycling (Dettori and Norvell, 2006; Kennedy et al., 2007).

Due to the lack of literature regarding cycling with OA, it is unclear if people with knee OA present the same cycling patterns as healthy individuals. In gait, it has been well established that people with OA do not present the same kinematics and kinetics as healthy individuals. Specifically, OA individuals show increased knee varus alignment and increased peak internal knee abduction moments (KAM) compared to healthy controls (Baliunas et al., 2002; Cerejo et al., 2002; Mundermann et al., 2005). Therefore, it is possible that differences in kinematics and kinetics between healthy and OA individuals may also exist in cycling as well. In experienced cyclists, Bailey et al. (2003) found that riders with a history of overuse knee pain had increased knee abduction angles when compared to the healthy controls. Thus, knee joint malalignment during cycling may be a concern for individuals with medial knee OA because it may exacerbate their OA symptoms or lead to other problems such as overuse injuries. If abnormal cycling kinematics and kinetics are present, it is possible that corrective measures can be taken to encourage normal riding patterns and reduce the chances of increased knee injuries while cycling.

During gait, the internal knee abduction moment (KAM), a surrogate measure for loading to the medial compartment of the knee (Schipplein and Andriacchi, 1991), has been shown to be an important factor associated with knee OA (Baliunas et al., 2002; Cerejo et al., 2002). Several researchers have shown it is possible to reduce the KAM through simple gait modification strategies (Fregly et al., 2007; Guo et al., 2007; Mundermann et al., 2008; Shull et al., 2013). Guo et al. (2007) attempted to reduce the KAM by requiring their participants to walk in an increased toe-out (foot progression) angle during walking. The results showed that participants were able to reduce their second peak KAM by 40% with a 15 degree increase of foot progression angle. However, no changes were noted for the first peak KAM which is a measure that is more closely related to loading response during gait and severity and progression of medial knee OA. Shull et al. (2013) attempted to reduce the KAM by having their participants walk in a toe-in foot progression angle (0.75 degree shank angle increase from baseline). They found that this method of walking reduced the first peak knee adduction moment by about 11% but the second peak KAM remained unchanged. The result provides a promising yet simple method to effectively reduce the KAM during walking, and may be a potential solution for reducing the KAM during cycling in the medial knee OA population.

To our knowledge, no studies have explored the effects of limb alignment alterations on the internal knee abduction moment of medial knee OA patients during cycling. Changes in lower extremity alignment using an increased toe-in foot progression angle could alter the frontal plane kinematics by placing the knee in a more medial position (Shull et al., 2013). This alignment change would decrease the length of the frontal plane moment arm of the pedal reaction force to the knee joint center, thus, decreasing the KAM. Previous literature has also suggested that the sagittal plane (Walter et al., 2010) and transverse plane (Boyd et al., 1997; Ruby and Hull, 1993) knee moments may be important variables for knee injuries. Therefore, kinematic and kinetic knee variables in all three planes of motion were analyzed in this study. Additionally, since we were directly manipulating the ankle joint by use of wedges, ankle kinematics in all three planes of motion were analyzed. Finally, PRF data were also analyzed in this study due to their direct influence on joint moments.Therefore, the primary purpose of this study was to investigate the effects of changes in toe-in foot progression angles on peak knee adduction angle and peak KAM in subjects with medial compartment knee OA during stationary cycling. It was hypothesized that toe-in foot progression angles would reduce the peak knee adduction angle and the peak KAM in subjects with medial compartment knee OA during stationary cycling compared to a neutral foot position.

METHODS

Participants

Eleven healthy male and female participants (age: 50.0 ± 9.7 yrs., height: 1.75 ± 0.12 m, weight: 80.17 ± 23.13 kg, BMI: 25.9 ± 5.4 kg/m^2) and thirteen participants with knee OA (age: 56.8 ± 5.2 yrs., height: 1.80 ± 0.14 m, weight: 83.2 ± 22.3 kg, BMI: 26.6 ± 3.6 kg/m^2) between the ages of 35 and 65 volunteered for participation in this study. Each participant with OA had medial compartment tibiofemoral OA in either one or both of their knees. To be included in the study, the OA participants must have had at least a grade 1 on the Kellgren-Lawrence score (Kellgren and Lawrence, 1957) (Grade 1: N=5, Grade 2: N=3, Grade 3: N=3, Grade 4: N=2) which was diagnosed and verified with radiographs by a rheumatologist. Participants were excluded from the study if they had OA in the hip or ankle joints, had previously had a lower extremity joint replacement, had knee joint arthroscopic surgery or intra-articular injections within 3 months prior to testing, had systemic inflammatory arthritis such as rheumatoid or psoriatic arthritis, had lower back pain that referred to the lower limbs. The participants were not excluded from the study if they had additional patellofemoral OA or OA in the lateral compartment of their knee (s). All OA subjects must have been experiencing pain the majority of the days of the week, for at least the previous 6 months. If subjects were taking any type of medication for their pain, they were asked to cease its use 2 days prior to the study. The healthy participants were pain free in their lower extremities for at least 6 months prior to the study and were not diagnosed with any type of lower extremity OA. All participants must have had a BMI of no more than 35 kg/m^2, and must have been able to walk and ride a stationary bike without aid. Each participant gave their informed consent which was approved by the Institutional Review Board.

For the X-rays, the OA participants performed bilateral standing in a semi flexed knee while anterior/posterior radiographs were taken of both knees in the frontal plane (Buckland-Wright et al., 2004). Additionally, a sagittal plane radiograph of each knee was collected while the participant stood in a semi flexed knee to determine the presence of patellofemoral OA.

Instrumentation

A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to acquire three-dimensional (3D) kinematics during the cycling test. The participants wore tight fitting spandex shorts and a t-shirt. To identify joint centers, anatomical markers were placed bilaterally on the 1st and 5th metatarsals, medial and lateral malleoli, medial and lateral epicondyles, left and right greater trochanters, left and right iliac crests, and left and right acromion processes. Four non-collinear tracking markers affixed to a semi-rigid thermoplastic shell was attached to the trunk, pelvis, thighs, and shanks using hook and loop wraps. For the feet, three markers were placed on the posterior and lateral side of heel counter of standard lab shoes (Noveto, Addidas).

A cycle ergometer (Excalibur Sport, Lode, Groningen, Netherlands) was used during testing. The ergometer was electro-mechanically braked which allowed for a precise workload setting that was independent of the pedal cadence. Additionally, the ergometer had removable pedals, and had the capability of adjusting the seat and handlebar to fit each rider.

A customized instrumented bike pedal was used on the Lode cycle ergometer, which allowed recordings of three dimensional forces and moments (Figure 3). The assembly contained two 3D force sensors (Type 9027C, Kistler, Switzerland) connected with two charge amplifiers (Type 5073A and 5072A, Kistler, Switzerland). The sensors could be placed in either the left or right pedal depending on the desired limb to be analyzed. A dummy pedal with the same mass and design was used on the opposite side. The pedal reaction forces and 3D kinematics were recorded through the Vicon Nexus system simultaneously.

Experimental Protocol

Upon arrival to the laboratory each participant filled out a KOOS (Knee Osteoarthritis Outcome Score) survey for each of their knees to assess knee pain and function during the week prior to the testing session. Participants then performed 3 minutes of treadmill walking at a self-selected pace which served as a warm-up and as a way to get a baseline VAS pain in their knees (one measurement before and one after the warm-up). Reflective markers were then placed on the individual's body segments for testing.

The seat height on the cycle ergometer was set so that the angle of the participant's knee was 30 degrees when the crank was at bottom dead center (Holmes et al., 1994). The horizontal seat depth was set so that the knee was in line with the pedal spindle when the crank was in the forward horizontal position (90°) (Burke, 2003). Each participant's trunk angle was also controlled by placing the handlebars in a position that created a 90° angle between the trunk and the thigh when the crank angle was at 90°. The participants were asked to warm up on the cycle ergometer for 3 minutes where knee pain levels were again recorded, one before and one after the warm-up.

The participants pedaled in 3 cycling conditions. The two toe-in conditions included 5 and 10 degree wedges placed between the pedal body and toe cage. The third was the control condition in which the participants pedaled with a neutral foot position established with a neutrally oriented pedal toe-cage. The testing conditions were randomized.

The cycling was performed for 2 minutes in each of the three conditions at a pedal cadence of 60 RPM and a workload of 80 Watts. Data were collected on 5 consecutive pedaling cycles from top dead center (TDC, 0°) to TDC (360°) for each condition, which began during the last 30 seconds of each trial. For the OA participants, an enlarged 0 to 10 numeric pain intensity scale was presented to the participants during this time, and they rated the pain in both of their knees (0 being no pain and 10 being worst pain possible). Participants could choose any real number between 0 and 10. Pain measurements for each knee were recorded at minutes 0, 1, and 2 during the cycling. Participants were given at least 2 minutes of rest between conditions.

Data and Statistical Analyses

Visual 3D (C-Motion Inc.), a 3D biomechanical analysis software suite, was used for signal processing and to obtain the 3D kinematic and kinetic computations for the lower extremity joints. A right hand rule was used to determine the polarity of the joint angles and moments and an X-Y-Z Cardan rotation sequence was used to compute joint angles. Both marker and pedal reaction force data were filtered using a zero lag, 4th order, digital Butterworth filter at 6 Hz (Gregersen et al., 2006a). Customized computer programs (VB_V3D and VB_Table) were used to determine critical events of the 3D kinematic and kinetic variables of interest from the output of Visual3D. For the kinematic and kinetic data, peaks were chosen at approximately 90° during the power phase of the crank cycle. This is the approximate time in the crank cycle when the rider is able to produce the most effort, which would have the greatest muscular impact on their joints. It should be noted that the moment variables were not normalized to any anthropometric feature (i.e. weight or height). In cycling we believe it is important not to normalize as the majority of the subject’s weight is carried by the cycle ergometer seat and handlebars. Thus, by not normalizing, we are able to get a better understanding of the actual moment value across the knee joint.

Independent samples t-tests were used to determine if KOOS scores for each subcategory were different between the two groups. A 2 x 3 (group x condition) mixed design analysis of variance (ANOVA) was used to detect differences between the cycling conditions and participant groups for pain and other selected biomechanical variables (IBM SPSS Statistics 20, Chicago, IL). When an interaction was present, a pairwise t-test was performed in post hoc analysis with Bonferroni adjustments to determine location of the statistical differences. An alpha level of 0.05 was set a priori. Additionally, a simple linear regression was performed for the OA patients to analyze the relationship between K/L score and the peak knee adduction angles and peak KAM. An alpha level of 0.05 was set a priori.

RESULTS

KOOS and VAS Knee Pain

All KOOS subscales (pain, symptoms, activities of daily living, sports, and quality of life) were lower in the OA group when compared to the healthy group (all p-values <0.001, Table 6). During treadmill walking, the VAS pain scores were 0.00 ± 0.00 for the healthy group and 1.19 ± 1.48 for the OA group. The VAS pain scores during cycling were 0.035 ± 0.08 cm, 0.046 ± 0.15 cm, and 0.11 ± 0.38 cm for the healthy group and 1.15 ± 1.10 cm, 1.25 ± 1.19 cm, and 0.96 ± 0.97 cm for the OA group for the neutral, 5° toe-in, and 10° toe-in conditions, respectively. The ANOVA revealed no interaction ($p=0.095$) or condition ($p=0.417$) effects. There was a group main effect found with the OA group experiencing more pain than the healthy group ($p=0.003$).

Pedal Reaction Forces

None of the PRF variables revealed a significant group or interaction effect. The peak vertical PRF was greater in the 5° toe-in condition compared to neutral ($p=0.028$, Table 7).

Ankle Joint Angles

The post hoc comparisons confirmed that the peak internal rotation angle was increased in the 5° ($p=0.003$) and 10° toe-in ($p<0.001$) conditions compared to the neutral condition (Figure 11). The ANOVA revealed a significant interaction for the peak plantarflexion angle ($p=0.044$). However, when the post hoc analysis was performed, no significant results were found. The peak eversion angle was increased in 10° toe-in condition compared to neutral ($p=0.023$), and overall the OA group presented less eversion than the healthy group (Table 8).

Knee Joint Angles

None of the knee angle variables revealed a significant group or interaction effect. The peak knee flexion angle was reduced in the 5° ($p=0.023$) and 10° ($p<0.001$) toe-in conditions compared to neutral (Table 9). There was also a significant reduction found in the 10° toe-in peak knee flexion angle compared to the 5° toe-in condition ($p=0.011$). Figure 12 shows representative curves for the knee adduction angle for one subject. Table 11 shows knee adduction angles for individual subjects as well as K/L scores for the OA subjects. The peak minimum knee adduction angle was decreased in the 5° ($p<0.001$) and 10° ($p<0.001$) toe-in conditions compared to neutral (Table 9). The results of the regression analysis revealed that the K/L score was a significant predictor for the peak knee adduction angle in all conditions ($r=0.810$ ($p<0.001$), $r=0.865$ ($p<0.001$), $r=0.847$ ($p<0.001$) for neutral, 5° toe-in, and 10° toe-in, respectively). Scatterplots of the relationship between K/L and peak minimum knee adduction angle are shown in Figure14.

Knee Joint Moments

None of the knee moment variables revealed a significant group or interaction effect. Figure 13 shows representative curves for the knee adduction moment for one subject. The peak internal rotation moment was reduced in the 10° toe-in condition compared to neutral (p=0.002, Table 10). The results of the regression analysis revealed that K/L score was a significant predictor of peak KAM in all conditions (r = -0.728 (p=0.002), r=-0.630 (p=0.011), and r=-0.812 (p<0.001) for neutral, 5° toe-in, and 10° toe-in, respectively). Scatterplots of the relationship between K/L and KAM are shown in Figure 15.

DISCUSSION

The purpose of this study was to examine the effects of 5 and 10 degrees of toe-in foot progression angles on knee joint biomechanics and pain in patients with medial compartment knee OA during stationary cycling. The primary hypothesis of this study was that decreasing the foot progression angle would decrease the knee adduction angle, internal knee abduction moment, and knee joint pain during cycling. Our hypothesis was partially supported with a significant decrease in the peak knee adduction angle for both the 5° and 10° toe-in conditions when compared to neutral. However, the noted changes in knee adduction angle were not accompanied by a decrease in KAM or pain when compared to the neutral condition.

There was very little difference in pain in the OA subjects during cycling compared to the treadmill walking warm-up. In the neutral and 10° toe-in conditions it appeared that pain decreased compared to walking (0.04 and 0.23 cm respectively). In the 5° toe-in condition, the pain appeared to increase (0.06 cm). However, these differences were very small, so it is unclear if the changes in pain level were actually a result of differences between walking and cycling, or if these particular subjects simply did not initially have much pain on the day of testing.

In the current study, the designed interventions through angled toe-cages did significantly increase the ankle internal rotation angle in both the 5° and 10° conditions (Figure 11). However, the change in ankle internal rotation angle from neutral was less than the actual change in toe-cage angle. While the toe-cage has straps to help constrict foot motion, there is still some flexibility for the rider to move their foot. Previous literature has shown the importance of allowing some freedom of movement between the foot and pedal; helping to reduce joint moments, and concomitantly, over-use knee injuries (Boyd et al., 1997; Ruby and Hull, 1993). Interestingly, we also found a significant decrease in peak ankle eversion angle in the 10° toe-in condition compared to neutral. This appears to be a result of the natural anatomy of the ankle joint since the ankle does not have joint axis that are perfectly perpendicular to the foot and shank segments (Lundberg et al., 1989). Due to the off-axis rotations, and because ankle movements occur in both the talocrural and subtalar joints, it is reasonable that increased internal rotation would be coupled with increased inversion (or reduced eversion). Another interesting result is that the peak eversion angle was greater for the healthy group than for the OA group. It is difficult to speculate why the eversion ankles of the two groups would be different. It is possible that the OA subjects' knees were closer to the cycle ergometer, thus, the frontal plane angle between the shank and the foot would be reduced in the OA subjects. However, we did not measure the distance between the knee and cycle ergometer and there were no group differences found in the peak knee adduction angle.

When the ankle was internally rotated in the toe-in conditions, we did find that the peak knee adduction angle also decreased in both the 5° and 10° toe-in conditions when compared to neutral. On average across all OA subjects, we found a 61.4% (2.7°), and 72.7% (3.2°) reduction in the peak adduction angle for the 5° and 10° toe-in conditions respectively. This is a potentially substantial result when considering a study by Bailey et al. (2003) that showed a 2.5° difference in knee abduction angle between asymptomatic and previously injured cyclists. We would like to note that in the neutral condition, the OA group in our study did have greater average peak knee adduction angle than the healthy group by 2.2°. However, a statistical difference was not found between the groups because of high variability (Table 9) and it is not clear if a 2° difference in adduction angle is clinically meaningful for knee OA subjects. The results from Bailey et al. (2003) suggest that subjects with large shifts in adduction angle from a more neutral position (i.e. hovering around zero) may be at increased risk of OA progression or other overuse knee injuries during cycling. Due to the large variations in frontal plane angle across subjects, we are cautious in making this claim, and suggest more research be done to examine the reasons for high frontal plane knee angle variability. We did include OA patients with all K/L grades in this study, and 5 of our OA participants had a K/L score of 1. Our subject's KOOS scores were similar to the KOOS's reference data which did not include a K/L of 1 (Roos et al., 1999). As suggested earlier, however, knee adduction angle increases with medial knee OA severity (Mundermann et al., 2005). Individual OA subject data from Table 11 supports this claim, with the participants with higher K/L scores producing larger adduction angles on average than those with lower K/L scores. The regression analysis revealed that the K/L score was significantly correlated with the peak knee adduction angles. There were strong and positive correlations between K/L score and peak knee adduction angle, with the K/L score explaining 65%, 75%, and 72% of variation in knee adduction angle for the neutral, 5° toe-in and 10° toe-in conditions respectively (Figure 14). Therefore, it is possible that a patient sample with a more severe medial knee OA would have greater adduction angles and would result in a greater deviation from the healthy group; perhaps making them more susceptible to OA progression or other injury. It is worth noting that in previous cycling studies of healthy subjects, the frontal plane knee angle hovered around zero, ranging between 2-4° of abduction to 1-6° of adduction (Bailey et al., 2003; Umberger and Martin, 2001). The average (but not for every individual) frontal plane angle of our healthy subjects remained adducted throughout the crank cycle, ranging between 2° and 13° of adduction (Table 11). The differences in healthy subjects between previous studies and our study could be due to the experience of the participants (previous studies used experienced cyclists), the age of the participants (previous studies age was about 28 years), or the fact that previous studies used clipless pedals while we used toe-cages. Additionally, the previous studies used a 2D analysis, while we used 3D.

During gait, the KAM has been found to be a surrogate measure for loading to the medial compartment of the knee (Schipplein and Andriacchi, 1991). Previous studies have found that an increase in the KAM is associated with an increased risk of OA severity and/or progression (Chakravarty et al., 2008; Miyazaki et al., 2002; Mundermann et al., 2005; Wada et al., 2001). Much emphasis has been placed on reducing the KAM for OA patients with the intent of reducing the risk of accelerated progression of OA and OA severity. Shull et al. (2013) showed promising results during gait with a 13% reduction in KAM by using a foot progression angle that was 5° less (toe-in) than the subject's baseline value. During cycling, previous literature has suggested a potential for a decreased risk of overuse knee injuries with a decrease in KAM when some amount of movement is allowed between the foot and the pedal (Boyd et al., 1997; Ruby and Hull, 1993) and when the cyclists adapts a more everted foot position (Gregersen et al., 2006a). The magnitude of our KAM during neutral cycling was similar to that of previous literature (Boyd et al., 1997; Gregersen and Hull, 2003; Gregersen et al., 2006a), however, we did not see a significant reduction in KAM in the toe-in conditions compared to neutral. Consistent with previous literature (Boyd et al., 1997; Gregersen and Hull, 2003), we did see large variability in the KAM which may be the reason why we did not find a significant decrease in KAM in the toe-in conditions compared to neutral. Even though sagittal plane cycling kinematics are relatively controlled by seat height, crank length, and bike frame geometry, there is much less restriction in the frontal plane, especially for the knee joint. Therefore, some individuals in this study might have benefited from a toe-in intervention while others showed no relative change. For example, similar to the peak minimum knee adduction angle, Table 11 suggests that individuals with higher K/L grades may have higher KAM. This observation was further confirmed by the moderate and positive linear relationships between KAM and K/L score (Figure 15). The results of the regression analysis for KAM suggest that differences in K/L score explain 53%, 40%, and 66% of the variation in the peak KAM for the neutral, 5° toe-in, and 10° toe-in conditions respectively. However, due to the large variability within OA subjects, we suggest that future cycling interventions be designed on individual basis. In addition to the KAM, we found a significant decrease in the peak internal rotation moment in the 10° toe-in condition compared to neutral. While the internal rotation moment has been less studied in bike related research, it is nonetheless an important variable as past work has shown that a decrease in non-driving intersegmental moments has potential to reduce overuse knee injuries (Boyd et al., 1997; Ruby and Hull, 1993). These researchers also found that a decrease in internal rotation and knee adduction moments occurs when a rider is free to internally/externally rotate their foot on the pedal when compared a condition in which the foot is fixed to the pedal (i.e. no rotations). Due to the lack of absolute restriction in our toe-cage, our toe-cage on the pedal did allow some transverse-plane rotations of foot.

It is important to remember that knee injuries are the leading complaint in cycling. Thus, while decreasing the KAM appears important, we should also be cognizant of the influence of interventions on other variables. In the current study, the peak vertical PRF was indeed increased in the 5° toe-in condition compared to neutral, but not the 10° condition, although the 10° condition was close to being significant ($p=0.058$). Previous gait literature has suggested that an increase in the knee extensor moment may negate the effects of a decreased KAM (Walter et al., 2010). We did not see an increase in the peak knee extensor moment in this study. Thus, our results imply that the increase in vertical PRF did not influence the net effort of the knee extensors to keep the cycle moving at the desired pace and workload. A musculoskeletal modeling analysis would be required to verify this finding.

There are a few limitations of this study. Due to the high variability in frontal plane variables, it is necessary to increase the sample size to get a better understanding of the true differences between OA and healthy subjects during cycling. Additionally, the pain levels in our OA subjects were relatively low. Thus, the effect of the intervention on the OA subjects may have been different if baseline pain values were higher. Also, we included OA subjects with a minimum K/L score of 1. This is technically considered a diseased joint, but generally a K/L of 1 represents a mild stage of OA. It is not clear if excluding subjects with a K/L of 1 would reveal different results. Similarly, we did not perform x-rays on the healthy subjects. Thus, while the healthy subjects reported no knee problems, we did not have radiographic support of no OA. Finally, since we included subjects with a BMI up to 35, difficulty in marker placement due to excess soft tissue may have introduced errors in calculating knee and hip joint centers. However, the same researcher placed markers on all subjects to ensure that marker placements were consistent across subjects.

CONCLUSION

The findings of this study indicate that the use of both 5° and 10° toe-in foot progression angles during seated cycling was effective in reducing knee adduction angles in medial compartment knee OA and healthy subjects. However, these interventions were not effective in reducing peak KAM or subjective knee pain. It is not known if the decrease in knee adduction angle has potential to influence the progression of OA. However, previous literature suggests small variations from a neutral knee joint during cycling have the potential to increase the risk of overuse knee injuries, and our results suggest that individuals with higher K/L scores have an increased knee adduction angle and KAM. . Additionally, even though there was a decrease in the knee adduction angle, there was also an increase in the vertical PRF which may be an indicator of increased joint loading. This is the first study to report changes in knee joint biomechanical variables with the use of toe-in interventions in knee OA patients during cycling. Further work is warranted to examine muscle forces and knee joint loading, why frontal plane knee joint variables are so variable, and the long term effects of toe-in foot progression angles during cycling.

CHAPTER V APPENDIX: TABLES AND FIGURES

Tables

Table 6: KOOS subscale normalized scores.

Subscale	Group	Mean	SD	p value
Pain	Healthy	95.73	8.27	<0.001
	OA	70.38	11.93	
Symptom	Healthy	93.09	10.00	<0.001
	OA	56.85	11.86	
ADL	Healthy	96.55	7.16	<0.001
	OA	72.92	15.71	
Sports	Healthy	91.36	15.51	<0.001
	OA	53.69	24.03	
QOL	Healthy	91.00	16.58	<0.001
	OA	51.62	13.06	

ADL: Activities of daily living; QOL: Quality of life

Table 7: Peak PRF during the downward phase of the crank cycle (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Toe-in	10° Toe-in	Neutral	5° Toe-in	10° Toe-in	Grp	Cond	Int
Peak Medial PRF (N)	-30.70±11.83	-33.60±9.27	-31.95±11.38	-28.59±8.87	-28.94±12.94	-25.55±11.07	0.275	0.193	0.653
Peak Posterior PRF (N)	-64.69±19.98	-71.58±24.23	-67.81±26.43	-80.46±17.67	-78.41±21.79	-75.61±24.54	0.258	0.361	0.204
Peak Vertical PRF (N)	236.35±46.60	246.16±44.87[#]	250.38±50.21	234.49±34.24	249.21±34.27[#]	242.97±26.62	0.896	**0.034**	0.245

[#]: Significantly different than neutral; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 8: Peak Ankle Joint Angles (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Toe-in	10° Toe-in	Neutral	5° Toe-in	10° Toe-in	Grp	Cond	Int
Peak Plantarflexion Angle (°)	-8.9±10.7	-8.2±10.3	-9.3±10.2	-6.0±8.5	-11.0±8.6	-11.7±8.2	0.834	0.152	**0.044**
Peak Eversion Angle (°)	-13.2±8.4	-11.8±7.0	-12.0±6.8[#]	-6.8±6.6	-5.2±6.4	-3.9±5.8[#]	**0.015**	**0.031**	0.383

[#]: Significantly different than neutral; Plantarflexion and eversion are negative; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 9: Peak knee joint angles during the downward phase of the crank cycle (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Toe-in	10° Toe-in	Neutral	5° Toe-in	10° Toe-in	Grp	Cond	Int
Peak Flexion Angle (°)	-44.9±7.8	-43.0±8.3[#]	-41.6±7.9[#&]	-39.8±8.1	-38.4±8.9[#]	-35.5±8.8[#&]	0.129	< **0.001**	0.586
Peak Adduction Angle (°)	2.2±5.3	0.5±4.8[#]	-0.2±4.5[#]	4.4±5.6	1.7±6.5[#]	1.2±6.7[#]	0.494	< **0.001**	0.573
Peak External Rotation Angle (°)	-5.0±5.2	-3.9±4.9	-3.8±4.5	-2.9±5.4	-1.4±4.9	-1.8±5.5	0.281	0.166	0.796

[#]: Significantly different than neutral; [&]: Significantly different than 5° wedge. Flexion, abduction, and external rotation are negative; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 10: Peak knee joint moments during the downward phase of the crank cycle (mean ± SD).

	Healthy			OA			P value (ANOVA)		
Variable	Neutral	5° Toe-in	10° Toe-in	Neutral	5° Toe-in	10° Toe-in	Grp	Cond	Int
Peak Extensor Moment (Nm)	26.27±9.60	27.54±10.11	27.75±9.97	27.97±7.42	28.33±9.66	26.70±9.64	0.899	0.587	0.190
Peak Abduction Moment (Nm)	-9.00±4.74	-9.54±4.92	-9.12±4.74	-7.72±4.76	-6.93±3.66	-6.69±4.18	0.242	0.562	0.484
Peak Internal Rotation Moment (Nm)	7.98±4.29	6.96±2.99	6.10±3.92[#]	6.58±3.32	4.88±4.40	2.92±3.88[#]	0.127	**0.004**	0.465

[#]: Significantly different than neutral. Extensor, abduction, and external rotation moments are negative; Grp: Group; Cond: Condition; Int: Group x condition interaction.

Table 11: Individual subject K/L scores, overall knee joint position during the neutral condition, and peak maximum and minimum knee adduction angles.

			Neutral			5° Toe-in			10° Toe-in		
OA Subject	K/L Score	Joint Pos	Peak (+)	Peak (-)	KAM (Nm)	Peak (+)	Peak (-)	KAM	Peak (+)	Peak (-)	KAM
1	1	Neutral	8.1	-5.9	-2.76	3.2	-7.7	-4.06	4.2	-6.4	-3.12
2	4	ADD	14.1	9.5	-12.39	9.4	5.8	-13.50	10.1	5.7	-14.90
3	3	ADD	21.3	10.2	-6.24	17.7	6.0	-7.50	24.5	6.9	-8.47
4	2	ADD	11.5	4.8	-1.92	8.2	3.5	-3.71	6.9	0.4	-2.16
5	1	ADD	11.8	4.9	-4.03	6.2	-2.4	-2.95	10.1	0.1	-2.80
6	2	ADD	16.8	8.2	-14.38	16.9	6.4	-11.41	13.9	5.5	-7.16
7	1	Neutral	8.2	-1.6	-5.02	3.7	-5.3	-9.02	-0.1	-8.9	-7.92
8	1	ADD	8.1	3.0	-2.58	2.5	-2.4	-3.08	0.9	-3.7	-2.96
9	1	Neutral	10.1	-1.8	-6.83	5.6	-4.0	-4.19	3.4	-4.0	-2.87
10	3	ADD	9.4	5.7	-14.10	8.3	4.1	-10.29	7.6	2.1	-11.29
11	2	Neutral	5.7	-0.3	-5.69	5.6	-3.7	-3.26	3.5	-4.1	-2.88
12	3	ADD	13.9	7.4	-9.91	11.1	6.7	-10.23	10.2	6.2	-10.69
13	4	ADD	19.4	13.7	-14.57	23.4	15.5	-6.87	21.5	15.4	-9.69
		Mean	12.2	4.4	-7.72	9.4	1.7	-6.93	9.0	1.2	-6.68
		STD	4.7	5.6	4.75	6.3	6.5	3.66	7.5	6.7	4.18

Healthy subject	K/L Score	Joint Pos	Peak (+)	Peak (-)	KAM	Peak (+)	Peak (-)	KAM	Peak (+)	Peak (-)	KAM
1	-	ADD	15.4	7.4	-13.69	15.3	5.4	-16.33	10.6	2.3	-17.90
2	-	Neutral	7.6	-4.2	-8.29	4.5	-6.4	-7.45	5.9	-7.0	-8.69
3	-	Neutral	11.2	-1.2	-9.15	12.5	-0.4	-7.76	14.1	0.4	-7.42
4	-	ADD	17.3	6.9	-17.34	18.2	6.7	-18.70	16.9	5.4	-15.29
5	-	Neutral	4.5	-1.6	-6.10	1.8	-4.0	-4.54	3.4	-3.7	-4.15
6	-	ADD	12.9	1.0	-7.54	9.9	2.3	-7.62	8.5	-0.2	-7.84
7	-	ADD	23.6	10.7	-14.74	21.9	6.7	-12.83	23.3	7.5	-13.43
8	-	ADD	10.8	3.3	-3.75	5.9	-2.8	-5.31	11.0	-1.2	-5.86
9	-	Neutral	5.3	-6.0	-1.43	6.2	-4.3	-2.98	5.3	-5.7	-2.06

Table 11 Continued

Healthy subject	K/L Score	Joint Pos	Peak (+)	Peak (-)	KAM	Peak (+)	Peak (-)	KAM	Peak (+)	Peak (-)	KAM
10	-	ADD	14.4	1.5	-7.12	13.5	-2.2	-11.42	12.4	-2.2	-9.65
11	-	ADD	18.1	6.6	-9.80	18.0	4.7	-10.00	16.8	2.6	-8.07
		Mean	12.8	2.2	-9.00	11.6	0.5	-9.54	11.6	-0.2	-9.12
		STD	5.8	5.3	4.74	6.5	4.8	4.92	5.9	4.5	4.74

Joint Pos: The overall position of the knee joint during the neutral condition. If the peak positive and peak negative joint angles surrounded zero, then the joint position was designated as Neutral. If the peak positive and negative values were both greater than 0, then the joint position was designated as continuously adducted (ADD). Peak (+): Peak maximum knee adduction angle (°); Peak (-): Peak minimum knee adduction angle (°); KAM: Peak knee abduction moment (Nm).

Figures

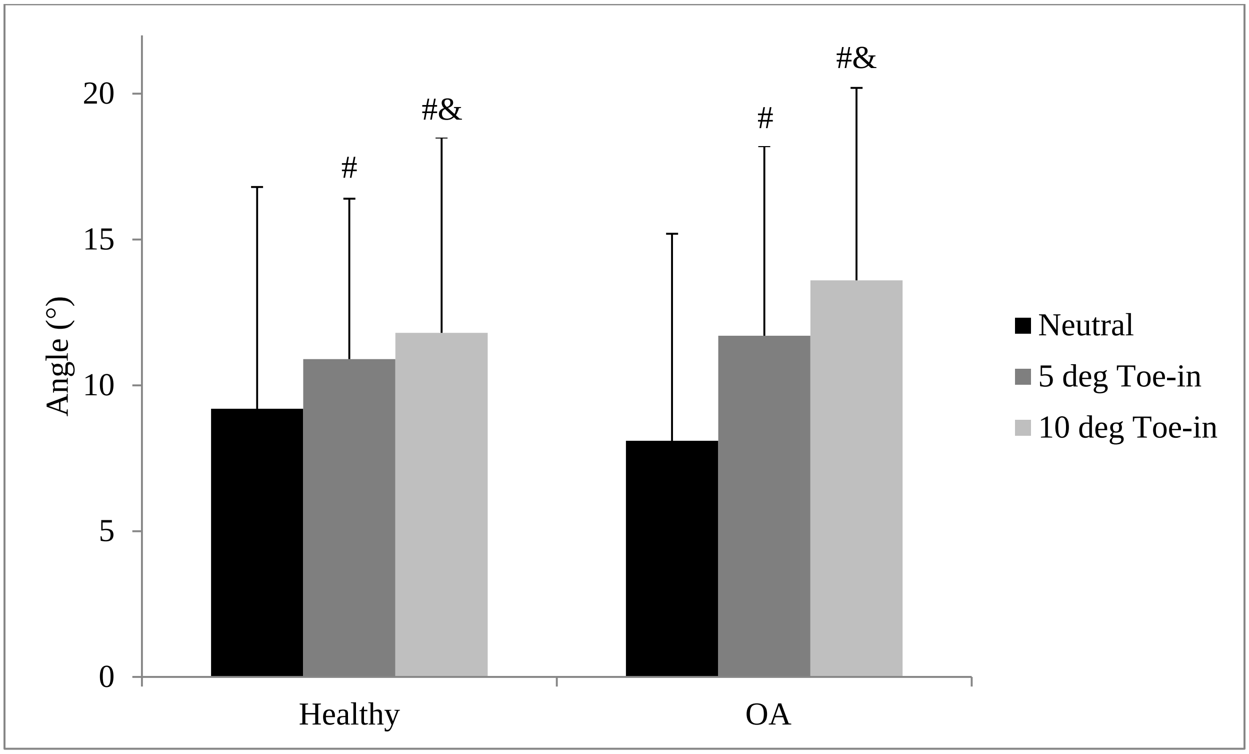

Figure 11: Peak ankle internal rotation angles. #: Significantly different than neutral; &: Significantly different than 5° Toe-in

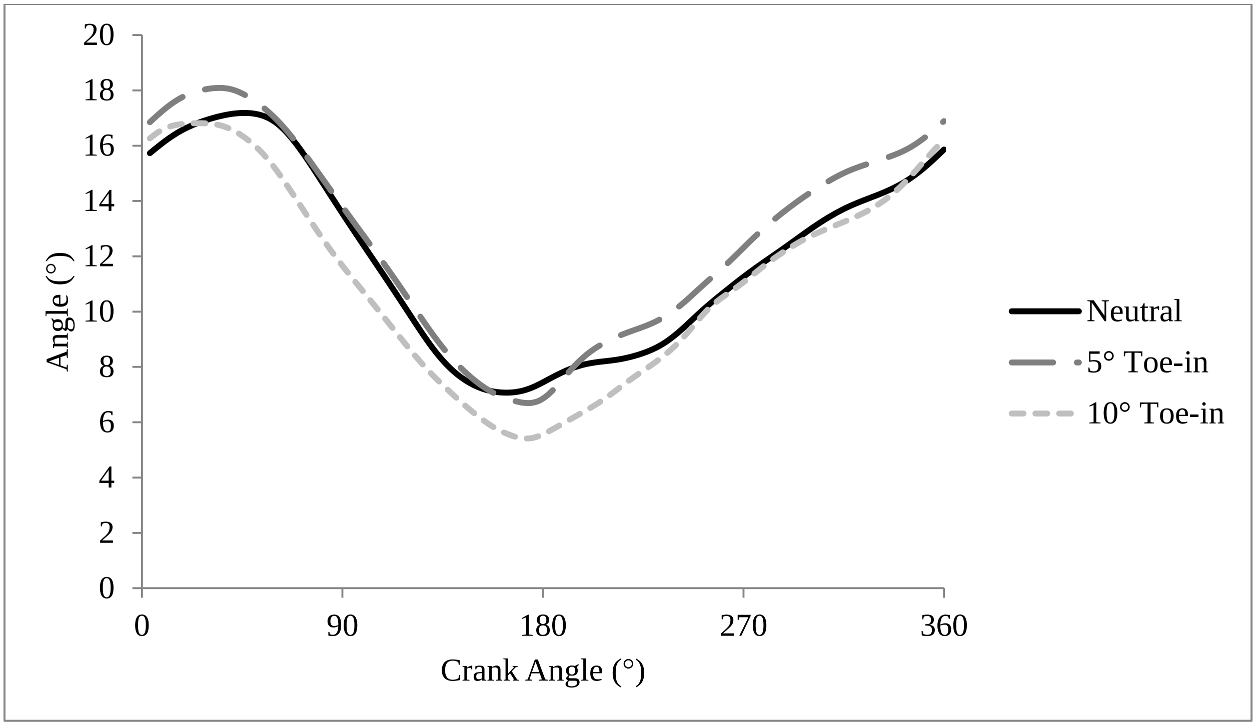

Figure 12: Representative knee adduction angles from one subject.

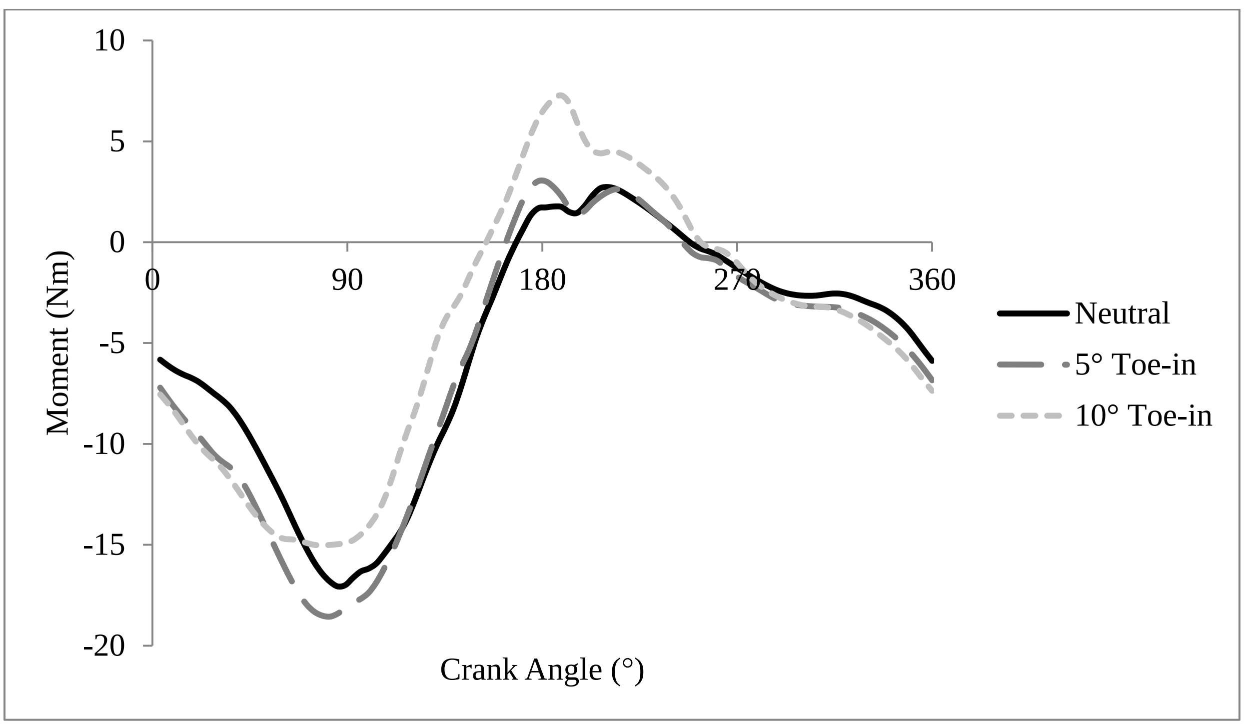

Figure 13: Representative knee abduction moments from one subject.

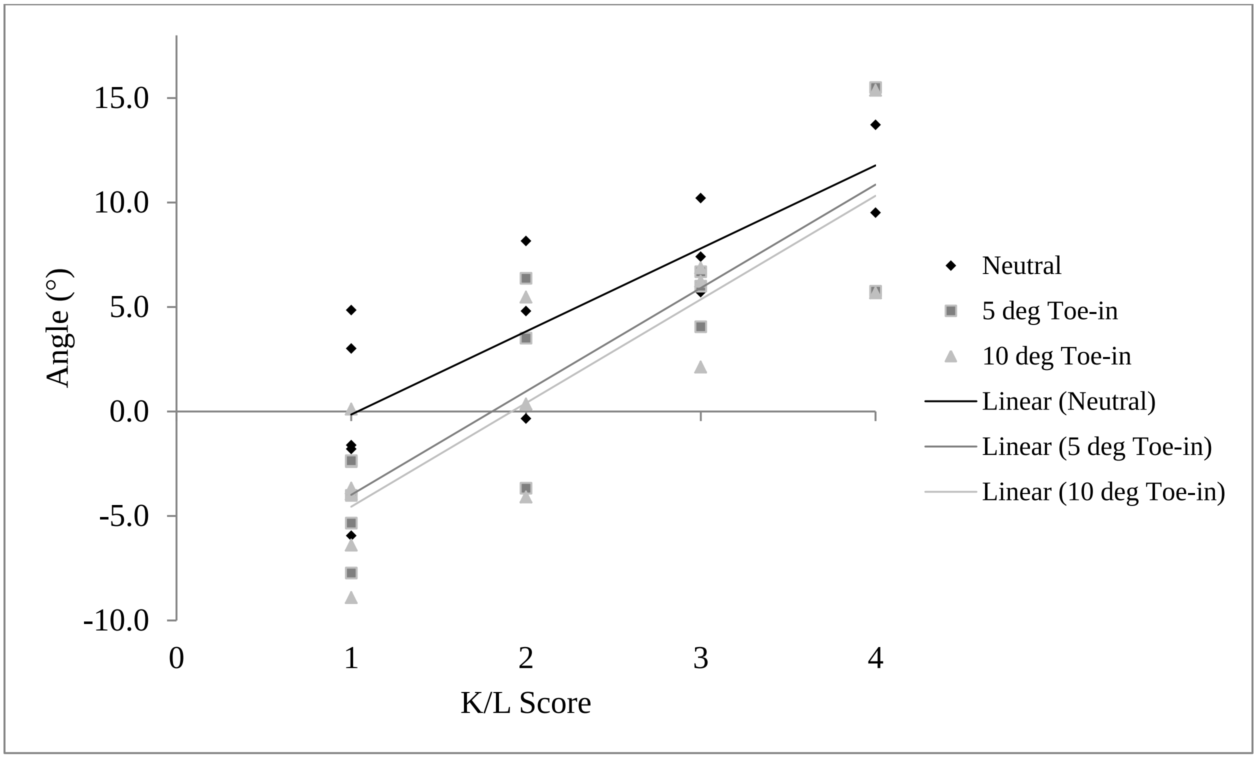

Figure 14: Scatterplots and linear trend lines for the relationship between minimum peak knee adduction angle and K/L score for individual OA patients across conditions. R^2 values are 0.66, 0.75, and 0.72 for neutral, 5° toe-in, and 10° toe-in, respectively.

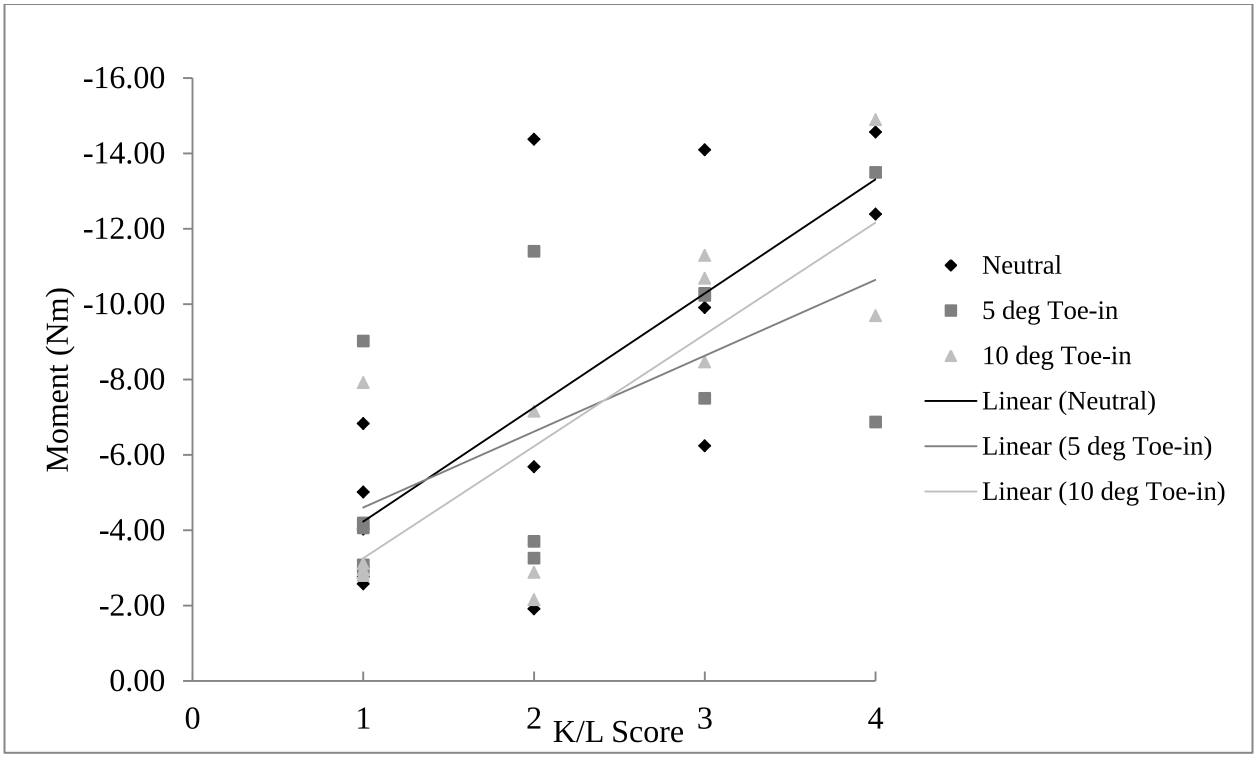

Figure 15: Scatterplots and linear trend lines for the relationship between peak KAM and K/L score for individual OA patients across conditions. R^2 values are 0.53, 0.40, and 0.66 for neutral, 5° toe-in, and 10° toe-in, respectively.

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APPENDICES

APPENDIX A: Individual Subject Characteristics

Table 12: Individual healthy subject characteristics

Subject #	Gender	Age (years)	Height (m)	Mass (kg)	BMI (kg/m2)
1	Male	35	1.83	104.54	31.2
2	Male	45	1.86	91.36	26.6
3	Male	42	1.69	68.41	24.0
7	Male	65	1.88	78.25	22.2
9	Female	49	1.64	57.27	21.4
15	Female	37	1.70	56.60	19.6
16	Female	55	1.66	87.73	32.0
17	Female	51	1.55	52.95	22.0
18	Female	54	1.70	93.18	32.2
19	Male	54	1.79	64.55	20.1
20	Male	63	1.95	127.05	33.4
Mean	-	50.0	1.75	80.17	25.9

Table 13: Individual OA subject characteristics

Subject #	Gender	Age (years)	Height (m)	Mass (kg)	BMI (kg/m2)
5	Female	58	1.80	99.55	30.7
6	Male	65	1.85	102.06	29.7
8	Female	63	1.57	52.95	21.5
10	Female	57	1.70	66.00	22.8
11	Female	59	1.69	58.18	20.5
12	Male	54	2.05	130.91	31.2
13	Female	55	1.69	85.00	29.8
14	Male	46	1.84	97.68	28.9
21	Male	54	1.79	91.82	28.8
22	Male	63	1.88	87.50	24.9
23	Female	59	1.57	61.59	25.0
24	Female	55	1.59	61.59	24.4
25	Male	51	1.78	87.27	27.6
Mean	-	56.8	1.8	83.2	26.6

APPENDIX B: Informed Consent Forms

Informed Consent Form for Healthy Subjects

INFORMED CONSENT FORM

Effects of Lateral Shoe Wedges and Toe-in Foot Progression Angles on the Biomechanics of Knee Osteoarthritis during Stationary Cycling

Principal Investigator: Jacob Gardner
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Faculty Advisor: Songning Zhang, Ph.D.
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Co-Investigator: Gary Klipple, MD
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Introduction
You are invited to participate in this research study because you are a healthy adult aged between 35 and 65 years old. The purpose of this study is to investigate the effects of changes in bicycle pedal wedge angles and toe-in angles on the motions and pain in the knee for individuals with knee osteoarthritis (OA) and compare them to healthy people. Please ask the study staff to explain any words or information that you do not clearly understand. Before agreeing to be a participant in this study, it is important that you read and understand the following explanation of the procedures, risks, and benefits.

Testing Protocol
Before testing, you will be asked to fill out a Physical Activity Readiness Questionnaire which assesses your readiness to participate in physical activity. If you answer “yes” to any question in the Par-Q, you will be asked to obtain written consent from your doctor that you are healthy enough to participate in this study. A physician permission form will be sent to your physician by the principal investigator to obtain the written consent. We will also measure your height and weight. If you qualify, you will be asked to attend one biomechanical test session (i.e. recording your joint movements) in the Biomechanics/Sports Medicine Lab on the UT campus that will take approximately 1 to 1.5 hours. Parking at the campus will be free to you. For the testing session, you will be asked to wear clothing appropriate for cycling exercise which includes spandex shorts. If you do not have this type of clothing, paper laboratory shorts will be provided.

Prior to data collection, you will warm up on the treadmill and on the stationary cycle for 3 minutes each to allow your joints and muscles to get ready for the cycling exercise. After the warm up, reflective markers will be placed on both sides of your feet, ankles, legs, knees, thighs, pelvis and trunk in order to capture your movements during cycling. Upon completion of all marker placement you will perform two minutes of cycling at an 80 Watt workload and a 60 RPM pedal cadence for each condition in the testing session. There are five conditions in the testing session so you will cycle up to 20 minutes including your warm up. You will be given at least a two minute rest in between conditions. You can end any condition early and are under no obligation to complete the test.

During the testing, biomechanics instruments such as reflective markers and motion capture cameras will be used to obtain measurements. The reflective markers will be placed on your body using double stick medical tape and hook and loop wraps. None of the instruments will impede your ability to engage in normal and effective motions during the test. The cameras will not record pictures of you. If you have any further questions, interests or concerns about any equipment, please feel free to ask the investigator

Potential Risks

Risks associated with this study are minimal. You will be required to pedal a stationary cycle up to 20 minutes including a warm up and cool down. If you are not used to regular cycling exercise, you may experience delayed onset muscle soreness (DOMS) in which the muscles are sore for a day or two following the exercise session. However, these conditions are normal for any person who is not accustomed to regular physical activity. Additionally, due to the demands of physical activity, there is risk for a cardiovascular event to occur (i.e. dizziness, shortness of breath, heart attack, or stroke). However, prior to the test you will fill out a Physical Activity Readiness Questionnaire that indicates you are ready for physical activity. Should any injury occur during the course of testing, standard first aid procedures will be administered as necessary. At least one researcher with a basic knowledge of first aid procedures will be present at each test session. All tests will be conducted and the equipment will be handled by qualified research personnel in the Biomechanics/Sports Medicine Laboratory. In the unlikely event a physical injury is suffered as a result of participation in this study (during the warm up and testing session), the University of Tennessee does not automatically provide reimbursement for medical care or other compensation and you will be responsible for any medical expenses. If physical injury is suffered in the course of research, or for more information, please notify Jake Gardner (974-2091).

Benefits of Participation

Results from the proposed study will help establish appropriate exercise protocols for people with knee OA in order to reduce pain while cycling. The findings may directly help you if you suffer from knee osteoarthritis, and may help you learn how to exercise in a way to avoid knee pain.

Compensation

You will be compensated $10.00 for completing this study. No partial compensation will be given for only completing part of the study. You will be eligible for payment once the cycling portion of the testing begins.

Voluntary Participation and Withdrawal

Your participation is entirely voluntary and your refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled. You may withdraw from the study at any time without penalty. It is your obligation to ask questions regarding any aspect of this study that you do not understand. You acknowledge that you have been offered the opportunity to have any questions answered. Your participation in this study may be stopped if you fail to follow the study procedures or if the investigators feel that it is in your best interest to stop participation.

Confidentiality

Your identity will be held in strict confidence through the use of a coded subject number during data collection, data analysis, and in all references made to the data, both during and after the study, and in the reporting of the results. The results will be disseminated in the form of presentations at conferences, and publications in journals. The consent form containing your identity information will be destroyed three years after the completion of the study. If you decide to withdraw from the study, your information sheet and consent form with your identity and injury history will be destroyed.

Contact Information

If you have any questions about the study at any time or if you experience adverse effects as a result of participating in this study you can contact Jacob Gardner at 1914 Andy Holt Ave. 136 HPER Bldg., The University of Tennessee (974-2091). Questions about your rights as a participant can be addressed to Compliance Officer in the Office of Research at the University of Tennessee at (865) 974-3466.

Consent Statement

The study has been explained fully to my satisfaction and I agree to participate as described. I have been given the opportunity to discuss all aspects of this study and to ask questions. Answers to such questions, if any, were satisfactory. I am eighteen years of age or older, in good health, am qualified for the study and freely give my informed consent to serve as a subject in this study. I have received a copy of this form.

Subject's Name: ___________________ Subject's Signature: ________________________ Date: _________

Investigator's Signature: ____________________________ Date: __________ Subject # ______

Informed Consent Form for OA Subjects

INFORMED CONSENT FORM

Effects of Lateral Shoe Wedges and Toe-in Foot Progression Angles on the Biomechanics of Knee Osteoarthritis during Stationary Cycling

Principal Investigator: Jacob Gardner
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Phone: (865) 974-4716

Co-Investigator: Gary Klipple, MD
Address: 1924 Alcoa Highway Box U-114
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Phone: (865) 305-9340

Introduction

You are invited to participate in this research study because you are an adult with knee osteoarthritis aged between 35 and 65 years old. The purpose of this study is to investigate the effects of changes in bicycle pedal wedge angles and toe-in angles on the motions and pain in the knee for individuals with knee osteoarthritis (OA). Please ask the study staff to explain any words or information that you do not clearly understand. Before agreeing to be a participant in this study, it is important that you read and understand the following explanation of the procedures, risks, and benefits.

Testing Protocol

If you qualify for the study based on the initial phone screening, you will be asked to attend one session at the Rheumatology division at the UT medical center where you will be assessed for BMI, and physical activity readiness. If you qualify based on these items, you will then be evaluated by a rheumatologist for knee OA and asked to have your knees X-rayed. The X-rays will be used to diagnose and confirm if you have knee OA and to assess OA severity. If you were previously evaluated for our recent cycling study, you will not be required to have X-rays taken or be evaluated again. If you qualify for the study based on your X-rays, you will be asked to attend one biomechanical test session (i.e. testing your joint movements) in the Biomechanics/Sports Medicine Lab on the UT campus that will take approximately 1 to 1.5 hours. Parking at the UT Medical Center and main campus will be free to you. You will also be asked to fill out the Knee injury and Osteoarthritis Outcome Score (KOOS) survey to assess your knee pain and function, and associated problems during common daily activities on both of your knees. For the testing session, you will be asked to wear clothing appropriate for cycling exercise which includes spandex shorts. If you do not have this type of clothing, paper laboratory shorts will be provided. Before testing, you will be asked to fill out a Physical Activity Readiness Questionnaire which assesses your readiness to participate in physical activity. If you answer “yes” to any question in the Par-Q, you will be asked to obtain written consent from your doctor that you are healthy enough to participate in this study. A physician permission form will be sent to your physician by the principal investigator to obtain the written consent.

Prior to data collection, you will be asked to warm up on the treadmill and on the stationary cycle for 3 minutes each to allow your joints and muscles to get ready for the cycling exercise. You will also be asked to rate your knee pain at several points during the warm-up and testing session. After the warm up, reflective markers will be placed on both sides of your feet, ankles, legs, knees, thighs, pelvis and trunk in order to capture your movements during cycling.

Upon completion of all marker placement you will perform two minutes of cycling at an 80 Watt workload and a 60 RPM pedal cadence for each condition in the testing session. There are five conditions in the testing session so you will cycle up to 20 minutes including your warm up. Several times throughout the experimental tests you will rate your knee pain level on a visual scale. You will be given at least a two minute rest in between conditions. You can end any condition early and are under no obligation to complete the test.

During the testing, biomechanics instruments such as reflective markers and motion capture cameras will be used to obtain measurements. The reflective markers will be placed on your body using double stick medical tape and hook and loop wraps. None of the instruments will impede your ability to engage in normal and effective motions during the test. The cameras will not record pictures of you. If you have any further questions, interests or concerns about any equipment, please feel free to ask the investigator

Potential Risks

Risks associated with this study are minimal. The radiation exposure of the X-rays you will receive is 0.003 milliSieverts, which is equivalent to three days of exposure to natural background radiation. You will be required to pedal a stationary cycle up to 20 minutes including a warm up and cool down. If you are not used to regular cycling exercise, you may experience swelling, tenderness, stiffness, or pain in your knees for a few days after the test. Additionally, you may experience delayed onset muscle soreness (DOMS) in which the muscles are sore for a day or two following the exercise session. However, these conditions are normal for any person who is not accustomed to regular physical activity. Additionally, due to the demands of physical activity, there is risk for a cardiovascular event to occur (i.e. dizziness, shortness of breath, heart attack, or stroke. However, prior to the test you will fill out a Physical Activity Readiness Questionnaire that indicates you are ready for physical activity. Should any injury occur during the course of testing, standard first aid procedures will be administered as necessary. At least one researcher with a basic knowledge of first aid procedures will be present at each test session. All tests will be conducted and the equipment will be handled by qualified research personnel in the Biomechanics/Sports Medicine Laboratory. In the unlikely event a physical injury is suffered as a result of participation in this study (during the warm up and testing session), the University of Tennessee does not automatically provide reimbursement for medical care or other compensation and you will be responsible for any medical expenses. If physical injury is suffered in the course of research, or for more information, please notify Jake Gardner (974-2091).

Benefits of Participation

Results from the proposed study will help establish appropriate exercise protocols for people with knee OA in order to reduce pain while cycling. The findings may directly help you if you suffer from knee osteoarthritis, and may help you learn how to exercise in a way to avoid knee pain.

Compensation

You will be compensated $10.00 for this study. No partial compensation will be given for only completing part of the study. You will be eligible for payment once the cycling portion of the testing begins.

Voluntary Participation and Withdrawal

Your participation is entirely voluntary and your refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled. You may withdraw from the study at any time without penalty. It is your obligation to ask questions regarding any aspect of this study that you do not understand. You acknowledge that you have been offered the opportunity to have any questions answered. Your participation in this study may be stopped if you fail to follow the study procedures or if the investigators feel that it is in your best interest to stop participation.

Confidentiality

Your identity will be held in strict confidence through the use of a coded subject number during data collection, data analysis, and in all references made to the data, both during and after the study, and in the reporting of the results. The results will be disseminated in the form of presentations at conferences, and publications in journals. The consent form containing your identity information will be destroyed three years after the completion of the study. If you decide to withdraw from the study, your information sheet and consent form with your identity and injury history will be destroyed.

Contact Information

If you have any questions about the study at any time or if you experience adverse effects as a result of participating in this study you can contact Jacob Gardner at 1914 Andy Holt Ave. 136 HPER Bldg., The University of Tennessee (974-2091). Questions about your rights as a participant can be addressed to Compliance Officer in the Office of Research at the University of Tennessee at (865) 974-3466.

Consent Statement

The study has been explained fully to my satisfaction and I agree to participate as described. I have been given the opportunity to discuss all aspects of this study and to ask questions. Answers to such questions, if any, were satisfactory. I am eighteen years of age or older, in good health, am qualified for the study and freely give my informed consent to serve as a subject in this study. I have received a copy of this form.

Subject's Name: ___________________ Subject's Signature: ________________________ Date: _________

Investigator's Signature: ____________________________ Date: __________ Subject # ______

APPENDIX C: Flyer

UT RESEARCHERS ARE LOOKING FOR PARTICIPANTS TO STUDY KNEE OSTEOARTHRITIS!

A team of researchers from the Department of Kinesiology, Recreation and Sport Studies at UT and the UT Medical Center are conducting research to study the influence of cycling on knee motions and pain in osteoarthritis. Test participants will receive a free evaluation of knee osteoarthritis by a medical doctor which will take approximately 45 min to 1 hour. Test and Control participants will be asked to attend one 1 – 1.5 hour testing sessions at the UT biomechanics laboratory.

Participants will receive $10 compensation

You may be able to participate if

- You are between the ages of 35 and 65
- Have knee OA (Test Group)
- Do not have knee OA (Control Group)
- Have not had ankle, knee or hip replacement
- Able to ride a stationary cycle without aid

If you'd like to participate, or would like more information, call **Jake Gardner** at the UT Biomechanics/Sports Medicine Lab.

- **Office: (865) 974-2091**
- **Cell: (406) 369-2885**
- **E-mail: jgardn25@utk.edu**

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APPENDIX D: Physical Activity Readiness Questionnaire (PAR-Q)

Par Q

1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor? Yes or No
2. Do you feel pain in your chest when you do physical activity? Yes or No
3. In the past month, have you had chest pain when you were not doing physical activity? Yes or No
4. Do you lose your balance because of dizziness or do you ever lose consciousness? Yes or No
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? Yes or No
6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition? Yes or No
7. Do you know of any other reason why you should not do physical activity? Yes or No

APPENDIX E: Knee Osteoarthritis Outcome Score (KOOS)

KOOS KNEE SURVEY

Today's date: ____/_____/_____ Date of birth: ____/_____/_____

Name: ____

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities.
Answer every question by ticking the appropriate box, only one box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms
These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S3. Does your knee catch or hang up when moving?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S4. Can you straighten your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

S5. Can you bend your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

Stiffness
The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S7. How severe is your knee stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Pain

P1. How often do you experience knee pain?

Never	Monthly	Weekly	Daily	Always
☐	☐	☐	☐	☐

What amount of knee pain have you experienced the **last week** during the following activities?

P2. Twisting/pivoting on your knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P3. Straightening knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P4. Bending knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P5. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P6. Going up or down stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P7. At night while in bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P8. Sitting or lying

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P9. Standing upright

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A2. Ascending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A4. Standing

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A5. Bending to floor/pick up an object

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A14. Sitting

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A17. Light domestic duties (cooking, dusting, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP2. Running

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP3. Jumping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP4. Twisting/pivoting on your injured knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP5. Kneeling

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never	Monthly	Weekly	Daily	Constantly
☐	☐	☐	☐	☐

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all	Mildly	Moderately	Severely	Totally
☐	☐	☐	☐	☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all	Mildly	Moderately	Severely	Extremely
☐	☐	☐	☐	☐

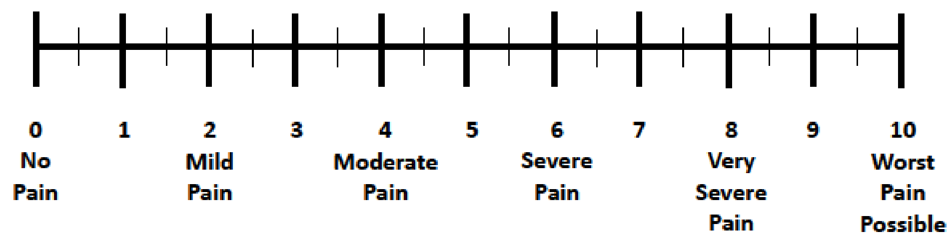
Q4. In general, how much difficulty do you have with your knee?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Thank you very much for completing all the questions in this questionnaire.

APPENDIX F: VAS Pain Scale

Pain Intensity Scale



APPENDIX G: Individual Results for Select Variables

Table 14: Peak medial pedal reaction force

		Peak Medial PRF (N)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-27.150±12.002	-35.757±8.360	-35.132±4.240	-38.346±12.533	-46.385±10.867
2	Healthy	-30.893±2.747	-32.432±2.253	-47.741±6.815	-36.587±2.188	-34.673±8.585
3	Healthy	-44.112±3.177	-46.376±4.102	-45.770±2.709	-42.145±4.323	-52.134±9.773
7	Healthy	-42.222±6.087	-45.880±13.090	-35.822±9.609	-36.726±5.862	-31.907±4.533
9	Healthy	-35.321±3.631	-37.509±6.583	-40.180±9.497	-38.341±5.896	-32.409±5.164
15	Healthy	-26.766±2.467	-33.580±3.917	-32.191±4.513	-22.829±4.775	-28.349±13.271
16	Healthy	-48.069±4.131	-46.507±1.970	-40.563±3.352	-37.811±3.085	-24.080±1.166
17	Healthy	-6.416±1.638	-12.559±6.481	-14.709±4.982	-11.108±2.660	-10.172±3.337
18	Healthy	-19.036±7.127	-18.038±4.052	-24.342±4.971	-28.348±3.826	-22.729±7.493
19	Healthy	-27.860±3.652	-39.287±2.110	-38.660±5.741	-39.985±5.987	-37.996±2.875
20	Healthy	-29.850±17.065	-36.813±6.003	-36.975±1.219	-37.344±14.116	-30.665±7.055
5	OA	-28.274±8.035	-28.314±4.573	-38.095±3.447	-47.341±2.920	-42.539±3.215
6	OA	-26.544±3.045	-32.651±1.874	-41.427±2.344	-28.266±2.581	-29.916±2.437
8	OA	-37.922±2.339	-42.823±2.489	-26.873±4.908	-36.360±7.636	-26.065±3.990
10	OA	-18.044±4.314	-22.997±2.501	-28.342±2.569	-22.511±3.236	-22.994±3.976
11	OA	-32.778±1.811	-35.319±2.515	-33.447±3.472	-37.283±6.356	-30.866±2.372
12	OA	-20.310±4.919	-9.598±4.416	-21.835±4.020	-18.515±7.554	-14.143±2.205
13	OA	-31.261±6.725	-30.807±4.419	-30.080±5.730	-47.907±7.379	-41.352±6.587
14	OA	-17.769±1.496	-28.799±4.792	-19.363±3.333	-25.410±3.476	-26.551±4.293
21	OA	-39.854±2.775	-42.489±4.479	-41.016±2.053	-33.901±4.552	-19.186±1.746
22	OA	-30.728±3.590	-40.642±1.745	-41.126±3.750	-31.978±1.755	-38.275±3.437
23	OA	-28.405±3.028	-32.386±2.897	-34.889±5.565	-23.276±5.112	-14.337±4.536
24	OA	-29.601±2.651	-28.071±3.137	-26.898±5.823	-25.234±4.435	-20.062±12.610
25	OA	-29.688±2.689	-7.559±2.891	-18.972±9.975	-15.796±1.984	-5.844±5.875
Mean±SD	Healthy	-30.696±11.825	-34.976±10.994	-35.644±9.395	-33.597±9.267	-31.954±11.383
Mean±SD	OA	-28.586±8.871	-27.420±14.158	-30.951±8.108	-28.937±12.943	-25.549±11.069

Table 15: Peak posterior pedal reaction force

		Peak Poster Pedal Reaction Force (N)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-66.619±16.794	-94.970±20.124	-78.499±15.389	-94.779±28.601	-96.119±21.720
2	Healthy	-82.081±4.985	-71.315±4.002	-77.660±17.920	-76.972±5.957	-81.265±14.546
3	Healthy	-87.904±10.290	-99.387±4.489	-94.450±9.043	-89.263±2.467	-85.511±12.413
7	Healthy	-87.564±62.012	-118.447±9.019	-108.575±18.001	-114.675±9.347	-108.543±8.416
9	Healthy	-57.262±3.379	-64.064±12.240	-62.824±16.125	-61.392±9.462	-55.029±12.550
15	Healthy	-60.286±6.834	-68.878±3.269	-61.998±3.376	-52.368±5.996	-57.656±18.844
16	Healthy	-52.065±6.337	-65.301±4.511	-27.559±3.534	-56.946±6.263	-19.666±3.700
17	Healthy	-18.960±2.778	-28.375±3.256	-24.681±3.845	-26.975±4.199	-30.940±7.952
18	Healthy	-79.188±10.436	-70.692±4.210	-85.122±6.423	-85.127±7.496	-76.717±5.390
19	Healthy	-64.419±8.533	-80.711±2.856	-77.409±4.489	-73.637±3.632	-69.079±3.964
20	Healthy	-55.268±6.966	-53.028±4.925	-55.204±15.686	-55.204±15.686	-65.423±19.970
5	OA	-83.926±11.250	-93.662±6.503	-99.954±8.500	-102.183±3.648	-103.704±6.063
6	OA	-99.906±2.700	-105.987±3.270	-108.155±3.959	-109.427±2.419	-105.609±1.430
8	OA	-73.262±4.393	-80.444±8.203	-42.228±8.177	-64.295±16.246	-68.645±5.794
10	OA	-60.508±12.946	-45.160±6.654	-55.186±7.433	-60.902±4.478	-44.230±9.889
11	OA	-98.621±5.264	-100.557±6.528	-97.737±11.123	-94.956±17.357	-94.483±4.592
12	OA	-89.402±5.223	-63.584±4.444	-74.950±11.026	-63.697±3.901	-75.775±7.737
13	OA	-117.029±18.804	-128.200±12.053	-125.771±14.114	-122.430±9.345	-109.802±11.549
14	OA	-59.553±6.145	-60.034±3.491	-51.752±6.791	-56.846±4.827	-47.646±5.615
21	OA	-78.050±1.130	-75.432±6.790	-23.374±5.951	-61.192±7.606	-64.149±3.527
22	OA	-85.999±6.369	-94.644±4.530	-87.251±6.572	-82.057±2.762	-86.423±8.640
23	OA	-60.786±6.357	-59.792±7.446	-65.955±7.099	-62.318±9.288	-31.770±6.979
24	OA	-64.143±9.043	-48.019±3.013	-70.493±4.771	-74.227±9.363	-78.785±10.508
25	OA	-74.787±18.363	-66.085±7.305	-73.514±7.982	-64.839±7.401	-71.843±9.582
Mean±SD	Healthy	-64.692±19.981	-74.106±24.176	-68.544±25.820	-71.576±24.228	-67.813±26.430
Mean±SD	OA	-80.459±17.669	-78.585±24.705	-75.102±28.494	-78.413±21.787	-75.605±24.536

Table 16: Peak vertical pedal reaction force

		Peak Vertical PRF (N)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	203.679±22.286	244.550±27.841	229.615±17.655	240.523±41.617	263.434±32.160
2	Healthy	268.140±9.726	249.287±10.676	301.020±47.607	258.073±11.341	286.511±34.349
3	Healthy	267.351±22.301	281.677±7.068	286.326±10.043	272.540±9.344	269.757±29.750
7	Healthy	192.569±61.511	201.662±13.886	194.591±30.994	196.513±16.132	200.224±14.319
9	Healthy	216.074±6.107	215.469±17.579	202.650±23.292	223.540±18.060	207.802±15.924
15	Healthy	218.667±16.729	248.115±11.845	214.690±6.840	219.038±14.275	234.366±31.966
16	Healthy	276.790±24.657	279.452±7.016	275.970±20.162	269.994±18.634	270.367±20.689
17	Healthy	190.790±9.466	200.070±24.536	193.434±20.915	188.297±25.580	186.102±29.629
18	Healthy	235.307±20.298	251.070±26.352	280.382±11.495	264.315±17.415	252.581±3.380
19	Healthy	192.245±13.039	228.904±8.224	223.354±13.911	224.578±15.037	217.794±9.838
20	Healthy	338.240±13.525	364.369±7.188	350.290±10.319	350.290±10.319	365.183±12.389
5	OA	219.741±29.146	255.606±18.593	264.423±10.958	285.271±12.550	280.154±11.738
6	OA	215.530±54.255	259.549±5.958	268.752±9.120	250.492±6.161	246.692±6.542
8	OA	201.781±8.928	235.449±11.944	211.939±15.738	251.015±15.893	224.736±12.557
10	OA	257.295±28.958	253.578±8.525	249.700±16.456	266.870±10.255	243.212±22.759
11	OA	191.539±12.296	190.088±14.549	182.067±17.693	214.560±9.508	199.054±15.253
12	OA	317.790±22.758	311.591±6.332	300.211±27.487	319.525±16.783	291.278±12.224
13	OA	254.405±18.264	263.810±22.212	271.807±31.495	277.024±16.617	268.423±13.080
14	OA	196.965±6.742	217.883±6.416	198.804±28.198	208.695±11.457	227.824±19.089
21	OA	244.060±16.438	240.070±15.112	241.621±21.685	231.501±17.172	235.490±10.392
22	OA	240.167±13.564	240.001±10.353	253.554±6.811	225.213±3.867	229.169±9.221
23	OA	209.851±17.581	207.026±23.104	210.210±19.715	200.954±9.894	211.584±14.693
24	OA	241.701±10.852	238.311±12.818	223.535±16.769	236.371±11.963	238.399±3.002
25	OA	257.587±27.431	271.489±22.159	297.587±17.574	272.225±20.015	262.530±31.355
Mean±SD	Healthy	236.350±46.603	251.330±46.297	250.211±51.441	246.155±44.874	250.375±50.214
Mean±SD	OA	234.493±34.244	244.958±30.640	244.170±37.002	249.209±34.266	242.965±26.620

Table 17: Maximum sagittal plane knee angle

		Maximum Sagittal Plane Knee Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-46.159±0.389	-51.392±0.427	-54.511±0.575	-44.482±0.426	-42.717±0.750
2	Healthy	-42.365±1.313	-45.725±0.878	-47.039±1.164	-42.229±0.418	-37.616±0.790
3	Healthy	-44.898±1.069	-41.631±0.699	-39.322±0.812	-46.056±1.326	-53.018±0.794
7	Healthy	-38.483±0.768	-42.329±0.915	-47.211±1.053	-35.558±0.322	-33.904±0.916
9	Healthy	-37.340±1.146	-43.301±1.205	-45.715±1.594	-32.584±1.394	-31.860±0.554
15	Healthy	-56.108±0.677	-56.442±1.381	-59.175±0.464	-55.785±1.276	-53.795±2.284
16	Healthy	-52.511±0.673	-55.998±0.096	-58.066±0.590	-51.713±0.464	-48.060±0.652
17	Healthy	-56.527±0.714	-56.712±1.123	-59.449±0.883	-52.551±1.076	-45.502±1.452
18	Healthy	-38.478±1.196	-38.906±4.023	-42.232±0.545	-34.977±1.075	-36.557±0.545
19	Healthy	-33.422±0.669	-37.607±0.214	-39.330±0.426	-32.245±0.557	-32.094±3.155
20	Healthy	-47.464±0.524	-49.069±0.171	-45.218±0.205	-45.218±0.205	-42.238±0.410
5	OA	-45.881±0.909	-50.420±0.482	-50.774±1.004	-44.857±0.390	-43.274±0.613
6	OA	-41.827±0.440	-45.679±0.338	-51.663±0.575	-42.004±0.412	-41.826±0.316
8	OA	-31.038±2.164	-46.980±3.405	-54.010±0.766	-37.603±1.847	-26.898±2.100
10	OA	-45.361±0.781	-51.100±0.796	-50.171±0.215	-45.057±0.439	-42.209±1.162
11	OA	-29.468±0.828	-35.025±0.764	-39.360±1.083	-25.294±0.889	-27.221±0.488
12	OA	-43.345±0.474	-49.166±0.142	-51.444±0.910	-43.354±0.331	-40.521±1.235
13	OA	-39.927±1.973	-43.297±1.385	-45.912±1.949	-35.945±0.988	-31.429±0.548
14	OA	-59.597±0.517	-61.849±0.484	-66.736±0.521	-57.798±0.523	-54.768±1.286
21	OA	-39.362±0.715	-44.697±0.537	-46.610±0.834	-40.926±0.463	-36.481±0.607
22	OA	-34.332±0.733	-37.626±0.914	-41.915±0.143	-31.963±0.452	-30.901±0.469
23	OA	-30.720±0.665	-36.587±1.161	-39.700±0.979	-24.603±0.933	-21.838±0.724
24	OA	-40.160±0.382	-43.899±0.330	-46.331±0.456	-35.925±0.593	-34.442±0.211
25	OA	-36.326±1.517	-39.127±0.651	-40.585±1.647	-34.244±0.500	-30.316±1.173
Mean±SD	Healthy	-44.887±7.776	-47.192±7.122	-48.843±7.682	-43.036±8.316	-41.578±7.896
Mean±SD	OA	-39.796±8.087	-45.035±7.278	-48.093±7.468	-38.429±8.850	-35.548±8.840

Table 18: Minimum sagittal plane knee angle

		Minimum Sagittal Plane Knee Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-111.512±1.351	-115.801±0.985	-116.627±0.939	-111.718±0.759	-109.702±1.096
2	Healthy	-112.385±0.331	-113.325±0.263	-114.043±0.354	-111.968±0.306	-110.513±0.201
3	Healthy	-122.620±0.522	-121.178±0.287	-119.029±0.217	-122.733±0.483	-125.694±0.409
7	Healthy	-103.108±31.860	-118.783±0.180	-120.337±0.195	-115.959±0.309	-115.540±0.177
9	Healthy	-115.457±0.374	-117.741±0.562	-118.594±0.430	-112.961±0.443	-111.659±0.782
15	Healthy	-120.666±0.474	-120.570±0.277	-122.105±0.316	-120.763±0.400	-119.169±0.200
16	Healthy	-122.730±0.300	-122.920±0.612	-121.256±0.384	-120.776±0.248	-117.353±0.851
17	Healthy	-121.909±0.513	-123.595±1.028	-123.416±0.803	-118.924±0.595	-115.614±0.809
18	Healthy	-109.776±2.786	-111.742±0.851	-111.511±0.511	-108.816±0.421	-107.879±0.646
19	Healthy	-107.522±0.103	-109.713±0.068	-110.819±0.321	-107.032±0.283	-104.516±2.120
20	Healthy	-107.494±0.589	-110.428±0.593	-107.364±0.187	-107.364±0.187	-105.626±0.199
5	OA	-114.697±0.413	-117.794±0.426	-117.224±0.300	-114.264±0.531	-111.038±0.511
6	OA	-117.955±0.196	-119.183±0.110	-120.697±0.250	-116.916±0.122	-116.407±0.225
8	OA	-127.702±0.607	-130.771±1.980	-127.871±0.536	-121.806±2.053	-114.593±1.848
10	OA	-112.832±1.015	-114.384±0.440	-111.076±0.801	-110.507±0.657	-108.324±1.261
11	OA	-113.355±0.165	-113.835±0.453	-114.956±0.158	-109.537±0.155	-109.603±0.334
12	OA	-105.566±0.110	-108.178±0.091	-110.057±0.102	-104.979±0.202	-101.941±0.892
13	OA	-116.487±0.636	-118.579±0.447	-119.204±0.470	-110.799±0.597	-107.869±0.908
14	OA	-125.110±0.472	-124.628±0.950	-126.838±0.687	-122.783±0.203	-121.130±0.337
21	OA	-110.638±0.441	-112.461±0.169	-109.934±0.544	-109.491±0.145	-106.238±0.130
22	OA	-108.732±0.165	-111.444±0.460	-112.310±0.146	-107.424±0.181	-106.468±0.132
23	OA	-105.200±0.620	-107.062±0.821	-109.269±0.652	-102.002±0.592	-102.340±0.388
24	OA	-110.471±0.294	-113.922±0.419	-115.458±0.317	-108.317±0.352	-107.778±0.361
25	OA	-109.465±0.457	-108.178±0.656	-110.716±0.687	-106.604±0.327	-107.770±0.254
Mean±SD	Healthy	-114.107±6.993	-116.891±5.012	-116.827±5.224	-114.456±5.689	-113.024±6.347
Mean±SD	OA	-113.708±6.800	-115.417±6.803	-115.816±6.300	-111.187±6.205	-109.346±5.394

Table 19: Maximum frontal plane knee angle

		Maximum Frontal Plane Knee Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	15.365±0.945	15.989±0.233	16.320±0.341	15.279±1.008	10.609±1.183
2	Healthy	7.638±0.457	5.389±0.615	6.508±0.415	4.467±0.546	5.876±1.133
3	Healthy	11.160±0.465	9.286±0.448	10.824±0.175	12.480±0.674	14.083±0.946
7	Healthy	17.310±0.510	18.474±1.230	16.637±0.596	18.153±0.427	16.858±0.281
9	Healthy	4.459±0.211	4.246±1.220	4.019±2.407	1.821±0.758	3.369±0.676
15	Healthy	12.923±0.773	12.700±0.849	11.946±0.806	9.928±0.454	8.450±0.617
16	Healthy	23.572±0.344	24.418±0.699	26.067±0.671	21.907±0.282	23.262±0.418
17	Healthy	10.819±0.665	11.079±0.604	11.908±0.733	5.948±1.334	10.973±1.243
18	Healthy	5.264±0.506	5.723±0.845	8.500±0.881	6.247±0.275	5.316±0.540
19	Healthy	14.398±0.251	13.779±0.333	12.900±0.313	13.512±0.565	12.443±0.566
20	Healthy	18.121±0.326	19.046±0.655	17.889±0.398	17.955±0.365	16.790±0.609
5	OA	8.133±0.452	7.165±0.811	8.675±0.392	3.242±0.277	4.155±0.920
6	OA	14.058±0.333	14.340±0.298	12.500±0.382	9.388±0.627	10.141±0.282
8	OA	21.329±0.657	18.280±0.344	20.134±0.947	17.698±0.494	24.489±1.113
10	OA	11.543±0.468	10.450±0.419	8.367±0.508	8.205±0.413	6.948±2.168
11	OA	11.804±0.172	11.973±0.162	11.766±0.426	6.239±0.732	10.093±0.949
12	OA	16.765±0.964	17.122±0.471	17.812±0.326	16.927±0.418	13.923±0.353
13	OA	8.169±0.701	7.720±0.789	8.860±0.751	3.720±0.861	-0.144±0.621
14	OA	8.065±0.701	6.511±0.702	7.540±0.437	2.541±0.744	0.871±0.701
21	OA	10.051±0.979	8.013±0.477	13.333±0.815	5.586±0.716	3.385±0.400
22	OA	9.399±0.348	8.851±0.722	7.969±0.351	8.262±0.221	7.620±0.216
23	OA	5.748±1.599	5.502±0.525	5.931±0.963	5.641±0.552	3.472±0.677
24	OA	13.949±0.639	14.709±0.722	15.356±0.361	11.113±0.975	10.171±0.462
25	OA	19.446±0.432	24.224±0.359	27.326±0.702	23.393±1.253	21.542±0.618
Mean±SD	Healthy	12.821±5.775	12.739±6.409	13.047±6.074	11.609±6.461	11.639±5.906
Mean±SD	OA	12.189±4.724	11.912±5.553	12.736±6.112	9.381±6.343	8.974±7.454

Table 20: Minimum frontal plane knee angle

		Minimum Frontal Plane Knee Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	7.413±1.069	7.803±0.440	7.258±0.355	5.361±0.941	2.268±0.488
2	Healthy	-4.235±0.635	-6.971±0.901	-7.455±0.649	-6.388±0.376	-7.040±0.540
3	Healthy	-1.184±0.260	-1.837±0.331	-0.811±0.538	-0.362±0.288	0.421±1.080
7	Healthy	6.894±0.436	6.877±0.436	7.008±0.287	6.683±0.305	5.401±0.186
9	Healthy	-1.581±0.999	-2.507±0.922	-2.539±1.359	-3.988±1.480	-3.663±0.338
15	Healthy	0.975±0.469	0.167±1.385	-0.888±1.209	2.293±0.786	-0.222±0.821
16	Healthy	10.671±0.343	10.442±0.326	9.189±0.617	6.675±0.267	7.546±0.577
17	Healthy	3.297±1.788	1.386±0.697	2.768±1.100	-2.828±0.371	-1.171±1.155
18	Healthy	-5.992±1.397	-6.511±0.452	-6.231±0.859	-4.335±0.738	-5.690±0.823
19	Healthy	1.534±0.277	2.294±0.231	0.319±1.787	-2.159±0.463	-2.162±1.556
20	Healthy	6.562±0.882	6.056±0.616	4.748±0.355	4.748±0.355	2.578±0.525
5	OA	-5.941±0.538	-5.966±1.245	-5.319±0.478	-7.727±0.618	-6.397±0.743
6	OA	9.515±0.196	9.732±0.232	7.399±0.447	5.753±0.451	5.679±0.236
8	OA	10.208±0.904	8.802±0.640	13.336±0.119	6.001±0.783	6.870±1.385
10	OA	4.809±0.395	3.800±0.816	1.366±0.354	3.513±0.405	0.357±0.566
11	OA	4.855±0.620	2.913±0.725	2.485±0.791	-2.401±1.162	0.113±0.305
12	OA	8.160±0.777	8.826±0.247	9.467±0.399	6.377±1.193	5.473±0.915
13	OA	-1.610±0.604	-2.566±0.268	-2.743±1.089	-5.342±0.331	-8.904±0.333
14	OA	3.012±0.722	-0.723±0.612	2.142±1.064	-2.350±0.554	-3.671±0.487
21	OA	-1.794±0.162	-3.042±0.676	-5.559±0.994	-4.016±0.599	-3.952±0.372
22	OA	5.707±0.338	5.296±0.220	5.246±0.356	4.053±0.691	2.124±0.501
23	OA	-0.336±1.086	-0.239±0.707	-1.528±2.410	-3.667±0.616	-4.088±0.328
24	OA	7.406±0.667	8.533±0.665	9.050±0.658	6.690±0.574	6.230±0.466
25	OA	13.714±0.498	14.612±0.879	17.718±0.736	15.502±0.533	15.370±0.794
Mean±SD	Healthy	2.214±5.267	1.563±5.787	1.215±5.501	0.518±4.807	-0.158±4.464
Mean±SD	OA	4.439±5.610	3.844±6.105	4.082±7.134	1.722±6.546	1.169±6.692

Table 21: Maximum transverse plane knee angle

		Maximum Transverse Plane Knee Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	5.066±0.404	8.597±0.487	8.647±0.763	5.904±1.073	6.040±1.297
2	Healthy	1.945±0.600	0.474±0.604	1.470±0.566	-0.224±0.166	1.217±0.778
3	Healthy	3.285±0.644	4.874±0.062	4.844±0.416	4.919±0.341	3.706±1.607
7	Healthy	12.785±5.784	14.724±0.804	13.847±0.425	16.293±0.621	16.395±0.294
9	Healthy	2.026±2.212	1.458±1.902	2.549±1.214	1.215±0.543	3.332±0.498
15	Healthy	6.680±0.578	9.565±1.252	8.697±0.826	5.570±0.350	3.923±1.423
16	Healthy	12.517±0.553	13.264±0.967	12.427±0.765	12.336±0.517	12.519±0.858
17	Healthy	5.835±0.761	10.865±0.722	10.373±0.527	7.351±1.065	12.948±1.240
18	Healthy	-9.703±1.395	-5.419±1.856	-8.881±1.527	-9.199±1.332	-12.600±1.545
19	Healthy	1.691±0.545	3.171±0.379	4.960±0.302	4.915±1.166	4.573±2.046
20	Healthy	2.093±0.616	6.210±0.579	2.519±0.194	2.519±0.194	3.249±0.797
5	OA	-3.166±0.753	-1.923±0.753	-1.972±0.466	-3.209±0.728	-4.483±0.655
6	OA	10.847±0.234	11.188±0.549	10.156±0.120	8.256±0.488	8.425±0.456
8	OA	7.974±1.623	7.489±0.845	5.922±0.802	6.104±1.375	5.851±1.597
10	OA	-1.143±0.534	-1.764±0.268	-2.918±0.265	-1.118±0.333	-0.903±1.065
11	OA	0.804±0.371	-0.557±1.020	-0.487±0.550	-0.021±0.481	1.853±0.657
12	OA	6.849±0.859	7.664±0.436	10.216±0.593	6.739±0.683	2.738±1.040
13	OA	1.429±1.208	1.849±1.203	3.403±1.758	0.174±1.088	-0.484±0.665
14	OA	11.101±1.233	9.587±0.565	11.121±0.667	10.141±0.972	9.555±0.370
21	OA	4.910±0.352	4.493±0.345	4.674±0.487	4.283±0.162	4.546±0.618
22	OA	2.998±0.345	3.572±0.332	3.372±0.348	4.091±0.321	4.899±0.221
23	OA	8.338±0.527	8.016±0.854	9.154±0.688	8.444±1.557	8.145±1.260
24	OA	12.022±0.288	13.221±0.454	14.784±0.481	11.952±0.424	11.149±0.518
25	OA	21.903±0.405	21.188±0.296	23.335±0.743	20.785±0.120	22.004±0.428
Mean±SD	Healthy	4.020±6.056	6.162±6.021	5.587±6.347	4.691±6.606	5.028±7.639
Mean±SD	OA	6.528±6.698	6.463±6.585	6.981±7.285	5.894±6.422	5.638±6.676

Table 22: Minimum transverse plane knee angle

		Minimum Transverse Plane Knee Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-12.096±1.049	-9.950±0.441	-9.690±0.505	-8.978±2.522	-5.453±0.955
2	Healthy	-1.576±0.401	-3.558±0.513	-3.719±0.473	-2.877±0.775	-2.335±1.096
3	Healthy	-3.059±0.539	-1.094±1.409	-0.286±0.866	-2.519±1.232	-4.453±0.927
7	Healthy	3.380±2.546	2.075±2.504	0.653±1.768	0.628±0.385	-0.107±0.446
9	Healthy	-3.117±0.376	-1.037±0.740	-0.015±0.533	0.265±0.739	0.563±0.672
15	Healthy	-0.613±1.437	-1.050±0.203	-1.419±1.978	3.062±0.655	1.927±1.533
16	Healthy	-7.919±0.792	-6.772±1.289	-7.004±0.246	-5.256±0.751	-5.958±0.419
17	Healthy	-0.001±0.868	-0.120±0.727	-0.231±0.551	0.162±0.606	-0.306±0.966
18	Healthy	-11.373±1.822	-9.113±1.057	-14.180±0.672	-12.240±1.471	-13.151±1.326
19	Healthy	-8.025±0.337	-7.125±0.106	-5.042±0.475	-5.132±0.278	-3.807±0.297
20	Healthy	-10.357±0.580	-9.053±0.570	-9.869±0.271	-9.869±0.271	-9.140±0.438
5	OA	-7.628±0.402	-7.650±0.553	-9.558±0.504	-7.276±0.705	-8.154±1.005
6	OA	-6.341±0.496	-7.051±0.400	-3.451±0.351	-4.716±0.327	-3.582±0.292
8	OA	-10.628±1.484	-7.540±0.900	-8.045±0.519	-4.894±0.814	-11.125±0.911
10	OA	-3.626±0.347	-4.665±1.082	-4.861±0.238	-3.553±0.308	-3.301±0.235
11	OA	-7.474±0.932	-7.793±0.368	-6.875±0.555	-3.826±0.572	-5.294±0.374
12	OA	-0.318±0.342	0.565±0.369	-0.299±0.497	-0.377±0.692	0.119±0.368
13	OA	-10.766±0.620	-9.661±1.145	-7.482±3.246	-2.007±1.629	-2.368±1.547
14	OA	5.215±0.709	4.609±0.773	5.325±0.835	6.030±1.093	5.796±0.491
21	OA	-1.471±1.049	-2.147±1.228	-4.327±0.967	-0.417±0.574	0.669±1.116
22	OA	-3.230±1.916	-1.827±0.792	-6.560±0.397	-8.645±0.312	-7.421±0.306
23	OA	2.773±0.371	3.307±1.159	4.584±1.902	5.888±1.411	5.639±1.179
24	OA	2.588±0.700	2.578±0.701	3.622±0.430	6.357±0.825	6.178±0.596
25	OA	3.298±0.552	-0.424±0.900	-1.945±0.671	-0.664±0.407	-0.879±0.709
Mean±SD	Healthy	-4.978±5.204	-4.254±4.265	-4.618±4.986	-3.887±4.900	-3.838±4.512
Mean±SD	OA	-2.893±5.421	-2.900±4.821	-3.067±5.006	-1.392±4.934	-1.825±5.509

Table 23: Peak Extensor Knee Moment

		Peak Extensor Knee Moment (Nm)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	28.634±5.450	39.250±9.339	33.346±7.871	36.550±10.670	39.465±6.847
2	Healthy	31.699±4.041	27.236±1.862	33.673±11.428	31.202±3.252	33.236±8.733
3	Healthy	33.690±4.884	36.885±2.987	34.167±3.794	31.684±1.595	32.567±4.848
7	Healthy	49.217±8.353	52.409±3.952	49.482±9.589	51.438±5.071	48.966±5.402
9	Healthy	21.289±1.223	26.977±5.447	27.513±6.897	24.202±4.003	22.251±4.679
15	Healthy	22.175±3.034	28.253±2.150	26.316±2.045	17.803±2.231	21.806±5.842
16	Healthy	25.197±2.319	30.191±2.322	22.020±0.469	24.755±1.629	23.142±0.563
17	Healthy	15.441±4.515	18.671±2.330	18.671±1.211	18.582±2.135	15.528±3.380
18	Healthy	22.940±2.353	22.576±2.918	24.544±2.544	23.345±2.135	24.914±3.239
19	Healthy	23.663±4.334	33.429±2.030	31.937±2.670	27.124±1.974	25.811±2.195
20	Healthy	15.019±3.546	17.978±2.551	18.075±1.564	16.256±3.777	17.535±7.268
5	OA	27.938±3.905	34.931±1.172	37.869±3.444	38.553±2.229	37.472±1.756
6	OA	34.566±1.519	39.037±1.283	42.872±1.904	40.640±1.516	38.747±1.593
8	OA	26.692±1.362	35.287±1.447	24.112±3.775	29.563±3.801	24.632±1.501
10	OA	22.971±5.945	21.099±3.068	25.892±3.713	27.301±2.197	18.327±4.570
11	OA	38.954±2.422	40.752±2.451	40.781±4.136	34.923±3.606	35.372±2.280
12	OA	24.133±1.858	13.287±1.537	22.431±3.728	11.786±2.527	15.809±4.295
13	OA	41.864±8.628	44.290±5.895	43.923±6.873	45.494±5.010	41.514±4.709
14	OA	22.606±2.390	26.081±0.716	21.773±4.417	22.805±2.235	19.830±4.018
21	OA	28.621±0.832	30.385±4.367	9.570±0.486	26.688±1.953	23.986±1.094
22	OA	33.973±0.996	35.447±3.340	34.174±3.248	27.682±1.242	33.285±3.154
23	OA	18.031±2.458	21.518±3.080	24.250±3.492	20.747±3.552	14.442±2.474
24	OA	24.445±2.924	18.047±2.309	25.940±1.815	26.185±3.304	27.793±3.868
25	OA	18.737±5.497	15.226±1.156	16.386±2.788	15.946±2.552	15.849±2.966
Mean±SD	Healthy	26.269±9.598	30.350±9.967	29.068±8.942	27.540±10.109	27.747±9.965
Mean±SD	OA	27.964±7.418	28.876±10.304	28.459±10.619	28.332±9.659	26.697±9.643

Table 24: Peak abduction knee moment

		Peak Abduction Knee Moment (Nm)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-13.692±4.182	-15.664±2.593	-13.345±2.278	-16.334±5.042	-17.898±4.252
2	Healthy	-8.293±1.244	-7.847±1.886	-8.805±1.663	-7.446±0.777	-8.690±3.231
3	Healthy	-9.146±1.061	-8.818±0.876	-10.189±0.397	-7.765±0.802	-7.419±0.944
7	Healthy	-17.336±3.135	-18.189±2.814	-13.281±2.827	-18.702±2.143	-15.286±1.800
9	Healthy	-6.099±1.344	-2.442±1.281	-2.816±1.775	-4.541±1.489	-4.154±2.250
15	Healthy	-7.540±0.488	-7.695±0.866	-6.509±0.515	-7.622±1.173	-7.843±3.360
16	Healthy	-14.738±1.498	-14.094±0.813	-7.951±0.878	-12.827±0.299	-13.427±0.396
17	Healthy	-3.745±0.801	-4.204±0.185	-4.527±0.279	-5.310±0.768	-5.863±0.835
18	Healthy	-1.429±0.720	-1.487±0.754	-0.763±0.283	-2.984±1.104	-2.059±0.946
19	Healthy	-7.124±1.232	-9.952±0.731	-8.827±0.826	-11.416±2.249	-9.652±0.679
20	Healthy	-9.804±2.634	-8.341±1.044	-12.413±1.486	-10.001±3.842	-8.075±1.777
5	OA	-2.760±1.001	-2.250±0.503	-2.320±0.350	-4.061±0.890	-3.116±0.872
6	OA	-12.391±0.979	-12.676±0.703	-12.213±0.712	-13.498±0.908	-14.897±0.235
8	OA	-6.242±0.452	-4.319±0.469	-5.046±0.817	-7.504±2.019	-8.467±0.758
10	OA	-1.915±0.354	-2.528±0.219	-2.493±0.564	-3.708±0.903	-2.155±0.151
11	OA	-4.030±0.569	-3.081±0.845	-2.577±0.308	-2.947±0.443	-2.799±0.222
12	OA	-14.378±2.411	-5.389±0.865	-10.047±2.573	-11.407±3.110	-7.159±1.114
13	OA	-5.016±1.408	-2.497±0.824	-2.851±1.129	-9.024±1.188	-7.918±0.807
14	OA	-2.579±0.220	-2.643±0.180	-2.503±0.362	-3.082±0.218	-2.960±0.116
21	OA	-6.833±1.083	-5.886±1.121	-4.863±0.826	-4.193±0.602	-2.869±0.235
22	OA	-14.095±1.156	-10.717±0.766	-9.836±1.945	-10.290±0.960	-11.293±1.613
23	OA	-5.686±1.666	-4.582±0.860	-4.061±0.793	-3.260±0.341	-2.882±0.243
24	OA	-9.910±0.713	-8.474±0.462	-9.277±1.277	-10.232±1.042	-10.688±1.235
25	OA	-14.570±2.775	-6.841±1.216	-9.805±0.635	-6.875±1.358	-9.694±3.236
Mean±SD	Healthy	-8.995±4.739	-8.976±5.307	-8.130±4.188	-9.541±4.917	-9.124±4.742
Mean±SD	OA	-7.723±4.755	-5.530±3.344	-5.992±3.658	-6.929±3.655	-6.685±4.183

Table 25: Peak internal rotation knee moment

		Peak Internal Rotation Knee Moment (Nm)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	10.160±2.954	12.202±2.350	12.269±1.878	11.689±4.021	12.500±2.529
2	Healthy	5.955±0.406	6.713±0.496	11.772±2.037	7.804±1.197	5.820±2.038
3	Healthy	11.154±1.216	9.447±1.083	9.687±0.758	9.195±0.989	12.273±2.507
7	Healthy	10.060±0.519	12.014±4.830	7.838±2.381	9.440±1.649	7.302±1.646
9	Healthy	6.294±0.863	5.879±1.751	6.355±1.848	3.646±0.932	3.934±1.767
15	Healthy	8.987±0.920	10.597±0.999	9.050±1.069	6.401±0.952	6.601±3.105
16	Healthy	16.459±1.550	14.925±0.683	13.182±1.272	8.289±0.928	6.142±0.437
17	Healthy	-0.159±0.313	1.520±1.242	2.119±0.784	0.918±0.469	-0.746±0.124
18	Healthy	4.261±1.824	2.230±1.199	4.219±1.167	4.653±0.546	5.838±1.348
19	Healthy	5.813±0.568	7.683±0.543	6.523±1.133	7.415±1.748	5.978±0.734
20	Healthy	8.752±3.621	5.618±1.551	10.178±0.499	7.137±3.672	1.473±1.169
5	OA	8.231±1.697	5.158±1.278	8.674±1.106	9.823±1.187	9.336±0.975
6	OA	8.522±1.325	10.170±0.735	12.949±1.087	9.521±1.060	9.223±0.581
8	OA	6.184±0.598	6.112±0.658	2.134±0.918	1.453±2.349	0.980±0.673
10	OA	0.186±0.957	-3.583±1.847	0.641±0.612	0.306±1.216	-1.097±0.282
11	OA	2.897±0.412	5.540±1.053	3.383±1.229	0.780±1.694	1.203±0.754
12	OA	9.174±2.122	4.535±1.755	8.881±1.772	9.231±3.486	2.398±0.308
13	OA	8.162±3.418	6.868±0.868	5.191±1.388	11.719±1.419	6.903±2.804
14	OA	2.922±0.420	0.927±1.967	0.909±0.509	-0.043±0.870	-1.203±0.580
21	OA	10.038±1.681	12.436±4.949	8.507±2.009	8.682±1.102	-0.192±1.319
22	OA	11.659±0.511	9.117±1.022	11.654±3.098	5.626±1.088	7.049±0.955
23	OA	4.170±1.408	6.661±0.616	6.806±1.881	1.569±1.259	0.193±1.340
24	OA	5.088±1.114	6.120±0.625	2.967±1.661	0.386±0.332	-0.099±1.053
25	OA	8.347±2.794	3.221±1.130	5.614±1.293	4.364±0.765	3.265±1.228
Mean±SD	Healthy	7.976±4.294	8.075±4.215	8.472±3.455	6.963±2.989	6.101±3.923
Mean±SD	OA	6.583±3.324	5.637±4.045	6.024±3.969	4.878±4.395	2.920±3.876

Table 26: Peak ankle plantarflexion angle

		Peak Ankle Plantarflexion Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-28.789±1.946	-19.780±2.376	-24.417±1.744	-21.089±2.310	-21.134±2.343
2	Healthy	-10.491±0.884	-6.909±1.204	-6.135±2.022	-10.175±1.076	-10.633±1.796
3	Healthy	-2.822±0.809	1.803±1.615	-2.282±0.478	2.082±1.658	0.210±2.011
7	Healthy	-5.467±1.549	-3.820±0.695	-4.163±2.123	-6.430±0.766	-3.903±1.344
9	Healthy	-21.928±4.528	-18.808±3.113	-14.919±2.594	-23.452±4.581	-23.036±1.781
15	Healthy	5.601±0.936	9.187±1.824	11.856±0.846	6.623±0.653	4.611±1.331
16	Healthy	-6.026±0.531	-9.989±1.125	-18.478±1.068	-4.391±0.915	-8.917±0.563
17	Healthy	-8.376±3.228	-23.089±2.646	-19.225±1.591	-13.391±2.086	-17.643±2.109
18	Healthy	7.011±2.636	11.889±9.334	8.144±0.919	4.765±3.010	5.147±2.655
19	Healthy	-16.764±0.516	-15.283±0.369	-12.530±0.725	-18.484±0.924	-19.925±1.002
20	Healthy	-10.235±0.849	-10.713±0.585	-5.921±1.154	-5.921±1.154	-7.295±0.548
5	OA	-5.141±1.297	-5.367±0.735	-2.278±3.005	-7.491±0.911	-8.580±1.955
6	OA	-0.076±0.891	2.867±0.170	3.958±0.702	0.635±0.804	-0.151±0.411
8	OA	9.412±0.376	8.511±4.965	-5.956±1.107	-5.822±10.483	-20.147±1.703
10	OA	-16.919±3.306	-26.814±1.082	-33.626±2.255	-28.496±1.719	-24.376±1.854
11	OA	0.813±1.762	-5.912±1.469	-2.232±1.179	-16.874±1.529	-13.283±1.520
12	OA	-0.865±1.162	1.335±0.445	2.200±0.534	-1.939±1.178	0.878±1.568
13	OA	4.695±4.456	4.899±0.992	7.808±2.848	-3.788±3.957	-9.185±2.936
14	OA	-8.213±0.735	-9.444±2.156	-4.589±0.548	-5.097±0.605	-4.398±1.418
21	OA	-17.702±1.107	-15.941±0.781	-30.496±2.005	-20.178±1.188	-23.350±0.680
22	OA	-13.248±1.074	-10.289±1.214	-11.940±1.029	-17.000±1.241	-17.400±1.493
23	OA	-5.546±1.546	-6.394±1.062	-1.609±1.000	-5.931±1.690	-5.799±2.119
24	OA	-13.011±1.238	-8.010±0.715	-6.534±2.470	-13.222±0.368	-12.665±1.204
25	OA	-12.211±1.979	-17.843±0.556	-20.454±3.961	-17.730±0.744	-13.919±0.986
Mean±SD	Healthy	-8.935±10.734	-7.774±11.631	-8.006±11.330	-8.169±10.255	-9.320±10.193
Mean±SD	OA	-6.001±8.476	-6.800±9.834	-8.135±12.764	-10.995±8.565	-11.721±8.187

Table 27: Peak ankle eversion angle

		Peak Ankle Eversion Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	-1.902±1.570	-4.180±0.553	-3.985±0.665	0.292±0.693	-0.419±1.175
2	Healthy	-2.061±0.857	-3.046±1.185	-4.361±0.461	1.105±0.358	-2.657±0.535
3	Healthy	-10.792±0.854	-8.563±0.406	-7.644±0.465	-14.059±0.187	-12.738±2.221
7	Healthy	-12.106±0.537	-14.435±0.406	-16.734±0.493	-13.818±0.215	-14.390±0.258
9	Healthy	-19.447±0.950	-21.493±1.321	-21.266±0.954	-15.722±0.858	-18.545±1.350
15	Healthy	-7.136±0.350	-14.842±0.608	-17.119±0.847	-15.014±1.298	-9.713±0.935
16	Healthy	-19.446±0.533	-24.423±0.074	-24.570±0.433	-15.166±0.440	-16.667±0.253
17	Healthy	-27.485±1.033	-34.365±1.533	-35.290±2.010	-15.417±1.028	-18.398±1.933
18	Healthy	-22.219±1.947	-28.174±0.809	-30.838±0.680	-19.316±1.369	-19.831±1.359
19	Healthy	-15.939±0.428	-21.252±0.445	-23.904±0.306	-16.911±0.252	-14.567±0.574
20	Healthy	-6.453±1.269	-9.754±1.109	-6.149±0.741	-6.149±0.741	-4.476±0.562
5	OA	-4.607±0.452	-8.220±0.715	-12.469±0.122	-1.662±0.787	-4.262±1.083
6	OA	-2.395±0.571	-5.338±0.312	-4.446±0.557	-2.856±0.168	-2.251±0.719
8	OA	-8.786±1.240	-9.247±1.157	-14.146±1.124	-6.668±2.560	-4.553±0.677
10	OA	-13.500±1.051	-16.123±0.625	-17.938±0.737	-14.096±1.471	-13.178±0.571
11	OA	-13.502±0.675	-16.509±0.502	-18.873±0.984	-6.345±0.695	-14.099±0.522
12	OA	-1.380±1.122	-2.267±0.574	-3.323±0.662	-0.384±0.393	3.965±0.172
13	OA	-15.193±1.348	-21.722±2.152	-25.025±1.011	-17.915±5.275	-9.253±1.253
14	OA	-14.025±1.078	-15.475±0.646	-19.660±0.355	-11.670±0.362	-4.929±0.642
21	OA	4.787±1.736	3.910±0.937	0.121±0.910	6.432±1.245	5.902±0.837
22	OA	0.395±0.234	-4.475±0.537	-5.954±0.279	-0.220±0.495	-1.549±0.218
23	OA	-12.058±0.735	-17.124±1.164	-24.030±0.550	-6.549±0.894	-4.332±0.735
24	OA	-7.351±1.013	-8.616±0.393	-13.570±0.259	-2.866±0.494	-2.388±0.848
25	OA	-1.243±0.319	-5.791±0.669	-4.736±2.745	-3.371±1.052	0.080±1.325
Mean±SD	Healthy	-13.181±8.437	-16.775±10.066	-17.442±10.870	-11.834±6.976	-12.037±6.804
Mean±SD	OA	-6.835±6.550	-9.769±7.224	-12.619±8.299	-5.244±6.442	-3.911±5.809

Table 28: Peak ankle internal rotation angle

		Peak Ankle Internal Rotation Angle (°)				
Subject	Group	Healthy	5 deg Wedge	10 deg Wedge	5 deg Toe-in	10 deg Toe-in
1	Healthy	6.071±1.469	3.041±0.565	1.021±0.681	5.401±0.551	7.142±0.371
2	Healthy	8.557±0.337	5.104±0.716	4.419±0.961	8.016±0.229	7.356±0.672
3	Healthy	12.003±0.236	11.914±0.336	11.498±0.049	12.323±0.707	13.966±0.733
7	Healthy	3.384±2.079	6.082±0.417	3.664±0.580	5.397±0.274	2.775±0.282
9	Healthy	12.314±2.328	11.482±1.451	10.746±1.553	12.311±1.611	12.790±1.514
15	Healthy	7.608±1.056	1.703±0.605	0.583±0.756	7.310±1.465	7.265±0.461
16	Healthy	21.552±0.350	19.813±0.165	14.188±0.425	18.644±0.330	20.924±0.130
17	Healthy	-4.775±0.517	-9.773±1.217	-7.318±0.428	4.391±0.751	4.939±1.047
18	Healthy	9.017±2.313	11.795±0.396	5.018±0.638	13.888±1.249	17.737±1.196
19	Healthy	4.349±0.499	-0.495±0.436	-0.715±0.550	10.800±0.172	11.442±0.262
20	Healthy	21.336±0.606	16.957±0.295	21.214±0.427	21.214±0.427	23.484±0.347
5	OA	15.063±0.717	13.628±1.277	15.837±0.587	17.444±0.623	18.344±1.116
6	OA	-6.376±0.383	-7.177±0.371	-9.351±0.813	-3.507±0.172	-3.134±0.144
8	OA	17.878±1.575	13.658±0.405	17.848±0.959	19.302±3.079	20.263±1.626
10	OA	7.206±1.404	8.506±0.796	6.715±0.269	9.245±0.461	11.791±0.156
11	OA	6.083±0.686	5.041±1.469	7.095±0.881	13.358±0.939	11.207±0.555
12	OA	9.570±0.367	6.949±0.568	5.913±0.671	10.390±0.725	14.871±1.265
13	OA	12.534±3.500	14.829±3.272	8.577±2.623	18.892±1.352	22.982±1.857
14	OA	7.390±0.302	6.343±1.207	4.307±0.290	7.472±0.264	11.905±0.304
21	OA	19.029±0.588	16.266±3.273	14.194±0.332	19.128±0.679	18.650±0.162
22	OA	6.356±0.285	4.070±0.264	2.859±0.742	11.253±0.375	13.331±0.609
23	OA	5.717±1.575	3.989±0.892	3.180±0.587	14.793±0.358	16.729±0.981
24	OA	-2.063±0.662	-3.101±0.857	-6.253±0.714	6.405±0.359	8.103±1.977
25	OA	7.301±0.602	5.573±0.953	6.079±1.263	8.057±0.612	11.636±0.189
Mean±SD	Healthy	9.220±7.634	7.057±8.463	5.847±7.978	10.881±5.508	11.802±6.718
Mean±SD	OA	8.130±7.147	6.813±6.857	5.923±7.741	11.710±6.509	13.591±6.569

VITA

Jacob Gardner was born in Canton, IL, to the parents of Derek and DeLynn Gardner. He is the second of three sons: Joshua, and Jesse. He grew up and attended elementary through high school in Corvallis, Montana. After his graduation, Jacob received his Bachelors of Science degree in 2005 from the University of Montana in Health and Human Development, with an emphasis in Exercise Science; this is where his interest in Biomechanics began. After graduation, Jacob worked for three years for a start-up company that designed and produced pediatric wheelchairs for children with cerebral palsy, where he gained valuable industry experience. He then followed his desire to learn more about Biomechanics by pursuing a Master's degree at Illinois State University; graduating in 2010. Finally, Jacob went on to pursue and ultimately complete a PhD in Biomechanics at the University of Tennessee. He graduated in 2013 and has accepted a tenure track, assistant professor position at Biola University in La Mirada, CA.